Biomechanical assessment of different surgical approaches of zygomatic implant placement on prosthesis stress

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Abstract. The treatment of severely atrophic posterior maxillae without bone augmentation by using zygomatic implants has received a major attention in prosthodontics due to great implant survival rates. However, mechanical implant system failures were still reported irrespective of surgical techniques used for zygomatic implant placement. Two main prominent approaches, the intrasinus and extramaxillary possess their own advantages and drawbacks with no particular indication found, to date, to highlight the best technique in relation to prosthesis stress. Thus, this study emphasised on the computational evaluation of both approaches with regards to the prosthesis responses. Two sets of finite element models comprising bones, soft tissue, implants, abutments, and prostheses were prepared accordingly. The models were then assigned with the material properties, contact modelling, and loadings as closely as possible with the real conditions. The results showed that the extramaxillary technique reported a more promising maximum stress value and distribution within the prosthesis than the intrasinus. Moreover, the prostheses in both approaches seemed to have a low tendency to failure as the stress levels were significantly less than the stress limit of the material.

1. Introduction

Over the last decade, the number of toothless patients has exhibited a significant increase [1, 2]. The edentulism phenomenon is normally attributed to the age increase or tooth extraction [3]. There are two types of edentulism which are full and partial. In comparison, it is noteworthy that the edentulous maxilla is approximately 35-times more prevalent than the mandible. The conventional treatment modality for those with edentulous maxilla or mandible is through the use of total complete denture to reinstate the functions, comfort, and aesthetic [4]. Nevertheless, several dissatisfactions from patients were reported in regard to less effective oral function and accelerated crestal bone resorption which could be caused by the reduced blood supply [4, 5]. An alternative of osseointegrated dental implants is therefore introduced to rehabilitate edentulous atrophic bone patients.

The osseointegrated dental implant can be used for a single or multiple restoration [5]. Prior to the implementation of this technique, anatomical bone evaluation in terms of quality and quantity is required to determine the most suitable type of implant treatment concept. Among the advantages promoted by dental implants are enhancing the occlusion and retention of removable prosthesis, preserving the bone and facial aesthetics, and improving the survival rate of prostheses [4]. The
implants placed in the maxilla possess a relatively lower success rate as compared to the ones placed in the mandible owing to low bone density and poor bone quality. Irrespective of the jawbones, the anterior region is reported to acquire higher bone density than the posterior. Besides, the posterior region also deals with greater bone resorption that resulting in limited bone quantity [6]. Bone density is important in the implantation sites as it serves as a key element in surgical approach, treatment planning, implant design, and healing time.

Commonly, the critical reduction in alveolar bone height is observed in the posterior maxilla and this could be because of periodontal disease prior to tooth loss. The quantity of available bone in that region is inadequate for the implant restoration. Bone augmentation technique is thus preferred to increase the amount of bone especially at the corresponding posterior maxilla regions [7]. However, this technique shows a lower implant performance when compared to the non-grafted bones, besides it needs longer treatment and healing time periods, and advocates morbidity risk on the harvested bones. Hence, Brånemark System® has taken an action by introducing zygomatic implants to prevent any related problems issued from the bone grafting method [8].

The placement of zygomatic implants for the prosthetic restoration can be categorised into four main approaches – intrasinus, sinus slot, extrasinus, and extramaxillary. Different later surgical techniques are developed to improve the common location of zygomatic implant head in the palatal area as shown in the traditional intrasinus method. The emergence of the implant head towards the palatal area induces mechanical resistance during mastication and also leads to poor aesthetic results of prosthesis. Each surgical approach appears to have their own unique benefits and characteristics to continuously improve the survival rate of zygomatic implants.

