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Original paper

Finite element modelling of the articular disc behaviour of the temporo-mandibular joint under dynamic loads

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The proposed biodynamic model of the articular disc joint has the ability to affect directly the complete chewing mechanism process and its related muscles defining its kinematics. When subjected to stresses from the mastication muscles, the disc absorbs one part and redistributes the other to become completely distorted. To develop a realistic model of this intricate joint a CT scan and MRI images from a patient were obtained to create sections (layers) and MRI images to create an anatomical joint CAD model, and its corresponding mesh element using a finite element method. The boundary conditions are described by the external forces applied to the joint model through a decomposition of the maximum muscular force developed by the same individual. In this study, the maximum force was operating at frequencies close to the actual chewing frequency measured through a cyclic loading condition. The reaction force at the glenoid fossa was found to be around 1035 N and is directly related to the frequency of indentation. It is also shown that over the years the areas of maximum stresses are located at the lateral portion of the disc and on its posterior rim. These forces can reach 13.2 MPa after a period of 32 seconds (s) at a frequency of 0.5 Hz. An important part of this study is to highlight resilience and the areas where stresses are at their maximum. This study provides a novel approach to improve the understanding of this complex joint, as well as to assess the different pathologies associated with the disc disease that would be difficult to study otherwise.

Key words: *biomechanics, articular disc, chewing, resilience, temporo-mandibular joints, muscle forces, 3D imaging, finite element analysis*

1. Introduction

The temporo-mandibular joint (TMJ) is one of the human body joints being subjected to strong stresses. Since the disc is located in the middle of the joint it is subjected to large compressive loads and stresses during the crushing phase of mastication [1]; this indeed is the focus of this study. Furthermore, the disc undergoes high peak stresses during bruxism periods triggering contractions at variable frequencies and amplitudes [2].

At some point of the lifetime of this joint, 28% of the population experience pain and/or temporo-mandibular disorders [3].

At this stage it is commonly understood that the biomechanical characteristics of the joint can affect the disc shape when it is under stress. It also plays an important role in load absorption and redistribution within this joint. But a lack of adaptability of tissues and an overload on the articular components are also multifactorial aetiologies of temporo-mandibular joint disorders [4]. The main objective of the study is to define the pathogenic processes that might be responsible for or trigger temporomandibular joint disorders, and in vertebral disc lesion. This second objective is to see whether joint resilience can have an impact on the occlusion. The biomechanical environment of the TMJ and its mechanical properties are key notions to understand the dynamics and kinematics of this joint, but also to find how its pathogenic process may trigger the myoarthropathy.

The modelling of bone and soft tissues using CT and MRI are gaining a lot of interest due to the soft-

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ware development and accuracy by which we are able to build the patient's specific models [5]. Usually, a computer program is used to trace different contours and discretized points and build a rendering image by smoothing the interconnected surfaces. This is needed in the mesh and finite element models to avoid gaps and singularities and allow the model to converge faster during processing [6]–[10]. In addition to the tracing of surfaces at each slice, which can be done automatically, the desired bone surface can also be traced manually or through markers. It is shown that in the tracing paper, contour lines can be used directly for the reconstruction of the elements on a numerical graphic pad [3], [12].

The MRI images can be directly processed by a 3D reconstruction software, but the images used should belong to the same plane; images taken from different planes cannot be combined [13]–[17]. Moreover, the MRI images' thickness is too large to obtain a model with a reliable geometry, without any considerable cleaning up of the mesh that would generate approximations of the forms. Images from the scanner have a better resolution and make modelling easier to perform. Yet, soft tissues are difficult to see on these images. The technique of image acquisition and image processing makes it possible to use modelling software that can compile images in the 3 directions of space [17].

