Finite element analysis of different surgical approaches in various occlusal loading locations for zygomatic implant placement for the treatment of atrophic maxillae


Abstract. The aim of this study was to compare two different types of surgical approaches, intrasinus and extramaxillary, for the placement of zygomatic implants to treat atrophic maxillae. A computational finite element simulation was used to analyze the strength of implant anchorage for both approaches in various occlusal loading locations. Three-dimensional models of the craniofacial structures surrounding a region of interest, soft tissue and framework were developed using computed tomography image datasets. The implants were modelled using computer-aided design software. The bone was assumed to be linear isotropic with a stiffness of 13.4 GPa, and the implants were assumed to be made of titanium with a stiffness of 110 GPa. Masseter forces of 300 N were applied at the zygomatic arch, and occlusal loads of 150 N were applied vertically onto the framework surface at different locations. The intrasinus approach demonstrated more satisfactory results and could be a viable treatment option. The extramaxillary approach could also be recommended as a reasonable treatment option, provided some improvements are made to address the cantilever effects seen with that approach.

Keywords: Finite element analysis; zygomatic implant; surgical approaches; edentulous atrophic maxilla.

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The number of edentulous or toothless patients has increased over the last decade. The prevalence of edentulism is usually proportional to age or may be due to tooth extraction. Traditionally, patients with edentulous maxillae and mandibles are treated with conventional, complete dentures to restore aesthetics, function and comfort, but the denture wearers always report dissatisfaction due to uncomfortable and inefficient oral function.
The use of partial or complete dentures also results in accelerated bone loss. Denture wearers have reported dissatisfaction with their dentures; greater dissatisfaction being reported by patients with lower dentures than by those with upper dentures. Dental surgeons have introduced a new alternative to rehabilitate the edentulous atrophic bone with osseointegrated dental implants. The maxilla is a difficult arch to restore with osseointegrated implants because of its morphology and configuration. The limited bone quantity caused by bone resorption, especially in the posterior region of the maxilla, has resulted in low success rates for such implants. The availability of bone quantity in edentulous sites is crucial in planning treatment, deciding on surgical approaches, implant design, healing time and initial progressive bone loading during prosthetic reconstruction.

An advanced surgical technique of bone augmentation has been suggested to increase bone volume for the placement of dental implant in the posterior region of the maxilla. Bone grafting is normally chosen as a standard procedure to treat atrophic maxillae, and bones from the iliac crest are normally used as bone graft, but this procedure is resource-intensive and requires a relatively longer treatment time and a longer healing period for the patients. Harvesting of bone grafts could also cause morbidity in the donor site. Although bone augmentation can improve the configuration for potential placement of the implant into affected maxillae, lower success rates have been reported for grafted maxillae compared with non-grafted maxillae. The success rate of implants placed in the anterior maxilla depends primarily on the pre-treatment of the bone present. Bränemark System has introduced an alternative of utilising zygomatic implants to overcome these problems by anchoring the implants to bone regions free from bone regeneration or remodeling. The original purpose of zygomatic implants was to rehabilitate patients who had undergone maxillectomy due to tumour resection, trauma or congenital defects. The function of this implant has been expanded for the rehabilitation of patients with edentulous resorbed maxillae.

The zygomatic bone has been chosen as an implant anchorage site and has been evaluated in terms of its anatomical and biomechanical aspects. By using this implant, the use of the bone augmentation procedure can be eliminated or slightly reduced because of the strength of the zygomatic arch to retain the implant and the prosthesis in position. Four types of surgical procedures are available for the placement of zygomatic implants: intrasinus; sinus slot; extrasinus; and extramaxillary. In the intrasinus approach, the position of the implant body has to be maintained within the boundaries of the maxillary sinus, which results in a bulky dental prosthesis as the implant head emerges in a more palatal aspect. The extrasinus approach is mainly used to treat patients who have a pronounced buccal concavity. In the extrasinus approach, the zygomatic implant head is positioned closer to the alveolar crest bone, which reduces the size of the prosthesis. The extramaxillary approach is significantly different from the other approaches because only the implant body is anchored in the zygomatic arch. Hence, the emergence of the implant head will be more prosthetically correct in comparison to the intrasinus or extrasinus approaches. The main reasons for the introduction of these different surgical approaches for the placement of zygomatic implants are the mechanical resistance arising during mastication due to the location of the implant head and aesthetics. Every surgical approach has its own unique characteristics to increase the survival times of zygomatic implants during physiological function. No specific indication has been found, to date, to determine the best approach for the placement of implants. The intrasinus approach is the most common approach used in the clinical setting; the new extramaxillary approach was introduced to simplify the previous protocol of implant surgery.

