Titanium implants represent an effective option for the rehabilitation of partially and totally edentulous patients.\textsuperscript{1,2} Fixed full-arch implant-supported prostheses are a predictable form of rehabilitation for totally edentulous patients. Four to six implants are placed and splinted using a metallic framework, which makes the stress distributions more complex than in single crowns.\textsuperscript{3–6} Thus, an occlusal load applied to one point of the prosthesis leads to different degrees of stress concentration in the surrounding implants and bone, which can cause rehabilitation failures such as fractures or loosening of the screws, abutment, and metallic framework, as well as the fracture or loss of implant osseointegration.\textsuperscript{6}

Stress distribution depends on several factors, such as the characteristics of the occlusal load (intensity, direction, duration, and speed), type of implant connection, bone quality and density, type of material, anterior-posterior spread, presence of cantilevers, as well as the number, diameter, and length of the implants.\textsuperscript{4,8–11}

The human cortical bone has a maximum physiologic limit of 140 to 170 MPa in compression and 72 to 76 MPa in traction.\textsuperscript{12} If the occlusal load exceeds the absorption capacity of the system, the implant or prosthesis will fail due to overloading and inadequate distribution of masticatory forces, among other factors.\textsuperscript{12}

The placement of long or regular-length implants in totally edentulous patients is sometimes impossible because of the high alveolar ridge resorption. Since the advent of short implants, < 10 mm long,\textsuperscript{7} dental implant rehabilitation in areas with significant resorption constitutes a less complex, costly, and traumatic treatment option for patients. When possible and correctly indicated, the use of short implants seems to be a safe choice in the treatment of edentulous areas with limitations in bone height.\textsuperscript{2,9,10}

It is vital to study the effects of the loading forces in implantology as a preventive measure and to improve the survival of implant-supported prostheses. Before large-scale production, each implant design must be evaluated based on the effects of a prototype on natural bone tissue. An important process entails the assessment of the generated stress in the implant and the surrounding bone. The finite element method (FEM) is an increasingly adopted biomechanical study approach to demonstrate and predict stress distribution.\textsuperscript{1,13–15}

Purpose: This research aimed to evaluate the stress and fatigue generated in short implants compared to regular implants in rehabilitation with fixed full-arch implant-supported prostheses in atrophic mandibles using the finite element method (FEM).

Materials and Methods: Four models were constructed with different implants lengths: 4, 6, 8, and 10 mm. A 100-N oblique load was applied to evaluate the stress on the bone, implant, and prosthetic components.

Results: Similar behavior was observed for all groups, except for 4 mm, which showed more discrepant values. During the fatigue test, all the groups exhibited infinite lives except G4. Conclusion: Based on the similarity of all the models, it is suggested that all short implants investigated are seemingly viable alternatives for the rehabilitation of atrophic mandibles. However, the 6-mm-long and 8-mm-long implants evinced more favorable mechanical behavior than the 4-mm-long type. Int J Oral Maxillofac Implants 2022;37:971–981. doi: 10.11607/jomi.9514

Keywords: biomechanics, complete denture, dental implants, finite element analysis
FEM is a computational mathematical technique for calculating deformations and stresses. The effect of the loading forces on a prosthesis or peri-implant region can be assessed using the equivalent stress (von Mises stress), expressed in megapascals (MPa), which is compared to the limiting resistance stress of the material. The stress levels around the peri-implant regions and prosthetic structures are presented in different colors to view these results, which enables assessment of the safety of these structures during chewing.\(^{14}\)

Short implants can still fail because of fatigue due to cyclic loading of the chewing process.\(^{16}\) Fatigue failure occurs after many loading cycles. The damage accumulates at a micromechanical scale, and a microcrack is formed. With additional loading cycles, the crack grows, and when it reaches a critical dimension, component failure occurs. Crack growth and propagation have been studied in fracture mechanics. Fatigue analysis, however, focuses on damage accumulation and enables the prediction of where the crack will grow and propagate.\(^{17,18}\)

To contribute to the literature regarding biomechanics against oblique loads, this research aims to evaluate the stress and fatigue generated in short implants compared to regular implants in rehabilitation with fixed full-arch implant-supported prostheses in atrophic mandibles using FEM.

