Patients with severely resorbed edentulous maxillae can be treated using conventional removable dentures, implant-supported fixed prostheses, or overdentures. An implant-supported fixed prosthesis or overdenture requires a minimum of four implants, two in the posterior region and two in the anterior region. The placement of the two implants in the posterior region can be achieved using sinus bone grafting or zygomatic implants. Bone grafting is normally chosen as the standard procedure for the treatment of severely atrophic maxillae before the placement of conventional implants. However, this procedure is resource demanding and requires a relatively long treatment time and a longer healing period for the patients. The harvest of bone grafts could cause morbidity at the donor site. Based on the literature, the implant survival rate is lower for grafted maxillae compared with non-grafted maxillae, especially in the posterior region. In the anterior region, the success rate of the implant depends mostly on the bone volume before the treatment. The Brånemark System (Nobel Biocare) has introduced an alternative system utilising zygomatic implants to overcome these problems. The original purpose of zygomatic implants was to rehabilitate patients who...
had undergone maxillectomy due to tumor resection, trauma, or congenital defects. However, the function of this implant had been expanded for the rehabilitation of patients with severely resorbed edentulous maxillae. A zygomatic implant for the treatment of edentulous maxillae comes in various sizes, lengths, diameters, and thread distributions. Two types of treatment are currently used: either two zygomatic implants are placed bilaterally, or two zygomatic implants are placed in conjunction with at least two conventional implants at the anterior region of maxilla. The selection of approach depends on the degree of bone resorption in the maxilla. The path of insertion of the zygomatic implant is usually from the alveolar bone in the second premolar or first molar region, going through the maxillary sinus or its wall into the zygomatic bone.

Several surgical approaches have been available in practice since the zygomatic implant was introduced, such as the intrasinus approach, sinus slot approach, extrasinus approach and extramaxillary approach. In the intrasinus approach, the position of the implant body must be maintained at the maxillary sinus boundaries, resulting in a bulky dental prosthesis since the implant head emerges in a more palatal aspect. This approach was originally defined by the Brånemark System in 1988, which involved the insertion of a long implant (between 35 mm to 55 mm) anchored to the zygomatic bone, following an intrasinusual trajectory. The penetration of the implant body through the maxillary sinus cavity needs to be considered because the soft tissue may be highly affected. Stella and Warner described a variant of the intrasinus technique in which the implant is positioned through the sinus via a narrow slot, following the contour of the malar bone and introducing the implant into the zygomatic process. In this way, the need for fenestration of the maxillary sinus is avoided, and the implant is on course to emerge over the alveolar crest at first molar level with a more vertical angulation. The extrasinus approach, on the other hand, is mainly used to treat patients who have pronounced buccal concavity. Through this approach, the zygomatic implant head will be positioned closer to the alveolar crest bone, and therefore the size of prosthesis could be reduced. The extramaxillary technique is the latest surgical approach introduced for the treatment of edentulous atrophic maxillae using zygomatic implants. This technique is significantly different compared with other approaches because the implant body is only anchored to the zygomatic bone. The coronal part of the implant body is placed externally to the maxilla and then covered with soft tissue using a zygomatic implant with a different thread distribution. The emergence of the implant head will be more prosthetically correct as it is located on or close to the alveolar ridge. The major difference between all approaches is the difference in implant path insertion.

The use of the classic surgical approach, the intrasinus approach, could result in a higher rate of complications as reported in clinical studies. Feedback from patients regarding discomfort was identified as the main problem because the bulky prosthesis may affect dental hygiene and increase mechanical resistance. Complications of peri-implant soft tissue bleeding and increases in probing depth may occur due to the inappropriate position of the zygomatic implant head and abutment. For the extramaxillary approach, implant mobility and fracture of abutment screw are among the complications that have been reported... It can be concluded that most complications are mainly caused by insufficient primary stability of the zygomatic implant in supporting the prosthesis.

A key factor for dental implant success or failure depends on stress transmission to the surrounding bone. Inappropriate loading may result in the concentration of stress in the bone around the implant, which could lead to bone resorption. It is known that the vertical load component plays a major role in masticatory loading. Conversely, the role of the horizontal load component cannot be compromised although its value is minimal, especially when an angled implant is used.

No consensus exists on the ideal approach for the placement of zygomatic implants in regard to the degree of bone anchorage and implant inclination. Little is known about the quantity of bone that accumulates around zygomatic implants through different techniques on the effects of mechanical implant stability. To the best of our knowledge, no comparative studies have examined these two surgical approaches via the finite element method. In this paper, the stress distribution and micromotion for two of the approaches - the intrasinus and extramaxillary - was investigated.

