

# Mechatronic Design Towards Investigation of the Temporo-Mandibular Joint Behaviour

Victor Creuillot, Cynthia Dreistadt, Krzysztof J. Kaliński and Paul Lipinski

**Abstract** A significant problem of the temporo-mandibular joint (TMJ) research is lack of data concerning geometry and position of TMJ discs. It leads to necessity of developing a driving method of the process optimization, which is based on chosen techniques of mechatronic design. In particular, the latter concerns a technique of experimentally supported virtual prototyping. On this stage, the research is characterized by well-verified constitutive characteristics of all components, engagement of real components of the research system, offline experimental identification of the TMJ system parameters, offline experimental determination of the appropriate parameters of the TMJ, online computation of the TMJ process of non-stationary model and assessment of the driving method's efficiency. In case of the medium opening-closing jaw motion, results of the FEM are close to the experimental data and validate discs' geometries and positions. Quality of the obtained solution is also verified by analysing the contact between discs and surrounding bones. In case of the immediate clenching simulation, a comparison of finite element model simulation and dental plaster mould occlusions testifies the correctly reproduced occlusal scheme. Values of the contact forces and the contact area surfaces enable calculation of the mean contact pressure values on the cranial and caudal faces of the discs. The obtained model allowed reproducing accurate anatomic mandible trajectories and also a physical occlusion.

**Keywords** Temporo-mandibular joint · FE modelling · Muscles' motion driving · Mechatronic design

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## 1 Introduction

Temporo-Mandibular Joint (TMJ) allows the relative motion between the mandible and the maxillary. It is one of the most utilized joints in the human body. According to Kamina [1], it endures approximately 10,000 cycles per day, and by consequence, it is also one of the most symptomatic joints. Indeed, up to 35 % of the worldwide population presents TMJ disorders. In 70 % of the pathologic cases, the symptom corresponds to a disc displacement. Nowadays, causes of these dysfunctions are still misunderstood. To elucidate the origin of such pathologies, the key factor, from the biomechanical viewpoint, is the knowledge of forces acting on the TMJ discs. However, for the evident ethical reasons, the direct TMJ force measurements are unworkable in living human. Mathematical modelling of the joint appears as the best approach to correctly quantify these forces.

Many 2D and 3D finite element models of TMJ have been developed during the last 15 years, see for instance [2–8]. The most finite element studies of the TMJ were performed on one side with the strong assumption of the jaw symmetry. They were dedicated to study the stress distribution in the TMJ for asymptomatic as well as pathologic cases and for different movements such as opening, closing, opening-closing cycle or jaw clenching. The first studies have been focused on healthy cases. Chen et al. [9] built a 2D model of one side of TMJ in order to establish a relationship between the condylar sagittal displacement and the stress distribution into the disc. A few years later, Tanaka et al. [10] proposed a method of geometry reconstruction based on MRI acquisitions. They developed a more realistic finite element TMJ model of a healthy person. In order to show the influence of the internal TMJ's disorders, Tanaka et al. [3] used the same method of reconstruction to compare the stress distribution in the TMJ between subjects with and without anterior disc displacement.

Since even the healthy subjects are not fully symmetric, the modelling of a complete jaw appeared necessary. Mori et al. [7] and Savoldelli et al. [8] demonstrated that the stress fields in the two discs are different, the condyle trajectories are not the same and the geometry itself of the left and right TMJs is dissimilar. Only the study of Savoldelli et al. [8] takes into account the full skull without assumption of the condyle geometry. This study simulated a 10 mm inter-incisal jaw closing and prolonged clenching. Force vectors oriented in muscular directions reproduced this simple movement. The crucial point of TMJ modelling is related to the discs description. The TMJ disc is constituted of cartilaginous tissue with collagen fibres oriented in function of their location [11]. In numerous studies [3, 7, 12], the disc geometry is retrieved from the MRI acquisitions. Otherwise, its geometry is reconstructed from the articular surfaces in the occlusal position [5, 13]. Recent studies were focused on prolonged clenching which is a current phenomenon that can appear during the sleep leading to the effect of a disc overload. A large variety of constitutive laws for the disc can be found in the literature. In the case of large movement, the disc is frequently considered as an elastic isotropic component [3],

whereas in clenching simulations, it is usually regarded as viscoelastic or hyper-viscoelastic constituent.

In some cases [4, 6, 13], the elevator muscles are taken into account or displacements are merely applied to the condyle. In our approach, the mandible was driven by a set of muscles. The FE software MSC MARC was used to perform simulations. However, this software does not provide elements reproducing muscles' actions such as those elaborated by Hill [14], or improved by Zajac [15]. This kind of "actuators" is implemented in biomechanical Multibody Software such as LifeMod or OpenSim.