Despite the high success rate reported for zygomatic implants, failures occur regardless of the type of surgical approach used. Many clinical reports describe a higher incidence of complications with the use of the classical intrasinus surgical approach. The bulky prosthesis in the intrasinus approach may affect dental hygiene and increase mechanical resistance to cause patient discomfort. Complications of peri-implant soft tissue bleeding and increased probing depth probably occur due to the inappropriate position of the zygomatic implant head and abutment. Mobility of the implant body and fracture of the abutment screw are some of the complications reported for the extramaxillary approach. It can be concluded that most of the complications associated with these surgical approaches for the placement of zygomatic implants are mainly caused by insufficient primary stability achieved by the zygomatic implants in supporting the prostheses.

Finite element analysis (FEA) presents several advantages, including reliable stress and strain distribution, accurate representation of complex geometries and simple model modification. It has also been proven to be an acceptable method to evaluate dental implant systems accurately in relation to other methods. The use of two-dimensional (2D) FEA is not recommended to simulate clinical situations because of the invalidity of model representation compared with three-dimensional (3D) FEA.

To the best of the authors’ knowledge, there have been no comparative studies between these surgical approaches. The objective of the study is to investigate two of the surgical approaches, intrasinus and extramaxillary, for stress and displacement distribution within the bones, framework and implants by using 3D FEA in various occlusal loading locations.

Materials and methods

A series of computed tomography (CT) image datasets was used to develop 3D models of a complete acrylic denture wearer with some degree of resorption by using Mimics 10.01 (Materialise, Leuven, Belgium), as depicted in Fig. 1. The CT images consisted of 460 axial slices at 0.7 mm intervals with a resolution of 0.7 mm/pixel. The selected region of interest was in the maxilla and the zygomatic bone on both sides, which also covered the infrayzygomatic crest, anterior nasal spine, zygomatic process, temporal process, frontal process and the orbital floor surface. The model dimensions were 111.9 mm in length, 46.5 mm in height and 52.4 mm in width. The cortical layer of the maxilla had a thickness ranging from 0.5 to 1.17 mm.

3D models of framework representing a partial prosthesis and soft tissues were also reconstructed based on similar CT image datasets. Subsequently, a partial framework with a flange was modelled to be 1.52–3.46 mm thick, 12.45–19.06 mm wide and 15.41–18.37 mm high. Two different designs of the framework were produced in which the design for the intrasinus approach was bulkier than that for the extramaxillary approach due to the expected emergence of implant heads in the palatal area. The gap along the maxillary arch between the palatal surface of the bone and the inside surface of the complete framework was used to develop a soft tissue model with a thickness of 0.39–5.58 mm.

The height and width of the atrophic maxilla to be tested 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 the measurements, the average heights of the anterior and posterior maxilla sections were 8.1 and 5.8 mm, respectively. The width of the alveolar ridge in the molar region was 9.7 mm. These dimensions fulfilled edentulous jaw classification criteria, described by Cawood and Howell, being Class III and Class V for the anterior and posterior maxillae, respectively. Therefore, the patient could be treated with a zygomatic implant placed bilaterally in conjunction with two conventional implants in the anterior region. The distance from the jugal point of the zygoma to the alveolar crest was measured to determine the length of the zygomatic implant to be used. The jugale point is defined as the point on the zygomatic lateral surface at the most depressed point of the transitional region from the lateral margin of the zygomaticofrontal process to the upper margin of the zygomatico-temporal process. The angulation of the zygomatic implant was determined between the long axis of the implant body and the plane through the bilateral infraorbital foramen, perpendicular to the mid-sagittal plane. The distance from the jugal point to the alveolar crest was 48.9 mm, and the angulation of the zygomatic implant was 45.7°.