**MATERIALS AND METHODS**

A series of computed tomography (CT) image datasets of a real complete denture wearer with a high degree of maxillary bone resorption was utilized to develop a three-dimensional (3D) model of the bones, prosthesis, and soft tissue using Mimics/Magics 10.01 (Materialise). The selected regions of interest were the left maxilla and zygomatic bone (Figs 1a and 1b), where the craniofacial model was assumed to be symmetric for the analysis. The maxilla and zygomatic bones consist of two layers, the cortical and cancellous with thicknesses ranging from 1.4 mm to 2.2 mm for cortical
The height and width of the atrophic maxilla were measured to determine a suitable approach for treatment, either through the use of zygomatic implants alone or in conjunction with conventional implants. Based on our measurements, the average height of the anterior and left posterior maxilla sections was 8.1 mm and 5.8 mm, respectively. The width of the alveolar ridge in the molar region was 9.7 mm. Hence, according to the Cawood & Howell edentulous jaw classification, the types of this patient's edentulism can be classified as Class III for the anterior maxilla and Class V for the posterior maxilla. Therefore, the patient could be treated with a zygomatic implant placed bilaterally in conjunction with two conventional implants in the anterior region. To determine the length of zygomatic implant to be used, the distance from the jugale point of zygomatic bone to the alveolar crest was measured. The angulation of zygomatic implant was determined between the implant body length and the plane through the infraorbital foramen. Results from this measurement were 48.9 mm for the length and 45.7 degrees for the angle.

A prosthesis superstructure with flange was modeled based on the original patient's complete denture design from the CT dataset. The design and geometry of the model were assumed to be symmetric at 1.5 mm to 3.4 mm in thickness, 12.5 mm to 19.1 mm in width and 15.4 mm to 18.3 mm in height. A single fixed-connection type was employed using two screws to secure the prosthesis to the implants. The prosthesis and its framework were modeled as one piece and assumed to be made of gold alloy for the analysis.

The construction of the implant model required a matched abutment model to connect the implant body to the prosthesis. One zygomatic implant, 46.5 mm in length, and a straight multi-unit abutment from the Branemark System (Nobel Biocare) were used in both surgical approaches. The design of the zygomatic implant in the extramaxillary approach is slightly different compared with the one in the intrasinus approach because the diameter of the implant body is larger and only the apical part is threaded. For the conventional implant, a 4.0 × 10.0 mm implant with an angled multi-unit abutment at 30 degrees was chosen from the same manufacturer as shown in Figs 2a and 2b.

The 3D solid implant designs from the CAD software were then transferred to other software, Abaqus/CAE 6.9-1 (Dassault Systèmes Simulia) to generate surface triangular elements before the virtual surgery simulation. All models were meshed with 0.5-mm triangular element mesh.

The conventional implant was virtually placed in the anterior region of the maxilla adjacent to the lateral incisor, whereas the zygomatic implant model was placed in the posterior region according to the earlier described techniques: adjacent to the first molar and second premolar for the intrasinus and extramaxillary approaches, respectively. The extramaxillary approach was found to have increased the distal cantilever length almost two times longer than the intrasinus. The reverse was seen for the buccal cantilever length.
where the intrasinus prosthesis was 1.3 times longer than the one in the extramaxillary (Figs 3a and 3b). All models were converted into four nodes of the tetrahedral element type in finite element analysis software, MARC 2007 (MSC Software). The total number of elements for the intrasinus model was 390,899 while the extramaxillary model had a total of 394,091 tetrahedral elements. The friction coefficient, \( \mu \), for all contact surfaces were set at 0.3 to simulate an immediate loading condition. The threaded part of the implant body for both approaches was simulated via contact properties accordingly and was assigned a friction coefficient of 0.5 to represent the strong attachment to the bones. All materials were assumed to be isotropic, homogenous, static and linearly elastic. The material properties of all models are shown in Table 1.

