Analysis of temporomandibular joint prosthesis using finite element method and a patient specific design

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A R T I C L E I N F O	A B S T R A C T
	In this study, a 3D finite element model of an intact mandible was used for the simulation of the movement of the lower jaw and analysis of the effects of TemporoMandibular Joint (TMJ) prosthesis replacement on the jaw movement. Seven bundles of muscle fibers were inserted in their appropriate positions following anatomical data. Digastric, geniohyoid and lateral pterygoid muscles were considered for opening the mouth while medial pterygoid, superficial masseter, deep masseter and temporalis muscles were used for closing the mouth. Then, the TMJ was replaced by two different types of TMJ prostheses in the same way as in a surgery operation. One of the prostheses was designed based on anatomical shape of the ramus and condyle of the mandible while the other one was taken similar to the commercial TMJ prostheses. Eventually, all three models underwent an opening jaw simulation and produced an identical range of motion while were mismatched in other parameters. The results show that since the anatomical TMJ prosthesis resembles the shape and structure of an intact mandible; therefore, it is more capable of simulating the motion of mandible compared to the commercial TMJ prosthesis. Furthermore, better contact between the anatomical TMJ prosthesis and the mandible leads to lower stress distribution in comparison with the commercial TMJ implant. Finally, as the amount of muscle forces and strain in anatomical TMJ prosthesis replacement are less than the commercial one, the patient needs to make less effort to move the mandible and open the mouth.

1. Introduction

Temporomandibular joint (TMJ) is one of the crucial joints in the human body which allows the rotation and translation of the mandible (May *et al.*, 2001). The complicated movement of the TMJ leads to significant tasks such as speech, swallowing and chewing. In the past decades, several attempts have been taken in order to enhance life-time expectancy among patients who suffer from maxillofacial problems. In spite of great endeavors, some problems are still remained unsolved or demand further investigations. In the realm of maxillofacial diseases, temporomandibular joint disorder (TMD) is one of the common issues which have been ensued from trauma, cancer or a sudden hit as in an accident (Hsu *et al.*, 2011; Ingawalé & Goswami, 2009; Chowdhury et al., 2011). In severe circumstances, TMJ replacement with an artificial one is an inevitable choice. In one study, the researchers analyzed the influence of TMJ replacement on maximum incisal opening of the jaw, jaw pain intensity and eating

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disorders in the long-run and the results showed good improvements in all of them (Sanovich *et al.*, 2014). TMJ prosthesis has had many modifications since its advent to the market up to now (van Loon *et al.*, 1995). However, in order to optimize the TMJ prosthesis, a comprehensive knowledge of maxillofacial anatomy; especially, the shape and structure of mandible and maxilla is required. In addition, the stress analysis of prosthesis under the action of muscle forces is another important factor for the optimum design. Hence the involved muscles in jaw opening and closing should be studied to understand the effect of muscle fiber orientations and the activation level of each muscle in the process of mastication and jaw opening (McFarland, 2014). Recently, Huang *et al.* (2015) have investigated the stress and stability of TMJ prosthesis and the whole mandible during alternative clenching tasks. They modeled the muscle actions through specified muscle forces applied on muscles' attachment points while have not studied the dynamics of jaw movement and the effects of variation of direction and magnitude of muscle forces during these tasks. In 2011, Ramos *et al.* (2011) studied three different bodies of a TMJ prosthesis with a same head and concluded that the highest displacement was noticeable in anatomical prosthesis body in the X and Z directions. These results were obtained while only deep and superficial masseter, medial pterygoid and temporalis were considered and digastric and geniohyoid muscles were not considered during the simulation. Also, kinematical features of the motion of mandible were not studied to compare the outcomes between these three models. In another similar study, Arabshahi *et al.* (2011) examined the TMJ prosthesis geometry and considered the effects of imitating the ramus of an intact mandible on the body of TMJ prosthesis in order to compare the produced stress distribution, mainly near screw holes. The obtained results show that the stress distribution in anatomical TMJ prosthesis is less than the commercial one. However, in the process of simulating the mandible incisal opening, the geniohyoid muscle was not considered during the simulation. Furthermore, no other parameters such as the reaction force, strain or range of motion were studied to see the real effects of this change on the desired items. In another study, the researchers investigated intramedullary TMJ prosthesis with an external fixation of the implant and concluded that the strain distribution is improved in the intramedullary implant compared to the commercial TMJ prosthesis (Ramos *et al.*, 2014). In addition to the shape of the TMJ prosthesis, it has been shown that the type and orientation of the screws' fixations play a pivotal role in the stress distribution on the implant (Roy *et al.*, 2011). By comparing the stress distribution between the parallel and zig-zag orientations of screws' positions along the TMJ prosthesis, it was concluded that the stress distribution in zig-zag orientation is much more than the parallel arrangement of screws. In addition to screw orientations, the number of fixed screws is another important issue. Ramos *et al.* (2015) have investigated different numbers of screws and found that three screws are fair enough to ensure about the stability and functionality of the TMJ prosthesis. Among various materials which are used to manufacture the TMJ prosthesis, titanium is known as the best due to its lower generated strain in the TMJ prosthesis compared to other materials (Arabshahi *et al.*, 2012). Bone quality of the mandible is another factor which greatly contributes to the stability and functionality of the TMJ implant, it is investigated that the higher the modulus of elasticity of mandible, the more stability is observed during different muscle activations (Hsu *et al.*, 2010).

