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DEVELOPMENT OF A TEMPOROMANDIBULAR JOINT PROSTHESIS BASED ON THE KINEMATIC BEHAVIOR OF A HEALTHY SYSTEM

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Abstract. The mandible is one of the structures of the maxillary buco system, which has two joints attached to the same bone. The joint presents a very particular geometry that develops several rotational, lateral and translational movements. Temporal Mandibular Joint (TMJ) is a small joint that is located in the base of the skull with the maxilla. There are several factors that result in malfunction of TMJ and consequently the use of prostheses. Due to these problems, the patient may undergo an operation to replace the condyle and the insertion of an artificial condylar fossa. However, the current prostheses partly supply the original movement. Among the main disadvantages of the current prosthesis can be mentioned the loss of amplitude of the mouth at the moment of maximum opening and the loss of protrusion in relation to the occlusal contacts. This work seeks to approximate the fixation geometry of the prosthesis according to the patient's resonance, in order to facilitate the fixation of the component and to approximate the kinematic movements to a healthy mandible.

Keywords: Mechanisms, BioCAD, Prosthesis, TMJ, Joint

1. INTRODUCTION

Temporal Mandibular Joint has the main function of allowing the sliding and, consequently, the functioning of the lower jaw (mandible), besides being one of the most used in the body, especially during speech and feeding. The symmetry of the TMJ region is formed by a superior fossa rounded at the base of the skull where the condyle fits. Between the condyle and the fossa there is a cartilaginous disc that allows the movement of the joint.

Among the main factors that cause the substitution of this joint, we can mention: accidents involving shocks, traumas, pathologies, congenital deformities of the condyle branch, bites with hard objects (Landes *et al.*, 2013), rheumatic arthritis (Ahmed, 2015), osteoarthritis (Lee *et al.*, 2017), wear on the surface of TMJ (Singh e Detamore, 2009; Mehra *et al*, 2009), and bruxism (Seaton, 1979).Patients with chronic pain use prosthetic surgeries as an alternative for an improvement in clinical status. According to Souza (2009), the main disadvantages of TMJ prostheses are the size limit of the prosthesis due to the region where it will be inserted, the loss of translation movement caused by laterality and the loss of protrusion due to the deinsertion of the lateral pterygoid muscle. Tanaka and Koolstra (2008) describe that throughout history the TMJ was not seen as an important joint and the first studies did not taken into account the loads involved in this system. For Grant (1973) the movements were poorly understood and illustrated. For this reason, experiments of lateral measurements of the skull with several fixed points of reference allows to conclude that the actual trajectory of the mandible is significantly different compared to the situation that uses the condyle as a rotating stationary center.

The changes in the thinking of the mechanical functioning of the mandible and later analysis with scientific rigor comprise a very recent path in the history of dentistry and medicine. The present prostheses are limited in terms of replacement the functions of the biological physiognomy (Van Loon *et al*., 1995; Mesnard and Ramos, 2012) and result in several problems due to contact with bone and stress propagation (Chowdhury *et al*., 2011).Among these limitations can be mentioned the loss of amplitude of the mouth at the moment of maximum opening and the loss of protrusion in relation to the occlusal contacts.

According to Posselt (1957), the opening stages of the mandible were divided into five stages: RC (Retracted Contact Position), A (End of Exclusive Hinge Movement), B (Maximum Opening of the Mouth), OC (Regular Occlusal Contact) and P (Previous Contact Position). These points are showed in Fig. 1.
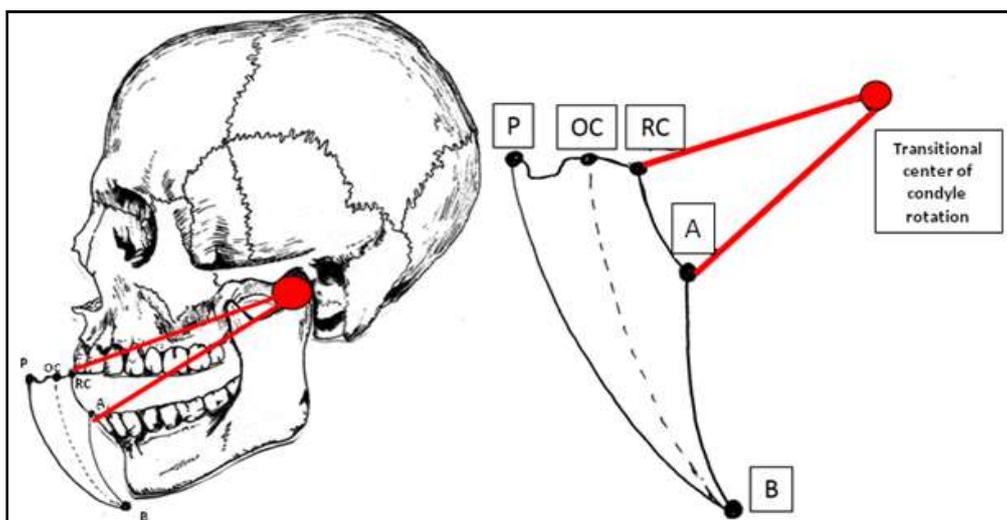
Figure 1. Posselt diagram projected on the sagittal plane

