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ORIGINAL ARTICLE

Stress distribution in the temporo-mandibular joint discs during jaw closing: a high-resolution three-dimensional finite-element model analysis

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Abstract
Purpose This study aims at analysing the stresses distribution in the temporomandibular joint (TMJ) using a complete high-resolution finite element model (FE Model). This model is used here to analyse the stresses distribution in the discs during a closing jaw cycle. In the end, this model enables the prediction of the stress evolution in the TMJ disc submitted to various loadings induced by mandibular trauma, surgery or parafunction.
Materials and methods The geometric data for the model were obtained from MRI and CT scans images of a healthy male patient. Surface and volume meshes were successively obtained using a 3D image segmentation software (AMIRA®). Bone components of skull and mandible, both of joint discs, temporomandibular capsules and ligaments and dental arches were meshed as separate bodies. The volume meshes were transferred to the FE analysis software (FORGE®). Material properties were assigned for each region. Boundary conditions for closing jaw simulations were represented by different load directions of jaws muscles. The von Mises stresses distribution in both joint discs during closing conditions was analyzed.
Results The pattern of von Mises stresses in the TMJ discs is non-symmetric and changed continuously during jaw movement. Maximal stress is reached on the surface disc in areas in contact with others bodies.
Conclusions The three-dimension finite element model of masticatory system will make it possible to simulate different conditions that appear to be important in the cascade of events leading to joint damage.

Keywords Temporomandibular joint · Biomechanics · Finite element analysis

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Introduction

Several situations as trauma, mandibular surgery [19], mandibular distraction osteogenesis [18] generate anatomical modifications in the temporomandibular joint (TMJ) discs. The resulting stress modifications can generate a temporary or permanently disc dislocation with degeneration and may cause severe oral and facial pain or masticatory dysfunctions [19]. Sometimes, the use of experimental devices is required in the mandible system to treat specific pathologies such as osteogenesis distraction. Few human experimental studies have been conducted on the masticatory system because its structures are difficult to reach for mechanical evaluation [5]. The use of these experimental devices inside the mandible structure may introduce damage to its tissues, which interferes with normal function and influences their mechanical behaviour. Finite element (FE) models have thus been extensively used to analyse the stress distribution in the TMJ because it allows studies in areas that are difficult to access without risks on a living subject of investigations [2, 9, 13]. FE analysis must account for the loading conditions in the

TMJ, since it has been suggested that they play a significant role in aetiology of TMJ disorders. Most of previous models assigned material parameters without reflecting differences between all the anatomical regions of the masticatory system. Most of the time, these models use a simplistic representation of maxilla, mandible and teeth. This representation has to be improved to optimize the assessment of physiological and pathologic strains in the joint disc. Mechanical parameters of the different bodies used to model the masticatory system also need to be accurately identified in order to achieve an optimal finite element analysis. The aim of this study is to analyse the stresses distribution in the temporomandibular joint (TMJ) disc during a jaw closing with dental occlusion using a complete masticatory FE model. Bone components of skull and mandible (cortical and cancellous bone), both joint discs, temporomandibular capsules and ligaments, maxillary and mandibulary dental arches were meshed as high-resolution separate bodies. This model will be extended in a near future to predict the stress modifications in the TMJ disc during various loading conditions resulting from mandibular trauma, mandibular surgery or parafunction.