**MATERIALS AND METHODS**

**Study Type and Location**

The experimental laboratory study was conducted under the Graduate Program in Dentistry and the Department of Mechanical Engineering at the Federal University of Pernambuco, Brazil.

**Construction of Geometric Models**

Four computational models of an atrophic mandible, with four implants splinted by a metallic prosthetic framework were built. The models were divided into four groups according to the implant length: 4 mm (G4), 6 mm (G6), 8 mm (G8), and 10 mm (G10) (Table 1). The G4 group had a smaller thread pitch than the others. All the implants were positioned vertically and well distributed in the interforaminal region, at least 6 mm mesial to the mental foramen.

**Geometric Construction**

**Geometric construction of the mandible.** The computational model of a human mandible was developed based on a study by Vajgel et al\(^{19}\) and adapted to the needs of this research. Atrophic mandibular axial tomographic sections of an unidentified woman 65 years of age were obtained from a radiologic clinic image database. The device saved the tomographic images in a standard DICOM format, which were then exported to a visualization and manipulation program for medical images called INVESALIUS (version 2.1, CTI, Ministry of Science and Technology, Brazil). This freely licensed software allows for the creation of a 3D mandibular bone volume from the total volume acquired from tomography, thereby allowing the exclusion of adjacent soft tissues (for example, skin, muscle, cartilage, and artificial material). The jaw model was built and saved to be imported into the SOLIDWORKS program (version 2016). This model was converted to a solid model and was divided (split tool) into parts that would serve as a basis for the construction of the spinal bone. These parts were copied, and smaller internal parts were created that kept their original shapes (scale tool). Finally, the internal parts were joined using the loft tool. Thus, the model was divided into the cortical bone with an average thickness of 2 mm, according to Vajgel et al,\(^{19}\) and the medullary bone, thereby matching it with reality (Fig 1).

Moreover, in this study, the mandible was subdivided into parts based on anatomical regions to designate the different properties of the material (bone) related to its site. The cortical bone was divided into 11 volumes (a single symphysis, as well as two each of the body, angle, branch, coronoid, and condyle), while a single volume was employed for the medullary bone. The mesh of the 6-, 8-, and 10-mm models consisted of approximately 2,390,000 nodes and 3,600,000 elements. The mesh on the 4-mm model consisted of 1,561,200 nodes and 2,456,111 elements.

---

Table 1 Groups Analyzed in the Research

<table>
<thead>
<tr>
<th>Groups</th>
<th>No. of implants</th>
<th>Diameter of implants (mm)</th>
<th>Lengths of implants (mm)</th>
<th>Thread pitch of implants (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>G4</td>
<td>4</td>
<td>4.1</td>
<td>4</td>
<td>0.56</td>
</tr>
<tr>
<td>G6</td>
<td>4</td>
<td>4.1</td>
<td>6</td>
<td>1</td>
</tr>
<tr>
<td>G8</td>
<td>4</td>
<td>4.1</td>
<td>8</td>
<td>1</td>
</tr>
<tr>
<td>G10</td>
<td>4</td>
<td>4.1</td>
<td>10</td>
<td>1</td>
</tr>
</tbody>
</table>

Fig 1 Cortical bone divided into 11 volumes according to anatomical regions.
Geometric construction of implants and protocol-type prostheses. Tissue-level implants (Straumann) with a diameter of 4.1 mm and different lengths (4, 6, 8, and 10 mm), associated with the prosthetic components of the synOcta type (Straumann) were chosen as retainers of the fixed full-arch implant-supported prostheses for the biomechanical analysis. The 3D geometries of the prosthetic components and the implants were modeled using SolidWorks (version 2016; Fig 2).

The systems for fixing the structure of the full-arch fixed prosthesis were developed using the same computer-aided device software. A 4-mm–high and 6-mm–wide Co-Cr prosthetic framework and 10-mm distal cantilevers were modeled.