Based on Posselt diagram, the researchers Molina (1995), Steenks and Wijer (1996) and Madeira (2004) describe four main movements of the healthy mandible by the sagittal plane of the individual face, showed in Fig. 2.
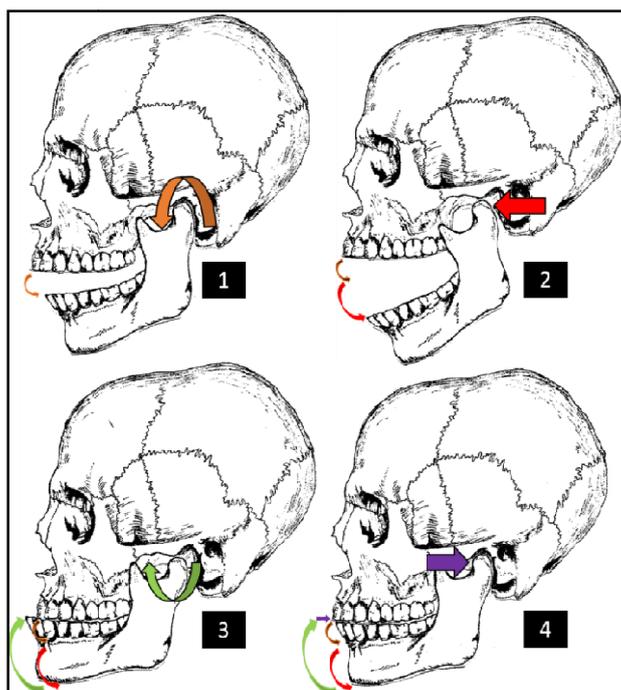

Figure 2. Four main moments of opening and closing the mouth. The movement has been indicated in the condyle and the incisor teeth of the mandible.

Understanding these movements is crucial to the design of more efficient prostheses. Within this context, this work aims to evaluate the kinematic movements of a healthy system to be based on a design of an innovative prosthesis.

2. METHODOLOGY

The methodology was divided into two main parts. The first part is a standard anatomical CAD model for the simulation of a new mechanism. The second part is a rigid body simulation of the proposed mechanical component inserted in the CAD model. The procedures of this methodology will allow plotting the time-displacement graphs of specific points that can be compared with the literature of the movement of a healthy jaw.

The anatomical models were obtained through Computer Aided Design (CAD). BioCAD is a CAD tool protocol adapted for biological structures and complex geometries. The database models of three-dimensional technologies research group (NT3D) of Renato Archer Information Technology Center represent standardized anatomical structures, but preserve a certain degree of interpersonal variability. This methodology is based on the identification and selection of the main anatomical landmarks, in the same way that this technique is also adopted in engineering geometric modeling. The use of modeling in BioCAD is due to the computational simplification of the model in terms of size for the subsequent processing of rigid body analysis.

The original *stereolithography* (*stl*) format of the jaw was obtained by capturing images from magnetic resonance imaging. From the Invesalius® software, the images were grouped and reconstructed to form the three-dimensional model. Although the *stl* model shows more details of anatomical structures, the large number of surfaces became the simulation unfeasible. To circumvent this problem, one of the techniques employed is to use NURBS surfaces. The method Non Uniform Rational Basis Spline (NURBS) is a mathematical representation of a 3-D geometry delineated through 2-D lines. The main advantage lies in the demand for complex surfaces, such as anatomical surfaces that do not follow a consolidated geometric pattern. Fig. 3 shows the steps to obtain the final geometry. The CAD software used was the Rhinoceros® version 5.0 that enables a wide range of surface tools that best fit the anatomical profile designed. The adaptation for NURBS surfaces allows a close configuration of reality without using a high number of surfaces in regions of little interest.