Materials and methods

Finite element modelling

The geometric data for the complete model were obtained from a 30 years old volunteer's healthy (normal occlusion and asymptomatic joints) and fully dentate male patient, who had skeletal Class I relationship and with no fixed prosthesis. The contours of the skull, compact and cancellous parts of the mandible, mandibulary and maxillary teeth were obtained from axial, coronal and sagittal CT images. The helicoidal, multislice CT scan was performed with a GE Lightspeed VCT 64 CT device (General Electric Medical System, Milwaukee, USA) at 120 kV and 160 mA, 0.625 mm thick tomographic slices were obtained using helicoidal technique. This procedure provides images in the axial plane, that are reformatted sagittaly and coronaly. Soft tissues as discs, capsule joint contours were constructed from sagittal MR images. MRI was performed with using a Gyroscan Intera 1.5-T (standard TMJ coil) MR system (Philips Medical Systems, Best, The Netherlands). The scanning parameters for sagittal images were: echo 1 fast spin echo sequences T2 weighted and a slicing thickness 2.5 mm. MRI and CT images acquisition was performed on a patient with a custom incisor splint to obtain 10 mm opening jaw (inter-incisal distance). AMIRA® software platform (Visage Imaging, Inc.) was used to identify and contour manually any relevant anatomical structures on the MRI and CT slices (Fig. 1). Automation was not accurate enough to differentiate some grey levels. Some tissues have indeed very close Hounsfield units values. The soft elements were located by a trained maxillofacial surgeon and radiologist. Soft elements' modelling (articular disc, connective tissue and capsule) was performed based on a set of sagittal slices (perpendicular to the condylar long axis). The upper boundary of the articular disc was the contour of glenoid fossa and articular eminence, and the lower boundary was