<table>
<thead>
<tr>
<th>Material</th>
<th>Young Modulus, ( E ) (MPa)</th>
<th>Poisson Ratio, ( \nu )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone(^{20,39})</td>
<td>13,400</td>
<td>0.30</td>
</tr>
<tr>
<td>Cancellous bone(^{22})</td>
<td>1000</td>
<td>0.30</td>
</tr>
<tr>
<td>Titanium alloy (implants and abutments)(^{20})</td>
<td>110,000</td>
<td>0.33</td>
</tr>
<tr>
<td>Soft tissue(^{23})</td>
<td>2.8</td>
<td>0.40</td>
</tr>
<tr>
<td>Gold alloy (prosthesis)(^{20})</td>
<td>100,000</td>
<td>0.30</td>
</tr>
</tbody>
</table>

where the prosthesis was 1.3 times longer than the one in the extramaxillary (Figs 3a and 3b). All models were converted into four nodes of the tetrahedral element type in finite element analysis software, MARC 2007 (MSC Software). The total number of elements for the intrasinus model was 390,899 while the extramaxillary model had a total of 394,091 tetrahedral elements. The friction coefficient, \( \mu \), for all contact surfaces were set at 0.3 to simulate an immediate loading condition. The threaded part of the implant body for both approaches was simulated via contact properties accordingly and was assigned a friction coefficient of 0.5 to represent the strong attachment to the bones. All materials were assumed to be isotropic, homogenous, static and linearly elastic. The material properties of all models are shown in Table 1.
There were two types of loading applied to the finite element models — static occlusal and masseter loadings. Simulated occlusal loading of 230 N and 50 N were applied separately as vertical and lateral loadings, respectively (different loading cases), on the top surface of the prosthesis in the first molar region. For the masseter loading, a 300-N load with force components of -62.1 N in the x-axis, 265.2 N in the z-axis, and -125.7 N in the y-axis was applied at the muscle attachment area on the zygomatic arch to represent the action of the masseter muscles.

For the boundary conditions, the mid-sagittal, posterior, and top cutting planes were constrained in the x, y, and z directions to prevent any movements. All loading and boundary conditions as shown in Fig 4.

In this study, three indices were used to verify the result of this finite element study. The first index was the total contact area between the zygomatic implant body with the surrounding bone tissues. The second index was a comparison of equivalent von Mises stress (EQV) distribution and its magnitude to assess the behaviour of bones and implants under simulated loadings. The results were presented in colored contour plots with blue representing low-magnitude stress and grey representing high-magnitude stress. The third index was the comparison of the displacement value of the zygomatic implant body for both approaches. All mentioned indices provided significant information regarding the influence of different surgical approaches on mechanical criteria where high contact area and low-magnitude stress and displacement are favorable for an encouraging interpretation of the results.

RESULTS

Figure 5 illustrates the value of total contact area between zygomatic implant body and bone tissues, both cortical and cancellous. It was clearly shown that the extramaxillary approach had higher bone-to-implant contact area compared with the intrasinus approach. The total contact area was increased by 40.3% through the extramaxillary approach, nearly a two-fold increase in the total number of mating surfaces as compared to the intrasinus approach.
lous bone compared with 39.9% for the extramaxillary approach.

Tables 2 and 3 summarize the maximum and average values of EQV measured at the bone-implant interface, zygomatic bone, and zygomatic implant body under vertical and lateral loading, respectively. Figures 6a to 6d show that the stresses within the bone model concentrated around the region of the implant and at the simulated loading location for both approaches. Under vertical loading, a high stress value of 35.9 MPa was generated at the edge of maxilla and deflected to the coronal part of the implant body and prosthesis for the extramaxillary approach. Similar observations were seen in the intrasinus approach; however, the magnitude of EQV was higher, 50.6 MPa. The stress was uniformly distributed in the zygomatic bone in two main directions, towards the temporal and frontal processes of the zygoma for both approaches. Maximum stresses of 270.3 MPa and 286.6 MPa were generated at the bone-implant interface in the extramaxillary approach as compared to the intrasinus. When loaded laterally, the EQV value increased at the bone-implant interface in the extramaxillary approach to a maximum of 106.5 MPa, whilst the stress value at a similar interface in the intrasinus approach was merely 42.9 MPa.

In terms of implant displacement, Figs 7a to 7d show that the zygomatic implant body in the intrasinus approach had a similar displacement value of 0.010 mm as the extramaxillary approach under vertical loading. Lateral loading significantly increased the displacement of the implant body in both surgical approaches. A higher displacement magnitude was found in the extramaxillary approach (0.024 mm) compared with the intrasinus approach (0.012 mm), which increased by 58% and 17%, respectively, as a result of lateral loading. Figures 6a to 6d also show that the coronal part of the implant body deformed more than the apical part for both intrasinus and extramaxillary approaches.
Figs 6a to 6d  Equivalent von Mises stress distribution within bone (front view) and implant model of intrasinus approach under (a) vertical loading and (b) lateral loading, and for the extramaxillary approach under (c) vertical loading and (d) lateral loading. All images are fit to scale.