The head of the mandible, the condyle and the mandibular fossa are not congruent, and the articular disc allows the adaptation of the movement in this region. There is also variation by age factor in newborns, the joint is shallow and the mandibular fossa is not well delineated. The geometric formation of this area is influenced by a large part of the occlusal contacts (Steenks and Wijer, 1996). For these reasons, the study of this articulation was modeled for a masculine healthy skull, with 30 years of age, where the forms are defined and represent a more stable situation for the study of prostheses. This group was also selected to be able to compare to the displacement curves of Missaka (2010).

(a)   (b)

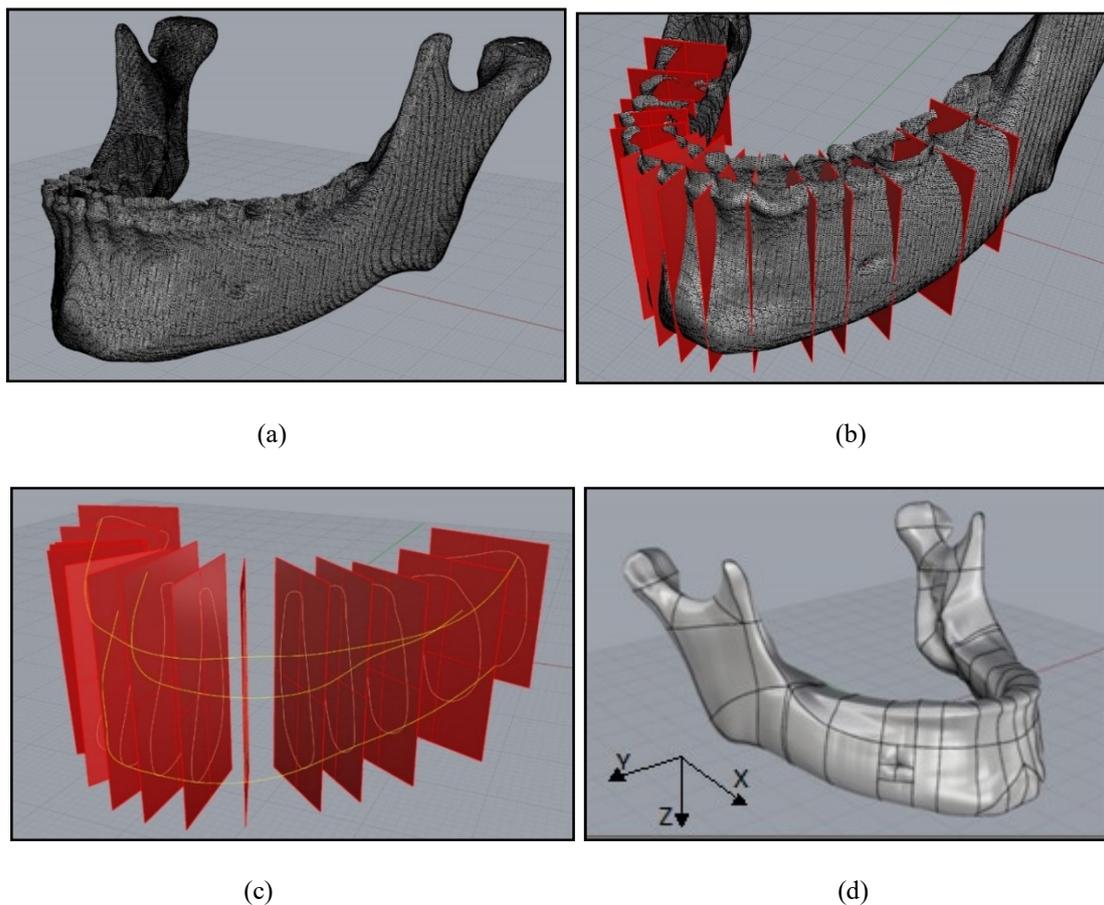(c)   (d)

Figure 3. BioCAD procedure of jaw design in Rhinoceros Software. (a): Model in *stl*. (b): Plans for the main anatomical frames. (c): Intersection lines between the planes and the *stl* model. (d): Reconstructed mandible with NURBS surfaces.

The next step is related to the delimitation of movement that the prosthesis will fulfill. In order to allow a comparison, the rotation and translation movements were analyzed together by the graphs in the X and Z coordinates, the same ones used in the comparative graphs of the literature described in the work of Missaka (2010).

