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The influence of bone quality on the biomechanical behavior of full-arch implant-supported fixed prostheses

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We evaluated the influence of bone tissue type on stress distribution in full-arch implant-supported fixed prostheses using a three-dimensional finite element analysis. Stresses in cortical and trabecular bones were also investigated. Edentulous mandible models with four implants inserted into the interforaminal region were constructed from different bone types: type 1 — compact bone; type 2 — compact bone surrounding dense trabecular bone; type 3 — a thin layer of compact bone surrounding trabecular bone; and type 4 — low-quality trabecular bone. The mandible was restored with a full-arch implant-supported fixed prosthesis. A 100-N oblique load was applied to the left lower first molar of the prosthesis. The maximum (σmax) and minimum (σmin) principal stress values were determined. The σmax in the type 4 cortical bone was 22.56% higher than that in the type 1 bone. The σmin values in the cortical bone were similar among all the bone types. For the superstructure, increases of 9.04% in the σmax and 11.74% in the σmin in G4 (type 4 bone) compared with G1 (type 1 bone) were observed. For the implants, the highest stress values were located in G4, and the lowest values were observed in G1. In the trabecular bone, the highest stress was generated in G1 and G2. In conclusion, the more compact bones (types 1 and 2) are the most suitable for supporting full-arch implant-supported fixed prostheses, and poor bone quality may increase the risk of biological and mechanical failure.

1. Introduction

Rehabilitation with implant-supported prostheses is considered a feasible therapy for edentulous patients [1–4]. Currently, approximately 300,000 patients per year are treated with dental implants in the United States [5]. The insertion of four to six implants into the interforaminal area of the edentulous mandible provides great stability for implant-supported prostheses with a distal cantilever [6,7]. However, factors such as oral hygiene, the inter-arch relationship, treatment cost, patient acceptance, and masticatory function may influence treatment options [8–10].

In this sense, the bone availability and inter-arch space should be appropriate for the placement of implants in the mandibular anterior region [11]. The volume and quality of the alveolar ridge influence the biomechanical and esthetic results, the stability of the implant-supported prostheses, and the health of the surrounding tissues [12]. After tooth loss, the structure of the mandibular bone undergoes a constant process of physiological resorption, which causes the decline of the alveolar perimeter and the expansion of the trabecular bone, decreasing the bone density. These factors may influence treatment with dental implants [13,14].

The quality of bone tissue is classified into the following four categories based on the ratio of cortical to trabecular bone: type 1 — primarily compact bone; type 2 — compact bone surrounding dense trabecular bone; type 3 — a thin layer of compact bone surrounding trabecular bone; and type 4 — a thin layer of cortical bone surrounding low-density trabecular bone [15]. The quality of the bone architecture influences the transfer and distribution of physiological forces, which dictates the treatment prognosis [16,17]. Low-quality bone tissue, especially type 4, is associated with a high rate of implant treatment failure [18] due to a reduced cortical/trabecular tissue ratio and low adhesion force, which jeopardizes osseointegration [16–21].

A three-dimensional finite element analysis (3D-FEA) has previously been used to evaluate the performance of bone tissue with different quality patterns in implant-supported single crowns attached to implants of different lengths [22] and in multi-unit prostheses with prefabricated bars [14,23]. For low-quality bone tissue, an increase in implant length has been shown to reduce stress distribution [19,22].

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Other studies [14,23] have demonstrated that bone types 3 and 4 generated the highest stress concentrations under axial and buccolingual loads. Type 4 cortical bone also exhibited high stress values under all loading conditions [14]. However, it has been suggested that bone quality is not the only factor that influences stress distribution because bone tissue helps to support implant-supported prostheses retained by prefabricated bars [14].

There are limited data concerning the factors that affect the biological performance of bone tissue and the stress patterns associated with different designs of implant-supported prostheses. Thus, the aim of this study was to evaluate the influence of different bone types (Types 1 to 4) on stress distribution in mandibular full-arch implant-supported prostheses using an FEA model based on computed tomography (CT) images. We hypothesized that the stress on the implant/superstructure assembly and on the peri-implant bone tissue would be significantly lower in bone types 1 and 2.

