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Biomechanical modeling of the human jaw joint

Maarten Beek
BIOMECHANICAL MODELING OF THE HUMAN JAW JOINT

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- Beek M, Koolstra JH, Van Eijden TMGJ (2001). Human temporomandibular joint disc cartilage as a poroelastic material. (submitted for publication). (Chapter 7)
Chapter 1

GENERAL INTRODUCTION

The research of the department of Functional Anatomy of the Academic Centre for Dentistry in Amsterdam (ACTA) is focussed on the relationship between form and function in the human masticatory system. This masticatory system is daily used during habitual tasks like talking and chewing. Its osseous framework comprises the upper jaw (maxilla) and the lower jaw (mandible). The movement of the mandible with respect to the skull is facilitated by two temporomandibular joints. Compared to other joints in the body, the temporomandibular joints are unique: they cannot function independently from each other, their anatomical geometry is very complex, and they contain a disc made of cartilage, which is considered to act more or less like a pillow. Jaw movements are caused by contraction of the masticatory muscles that are attached between the mandible and the skull (jaw closers) or the hyoid bone (jaw openers). The forces, generated by these masticatory muscles, are not exclusively used for crushing food, but are also absorbed by the jaws and joints. The resulting deformations of the stiff jaws are negligible, which enables them to serve as rigid foundations for the teeth. The large deformations of the soft structures in the joint, on the other hand, ensure that loads are absorbed and spread over larger contact areas, thus preventing harmful peak loads.

Biological structures are adaptive and continuously experience changes in their geometry and internal structure. In addition to age (Roth and Mow, 1980) and hormones, this adaptation is also influenced by mechanical stimuli (Carter et al., 1998; Turner, 1998; Huiskes, 2000). Because the temporomandibular joints are loaded during function, they experience such stimuli. Therefore, the structures in these joints are also subject to adaptation (Moffett et al., 1964; Tuominen et al., 1996; Mao et al., 1998; Jonsson et al., 1999). On the other hand, these loads might cause wear and other kinds of damage. The degeneration of the structures might lead to discomfort to the person involved, both during masticatory activity and rest (Stegenga et al., 1991). In his review of the epidemiological literature related to
temporomandibular disorders, Carisson (1999) found the average value for perceived dysfunction to be 30% among 15,000 subjects and for clinically assessed dysfunction to be 44% among 16,000 subjects. Despite these figures, the causes for joint dysfunction have not been established unambiguously yet. One of the main reasons for this lack of knowledge is the inaccessibility of the joint, which complicates experimental research.

Detailed information about deformations in the structures in the human temporomandibular joint during joint loading can be assumed to be crucial to unravel the adaptation and degeneration processes occurring in these structures. Therefore, the purpose of the present research was to determine the distribution of deformations in the structures in the joint that occur during joint loading. While large joint loads were predicted during static biting (Koolstra et al., 1988) and the dynamical behavior of the relevant tissues is still not sufficiently known, the present study was initially limited to statical loads. To overcome the inability to measure deformations experimentally, a biomechanical modeling approach was applied. The three-dimensional anatomy of the joint and the presumed three-dimensional nature of the loads and deformations enforced the application of a three-dimensional model. This model was created from anatomical data, obtained from measurements on a human cadaver head. In order to apply these measurements, the development of a suitable method was required to generate accurate surface representations. Biological structures generally show complex time-dependent mechanical behavior. For the temporomandibular joint disc, this particular behavior has not been subject to adequate quantitative description yet. Therefore, in situ experiments were performed to quantify important features of the disc's tissue behavior. An appropriate mathematical material model was created to describe the experimentally obtained specific tissue behavior of the disc.

Anatomy of the human temporomandibular joint

The temporomandibular joints are located at the skull-base in front of the auditory canals (Fig. 1, inset). These joints are so-called diarthrodial joints, which enable large relative movements between separate bones. The integrity of the joints is
maintained by a fibrous capsule having intrinsic ligamentous thickenings and by accessory ligaments. The capsule is slack and the ligaments are only starting to get taut at the extremes of jaw movement and thus have limited influence on the mechanics of normal more or less symmetrical movements (Koolstra and Van Eijden, 1995, 1997). The articular surfaces are covered with thin cartilage layers, which have a very low coefficient of friction (Fung, 1981). Synovial fluid inside the joints is concerned in the maintenance of living cells in the cartilage layers and acts like a lubricant during movement.

The articular surfaces of the temporomandibular joint reside on the mandibular fossa and the articular eminence of the temporal bone above and on the mandibular condyle below (Fig. 1). These surfaces are highly incongruent, which enables the mandible to perform open-close movements, laterodeviations and pro- and retrusions or a combination of them relative to the skull with a high degree of movability (Ostry and Flanagan, 1989). Evidently, the shape and size of the contact areas of the opposing articular surfaces change considerably during jaw movement. Presumably to prevent high peak loading in these contact areas, an additional cartilaginous disc is present inside the joint. It fits like a baseball cap on the

**Figure 1.**
A: Sagittal cross section of the human temporomandibular joint. The disc and joint capsule are represented in blue. A: anterior band, I: intermediate zone, P: posterior band. The articular surfaces of the bones are covered with thin cartilage layers.
B: Three-dimensional artist's impression of the human temporomandibular joint.
Inset: location of the joint with respect to the skull.
mandibular condyle. The disc consists of two thicker regions, referred to as anterior and posterior bands, which are oriented medio-laterally. The bands are separated by a thinner intermediate zone. At the medial and lateral side of the condyle, both bands are attached to the medial and lateral poles of the condyle, ensuring that the disc and condyle move together during anteroposterior translation. When the teeth are in occlusion, the disc is located in the mandibular fossa of the temporal bone and the condyle touches its posterior band. During jaw opening, the condyle moves together with the disc in anterior direction (Fig. 2). Simultaneously, the condyle rotates relative to the disc. When the jaw is in the open position, the condyle is in contact with the disc's anterior band, which is located on the articular eminence in this situation (Rees, 1954). A comprehensive review concerning the anatomy of the temporomandibular joint was published by Piette (1993).

The presence of the cartilaginous disc inside the temporomandibular joint can be expected to influence the mechanical behavior of the joint considerably. The disc is easily deformable which enables it to continuously adapt its shape to fit in the space between the articular surfaces. Therefore, the disc seems a suitable means for
preventing the existence of small contact areas on the opposing articular surfaces. Small contact areas would increase the possibility of high peak loads and thus accelerate the processes causing wear and other damage inside the joint. Furthermore, its internal structure might supply the disc with load bearing capabilities and the opportunity to dissipate energy during impact loading. However due to a lack of research, the various mechanical functions of the disc are still hypothetical.

**Human temporomandibular joint mechanics**

To elucidate the mechanics of the temporomandibular joint, the distribution of the loads and deformations in the structures of the joint have to be determined. Experimental studies have been performed using animal models (e.g., Hylander, 1979; Brehnan et al., 1981; Hohl and Tucek, 1982; Boyd et al., 1990). The number of such experimental studies is limited, because (1) the joint is difficult to reach, (2) the application of experimental devices will alter the mechanics of the joint by introducing damage to its tissues, and (3) the results are hard to extrapolate to the human case.

To overcome these difficulties, others have tried to investigate the mechanics of the temporomandibular joint by means of numerical models, which enable the analysis of internal deformations of its separate structures (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). It was found that the disc has a load distribution function inside the joint. During jaw opening, the disc is able to move together with the condyle in anterior direction even without the presence of ligaments (DeVocht et al., 1996). The studies mentioned were two-dimensional, supposing that the deformations (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. It is questionable whether this simplification is allowed, because the anatomical geometry and the loading (e.g., during laterodeviations) of the joint are truly three-dimensional. To accurately investigate the mechanical behavior of the temporomandibular joint during loading, a numerical model is needed, which is based on the actual three-dimensional anatomy of the joint and which is capable of estimating the deformations of the structures in the joint in all three dimensions. Until now, however, such a model has not been available.
Gaining insight into the mechanics of the temporomandibular joint is also impeded by a limited knowledge of the tissue behavior of its cartilaginous structures. In contrary to other diarthrodial joints, the osseous surfaces inside the temporomandibular joint are covered with fibrocartilage instead of hyaline cartilage (Moffett et al., 1964). Fibrocartilage is softer than hyaline cartilage (Benjamin and Evans, 1990). Nonetheless, cartilage layers on articular surfaces of bones play an important role in the load transmission in diarthrodial joints (Schreppers et al., 1990). Until now, only a few studies have been performed to elucidate the mechanical tissue behavior of the temporomandibular joint disc. Using animal or human material, some elastic (Tanne et al., 1991; Lai et al., 1998) as well as viscoelastic (Teng et al., 1991; Chin et al., 1996; Scapino et al., 1996; Kuboki et al., 1997; Tanaka et al., 1999) properties of the disc have been determined by means of mechanical testing experiments. However, the results of these experiments are not easily applicable. Most experiments were (quasi-) statical, although the physiologic loading of the tissues of the temporomandibular joint is highly dynamical (e.g., talking, chewing). Secondly, the internal structure of the disc, like all biological structures, is inhomogeneous. Therefore, the mechanical behavior of the disc presumably demonstrates regional variations. The location of mechanical testing was mostly varied mediolaterally, while the morphology of the disc varies more in anteroposterior direction (Fig. 1 and Teng and Xu, 1991; Mills et al., 1994; Scapino et al., 1996). For a complete insight into the tissue behavior of the temporomandibular joint disc, additional experiments are needed in which the disc is loaded in a more physiologic manner and the testing locations are varied in anteroposterior direction.

**Finite Element Method**

The mechanical behavior of each deformable structure is described by the relationship between the relative deformation (strain) and the relative load (stress). This relationship is difficult to determine analytically. The finite element method has proven to be a suitable and usable numerical tool for obtaining good approximations for the strain and stress distributions in complex structures (orthopedics: Huiskes and Chao, 1983; diarthrodial joints: Mow et al., 1993; masticatory system: Korioth
and Versluis, 1997; Van Eijden, 2000). In the following, this finite element method is described briefly.

The application of this method involves the subdivision of the volume of each structure into many, simply shaped pieces (elements) with a small yet finite volume (Fig. 3, left). The elements are connected to each other by nodes on their edges and are loaded by internal and external forces acting on these nodes. These nodal forces determine, together with the element’s material behavior, the displacements of the nodes and thus the deformation of each element. Due to its simple geometry, the differential equations describing the mechanics of an element can be transformed into algebraic equations without introducing errors that are unacceptably large. The large amount of elements needed to describe a complex geometry accurately, results in a large set of linear equations which needs to be solved simultaneously. A (super-) computer is needed to find the solutions of all equations. The deformation of the total structure is determined by merging the deformations of the separate elements.

When more than one structure is involved in the model, the possibility exists that contact occurs between the different structures. This phenomenon makes the calculations more complex, because contact results in additional deformations of the structures (MARC, 1996). Because it is unknown where contact occurs between the structures a priori, the total load is applied to the model in many tiny steps (increments). During each increment a numerical check for contact is performed. The

Figure 3.
The finite element method. Left panel: The geometry of the model is subdivided into elements and the boundary conditions and external forces are applied to the nodes. Right panel: After the successful simulation, the geometry is deformed and the magnitude of the calculated stresses is represented by different colors, varying from blue for low stresses to yellow for high stresses.
increments have to be sufficiently small for correctly determining the contact. Each increment is finished successfully, when the solutions converge, i.e., when the difference between the applied force and the estimated reaction force is sufficiently small. When all increments have been finished successfully, the total simulation has been completed. The resulting strain and stress fields can be visualized by means of different colors (Fig. 3, right).

In conclusion, the finite element method can be used as a numerical tool for determining the deformations of various structures in the temporomandibular joint during loading. However, simulations with the finite element method can only generate valuable results, when the model is based on the actual three-dimensional anatomy of the joint and includes a material model accurately describing the complex mechanical behavior of the included biological structures.

**Modeling of anatomical geometry**

To create a numerical model of the temporomandibular joint, a mathematical description of its geometry must be available. The first step is to determine the geometry of its articular surfaces by accurate measurements. Using the electromagnetic tracking devices available nowadays (An et al., 1988; Milne et al., 1996), surfaces of various complexities can be scanned. During this scanning the relevant surface is followed with a stylus. It requires almost no pressure, herewith preventing deformation of soft structures. With these devices, many measurements are obtained in a relatively short time. Using adequate averaging methods, the precision of these measurements is relatively large (An et al., 1988; Milne et al., 1996). Because the measurements are gathered randomly, they are unstructured. This means that the position of a particular measurement is unknown with respect to the other measurements.

The second step involves obtaining a mathematical surface representation from the measurements. Generally, two different methods can be applied for this purpose. Interpolating models connect the measurements with smooth functions (Scherrer and Hillberry, 1979; Ateshian, 1993). Because this method forces the function through the measurements which unequivocally contain noise, the surface description
will be rough and additional calculations are required to smoothen this description. The second method involves approximating models (Ateshian et al., 1991). A specific function, considered suitable for describing the surface, is iteratively adapted by fitting its values through the measurements.