2. Materials and methods

2.1. Design method and experimental apparatus

CT scans and MRI images of a male volunteer individual in his early thirties with no prior articular or muscular medical history were used for the initial design of the TMJ joint. The examinations were performed at the Reims hospital in Reims. A SIEMENS® CT scanner, with a thickness of 0.75 mm, provided images with a 512×512 pixel resolution. They were saved under DICOM and JPEG format(s) muscular window and osseous window. On the other hand, the modelling of the left TMJ was performed using 20 images obtained from Phillips MRI 3 Tesla. The latter was retained in a sequence of «T1-3D-Wats-Clear», with a repetition time (TR) of 20 milliseconds, and an echo time (TE) of 3.69 milliseconds within a range of vision of 515×512 pixels – it important to note that the segmentations had a thickness of 3.4 mm and a 1.7 mm of overlap. Just as scan images, MRI images are saved under DICOM and JPEG format(s). The segmentation of the manducatory apparatus, digitization, and interpolation were performed with a 3.1 Amira software (TGS Corporation). ABAQUS® software was then used to run the analysis and simulation. The purpose of the analysis was to determine the stresses, deformations, displacements, and reaction forces within the TMJ.

2.2. Modelling with ABAQUS®

Globally, this approach is very complex. So as to simplify the simulation on the ABAQUS® software as much as possible, the modelling of the skull and of the condyle was confined to one part (figure 1).

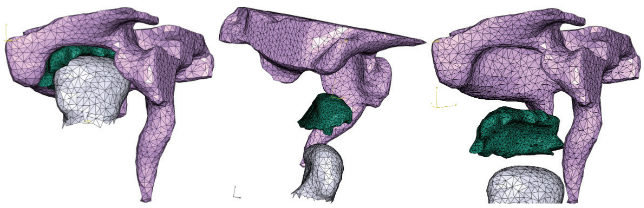


Fig. 1. Meshing of the mandibular disc with its environment

The skull and the condyle were supposed to be rigid and non-deformable, and for their discretization a rigid triangular element (R3D3) was used. The skull includes 5567 nodes and 11167 elements, whereas the condyle includes 5415 nodes and 10822 elements. As for the disc itself, it is considered to be deformable. It includes 1908 nodes and 7327 tetrahedral elements C3D4. The skull base was considered to be completely blocked. Taking into account the complexity of the approach, we noted that the number of nodes used in our model was higher increasing the precision of our results.

For the simulation, it was necessary to assign to the articular disk a behaviour law close to the living to perform the simulation.

The constitutive law chosen is the result of an experimental study comparing the disc to a hyperelastic material. The behaviour law chosen was taken from an experimental study which compared the disc to a hyperelastic material [18].

Compared with the experiment, the behaviour law applied to a model has to reproduce qualitatively the same decrease in the time of maximum forces, of the amount of dissipated energy as well as the relationship between the stress amplitude and excitation frequencies. The behaviour law chosen makes the connection between the Green–Lagrange deformation and the

Piola–Kirchhoff stress in a polynomial equation of the third order that can reproduce the hyperelastic behaviour of the disc: $\sigma = c_3\varepsilon_{eq}^3 + c_2\varepsilon_{eq}^2 + c_1\varepsilon_{eq} + c_0$. The values of the coefficients c_3, c_2, c_1 and c_0 are 10^4, 10^4, 0 and 1, respectively.

3. Results

The forces applied have to correspond to the sum of the forces developed by the levator muscles in the 3 directions of space. For this, these forces have to be considered in the vectorial form, and a method of appropriate vectorial decomposition has to be applied. In order to achieve this, we were interested in the crushing phase during occlusion on the working side [1].

So, the muscles are isometrically contracted and develop their maximum load [1]. During isometric contraction the muscle length is constant, but its contraction intensity increases until reaching its maximum value. It is possible to calculate the force/tension developed by a muscle during contraction thanks to its electromyographic activity and its surface of maximum area [19]. However, electromyographic records were not beneficial for this study. A linear relationship between the maximum strength and the maximum area of the muscle was set by using a coefficient $\Gamma = 124$ N/cm^2 as being the average values taken from the literature [19].