The main aim of this study is to manipulate the movement of mandible according to different muscles' activations in order to control the motion of the lower jaw. Two different TMJ prostheses were implanted in their appropriate positions and fixed by screws to imitate the function of the left condyle. Ultimately, both TMJ prostheses underwent the same simulation, then, different parameters involved in the movement of the mandible were compared and the results were discussed.

2. Materials and Methods

2.1 CAD model

The 3D model of an intact mandible, maxilla and hyoid bone were obtained by means of CT (Computational Tomography) scan images of a healthy patient. In this model more than 250 slices were

used in order to form the complete point cloud model of the skull. In the next step, the point cloud model was converted into a solid model. Finally, seven bundles of muscle fibers were inserted in their appropriate positions following anatomical data. Three muscles (Digastric, Geniohyoid and Lateral Pterygoid) were considered for opening the mouth while four muscles (Medial Pterygoid, Superficial Masseter, Deep Masseter and Temporalis) were used for closing the lower jaw (Fig. 1(a)) (McFarland, 2015). Each muscle has a different physiological cross section area. The psychological cross section of each muscle was distributed uniformly between the muscle fibers of every individual muscle. Table 1 shows the muscles of mastication and the corresponding cross section area (Brugman, 1997).
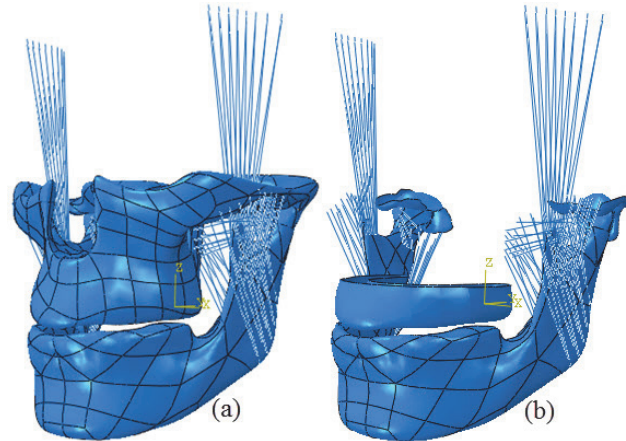

Fig.1. (a) The 3D model of mandible, maxilla, hyoid bone and the muscles of mastication, (b) The simplified model of mandible and maxilla

Table 1. Muscles of mastication and the corresponding physiological cross section area (Brugman, 1997)

Muscle	PCSA (cm^2)	Number of Fibers	PCSA (cm^2) for every individual muscle fiber
Sup Masseter	6.82	10	0.682
Deep Masseter	3.488	8	0.436
Temporalis	5.536	16	0.346
Medial Pterygoid	2.97	9	0.33
Lateral Pterygoid	1.884	8	0.2355
Geniohyoid	0.968	8	0.121
Digastric	1.16	4	0.29

In order to decrease the cost of computation, only the interacting segments of maxilla which are in touch with the mandible were considered and the rest of maxilla was omitted from the simulations. Three major parts of maxilla which were left in the simulations are the left articular fossa, right articular fossa and the upper jaw teeth. Figure 1(b) shows this simplified model.