Apart from the spatial coordinates, both methods require the recording of the surface coordinates of each measurement and therefore cannot be applied with the unstructured measurements obtained with devices like the electromagnetic tracking device. Therefore, a new method had to be developed, capable of generating an accurate surface representation from unstructured measurements. After mathematical descriptions of all relevant surfaces had been obtained, they were combined to define three-dimensional volume representations of the anatomical geometry of the structures considered deformable. These volumes were filled with three-dimensional finite elements. Finally, they were put together to form a complete finite element model of the temporomandibular joint.

**Modeling of tissue behavior**

The relationship between the relative loads (stresses) and the relative deformations (strains) inside a structure is determined to a large extent by the material of which that structure consists. Cartilaginous structures, like the cartilage layers on the articular surfaces of the mandibular condyle and temporal bone, and the disc between these surfaces, consist of a network of collagen fibers and proteoglycans. The collagen fibers are formed by three polypeptide chains and are very strong in tension (Roth and Mow, 1980). In the disc, the collagen fibers in the intermediate zone are mainly oriented anteroposteriorly, whereas they have a mediolateral orientation in the anterior and posterior bands (Teng and Xu, 1991; Mills et al., 1994; Scapino et al., 1996; Berkovitz, 2000). Proteoglycans become negatively charged in an aqueous fluid and thus are highly hydrophilic, which means that they attract water. The space between the collagen fibers is filled with an interstitial fluid (water and dissolved electrolytes), which comprises between 60% and 85% of the wet weight (Cohen et al., 1998). Recently, Sindelar et al. (2000) have determined that the average water content of the temporomandibular disc is about 77%. In rest,
tensile forces in the collagen network balance the hydrostatic pressure in the interstitial fluid generated by the hydrophilic character of the proteoglycans. The small permeability of the collagen network (pore size of 10 - 60 Å, Mow et al., 1984, 1993) impedes this fluid to flow through the collagen network. Therefore, the loads acting on cartilaginous structures are initially carried by pressurization of the incompressible fluid without much deformation of the collagen network (Soltz and Ateshian, 1998). This mechanism protects the collagen network against extreme local deformations during impact loading. Fluid flow through the network is nonetheless possible, which leads to a gradual transfer of the load from the fluid to the collagen network. When loaded, the collagen network deforms. This enables the disc to continuously adapt its shape to fit in the space between opposing articular surfaces. In a recent review, Mow et al. (1999) gave an extensive description of the relation between the internal structure and the mechanical behavior of cartilaginous structures. It is clear that the multi-phasic nature of cartilage causes its mechanical behavior to be highly nonlinear and time-dependent.

When the pressure gradients in the fluid have disappeared, the fluid flow through the collagen network stops. This means that the long-term mechanical behavior of cartilaginous structures during statical loading can be approximated by a linear single-phase material model. However, when one's interest is focussed on the short-term or dynamical behavior, a more sophisticated material model is needed. Various theories have been applied to describe the time-dependent nature of the mechanical behavior of cartilaginous structures: e.g., viscoelastic (Kovach, 1996), biphasic (Mow et al., 1980, 1984) or poroelastic (Simon, 1992). Application of the viscoelastic theory seems not preferable for cartilage in general, because it does not include the effect of fluid flow through and out of the structure. The latter two theories assume that the material consists of two inmiscible constituents, which can move with respect to each other. Both theories are equivalent provided that the fluid is inviscid (Simon, 1992). However, none of these theories have yet been applied in finite element simulations of temporomandibular joint loading. To improve the results of future simulations, an effort was made to adequately model the dynamical tissue behavior of the human temporomandibular joint disc found in indentation experiments.
Outline

In order to obtain a finite element model of the human temporomandibular joint of an accurate anatomical geometry, the articular surfaces of the right joint from a male cadaver were scanned using an electromagnetic tracking device (Polhemus). A numerical algorithm was developed to fit polynomial surfaces through the unstructured measurements obtained with this device. This algorithm is the subject of chapter 2. In chapter 3 the accuracy of the procedure of scanning and fitting is described. The development of a three-dimensional finite element model of the temporomandibular joint and its application during static clenching are presented in chapter 4. Initially, the temporomandibular joint disc was considered to be the only deformable structure in the joint. Implementation of deformable cartilage layers enabled the usage of the model to distinguish between the different functions of each cartilaginous structure. In chapter 5 this enhanced model is applied in static loading tasks with different jaw positions. Concerning the dynamical behavior of the tissue of the human temporomandibular joint disc very little is known. Therefore, dynamical indentation experiments were performed on fresh human discs; the results of these experiments are shown in chapter 6. Chapter 7 deals with the acquiring of an accurate material model with appropriate parameters describing the behavior of the disc, which can be applied in finite element models of human temporomandibular joints. Finally, general conclusions drawn from the finite element simulations and experiments are included in chapter 8.
Chapter 2

FITTING PARAMETRIZED POLYNOMIALS
WITH SCATTERED SURFACE DATA

Abstract- Currently used joint-surface models require the measurements to be structured according to a grid. With the currently available tracking devices a large quantity of unstructured surface points can be measured in a relatively short time. In this study a method is presented to fit polynomial functions to three-dimensional unstructured data points. To test the method spherical, cylindrical, parabolic, hyperbolic, exponential, logarithmic, and sellar surfaces with different undulations were used. The resulting polynomials were compared with the original shapes. The results show that even complex joint surfaces can be modeled with polynomial functions. In addition, the influence of noise and the number of data points was also analyzed. From a surface (diameter: 20 mm) which was measured with a precision of 0.2 mm, a model could be constructed with a precision of 0.02 mm.
Introduction

Several mathematical models have been used to describe joint surfaces. All of these methods divide the surface into small quadrangles, called patches. Coons' bicubically blended patches (Scherrer and Hillberry, 1979; Hirokawa, 1991; Hefzy and Yang, 1993) can be used to model the patches individually. They can be used for any shape, but the borders between the patches only have first-order continuity. This makes them unsuitable for curvature analysis. B-splines (Ateshian, 1993; Kwak et al., 1997) offer any-order continuity, but this is accomplished at the cost of some of the flexibility of Coons' patches. Since both methods force the surface through the measured points and measurements always contain noise, these methods create rough surfaces. To overcome this problem the B-splines can be smoothed (Ateshian, 1993).

Another approach is to use a polynomial function as a model for the whole joint surface (knee: Wismans et al., 1980; Blankevoort et al., 1991; carpometacarpal joint: Ateshian et al., 1992). These methods are less suitable for complex shapes. A fifth-degree polynomial \( z = z(x,y) \) was used by Wismans et al. (1980). Ateshian et al. (1992) extended this single function to three parametrized polynomials \( S = S(u,v) = (S_x(u,v), S_y(u,v), S_z(u,v)) \), making it suitable for more complex shapes. The main advantages of these methods are their simplicity, their higher-order continuity and their capability to smooth noisy data.

Most methods mentioned need a grid that has to be applied to the surface, after which the spatial coordinates as well as the grid position are recorded for every grid point. These grid positions are essential for the construction of the surface model. With the magnetic tracking devices now available (An et al., 1988; Milne et al., 1996) a large quantity of surface coordinates can be measured. This results in a large thin cloud of scattered surface points. Since these measurements are done without measuring grid positions, the data are unsuitable for the models mentioned.

This chapter will introduce a method with which scattered surface data are structured and polynomial functions are fitted with the surface data. To test the method a large number of different surface shapes were used. The resulting
polynomials were compared with the original shapes. In addition, the influence of noise and the number of data points was also analyzed.

Materials and Methods

Theory
A common way of structuring surface points is to define a surface-coordinate system, and to assign surface coordinates to the points. Note that this is the same as applying a grid to the surface and to assign grid coordinates to every data point. Since every point on the surface has unique surface coordinates as well as unique spatial coordinates, there is a one-to-one relationship between surface coordinates and spatial coordinates. This means that the spatial coordinates can be written as functions of the surface coordinates, but also that the surface coordinates can be written as functions of the spatial coordinates. If we keep the domains of both functions limited to the surface, one function is the inverse of the other. To obtain an approximation of both functions polynomials will be used. If the three-dimensional position of data point \( n \) is \( S_n = (s_{xn}, s_{yn}, s_{zn}) \), then its surface coordinates are:

\[
\begin{align*}
  u_n &= U(s_{xn}, s_{yn}, s_{zn}) = \sum_{i,j,k=0}^{N_{uv}} p_{ijk} s_{xn}^i s_{yn}^j s_{zn}^k \\
  v_n &= V(s_{xn}, s_{yn}, s_{zn}) = \sum_{i,j,k=0}^{N_{uv}} q_{ijk} s_{xn}^i s_{yn}^j s_{zn}^k
\end{align*}
\]

(1)

Here \( p_{ijk} \) and \( q_{ijk} \) are constants that depend on the shape of the surface, and \( N_{uv} \) is the degree of the polynomials.

Similarly, if \( (u,v) \) are the surface coordinates of a point on the surface, then its spatial coordinates are:

\[
S = (s_x(u,v), s_y(u,v), s_z(u,v)) = \left( \sum_{i,j=0}^{N_{xyz}} a_{ij} u^i v^j, \sum_{i,j=0}^{N_{xyz}} b_{ij} u^i v^j, \sum_{i,j=0}^{N_{xyz}} c_{ij} u^i v^j \right)
\]

(2)

\( N_{xyz} \) is the degree of the polynomials, and the constants \( a_{ij} \), \( b_{ij} \) and \( c_{ij} \) again depend on the shape of the surface. Eq. (2) is the mathematical model of the
surface. Apart from the summation boundaries this is the same model as was used by Ateshian et al. (1992). For all the models in this chapter the degrees of polynomials (1) and (2) were four and eight, respectively.

The unknowns $a_{ij}$, $b_{ij}$, and $c_{ij}$ of Eq. (2) were determined by means of a linear least-squares fit, which was nested in a nonlinear least-squares fit that was used to determine the unknowns $p_{ijk}$ and $q_{ijk}$ in Eq. (1). The result was considered to be acceptable, if the root mean square of the distances between the measured coordinates and the coordinates calculated with Eq. (2) was smaller than 0.01 mm. To improve the stability of the algorithm, the data points were normalized ($-1 < s_{xn}, s_{yn}, s_{zn} < 1$) prior to the fitting procedure. After the procedure was finished, the polynomials were scaled back.

![Figure 1.](image)

Seven original surfaces are shown with below their reconstructions with the polynomial functions. Gray values are used to indicate the distances between the original surface and the reconstruction. The axes of the axis system have a length of 10 mm.
To reconstruct the joint surface, the border of the surface must also be determined in the surface-coordinate system. For this border a grid was constructed on the surface, and each square was tested on the presence of data points using Eq. (1), and was labeled accordingly as 'surface' or 'empty'.

**Test protocol**

The testing had two purposes. Firstly, the grids produced by Eq. (1) had to be checked and, secondly, we wanted to investigate in what way the model precision depends on the shape of the surface, the measurement error, and the number of samples taken from the surface. For the tests seven basic shapes were selected, namely a cylinder, a saddle, a sphere, a paraboloid, a logarithmic shape, an exponential shape, and a hyperboloid (Fig. 1 and Table 1). Measurements of these surfaces were simulated with a computer program. The program generated data points that were homogeneously distributed over the surface (see the appendix). All these shapes were measured with 0.5 samples/mm². Subsequently, the method was applied to the data points. Then, the error was calculated for all the points from the surface grid as the difference between the polynomial functions and the original shape. The peak error was the maximum of the calculated distances and the RMS error the Root Mean Square of these distances. These values were used to determine the precision of the mathematical model.

To analyze the influence of the measurement error and the density of the data points on the precision of the model, measurements were simulated with measurement errors of 0.0, 0.2 and 1.0 mm and densities of 0.5, 1, 2, 5, 10 and 20 samples/mm².