In order to find the resultant forces applied to the condyle according to the 3 directions of space (x, y and z), a vectorial decomposition of the forces applied to the mandible in 3 directions must be used. To do this vectorial decomposition, the maximum intensity, as well as the orientation of the force developed by a muscle, must be known.

In this model, only the joint is subjected to numerical simulation. The modelling of the levators' muscles with the Amira software made it possible to calculate the angles which are necessary to calculate resultant forces in space. However, these components concerning vectorial decomposition of the resultant applied to the mandible, but there is a distribution of this force on teeth (65%) and on the TMJ (35%) [20].

The resultant muscular force chosen and applied to the TMJ, whose components are $F_x = -3.43$ N, $F_y = -190.14$ N, $F_z = 631$ N, are shown in figure 2. As the mastication frequency ranges from 0.75 to 1.5 Hz, the disc was subjected to indentation forces at arbitrarily chosen frequencies of 0.5 Hz, 1.0 Hz and 2.0 Hz.

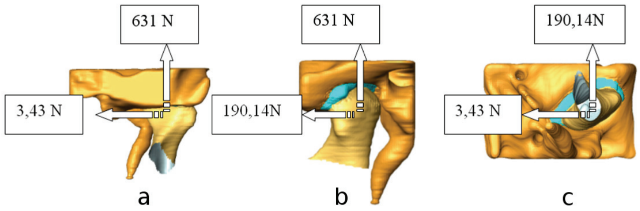


Fig. 2. Illustration of resultant forces:
(a) front view, (b) left side view, (c) bottom view

Figure 3 shows the evolution of the resulting force of reaction calculated at skull base level. It gives information on the transmission, through the disc, of the forces applied by the condilar head.

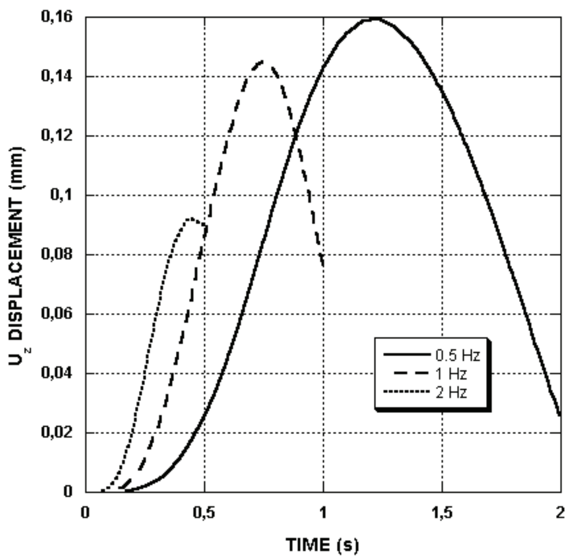


Fig. 3. Resulting force evolution of the skull during the first cycle

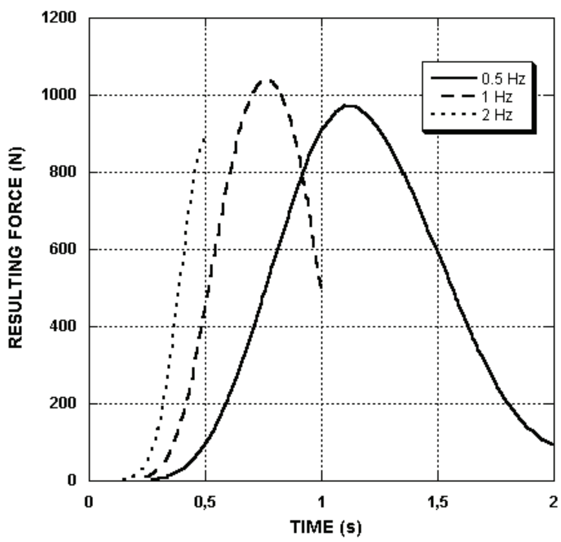


Fig. 4. Illustration of the U_z resulting displacement of the condylus in the first cycle

• At 0.5 Hz the resultant force of reaction reaches its maximum (973 N) just after the half period (1.1 s) and then becomes almost null (92 N) at the end of the period. It was noticed that the forces applied are generated over a sufficient time span to increase the disc stiffness, so the force is transmitted to the glenoïd fossa. During force relaxation, the energy completely dissipates and the disc does not distribute force on the skull base anymore.