Patients who undergo prosthetic installation operations lose the lateral pterygoid muscle action that is responsible for the laterality movement because it is inserted over the capsule that ruptures. Thus, this study will not analyze the Y-axis motion graph, which represents the lateral movement. The second delimitation made is the kind of analysis: a planar form mechanism (cam system) was adopted in order to facility the visualization of a profile. The cam is a solid body whose movement of its trajectory defines the movement of the second body attached to it, the follower (Doughty, 2001). The cam provides a similar form of movement in which an analogy can be made to the study system. Considering the opening angle as the primary coordinate and the displacement of the mandible in the Z-direction as the secondary coordinate, then the system motion can be described from the displacement functions. The displacement function is the mathematical description that relates the follower's displacement to the angular position of the cam.

The displacement diagram is a reference to a rotation cycle. According to Doughty (2001) choosing a curve that best meets the design of the displacement function of a given design is an art. In the literature there are studies and well-defined functions, among which can be mentioned the sine, parabolic, cubic, etc. However, in some works, a curve must be adjusted according to positions obtained through experimental data. In this case, the work done by Missaka (2010) can be used as a data source for the lifting of the displacement function that the mandible should perform.

The software used in the modeling and simulation of the mechanisms was the PTC Creo Parametric®, version 3.0, in the section of Rigid Analysis of Mechanisms. Among the featured tools, the software has a specific area that determines the tracks of the cam and follower. The chosen cam profile was defined using a plane of intersection of the condyle of the *stl* model. The plan adopted should intersect the main region of contact of the condyle with the condylar fossa. The trajectory of the condylewas defined from the work of Ramos and Mesnard (2014). By simulating the opening of the Christensen® Prosthesis was delimited the prosthesis contact line along the movement over the portion of the condylar fossa. Even based on the opening movement of prosthesis, the contact line between the prosthesis and the fossa was considered similar to the opening of a healthy mouth due to the application of a personalized prosthesis. Taking into account the method proposed by Ramos e Mesnard (2014), the Fig. 4 shows the steps accomplished in this work to obtain the intersection line between the fossa and condyle that results the final condyle profile used in these numerical simulations.
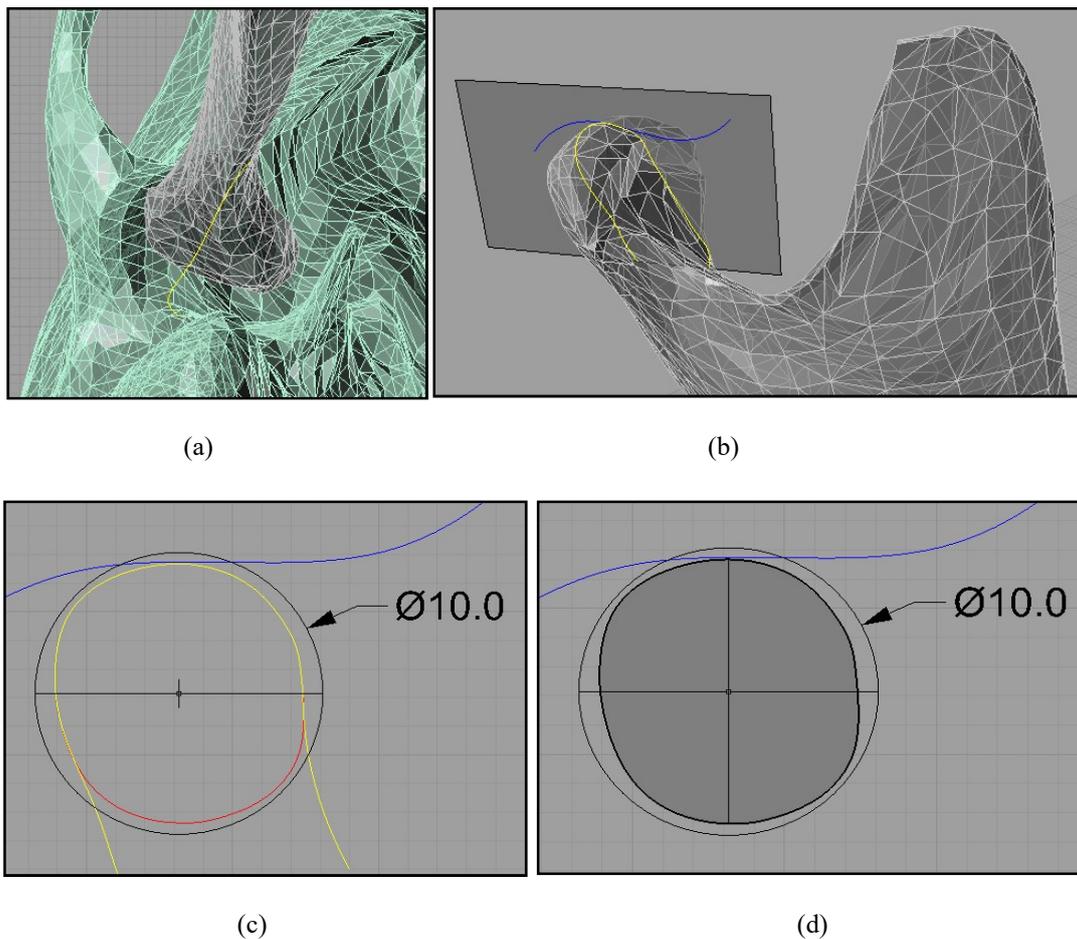