To increase the complexity of the shapes, undulations were introduced to the

<table>
<thead>
<tr>
<th>Name</th>
<th>(X(u,v))</th>
<th>(Y(u,v))</th>
<th>(Z(u,v))</th>
<th>Domain</th>
<th>(E_1(u))</th>
<th>(E_2(v))</th>
<th>Area ((\text{mm}^2))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Paraboloid</td>
<td>(u \sin(v))</td>
<td>(u^2)</td>
<td>(u \cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(u^2 \sqrt{(4u^2+1)})</td>
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<td>580</td>
</tr>
<tr>
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<td>(u \sin(v))</td>
<td>(u^2 \cos(2v))</td>
<td>(u \cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(u \sin(4u^2+1))</td>
<td>1</td>
<td>532</td>
</tr>
<tr>
<td>Hyperboloid</td>
<td>(u \sin(v))</td>
<td>(\sqrt{(u^2+1)})</td>
<td>(u \cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(u \sin(2u^2+1/(u^2+1)))</td>
<td>1</td>
<td>358</td>
</tr>
<tr>
<td>Cylinder</td>
<td>(u \cos(v))</td>
<td>(\sin(v))</td>
<td>(\sin(u)\cos(v))</td>
<td>-1&lt;(u&lt;1); -0.7&lt;(v&lt;0.7\pi)</td>
<td>1</td>
<td>1</td>
<td>875</td>
</tr>
<tr>
<td>Sphere</td>
<td>(\sin(u)\sin(v))</td>
<td>(\cos(u))</td>
<td>(\sin(u)\cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(\sin(u))</td>
<td>1</td>
<td>997</td>
</tr>
<tr>
<td>Exponent</td>
<td>(u \sin(v))</td>
<td>(\exp(-4v^2))</td>
<td>(u \cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(u \sin(6u^4 \exp(-8v^2)+u^2))</td>
<td>1</td>
<td>431</td>
</tr>
<tr>
<td>Logarithm</td>
<td>(u \sin(v))</td>
<td>(0.1 \ln(16u^2+1))</td>
<td>(u \cos(v))</td>
<td>0&lt;(u&lt;1); 0&lt;(v&lt;2\pi)</td>
<td>(u/(16u^2+1))</td>
<td>(u/(16u^2+1))</td>
<td>325</td>
</tr>
</tbody>
</table>

Table 1: Surface definition of the used shapes
cylinder and the sphere. For the amplitude $A$ of the undulations the values -5, -2, 2 and 5 mm were used, and for the undulation width $e$ the values 2, 5 and 10 were used. All these surfaces were measured with densities of 0.5, 1, 2, 5, 10 and 20 samples/mm$^2$.

No exact criteria exist to make a distinction between good and bad surface-coordinate systems. So we limited ourselves to display the surface grids produced by our method as wire frames. The lines of the wire frame represent constant $u$ and constant $v$ levels. It is to be expected that the best grid is a grid in which the quadrangles mostly resemble squares, i.e., all the edges have almost the same length and the angles between the edges are close to $90^\circ$.

For the fitting we used an Intel Pentium computer and Labview for Windows (v4.0, National Instruments). Matlab (v5.1, The Mathworks) was used to reconstruct the surfaces with the results of the fitting program.

![Figure 2.](image)

The dependence of the RMS error on the measurement error and the sample density for the different shapes. The markers, which are connected by straight lines, represent the different simulations.

## Results

Figure 1 depicts the wire frames of the seven basic shapes and the fitted polynomial functions. The shapes were measured with 0.5 data points/mm$^2$. As can be seen from this figure, all the quadrangles in the grid are smoothly shaped in all test cases.
Six of the seven models had a mean error of less than 0.01 mm, which is below the stop criterium used in the fitting program. The logarithmic surface had a mean error of 0.02 mm.

The influence of the noise and number of samples on the RMS error can be seen in Fig. 2. Without any measurement error the RMS error was independent of the number of samples. An increase of the measurement error from 0.2 mm to 1.0 mm led to approximately a five times higher RMS error. An increase of the sample density from 0.5 to 10 samples/mm² resulted coarsely in a four times smaller RMS error. Above 10 samples/mm² the error did not reduce anymore. When the surface was sampled with 10 samples/mm², the error of the mathematical model was roughly ten times less than the measurement error.

In Fig. 3 a few examples of shapes with undulations are shown. In Fig. 4 the peak error of all the simulations with undulations is shown as a function of the undulation amplitude and width. As long as the width of the undulation was larger than its amplitude the error remained low, but when the amplitude became equal or larger...
this error increased fast. In these cases the peak error was always located at the center of the undulation. Sharp undulations were clearly smoothed by the method and in these cases a deterioration of the surface grid was often seen. A possible reason of this smoothing is that polynomials get into problems if a large part of the surface has a constant curvature, while another part has a much higher or lower curvature. To check this hypothesis simulations were performed in which the area of the original sphere or cylinder was decreased. Figure 5 shows the results in case of a sphere. A sharp decrease of the error is the result. The peak errors are also shown in Fig. 4 (median and small). Only with an amplitude of -5 mm and a width of 2 smoothing occurred. Clearly the undulation itself was not the problem, but the size of the spherical or cylindrical surface.

![Figure 4](image)

Figure 4.
Peak errors as a function of the undulation amplitude and width. An amplitude of zero mm indicates that no undulation was applied. The additions "median" and "small" signify that respectively a small part or a large part was removed from the surface (see also Fig. 5).

In Fig. 6 an example of a model of the mandibular fossa and the articular eminence of the human temporomandibular joint is shown. The measurements were done with a magnetic tracking device. The surfaces were measured with approximately 40,000 samples. The RMS error of the device was 0.2 mm. The RMS error of the model could not be calculated, because independently measured high precision surface data for the comparison was not available. A comparison of the model with the data points showed that the smoothing was smaller than the noise. Therefore, it can safely be assumed that the error of the model was much less than 0.2 mm.
Discussion

For the selection of the surfaces used in this study three criteria were used. Firstly, the surfaces had to be continuous. Secondly, we wanted to incorporate a large set of mathematical functions in the surfaces definitions. And thirdly, it had to be possible to generate random data points that were distributed homogeneously over the surface. This last criterium, for instance, caused the paraboloid, hyperboloid, saddle surface, logarithmic surface and exponential surface to be rotational symmetric. Only the paraboloid and the saddle surface can be written as polynomials without making an error. The other surfaces can only be approximated with polynomials. Although the grid of the logarithmic surface looked much better than the grid of the sphere, the sphere was fitted much better by the model. It appeared that logarithmic functions are more difficult to approximate with polynomials. As far as we know there is no mathematically correct way to predict the errors beforehand. So a visual inspection afterwards to detect undershoots and overshoots was always required. If the function gives a good fit, it is not to be expected that a nonhomogeneous distribution of data points will have a negative influence on the result.
Chapter 2

A reconstruction of the mandibular fossa and the articular eminence. The same grid is plotted in three different projections. In (2) seven grid lines are marked (Aa, Bb, Cc, Dd, Ee, Ff and Gg). In (4) these grid lines are projected on a sagittal plane, together with the data points in the immediate surroundings of these lines.

The shape of the surface-coordinate system was determined indirectly by the possibility to fit good polynomials through the data set \((s_x, s_yn, s_2n, u_n, v_n)\). Factors as regularity and rectangularity were not used during the selection of the surface grid. Still, the results show that the method results in surface grids with well shaped quadrangles. An advantage of our method is that the surface grid can be adjusted afterwards. If the polynomials in Eq. (1) do not satisfy, other models (e.g., a transformation to polar coordinates) can be tested.

A major problem was the numerical stability, because a least-squares fit was nested in a nonlinear least-squares fit. For such an algorithm a precision of 14 digits is tight, especially when there are many data points. Much effort was put into the stabilization of the algorithm (gradual increase of the model complexity, selection of proper initial values, normalization of the data). Although all tests described in this chapter converged, in some cases the results would have been better if a higher numerical precision was available. For instance, Fig. 2 clearly shows that measuring more than 10 points/mm\(^2\) did not improve the quality of the surface model. Theoretically, however, the error should continue to decrease with the root of the
Fitting parametrized polynomials

sample density until the same error was reached when no noise was added. Most likely this effect was counteracted by an increase of the numerical problems caused by the extra data points. If the numerical precision was large enough, measurement noise could be eliminated by extra measurements, and the precision of our method would not be limited by the measurement noise.

We also looked at the distribution of the error over the surface. Without noise the error of the saddle and the paraboloid showed a noisy pattern. This is as expected, because these shapes are simple second-degree polynomials. With the other five surfaces, the mathematical model oscillated around the original surface, with the largest errors at the border of the surface. Addition of noise caused larger errors on the whole surface, but near the borders the increase was largest. Since a data point on the border is only surrounded by other points in one direction, it is probably easier to bend the surface towards individual data points. Also due to the noise sharply defined borders change into wide borders with a slowly decreasing density of data points. This also makes it favorable to concentrate the errors on the border.

Appendix

In order to have complete control over the data points a computer program was written to simulate measurements. Since the quality of the mathematical model turned out to depend on the distribution of the data points over the surface (areas with many data points are fitted more precisely than areas with only a few data points), the data points had to be distributed homogeneously over the surface. If $S = (S_x(u,v), S_y(u,v), S_z(u,v))$ is a point on a surface, with partial derivatives $S_u = dS/du$ and $S_v = dS/dv$, then the area $dA$ of an infinitesimal square on this surface defined by the interval $[(u,v),(u+du,v+dv)]$ is given by: $dA = |S_u \times S_v| \, du dv$ ($\times$ means the outer product). If the data points are generated according to the normalized probability distribution $E(u,v) = |S_u \times S_v|/A$, then the data points will be distributed homogeneously over the surface. Using the surface definitions from Table 1, we were able to write the distribution as $E(u,v) = E_1(u) \cdot E_2(v)/A$. Numerical integration of the functions $E_1$ and $E_2$ resulted in the surface areas. The functions $E_1(u)$ and $E_2(v)$ as well as the surface areas $A$ are shown in Table 1. The distribution functions $E_1$
and $E_2$ were used separately by a random generator to generate the surface coordinates $u$ and $v$ respectively. These values were combined to calculate surface points with the surface definitions from Table 1.

It should be noted that the surface area is influenced by the undulations, and thereby the density of the data points. This density decreases as the area of the surface increases.
Chapter 3

THE ACCURACY OF JOINT SURFACE MODELS CONSTRUCTED FROM DATA OBTAINED WITH AN ELECTROMAGNETIC TRACKING DEVICE

Abstract- Electromagnetic tracking devices are widely used in biomechanics. In this study a method is evaluated to construct models of articular surfaces using an electromagnetic tracking device. First, the accuracy of the space tracker was examined and optimized. Then, from several joint surfaces random points were measured and eighth-degree polynomials were fitted through these measurements. To check if the fit converged well, plots of cross sections of the model with corresponding data points were examined. The accuracy of the models was determined by comparing them with computed tomography data and by reproducibility tests. All the fits converged well to the data. The root mean square error of the models varied from 0.07 mm to 0.18 mm, and was proportional to the size and complexity of the surface. This was mainly due to systematic errors made by the space tracker, which were also proportional to the size and complexity of the surface.
Introduction

The general approach to model articular surfaces is to measure the position of a series of surface points, and to construct a mathematical function that matches these points (e.g., Huiskes et al., 1985; Ghosh and Poirier, 1987; Hirokawa, 1991). The instruments used for the measurements were mostly optical (e.g., Ateshian et al., 1994; Blankevoort and Huiskes, 1996) or mechanical (e.g., Scherrer and Hillberry, 1979). An accuracy of less than 0.09 mm (95% confidence level) has been reported for optical measurements (Ateshian et al., 1991). Mechanical instruments are available in a wide range of precisions. Scherrer and Hillberry (1979) reported a precision of 0.01 mm. Electromagnetic instruments are not costly, widely used in biomechanics, and can sample surface points fast. They are, however, seldom used to reconstruct articular surfaces (Hefzy and Yang, 1993). This is probably due to their poor precision, and the lack of a method to compensate for this inaccuracy. Generally a root mean square (RMS) error of 1.5 mm or worse is reported by the specifications, but this can be improved (An et al., 1988; Zoghi et al., 1992; Luo et al., 1996; Milne et al., 1996); with a standard normal error distribution a RMS error of 1.5 mm corresponds to an error of 3 mm (95% confidence level).

The mathematical models can be subdivided into interpolating and approximating models. Interpolating models connect the data points with smooth functions. Examples are Coons’ blended patches (Scherrer and Hillberry, 1979) and basic splines (Ateshian, 1993). Extra calculations are required when they are used with noisy data. Approximating models start with a general function, containing several constants. For every surface the constants are determined by fitting the function to the data points (Ateshian et al., 1991). A problem of this method is that beforehand it is unknown if a general function will converge well enough. This may lead to systematic errors such as smoothing of small undulations and sharp edges. Recently, we developed a method to fit polynomial functions with random surface points (chapter 2 of this thesis). By this method, data is automatically filtered and the accuracy of the mathematical reconstructions can be improved by increasing the number of data points. This makes the method very suitable for high-density
Accuracy polynomial surface models

Accurate measurements (i.e., the distance between the data points is smaller than the noise) can easily be obtained with an electromagnetic tracking device.

In the present study, the accuracy of that method applied to several articular surfaces was determined. Data points were measured with an electromagnetic tracking device. First, the precision of the tracking device was analyzed in more detail. Then a number of articular surfaces was measured, and models were fitted to these measurements. Finally, the precision of the models was evaluated by comparing one of the models with a micro-computed tomography (CT) scan of a surface and by analyzing the reproducibility of the method.

Material and Methods

Accuracy of the tracking device

The 3SPACE® FASTRAK™ System (Polhemus Inc., USA) was used to measure all surfaces. This instrument measures the three-dimensional position of the tip of a stylus and transfers these coordinates to the computer. The surfaces were measured by moving the tip of this stylus over the surface while continuously recording its position with a frequency of 30 Hz.