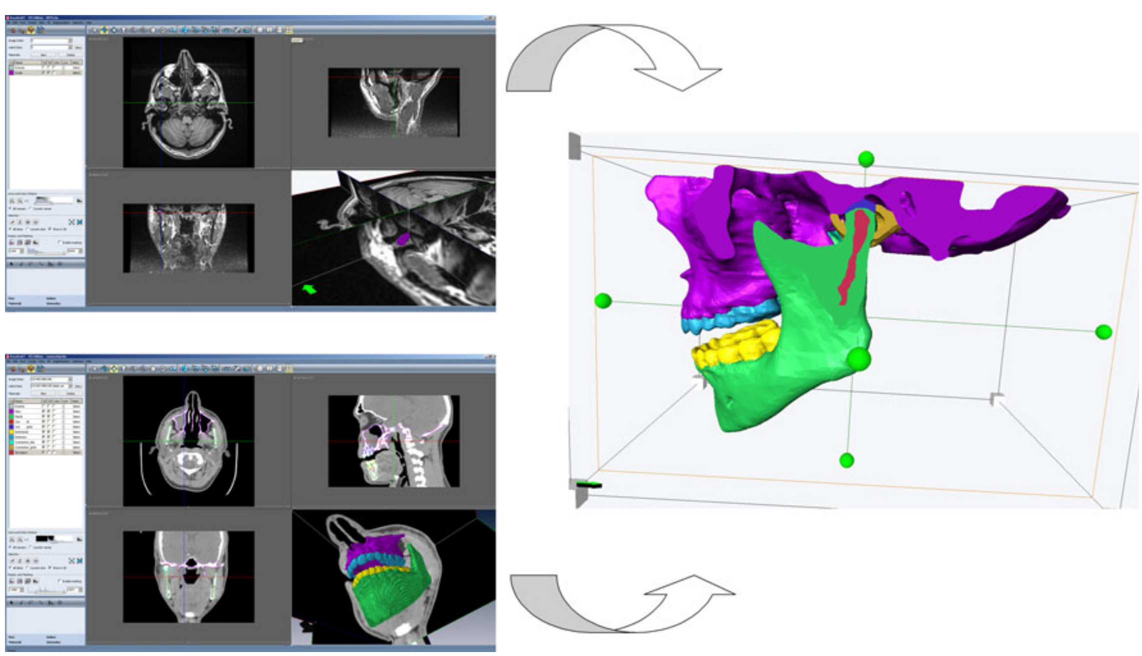

Fig. 1 Surface modelling using AMIRA® from the MRI and CT images

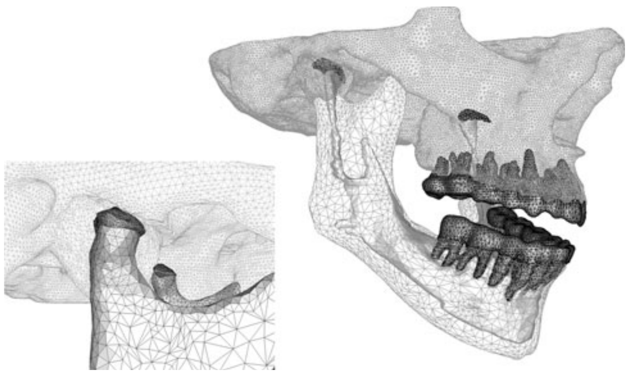

Fig. 2 Tetrahedral element volume meshes for each anatomical body

the articular surface of the condyle. The detection of capsule and ligaments was more difficult. The connective tissues (capsule and temporomandibular ligament) have been located from the MR images, and then manually generated as one single body. Three-dimensional surface meshes (hxsurface ascii format) and tetrahedral element volume meshes (I-DEAS Universal File format) were successively obtained using AMIRA® software (Fig. 2). The FE model consisted of 386,092 tetrahedral elements. The number of elements for each anatomical material is shown in Table 1. The volume meshes were exported into the FE analysis program FORGE® (Transvalor, GLpre 2005). The mechanical properties of bone mandibular components, skull, teeth and articular disc and connective tissue with capsule were assumed, as a first approximation, to be homogeneous and isotropic. A linearly elastic (Hooke's law) behaviour law was also considered since tissues deformation is limited due to applied boundary conditions.

Table 1 Number of tetrahedral elements and mechanical properties of different components in the model

Anatomical region	Tetrahedral elements (n, total = 386,092)	Young's modulus (E) MPa	Poisson's ratio (γ)
Mandible			
Compact bone	68,800	13,700	0.3
Cancellous bone	31,632	7,930	0.35
Skull	202,595	14,000	0.3
Dental arches			
Maxillary	29,761	20,000	0.3
Mandibulary	40,831	20,000	
Articular discs			
Right	1,456	40	0.4
Left	1,400		
Connective tissue-joint capsule			
Right	4,790	1.5	0.49
Left	4,827		

MPa megapascals

The material properties are summarized in Table 1 by referring to previous studies [29]. The accuracy of the three-dimensional modelling technique was checked by comparing the numerical distance values of the upper and lower incisor of the AMIRA© model and human model with anterior occlusion cap splint.

Contact conditions

A Coulomb's law was used to account for friction between the different bodies of the model. Different friction coefficient values, μ, can be used depending on the parts in contact. A bilateral sticking contact was imposed between temporal bone and mandibular condyles with the connective tissues, maxillar and mandibular (cortical and cancellous) bone with the teeth and also between cortical and cancellous bone of the mandible. The cartilage layer was taken into account by considering a low friction contact between different bodies: a unilateral low friction contact ($\overline{m} = 0.05$, $\mu = 0.02$) was imposed between temporal bone (glenoid fossa and temporal eminence) and mandibular condyles with the articular discs (Fig. 3). A low friction contact was also imposed between mandibular and maxillary teeth during clenching.

Boundary conditions and simulations

Force vectors were applied to the model to account for the bilateral masticatory muscles (masseter, temporalis and medial pterygoïd). The magnitude of each muscle force was assigned according to its total physiological cross section (PCS) (Fig. 4) and the scaling factors [17]. The origin and direction of each muscle force were defined from anatomical measurements [7]. Loading conditions were the same on both sides and the different forces were applied linearly. The PCS and the load assumed are resumed in Table 2. Calculation

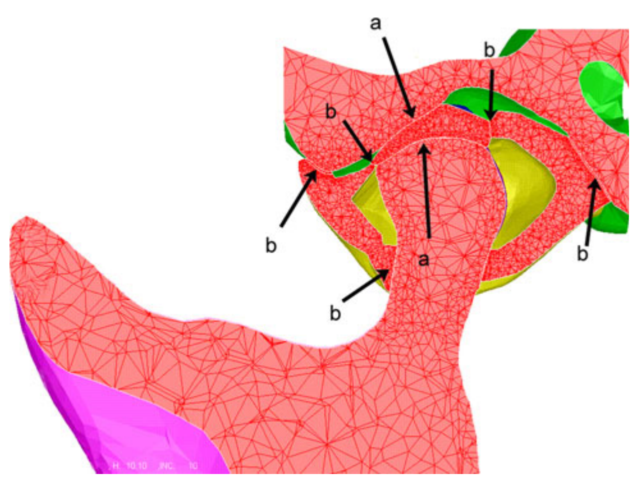

Fig. 3 Friction coefficients imposed on different contact areas. **a** Low friction. **b** Bilateral sticking