To date, there is no findings found on the superior technique for the placement of zygomatic implants especially in terms of prosthesis responses. The present study, therefore, focuses on the assessment of the mechanical stress magnitude and distribution within the prosthesis for different surgical techniques, intrasinus and extramaxillary. The evaluation was made via three-dimensional (3-D) finite element analysis (FEA) under simulated inclined occlusal and masseter loadings. Finite element analysis is performed to evaluate the problems of complex geometry which cannot be solved by analytical solutions. The application of FEA is not only limited to the field of implant dentistry, but it is also widely applied in heat transfer, electromagnetic, and thermo-fluid areas [9-29].

2. Materials and methods

2.1. Construction of 3-D bone model
A 3-D model of craniofacial bone with the selected region of interest was created based on a series of computed tomography (CT) image datasets. The CT images were taken from a patient experienced with a high degree of bone resorption and the images were saved in DICOM file format. Mimics software which is an image-processing tool was used to develop the 3-D shape of the required model using appropriate features. As a result, the bone model comprised two different layers, cortical and cancellous, with the dimension of 52.4, 46.5, and 111.9 mm for the width, height, and length, respectively, as illustrated in figure 1. The cancellous bone, however, only distributed and covered around the maxilla and zygoma regions due to high complexity of the anatomical structure.

2.2. Construction of 3-D prosthesis and soft tissue models
There were two different designs of prosthesis developed representing the features of each surgical approach. The intrasinus one was directly created based on the CT image datasets of the patient’s complete upper denture with slight modifications. Subsequently, the prosthesis model with flange was 15.4 to 18.4 mm in height, 12.5 to 19.1 mm in width, and 1.5 to 3.5 mm in thickness. Whilst, for the extramaxillary approach, an extension around the second premolar and first molar of the prosthesis was made towards the palatal area to signify the expected location of the zygomatic implant head. Meanwhile, the gap between the outer surface of the prosthesis model and the palatal area of the maxillary bone was considered to construct the soft tissue model with a thickness of 2.2 (alveolar ridge) to 5.6 mm (hard palate).
2.3. Construction of implant and abutment models
The anterior and posterior regions of the bone model were placed with two standard dental and zygomatic implants, respectively. Similar design of standard dental implants was considered for both surgical techniques, however, significant different designs were chosen for that of zygomatic implants relevant to the respective technique evaluated. Both designs of zygomatic implant have the same length dimension of 46.5 mm with 45º-head angulation and attached with identical straight multi-unit abutments from Brånemark System®. They are differed in diameter and body thread distribution. On the other hand, the standard dental implant body model placed anteriorly possesses a dimension of 4.0 mm x 10.0 mm that connected with an angled multi-unit abutment 30º at the height of 3.5 mm. The threaded portions on the all implant bodies were ignored and simulated via contact properties. Figure 2 depicts all implant and abutment design models used in the analysis.

![Figure 1. Three-dimensional models of (a) bones, (b) prosthesis, and (c) soft tissue.](image)

![Figure 2. Three-dimensional models of (a) zygomatic implant, (b) straight multi-unit abutment, (c) standard dental implant, and (d) angled multi-unit abutment 30º.](image)

2.4. Simulation of virtual implant placement
As all the 3-D models have successfully been developed, a virtual implant placement was then performed for each surgical technique. Regardless of surgical technique, the zygomatic implant head must start appearing around the alveolar ridge before passing through the maxillary lateral wall and ends in the zygomatic bone. In the intrasinus approach, the zygomatic implant bodies slightly penetrated the maxillary sinus cavity and the apical portions were directed dorsally towards the infratemporal fossa. Whilst, for the extramaxillary approach, the coronal part of zygomatic implant body accommodated the maxilla externally and the body only secured in the zygoma by the apical part. In comparison, the extramaxillary technique was observed to have considerably increased the prosthesis cantilever length for approximately 23% as compared to the intrasinus. The reverse was seen for the horizontal implant offset where the intrasinus approach exhibited about 45% longer than the extramaxillary. The zygomatic implant heads in both surgical techniques were attached with similar straight multi-unit abutments to support the prosthesis. Besides, the standard dental implants were placed in the anterior regions near the lateral incisors and canines, connected with the angled
multi-unit abutments 30° due to limited bone quantity. The reason of placing the standard implants at the respected regions was to evenly distribute the whole implant configuration within the maxillary arch to secure an ideal support for the prosthesis stability.