3D computer-aided design (CAD) software, SolidWorks 2009 (Dassault Systèmes SolidWorks Corp., Concord, MA, USA) was used to develop implant models. The construction of the implant models required matched abutment models to connect the implant body to the prosthesis. Two 46.5 mm zygomatic implants with different diameters and thread distributions and two multi-unit abutments from Bränemark System® (Nobel Biocare AB, Goteborg, Sweden) were used (Fig. 2). Two 4 mm × 10 mm implants with a 30°-angle multi-unit abutment from the same manufacturer were chosen for the conventional implant. The abutment body and screw were modelled as one part. The original feature of the implant body with its thread distribution was ignored and simplified to that of a step cylinder for the zygomatic implant models and a taper cylinder for the conventional implant models. The 3D solid implant models were transferred to other CAD software from ABAQUS, Inc. to generate surface triangular elements before the virtual surgery simulation. All models have been meshed with 0.5 mm triangular element size, which was the size used in a study conducted by Cattaneo et al.22

Both the intrasinus and extramaxillary virtual surgical approaches were carried out in accordance with the procedure described by the Bränemark System®.

Fig. 1. (a) CT of craniofacial structure. (b) 3D model of craniofacial. (c) Region of interest shown in blue. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of the article.)

Fig. 2. 3D model of zygomatic implant body and straight multi-unit abutment used in (a) intrasinus and (b) extramaxillary approach. (c) 3D model of conventional implant body and 30°-angle multi-unit abutment.
protocol. Two points were identified for the placement of the implants: the starting and end points. The point of incisura (end point) was identified prior to the placement of the zygomatic implant model into the bone site as the implant had to be installed as posteriorly as possible or close to the point. Several minor adjustments can be made to the position of the implant so that the implant head and its apical part are surrounded by bones. It was important to make sure that the intermediate part of the implant body did not perforate the maxillary anterior wall and that the end

Fig. 3. The emergence of conventional implant abutments (orange) and zygomatic implant abutments (purple) on the maxillary arch in (a) intrasinus and (b) extramaxillary approaches. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of the article.)
of the implant body did not emerge into the infratemporal fossa. The implant body had to be adjusted so that the implant platform could be positioned parallel to the occlusal plane to achieve a satisfactory prosthetic outcome. As a result, the position of the apical part of the implant was slightly diverging in the dorsal direction towards the infratemporal fossa. Similar steps were repeated for inserting another zygomatic implant model into the opposite side. The extramaxillary approach was introduced to simplify the earlier classical surgical approaches and was much easier to perform and simulate in comparison to the intrasinus approach. Zygomatic implants of a different design were placed externally into the maxilla and only anchored in the zygoma. The final positions of the zygomatic implants in the extramaxillary approach were found to have increased the framework cantilever length (23% longer) and decreased the horizontal implant offset (45% shorter) and anterior–posterior (A–P) distance. The A–P distance is determined between the centre of the most anterior implant and the most distal aspect of the posterior implant (41% shorter) (Fig. 3). Conventional implants were located in the anterior region adjacent to the lateral incisors. The rationale was to distribute the overall implant configuration equally within the arch to achieve optimal support for the stability of the prosthesis.

The total number of elements for the intrasinus model was about 791,000, whereas the extramaxillary model had 787,000 tetrahedral elements. There were approximately 196,000 and 194,000 nodes for the intrasinus and extramaxillary approaches, respectively. The friction coefficients, μ, for all contacting surfaces of the implants and the framework model were set to 0.3 to simulate the immediate loading condition. The thread distributions of the zygomatic implant designs were different for the two approaches and were accordingly simulated via their contact properties. For both the approaches, the threaded part of the implant body was assigned a friction coefficient of 0.5 to represent its strong attachment to the bone. The contact surfaces between cortical cancellous and cortical-gingival soft tissues were assumed to be as perfectly bonded by merging the nodes between the two contacted models.

All materials for FEA models were assumed to be isotropic, homogeneous, and linearly elastic. The material properties of all models are shown in Table 1. The implants and abutments were made of Ti6Al4V titanium alloy, whereas the framework was made of gold alloy. Two types of load were applied to the finite element models being simulated: occlusal and masseter loads. A simulated occlusal load of 150 N was applied as a vertical load at different locations on the framework surface, specifically at the central incisor (L1), first premolar (L2), first molar (L3) and second molar region (L4) to represent more realistic food positions during biting or chewing. All vertical loads were applied along the z-axis, which is parallel to the standard implant axis. A simulated masseter load of 300 N with force components of −12.42 N along the x-axis, −53.04 N along the y-axis and 25.14 N along the z-axis was applied to the left side of the bone. For the right side, similar magnitudes of forces were applied but in the opposite direction, except for the z-axis force component. The simulated masseter load represented the masseter muscle action and was applied to the muscle attachment area on the zygoma as a distributed load (Fig. 4).