• At 1 Hz the maximum reaction force (1035 N) appears after three-fourths of the period (0.78 s); it is the highest at the skull level and remains considerable at the end of the period (495 N). In fact, the energy stored by the disc was not totally dissipated at the end of the period; force-relaxation time should be increased to make resultant reaction force tend towards zero.

• At 2 Hz the maximum resultant reaction force is the weakest at the skull level (889 N); moreover, it appears at the end of the period (0.5 s). This is due to the fact that the disc rigidity, because of the speed of the loading, becomes more significant. The maximum force, however, would have been expected to be the highest at this frequency, but the speed of the test does not make it possible to reach the maximum force at 2 Hz (figure 3).

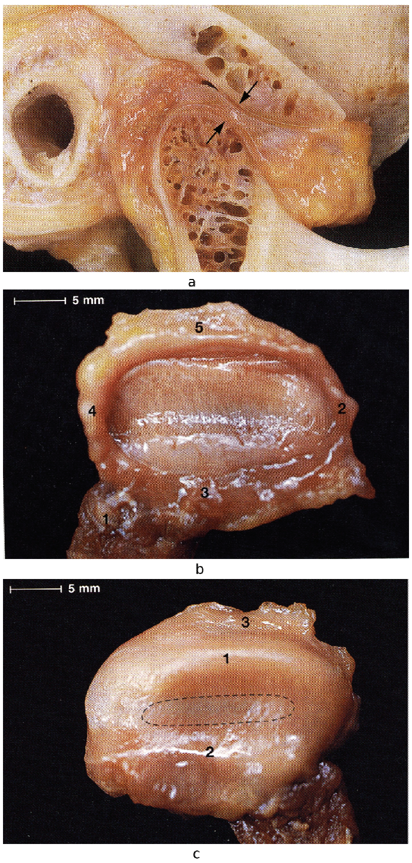


Fig. 5. Location of the area of intimate contact between the condylus and temporal condyles: (a) bottom view, (b) superior skull view, (c) view after dissection

In general, the disc plays the role of a real damper, which partly absorbs the forces applied to it, before transmitting them. Figure 6 shows the distribution of the von Mises stresses obtained in the 16th charge cycle. For each frequency, the ventral view shows the contact area between the disc and the condyle (that is, the underside of the disc of the left joint), whereas the dorsal view shows the contact area between the disc and the skull (figure 5) [21]. After 16 cycles, it was noticed that maximum stresses appear at a frequency of 0.5 Hz because the time of mastication is the longest (32 s). The maximum stresses, at the level of intimate contact zone between the condyles and the disc, are equal to 13.2 MPa. This stress can be found on the upper surface of the disc. Attention is not paid to the disc borders because these zones are not in contact with the mandibular condyle.