Figs 7a to 7d: Displacement of implant body in intrasinus approach under (a) vertical loading, (b) lateral loading; and for extramaxillary approach under (c) vertical loading and (d) lateral loading. The original implant positions are shown in pink. In comparison, the coronal part of the implant body in the intrasinus approach showed the most significant bending effect in the buccal direction under vertical loading, and the reverse is seen under lateral loading. All images are fit to scale.
DISCUSSION

As mentioned above, several types of surgical approaches exist to treat edentulous atrophic maxilla—the intrasinus, sinus slot, extrasinus, and extramaxillary approaches. Even though all approaches have shown successful outcomes, there are still complications reported in the literature.\(^1\)-\(^2\),\(^25\)–\(^26\) Comparative study between the most popular, the intrasinus approach and the latest, the extramaxillary approach, can highlight their strengths and weaknesses and provides crucial information to improve clinical outcomes as very few studies have addressed these issues.\(^20\),\(^27\) Through finite element analysis, a simulation can be made to determine the type of surgical approach that may provide better stress distribution and implant stability in the maxilla and zygoma.

In this study, the concept of immediate loading was simulated through finite element analysis since it has become popular among dental surgeons for the treatment of fully and partially edentulous patients. This concept has been shown to have good outcomes and high success rates have been reported in numerous clinical-based studies.\(^3\),\(^9\),\(^28\)–\(^31\) The types of surgical techniques, implant designs, implant surface roughness, bone quality, and bone quantity are some of the factors that contribute to the primary stability of implants in immediate loading cases.\(^28\) Primary or initial stability is defined as the strength of anchorage or engagement of the implant body to the bone site without any critical movement that can cause implant failure after implantation.\(^28\),\(^32\)

Many clinically based studies have been conducted on classic surgical approaches, especially the intrasinus technique.\(^1\)-\(^2\),\(^27\),\(^33\) According to Aparicio et al,\(^1\) the cumulative failure rates of zygomatic implants and conventional implants were 1.6% and 5.2%, respectively, with follow-up time periods of six months to 12 years. Reports on classic surgical approaches showed that the survival rate of zygomatic implants was higher than that of conventional implants.\(^25\),\(^33\)–\(^36\) Ahlgren at al\(^37\) reported a 100% success rate for zygomatic implants achieved for a follow-up period of 11 to 49 months. Another study by Aparicio et al\(^32\) found similar results when 69 patients treated with 131 zygomatic implants within a period of 6 months to 5 years follow-up. There were also cases of 2 zygomatic implants placed bilaterally without additional retention from anteriorly placed conventional implants. A study conducted by Duarte et al\(^31\) showed that out of 48 zygomatic implants, 2 implants had failed after 30 months. Among the complications identified through the use of intrasinus approach were bleeding of peri-implant soft tissue, increased probing depth and sinusitis.\(^1\),\(^2\),\(^6\),\(^38\)

These problems could be caused by inappropriate positioning of the zygomatic implant body and abutment resulting from the chosen surgical technique. The design of prosthesis also plays an important role for successful clinical outcomes.

Malo et al\(^9\) reported 98.5% and 100% cumulative survival rates for implants (conventional and zygomatic) and prosthetics, respectively, in their 1-year follow-up study. They investigated the application of the extramaxillary approach for the treatment of atrophic maxillae using a new zygomatic implant design with immediate loading. Their results showed that only one zygomatic implant failure was caused by implant mobility due to a disconnection between the implant and prosthesis.

Our results showed that the intrasinus approach increased the stress magnitude at the bone-implant interface and zygomatic implant body under vertical loading approximately 1.41- and 4.27-fold higher, respectively, compared to the extramaxillary approach. Although the applied loading is close to the implant head, it was more towards the buccal aspect (the implant head was positioned palatal to the ridge), thereby creating buccal cantilever force. The implant body had to sustain higher loads to counter the bending moment from vertical loading, and this resulted in a high concentration of stress at the bone-implant interface and the coronal part of the implant body. For the extramaxillary approach, even though the location of implant head was on the alveolar ridge, it was further away from the loading point, which produces the distal cantilever effect. The prosthesis transferred a large amount of stress to the maxilla and zygomatic bone, which resulted in a wider stress distribution within the bone at the implant site (premolar region). However, the maximum stress value is lower than that of intrasinus approach. The stress peaked at the abutment-implant body connection for both approaches due to the connection between the two parts.