The accuracy of the instrument can be enhanced by limiting the measurement space (An et al., 1988; Luo et al., 1996; Milne et al., 1996; Bull and Amis, 1997). To find the optimal spatial volume for the measurements, a number of tests was done with a Plexiglas cylinder and a Plexiglas spherical cavity. The size of the resulting volume was 50×100×100 mm³, which was in agreement with the region used by Luo et al. (1996). Furthermore, the offset of the tip of the stylus along the axis of the stylus was calibrated. This was done by fitting cylinders and spheres with variable radii through the data points. The mean difference between the fitted radii and the real radii was used to correct the offset of the stylus. The precision of the instrument for the measurement of surfaces in the optimal domain was determined by measuring the cylinder and the spherical cavity at 14 and 8 different positions, respectively. Mathematical models of these shapes with the previously mentioned dimensions were fitted to the data points, and finally the distances between the data...
points and the fitted surfaces were calculated. The RMS of these distances was used as an estimate for the precision of the system when measuring surfaces.

**Construction of articular surface models**

The surfaces of the tibiofemoral joint (femur: condylus medialis, condylus lateralis, facies patellaris; tibia: condylus medialis, condylus lateralis) and the shoulder joint (cavitas glenoidalis, caput humeri) obtained from a human cadaver were used for the testing. In addition, the mandibular condyle (caput mandibulae) of a dried skull was used. The articular surfaces of the femur and the tibia were measured and modeled separately. In order to know the positions of these different surfaces relative to each other, a set of reference points was applied to the bones. Before the joint was opened, the positions of these points were measured relative to each other. Later the surfaces were measured relative to these reference points. This way the different articular surfaces of a joint could be positioned relative to each other. The number of random data points ranged from 3,000 to 11,000, depending on the size of the articular surface. The scanning of a surface varied from 2 to 10 minutes.

The method described in chapter 2 of this thesis was used to construct models of the surfaces. In this method a surface \( S \) is modeled by eighth-degree polynomial functions

\[
S = \left( \sum_{i,j=0}^{8} a_{ij} u^i v^j, \sum_{i,j=0}^{8} b_{ij} u^i v^j, \sum_{i,j=0}^{8} c_{ij} u^i v^j \right)
\]

Here \( u \) and \( v \) are the parameters of the surface, and \( a_{ij}, b_{ij}, \) and \( c_{ij} \) the constants that determine the shape of the surface. The constants are determined iteratively with a nonlinear least-squares fit. The fitting error was defined as the RMS of the distances between the data points and the surface. These distances were not calculated exactly, but estimated with a special algorithm. This algorithm had a tendency to overestimate the distances. To check the convergence of the fit, plots of cross sections of the model were visually compared with data points from the cross sections.

**Precision of the surface models**

The caput mandibulae was also measured with micro-CT (Rüegsegger et al., 1996) with a voxel size of 34x34x34 \( \mu \text{m}^3 \). From this scan the cranialmost part, including the
Accuracy polynomial surface models

Figure 1.
A perpendicular view on two cross sections of a fitted cylinder (continuous line) and the data points in this cross section (dots). The right panel illustrates the worst example of systematic deviations of the data points from the cylindrical shape, the left panel the best example.

Articular area, was selected using an oblique cutting plane, after which the surface voxels were extracted. For each voxel with five or less neighboring voxels (surface voxels) the three-dimensional position was calculated, and the set of voxels was positioned such that it matched the polynomial surface optimally. For this purpose, a transformation consisting of the three rotation angles and the three translation distances was calculated such that the difference between surface voxels and the polynomial surface was minimal. This difference was defined as the mean of the distances of all the individual surface voxels to the polynomial surface.

Reproducibility tests were done on the surfaces of the caput mandibulae and the cavitas glenoidalis. The caput mandibulae was measured twice at one position, and a few days later twice at another position. The cavitas glenoidalis was measured at three different positions. The surface models derived from these measurements were compared in pairs. The difference between two models was calculated by extracting approximately 2000 points from the first model, and calculating the RMS distance from these points to the second model with the method described for the CT data. Statistically, subtraction of two surfaces with an error of \( \sigma \) gives zero vectors, with an error of \( \sigma \sqrt{2} \). Therefore, we divided the RMS of the distances with \( \sqrt{2} \) to obtain an estimation of the error of the model. The reproducibility tests will not reveal any convergence problems; such problems only depend on the shape of the surface and not on the location where the surface was measured.
Chapter 3

Figure 2.
The models of the surfaces of the knee joint seen from different directions. The cross sections in the bottom panel together with many others were visually inspected to check the quality of the fit. In the two middle panels the positions of the cross sections are indicated. (Inf = inferior, Sup = superior, Med = medial, Lat = lateral).

Results

The measurements of the cylinder and spherical cavity had a precision of 0.16 mm and 0.08 mm, respectively. The visual comparison of the data points with the fitted surfaces revealed that at some positions in the measurement space the systematic errors in the measurements of the cylinder were larger than the random errors (Fig. 1). No systematic errors were found for the spherical cavity.

In Fig. 2 the reconstructed articular surfaces of the knee joint including a few cross sections together with the corresponding data points are shown. The average fitting errors of the femoral surfaces were 0.26 mm (condylus lateralis), 0.19 mm (condylus medialis), and 0.12 mm (facades patellaris), and of the tibial surfaces 0.15 mm (condylus lateralis), and 0.12 mm (condylus medialis). The cross sections in Fig.
2, as well as many others analyzed, showed good convergence of the fit. The reconstructions of the articular surfaces of the shoulder joint are shown in Fig. 3. The fit error of the caput humeri was 0.17 mm, and the average fit error of the cavitas glenoidalis was 0.20 mm. Again good convergence of the fit was observed in the cross sections. The reproducibility tests showed that the differences between the models of the cavitas glenoidalis were 0.06, 0.09 and 0.08 mm for the three pairs, respectively.

Figure 4 shows the results of the first measurement of the caput mandibulae. The fit errors were 0.13, 0.18, 0.11 and 0.13 mm for the four measurements, respectively, and again the cross sections showed good convergence. The differences between the models of the caput mandibulae measured at the same position at different times were 0.07 mm and 0.08 mm for the two different positions tested. The mean of the differences of the four combinations measured at different positions was 0.07 mm,
indicating an error of 0.05 mm. The differences between the four models and the CT scan were 0.07, 0.07, 0.06 and 0.08 mm, respectively. Also shown in Fig. 4 are cross sections of the surface model with points extracted from the CT scan. The largest difference was observed in the cross section marked “e”; this cross section, however, included a part of the surface area that did not belong to the articular area. A similar artifact was found with the other three models of the caput mandibulae.

**Figure 4.**
A and B show the surface of the caput mandibulae from two directions. C and D show cross sections of the surface as indicated in 4B. In C these cross sections are accompanied with the surface points extracted from the CT scan. In D they are accompanied with the data points measured with the space tracker. (Sup = superior, Lat = lateral, Med = medial).

**Discussion**

**Precision of the tracking device**

The error of the space tracker consists of two components: a systematic error and a random error. The systematic error appeared to be proportional to the size of the surface (especially the range of angles needed to measure a surface). In that case the cylinder reflects the worst case (it fits hardly in the selected region and the stylus
Accuracy of polynomial surface models

has to be rotated over 360 degrees) and the spherical cavity the best case. Assuming that for the spherical cavity the systematic errors can be neglected, then the random error can be calculated to be 0.08 mm, and the systematic error to be maximally 0.14 mm. So the total error can be estimated to range from 0.08 mm to 0.16 mm. A table with corrections for the systematic error will improve the precision of the space tracker considerably.

**Precision of the models**

The precision of the surface models is determined by the goodness of fit, and errors of the tracking device. To check the fit, the models were visually compared with the data points. With this comparison the maximal difference was estimated to be 0.08 mm (the random error of the data points). So the average smoothing is less than 0.08 mm. The random error of the tracking device is filtered by the usage of a least-squares fit, and its contribution to the model precision is substantially less than 0.08 mm (chapter 2). The systematic error of the tracking device ranged from almost 0.00 mm for small surfaces to 0.14 mm for large surfaces. So the total error in the models can be estimated to range from much less than 0.11 (=$\sqrt{0.08^2+0.08^2}$) mm for small surfaces to about 0.18 (=$\sqrt{0.14^2+0.08^2+0.08^2}$) mm for large surfaces. These estimations are in good agreement with the results presented. With reference to the CT data the precision of the models of the caput mandibulae ranged from 0.06 to 0.08 mm. The repetition tests gave a precision of 0.05 mm. For the cavitas glenoidalis the repetition tests showed the same result.

Until now only Scherrer and Hillberry (1979) did an extensive analysis of the precision of their surface models. They measured the scapula of a dog and interpolated the data with Coons' bicubic patches. The precision of their measurements was 0.01 mm while their model had a precision of 0.05 mm (RMS). So in fact their method was less accurate than their measurements. The main advantage of electromagnetic tracking devices is that they can collect surface points with a high frequency. In the present study a method is described that eliminates the main drawback of these devices, i.e., their poor precision. It is shown that the models, which were fitted through the random measurements, can have a higher accuracy than the measurements themselves.
Chapter 4

THREE-DIMENSIONAL FINITE ELEMENT ANALYSIS OF THE HUMAN TEMPOROMANDIBULAR JOINT DISC

Abstract- A three-dimensional finite element model of the cartilaginous disc of the human temporomandibular joint has been developed. The geometry of the articular cartilage and cartilaginous disc surfaces in the joint was measured using a magnetic tracking device. First, polynomial functions were fitted through the coordinates of these scattered measurements. Next, the polynomial description was transformed into a triangulated description to allow application of an automatic mesher. Finally, a finite element mesh of the disc was created by filling the geometry with tetrahedral elements. The articular surfaces of the mandible and skull were modeled by quadrilateral patches. The finite element mesh and the patches were combined to create a three-dimensional model in which unrestricted sliding of the disc between the articular surfaces was allowed. Simulation of statical joint loading at the closed jaw position predicted that the stress and strain distributions were located primarily in the intermediate zone of the disc with the highest values in the lateral part. Furthermore, it was predicted that considerable deformations occurred for relatively small joint loads and that relatively large variations in the direction of joint loading had little influence on the distribution of the deformations.
Introduction

The human mandible is connected to the skull by two temporomandibular joints. The articular surfaces of these joints are highly incongruent, which provides the mandible with a large degree of movability with respect to the skull (Ostry and Flanagan, 1989; Koolstra and Van Eijden, 1999). Between the articular surfaces a cartilaginous disc is situated, which is assumed to decrease the contact pressure by increasing the contact area between the incongruent joint surfaces, similar to the menisci in the knee joint (Scapino et al., 1996).

Experimental as well as analytic studies have demonstrated, that the human temporomandibular joint is loaded during masticatory function (Hatcher et al., 1986; Smith et al., 1986; Faulkner et al., 1987; Koolstra et al., 1988; Ferrario and Sforza, 1994; Throckmorton and Dechow, 1994). Development and degeneration of joint tissues and overloading caused by parafunctions like bruxism, are supposedly influenced by these loads (Moffet et al., 1964; O'Ryan and Epker, 1984; Nickel et al., 1988; McCormack and Mansour, 1998; Newberry et al., 1998). However, detailed data about the distribution of the loads are still lacking, which means that the main causes of these processes mentioned cannot be fully understood.

Experimental studies regarding the distribution of the loads in the temporomandibular joint have been performed in animal models (e.g., Hylander, 1979; Brehnan et al., 1981; Hohl and Tucek, 1982; Boyd et al., 1990). The number of experimental studies is limited, because the joint is difficult to reach and the application of experimental devices, such as strain gauges, inside the joint will introduce damage to its structures, which will influence their mechanical behavior.

Mathematical models of the human masticatory system including the temporomandibular joint have been demonstrated to be a powerful tool to predict the loads acting on this joint. Many studies, however, have oversimplified the geometry of the articular surfaces and assumed them as being rigid (Koolstra et al., 1988; Ferrario and Sforza, 1994). Therefore, the deformations and the distribution of loads inside these structures could not be analyzed.
The finite element method has been proven to be a suitable tool for approximating such mechanical quantities in structures with a complex geometry (Huiskes and Chao, 1983). Few finite element analyses of the temporomandibular joint including a movable disc have been published (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). These analyses, however, were limited to the two-dimensional sagittal plane and thus unsuitable to investigate, for example, the influence of variations of the loading direction out of the sagittal plane or the development of peak stresses located medially or laterally.

The purpose of the present study was to develop a three-dimensional finite element model of the human temporomandibular joint in which unrestricted sliding of a deformable temporomandibular joint disc between the articular surfaces was allowed. This model was used to investigate the three-dimensional load distribution in the disc during statical loading tasks. The results might contribute to a better understanding of the normal and abnormal functioning of the disc.