The aim of this study is to validate an original driving method of the mandible movement by the muscular contractions reproducing muscles' activation determined using the LifeMod code. In order to improve performance of this method classical mechatronic design techniques [16] were employed. Mechatronic approach to the design process is characterized by the fact that several components of the system/process can be designed in parallel (i.e. concurrently), provided that there will be a method for checking the compatibility of each element. The benefits of concurrent design are: shortening the design phase, simplifying and accelerating the implementation and the possibility of flexible implementation of individual functions.

The correctly carried concurrent design, with careful verification of the components (virtual prototyping) allows us to avoid problems with the cooperation of the individual components of the system [17]. When all components are developed simultaneously, one can be more flexible to implement various functions of the device/process in various fields.

A standard mechatronic design procedure was successfully applied in this work, similarly to that proposed by Kaliński and Buchholz [18] in scope of development of 3-wheeled mobile platform.

In the paper, two simulations are presented—one reproducing a medium opening–closing movement and another mimicking clenching with large occlusal forces. The simulated condylar trajectories have been compared with experimental data of Alvarez et al. [19] to validate the driving method used in this paper. On the other hand, TMJ forces have been analysed by comparing the obtained values with data from the open literature.

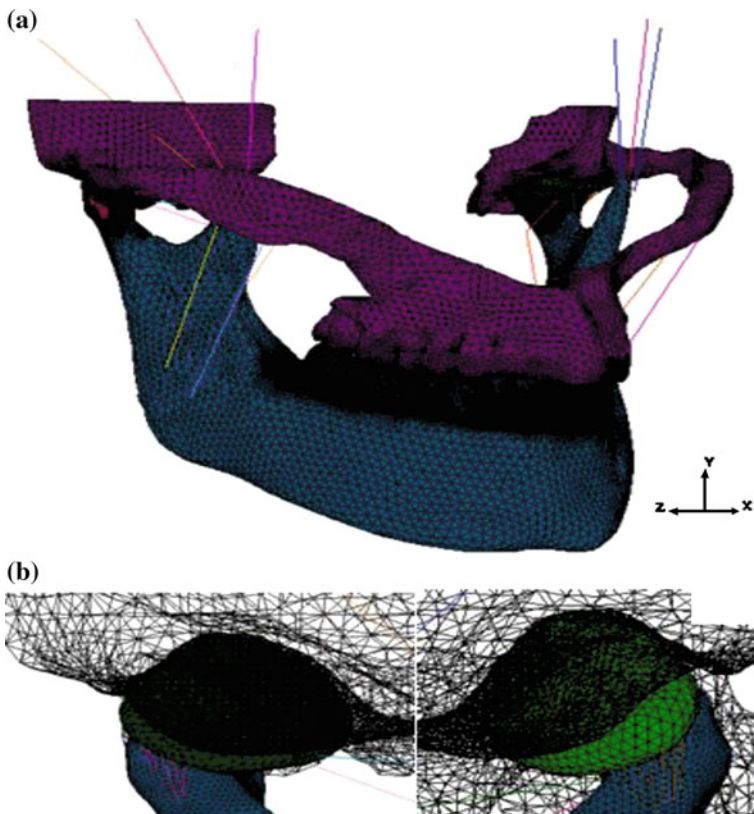
## 2 Materials and Methods

A 31-year-old female healthy volunteer with no history of present and past TMJ disorder (Skeletal class I and asymptomatic joint) has been selected for her relatively symmetrical condylar trajectories during an opening-closing movement of the mouth. The movement recording has been made by the WinJaw system Zebris GmbH and was briefly presented in Alvarez et al. [19].

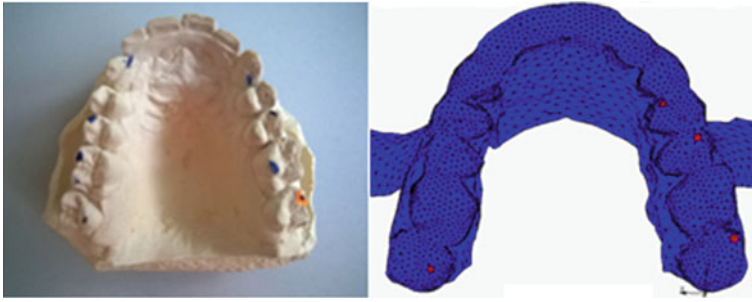
## 2.1 Numerical Model

A cone beam computed tomography with 0.25 mm thick slices was carried out in close mouth position on this volunteer. The segmentation and the reconstruction tasks were done using Mimics 14.12 and 3Matic 5.1 software packages (Materialise, Leuven, Belgium). Only the bone structures, maxillary and mandible can be accurately recovered with the tomography. These structures have been imported into the Hypermesh software 11 (Altair, Troy, Michigan, USA) and meshed using linear tetrahedral elements with a size variation depending on location in the mandible. The condyles and teeth have been meshed with a small element size (i.e. 0.5 mm) to increase the precision of obtained mechanical fields in these areas. The obtained mesh is illustrated in Fig. 1a, b.