(a)

(b)

(c)

(d)

Figure 4. Condylar profile procedure. (a): Placement of the mandible at the maxilla. (b): Intersection planes following the trajectory line (blue) in the condylar geometry (yellow). (c): Definition of the lower cam region. (d): Final planar profile

As can be observed in Fig. 5, the final prosthesis is mounted to the circular base in order to prevent the deinsertion under critical displacement. In Fig 5, the point P3 represents the point analyzed by Missaka (2010) and will be the standardized distance used to compare the graphs plotted.
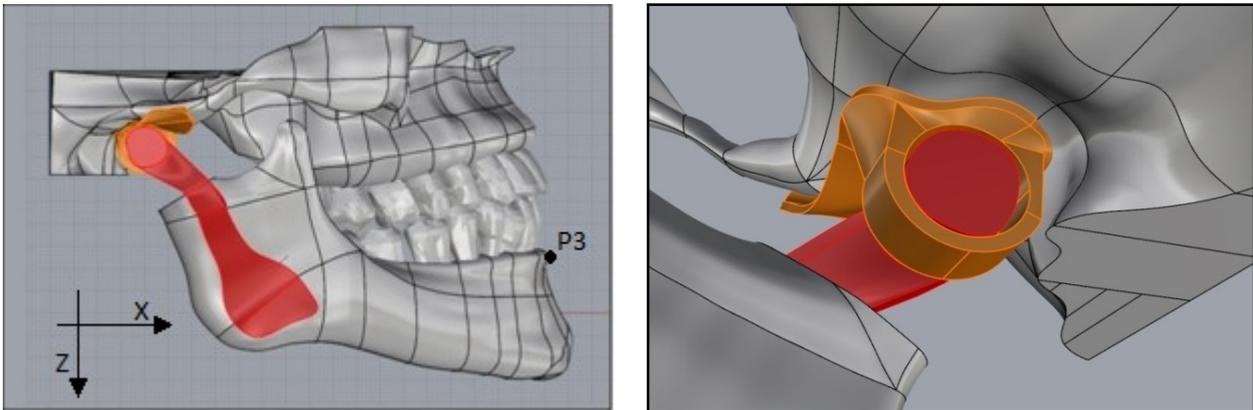

Figure 5: Final set of jaw model with the proposed prosthesis mechanism.

In the previous step, the anatomical modeling was made for half the jaws to restrict the case study for patients with a single fracture and a healthy opposite side. The constraint conditions imposed on the model were symmetry in the medial plane and the fixed condition in the upper region of the maxilla. In order to perform numerical simulations, a function of the cyclic movement for opening and closing in the maximum opening has been defined using a servo motor to lead the movement in the prosthesis. Comparing the graphs in the Z direction obtained by Missaka (2010) and restricting the time of the motion cycle, the Eq. (1) and (2) define the function used in the PTC Creo Parametric® program.

$$F(t) = P*\cos (360*t/T + B) \tag{1}$$

$$F(t) = -20*\cos (360*t/40 - 90) \tag{2}$$

Where P is the peak value, t is the time in seconds, T the period of the function and B is the constant that determines the start of the function with respect to 0 °. Among the options of functions provided by the program, F (t) was the one that best represented the cyclical way of opening and closing movement and respects the displacement per time.

Several simulations were performed by varying the internal diameter of the prosthesis base. Starting from the value of 10 millimeters, simulations were also made with the diameters of 10.35 mm and 9.85. The diameter of 9.85 mm represents the smallest diameter that allows the free motion of the condylar prosthesis during the rigid body simulation. Smaller dimensions are desired in this type of design because the prosthesis implantation region is critical in terms of surgery and physical comfort of the patient.