2.5. *Pre-processing settings of finite element analysis*

In this study, a FEA software, MSC/MARC was used to prepare the pre-processing settings on the models to be analysed. All the geometrical models of tissues and prosthetic components were converted from surface triangular into four nodes of solid tetrahedral elements with three degree of freedom. As a result, the FEA models in the intrasinus and extramaxillary approaches consisted of 791,000 and 787,000 tetrahedral elements, respectively. In terms of node numbers, a total about 196,000 and 194,000 nodes were generated for the intrasinus and extramaxillary techniques, respectively.

The present study only considers an immediate implantation loading in predicting the mechanical responses of the prosthesis. Thus, the interface between the implant body and bone models were set with a friction coefficient, $\mu$ of 0.3. Similar value of the friction coefficient was also applied to all mating surfaces of tissues and prosthetic components such as implant-abutment, abutment-prosthesis, and soft tissue-prosthesis interfaces. Nevertheless, frictionless contact behaviour was assigned for the contacting surface of cortical-cancellous and cortical-soft tissue by merging the interfacial nodes. In overall, the intrasinus and extramaxillary showed a total of 12 and 14 contacted surfaces, respectively.

It is noteworthy that the type of bone quality considered in the analysis is the one simulating D4. The value of elastic modulus, $E$ and Poisson’s ratio, $\nu$ for the cortical and cancellous bones are 13,400 MPa and 0.3, and 1,000 MPa and 0.3, respectively. Furthermore, Ti6Al4V titanium alloy ($E$: 110,000 MPa; $\nu$: 0.33) was selected as the material for all implant bodies and abutments, whilst the prosthesis was assumed to be made of gold alloy ($E$: 100,000 MPa; $\nu$: 0.3). The soft tissue recorded the smallest magnitude of $E$, 2.8 MPa and $\nu$ of 0.4. All the material properties were assumed to be homogenous, isotropic, and linearly elastic.

There were two types of loading applied to the finite element models namely occlusal and masseter loadings. The occlusal load of 150 N at the inclination of 15° was subjected to the first molar of the left-side prosthesis as a concentrated force. Meanwhile, a slightly higher value of resultant load, 300 N ($F_x = 12.42$ N; $F_y = 53.04$ N; $F_z = 25.14$ N) was applied at the muscle attachment areas on both sides of the zygoma to represent masseter loadings as a distributed load. Top cutting (x-y plane) and posterior (x-z plane) planes of the bone model were fixed in all axes (x, y, and z) to prevent translational and rotational movements. Figure 3 shows the applied loadings and boundary conditions in the models for both surgical approaches.

![Figure 3. Inclined occlusal load, masseter loads, and boundary conditions applied to the finite element models in the (a) intrasinus and (b) extramaxillary approaches.](image)

3. **Results and discussions**

The results of FEA was presented in terms of equivalent von Mises stress value and distribution within the prosthesis for its reaction due to the applied mechanical loadings in both surgical techniques. The
spectrum contour plot was also used to well describe the results with grey colour representing high stress magnitude whilst blue colour representing low stress magnitude.

Figure 4 exhibits the comparison of maximum equivalent von Mises stress values generated within the prosthesis between the intrasinus and extramaxillary approaches. It was observed that the prosthesis in the intrasinus recorded greater maximum stress level (290.3 MPa) for about 49.2% than that in the extramaxillary (147.4 MPa). In other words, it was approximately 2-fold increase in the stress level when the zygomatic implants placed through the intrasinus method.