The model has been set in the occlusal position by comparing with dental plaster cast executed on the volunteer (Fig. 2). The computed tomography does not permit to visualize disc, so they were designed following indications from literature data,



**Fig. 1** The mesh of finite element model: **a** global view, **b** details of left and right TMJs



**Fig. 2** Comparison of dental plaster and numerical occlusal positions

shape was taken from Athanasiou et al. [11] and occlusal positioning from Bumann and Lotzmann [20]. Disc volumes have been created in accordance with the data of Tanaka et al. [10]. They were meshed by using tetrahedral quadratic elements with a size of 0.5 mm.

The articular capsules were modelled by using 3D truss elements representing medial and lateral parts of the capsule with a position taken from Bumann and Lotzmann [20]. Literature provides stiffness for these ligaments. Cross-sections associated to the truss elements have been defined in function of this stiffness as

$$A = \frac{k \cdot L}{E \cdot n} \tag{1}$$

where  $A$  is the unknown ligament cross-section area,  $k$  is capsular stiffness,  $L$  is length of ligaments,  $n$  is number of parallel ligaments modelling the capsular part and  $E$  is Young’s modulus assigned to the ligaments representing the capsule.

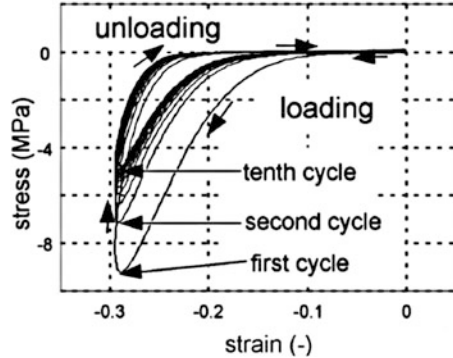
## 2.2 Material Properties

Generally, bones are considered as elastic, macro-heterogeneous and anisotropic media. However, in this work, which is focused on the discs behaviour, isotropic elastic properties have been assigned to the mandible and the maxillary. Distinction between cortical and cancellous bone was not implemented in this model, because the mechanical fields in the bone are not analysed. The corresponding data is summarized in Table 1. A density of 1.4 g/cm<sup>3</sup> has been attributed to the mandible in order to obtain a mass of 74 g in accordance with rigid body model of Alvarez et al. [19].

TMJ discs are modelled as a hyper-elastic Marlow continuum [13]. The corresponding constitutive relationship is implemented in the MSC MARC software. The constants of this law were deduced from the first cycle of the stress–strain curve from experimental data of Beek et al. [21] presented in Fig. 3.

**Table 1** Material properties of model components

Component	Young's modulus (MPa)	Poisson's ratio
Mandible	13,700	0.3
Maxillary	13,700	0.3
Disc	Data from Beek et al. [19]	
Capsule	0.2	0.3

**Fig. 3** Experimental data describing the disc stress–strain relationship, Beek et al. [21]

As mentioned in previous section, the capsule has been modelled by elastic 3D truss elements. The last line of Table 1 specifies the corresponding elastic properties of these elements.

### 2.3 Contact Management

Three interacting deformable bodies, namely condyle, disc and fossa, form each temporomandibular joint. All these contacts are supposed frictionless as the synovial fluid plays the role of quasi-perfect lubricant; see Chen et al. [9].

Teeth have been assumed as the continuation of the maxillary and mandible bones. The upper and lower teeth arcs are considered as two deformable contact bodies in the clenching case simulation. The friction coefficient  $\mu = 0.1$  was supposed for this interaction.

### 2.4 Muscular Action Modelling

In numerous finite element studies, the TMJ behaviour was analysed in various configurations, but anyone used a muscular activation to drive the mandible. In this study, our intention was to drive the mandible by activation of eleven pairs of muscles, illustrated in Fig. 1. The MSC MARC was used to reproduce the motions. However, as it was mentioned in the Introduction paragraph, this general-purpose

software does not include muscle elements. To circumvent this difficulty, the Multibody System software LifeMod was first worked to determine the simultaneous forces and elongations of all Hill-type muscles. First, experimental trajectories of the condyles and incisor point, determined by Alvarez et al. [19], for the same volunteer, have been used throughout the inverse dynamic simulation. The activation profiles of all muscles were obtained in this way. Next, during the direct dynamic analysis stage, these activation profiles were used to simulate a required cycle of mandible motion. As the output, files of the current length  $l(t)$  and force  $F(t)$  experienced by the muscles were generated. To reproduce the same behaviour of muscles, modelled by 3D truss elements in MARC, the thermal contraction capability of these elements was exploited. Let  $l_o$  be the initial length of a muscle. Its total logarithmic strain  $\varepsilon^T(t)$  is defined by its current and initial lengths:

$$\varepsilon^T(t) = \ln \frac{l(t)}{l_o} \quad (2)$$

The elastic part of this strain  $\varepsilon^e(t)$  is due to the force exerted by the muscle:

$$\varepsilon^e(t) = \frac{F(t)}{SE} \quad (3)$$

where  $S$  and  $E$  are the muscle cross-section area and Young's modulus, respectively.