Fig. 4 Physiological cross section (PCS) of bilateral masticatory muscles

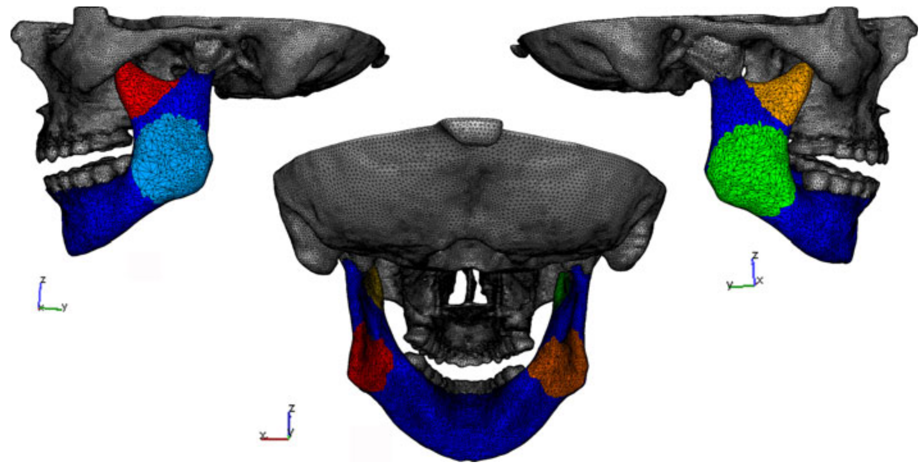

Table 2 Boundary conditions

Muscles	Total physiological cross sections (mm^2)	Load in different directions (N)		
		x	*y*	*z*
Masseter (deep and superficial parts)	9.1×10^2			
Left		50	−50	200
Right		−50	−50	200
Temporalis (anterior, middle and posterior parts)	9.762×10^2	0	50	100
Medial pterygoïd	5.095×10^2	0	−100	200

N Newton

was done on a 16 processors cluster computer with the FE analysis program FORGE®. The finite element analysis software FORGE® has already been used with accuracy for biomechanical models [22].

Assessment criteria

The von Mises stress distribution (mathematical combination of all components of both axial and shear stresses) in the articular discs was successively evaluated at 8, 4, and 2 mm inter-incisal distances (these values measure the relative displacement between central incisors of the mandible and the upper jaw) and during 500 ms period of clenching (contact between all the teeth in the maxillary and mandibular dental arches). The results were analysed and visualized with the post-processor of the Forge® software, GLview Inova 7.1 (Ceetron).

Size and orientation validation

The accuracy of the three-dimensional modelling technique was checked by comparing the numerical distance values of the upper and lower incisor of the AMIRA® model and human model with anterior occlusion cap splint. Orientation mismatch was found and needed for a correction of 3° angle in *z* axis to be correctly positioned in a Cartesian orthogonal coordinate system.

Results

The isovalues of von Mises stress in the articular discs are displayed in Fig. 5 at four different time steps during jaw clothing and during clenching. Stress distribution in both articular discs for this particular person is asymmetric. The patterns of stresses in the TMJ discs changed continuously during jaw movement. The largest stress for the left discs were located in the external part and progressively moved to the central band. It can be seen that in the right disc the highest stresses were located in the external part, anterior (Fig. 5) and then in the central band of the disc. Maximal values were 5.1 MPa (megapascals) for the left disc and 4.89 MPa for the right disc. Figure 6 shows the volumetric von Mises stress (cross section view of the right disc). The maximal stress is reached on the disc surface, in areas in contact with condyle surface and glenoid fossa. "*z* axis" displacement analysis presented in Fig. 7 shows a symmetrical movement of mandible.