Finally, the mandible models with the implants were discretized using tetrahedral finite element meshes. The mesh was refined in the region of the most interest for the study to reproduce the stress distribution generated in the implant, prosthetic components, and peri-implant bone (Fig 3).

Material Properties and Interface Conditions

The mechanical properties of the cortical bone were considered orthotropic, assuming a specific value for each of the three directions in space, as proposed by Schwartz-Dabney et al.21 These properties were associated with a referential coordinate system for each anatomical region, as suggested by Sugiura et al.,22 Fernández et al.,23 and Lovald et al.24 The mechanical properties adopted in this work are listed in Table 2. The implants and prosthetic components are made of a titanium (Ti) alloy, while the metal framework is made of the cobalt–chromium (Co–Cr) alloy. The implants were considered to be fully osseointegrated; therefore, a perfect mechanical interface was assumed.

Table 2 Mechanical Properties of Materials Used in Study

<table>
<thead>
<tr>
<th>Material Properties</th>
<th>Cortical bone</th>
<th>Cancellous bone</th>
<th>Co-Cr alloy</th>
<th>Titanium</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ex (MPa)</td>
<td>20.492</td>
<td>21.728</td>
<td>1.500</td>
<td>220.000</td>
</tr>
<tr>
<td>Ey (MPa)</td>
<td>12.092</td>
<td>12.700</td>
<td>1.500</td>
<td>220.000</td>
</tr>
<tr>
<td>Ez (MPa)</td>
<td>16.350</td>
<td>17.828</td>
<td>1.500</td>
<td>220.000</td>
</tr>
<tr>
<td>√xy (√)</td>
<td>0.43</td>
<td>0.45</td>
<td>0.3</td>
<td>0.30</td>
</tr>
<tr>
<td>√yz (√)</td>
<td>0.22</td>
<td>0.2</td>
<td>0.3</td>
<td>0.30</td>
</tr>
<tr>
<td>√xz (√)</td>
<td>0.34</td>
<td>0.34</td>
<td>0.3</td>
<td>0.30</td>
</tr>
<tr>
<td>Gxy</td>
<td>5.317</td>
<td>5.533</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Gyz</td>
<td>4.825</td>
<td>5.083</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>Gxz</td>
<td>6.908</td>
<td>7.450</td>
<td>–</td>
<td>–</td>
</tr>
</tbody>
</table>

References: Schwartz-Dabney et al.21 Sugiura et al.22 Schwartz-Dabney et al.21 Stegaroiu et al.25

E = Young’s modulus (elasticity); √ = Poisson’s ratio; G = Shear modulus.

Load Restrictions and Conditions

The directions and positions of the forces considered in the simulations of the groups were those resulting from chewing, as well as the masticatory muscles based on previous studies.19,25,26 The force generated from the molar chewing considered in this study consisted of an oblique load of 100 N based on previous studies.1,5,14 For this, a force was applied at 45 degrees on the framework in the regions of the molar cantilever on the left side (working side), while the right side was the balancing side.1,13,14,27,28 The magnitudes of the muscular forces were adjusted to counterbalance the oblique load of 100 N resulting from mastication to guarantee the static conditions of the jaw for carrying out active stress analysis.28 This adjustment was made through an analysis of rigid jaw dynamics in Ansys Mechanical (version 18.2, Ansys). In this analysis, the regions of the condyles were considered as the axis of rotation of the mandible, whose movement is generated by the resulting force of 100 N opposing the imposed muscle forces. The objective was to evaluate the resulting rotational acceleration. The magnitudes of the forces were kept proportional to each other, according to previous studies,19,26 with their values being adjusted only by one parameter, which was changed manually. This parameter was changed until the resulting rotational acceleration was null, thus guaranteeing the static condition of the mandible, ie, the balance between the resulting force and the muscular forces. Muscle strength magnitudes were then collected (Table 3) and used as input data in the static analysis of this work. Finally, in the static analysis, the mandible was restricted to...
movements in both condyles in all directions. Complete temporomandibular joint movements were unnecessary for this model (Fig 4).