Materials and methods

Geometry

The geometry of the model was obtained from the right temporomandibular joint of an embalmed male cadaver (age: 77 years), showing no abnormalities, using a magnetic tracking device (Polhemus 3SPACE Digitizer). To avoid disturbing the geometry of the soft structures by applying force with this device during the measurements, tight fitting plaster casts of the articular surfaces were made. To ensure that the separate measurements of the casts could be combined into one joint model afterwards, four reference points were measured on both the mandible and the skull before separation. Thereafter, the articular capsule was removed and the mandible was separated from the skull. This way the articular surface of the mandibular condyle and the lower surface of the temporomandibular joint disc, which remained in contact with the articular surface of the skull, were revealed and casts could be made. After removing the disc from the preparation the articular surface on the skull (fossa plus eminence) was revealed and a third cast could be made. The
casts were sampled in a random sequence with about 10,000 points with a frequency of 30 Hz and an accuracy of about 0.1 mm (chapter 3).

The articular surfaces were reconstructed from the unstructured measurements according to the method described in chapter 2 and 3 of this thesis. Briefly, this method presumes that the spatial coordinates of the surfaces could be approximated by eighth-order polynomial functions of their surface coordinates. The coefficients of these functions were determined simultaneously using a nonlinear least-squares optimization method, by minimizing the difference between the measured spatial coordinates and their equivalent in the polynomial approximation. This reconstruction procedure filtered the measurement noise by a factor ten due to the large amount of measurements. The resulting reconstructions of the articular surfaces on the condyle and the skull are displayed in Fig. 1.

![Figure 1](image)

**Figure 1.**

The polynomial reconstructions of the articular surfaces of the condyle and of the skull (mandibular fossa and articular eminence) in the human temporomandibular joint. Both panels display a frontal view of the reconstructions. To improve clarity the line width on the border was increased (sup = superior, inf = inferior, med = medial, lat = lateral).

**Finite element mesh**

Removal of the temporomandibular joint disc from the articular surface of the skull deformed its geometry in such a way that direct measurement of its top surface was unreliable. This limited direct measurement of the disc geometry to its bottom surface. In a closed jaw position, the top surface of the disc is situated against the articular surface of the skull. Therefore, the geometric data of the latter surface were used to represent this top surface.

Due to the complex geometry of the disc, an automatic mesher (Mentat 3.2, MARC Analysis Research Corporation, Palo Alto, USA) was needed to create the
finite element mesh. To be able to apply this mesher, the polynomial descriptions of the surfaces were transformed to triangular patches. This was performed by applying a two-dimensional Delaunay algorithm (Lee and Schachter, 1980; Cavendish et al., 1985; Chew, 1989) upon the two-dimensional surface coordinates of the reconstructed surfaces. After transformation of the patches to the three-dimensional space, occasionally occurring surface distortions (see Fig. 2) were corrected by applying the following procedures. Pairs of adjoining triangles were redivided according to the shortest diagonal. Edges larger than 1.7 times the mean edge length were subdivided. The new nodes were placed on the surface using the polynomial description. The resulting triangulated reconstructions of the surfaces are displayed in Fig. 3. The borders of the top and bottom surfaces of the disc were then connected to each other by triangular patches of the same size as those on the surfaces themselves. After removal of a small region on the lateral side of the disc that appeared to be too thin to allow the creation of a valid mesh, the geometry of the disc was totally surrounded by 3,028 two-dimensional triangular patches and was subsequently meshed with three-dimensional linear hexahedral elements. The mesh consisted of about 16,000 elements. The joint surfaces on the condyle and the skull were modeled by quadrilateral patches.

![Figure 2.](image)

A: typical example of possible deterioration of the surface representation after transformation from two-dimensional surface coordinates to three-dimensional spatial coordinates. The dotted line identifies the mathematical shape of the surface.
B: surface representation after correction by the smoothing procedures mentioned in the text.

The local incongruencies due to the flat patches used in the triangular description, caused point contacts generating large stress gradients and thus resulting in poor convergence. Therefore, an initial adaptation step was performed by moving the articular surface of the condyle towards the articular surface of the skull over a
distance according to the incongruencies. During this operation, the Young's modulus of the disc was temporarily decreased considerably. This way, the disc regained its congruency with the articular surfaces. Thereafter, the strains and stresses were neutralized and the disc was remeshed. This way, an initial state with conditions necessary for a proper use of a contact algorithm was obtained. Figure 4 displays the finite element model in its initial state.

**Simulations**

In order to investigate load distributions and deformations in the temporomandibular joint disc during joint loading, simulations were performed using the commercially available finite element software MARC K6.2 (MARC Analysis Research Corporation, Palo Alto, USA). This software was installed on the Cray C916 National Supercomputer at SARA in Amsterdam. Because large deformations were expected to occur, the Cauchy stresses and the logarithmic strains were calculated using the finite deformation theory (Marc, 1996). The contact phenomenon occurring at the condyle-disc and the disc-fossa interfaces, was solved using the Direct Constraint Method, in which direct constraints are placed on the motion of a body in contact using boundary conditions -kinematic constraints as well as nodal forces (MARC, 1996).

The material behavior of the disc was assumed to be linearly elastic. The values of the Young's modulus and the Poisson's ratio applied were 6 MPa and 0.40, respectively. Due to lack of consensus in the literature and to allow comparison with
other studies, the value of the Young's modulus was chosen between the values used by other investigators (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). A relatively low value was chosen for the Poisson's ratio, because the fluid inside the disc was supposed to play a minor role in determining its mechanical behavior in (quasi-) static situations. The joint surfaces on the condyle and the skull were assumed to be rigid and the joint capsule was not included. Furthermore, no friction between the disc and the articular surfaces was applied (Mow et al., 1993; Nickel and McLachlan, 1994).

In a configuration concomitant with the jaw in a closed position, the condyle was displaced in a direction according to the estimated direction of the joint reaction force, i.e. a line originating from the center of the condyle and perpendicular to the joint surface and to the axis through the poles of the condyle (Koolstra and Van Eijden, 1992). During the simulations, the articular surface of the skull was fixed.

Four additional simulations were performed in which the direction of the movement was changed with respect to the reference simulation over 30° in anterior, medial, lateral and posterior direction, respectively. The results of these variations will yield information about the sensitivity of loading direction on the stress and strain distributions in the disc. The influence of the material properties of the disc on the

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**Figure 4.**

Three-dimensional lateral view of the finite element model of the right human temporomandibular joint. The articular surfaces of the condyle and the skull are represented by quadrilateral patches and the geometry of the disc is filled with approximately 16,000 tetrahedral elements.
load distribution in the joint was investigated by increasing the value of the Young's modulus by a factor 10, and by applying values of 0.30 and 0.47 for the Poisson's ratio. Finally, to investigate the convergence of the simulations, the elements in the region that was mostly deformed in the reference simulation were transformed from linear (containing four nodes) to quadratic (ten nodes) tetrahedral ones.

Results

Reference simulation

When the joint was loaded in a direction according to the estimated joint reaction force, a displacement of 0.20 mm was simulated. This distance corresponded with almost half the thickness of the so-called intermediate zone of the temporomandibular joint disc (thickness: 0.46 mm). In this reference simulation, a force of 35.5 N was acting on the condyle. This force was calculated by summation of the nodal forces that were generated by the contact algorithm. In Fig. 5 the resulting stress distribution (Von Mises) on the surface of the disc is displayed. The disc appeared to be loaded predominantly in the intermediate zone. The largest values of the stresses were located in the lateral part of this zone. The strains were not as concentrated as the stresses, but were also mainly located in the intermediate

Figure 5.
Distribution of the Von Mises stress ($\sigma_{VM}$) after nine increments of joint loading in a frontal and inferior view of the temporomandibular joint disc. The stress is concentrated in the lateral part of the thin (=intermediate) zone of the disc (sup = superior, inf = inferior, med = medial, lat = lateral).
zone. The maximum value of the equivalent elastic strain was 0.44.

**Variation of the direction**

A change in the direction of joint loading over 30° had little influence on the distribution of strains and stresses in the disc (Fig. 6). Only in the cases with a laterally or medially directed load, a small shift of the strain pattern in the concomitant direction could be noticed visually. However, the direction influenced the amount of displacement that was needed to obtain a similar amount of deformation. This displacement was smallest (0.18 mm) when the direction of the movement was changed over 30° in anterior direction and largest (0.34 mm) when the direction was changed in posterior direction.

![Figure 6](image)

Influence of the direction of the displacement of the condyle on the distribution of the equivalent elastic strain ($e_{eq}$) in the temporomandibular joint disc. The displacement was directed 30° anteriorly (A), posteriorly (B), medially (C), or laterally (D) with respect to the reference simulation.

**Variation of material properties**

A force of 327 N was acting on the condyle after a displacement of 0.2 mm, when the Young's modulus had a value of 60 MPa. The predicted strains did not differ
substantially from the strains predicted in the reference simulation. The predicted stresses increased proportionally with the Young's modulus.

By comparing the results of the simulations obtained by application of the three different Poisson's ratio values, we observed that the predicted force acting on the condyle was proportional to the applied Poisson's ratio, when the same amount of displacement was simulated.

**Mesh accuracy**

In Fig. 7 the Von Mises stress at node 1512 is displayed as a function of the increment number. This node was located on the top surface of the disc in the middle of the region which bore the highest loads. In this area elements were transformed from linear to quadratic in order to check the convergence. As can be seen, the Von Mises stress at this node predicted in the model containing quadratic elements, increased similarly with the joint load as in the original model. The amount of stress at the final increment differed by only 0.36% from the reference simulation.

![Figure 7](image)

**Figure 7.**

The Von Mises stress ($\sigma_{VM}$) at the location of node 1512 as a function of the increment number. The results obtained using quadratic elements (striped line) do not differ very much from those using linear elements (solid line).
Discussion

The present model, as far as we know, is the first three-dimensional finite element model of the temporomandibular joint disc. The disc was shaped according to its anatomical geometry which was sampled with high resolution. The transition zone between the fibrous attachment of the superior head of the lateral pterygoid muscle and the disc was also included in the present model. This probably explains why the disc of the present model and that reported by other researchers (e.g., Chen and Xu, 1994) had a thicker appearance anteriorly. The disc was allowed to move unrestrictedly between the articular surfaces of the condyle and the skull, only kept in place by the contact occurring between the disc and the surfaces.

Although movement of the disc was not impeded by ligaments or friction, the disc remained in the fossa during loading in this position. Variation of the loading direction over 60° led to only small changes in the stress and strain distributions. Therefore, it can be concluded that during clenching, the highest deformations are always located in the same region. The sketches in Fig. 8 visualize the possible reason why a large variation of loading direction has little influence on the deformation of the disc. Whatever the loading direction, the deformation of the intermediate zone remains large compared to the other regions in the disc. Fig. 6 might suggest that during

![Image](image_url)

**Figure 8.**

Sagittal sketches of the human temporomandibular joint. Left panel: anterior variation of the direction of the displacement over 30° compared to the reference direction. Right panel: posterior variation of the direction over 30° compared to the same reference direction. It is shown that the deformation of the intermediate zone remains large compared to other regions of the disc.
various, both symmetric as well as asymmetric, loading conditions in a closed jaw position, large stress concentrations occur predominantly in the lateral side of the intermediate zone of the disc. This is supported by a dissection study by Werner et al. (1991), who reported that wear leading to perforations was mainly located in this region.

Thus far, no experimental measurements of stress and strain distributions in the loaded disc have become available. Therefore, it is yet impossible to compare our predictions with such data. It is also still unknown whether the disc is preloaded when the jaw is in the closed position. Therefore, we assumed the initial loads to be zero and our results should be interpreted as relative values with respect to the initial configuration.

Comparison of the results with previous finite element analyses (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998) is difficult, because these simulations differ substantially from those described in the present study. The studies were two-dimensional and did not describe the strain distributions. Furthermore, they mutually differed in the direction and the amount of the displacement applied to the condyle, and in the mechanical properties of the disc. Although the Von Mises stress is not a reliable representative of the mechanical stress in a structure when the deformations are very large, it was nevertheless calculated in the present study to allow comparison with the studies mentioned. When we applied a Young’s modulus according to Chen and Xu (1994), we obtained almost similar results. DeVocht et al. (1996) prescribed a displacement that was much larger than the ones applied in the studies by Chen and Xu (1994) and Chen et al. (1998) and the present one. While this displacement was almost parallel to the articular surface of the skull only small deformations of the disc occurred.

**Model assumptions**

The present finite element model was based on the geometry of the right temporomandibular joint of only one male cadaver. This particular joint did not deviate from normal morphology and thus the model can be expected to yield generally applicable results. Due to the complexity of the joint, several additional assumptions had to be made to ensure convergence of the model.
Comparison with micro-computed tomography data indicated that the mandibular condyle could be reconstructed with a root mean square error of about 0.07 mm (chapter 3). An extremely different accuracy of the reconstructions of the other relevant articular surfaces is not to be expected. The question remains whether the applied eighth-order polynomials were adequate to reconstruct the articular surfaces of the temporomandibular joint accurately. Visual inspection ensured us that the reconstructions did not show detectable aberrations.