Discussion

Finite elements models have their current origin and real use in mechanical engineering analysis and design. They provide interesting local information in terms of displacement, strain and stress. This local information are generally difficult to obtain experimentally. The invasive nature of the direct methods decrease their reliability: insertion of experimental devices, such as strain gauges, inside the structure can induce damage to its tissues, while placing

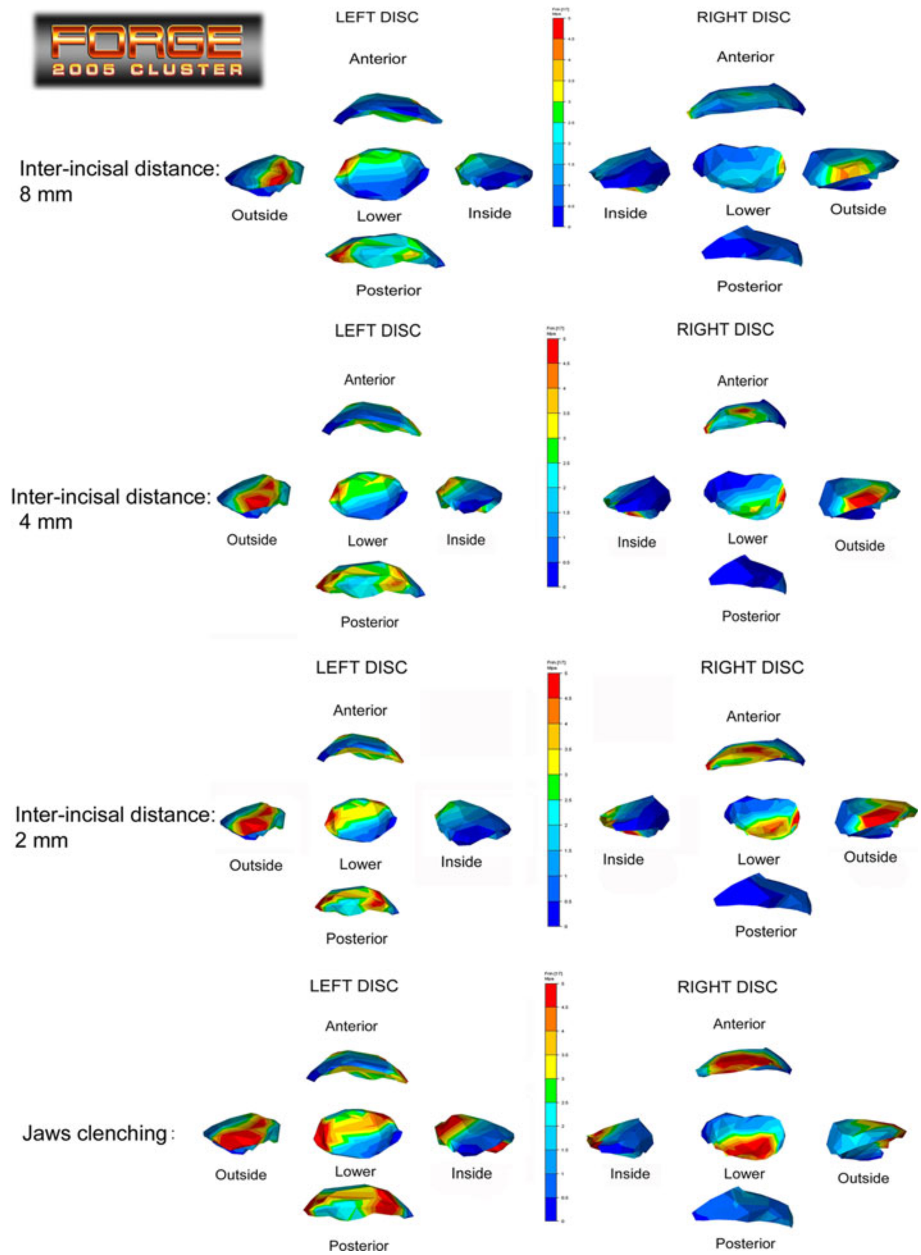

Fig. 5 von Mises stress distribution in the discs during the jaw closing

the measuring device in or between the dental arches can be inefficient. Furthermore, these experimental techniques deliver local measurements in specific points, giving an approximation of the biomechanical behaviour [1]. Accordingly, experimental studies of the biologic effects of various magnitudes of force acting on the condyle, discs or the fossa are not available in vivo.