The muscle contraction is reproduced by the thermal expansion capability of the truss elements. The contractile strain of the muscles  $\varepsilon^c(t)$  is calculated using the following expression:

$$\varepsilon^c(t) = \alpha\theta(t) \quad (4)$$

where  $\alpha$  is an arbitrarily chosen contraction coefficient and  $\theta(t)$  the unknown muscle excitation. Assuming the additive decomposition of the total strain into elastic and contractile parts:

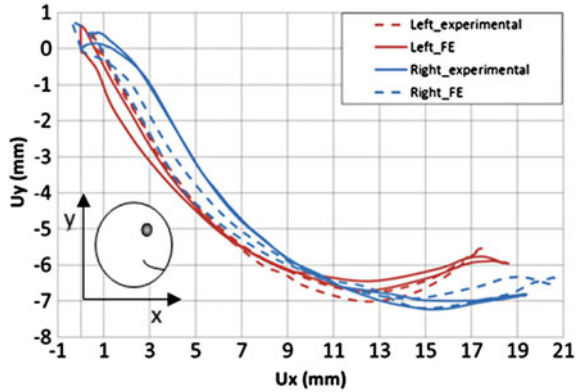
$$\varepsilon^T = \varepsilon^e + \varepsilon^c \quad (5)$$

and using definitions (2) to (4), the determination of the excitation profiles for all muscles of the model can be expressed as follows:

$$\theta(t) = \frac{1}{\alpha} \left( \ln \frac{l(t)}{l_o} - \frac{F(t)}{SE} \right). \quad (6)$$

These excitation profiles were calculated for every muscle using the files of the current length and force procured by the LifeMod. The method was first verified in the case of ample opening-closing jaw motion. The comparison of the experimental and finite element trajectories of condylar centres is illustrated in Fig. 4, taken from Alvarez [22]. It validates the method proposed.

**Fig. 4** Comparison of experimental and predicted trajectories of condyles' centres



## 2.5 Definition of Disc Shape and Positioning by Mean of Experimentally Supported Virtual Prototyping

Simulation model is an idealized mathematical description of real process and, because of a class of phenomena planned for investigation, is not purposed for the other classes. Thus, there is a need of creating a great number of models for simulation of various phenomena with various levels of particularity. Regardless of the software used, due to unavoidable inaccuracies of the built models and numerical calculations, it should be carried out verification of the used models, the obtained results and used design tools. The best way, as far as possible, is to experiment on a real object.

Since the computed tomography achieved on the volunteer did not permit to visualize soft tissues, various discs' geometries were created as following the method presented above. Because of the lack of data concerning the shape and occlusal position of the TMJ discs, the choice of these parameters was done in function of experimental trajectories of the condyle centres and the incisal point, i.e. the method of assurance assessment (validation) with the real object behaviour was applied. For this purpose, the offline identification of the model parameters of the TMJ was utilized.

While the precise design procedure depends on the particular device or process, we can distinguish certain techniques, which enable the achievement of the mechatronic design idea. They are as follows:

- virtual prototyping,
- experimentally supported virtual prototyping and
- rapid prototyping (i.e. real implementation) on the target system.



It is possible to design without the use of these techniques but, in many situations, it is inconvenient or very difficult as for example in the cases when:

- the monitoring system is being developed for the physically non-existent system or it is unacceptable to experiment on a working device,
- the system is unstable or weakly damped,
- the system is characterized by complex dynamic properties, being difficult for modelling.

In the TMJ case studied here, experimentally supported virtual simulation has been used. The latter, together with simulation implemented only on computing equipment having office software (i.e. virtual prototyping), are important techniques for mechatronic design.

Prediction of the results of the driving method, on a basis of computer simulation of the TMJ model commonly used in many former scientific and research works, is mainly affected by the disc parameters. Generally, the real time (RT) simulation should significantly improve precision of the prediction. However, the system/process studied is so complex that, for the reasons mentioned above, the whole system is not controllable in case of the present application. Behaviour of the TMJ is investigated in convention of open loop system. Thus, because of a lack of closed loop interaction between the measured and controlled signals, the RT simulation is not recommended in such a situation.

Virtual prototyping allows for fast and low cost analysis of many alternatives of the TMJ process performance, at the demanded accuracy level. Also it consents the concurrent design, thanks to supported means of verification of correct cooperation between all components (mandible, maxillary and discs).