The convergence of the simulations was investigated by application of higher-order (quadratic) tetrahedral elements in the region that was deformed mostly in the reference simulation. While the desired mechanical quantities are related to the displacements of the nodes, such higher-order elements lead to more accurate results. The results obtained using quadratic elements differed by only 0.36% from those obtained using linear elements. Therefore, we concluded that mesh refinement was not necessary. On the other hand, the results showed relatively large deformations of the disc upon joint loading. Loading of the joint beyond the values presented here caused divergence of the calculations due to severe deterioration of the elements. Therefore, refinement of the mesh in this particular region would obviously lead to more accurate results.

It has been demonstrated that the mechanical behavior of cartilage is time-dependent, which can be modeled as viscoelastic (Kovach et al., 1996) or biphasic (Mow et al., 1984). Regarding the temporomandibular joint disc, details about these characteristics are still lacking. Therefore, it was considered inappropriate to simulate dynamic situations like jaw opening where the condyle moves together with the disc in anterior direction along the articular eminence. The present study was principally aimed at the (quasi-) statical behavior of the disc.

The model did not include deformable cartilage layers on the articular surfaces of mandible and skull, while we were principally aiming for the mechanical behavior of the temporomandibular joint disc. It has been suggested that the cartilage layers in joints may play an important role in the load transmission during joint loading (Schreppers et al., 1990). According to the two-dimensional study by Chen et al. (1998) the implementation of cartilage layers in the model had little influence on the stress distribution in the disc. The joint capsule was also neglected in our model. The
study by DeVocht et al. (1996) showed that even during jaw opening, the discal attachments had little influence on the mechanical behavior of the disc.
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; Koriith 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 (Mentat 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.
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 \( \varepsilon_{\text{eq}} = \sqrt{3/2 \varepsilon_{ij}^{\text{ij}}} \), 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.](image)

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 $\varepsilon_{eli}$. 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-url)

_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 $\varepsilon_{eli}$."

**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.
Table 1.
Contact areas and the maximum values for the equivalent elastic strain (\(\varepsilon_{eq}\)) and the Von Mises stress (\(\sigma_{VM}\)) estimated at a condylar load of 1 N. The contact areas are given for two different stress values.

**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
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
Cartilaginous deformation in various joint configurations in a frontal view. A: jaw-closed configuration. B: retruded jaw-open configuration. C: protrusive 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 protrusive position. The joint was loaded in the same direction as in the protrusive configuration. The strain distributions in the temporal cartilage and the disc were unchanged compared to the protrusive 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
FE analysis cartilaginous structures in TMJ

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.
DYNAMICAL PROPERTIES OF THE HUMAN TEMPOROMANDIBULAR JOINT DISC

Abstract- The cartilaginous intra-articular disc of the human temporomandibular joint, shows clear anteroposterior variations in its morphology. However, anteroposterior variations in its tissue behavior have not been investigated thoroughly. To test the hypothesis that the mechanical properties of fresh human temporomandibular discs vary in anteroposterior direction, dynamical indentation tests were performed at three anteroposteriorly different locations. The disc showed strong viscoelastic behavior dependent on the amplitude and frequency of the indentation, the location, and time. The resistance against deformations and the shock absorbing capabilities were larger in the intermediate zone than in regions located more anteriorly and posteriorly. Because a number of studies have predicted that the intermediate zone is the predominantly loaded region of the disc, it can be concluded that the topological variations in its tissue behavior enable the disc to effectively combine the functions of load distribution and shock absorption.
Introduction

The articular surfaces of the temporomandibular joint are highly incongruent (Ostry and Flanagan, 1989). Due to this incongruency, the contact areas of the opposing articular surfaces are very small. Upon joint loading this may lead to large peak loads, which may cause damage to the cartilage layers on these articular surfaces. The joint contains a cartilaginous disc that prevents these peak loads (chapter 4 and 5 of this thesis). Besides load distribution, other functions have also been attributed to the disc, such as shock absorption (Tanaka et al., 1999) and lubrication (Nickel and McLachlan, 1994). Biomechanical quantities like stress and strain are considered as initial factors for changes in the morphology of cartilaginous structures, like for example changes in the amount of proteoglycans (Mao et al., 1998; Quinn et al., 1998) and collagen (Tuominen et al., 1996) in the solid matrix of these structures. In the temporomandibular joint, such changes might play a role in the development of temporomandibular disorders. However, the exact contribution of the disc to the mechanics of the joint is still poorly understood. This is partly caused by a lack of knowledge about its mechanical characteristics.

Until now, only a few studies have been performed to elucidate the mechanical tissue behavior of the disc. Tanne et al. (1991) and Lai et al. (1998) approached the disc as a structure which has linearly elastic behavior. On the other hand, numerous studies have reported that cartilaginous tissues display viscoelastic characteristics. This time-dependent behavior is assumed to be caused by the movement of interstitial fluid through the matrix of macromolecules (e.g., collagen and proteoglycans) of these tissues (Kovach, 1996; Fithian et al., 1990; for a review: Mow et al., 1984). Therefore, the viscoelastic behavior of the disc has also been investigated (Teng et al., 1991; Chin et al., 1996; Scapino et al., 1996; Kuboki et al., 1997; Tanaka et al., 1999). Using quasi-static experimental set-ups, they were only able to determine linear viscoelastic behavior. It might be expected, though, that the disc, like other soft tissues, exhibits nonlinear tissue behavior including a dependency on frequency of load.
Various cartilaginous structures have been demonstrated to exhibit regionally different tissue behavior (Athanasiou et al., 1995; Arokoski et al., 1999). In the temporomandibular joint disc, such variations in tissue behavior have been investigated mainly in mediolateral direction (Tanne et al., 1991; Chin et al., 1996; Lai et al., 1998). Macroscopically, however, the morphology of the disc varies more in anteroposterior direction than in mediolateral direction. The so-called intermediate zone, for example, is much thinner than its adjacent anterior and posterior bands. During movement of the mandible, the disc moves predominantly in anteroposterior direction relative to the articular eminence of the temporal bone and the mandibular condyle. Therefore, it may be expected that anteroposterior variations in the characteristics of the disc are of greater influence on the mechanics of the joint than variations in mediolateral direction. Anteroposterior variations have been demonstrated to exist in canine discs for mediolaterally applied tensile loads (Teng et al., 1991).

The purpose of the present study was to determine the biomechanical tissue behavior of the human temporomandibular joint disc. Therefore, the resistance against deformation and the energy dissipation capabilities of fresh discs were measured during dynamic loading conditions. The discs were subjected to sinusoidal compression cycles of different frequency and amplitude. To test the hypothesis that the mechanical properties vary in anteroposterior direction, each disc was tested at three anteroposteriorly different locations.

Materials and methods

Specimen preparation

Seven discs from seven fresh human cadavers (five male and two female, age: 73-86 years) with no known history of temporomandibular disorders were used to study the mechanical properties. The use of human discs conforms to a written protocol that was reviewed and approved by the Department of Anatomy and Embryology of the Academic Medical Center of the University of Amsterdam. The discs were carefully dissected within 48 hours after death, rapidly frozen in liquid nitrogen
cooled isopentane, and stored at -80°C until further processing. Generally, they did not show any signs of arthritis or other degenerative diseases, with the following exceptions: one specimen had a lateral perforation, another specimen appeared to be superficially roughened in its posterior band biased laterally, and a third one had a sphere-shaped hardening (calcification?) in its posterior band. All three affected regions had a diameter of less than 5 mm. Care was taken to exclude these regions from testing.

**Experimental apparatus**

![Figure 1. Schematic drawing of the experimental set-up. The temporomandibular joint disc is glued between two indenters and submerged in a saline solution. The displacement of the top indenter relative to the bottom indenter is measured by means of two linear variable differential transducers (LVDT). A load cell was used to measure the force exerted to the top indenter.](image)

Indentation experiments were performed using a modified universal mechanical testing machine (Hounsfield, Salfords Surrey UK) (Dauvillier et al., 2000). The whole disc (dimensions (mean ± SD): anteroposterior: 18.4 ± 3.4 mm; mediolateral: 23.0 ± 2.1 mm) under examination was placed between two cylindrical indenters with a diameter of 3.94 mm. In Fig. 1 a schematic representation of the experimental set-up is shown. The disc was compressed by a cyclic displacement of the top indenter. The testing machine was capable of generating a cyclic movement that differed less than 1.5% from a true mathematical sine for all amplitude-frequency combinations.
applied in this study. To measure this displacement relative to the bottom indenter, two linear variable differential transducers (LVDT; Mahr, millitron type 1202; range: ±2000 μm and resolution: ±1 μm) were used. They were positioned at each side of the specimen and the measurements of both were averaged. The force exerted to the top indenter was recorded using a load cell (Hounsfield 1000 N). The combination of load cell and data acquisition equipment had a resolution of 0.025 N. During the experiments, the disc was immersed in phosphate buffered saline (0.9% NaCl, pH 7.0). The temperature of this solution was kept at about 37.7°C, monitored using a thermocouple. The data acquisition was performed using custom-made software, which was also used for controlling the displacement of the top indenter. The measurements were analysed offline using Labview (version 4; National Instruments, USA) and Matlab (version 5.1; Mathworks, USA). Deformation of the disc was defined by its strain $\varepsilon$ ($\varepsilon=\Delta h/h_0$) as the displacement of the top indenter ($\Delta h$) relative to the initial thickness of the disc between the indenters ($h_0$). The load was defined by its stress $\sigma$ ($\sigma=F/A$) as the force $F$ relative to the cross-sectional area of the indenter ($A$). The area enclosed by the stress-strain curve of each loading cycle was calculated to determine the dissipated energy per unit of volume.

**Measurement protocol**

One hour before the start of the experiments, the disc was slowly thawed. To prevent slippage, it was glued between both indenters using cyanoacrylate adhesive (Permacol, Ede, the Netherlands) under a small compressive load (<2 N). Thereafter, it was submerged in saline and allowed to equilibrate for 5 minutes without any load. Then, the thickness of the disc portion between the indenters was determined with a digital vernier caliper with a resolution of ±0.01 mm. The displacements of the top indenter were measured relative to this unloaded thickness. To determine the regional differences in the behavior of the temporomandibular joint disc all discs were tested at three different locations, namely at the anterior band, the intermediate zone and the posterior band. Care was taken to prevent the different locations from overlapping by varying the locations in mediolateral direction.

To elucidate the time-dependency of the behavior of the disc, each experiment consisted of ten sinusoidal compression cycles. To investigate the influence of the velocity of deformation, indentation frequencies of 0.02 Hz, 0.05 Hz, and 0.1 Hz were
applied. The amplitude of the indentation was varied relative to the unloaded thickness of the disc portion between the indenters in such a way that a maximum compressive strain between 20% and 40% was obtained. After each experiment of ten compression cycles, relaxation of the disc was allowed for 300 s. Then a control experiment consisting of one sinusoidal compression cycle with a frequency of 0.05 Hz and a strain of 30% was performed, followed by another 120 s relaxation. This control cycle was repeated after each experiment to check for sufficient relaxation. Hereafter, the next experiment of ten compression cycles with another frequency-amplitude combination was started. After all the amplitude-frequency combinations were performed at the same location, the disc was carefully removed from the indenters. The disc was then tested at another location by the same protocol. Remains of glue were used as an indication for a processed location to prevent the next locations from overlapping. The order of the variations in location, frequency and amplitude was randomized.

Figure 2.
Measurement signals of a typical experiment. The disc was subjected to ten compression cycles with a strain amplitude of about 30% and a frequency of 0.05 Hz.
A: Displacement (in μm) and strain vs. time (in s).
B: Force (in N) and stress (in MPa) vs. time (in s).
Results

In Fig. 2 typical time series of the measured quantities are displayed. The measurements of the displacement of the top indenter showed a fairly true sinusoidal shape. The shape of the force measurement curve, on the contrary, showed a sharp tip near maximum compression and was very broad at low compression. Also, a small phase shift was present between the displacement and force signals; the force reached its maximum earlier than the displacement. In subsequent cycles the value of the maximum force decreased asymptotically, but equilibrium was hardly ever achieved after ten cycles.

![Figure 3](image-url)

**Figure 3.**

Characteristic stress-strain relationships for tissue of the anterior band (A), intermediate zone (B) and the posterior band (C) of a human temporomandibular joint disc for ten successive indentation cycles. The indentation frequency was 0.05 Hz and the strain amplitude was 30%. The resistance against deformation starts building up at about 10% strain for the intermediate zone, at about 15% strain for the anterior band, and at about 20% strain for the posterior band. The stress increases superproportionally. The hysteresis loops show that energy is dissipated inside the disc. The maximal stress as well as the amount of energy dissipated decrease in subsequent cycles.