Table 1. Material properties used in FEA. All material properties were assumed to be isotropic, homogenous and linearly elastic throughout.

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus, E (MPa)</th>
<th>Poisson’s ratio, ν</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical</td>
<td>13,400</td>
<td>0.30</td>
<td>Geng et al.18</td>
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<tr>
<td>Cancellous</td>
<td>1000</td>
<td>0.30</td>
<td>Meyer et al.7</td>
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<tr>
<td>Soft tissue</td>
<td>2.8</td>
<td>0.40</td>
<td>Cheng et al.17</td>
</tr>
<tr>
<td>Framework</td>
<td>100,000</td>
<td>0.30</td>
<td>Ujigawa et al.23</td>
</tr>
<tr>
<td>Implants</td>
<td>110,000</td>
<td>0.33</td>
<td>Geng et al.19</td>
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**Fig. 4.** Boundary conditions and loading configurations shown in (a) frontal and (b) sagittal view. L1, L2, L3 and L4 are loadings applied at specific location on the framework.
Table 2. Magnitudes of stress (MPa) recorded within each model at different loading locations for the intramus (IA) and extramuscular approach (EA).

<table>
<thead>
<tr>
<th>Models</th>
<th>IA</th>
<th>L1</th>
<th>L2</th>
<th>L3</th>
<th>L4</th>
<th>EA</th>
<th>L1</th>
<th>L2</th>
<th>L3</th>
<th>L4</th>
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<tbody>
<tr>
<td>Cortical bone</td>
<td>1.96</td>
<td>1.79</td>
<td>1.65</td>
<td>1.97</td>
<td>2.02</td>
<td>1.70</td>
<td>1.69</td>
<td>2.04</td>
<td></td>
<td></td>
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<tr>
<td>Cancellous bone</td>
<td>1.64</td>
<td>1.46</td>
<td>1.34</td>
<td>1.35</td>
<td>1.67</td>
<td>1.37</td>
<td>1.35</td>
<td>1.39</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Framework</td>
<td>158.00</td>
<td>193.00</td>
<td>196.68</td>
<td>294.27</td>
<td>136.05</td>
<td>226.25</td>
<td>223.57</td>
<td>459.35</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ZI1</td>
<td>32.51</td>
<td>64.80</td>
<td>64.48</td>
<td>63.73</td>
<td>42.07</td>
<td>37.61</td>
<td>53.58</td>
<td>91.30</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ZI2</td>
<td>22.35</td>
<td>23.50</td>
<td>15.40</td>
<td>15.12</td>
<td>59.38</td>
<td>32.91</td>
<td>31.81</td>
<td>40.47</td>
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<td>CI1</td>
<td>20.14</td>
<td>23.60</td>
<td>7.48</td>
<td>4.80</td>
<td>22.49</td>
<td>10.04</td>
<td>4.50</td>
<td>5.04</td>
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<tr>
<td>CI2</td>
<td>17.42</td>
<td>16.83</td>
<td>5.73</td>
<td>4.49</td>
<td>23.45</td>
<td>12.00</td>
<td>6.97</td>
<td>16.83</td>
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</tbody>
</table>

Table 3. Magnitudes of displacement (mm) of each prosthesis component recorded at different loading locations for the intramus (IA) and extramuscular approach (EA).

<table>
<thead>
<tr>
<th>Models</th>
<th>IA</th>
<th>L1</th>
<th>L2</th>
<th>L3</th>
<th>L4</th>
<th>EA</th>
<th>L1</th>
<th>L2</th>
<th>L3</th>
<th>L4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Framework</td>
<td>0.0382</td>
<td>0.0333</td>
<td>0.0238</td>
<td>0.0264</td>
<td>0.0549</td>
<td>0.0213</td>
<td>0.0174</td>
<td>0.0377</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ZI1</td>
<td>0.0066</td>
<td>0.0074</td>
<td>0.0090</td>
<td>0.0088</td>
<td>0.0102</td>
<td>0.0081</td>
<td>0.0055</td>
<td>0.0057</td>
<td></td>
<td></td>
</tr>
<tr>
<td>ZI2</td>
<td>0.0079</td>
<td>0.0052</td>
<td>0.0038</td>
<td>0.0041</td>
<td>0.0096</td>
<td>0.0044</td>
<td>0.0042</td>
<td>0.0064</td>
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</tr>
<tr>
<td>CI1</td>
<td>0.0195</td>
<td>0.0145</td>
<td>0.0049</td>
<td>0.0018</td>
<td>0.0225</td>
<td>0.0099</td>
<td>0.0019</td>
<td>0.0019</td>
<td></td>
<td></td>
</tr>
<tr>
<td>CI2</td>
<td>0.0168</td>
<td>0.0068</td>
<td>0.0013</td>
<td>0.0018</td>
<td>0.0206</td>
<td>0.0063</td>
<td>0.0029</td>
<td>0.0087</td>
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</tr>
</tbody>
</table>