2. Materials and methods

This study was approved by the Human Research Ethics Committee of Aracatuba Dental School-UNESP, Brazil (process number: 2008–00939).

The geometry of the completely edentulous mandible of a 60-year-old man was reconstructed from cone-beam CT images (I-Cat Cone Beam Volumetric Tomography and Panoramic Dental Imaging System, Imaging Sciences International, Hatfield, PA, USA). The patient was informed about the procedure and signed an informed consent form. The mandibular section was imaged with 2-mm slices, and the patient was rehabilitated with a conventional complete denture. The denture was duplicated in self-polymerized acrylic resin mixed with barium sulfate in a 3:1 ratio to allow for the radiopacity of the denture during the CT scan. After the duplicated denture was adjusted, the CT scan was performed.

The CT assessment data were imported into the Simpleware 4.1 software package (Simpleware Ltd, Rennes Drive, Exeter, UK) for the construction of the 3D solid geometries of the edentulous mandible and denture. Based on the actual positions of the mandible and denture, the mucosal geometry was deduced, and the mucosa remained in contact with the inner surface of the denture [22]. In the edentulous mandible, both the cortical and trabecular bones were determined according to the CT data. The mucosa and the cortical bone were approximately 3.0 and 1.5 mm thick in the interforaminal area, respectively.

A 3D constitutive model of an edentulous mandible was obtained. Four different types of bone (types 1, 2, 3 and 4, with varying elastic moduli of the bone tissue) were used based on the bone quality classification system suggested by Lekholm and Zarb [15]. The model was rehabilitated with a fixed full-arch implant-supported prosthesis.

Four implants, 11.5 mm in length and 3.75 mm in diameter, were modeled using CAD software (SolidWorks 2010, Dassault Systèmes SolidWorks Corp., Concord, MA, USA) and were virtually inserted into each model. In all the models, the implants were placed in the center of the mandibular alveolar crest, 10 and 20 mm away from the midline on both sides of the mandible [24], as shown in Fig. 1.

The implants and prosthetic components were imported into the Simpleware software and were merged with the edentulous mandible and prosthesis in all the groups according to the level of bone quality (G1 — bone type 1; G2 — bone type 2; G3 — bone type 3; and G4 — bone type 4). Finally, a finite element mesh of the models was obtained using the Simpleware software. The mesh refinement was established based on a convergence analysis (5%) [25]. The models contained a total of 244,388 elements and 70,387 nodes, as shown in Fig. 2.

The meshed models were imported into finite element analysis software (Abaqus 6.10-EF1, Dassault Systèmes Simulia Corp., Providence, RI, USA) to evaluate the stress distribution. The mechanical properties (elastic modulus and Poisson coefficient) of the materials are presented in Table 1 [23,26–29].

Complete bonding between the bone tissue and implants was assumed to simulate osseointegration with no motion between the structures during loading [18,30–33]. To reproduce the clinical setting, a contact was applied between the implants and the superstructure [34]. The superstructure was glued to the acrylic resin prosthesis [35].

The models were supported by the masticatory muscles and temporomandibular joints. The forces generated by the masticatory elevator