Fatigue Analysis
Fatigue analysis was performed based on the number of chewing cycles exercised by a person at a given time. These data were established according to statistical evaluations that discovered that years of chewing in vivo corresponded to 5 million chewing cycles.29,30 In addition, the infinite life criterion for metal alloys was 10^9 (1 billion) cycles.31–33

The structure sets (prosthetic components and implants) were evaluated for their ability to withstand the load associated with the cyclic movement of the jaw during chewing, consisting of fatigue analysis. The Ansys Mechanical postprocessing tool (version 18.2, Ansys), known as the fatigue tool, was used for this analysis.

After obtaining the equivalent Von Mises stresses, through the structural analysis of the model, the cyclic loading in this tool was defined. The masticatory cycle was considered as a wave of sinusoidal load whose minimum value was 0, corresponding to the moment when the patient was not exerting masticatory force. The maximum value was associated with the resulting stress state, referring to the moment when the oblique load of 100 N was acting on the molar region.

In this work, it was decided to adopt a more conservative failure theory, which is often used as a criterion for device designs. While the Gerber curve is a measure of average behavior for fatigue failure, the Goodman curve represents a measure of minimal behavior.31 Thus, in this work, Goodman’s theory was used as a failure criterion and methodology for calculating the fatigue strength of each component. The stress–number of cycles (S–N) method was adopted to predict component durability, and S–N diagrams related to the fatigue strength limits of the Ti and Co–Cr alloys, 825 and 552 MPa, respectively,30,33,31 were entered as input data in Ansys Mechanical for performing calculations (Fig 5). Consequently, the results of the expected life were found for the components in question.

Rating Criteria
Computational analyses of the finite element models were performed using the Ansys program (version 18.2, Ansys). The following evaluation criteria were adopted to interpret the data obtained:

- Implants and prosthetic components failed if the stress exceeded 825 MPa, which represents the yield strength of the Straumann Ti alloy (Institut Straumann).
Tensile stresses > 76 MPa and compression > 170 MPa in the mandible will cause bone resorption.12

A von Mises tension > 552 MPa (flow limit of the Co–Cr alloy)32 in the metallic framework region leads to its failure.

An estimated service life of < 5 × 10⁶ cycles, for any component in the set of structures, will indicate fatigue failure of that component before 5 years of prosthetic rehabilitation.29

RESULTS

Analysis of Stresses After Oblique Loading

Because of the application of the resultant load of 100 N, the frequently used areas of the implant, prosthetic components, and peri-implant bone were concentrated on the first molar in the region of the left cantilever.

Implants

The general profile of the stress in the implant in the first molar region was similar in all groups except G4, which presented much higher values. The result of von Mises stress in the implant and its resistance limit after loading in the molar region on all models did not exceed the resistance limit of 825 MPa. G4 had a von Mises stress of 273.79 MPa, which was more than double the stresses found in G6, G8, and G10, which were 131.74, 119.57, and 111.80 MPa, respectively. A stress range from 30 to 45 MPa was predominant in the transition region from the body to the cervical part of all implants. The highest stress region in the G4 implant was concentrated internally, which is in close contact with the abutment (Figs 6 and 7). This site was characterized as a high-stress concentration region, with a maximum value of 273.79 MPa. As the length increased, the stresses of groups G6, G8, and G10 were concentrated in the region of the first threads of the implant with the mandible. It was also observed that none of the implants exceeded the resistance limit.

Abutment

The result of von Mises stress in the abutment and its resistance limit after loading in the molar region on all models did not exceed the resistance limit of 825 MPa. G4 had a von Mises stress of 222.80 MPa, while G6, G8, and G10 had 299.45, 267.93, and 239.61 MPa, respectively. For the whole body of the abutment, G6, G8, and G10 were predominantly < 50 MPa. Nonetheless, G4 had the lowest maximum tension peak. The location of the abutment that had the highest tension for all groups was in the region of the first threads in contact with the prosthetic screw (Figs 8 and 9).