These findings could also be attributed to the total contact area between implant and bone tissues. According to Javed et al,\(^28\) the threaded implant design could increase the primary stability by reducing the micromotion of the implant. The implant used in the intrasinus approach is likely to have high contact area since the implant surface area increases due to the thread along the implant body. However, the implant-bone contact area only occurred at the alveolar ridge, slightly in the palatal aspect and at the jugale point of the zygoma, which resulted in a smaller mating surface compared to the extramaxillary approach. The percentage of bone-implant contact area for the extramaxillary approach was higher (40.3%) due to the placement of the implant body external to the maxilla as well as at the maxillary sinus wall. It is noteworthy that the coronal part of the implant body had no threads to avoid in-
fections of the soft tissue. However, the insertion path of the zygomatic implant increased the contact area of the implant body to the bone and therefore reduced the stress at the bone-implant interface.

Lateral loading contributed more to the increase in the magnitude of stress at the maxilla around the neck of implant and a wider stress dispersion area on the implant body in the extramaxillary approach due to large rotational effects caused by the torque generated. Since the distance from the implant head to the loading point is longer than in the intrasinus approach, the magnitude of torque produced increases proportionally, resulting in a high stress level on the surrounding bone (about 2.48-fold higher). The high stress value in the maxilla is a concern because it could lead to marginal bone loss in the long-term.

Under both the vertical and lateral loadings, the maximum stress was generated on the zygomatic bone for both approaches with a higher value found in the extramaxillary approach. Most of the stress produced from simulated occlusal loading was borne by the zygoma and appears to be independent of the maxillary anchorage. These results were in agreement with the findings of Ujigawa et al., who reported that the applied loading will be transferred through the infraglyomatic crest and directed into the temporal and frontal processes of zygomatic bone.

In all models tested, the highest stress value was recorded within the implant bodies as compared to the bone model. It seems possible that the results are due to the high modulus of elasticity of titanium alloy (Ti6Al4V), 110,000 MPa compared to 13,400 MPa for cortical bone. The maximum stress values generated within the implant bodies in all loading conditions for both approaches have no tendency to cause implant failures since titanium alloys are known to be able to tolerate stress up to 900 MPa. However, the reverse was observed for the bone, as the peak stress magnitude exceeded the yield strength of cortical bone, 69 MPa. The percentage of nodal stress higher than 69 MPa, however, was only about 0.1% for both extramaxillary and intrasinus approaches.

All displacement magnitudes of the zygomatic implant body found in the intrasinus and extramaxillary approaches under both loading conditions are lower than the threshold motion limit reported in the literature. The maximum value of 24 µm generated by the extramaxillary approach under lateral loading was relatively higher than the intrasinus approach (12 µm); however, the implant bodies have a low tendency for failure as the value of micromotion between 50 µm and 150 µm may negatively influence osseointegration and bone remodelling at the bone-implant interface. Moreover, the zygomatic implant has a higher tendency to displace and bend under lateral loading due to its increased length to width ratio. The present study showed that more deformations occurred in the buccal direction for both approaches. The findings also demonstrated that the implant body deformed from the coronal to the middle of the implant body without significantly affecting the apical part. This observation was parallel with the implant stress and displacement contour plots. It shows that adequate strength for anchorage in the zygomatic bone is achievable for both approaches without critical deformation in the apical part. This is in agreement with the work of Stievenart et al. who reported that the success rate of treatment using the zygomatic implant depended mostly on the strength of the zygomatic cortical bone. Anchorage within the trabecular bone was less important since the strength of the anchorage in the cortical layer was able to retain the prosthesis successfully. The implant body also penetrates through a small volume of cancellous bone depending on the zygoma's anatomy. Another important reason for the placement of implant within the zygomatic bone as it has very little tendency towards resorption or regeneration.

Limitations of the present study were the following: (1) the simplification of material properties used in the analysis, which were assumed to be homogenous, isotropic and linearly elastic; and (2) only unilateral modelling was conducted, whereas bilateral simulation would be more clinically relevant.

CONCLUSIONS

In conclusion, the intrasinus approach generated 1.41- and 4.27-fold higher stress at the bone-implant interface and the zygomatic implant body, respectively, under vertical loading than the extramaxillary approach. However, the reverse was seen under lateral loading where the extramaxillary approach showed an increased stress level at the bone-implant interface by 2.48-fold. The zygomatic implant body in the extramaxillary approach also exhibited micromotion with a magnitude two-fold higher than those with the intrasinus approach under lateral loading. Both techniques may be used for the treatment of severely atrophic maxillae; however, the intrasinus approach is more favorable if lateral loading is a major concern.

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