Abstract

The aim of this study is to examine the effect thickness and contact surface geometry of condylar stem of TMJ implant on its stability in total reconstruction system and evaluate the micro-strain resulted in bone at fixation screw holes in jaw bone embedded with eight different designs of temporomandibular joint implants. A three dimensional model of a lower mandible of an adult were developed from a Computed Tomography scan images. Eight different TMJ implant designs and fixation screws were modeled. Three dimensional finite element models of eight implanted mandibles were analyzed. The forces assigned to the masticatory muscles for incisal clenching were applied consisting of nine important muscular loads. In chosen loading condition, The results indicated that the anatomical curvature contact surface design of TMJ implant can moderately improve the stability and the strain resulted in fixation screw holes in thinner TMJ implant was diminished in comparison with other thicknesses.

Key words: Finite element analysis; Mandible; Temporomandibular joint; TMJ implant, Strain; Stability

1. Introduction

The temporomandibular joint (TMJ) is a joint in the body that is composed of a hinge and a sliding compartment [1]. It has a frequency of motion indicated up to 2000 times per day during talking, chewing, swallowing and snoring [1] [2]. Consequently, The TMJ is the most exerted joint in the body subjected to functional loads during physiological movements [3]. In spite of the fact that a large number of people
who are suffering from TMJ disorders, the TMJ field of research has not been deeply explored and it is one of the least studied joints in the human body [4]. According to the epidemiological studies, symptoms of TMJ disorders can be seen at 20–25% of the population [5], however, only 3–4% of them look for treatment [6]. The most common TMJ disorders are trauma or fracture, advanced degenerative disease, tumors, developmental anomalies, and ankylosis. In order to healing injured TMJ joint, total TMJ replacement (Alloplastic Replacement) has been developed to recover mandibular function and diminish disability [6]. Alloplastic replacement of the TMJ generally is consisting of a condylar implant with an articulating glenoid fossa, in which the nonfunctional joint has been replaced by an artificial one [1]. Since that this joint involves complex anatomical structures, the design, material and geometry of the TMJ implants are immensely important for long-term success of implants [7]. Nowadays different TMJ implants are used in surgeries. Some of the TMJ implants are easily can be bent so that surgeons consume some hours to manipulate the plate to fit to the curvature of the mandible. Reversely, some other TMJ implants such as Christensen implants, are manufactured as flat and rigid plates and it is hard to be bent [8].

Although, There are three commonly available prosthesis systems (namely: TMJ Concepts, TMJ Systems, Biomet /Lorenz) [1], none of them has been acknowledged as a universally accepted implant for replacement of the TMJ [9]. So that, the field of TMJ alloplastic replacement is highly demanding further research to characterize the essential design features and biomechanical requirements of these implants [2]. Therefore, Finite element analysis (FEA) is a useful means that can be applied to quantify the TMJ implants. Several authors have developed finite element (FE) models for the TMJ, including the articular cartilage; however, few studies have used mathematical or FE analysis to investigate TMJ implants [10] [11]. To determine the optimum design of an artificial TMJ implant, the different aspects should be carried out. Whereas, TMJ implants applied on patients are variable in thickness; the goal of this study was to determine the effect of various thicknesses and the geometry of the TMJ implant on its stability by evaluating the strain in bone at fixation screw holes in each model.

2. Materials and Methods

The geometry of the mandible of an adult, including cortical and cancellous bone, was obtained from a CT scan images set of 98 slices, with 1mm slice thickness. Using an image processing software package (Mimics, Materialise NV, Leuven, Belgium) and based on the Hounsfield Unit, cortical and cancellous bone were separated and reconstructed. Then three dimensional (3D) model of the lower jaw imported into a commercially 3D modeling software (SolidWorks 2009, Dassault Systèmes, USA). Based on geometry of a commercial TMJ implant (TMJ Implants, Inc.,Golden, CO, USA) [12], flat standard implant (FSI) was created (Fig 1(b)) and fixed on lower jaw (Fig 1(a)). The dimensions of standard implant were as follow: implant slender part (thickness = 2.5 mm and length= 44.6 mm); implant condyle (diameter = 8.7 mm and length/height = 10.03 mm); and the 10 screw holes were 3.02 mm in diameter [12]. To compare different thicknesses of implant, extra three models of TMJ implant were developed with the same dimensions of the standard one but vary in thicknesses (Thickness=1mm, 1.5mm, 2mm).

In order to differ in geometry, another implant was designed based on the anatomical curvature surface (ACSI) of the mandible (condyl and ramus) and thickness of implant stem kept constant. Other features, such as holes, for this design of implant were similar to flat one in all thicknesses (Thickness=1mm, 1.5mm, 2mm, 2.5mm) (Fig 1(c)). In comparison, ACSI has larger contact surface with the jaw bone rather than FSI. Collectively, there are eight different designs of implants.

To simulate the real patient (surgery), the condylar part of the mandible was cut and all eight implants were aligned to the left side of the mandible. According to pervious findings [13], three screws can supply optimum implant stability. Hence, three screws were used to fix implant on the jaw.

For static assessment of the model of intact mandible (Fig 2(a)) and mandible with TMJ implant (Fig 2(b)), finite element analysis (FEA) method was established by utilizing FEA software (CosmosWorks
The models were meshed using 1.2 mm parabolic tetrahedral elements with number of 133,234 elements and 196,599 nodes.

The models were used to simulate the static biting task. This task involves incisal clenching in which the four incisor teeth were constrained, permitting freedom of displacement in horizontal plane and no upward translations. The loading configuration consisted of nine principal muscles [14]. The magnitude of muscular forces applied, relative to its maximum possibility, and their corresponding unit vectors are presented in Table 2. Data on the material properties of all TMJ parts were taken from previously publications. The implant and screws were made from titanium alloy. All material properties assigned to the components were considered to be homogenous, isotropic and linearly elastic [15] [16]. They are listed in Table 1.