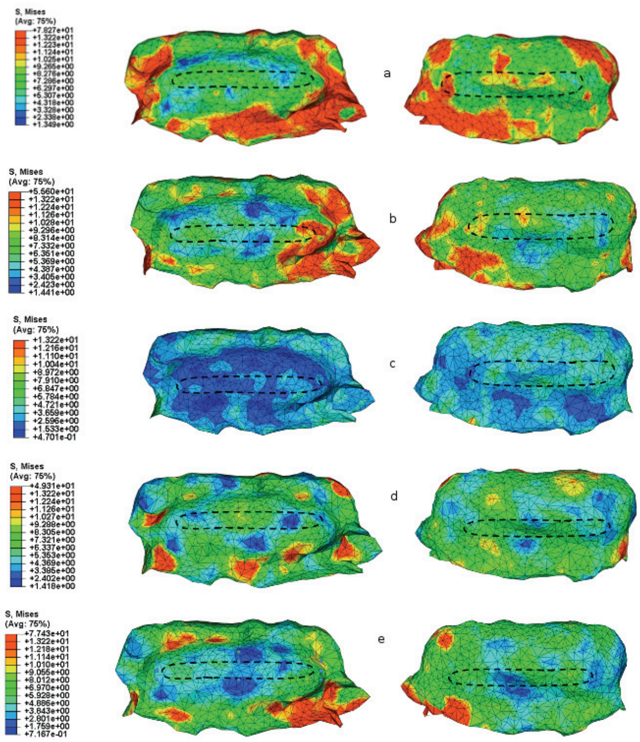


Fig. 6. Distribution of the von Mises stress (MPa) in lower and upper views for different frequencies: (a) 0.5 Hz ($T = 32$ s), (b) 1 Hz ($T = 16$ s), (c) 2 Hz ($T = 8$ s), (d) 0.5 Hz ($T = 8$ s), (e) 1 Hz ($T = 8$ s)

On the other hand, after 8 seconds, it was noticed that maximum stresses appear this time at a frequency of 1 Hz. At a frequency of 0.5 Hz there is enough time for the disc to partly discharge, whereas at a frequency of 2 Hz (figure 6) the charging speed is very considerable, and this does not allow the disc to load sufficiently. Thus, for a time of 8 seconds, the maximum stress zones are located on the lateral part of the disc as well as on its posterior rim. In fact, the results showed that, on the whole, the simulation of the joint behaviour with imposed force is not coherent with the physical reality. In actual fact, after having observed the evolution of the displacement of the application point of the force at condyle level in the *z* direction over a period of time (figure 4), it was noticed that the initial position is not the same, which is incoherent with the physical reality of the mastication phenomenon. It would be preferable for the simulation to lean on an imposed displacement of the condyle, ideally taken from experimental results.

4. Discussion

The muscular forces used in our approach derived from a linear relation between the muscular forces and the muscular areas, which is arguable because muscles cannot be considered as mere springs, nor as generators of classical forces [19]. However, even if the exact forces developed cannot be found, the distribution of the developed forces between the different muscles can still be considered.

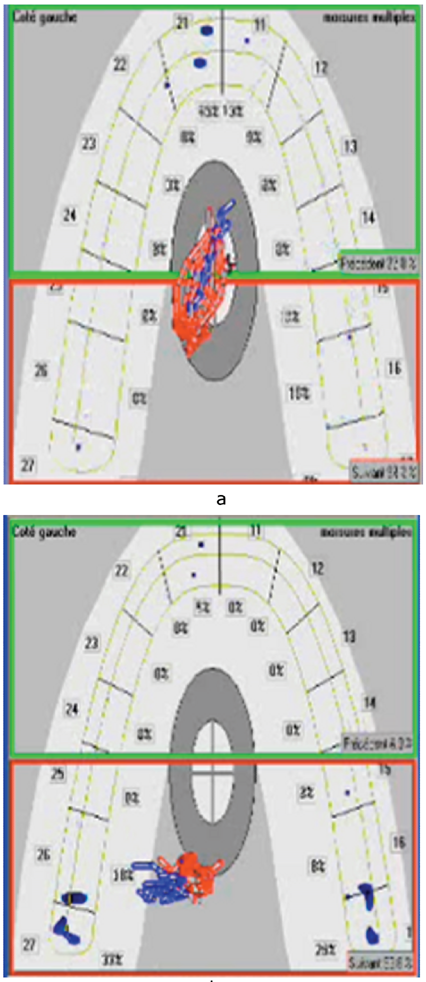
Actually, the force developed by a muscle is calculated thanks to its area and its direction. Both of these quantities are reliably evaluated due to the imagery taken from the scanner for each muscle. The calculation of the resultant force is made based on the forces developed by the levator muscles. Even if the factor varies, only the intensity of this force will vary, not its direction. Schumacher and Grant, as well as TANAKA et al. [10], MEYER et al. [19], and TANAKA [22] estimate that the resultant of the forces of mastication muscles adds up to 500 N. The evaluated resultant of the forces of mastication was equal to 631 N according to the vertical *z*-axis on the condyle level. The osseous structures like the mandible and the skull basis could be considered to be deformable in the modelling and not rigid anymore.