3. RESULTS AND DISCUSSION

When the profile was adopted, three diameters of the circular base were simulated. Starting from the diameter of 10 mm the first simulation was made to verify the displacement of the direction X (Fig. 6) and Z (Fig. 7).
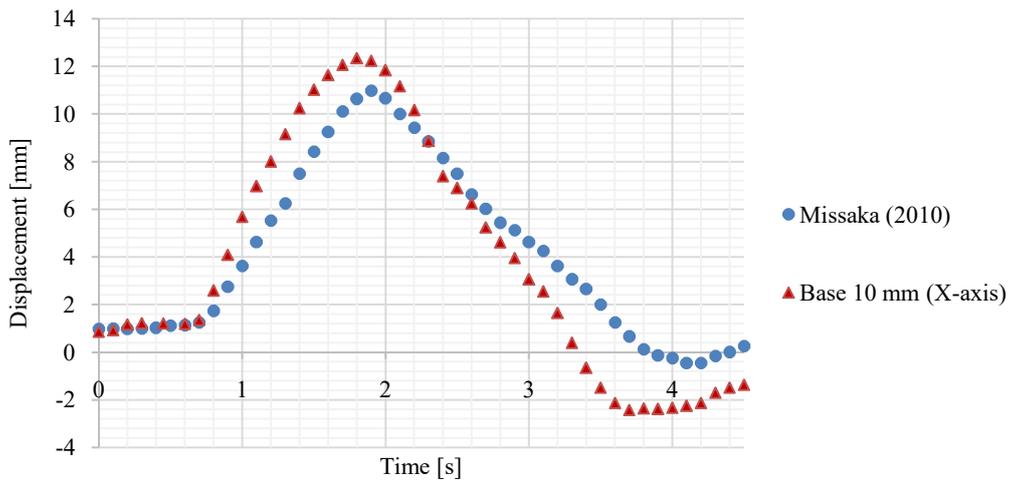
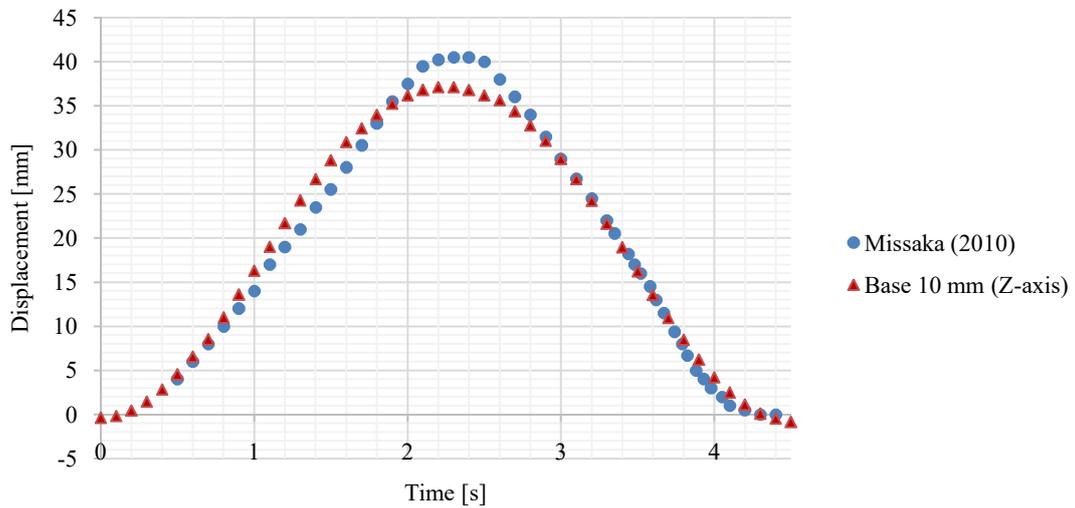Figure 6: Comparative graph of displacement for X-axis in 10 mm diameter.

Figure 7: Comparative graph of displacement for Z-axis in 10 mm diameter.