Metallic Framework

The result of von Mises stress in the metallic framework and its resistance limit after loading in the molar region on all models did not exceed the resistance limit of 522 MPa. G4 exhibited a maximum stress value that surpassed those of the other groups with a value of 379.34 MPa, which was approximately fourfold higher (Fig 10). However, its location was in a small region inferior to the metallic framework (Fig 11). Furthermore, its value corresponded to 72.6% of the yield strength of the material. In the other regions of the G4 metallic framework, the von Mises stress attained a maximum value of approximately 100 MPa. The stress profile in groups G6, G8, and G10 was practically the same, with its maximum value occurring in the upper region of the coping (Fig 11). A lower stress was observed in all the groups relative to the yield strength of the Cr–Co alloy metallic framework.

Prosthetic Screw

The maximum stress in the G4 group occurred in the region of the first threads of the prosthetic screw with the abutment. In the other groups, this stress was located in the rounded screw heads (Fig 12). The result of von Mises stress in the prosthetic screw and its resistance limit after loading in the molar region on all models did...
not exceed the resistance limit of 825 MPa. G4 had a von Mises stress of 300.39 MPa, while G6, G8, and G10 had 296.45, 265.82, and 257.51 MPa, respectively.

Peri-implant Bone
The stress peaks were distributed at the level of the peri-implant cortical bone of all the groups in the region close to the load application, and they decreased as the length of the implants increased. However, there were small differences between the G6, G8, and G10 groups (Fig 13), while the highest stress values were observed with G4, thereby suggesting a probability of peri-implant bone resorption (Table 4). For oblique loading, the main maximum and minimum stresses in the cortical bone in the studied models showed that G4 had a higher minimum normal stress (compression) and maximum normal stress (tensile) than the other groups.
Fatigue Analysis

Before the fatigue analysis, the von Mises stresses obtained based on the applied loads were compared with previous studies to validate the model and guarantee its safety against failures caused by static loading. All fatigue analyses were performed using the infinite life criterion, since the range corresponds to a number > 10⁹ cycles.

The FEM conducted in this study revealed that for the loading conditions tested, the entire geometry of the groups of implants and prosthetic components would have a useful life that surpasses 10⁹ cycles. However, the metallic framework of G4 will resist 4.8415 × 10⁷ (or 48.4115 million) cycles, which corresponds to 48.415 years before failing due to fatigue.

DISCUSSION

Studies with implant-supported prostheses have shown, through finite element analysis, that oblique loads cause more damage to the implant or prosthesis system, thereby revealing a tendency to generate high stresses due to system fatigue. Therefore, the present study investigated the stresses generated by oblique loads in fixed full-arch implant-supported prostheses rehabilitated with different implant heights in the atrophic mandible, thus suggesting more discrepant behavior of 4-mm–long implants aside from bone resorption.

Notably, the results presented herein refer to the stresses generated by the application of only a 100-N oblique load at an angle of 45 degrees in the cantilever region of the working side. Some authors argue that, in cantilevered extensions, the load increases in the prosthetic components and adjacent implants, justifying the simulation of the oblique load in this area in the present study.

For each model, the connections between the prosthetic screws and abutment, as well as between the implants and the mandible, were defined through a thread correction feature, which represents the behavior of screw connections through their step data, threads, and nominal diameter. Consequently, the connections between the implants and the abutment were defined as juxtaposed. Therefore, models were developed under necessary conditions to obtain a better representation of the real physical situation to have a result closer to the expected performance in the patient’s mouth.

In the present study, the properties of the Co–Cr alloy were applied to the metallic framework because this type of alloy exhibits more satisfactory results, including good corrosion resistance, reduced cost, and a high modulus of elasticity. Therefore, it favors a more uniform stress distribution throughout the entire metallic framework, which is also noticeable from the results obtained in this work. The metallic framework in groups G6, G8, and G10, along with the other structures, tends to deform predominantly with a translation movement; Fig 14 represents the displacements. The G4 group, however, presents a completely different behavior. In this group, the metallic framework has a much more pronounced rotational movement (Fig 15). This is likely caused by the shorter lengths of the implants not providing as much of a lever arm as the longer ones, and the total moment applied to the framework being resisted by the small contact area between the cervical part of the implant and the internal part of the coping, hence justifying the high stress in this small framework region. The discrepancy in the displacement behavior of the G4 metallic framework can be justified by the difference in implant design, as this is the only factor that varies between the models.