Two different types of TMJ prostheses were used for TMJ replacement in the mandible. One was modeled based on available commercial TMJ implants and the other one was designed using patient's anatomical specifications. For improving the design of the TMJ prosthesis, the anatomical features of mandible were taken into account and applied to the new design. According to the published data, three screws are fair enough to fix the TMJ prosthesis to the mandible (Ramos *et al.*, 2015). To install the TMJ prosthesis in its proper position, the steps of surgery were simulated to complete the model. Lateral Pterygoid muscle was removed, the damaged condyle of the mandible was cut and the TMJ was fixed to the jaw using three screws. Figure 2(a) and 2(b) shows the commercial and anatomical TMJ prostheses and Figure 3(a) and 3(b) represents the TMJ prostheses in their desired position.
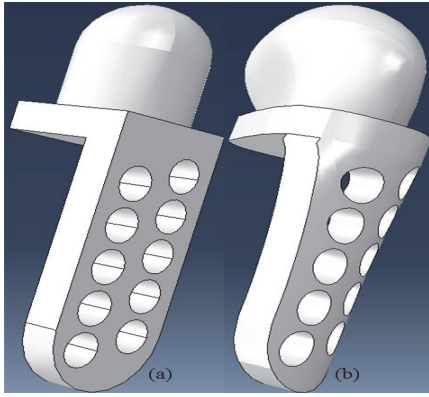
Fig. 2. (a) Commercial TMJ prosthesis, (b) Anatomical TMJ prosthesis
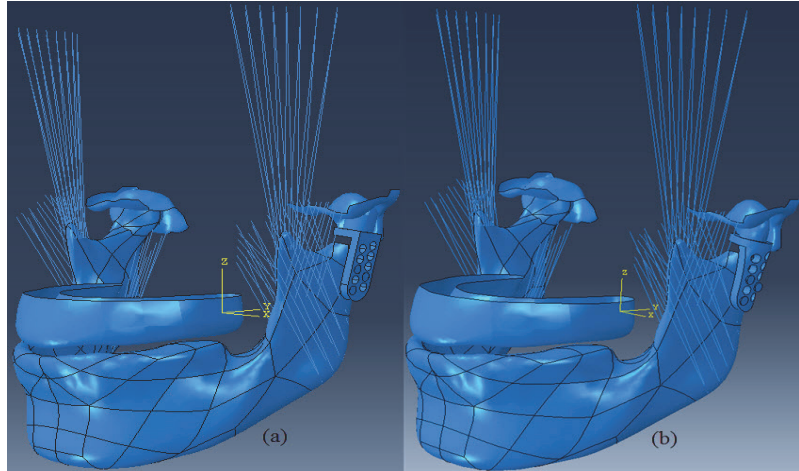
Fig. 3. (a) Insertion of commercial TMJ prosthesis in its proper position, (b) Insertion of anatomical TMJ prosthesis in its proper position

2.2 Boundary Conditions

During the simulation, the left articular fossa, right articular fossa, upper jaw teeth and hyoid bone were fixed in all six degrees of freedom. Besides, a surface to surface contact was defined between the articular fossa and the condyle, and between lower jaw teeth and upper jaw teeth to avoid the penetration of these parts during movement. Furthermore, one end of each muscle fiber was tied to the mandible while its other end was attached to the corresponding anatomical attachment point. Consequently, the movement of mandible could be manipulated by the activation of different muscles and controlled via the contacts between the lower jaw and the upper jaw teeth, and between the articular fossa and the condyle.

Apart from the aforementioned boundary conditions, an interaction was defined between the screws and the prosthesis, and between TMJ prosthesis body and ramus of the mandible. An initial stress was also exerted to fix the screws in the mandible.

2.3 Material properties

Titanium alloy is considered as the TMJ prosthesis material due to its fine compatibility to human body and good mechanical properties (Arabshahi *et al.*, 2012). Linear elastic material property was considered for different parts of the model. Bone, muscle and titanium alloy properties of the finite element model are listed in Table 2.