For the second simulation, the diameter of the base hole was increased to 10.35 mm, in order to check if the value was restricting the movement or whether the adopted profile should be modified.
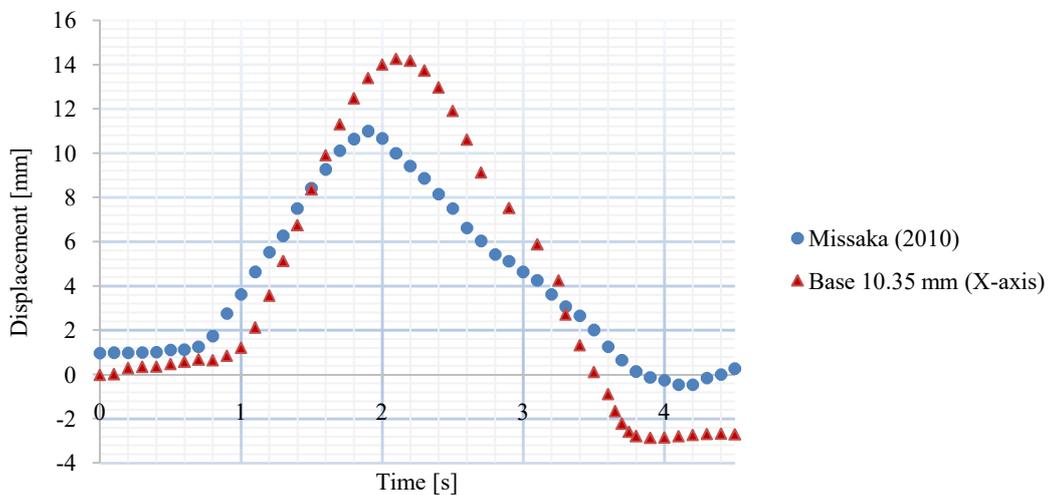

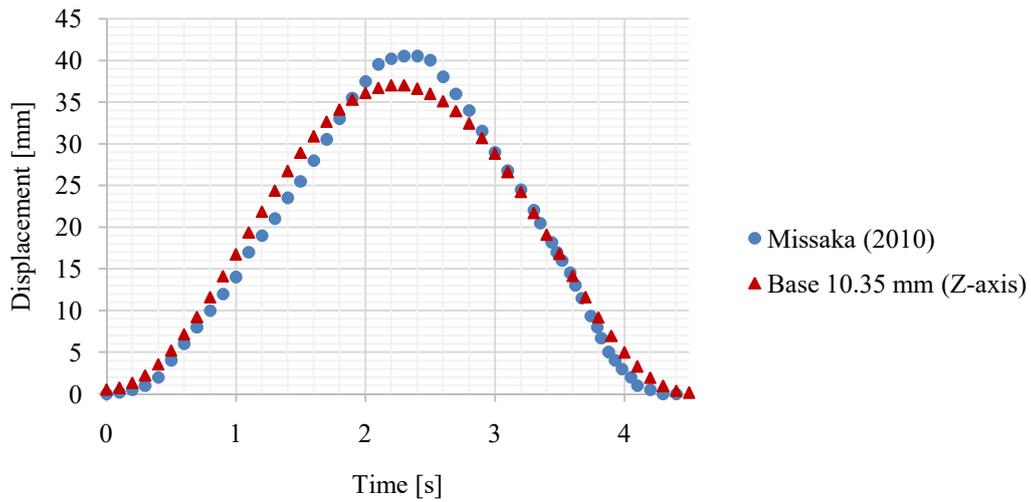Figure 8: Comparative graph of displacement for X-axis in 10.35 mm diameter.

Figure 9: Comparative graph of displacement for Z-axis in 10.35 mm diameter.

In terms of displacement amplitude, the behavior of the Z direction of the 10.35 mm (Fig. 9) and 10 mm diameter are very similar. However, the gradient of displacement in the X direction (Fig. 8) has increased, moving away from the curves of Missaka (2010). Therefore, in this new simulation, the smallest diameter value is used. In this simulation was considered the minimum circumference that allowed the free movement of the cam. The diameter of 9.85 mm was adopted in this strategy in order to make a new verification.
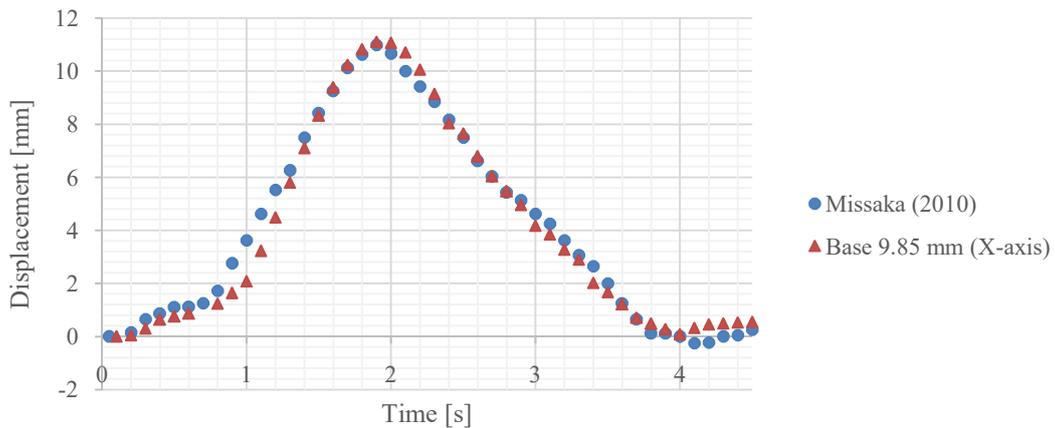