Biological applications of FE analysis have been successful in biomechanical field such as maxillofacial in modelling human joint [25]. Obviously, biomechanical models of the human masticatory system are not perfect yet because they are based on a number of assumptions and simplifications [9]. The accuracy of finite element analyses increases thanks to the highly refined meshes of the models components used. However, an accurate model requires large computer resources for calculations. For instance, in this study, parallel computing was necessary and a computer cluster with 16 processors was used.

Several models of TMJ are already available in literature. Nagahara [20] analysed the biomechanical reactions in the mandible and in the TMJ during clenching under various approximated boundary conditions. Other FE models have been developed more recently. They considerably improved the analysis accuracy but usually only considered one of the two joints [2, 6, 27, 29]. These simulations considered symmetrical movements of the mandible but did not take into account the asymmetry of both joints. More realistic FE models of the TMJ

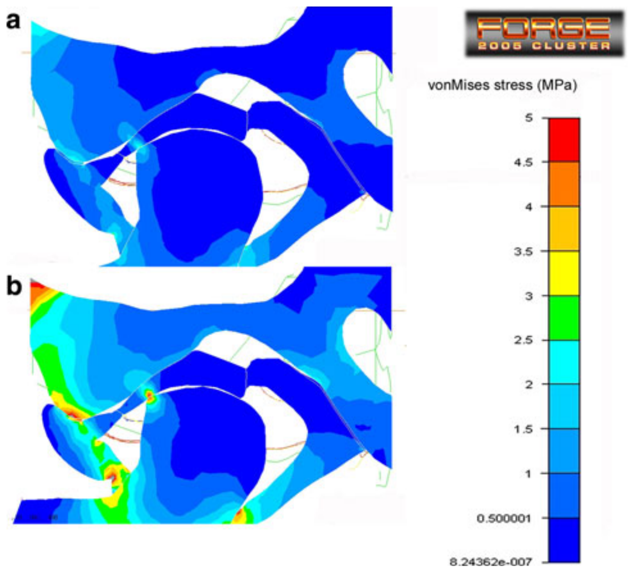

Fig. 6 von Mises stress distribution of the right disc (*cross section view*). **a** Inter-incisal distance: 8 mm. **b** Inter-incisal distance: 4 mm

(accounting for the two joints) with more complicated constitutive models for the soft components were studied later. Koolstra [15] developed a combination of a rigid-model with a FE model of both discs and the articulating cartilaginous surfaces to simulate the opening and closing movement of the jaw in symmetric conditions. Pérez Palomar [23] reported another model in which both (non-symmetrical) TMJs were solicited with a non-symmetrical movement of the jaw (lateral excursion). The latest model published by Koolstra [13], studied the direction of the first principal stress in the articular disc of the TMJ in a very accurate model. The geometry of the cartilaginous structures was obtained from the right TMJ of one cadaver [2] and the left-side joint was reconstructed as a mirror image.

In our present study, we have also considered both joints but we have worked on real geometries. The geometry of each joint was reconstructed from a healthy volunteer model in order to analyse the differences of strain results on both sides related to the anatomical asymmetry. The results of this study are thus specific to this model and other FE might give slightly different values and trends.

Mandibular fossa and articular eminence of the temporal bone and condylar head of the mandible were modelled with their irregularly shapes. Shapes of the upper and lower surfaces differ considerably. This allows for a large amount of motion but implies a lower joint stability and relatively small areas of joint contact [12]. The articular disc is supposed to reduce joint incongruency and to increase joint stability by enlarging the contact area [12]. Therefore, joints morphology is highly complex and naturally asymmetric. Thus, stresses and strains are supposed to be heterogeneous in both articular discs during masticatory movements [12, 28]. Previous models [23] show non-symmetric stress distribution for any type of motion. In our study of a closing movement, the von Mises stress distribution was almost symmetric in external and central areas but was higher in posterior area for the right disc, and higher in anterior area for the left disc. This result may change if one takes into account non-symmetrical applied forces. Indeed, generally, asymmetry may be associated with functional asymmetry, i.e. an asymmetry of muscle forces and unilateral chewing habit may induce morphological asymmetry. In this study, the results do not account explicitly for functional asymmetry, but incorporate the induced morphological asymmetry, which may be one of its consequences.