For the boundary conditions, the posterior and top cutting planes were constrained in the x, y, and z directions to prevent any movements.

A comparison was made in terms of mechanical stress and displacement pattern to determine the behaviour of the bones and implant reactions under the applied simulated mastication loading conditions. Equivalent von Mises (EQV) stress is an appropriate stress measurement to be chosen for ductile materials such as titanium. The data for EQV are shown in Table 2, and the displacement data are presented in Table 3. The results have also been presented in a colour band contour plot with blue representing low stress or displacement magnitude and grey representing high magnitude.

Three sets of symbols were used to differentiate easily between the models. The first symbol denotes the different surgical approaches: IA for the intramus approach and EA for the extramuscular approach. The second symbol represents the types of implants: ZI for the zygomatic implant and CI for the conventional implant. The third symbol indicates the location of the implants: 1 and 2 symbolises the placement of the implant on the left and right side of the bone, respectively.

Results

EQV distribution within bones

Figure 5(a) shows that anterior load led to a high magnitude and concentration of stress within the cortical bone in the anterior nasal spine (ANS) region. A similar pattern of stress was found in the cancellous bone wherein stress values decreased as the load moved posteriorly (Fig. 5(b)). When load was applied on the second molar, the stress tended to increase drastically in both IA and EA when compared to loading of the first molar and first premolar. The magnitude of EQV was much lower in cancellous bone in comparison to cortical bone. Posterior loads seemed to mainly disperse stress in the molar areas, particularly around the region of zygomatic implant placement (Fig. 6). When the results are interpreted in terms of the type of surgical approach, it is evident that there was insignificant discrepancy in the stress values of the bones between both approaches: EA resulted in a slightly higher magnitude of EQV within cortical and cancellous bone compared to IA in almost all load locations.

EQV distribution within the framework

As observed in IA, the magnitude of EQV increased as loads moved posteriorly (158.00 MPa to 294.27 MPa). This situation was also seen in EA where EQV values increased from 136.05 MPa to 459.35 MPa (Fig. 5(c)). The peak EQV value within the framework was recorded under L4, with EA being 36% higher than IA. In comparison, the framework in EA demonstrated higher values of EQV than that in IA under vertical load at all load locations, except in L1 (being 14% lower). As illustrated in Fig. 7, the stresses were generally dispersed in the mesio-distal direction and strongly concentrated at the framework-implant connection, with larger stress dispersion and the creation of a concentration region when loading was exerted in the posterior region for both approaches. The smallest stress concentration region within the framework was almost dependent on the application of anterior load.

EQV distribution within the implants

It appears that the second molar load (L4) configuration led to the highest stress values for ZI1 in both IA and EA (Fig. 5(d)). The EQV values under the three posterior loads were nearly similar with 64.80 MPa (L2), 64.48 MPa (L3) and 63.73 MPa (L4) being seen in IA. As a result, EA exhibited 30% higher levels of maximal stress generated within ZI1 compared to IA. In contrast, the stresses were significantly decreased in ZI2 as the load was applied posteriorly in both approaches (Fig. 5(e)). The high stresses were mainly concentrated at the abutment-implant connection and spread out towards the coronal and intermediate parts of the implant body on the buccal-lingual side (Fig. 8). The apical part seemed to have a smaller EQV dispersion than the coronal part, and the stress concentration region became wider as the applied load moved posteriorly. For the conventional implants (CI1 and CI2) placed in the premaxillary region, both implants sustained high stress values and stress concentration when loaded by anterior forces than by posterior forces (Fig. 5(f) and (g)).