On this stage, the research was characterized by:

- the well-verified constitutive characteristics of all components,
- engagement of real components of the research system,
- offline experimental identification of the TMJ system parameters,
- offline experimental determination of the appropriate parameters of the TMJ,
- online computation of the TMJ process of non-stationary model,
- assessment of the driving method's efficiency.

If the discs are not correctly shaped and positioned, there is a risk of their dislocation during the opening on the model leading to the wrong final trajectories and positions. A few adjustments on the discs shape and position were done until obtaining satisfactory trajectories of the mandible compared to the experimental records.

### 3 Results

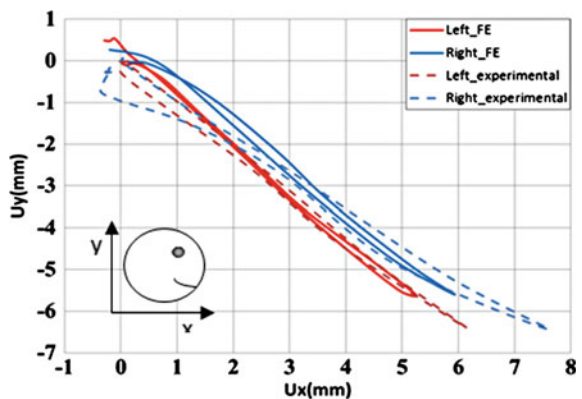
#### 3.1 Medium Opening-Closing Jaw Motion

The jaw motion of this finite element simulation was driven using the method presented in Sect. 4, that is, the mandible trajectory was not imposed as a boundary condition but resulted from the muscular activation and contact management between discs, fossa and condyles. Figure 5 shows the condylar trajectories during a medium opening-closing jaw motion. They reflect the best solution obtained in this study and correspond to shapes and positions of disc illustrated in Fig. 1. The blue (respectively red) continuous line represents the right (respectively left) condylar trajectory obtained by using the FEM. The blue (respectively red) dotted line represents the experimental trajectory of the same condyle at the same coordinate system. Results from the FEM are close to the experimental data and validate discs' geometries and positions.

Another way of verifying the quality of the obtained solution could be done by analysing the contact between discs and surrounding bones. The MSC MARC enables the visualization of contact status. The contact status, as all mechanical fields such as stresses or strains, evolves during the motion. The results presented below were exploited for the final opening of the jaw corresponding to 22 mm inter-incisal space. Figure 6 illustrates the contact status. Red zones indicate the part of the discs in contact with fossa (cranial view) or condyles (caudal view). The areas of contact localization are not symmetric (left and right discs) and are different on cranial and caudal faces. The cranial face of the right disc is in contact with the fossa on the lateral side while the contact between the left disc and the fossa is on the posterior area. The contact between right disc and condyle is situated in the middle of the disc when for the left pair this contact is experienced rather on posterior area.

The size of the contact areas is relatively limited indicating the low contact forces transmitted by the discs. The evolution of contact normal forces in the TMJ

**Fig. 5** Condyle trajectories in the case of a medium opening-closing jaw motion



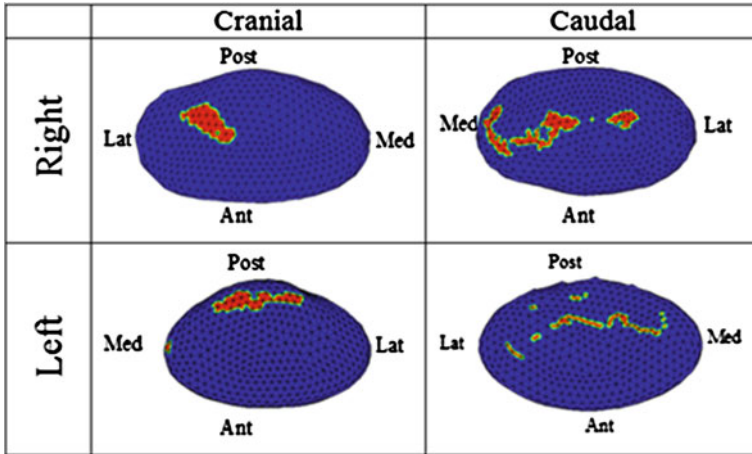


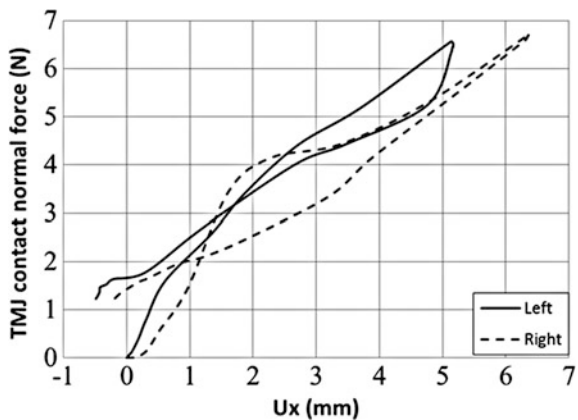
Fig. 6 Contact zone on the TMJ discs during the final medium opening

is illustrated in Fig. 7. This evolution is drawn in function of the condyles antero-posterior movement. Right condyle (dotted line) demonstrates higher motion magnitude than the left one (continuous line). The maximal antero-posterior displacements are of about 6.4 mm and 5.1 mm for the right and left condyles, respectively. Maximal contact normal force was reached at the maximum opening. Similar maximal values of this force were obtained for both discs equal to 6.39 N and 6.61 N, respectively, on the right and left one.