**Maximal stress**

The stress-strain curves obtained from ten successive compression cycles at three different locations all showed a clear hysteresis (Fig. 3). For small strains the disc hardly showed any resistance against compression. Then the stress increased superproportionally towards a maximum, which was reached before the total compression was completed. This maximum stress was largest in the intermediate zone. During unloading the stress disappeared almost instantly. The stress response was the strongest during the first cycle of each experiment. In subsequent cycles this
response diminished monotonically. After ten compression cycles the maximum stress was decreased by about 50% on average. Roughly the overall stiffness of the intermediate zone was two and three times as large as that of the anterior and posterior band, respectively.

Figure 4A shows the average maximum stress in the first cycle as a function of the deformation frequency and amplitude for three different locations. The standard

Figure 4.

Means and standard deviations (top bars) of the maximum stress (A) and energy dissipation (B) in the first cycle as functions of indentation frequency and amplitude for three different locations. Grey: intermediate zone. Dots: anterior band. Stripes: posterior band. Because some discs were tested at strain amplitudes not shown in the graphs, not all bars are based on the same number of discs (N = 4.4 ± 1.1). The intermediate zone exhibits different tissue behavior compared to the anterior and posterior bands. Asterisks: significance of difference in normalized maximal stress or energy dissipation between the relevant region and the intermediate zone (*: p<0.1; **: p<0.05) as tested with a paired Student's T-test. The indentation amplitude and frequency (although to a smaller extent) both have a proportional effect on the value of the maximal stress and the amount of energy dissipation.
deviation bars indicate that there was a large variation between the specimens. This figure clearly shows that the average maximal stress in the first cycle was much larger in the intermediate zone than in the anterior and posterior bands. Application of the Friedman test showed that the maximum stress in the intermediate zone, irrespective of the applied test protocol, was significantly larger than that in the anterior and posterior band (p<0.05). Furthermore, there was a tendency that the maximal stress is proportionally dependent on the strain and (to a smaller extent) on the frequency.

*Energy dissipation*

The hysteresis in the stress-strain curves of ten successive indentation cycles indicates that energy was dissipated in the disc (Fig. 3). Comparison of the curves of the successive compression cycles indicates that the amount of energy dissipated in the disc asymptotically decreased in time. After 10 cycles the amount of energy dissipation ranged between 20% to 40% of the initial amount. Figure 4B clearly shows that on average the amount of energy dissipated in the first indentation cycle was larger in the intermediate zone than in the anterior and posterior bands. Application of the Friedman test showed that the over-all energy dissipation in the intermediate zone was significantly larger than that in the anterior and posterior band (p<0.05). Furthermore, the amount of energy dissipation increased with the amount of strain. A frequency dependency of the amount of energy dissipation was not evident.

**Discussion**

**Measurements**

The mechanical properties of the human temporomandibular joint disc were obtained by performing cyclic compression experiments. Experimental as well as numerical studies have shown that during masticatory function the joint is loaded, which means that the disc is compressed between the mandibular condyle and the articular
eminence of the temporal bone (e.g., Koolstra and Van Eijden, 1995; Throckmorton and Dechow, 1994). Although the disc slides along the articular eminence during jaw opening, the loading of the disc by shear was considered to be negligible due to the very low friction (Nickel and McLachlan, 1994). The loads acting on the disc during rest were considered to be negligible. Therefore, a dynamical compressive set-up without preload was chosen. The upper limit of the indentation frequencies (0.1 Hz) was determined by the capabilities of the mechanical testing machine.

**Tissue behavior**

The present results demonstrate that the tissue behavior of the temporomandibular joint disc is largely dependent on the amount of deformation, the location, and time. Due to the large inter-species variations and different experimental protocols applied in previous studies (Teng et al., 1991; Tanne et al., 1991; Tanaka et al., 1999), a direct comparison with these studies cannot be made. Nonetheless, the present results can be qualitatively compared with the results from other studies. The observed superproportional stress-strain relationship has also been described by Teng et al. (1991) and Tanne et al. (1991) using tensile tests, and by Tanaka et al. (1999) using compression experiments. Teng et al. (1991) also found that the tensile elastic modulus in mediolateral direction of the intermediate zone differed significantly from the equivalent moduli of the anterior and posterior bands. However, in contrast to the results of the present study in which the disc was compressed in superoinferior direction, they found that the anterior and posterior bands were stiffer than the intermediate zone. This apparent inconsistency is indicative for the anisotropic nature of the temporomandibular joint disc.

Recent studies have discussed a possible shock absorbing function of intra-articular discs facilitated by a large amount of proteoglycans combined with a low permeability (e.g., Fithian et al., 1990; Tanaka et al., 1999). The results of the present study indicate that some of the energy needed to compress the disc is not released during unloading. Therefore, the disc possesses the capability to dissipate energy, which is advantageous during biting to suppress the effects of sudden impact or unloading when crushing brittle food. The results from the successive indentation cycles show that the amount of energy dissipation in subsequent cycles was reduced to about 50% of the original amount after 1 or 2 cycles and to about
30% after ten cycles (Fig. 3). This indicates that the return of the fluid squeezed out of the disc portion under loading is relatively slow. Because of this weak recovery and the long relaxation times associated, the loaded disc portion becomes relatively soft when it is cyclically loaded, during for example chewing and speaking.

Wear of the temporomandibular joint disc has been found primarily in its anterior band (Kopp, 1976) or intermediate region (Werner et al., 1991). The latter is in agreement with various studies predicting that the disc is predominantly loaded in its intermediate zone (DeVocht et al., 1996; Nagahara et al., 1999; chapter 4 and 5). In contrast, it has also been predicted that the anterior region receives the largest loads (Tanaka et al., 1994). The presently observed relatively large resistance against deformations of the intermediate zone, however, will be advantageous when the disc is loaded in this zone by shifting the distribution of the deformations to regions with a lower resistance. In this way a larger part of the disc may deform which will increase the contact areas between the articular surfaces (see chapter 4 and 5) and thus prevent large loads acting locally and herewith the risk for damaging.

**Histological interpretation**

The tissue behavior of the temporomandibular joint disc is determined by its internal composition and structure. Histological studies indicate that the tissue of the disc contains very few elastic fibers in its intermediate zone (Gross et al., 1999) and that the density of collagenous fiber bundles in this region is greater than in the anterior and posterior regions (Mills et al., 1994). These collagen bundles are mostly oriented in anteroposterior direction in the intermediate zone and mediolaterally in the anterior and posterior bands (Teng and Xu, 1991; Mills et al., 1994; Scapino et al., 1996). This internal orientation is confirmed by the different values for the mediolateral and anteroposterior elastic moduli found by Teng et al. (1991) and Tanne et al. (1991), respectively. The amount of proteoglycans in the temporomandibular joint disc is less than in hyaline cartilage (Mills et al., 1994). Therefore, it is suggested that the restorative swelling pressure in the disc is less. This suggestion supports the findings of the present study concerning the weak recovery during unloading and the relatively long relaxation times.

The larger resistance against compression and larger energy dissipation capabilities of the intermediate zone can be attributed to an increased collagen
density compared to the other regions. During compression the fluid flow inside the
disc can be assumed to be directed predominantly anteroposteriorly, guided by the
equally oriented collagen bundles. Due to the mediolaterally oriented collagen
bundles in the anterior and posterior bands, the squeezing of the fluid through the
disc is assumed to be inhibited in the anteroposterior direction. This phenomenon
might provide the intermediate zone with increased energy dissipation capabilities
despite its low proteoglycan concentration. The elastic fibres in the disc are
supposed to play only a minor role in restoring the original shape of the disc after
removal of the loads (Scapino et al., 1996).

Static loading decreases the proteoglycan synthesis in cartilaginous structures
whereas dynamic loading is positively related to this synthesis (Quinn et al., 1998).
Changes in the internal structure of cartilage, like changes in the amount of
proteoglycans, affect its mechanical properties. Therefore, as opposed to dynamic
loading like during talking and chewing, static loading of the temporomandibular joint
like during clenching and grinding might lead to degeneration of the cartilaginous
structures involved. This degeneration negatively affects the stiffness or the
dissipation capabilities of the temporomandibular joint disc which eventually might
lead to temporomandibular joint disorders.
Chapter 7

HUMAN TEMPOROMANDIBULAR JOINT DISC
CARTILAGE AS A POROELASTIC MATERIAL

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-) static 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 static or quasi-static nature. However, the behavior of cartilaginous tissues during static 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
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:

\[
E_{tg} = c3\varepsilon_{eq}^3 + c2\varepsilon_{eq}^2 + c1\varepsilon_{eq} + c0
\]

in which:

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

This hyperelasticity was implemented in the poroelastic material model through a user subroutine. The applied values for \(c3, c2, c1,\) and \(c0\) 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.

![Figure 3](image)

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.](image)

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 $k_{ref}=7.5 \times 10^{-15}$ 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
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.
Chapter 8

SUMMARY AND CONCLUSIONS

In the studies described in this thesis, mechanical modeling techniques were developed and applied to investigate the mechanics of the cartilaginous structures in the human temporomandibular joint.

Background

The articular surfaces of the temporomandibular joint reside on the temporal bone above and on the mandibular condyle below. These surfaces are highly incongruent, which enables the mandible to perform a wide variety of movements. Evidently, the shape and size of the contact areas of the opposing articular surfaces change considerably during jaw movement. Presumably to prevent high local peak loading in the contact areas that otherwise would be small due to the incongruency, an additional cartilaginous disc is present.

During functioning the joint is loaded causing deformations in various structures in the joint. These loads and deformations are assumed to play a dominant role in adaptation and degeneration processes. Therefore, detailed information about these loads and deformations is crucial for improving our insights into these processes. However, the joint is inaccessible for direct measurements. To overcome this inability, loads and deformations have to be estimated by application of biomechanical modeling techniques. The finite element method has proven to be a valuable numerical tool for obtaining adequate estimations of the distribution of loads and deformations in complex structures. Therefore, this method was applied in studies described in this thesis. In the development of a reliable finite element model four separate processes can be distinguished. Firstly, the acquisition of the geometry of the separate structures, secondly, the mathematical reconstruction of this
geometry, thirdly, the filling of the geometry of each structure with tiny elements, and fourthly, the acquisition and incorporation of the relevant material properties.

Various methods have been presented in the literature to obtain a mathematical description of articular surfaces of human joints (patches: Scherrer and Hillberry, 1979; Hirokawa, 1991; Hefzy and Yang, 1993; Ateshian, 1993; Kwak et al., 1997; polynomials: Wismans et al., 1980; Blankevoort et al., 1991; Ateshian et al., 1992). Unfortunately, these methods were unsuitable to apply in the present study for assessing the geometry of the articular surfaces of the human temporomandibular joint. Therefore, one of the purposes of the present study was to develop a method for adequate mathematical reconstruction of the relevant articular surfaces.

The present finite element models of the human temporomandibular joint are two-dimensional (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). While these were considered inadequate to assess the expected three-dimensional distribution of joint loads and deformations, a three-dimensional finite element model was developed and applied in this study. It was based on the actual three-dimensional anatomy of a human temporomandibular joint and capable of estimating deformations in the cartilaginous structures in the joint in all three dimensions.

The present knowledge concerning the material properties of the temporomandibular joint disc is based on (quasi-) statical experiments (Tanne et al., 1991; Teng et al., 1991; Chin et al., 1996; Scapino et al., 1996; Kuboki et al., 1997; Lai et al., 1998; Tanaka et al., 1999). However, the physiologic loading of the temporomandibular joint is highly dynamical (e.g., talking, chewing). Therefore, in this thesis experiments are described in which the material properties of the disc were obtained for more physiologic conditions. These experiments were also simulated numerically to determine an adequate material model.

**Geometric modeling**

In the present study, the geometry of the relevant articular surfaces of the right temporomandibular joint of a human cadaver was scanned with an electromagnetic tracking device. At this stage a surface is represented by a large collection of
unstructured surface points. This means that the location of a particular measurement is a priori unknown with respect to the others.

An algorithm was developed that iteratively fitted polynomial functions through such a set of unstructured surface points. It was capable of generating accurate reconstructions of mathematical surfaces like spherical, cylindrical, hyperbolic, exponential, logarithmic, and sellar surfaces. It appeared to improve the accuracy of the surface representations with an increase of the number of surface points (chapter 2). This algorithm was applied and tested for the reconstruction of articular surfaces of various human joints like the knee, shoulder and temporomandibular joint. The maximum root mean square error of the reconstructed surfaces was about 0.18 mm. This error was dependent on the size and complexity of the surface. Comparison of reconstructions of the mandibular condyle reconstructed with this algorithm and from micro-computed tomography scans (voxel size: $34 \times 34 \times 34 \, \mu m^3$) indicated a root mean square error of about 0.07 mm (chapter 3).