Displacement results

Depending on the location of load application, the magnitude of framework displacement decreased proportionally as the load moved posteriorly, except for L4 where the stresses were increased by 10% and 54% in IA and EA, respectively (Fig. 9(a)). The lowest magnitude of displacement was recorded for L3: 0.0174 mm for EA and 0.0238 mm for IA, respectively. A larger displacement concentration area was produced at the anterior aspect of the framework model for both surgical approaches. When loaded at the first premolar (L2) and first molar (L3), IA recorded a higher displacement magnitude of the framework compared to EA, and the reverse was seen when anterior and most posterior loads were applied.

When posterior loads were applied (L2, L3 and L4) for zygomatic implants, ZI1...
Fig. 5. Comparison of stress distribution within (a) cortical, (b) cancellous, (c) framework, (d) zygomatic implant on the working side, (e) zygomatic implant on the non-working side, (f) conventional implant on the working side and (g) conventional implant on the non-working side.
showed higher displacement in IA and the reverse was observed in EA (Fig. 9(b)). A wide displacement concentration area developed, starting from the implant-abutment connection in the coronal part and spreading out towards the apical part of the implant body, indicating a low tendency for the zygomatic implant body to deform at the intermediate and apical portions in both groups. Maximal implant displacement was found in L1 (0.0102 mm) and L3 (0.009 mm) for EA and IA, respectively. For the right zygomatic implant (ZI2), anterior loading resulted in a high displacement magnitude for both IA and EA (Fig. 9(c)). Although displacement values decreased as loading moved posteriorly, this was not evident in L4 where the values increased to 0.041 and 0.064 mm for IA and EA, respectively. A similar observation was seen for the CII implant. The displacement of the CII implant decreased proportionally as the applied loads moved towards the posterior region (Fig. 9(d)). In comparison, EA showed considerable implant displacement compared to IA at almost all load locations.

Fig. 6. Comparison of displacement of (a) framework, (b) zygomatic implant on the working side, (c) zygomatic implant on the non-working side, (d) conventional implant on the working side and (e) conventional implant on the non-working side.


Discussion

A higher level of stress was expected within the cortical bone on the left side when the load was applied in posterior regions of the corresponding side at the first premolar, first molar and second molar. As far as stress within the bone was concerned, the highest value was recorded for the most posterior loading (L4) in both the intrasinus and extramaxillary approaches. An unexpected situation was observed when a high magnitude of stress was generated with the application of an anterior load for both surgical approaches, compared to first premolar and first molar loading. This finding is probably due to the limited area for stress dispersion at the anterior nasal spine region. In contrast, the maximum stresses induced by the most posterior loading

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Fig. 8. Comparison of stress distribution within framework for the intrasinus and extramaxillary approach as viewed from top axial.
could be initially transferred to the alveolar bone, and subsequently dispersed to the zygoma.

This finding is in agreement with Bonnet et al. who reported that higher stress was observed in the molar region compared to other regions (canine and incisor). The results of this study also supported the concept of a class 3 lever wherein a higher magnitude of force is required to lift the weight as the force moves nearer to the fulcrum. In clinical situations, the force, fulcrum and weight are represented by occlusal forces, the temporomandibular joint and food, respectively.

The cantilever effect is one of the crucial issues that cannot be compromised in implant-supported fixed prostheses. The present results showed that the highest stress was within the left zygomatic implant body when load was applied on the second molar region in the extramaxillary approach. It is noteworthy that the framework used in that approach had a longer distal extension of a cantilever compared to that used in the intrasinus approach, thus, increasing the bending moment generated. Rubo et al. drew similar conclusions that the increase in stress in implants is proportional to the increase in cantilever lengths. In another study, Bevilacqua et al. suggested that the use of shorter cantilevers, associated with distal tilted implants, could preserve prosthetic components from overload. As shown in this study, it can be suggested that the influence of cantilever effects may be more significant in the extramaxillary approach than in the horizontal implant offset for implant restoration.

The findings of this study also showed that conventional implants placed in the anterior region were mostly responsible for the high stress produced by anterior loads rather than by posterior loads. Also, both conventional implants placed via the intrasinus approach sustained higher stresses under nearly all loading conditions to retain the whole structure of restoration compared with the extramaxillary approach. This finding could be attributed to the appearance of the posterior implant head, which is slightly out of the maxillary arch alignment, being inclined towards the distal direction. This resulted in an increase in the A–P distance, causing lower retention by the framework to sustain interimplant loads, except when the load was applied on the second molar.

Load applied in this region was better tolerated in the intrasinus approach as the force was more dispersed throughout the arch and the zygomatic implant body, unlike the extramaxillary approach, which was more affected by cantilever effects.