E: Modulus of elasticity

ρ: Density

ν: Poisson's ratio

Table 2. Material properties of bone, muscle and titanium alloy

	E (GPa)	ρ (Kg/m^3)	ν
Bone (Pal, 2014)	18	1810	0.3
Muscle (Raustia *et al.*, 1998)	4.00E-05	1075	0.499
Titanium (Arabshahi *et al.*, 2011)	110	4500	0.3

2.4 Muscle activation

In a Hill-type model the muscle active force can be expressed as:

$$F = Af(l)f(v)F_{max},\tag{1}$$

where A is the activation command for the muscle, $f(l)$ and $f(v)$ are force-length and force-velocity characteristics curves of active muscle force and F_{max} is the maximum voluntary force (MVF) of the muscle. This internal force of muscle can be modeled through a virtual thermal strain idea. The thermal strain as a function of temperature difference (ΔT) is given as:

$$\varepsilon = \alpha \Delta T,\tag{2}$$

where α shows the thermal expansion coefficient. The change in its length from its initial length (ℓ_0) becomes: $\Delta \ell = \alpha \ell_0 \Delta T$. Hence the force exerted by the muscle in an isometric way can be computed as:

$$F = SE\alpha \Delta T\tag{3}$$

where S is the cross section area and E is the modulus of elasticity.

According to these formulas, the virtual thermal strain idea can be easily replicated with the Hill-type muscle model. In other words, parameters which are involved in exerting muscle forces in the virtual thermal strain theory can be matched with other parameters in the Hill-type muscle model. In equation number (3) the temperature variation as a function of time can be regarded as the activation command in the equation (1). Furthermore, the multiplication of thermal expansion coefficient with the cross section area of the muscle in equation (3) can be a representative of maximum voluntary force (MVF) in equation (1). Finally, the modulus of elasticity as a function of length in equation (3) can be indicated as a function of length in the Hill-type muscle model. In this way, the virtual thermal strain model has the capability of demonstrating muscle forces in the Hill-type muscle model (Nazari *et al.*, 2010).

In this research, the virtual thermal strain idea was the main approach toward muscle activation; however, the full aspects of a Hill-type muscle model were not implemented. Since the main concern in this research was focused on quasi-static analyses hence this simplification does not produce large differences in the final results.

2.5 Method

First, the prepared model was imported into ABAQUS FEA™ software and all the boundary conditions and interaction properties were applied to every individual part precisely. Then, the movement of the lower jaw was manipulated by a virtual thermal strain produced in muscle fibers. In other words, since the end of each muscle fiber is tied to the mandible, the temperature difference causes thermal strain in muscle fibers which leads to movement of the lower jaw. The activation of each bundle of muscle depends on the specific task we expect to see. For opening the mouth, three bundles of muscle fibers (Digastric, Geniohyoid and Lateral Pterygoid) are involved and the temperature variation ($\Delta T = -15000$) was applied to every individual muscle fiber in this group. In this way, we witness a complete movement of the jaw which is in agreement with the previous researches (Koolstra & Van Eijden, 1997). The magnitude of temperature difference in muscles varies for different tasks and various situations.

In the next stage of simulation, both commercial and anatomical TMJ prostheses were implanted in their appropriate positions and all of the aforementioned assumptions were applied in both models. In the analysis procedure of the operated mandible, the solution time is divided into two separate time-steps. The first time-step is a short and slight activation of muscles to make the initiation of contacts in order to fix the TMJ prosthesis in its position whereas the second time-step is the main step at which

muscles undergo the desired activation in a physical period of time to generate the movement of mandible and all of the desired outputs during the complete range of motion. The whole mentioned procedures were applied for both TMJ prostheses and the output results were compared and analyzed with an intact mandible.

An aggregation of approximately 400000 quadratic tetrahedral elements was generated for all components of the model. Figure4 shows the discretized 3D model of all three assemblies. In order to improve the accuracy of simulation, second order elements with mid-side nodes were used in all models.
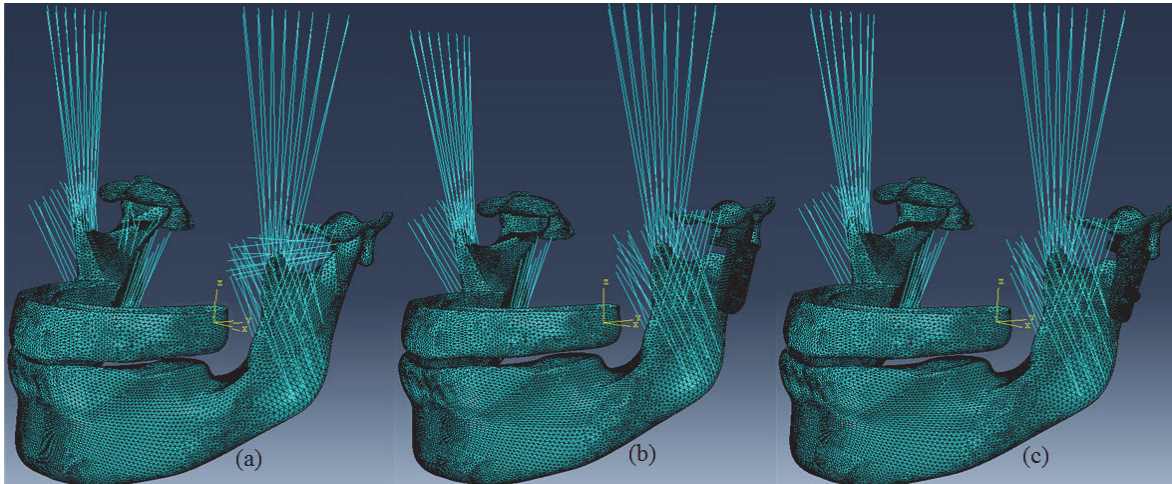