Figures 5 and 7 indicate that condyles did not come back to their initial positions and are submitted to some residual contact forces of about 1.2 N at the end of the opening-closing loop.

Since during the motion discs are compressed between the condyles and fossa, the minimal eigen, or principal, stress  $\sigma_{III}$  values developed in the discs are analysed

Fig. 7 Condylar contact forces evolution as a function of the condylar anterior displacement



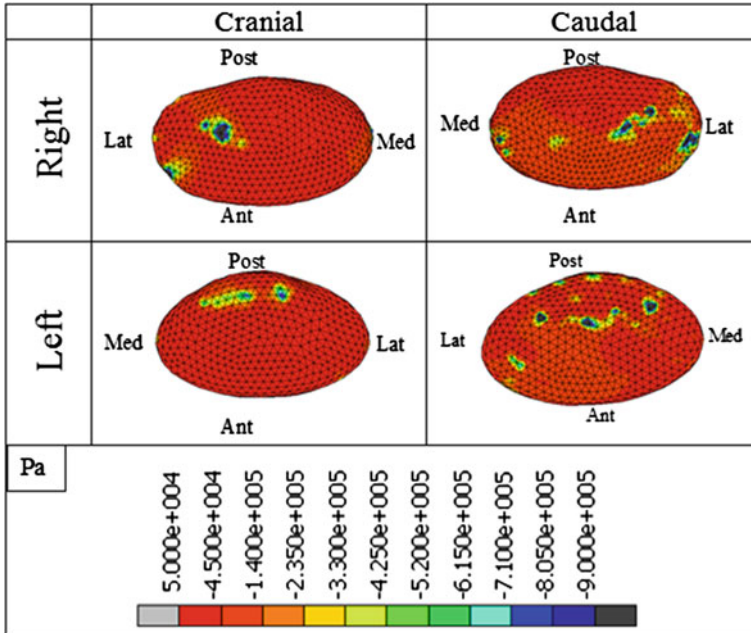


Fig. 8 Fields of minimal eigen stress  $\sigma_{III}$  on discs for 22 mm inter-incisal space

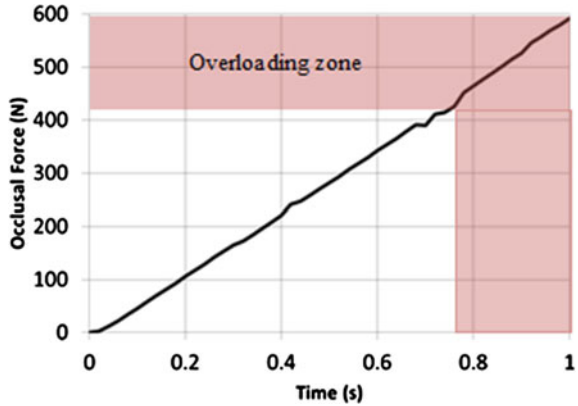
here. These stresses are expressed in Pa. Fig. 8 shows a comparison between the right and left sides.

Obviously, the contact areas identified above are submitted to the most intense compression. It can be observed that the stress concentration on the right disc is located on its lateral part while the maximum stress concentration appeared on the posterior location of the left disc. The value of the minimal principal stress exceeds locally  $-0.9$  MPa on both discs and its mean value on the contact zones was evaluated to be  $-0.15$  MPa.

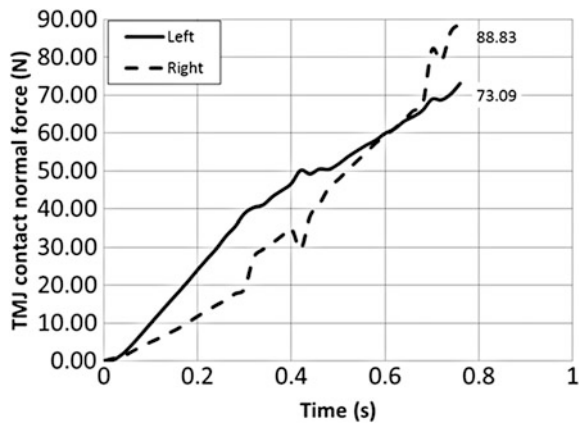
### 3.2 Immediate Clenching Simulation

Figure 2, comparing the finite element model and dental plaster mould occlusions, testifies the correctly reproduced occlusal scheme. Four anatomic occlusal points were detected with the finite element simulation; one on the second right molar, one on the left canine, one on the second left premolar and one on the second left molar. The elevating muscles of the model were activated to reproduce a typical normal clenching force observed during bruxism. The evolution of this force in function of time is illustrated in Fig. 9. Also, the overloading zone proposed by Nishigawa et al. [23] is indicated in this figure. This zone is defined by maximum or threshold force of 420 N.