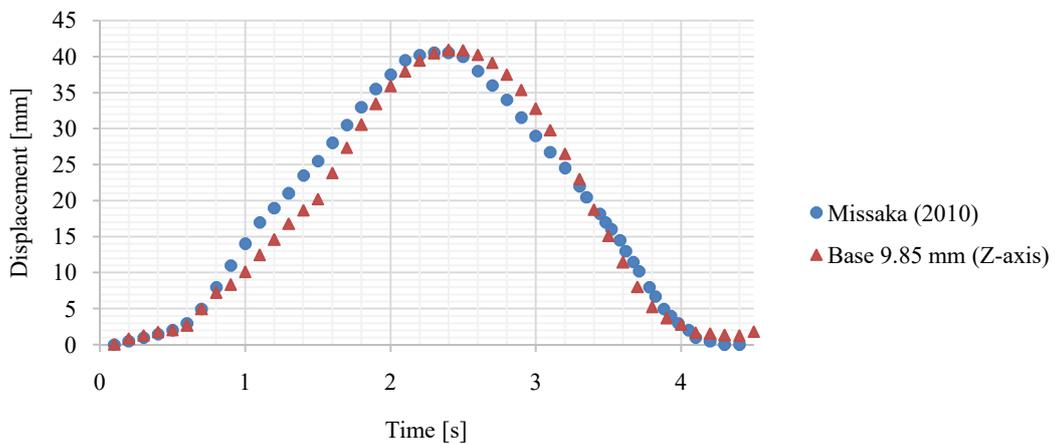Figure 10: Comparative graph of displacement for X-axis in 9.85 mm diameter.

Figure 11: Comparative graph of displacement for Z-axis in 9.85 mm diameter.

Regarding the X direction, there was a decrease in amplitude, with a difference of 5.8% at the instant of 1.8 seconds between the points obtained and that of the literature. The graphical displacement results (Fig. 10 and 11) were close with the Missaka (2010) charts.

Once the graphs were checked, it was possible to observe that the points of the 9.85 mm have satisfactorily followed the results of Missaka (2010). In general for the condylar profile joint, the decrease in the base diameter causes a decrease of the X and Z direction displacement. For values starting from 10.00 mm and increasing up to 10.35 mm, the Z amplitude remains stable, while the amplitude in X suffers a considerable increase (distancing from the natural displacement obtained by Missaka (2010)). The movement restriction within the base is predominantly at the top of the base. Mean relative error indicates the variability between samples and can be checked for each simulation in Tab.1.

Table 1: Mean relative error for the three bases according to the diameter.

Diameter [mm]	X-axis [%]	Z-axis [%]
9.85	5,08	6,19
10.00	16,41	10,25
10.35	51,22	11,40

4. CONCLUSION

The present study allowed the mechanical analysis of a biological component (TMJ) with emphasis on the kinematic recovery of the assembly through a joint. Between the difficulties encountered throughout the work, two points can be mentioned: the representation of the anatomical condition (bone surfaces) of a patient with malfunctioning of the TMJ and the delimitation of a profile capable of recovering the rotation and protrusion. Both characteristics are implicit in the work and the understanding of simulated physical effects was the solution to such difficulties. The results throughout the project allowed the use of the cam profile gasket as a possibility of prosthesis mechanism for cases involving the set of rotation and translation and verify an approximation in the response of the aperture displacement curves.

As a future proposal the simulation in finite elements will allow to evaluate the regions with greater mechanical stress, taking into account the elastic deformation of the joint. Main maximum stress variation in contact regions of the base gasket indicate regions required in traction and compression that can cause wear and fracture.

5. ACKNOWLEDGEMENTS

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