Fig. 4. (a) The meshed model of an intact mandible, (b) The meshed model of the operated mandible replaced with commercial TMJ prosthesis, (c) The meshed model of the operated mandible replaced with anatomical TMJ prosthesis

3. Results

In conclusion, the strain of each muscle fiber which ensued from its activation resulted in complete movement of the mandible. This result indicates that the model and the assumptions are capable of simulation of mandible movement which is in agreement with previous researches (Omidi *et al.*, 2017; Koolstra & Van Eijden, 1997).

As it can be seen from the simulation outcomes, it is obvious that the range of motion in both TMJ prostheses is similar to each other and the intact mandible (Fig. 5) while the muscle forces, strain and stress distribution and reaction forces are different.
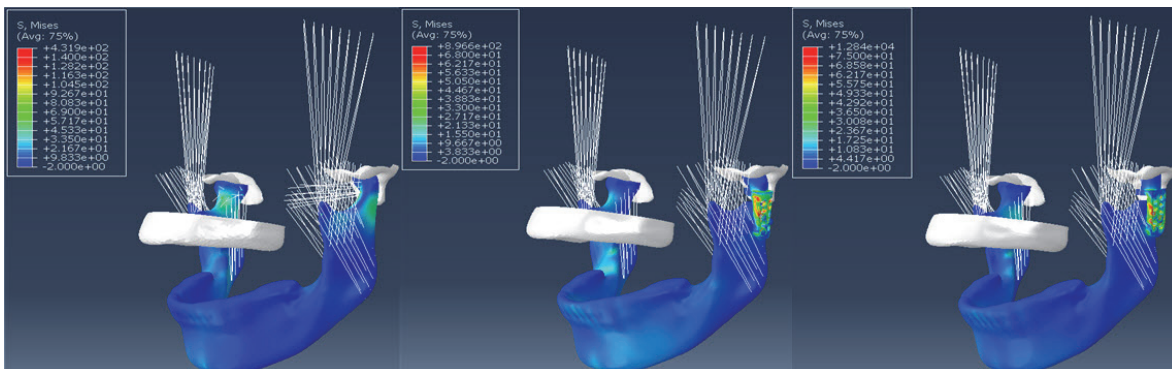
Fig. 5. Range of motion and stress distribution in all three models

Due to removal of Lateral Pterygoid muscle in the operated mandible, other muscles have to produce extra forces to compensate the lack of Lateral Pterygoid muscle and produce the same range of motion as in an intact mandible. Tables 3 and 4 show the amount of muscle forces for all three models.

Table 3. Muscle forces along fibers directions in both intact mandible and the operated one with commercial TMJ prosthesis

Muscle	Intact Mandible Force (N)	Commercial TMJ Prosthesis Force (N)	% of Difference
Superficial Masseter	0.6206	0.682	9.89
Deep Masseter	0.2528	0.2706	7.14
Temporalis	0.1903	0.2352	23.59
Medial Pterygoid	0.2343	0.2805	19.72
Geniohyoid	7.018	7.957	13
Digastric	16.82	17.675	5
Lateral Pterygoid	14.13	-	-

Table 4. Muscle forces along fibers directions in both intact mandible and the operated one with anatomical TMJ prosthesis