The implants evaluated in this work are made of a titanium alloy, and they differ in length. The G4 implant, in addition to having a smaller body, also has a smaller thread pitch. These two characteristics make the implant more fixed to the mandible. Therefore, it is difficult for the metallic framework to impose oblique deformation because it cannot move the implants in the same direction. These implants functioned as flexion points in the metallic framework, causing it to rotate. This generated different physical behaviors in practically the entire prosthetic structure, with a maximum stress value higher than that of the other groups, which predominantly occurred in the internal region of the cervical implant in G4.

The implants in G6, G8, and G10 allowed for a greater relative displacement with the mandible and provided an easier deformation in the direction imposed by the

<table>
<thead>
<tr>
<th>Group</th>
<th>Bone maximum normal stresses (MPa)</th>
<th>Bone Minimum normal stresses (MPa)</th>
<th>Resistance limit (traction)</th>
</tr>
</thead>
<tbody>
<tr>
<td>G4</td>
<td>125.38</td>
<td>–142.7</td>
<td>72–76</td>
</tr>
<tr>
<td>G6</td>
<td>74.82</td>
<td>–77.26</td>
<td>140–170</td>
</tr>
<tr>
<td>G8</td>
<td>58.434</td>
<td>–71.498</td>
<td></td>
</tr>
<tr>
<td>G10</td>
<td>65.014</td>
<td>–67.19</td>
<td></td>
</tr>
</tbody>
</table>

MPa = Megapascal.
oblique load. Therefore, maximum stresses occurred in the first threads of the implant closest to the cantilever in the region of contact with the cortical bone, as previously observed by other authors.37,41,42 However, all the groups presented stress values below the threshold of 825 MPa, suggesting that the oblique load applied to this model did not exceed the absorption capacity of the system and consequently will not fail due to overloading nor inadequate distribution of the masticatory forces.

The prosthetic components (abutment and prosthetic screw) were also made up of the same titanium alloy, and their stress distribution was strongly related to the behavior of the metallic framework. The prosthetic screw in G4 suffered a significant and predominantly horizontal flexion deformation due to the rotational movement of the metallic framework. This bending deformation provided greater contact between the abutment and the internal cervical region of the implant, thereby justifying the values of the significant stresses in the abutment body, which were higher in the internal cervical part of the implant relative to the other groups, wherein screw displacement was predominantly oblique (Fig 16), similar to the deformation of the metallic framework. On the other hand, the screws in the G6, G8, and G10 groups exhibited a much more significant vertical displacement than in G4, leading to higher stresses in the internal threads of the abutment with the screws (Fig 17).

The most frequent mechanical failures in implant prostheses are looseness and fracturing of the prosthesis and abutment screws.5 Notably, in this study, the stresses were more concentrated in the first threads of the G4 prosthetic screw, which was suggestive of a greater probability of this screw suffering looseness or fractures. Notwithstanding the G4 results, it was discovered in this study that there was no significant difference in the distribution of the stresses in the prosthetic screws of the other groups. Furthermore, the importance of more robust screws rather than thinner screws to support greater loads, especially in 4-mm implants, is also noteworthy.
The present study followed the structural pattern already described in the literature. The study demonstrated that increasing the length of the implants in the rehabilitation of the atrophic edentulous mandible with a fixed full-arch implant-supported prosthesis reduced the stress intensity for the oblique loads in the molar region at the bone and implant levels, as well as in most of the prosthetic components, for all groups except G4, which had unique metallic framework behavior.

Korioth and Johann observed that implant stress could be significantly affected by the shape of the metallic framework under different mandibular loading conditions. In the present study, the shape of the metallic framework selected for this analysis was rectangular, compared to other configurations. It is important to note that the unbalanced occlusal pattern influences the stress distribution in the metallic prosthetic framework. Therefore, if an implant-supported fixed prosthesis is not well planned and adjusted, resulting in some premature contact or interference, the prosthetic framework will inevitably fail.