**Fig. 9** Occlusal force evolution during clenching



**Fig. 10** Evolution of condylar contact normal forces during clenching



The evolution of the corresponding condylar contact forces is presented in Fig. 10 only for normal load zone. It can be observed that the forces grow linearly and quite symmetrically during the clenching. The maximum contact forces rise up to 88.8 N on the right condyle and 73.1 N on the left one. During the clenching process, the sum of these forces represents approximately 40 % of the clenching force. These important forces are transmitted through the disc soft tissue to the fossa bone.

To estimate the risk of mechanical damage of the discs, the contact areas and stress state are analysed below for the maximal admitted clenching force of 420 N.

According to Fig. 11, the contact areas on the cranial and caudal faces of the discs are more pronounced during the bruxism than during the opening-closing cycle.

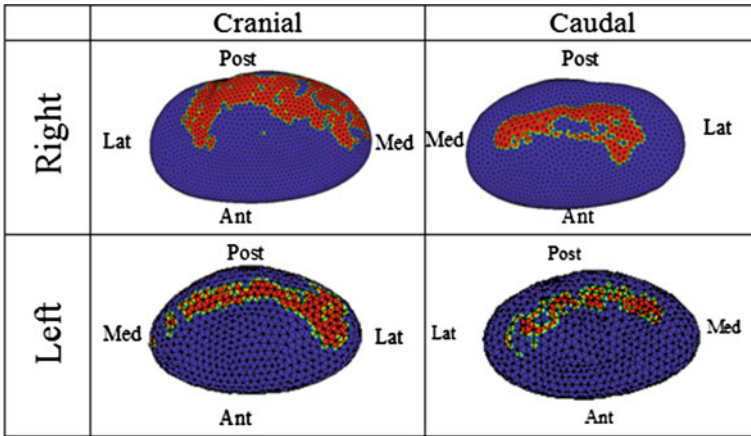


Fig. 11 Contact areas on discs at the maximal clenching state

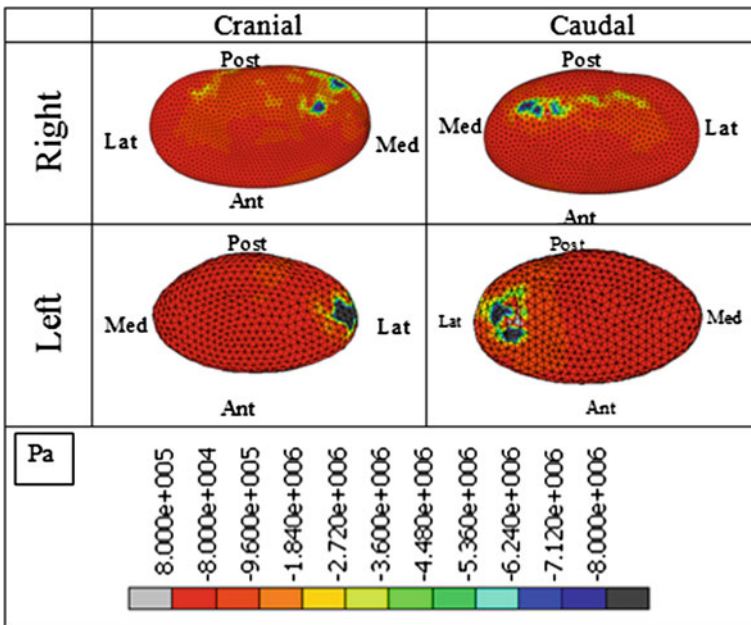


Fig. 12 Minimal principal value of stresses on discs during the maximum clenching state

These contact areas are quite similar on both discs. However, the contact of the right disc is mainly located on its central part while that of the left one is slightly shifted in posterior direction. The surfaces of the contact areas were estimated to be 38 and 100 mm<sup>2</sup>, respectively, on the left and right cranial faces and to 35 and

54 mm<sup>2</sup> on the left and right caudal faces. The values of the contact forces and the contact area surfaces enable calculation of the mean contact pressure values on the cranial and caudal faces of the discs. They are equal to:

- 0.89 and 1.04 MPa for the right disc on the cranial and caudal faces, respectively.
- 1.92 and 2.09 MPa for the left disc on the cranial and caudal faces, respectively.