The articular capsule is composed of loose fibrous structures that are connected superiorly to the temporal bone and inferiorly to the mandible. The temporomandibular ligament reinforces the capsule laterally. These

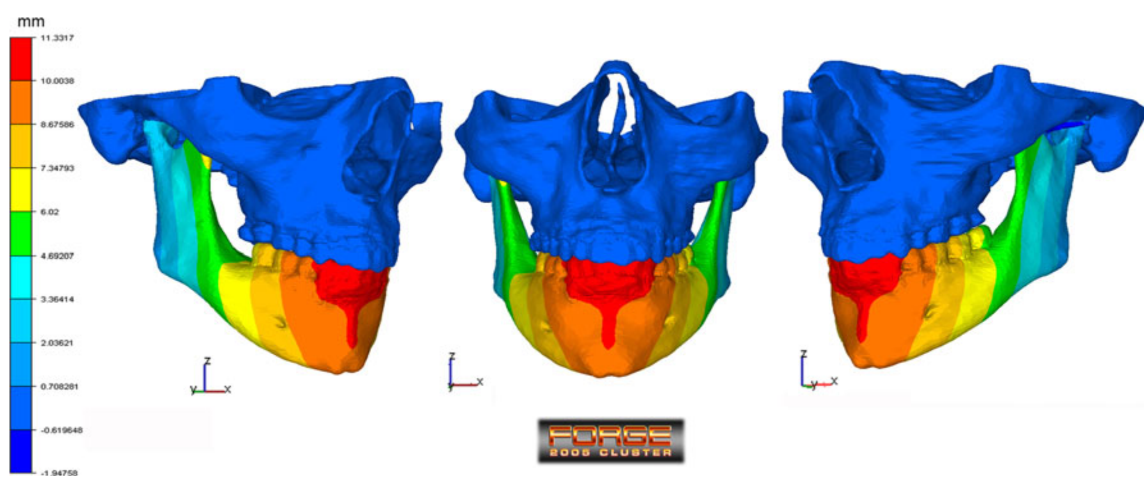

Fig. 7 Symmetrical displacement in “z” axis after jaw closing

different components of the articular capsule have been deliberately modelled as one single body. This is due to the difficulty to separate both anatomic components from MRI slices. MRI indeed enables the analysis of disc position and movement during opening and closing [3, 4], but the spatial resolution of MRI is not accurate enough for a detailed imaging of all structures of TMJ, especially the ligaments. Modelling these ligaments as a single body is an acceptable assumption since they have the same biomechanical behaviour. Sphenomandibular and stylomandibular ligaments have an important biomechanical role on jaw opening but have a minor influence on guiding the jaw during regular closing movements [12]. Therefore, they were not included in our model especially because their geometry definition and meshing are really difficult based on MRI [23]. To improve even more the model accuracy, it will be necessary to add these components in a subsequent FE model when an optimal imaging technique will be available.

The articular surfaces of the condyle and the fossa-eminence complex are covered with cartilage layers that are separated by articular disc. Previous studies have modelled the cartilage joint surfaces [30] and the influence of the cartilage was studied extensively [10]. The cartilage surface was not modelled as a separate body in our work. However, the interaction between the cartilage layers and the disc is the key point in the biomechanical simulation of the TMJ [27]. Thus, cartilage layer was modelled using a low-friction coefficient between upper and lower interfaces of the discs. Using the right friction coefficient is very important to obtain correct results in the stress analyses. The von Mises stresses are significantly higher in the entire region of the disk as the friction coefficient increases [30]. For a clinical practise, the biomechanical environment changes inducing modifications of friction conditions in the TMJ may explain the origin and progression of temporomandibular disorders. According to the present study, relatively high stresses were observed in the lateral regions of the disk and, therefore, we can speculate that wear might occur in these regions during repeated loadings presumably inducing derangements. Anyway, it would be necessary to confirm the values of the friction coefficients by analysing the difference between the disc displacement computed by FE analysis and the one measured on the MRI images after jaw closing.