During the mandible result analysis, it was observed that the stresses were concentrated in the cortical bone for all models and regions of load application. Pregraded studies also show a predisposition to concentrate higher levels of tension in the cortical bone. This demonstrates that, among other factors, the inherent characteristics of the cortical bone, which has a greater modulus of elasticity compared with the cancellous bone, predispose it to greater stress. When applied to oblique loads, this cortical bone acts as the fulcrum of the system (moment of flexion), concentrating the highest stresses compared with the cancellous bone, indicating that the peri-implant bone may be more susceptible to bone loss, according to some studies. In the G6, G8, and G10 models, the stress distribution was theoretically within the physiologic limits of the resistance of the human cortical bone of 140 to 170 MPa of compression and 72 to 76 MPa in tensile strength. Notably, these limits are based on a study of the femoral bone in 1996; therefore, it is suggested to update the studies in this regard. For the G4 model, the maximum and minimum main stress levels in the bone, 125.38 and 142.7 MPa, respectively, revealed a pathologic microdeformation trend. The most significant increase in the values of the maximum stresses in G4 compared with the other groups is because of the shorter implant length of 4 mm and smaller thread pitch. For the other groups, the levels of the maximum and minimum main stresses in the bone were consistent with the bone compatibility based on resistance to the loading.

All models in this study showed that there was a higher concentration of stress in the contact region between the cortical bone and the first threads of the implants. The analyses presented adopted the von Mises theory of failure as a reference. This theory is the most accurate for studying materials whereby the effect(s) of internal friction is negligible. For instance, the effect of internal friction can be neglected in bones because they have a porous aspect and anisotropic properties. Thus, the von Mises theory has been used in structural analysis studies on bones and is presented herein as a criterion for comparing the results between different implant models. However, due to the fragility of the bones, the maximum and minimum principal stresses were also analyzed in a previous study to better elucidate the effects of implants on the mandible.

It is important to emphasize that all analyses, both static loading and fatigue, were done based on a single atrophic mandible of an edentulous patient and under the specific condition of an oblique masticatory load of 100 N. This condition idealized the models analyzed in this work, so the results should be seen as preliminary for understanding the mechanical performance of the implants adopted. Therefore, the results presented by these analyses can provide the probable durability of fixed full-arch implant-supported prostheses when used in the treatment of edentulous patients. New simulation work is suggested with other possible input data, such as other anatomical models of the atrophic mandible.

Based on the similarity of the stress distribution and fatigue of implants and prosthetic components for all the models, it is suggested that short implants are seemingly viable alternatives for the rehabilitation of atrophic edentulous mandibles with fixed full-arch implant-supported prostheses when clinical reality inhibits the use of longer implants. However, the 6-mm–long and 8-mm–long implants showed a more favorable mechanical behavior than the 4-mm–long type. It is also suggested that long-term clinical studies should be conducted to evaluate the screw loosening index using computed tomography scans showing the level of bone resorption in the rehabilitation of atrophic edentulous mandibles with fixed full-arch implant-supported prostheses with short implants to provide more substantial support of these results.
CONCLUSIONS

Through this in vitro study, it was concluded that:

- The quantified stress in all implant length models could not generate failure in the implants after applying a 100-N oblique load, as the values did not exceed the yield strength of 825 MPa of the material of the implant.
- The stresses in the prosthetic components presented approximate values for the G6, G8, and G10 groups, but not for G4, which, in most situations, presented more discrepant results. Nonetheless, no group exceeded the yield strength for the materials of these components.
- For the cyclic loading conditions tested, the prosthetic and implant components will have infinite life, except for the G4 metallic framework.
- Because of the similarity in the behavior of the stresses with the oblique loads and the fatigue test between the models, it is suggested that all implants are viable alternatives, but that biomechanically, the 6-mm and 8-mm implants behave better than the 4-mm implants for cases of atrophic edentulous mandibles that need to be rehabilitated with fixed full-arch implant-supported prostheses, without the possibility of placing longer implants, according to the FEM.

ACKNOWLEDGMENTS

The authors reported no conflicts of interest related to this study.

REFERENCES