Muscle	Intact Mandible Force (N)	Anatomical TMJ Prosthesis Force (N)	% of Difference
Superficial Masseter	0.6206	0.6479	4.40
Deep Masseter	0.2528	0.26	2.84
Temporalis	0.1903	0.2214	16.34
Medial Pterygoid	0.2343	0.2607	11.27
Geniohyoid	7.018	7.018	0.00
Digastric	16.82	16.82	0.00
Lateral Pterygoid	14.13	-	-

In the intact mandible, the muscle forces are minimum compared to the mandible containing TMJ prosthesis. Between the two TMJ prostheses, the commercial TMJ implant needs more muscle forces to move the mandible while the anatomical one can imitate the movement of mandible with less muscle forces. Furthermore, the discard of Lateral Pterygoid muscle affects the strain produced in muscle fibers. As it can be seen from Table 5 and Table 6, the amount of strain is different in all three models.

Table 5. Strain along fibers directions in both intact mandible and the operated one with commercial TMJ prosthesis

| Muscle	Intact Mandible Strain	Commercial TMJ Prosthesis Strain	% of Difference
Superficial Masseter	0.23	0.25	8.70
Deep Masseter	0.15	0.1875	25.00
Temporalis	0.14	0.175	25.00
Medial Pterygoid	0.18	0.2125	18.06
Geniohyoid	-0.82	-0.85	3.66
Digastric	-0.69	-0.7	1.45
Lateral Pterygoid	0.063	-	-

Table 6. Strain along fibers directions in both intact mandible and the operated one with anatomical TMJ prosthesis

Muscle	Intact Mandible Strain	Anatomical TMJ Prosthesis Strain	% of Difference
Superficial Masseter	0.23	0.23	0.00
Deep Masseter	0.15	0.185	23.33
Temporalis	0.14	0.16	14.29
Medial Pterygoid	0.18	0.2	11.11
Geniohyoid	-0.82	-0.82	0.00
Digastric	-0.69	-0.7	1.45
Lateral Pterygoid	0.063	-	-

Besides the muscle forces and strain in fibers, stress distribution is also compared in all three models. It is evident that the stress distribution is totally different in the intact mandible and the one with TMJ replacement. This difference is more noticeable around screw holes which are responsible to fix the TMJ prosthesis in its proper location. Figure5 demonstrates the stress distribution in the same range of motion for all three models. In the anatomical TMJ prosthesis, since the contact surface between the mandible and the implant matches precisely, the stress distribution along the anatomical TMJ prosthesis and screw holes is lower in comparison with the commercial TMJ prosthesis (Fig. 6).
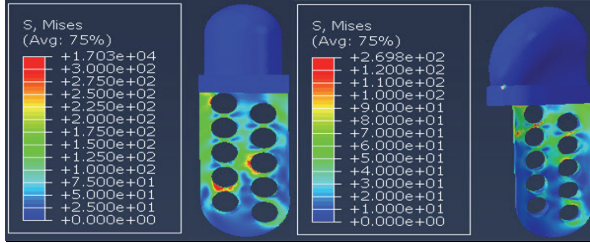

Fig. 6. The stress distribution along both TMJ prostheses (Left: commercial TMJ prosthesis, Right: anatomical TMJ prosthesis)
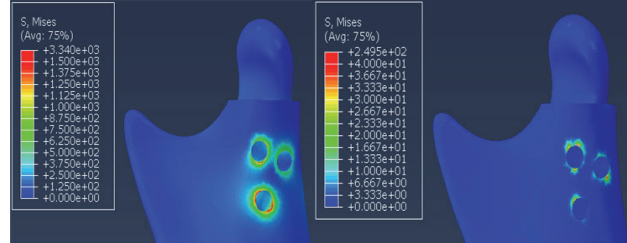
Fig. 7. The stress distribution near screw holes on the ramus of the both operated mandibles (Left: commercial TMJ prosthesis, Right: anatomical TMJ prosthesis)

Moreover, the stress distribution in the ramus of the mandible is lower when the anatomical TMJ implant is used for TMJ replacement (Fig. 7).