In regard to the stress dispersion pattern in the prosthesis, it was clearly seen that a larger high stress concentration region indicated by grey colour spectrum scale, was developed in the intrasinus technique as compared to the one in the extramaxillary as illustrated in figure 5. However, the peak prosthesis stress magnitude in both approaches occurred at similar location which was around the prosthesis-abutment connections on the working side (adjacent to the point of occlusal load application). Minimally-stressed regions indicated by blue colour contour plot appeared on the coronal portion and distal areas of the prosthesis.

![Figure 4](image1.png)

**Figure 4.** Comparison of maximum equivalent von Mises stress value within the prosthesis for both surgical approaches.

![Figure 5](image2.png)

**Figure 5.** Comparison of equivalent von Mises stress distribution within the prosthesis for both surgical techniques.

One of the possible explanations for the higher level of mechanical stresses produced within the intrasinus’s prosthesis is owing to the location of opening path for the zygomatic implant placement on the alveolar ridge away from the point of occlusal load application. This situation has therefore increased the clinical moment arm due to the increase in occlusal width palatally (7 mm) in the buccolingual or x-axis, that leading to a greater resistance towards the applied loading. The reverse was seen in the extramaxillary technique where the starting insertion path in the bone was around the second premolar. The resulting shorter distance between the zygomatic implant head and point of occlusal load application (5.4 mm) in this approach does not significantly affect the value of stresses generated. As a result, an adequate stress dissemination was observed in the prosthesis. It is noteworthy that the greater the bending moment, the more the unfavourable stress produced [30]. Moreover, the torsional
moment effect is also inevitable by which its value can be increased up to 15% for every 1 mm of horizontal implant offset [31].

Another influencing factor contributes to the greater prosthesis stresses in the intrasinus approach is related to the total contact area of zygomatic implant-to-bones. The respective mating surfaces in the intrasinus technique were slightly smaller than that in the extramaxillary which only occurred at the alveolar bone towards palatal aspect and in the zygoma. In contrast, a higher percentage of the contact region was achieved in the extramaxillary from the maxillary sinus wall and end up in the zygomatic bone. This has also been shown by the corresponding contact area between the prosthesis and soft tissue with the intrasinus leads to a lower percentage (10.6%) as compared to the extramaxillary (16.2%). From the biomechanical point of view, the increase in total contact area will decrease the stress magnitude accordingly [30]. This is supported by the fact that the normal or shear stress relates proportionally with the applied load and inversely with the cross-sectional area. Past studies have evident that the implant design with threaded portion (high mating surface) promotes an improved primary stability due to evenly distributed stress and minimal micromotion of the implant system [32].

In general, the peak stress magnitude in the prosthesis was recorded at the value of 290.3 MPa in the intrasinus approach as compared to the one in the extramaxillary which merely 147.4 MPa (about 2-times lower). However, regardless of surgical technique types, the maximum prosthesis stresses do not tend to contribute to prosthesis failure since gold alloy may sustain stresses up to 786 MPa. On top of that, those peak stress values were also even way lower than the yield strength of the material.

Among the limitations of this study are the simplification of the cancellous bone distribution within the cranial bone specifically at the infrayzygomatic crest region and maxillary sinus wall. No cancellous bone composition was modelled at those areas because of the thin and complex anatomical structures. Besides, the material properties of the models were assumed to be isotropic, homogenous, and linear elastic wherein they appear to be different in reality. Then, the occlusal and masseter loads has been simulated as static, contrasting with the dynamic loads confronted in real mastication. Furthermore, experimental works and clinical trials can be executed to substantiate the findings of the present study as the living tissues are incomputable entity.

4. Conclusions
The outcomes of this numerical analysis support the following conclusions. The mechanical stress (equivalent von Mises) value and distribution within the prosthesis was found to be more encouraging in the extramaxillary approach (147.4 MPa) than those within the intrasinus (290.3 MPa). Also, the maximum prosthesis stress magnitudes recorded in both techniques were evident to be significantly smaller than the yield strength and stress limit of the respective material that preserving the prosthesis from failure.

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