**Biomechanical modeling**

The articular surfaces of the temporomandibular joint were mathematically reconstructed by fitting eighth-order polynomials through measured surface points. The volumes of the deformable cartilaginous structures (cartilage layers on the articular surfaces and disc) were synthesized by filling these surface reconstructions with tiny tetrahedral elements. The temporomandibular joint disc was modeled as a separate structure, which could slide unrestrictedly between the articular surfaces of the mandibular condyle and the temporal bone. The finite element model was applied in statical loading conditions comparable with clenching, while this is considered one of the predominant risk factors for joint overload.

The results of the simulations showed that the temporomandibular joint disc had a clear load distribution function and that it was mainly loaded in its intermediate zone. The load distribution capability of the disc appeared to be proportional to the value of its Young's modulus. Variation of the loading direction revealed that the distribution of the deformations in the disc was hardly influenced by the loading direction, when the jaw was closed (chapter 4). The predominantly deformed area shifted from the
entire intermediate zone when the condyle was located in the mandibular fossa of the temporal bone (jaw closed) to the lateral side of the intermediate zone when the condyle was located on the articular eminence of the temporal bone (jaw open). It was also found that the cartilage layers on the osseous structures in the joint increased the effectiveness of the load distribution function of the disc (chapter 5).

Tissue behavior modeling

Biological structures in general exhibit a nonlinear and time-dependent tissue behavior. Presumably this also concerns the cartilaginous structures in the human temporomandibular joint, although quantitatively little is known about their behavior. Therefore, in the present study dynamical indentation experiments were performed with fresh human temporomandibular joint discs. In these experiments, the amplitude, frequency, and location of the cyclic indentation were varied. This allowed for identification of the nonlinearity, the time-dependency, and the heterogeneity of the tissue behavior of the disc. It was shown that the resistance of the disc against deformation is superproportional to its deformation. This resistance was in the intermediate zone of the disc roughly two and three times larger than in its anterior and posterior band, respectively. While the behavior during loading was different from the behavior during unloading, energy was dissipated in the disc. The resistance against deformation as well as the amount of energy dissipation asymptotically decreased in time. These characteristics make the disc suitable for absorbing impact loading as well as for allowing smooth cyclic movements (chapter 6).

When interest is focused on the short-time mechanics or the dynamical behavior of the temporomandibular joint, this nonlinear and time-dependent tissue behavior needs to be included in the joint model. Therefore, the obtained characteristics had to be incorporated in a suitable material model. Because cartilaginous structures consist of a collagen network surrounded by interstitial fluid, it was obvious to choose a material model that incorporates both constituents. These conditions were met in a so-called poroelastic model. By modeling the collagen network as hyperelastic, the
nonlinear elasticity that the temporomandibular joint disc exhibits, could be approximated (chapter 7).

Conclusions

- The complex geometry of the articular surfaces in the human temporomandibular joint can be accurately reconstructed by eighth-order polynomials.
- Polynomial reconstructions of the articular surfaces are applicable to create a three-dimensional finite element model of the temporomandibular joint for biomechanical analysis.
- Finite element simulations indicate that the temporomandibular joint disc and the cartilage layers on the articular surfaces play an important role in the distribution of loads.
- During statical loading the disc is mainly deformed in its intermediate zone; when the jaw is closed the region with the largest deformations is located in the central region of this zone and when the jaw is opened it shifts to the lateral side.
- Dynamical indentation experiments demonstrate that the tissue behavior of the temporomandibular joint disc is nonlinear and time-dependent (viscoelastic).
- The dynamical material properties of the disc are heterogeneously distributed. The intermediate zone is two to three times stiffer than the anterior and posterior bands. Also more energy is dissipated in the intermediate zone than in the anterior and posterior bands.
- When the disc is cyclically loaded, as during, for example, chewing and speaking, it becomes increasingly softer. By this softening the contact areas between the articular surfaces will increase and thus prevent loads acting locally and herewith the risk for damaging.
- The disc is stiffest and absorbs the largest amount of energy during the first of a number of loading cycles. This makes the disc suitable for absorbing impact loading, that might occur during, for example, crushing a nut.
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- The dynamical material properties of the disc can be approximated adequately by a poroelastic model.
SAMENVATTING EN CONCLUSIES

In het onderzoek dat in dit proefschrift wordt beschreven is een aantal mathematische modellen ontwikkeld en toegepast om het mechanische gedrag van de kraakbeenstructuren in het menselijke kaakgewricht te bestuderen.

Achtergrond

De gewrichtsoppervlakken van het kaakgewricht bestaan uit kraakbeen en bevinden zich onderop het slaapbeen van de schedel en bovenop het kopje van de onderkaak. Deze oppervlakken zijn ongelijkvormig en dit draagt bij aan de relatief grote bewegingsmogelijkheden van de onderkaak. Tijdens bewegingen van de onderkaak zullen de vorm en grootte van het contactoppervlak tussen de beide gewrichtsoppervlakken aanzienlijk kunnen veranderen. Tussen de gewrichtsoppervlakken is een extra kraakbeenschijf aanwezig, de zogenaamde discus. Door de aanwezigheid van deze discus wordt vermoedelijk voorkomen dat er te hoge lokale piekbelastingen in het gewricht optreden.

Tijdens bijvoorbeeld kauwen wordt het kaakgewricht belast, waardoor de kraakbeenstructuren zullen vervormen. De optredende belastingen en vervormingen spelen waarschijnlijk een belangrijke rol bij adaptatie- en degeneratieprocessen in het gewricht. Om deze processen beter te kunnen begrijpen is een gedetailleerde kennis over deze belastingen en vervormingen nodig. Het gewricht is echter niet toegankelijk voor directe metingen. Daarom moet er van modelleringsmogelijkheden gebruik worden gemaakt om de grootte en de plaats van de belastingen en vervormingen in de kraakbeenlagen op de gewrichtsoppervlakken en in de discus te kunnen benaderen. De eindige elementen methode is zo’n techniek. Deze methode wordt dan ook toegepast in dit proefschrift. Om een betrouwbare eindige elementen model te kunnen ontwikkelen moeten vier achtereenvolgende stappen worden


De momenteel beschikbare eindige elementen modellen van het menselijke kaakgewricht zijn tweedimensionaal (Chen and Xu, 1994; DeVocht et al., 1996; Chen et al., 1998). Ze zijn ongeschikt om de driedimensionale verdeling van belastingen en vervormingen in de kraakbeenstructuren van het gewricht te bestuderen. Daarom werd een driedimensionaal eindige elementen model ontwikkeld. Het model werd gebaseerd op de werkelijke driedimensionale bouw van een menselijk kaakgewricht.

De huidige kennis omtrent de materiaaleigenschappen van de discus in het kaakgewricht is gebaseerd op (quasi-) statische experimenten (Tanne et al., 1991; Teng et al., 1991; Chin et al., 1996; Scapino et al., 1996; Kuboki et al., 1997; Lai et al., 1998; Tanaka et al., 1999). De normale belasting van het kaakgewricht is echter meestal dynamisch (bijvoorbeeld tijdens praten en kauwen). Daarom werden in het onderzoek dat in dit proefschrift beschreven wordt experimenten verricht waarmee de dynamische materiaaleigenschappen van de discus konden worden bepaald. Deze experimenten werden vervolgens numeriek gesimuleerd om zo een geschikt materiaalmodel te verkrijgen, dat in de toekomst eventueel aan een dynamisch eindige elementen model van het kaakgewricht toegevoegd kan worden.
Modellering van de geometrie

In het huidige onderzoek werd de geometrie van de gewrichtsoppervlakken van het rechter kaakgewricht van een menselijk stoffelijk overschot afgetast met een elektromagnetische meetpen. Zo werd voor elk oppervlak een grote verzameling van ongeordende punten verkregen. Er werd een algoritme ontwikkeld dat iteratief polynomen door de verzameling ongeordende punten kon passen. Met deze methode konden nauwkeurige reconstructies van bolvormige, cilindrische, hyperbolische, exponentiële, logaritmische en zadelvormige oppervlakken gemaakt worden. De nauwkeurigheid van de methode nam toe met het aantal meetpunten (hoofdstuk 2). Dit algoritme werd ook getest op de gewrichtsoppervlakken van een aantal menselijke gewrichten, zoals knie-, schouder- en kaakgewricht. De maximale gemiddelde fout van de gereconstrueerde oppervlakken bleek ongeveer 0.18 mm en was afhankelijk van de grootte en complexiteit van het oppervlak. Voor het kaakkopje werden de aldus verkregen reconstructies vergeleken met reconstructies verkregen met behulp van micro-CT (micro-computed tomography; voxel-grootte: $34\times34\times34 \, \mu m^3$). Deze vergelijking leverde een gemiddelde fout op van ongeveer 0.07 mm (hoofdstuk 3).

Biomechanische modellering

De gewrichtsoppervlakken van het kaakgewricht werden wiskundig gereconstrueerd met behulp van achtste ordre polynomen. Vervolgens werd het kaakbeen van de gewrichtsoppervlakken en de discus gevuld met kleine piramidelvormige elementen. De discus werd als een aparte structuur gomodelleerd, waardoor deze ongehinderd tussen de gewrichtsoppervlakken kon glijden. De berekeningen met het eindige elementen model werden verricht voor statische belastingen, vergelijkbaar met die tijdens klemmen. Statisch klemmen wordt namelijk als een van de voornaamste risicofactoren voor overbelasting van het kaakgewricht beschouwd.

Uit de simulaties bleek dat de discus een belangrijke rol speelt bij het verdelen van de belastingen over de gewrichtsoppervlakken. De discus werd vooral in het middendeel, in de zogenaamde intermediaire zone, vervormd. De verdeling van de
vervormingen in de discus was nauwelijks afhankelijk van de belastingsrichting (hoofdstuk 4). Tijdens een kaakopening verschoof het gebied in de discus dat het meest vervormd werd naar de buitenkant van de intermediaire zone. Behalve de discus bleken ook de kraakbeenlagen op de gewrichtsoppervlakken een positieve bijdrage te leveren aan het verdelen van de belastingen in het gewricht (hoofdstuk 5).

**Modellering van het weefselgedrag**

In kwantitatief opzicht is er weinig bekend over de materiaaleigenschappen van kraakbeen. Daarom werd in het huidige onderzoek de dynamische relatie tussen spanning en vervorming onderzocht. Hiervoor werden experimenten verricht aan verse menselijke discs die op verschillende plaatsen cyclisch werden samengedrukt met variërende amplitude en frequentie. Hierdoor kon de niet-lineariteit, tijdsafhankelijkheid en de heterogeniteit in het gedrag van de discus onderzocht worden. De weerstand die de discus tegen de opgelegde vervorming bood bleek meer dan evenredig toe te nemen met de grootte van deze vervorming. De weerstand was ook niet overal hetzelfde in de discus. In de intermediaire zone was deze twee tot drie keer groter dan in de zogenaamde voorste en achterste band. Tijdens een samendrukkingscyclus kwam er energie vrij uit de discus. Zowel de grootte van de weerstand tegen vervorming als de hoeveelheid vrijkomende energie namen asymptotisch af in de tijd. Deze eigenschappen maken de discus buitengewoon geschikt voor het absorberen van impulsbelastingen en het soepel doen geleden van cyclische bewegingen (hoofdstuk 6).

Conclusies

- De complexe geometrie van de gewrichtsoppervlakken in het menselijk kaakgewricht kan nauwkeurig worden gereconstrueerd met behulp van achtste orde polynomen.
- Polynoomreconstructies van de gewrichtsoppervlakken kunnen in een driedimensionaal eindige elementen model van het kaakgewricht worden toegepast.
- Eindige elementen simulaties tonen aan dat de discus en de kraakbeenlagen op de gewrichtsoppervlakken van het kaakgewricht een belangrijke rol spelen bij het verdelen van belastingen.
- De discus wordt tijdens statisch klemmen vooral vervormd in de intermediaire zone. De vervorming is bij een gesloten kaak het grootst in het centrale gebied van deze zone en dit gebied is bij een geopende kaak meer naar de buitenkant verschoven.
- Dynamische compressie-experimenten tonen aan dat het gedrag van de discus niet-lineair en tijd afhankelijk (visco-elastisch) is.
- Deze dynamische materiaaleigenschappen zijn heterogeen verdeeld in de discus. De intermediaire zone is twee tot drie keer stijver dan de voorste en achterste band. Ook komt er meer energie vrij in de intermediaire zone dan in de voorste en achterste band.
- Wanneer de discus cyclisch wordt belast, zoals bijvoorbeeld tijdens kauwen en praten, wordt deze steeds zachter. Hierdoor wordt het contactoppervlak tussen de gewrichtsoppervlakken steeds groter, waardoor lokale piekbelastingen en mogelijk schade worden voorkomen.
- Bij cyclische belastingen is de discus tijdens de eerste compressie-cyclus het stijfst en absorbeert deze de meeste energie. Hierdoor is de discus geschikt voor het absorberen van impulsbelastingen, zoals bijvoorbeeld bij het doorbijten van een noot.
- De dynamische materiaaleigenschappen van de discus kunnen benaderd worden met behulp van een poro-elastisch materiaalmodel.
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