From a classic anatomic perspective, the human masticatory system contains elevator and depressor groups. There are spatial and complex requirements to the construction of the muscular system. From a mechanical point of view, muscles can perform almost any tasks in various ways. For a closing jaw motion task, only the elevator group is required and consists of the masseter, the temporalis and the medial pterygoid muscles. The muscle activations during clenching are represented through the force vectors of the bilateral masticatory muscles corresponding to centric occlusion. The magnitude of each muscle force was assigned according to its physiological cross section and was in good agreement with the validated model of Koolstra [14]. This numerical approach does not consider the depressor muscles behaviour. No current model studies the depressors relaxation during jaw-closing. Gupta's [8] model showed another approach of the mathematical modelling of muscle activation. Muscles (masseters) were reconstructed from MR images to obtain a complete 3D representation. Physical properties (Young's modulus and Poisson's ratio) were taken from literature. Applied boundary conditions are not explained. The complexity of muscles mechanical behaviour laws cannot be simplified into a linear Hooke's law. The physiological complexity of the chewing system and the fact that the muscle forces cannot be easily measured experimentally make mathematical models of the mastication muscles essential for analysing human chewing [24]. Furthermore, the location, direction and magnitude of the moments applied to the teeth are important causes of errors of muscle force predictions. In order to eliminate any modelling errors associated with the use of fixed datasets, Iwasaki [11] developed a patient-specific model. The 3D computer model was derived from radiographs of patient from whom EMG data set were recorded. In the future, our model will be optimised by comparing the predicted muscle activations to in vivo data.

Some anterior models [9] did not consider the load and the contact between the teeth. However, the required jaw-closing forces are large and the loading of the joints increased concomitantly. Therefore, the behaviour laws parameters of maxillary and mandible teeth were considered here in order to analyse the influence of the magnitude and the stresses distributions in the disc during clenching. The model includes the combination of the trabecular bone and the root of the teeth because this anatomical region absorbs occlusal loadings and distributes the stress over the alveolar bone. These occlusal loads also influence the value of stresses in temporomandibular discs during clenching. As far as we know, this is the first 3D finite element model of masticatory system with closing jaw simulation that includes body contact between the root of the teeth and the trabecular bone. Nevertheless, our model should have taken into account the periodontium tissue because of its important mechanical role [21]. However, this tissue is invisible using standard medical imaging condition. It requires a microtomography system, which is only applicable on small size sample. Such approaches were used for dental finite element analysis studies [22].

The main limitation of our model was the material behaviour of the articular discs, which was approximated by a linear elastic model and not as a non-linear visco-hyperelastic material model. As highlighted by Tanaka [27], developing a model that describes the real visco-elastic behaviour of the discs is still challenging. According to in vitro experimental studies [26], some poro-elastic finite element models have been introduced in order to simulate its biphasic behaviour, which accounts for the shock-absorbing properties. Viscoelastic properties are important to analyse the influence of velocity and subsequent cycles [16]. In this present analysis, these influences were not investigated, and consequently the use of a linear behaviour law was more appropriate. Moreover, we assumed the linear mechanical behaviour because of the small deformation of the discs. For larger deformation, the use of non-linear elastic behaviour is required.

However, even with the above-mentioned limitations, our model including a highly refined mesh of each component was able to demonstrate the effect of jaw closing and clenching in physiologic conditions on the stress distribution in the TMJ discs. Comparison of our results to previous studies was difficult because applied load and material properties of the articular discs varied among these models. Nevertheless, stress levels given by our model were within the range of magnitude than reported stress. Previous experimental [5] and numerical studies [13] reported indeed that the maximum stresses in the disc were of the order of 0.85–9.9 MPa as compared to 5.1 MPa in the present study.

Conclusion

The authors are not aware of any other work that produced a 3D finite element (FE) model of the mastication system with this level of anatomical accuracy. In particular, this model accounts for the morphological asymmetry of the temporomandibular joint and its influence on the stress distribution during clenching. It also incorporates teeth as separate bodies as they play an important role on stress distribution during clenching.

The reconstruction of all components of the masticatory system makes it a high level of anatomical accuracy. Appropriate contact and friction conditions have to be used between each of these components to get reliable results and investigate distribution of stresses and strains in the discs.

Future developments of this work will be to detect TMJ disorders in simulated situations, such as trauma, mandibular surgery, mandibular distraction osteogenesis or simply disturbed occlusal equilibrium.

Conflict of interest None.

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