Articular fossa is one of the most significant parts in manipulating the movement of mandible. Due to interaction of condyle with articular fossa, the reaction force is produced as a result of this interaction. Fig. 8 shows the different amounts of reaction force in all three models for the left articular fossa.
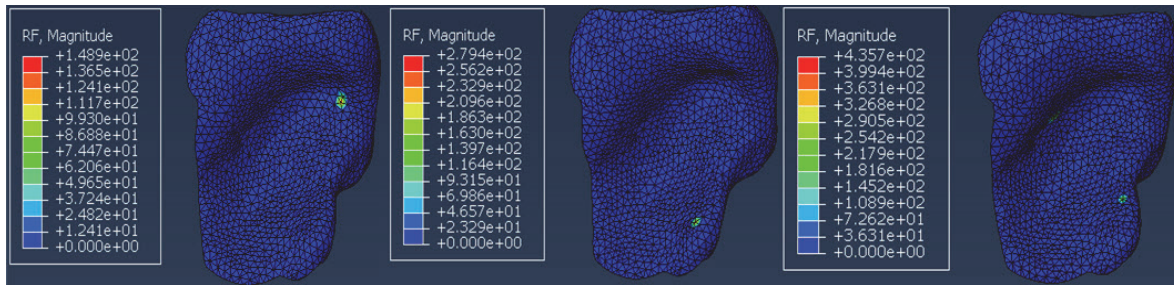
Fig. 8. From left to right: the reaction force on the left articular fossa of an intact mandible, operated mandible with anatomical TMJ prosthesis and operated mandible with commercial TMJ prosthesis

It is conspicuous that as the head of TMJ prosthesis in anatomical implant is similar to the condyle of an intact mandible, the produced reaction force in anatomical TMJ prosthesis and intact mandible are less than the commercial TMJ prosthesis.

4. Discussion

The results show good agreement between the range of motion of the intact mandible, the operated one and Koolstra's et al. results (Koolstra & Van Eijden, 1997). Since the replacement of the TMJ prosthesis consequence in removal of Lateral Pterygoid muscle, some variations in different parameters are inevitable. In fact, due to contribution of Lateral Pterygoid muscle in opening the mandible, the removal of this muscle and replacement of TMJ prosthesis demands extra forces to move the mandible in the same range of motion as an intact mandible. However, since the anatomical TMJ prosthesis is similar to the anatomy of the ramus and condyle of the mandible and have a better contact with the lower jaw, these features provide a desirable environment which influences the behavior and performance of the TMJ prosthesis. In this regard, the amount of muscle forces in the anatomical TMJ prosthesis is lower compared to the commercial one. Furthermore, this variation is evident in strain in

muscle fibers because of the same reason. Therefore, the patient needs to apply less effort to move the mouth for alternative tasks such as clenching and swallowing.

In addition, the stress distribution variation between all three models is conspicuous. In other words, since the operated mandible contains TMJ prosthesis fixed by screws to the mandible, this fixation process produces stress concentration around screw holes. Besides, the contact between the TMJ prosthesis and ramus of the mandible and movement of mandible yields to stress on TMJ implant body and ramus of the jaw. However, because of the anatomical shape of the anatomical TMJ prosthesis and its appropriate contact with the mandible, the produced stresses are less than those generated in the commercial TMJ prosthesis.

The good contact and similarity between the head of the TMJ prosthesis and the condyle of the mandible provide additional advantages to the performance of the proposed implant. As it was shown in Figure 8, the reaction forces on the left articular fossa are lower in the anatomical TMJ prosthesis compared to the commercial one. Moreover, it has been proved that good contact helps better bone absorption (Ramos *et al.*, 2011), therefore, the anatomical TMJ prosthesis provides better environment in patient's mouth and helps its durability.

Therefore, the ultimate inspections declare that the anatomical TMJ prosthesis is suitable and useful for patients who suffer from TMJ disorders. In fact, the anatomical design of this prosthesis prepare conditions which help increasing the life-time of TMJ prosthesis and enhancing the patient's satisfaction.

5. Conclusion

According to the obtained results, it can be concluded that the anatomical TMJ prosthesis is capable of imitating the function of left condyle while generating lower stress in comparison with the commercial TMJ prosthesis. Moreover, it can be seen that in anatomical TMJ prosthesis, the muscles have to produce less efforts to move the mandible. All in all, anatomical TMJ prosthesis is preferable in TMJ replacement surgery since it can produce the same range of motion, generate the minimum stress on the mandible and make less muscle forces to move the lower jaw. All are benefits of using anatomical TMJ prosthesis instead of the commercial TMJ prosthesis. It should also be mentioned that due to special characteristics of anatomical TMJ prosthesis, the process of manufacturing this type of prosthesis might be an issue; however, by developing new methods of manufacturing such as 3D printing it is not an obstacle anymore.

Acknowledgment

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