The field of eigen stress  $\sigma_{III}$  exerted in the discs is illustrated in Fig. 12.

The most important compressive stress is concentrated in contact areas. Locally its value can be lower than  $-8$  MPa for both discs on cranial and caudal sides. The mean values of this stress on contact areas are comparable with the mean contact pressures presented above.

## 4 Discussion

The use of the finite element method for simulation of the jaw motion driven by muscular actions leads to very realistic models contrarily to the frequently presented simulations driven by the imposed mandibular displacement or force vectors. Even if the selected volunteer was a healthy case, the bone structures and mandible placement with respect to the maxillary appeared to be asymmetrical. Different initial locations and shapes of the TMJ discs reflected this asymmetry. These parameters play a crucial role on the disc contact area and contact forces during the motion and particularly for the final jaw medium opening state. In their study, Tanaka et al. [3] explained that the anterior disc dislocation has an influence on the load level transmitted by the disc. Here, for the same reason, the contact areas and pressures were found different on the left and right discs because of the differences evoked above. Nevertheless, Fig. 7 shows that these forces remain globally balanced during the all opening-closing motion, as expected for a healthy person without TMJ trouble. The residual contact forces and displacements obtained at the end of the simulated motion indicate that more than one cycle is necessary to stabilize the opening-closing loop. This fact is well confirmed experimentally [22].

The contact areas on discs obtained for the clenching load are pretty similar. In the literature [8], contact zones appeared generally on the middle part of the disc since the discs are usually constructed in occlusal position, i.e. with teeth in contact [7]. In our study, these zones are slightly posteriorly displaced. This is probably due to, the initial disc position and shape, which have been defined in rest position (with jaw slightly opened) and muscular actions were used to bring the mandible in occlusion. Significant condylar contact forces were obtained during the clenching simulation (about 80 N per disc). These forces represent approximately 40 % of the clenching load. Such a high level of compressive force can be exploited to explain the troubles and pains due to the clenching. No studies were found in the open literature analysing the contact forces on TMJ discs. Generally, the authors discuss the stress distribution in discs; see for instance Aoun et al. [13], Savoldelli et al. [8].

In accordance with Aoun's results, maximal compressive stress is located on the caudal face of both discs.

However, in our case, the magnitude of this stress is approximately five times lower than in case of Aoun's results. This difference is mainly due to the muscular action introduced and that is higher in Aoun's study than in this work. At the maximum clenching, forces on the elevator muscles are 55 N and 290 N, respectively, in this study and Aoun et al. [13] simulation.

## 5 Conclusions and Prospects

The volunteer treated in this study was identified as a person with symmetric condylar motion. However, the CT scanning revealed slight differences in morphology of the left and right TMJs. Consequently, the whole mandible with two discs and fossa was reconstructed and meshed.

The general-purpose finite element software, such as ABAQUS, MARC or ANSYS, does not have the ability to model the muscle's actions. An original method of muscular activation was developed and used to simulate two load cases, namely moderate jaw opening-closing cycle and teeth clenching.

The lack of data concerning the geometry and position of TMJ discs led to an optimization process based on the well-established method of the mechatronics design. In particular, the latter concerns one technique of the mechatronic design, called "experimentally supported virtual prototyping". Indeed, the experimentally determined condyle trajectories were used as a selection tool. The exploitation of this method enabled the optimal choice of these parameters, different for the left and right discs.

The obtained model allowed reproducing accurate anatomic mandible trajectories and also a physical occlusion. Values of forces determined on the basis of a TMJ model are consistent with literature data. The results of simulations allowed for:

- further improvement of the driving method and its evaluation,
- the evaluation of improvement of the driving method results prediction precision.

In the future works, the model can be improved by a stronger stabilization of the discs by inserting a 3D model of TMJ capsules, instead of the 1D representation of these organs. Moreover, the "experimentally supported virtual prototyping" procedure was applied in a quasi-manual manner. This task can be highly improved and accelerated basing on the following general indications. For the purpose of online computer simulations, we envisage to prototype some authorial computer programs, written in a high-level programming language (e.g. Fortran, C, C++), and being supported by commercial software used. Promising results of the simulation shall allow developing and implementing the driving method on the target system



(i.e. the examined patient), in which all components are real. Research on the target system will enable increasing:

- the performance of the driving method of the mandible motion by the method of muscular contraction, on selected volunteer,
- the assessment of efficiency of the driving method during real performance of the TMJ.

It is expected that the obtained results will confirm performances of the driving method in case of one TMJ of the volunteer. The solution enables the monitoring of the TMJ behaviour during routine functioning without the need for interference with the structure of the body, but only by adapting a portable bench on the mandible. Thanks to the above, the proposed solution will entail lower price in comparison with the apparatus investment.

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