3. Results and discussion

This study attempted to investigate the stability of eight various designs of TMJ implant. FEA is an applicable means that can be applied to evaluate such these structures which are complicated to be experienced in real world. In this regard, a critical factor, strain in bone at fixation screw holes for quantifying the models had been considered. The screws utilized for fixation of TMJ implant to the bone, transfer the stresses to the bone during functions. Other researchers reported failures of experimental titanium alloy implants and others on the market. In which the geometry of the first screw hole caused the implant fracture [17] [18].

![Fig. 1](image1.png)
(a) Implanted lower jaw; (b) FSI; (c) ACSI

![Fig. 2](image2.png)
(Von Mises Stress (a) intact mandible; (b) mandible with TMJ implant

Table 1. Directions of unit vectors (i.e., direction cosines) of muscular forces and forces assigned to the masticatory muscles for incisal clenching tasks [14]: When seen from the front, the x-z plane was parallel to the floor, with the +x axis oriented toward the right, the +y axis running upward, and the +z axis oriented forward (anteriorly).

| Right Side | Left Side |
|------------|-----------|
| a          | b         |
Figure 3 illustrates the micro strain (deformation) in bone at screw holes locations. As previously mentioned, the screws were the same positions in all simulations. We applied three screws in which the first screw is the closest one to the condyle and the farther ones are the second and third screws. The strain higher than 4000 μstrain can cause hypertrophy [19]. In screw holes that strains are less than bone resorption limit (4000 μstrain), might be a deformation of circumferential bone.

For $T=1$, in both FSI and ACSI patterns, the trend for strain distribution were the same. The highest strain was at first screw location. This was followed by second and third screw locations. In comparison, ACSI has reduced the amount of deformation at all screw locations. However, all micro strains were below 3500μs.

For $T=1.5$, in FSI model, the uppermost deformation happened at second screw location, 4031 μs, and the lowest one had happened at first screw location, 3124 μs. The strain in third screw hole was 3788 μs. The stain distribution at first, second and third screw holes in ACSI, were 4930, 3413 and 3815 μstrain respectively.

For $T=2$, in FSI, it was observed that the upmost strain, 4343 μstrain, generated at second screw hole. The strain in third and first screw holes were 3899 and 3082 μstrain, in order. The strain obtained at first, second and third screw locations of ACSI, were 4375, 2746 and 3200 μstrain.

| Muscle                  | Cos-x | Cos-y | Cos-z | Force [N] | Cos-x | Cos-y | Cos-z | Force [N] |
|-------------------------|-------|-------|-------|-----------|-------|-------|-------|-----------|
| Superficial Masseter    | -0.207| 0.884 | 0.419 | 76.16     | 0.207 | 0.884 | 0.419 | 76.16     |
| Deep Masseter           | -0.546| 0.758 | -0.358| 21.216    | 0.546 | 0.758 | -0.358| 21.216    |
| Medial Pterugoid        | 0.486 | 0.791 | 0.373 | 136.34    | -0.486| 0.791 | 0.373 | 136.34    |
| Anterior Temporalis     | -0.149| 0.988 | 0.044 | 12.64     | 0.149 | 0.988 | 0.044 | 12.64     |
| Middle Temporalis       | -0.222| 0.837 | -0.5  | 5.736     | 0.222 | 0.837 | -0.5  | 5.736     |
| Posterior Temporalis    | -0.208| 0.474 | -0.855| 3.024     | 0.208 | 0.474 | -0.855| 3.024     |
| Inferior Lateral pterygoid | 0.63 | -0.174 | 0.757 | 47.499   | -0.63 | -0.174 | 0.757 | 47.499   |
| Superior Lateral pterygoid | 0.761 | 0.074 | 0.645 | 14.35     | 0.761 | 0.074 | 0.645 | 14.35     |
| Anterior Digastric      | -0.244| -0.237| -0.94 | 20        | 0.244 | -0.237| -0.94 | 20        |
| Superficial Masseter    | -0.207| 0.884 | 0.419 | 76.16     | 0.207 | 0.884 | 0.419 | 76.16     |

Table 2. Isotropic material properties assigned to the components in FEA models

| Material                | Elastic Modulus [MPa] | Poisson’s ratio |
|-------------------------|-----------------------|-----------------|
| Cortical Bone [15]      | 13700                 | 0.3             |
| Cancellous bone [15]    | 1370                  | 0.3             |
| Dentin [15]             | 18,600                | 0.31            |
| Titanium Alloy [16]     | 110,000               | 0.3             |
For T=2.5mm, the strain distributed at screw locations were under 3000 μstrain for FSI case. In FCSI, the highest stain generated at first screw at 4031 μstrain which is followed by 3576 μstrain at second screw and 3485 at third screw.

Based on the micro strain resulted in bone at screw locations (Fig 3), there is no bright tendency of strain distribution in different implants. It was observed that ASCI implant with T=1mm was the best design among them and it is safe in term of hypertrophy. For best outcomes, it will be beneficial if each kind of implant separately evaluated in terms of exploring most stress sites at implant.

Due to the fact that one of the crucial factors for a successful implant is stability [20], for reduction of strain around these holes the application of changing the location might be functional. Generally, the ASCI design of implant is recommended because it designed according to the anatomical curvature of the jaw and it has more contact surface with the bone which might diminish the possibility of bone loss.

4. Conclusion

This study shows that there was significant correlation between the strain distribution in bone at screw hole locations and its prosperity. It is anticipated that in ACSI that the implant was fully matched to the bone, the strain had been mildly decreased which would promote the bone resorption process and long-term success of implant.
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6. References

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