Past studies on implant-supported fixed prostheses indicated that the success of prostheses and bone-implant stress distribution can be influenced by implant inclination, number, position, prosthesis splinting schemes, occlusal surfaces, framework material properties and different cross-sectional beam shapes. The stress distribution within the framework indicated that posterior loads resulted in high magnitudes of stress and the concentration of stress around the connection between

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the framework and the implants. This outcome was parallel to the stress results obtained within the bodies of zygomatic implants wherein posterior loads were the major influencing factors. The load applied to the top surface of the framework in the molar region resulted in high stress concentration that could be due to resistance to high bending moments generated by the zygomatic implants. A balanced stress distribution was seen in the framework model when it was loaded anteriorly. A possible explanation for this observation is that there may be a progressive transfer of occlusal load from the load point to the conventional and zygomatic implants in the mesio-distal direction.

Anterior loading seemed to increase the magnitude of zygomatic implant displacement in the extrammary approach considerably as the implant body was only supported by the outer surfaces of the maxilla. This could be explained by the lack of bony support provided at the coronal portion of the implant to resist loads in the buccal direction. A different pattern of displacement was observed in the intrasinus approach as the implant head was highly restricted by the alveolar bone in the buccal-lingual and the mesio-distal directions.

The purpose of a zygomatic implant is almost similar to that of the angled implant application, which is to rehabilitate the edentulous atrophic maxillae without undergoing a bone grafting procedure. The theoretical successes of angled implants are dependent on several factors, such as implant length, cortical anchorage and implant distribution within the arch. The longer the implant is used, the more the bone-implant contact interface will be generated. Immediate loading can be secured when the implant body achieves one or more cortical anchorage points. Through the use of angled implants, prosthetic restorations can also be directed more posteriorly, thus, increasing the A-P distance without compromising the load distribution throughout the maxillary arch. The body of a zygomatic implant has a high tendency to deform from its coronal part to the middle of the implant body, without significantly affecting the apical portion, regardless of the type of surgical approach used. This finding indicates that adequate strength is achievable for anchorage in the zygomatic bone for both intrasinus and extrammary approaches, with no critical deformation in the apical part. The zygomatic bone has a wider and thicker cancellous bone that can confer initial stability to the implant. Nkenge et al. have revealed that the parameters of the zygomatic bone are poor for implant placement. It has also been reported that the success of zygomatic implants can be guaranteed by inserting the implants into four cortical portions. This finding is in agreement with the work of Stienvenet et al., who reported that the success of treatment using zygomatic implants mostly depends on the strength of the zygomatic cortical bone. The models prepared in this study demonstrated that a total of four cortical layers were penetrated in both surgical approaches. The literature reports that the initial stability of the implants could be increased if more than one cortical anchorage were achieved. Anchorage is not required within the trabecular bone as the strength of anchorage in the cortical layer could successfully retain the prosthesis. Depending on the anatomy of the zygoma, the implant body also penetrates through a small volume of cancellous bone, which is another important reason for the placement of implants within the zygomatic bone as it has very little tendency towards resorption or regeneration.

The findings of this study suggest that the cantilever effect contributes to high stress distribution and displacement in the extrammary approach, which could be reduced by using a support system, such as the IL system. This system uses a short implant and a ball-type attachment as an additional retention for distal extension of a cantilevered prosthesis. Through a 2D FEA, this system has shown desirable results of stress dispersion within the bone and the implant when compared to conventional cantilevered prostheses. The effects of horizontal implant offset could be reduced by placing the prosthesis in cross-occlusion in the intrasinus approach. The shallow cusp obtained by reducing cusp inclination can be incorporated into the construction of the prosthesis because it can disperse horizontal load more efficiently, thus reducing torque-producing forces.

As this is a new study on the comparison of surgical approaches for the placement of zygomatic implants, further work has to be done to validate these results. The results of simulated static loadings and non-linear analysis support the following conclusions. The zygomatic implant was able to absorb most of the posterior load while the anterior conventional implants absorbed most of the anterior load. The extrammary approach exhibited higher stress values than the intrasinus approach within the bones and the implants, especially, when applied with posterior loading due to the cantilever effect. The prosthetic components in the extrammary approach promoted a higher displacement tendency than the intrasinus approach when loaded anteriorly, and also when loading occurred in the most posterior region.

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