This study highlights the distribution of stresses on the disc level as well as the displacement of the condyle. The temporal factor shows how the disc tissue becomes accustomed to load, as well as the evolution with time of the deformation and of the quantity of the dissipated energy. The disc behaviour is subjected to a behaviour law of the material experimentally defined that should be considered as a poro-hyperelastic material, strengthened by fibers [23]. But this behaviour law cannot affect the whole disc because the collagen fibers are not uniformly distributed and directed. Its behaviour, when subjected to different tensile or shear forces, is not uniform according to the strained area. It has to be taken into consideration that when the tissue is compressed, the porosity is lowered and consequently, the permeability decreases. In other words, a non-linear function which characterizes permeability, and which depends on deformation, should be incorporated [24], [25]. At the sight of the results, the questions about the relationship between the presence or absence of facial pain, associated with amplitudes and frequencies of contraction higher or lower during episodes of bruxism [24], and about the degradation of the articular disc could be raised. In this case, the increase, on the one hand, of the indentation frequencies of the disc and, on the other hand, of the loading speed increase the stresses on the top of the disc at the level of the posterior rim as well as the reaction forces at skull basis level. A centric bruxism, which includes muscular contractions very close together, could play a role in the disc damage or in the fibroelastic cartilage destruction of mandibular and/or temporal condyles. In order to reinforce these hypotheses, the behaviour law of the disc should be refined by incorporating a new routine on ABAQUS®.

In these experiments of disc solicitation, the condyle moves up to a maximum of 0.37 mm. This is due to the disc strain; this phenomenon has already been put forward by other authors [27]. In vivo, this displacement can be expressed by a posterior shift of the mandibular body and an increase of loads' distribution on the surface of posterior teeth.

In fact, the observation of the evolution, for a period of time, of the load application point displacement of the condyle in the *z* direction (figure 7) shows that the initial position is not found again. This strain has an effect on the spatial position of the mandibular body. The hypothesis of the posterior displacement of occlusal loads at the level of molars occlusal surfaces due to a shift (tilt) of the mandibular body can be confirmed by the Tscan. This system measures the loads' distribution during occlusion and performs dynamic records of it [28], [23].

The load test consisted in masticating a chewing gum of a quite tough consistency for 2 minutes and was repeated during the day. Phases of rest for several hours have been respected. The average result during the various performances shows an increase of 8% of occlusal loads in the posterior region (red box). Occlusal loads amounting to 1171.85 N for this person, the increase represents an additional load of 93.74 N. It would be interesting to carry out this load test on



a

b

Fig. 7. Barycentre position of occlusal loads before displacement (a) and after displacement (b)

a statistically significant sample in order to validate our hypothesis. We find this phenomenon in the case of unstable occlusion characterized by tooth contacts' increase during an increase in muscle contraction. This is not only due to the periodontal ligament depressibility but, as we have shown earlier, also due to the disc strain associated with a condylar displacement. For some authors, this tension overload phenomenon at the articular level can be of dysfunctional nature. Our model is interesting to better understand, in different situations, how the disc redistributes the loads. Although our model is limited in size and it would be necessary to extend it to the dental arches, it offers a better understanding of the articular biodynamics, therefore the disease processes.

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