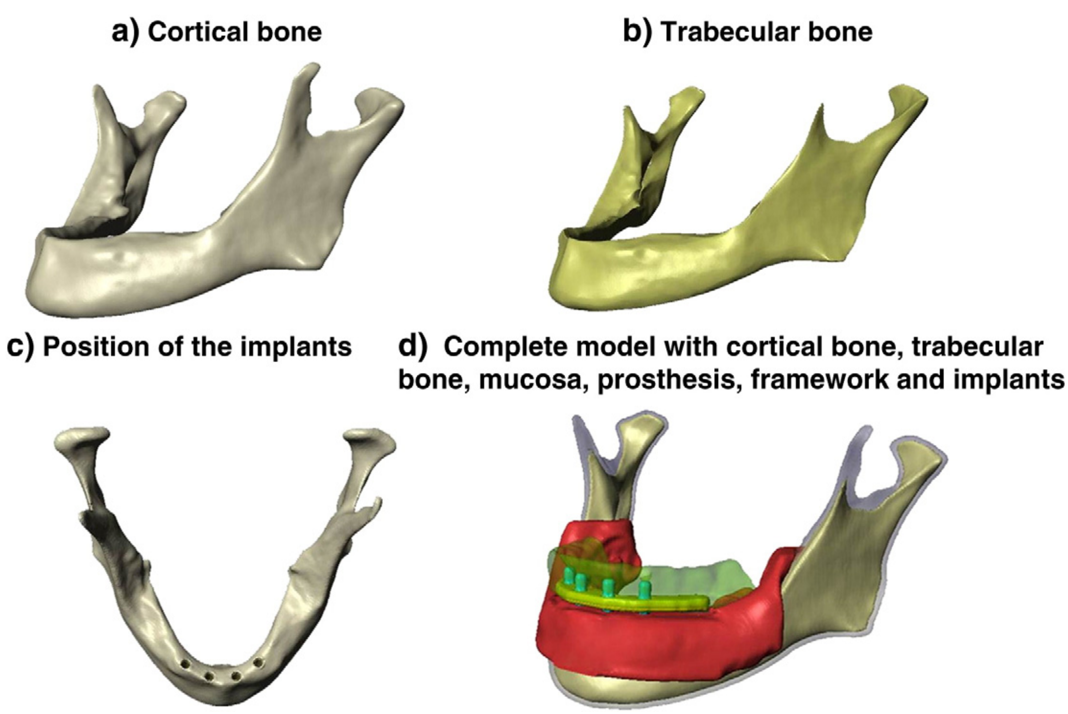

Fig. 1. Models representing a) cortical bone, b) trabecular bone and c) implants inserted into the interforaminal region. d) Complete model with cortical bone, trabecular bone, mucosa, prosthesis, framework and implants.

Fig. 2. Finite element meshes of the groups. The models contained a total of 244,388 elements and 70,387 nodes.

muscles (temporal, masseter, medial pterygoid and lateral pterygoid) and their positions on the mandible were established according to previous studies [30–34]. The mandibular coronoid process was used to insert the temporal muscle, whereas the angle and lower half of the lateral surface of the ramus underwent masseter insertion. The medial pterygoid was inserted into the lower and back areas of the medial surface of the ramus and angle of the mandible, whereas the lateral pterygoid was inserted into the neck of the mandibular condyle [36]. A total of 10 nodes of finite elements were used to define each muscular area. The force directions were established by the cosines α, β and γ [31], representing the x-, y- and z-axes of each force, respectively. Muscular forces of 59.23 N for the masseter, 39.60 N for the medial pterygoid, 34.44 N for the lateral pterygoid, and 34.09 N for the temporal were applied [32–34]. A rigid contact between the condyle and the glenoid fossae was simulated in the temporomandibular joint [34]. The nodes placed in the front bevel face of the condyles were fixed in three degrees of freedom to support the temporomandibular joints [37] (Fig. 3).

An oblique load of 100 N (30 degrees of inclination in relation to the long axis of the implant) was applied to simulate the mean value of the posterior bite force in humans [22,27,34]. The load was applied in the buccolingual direction to the left lower first molar of the prosthesis in each bone type (Fig. 3). The maximum (σmax) and minimum (σmin) principal stress values in the components of the implant/superstructure assembly and the cortical and trabecular bones were determined to better understand the influence of bone quality on stress distribution in the system.

3. Results

The stress values and distribution (σmax and σmin) in the implant-supported prosthesis system are shown in Figs. 4, 5, 6 and 7. The σmax is related to the tensile force and is therefore a positive value, whereas the σmin indicates the compression force and is a negative value. To quantify the highest stress values (depicted in Fig. 4), the node with the maximum value in any case was selected for each structure.

For the cortical bone, superstructure and implants, the highest stress values were observed in type 4 bone (G4) and the lowest values in type 1 bone (G1). The σmax in type 4 cortical bone was 22.56% higher than that in type 1 bone. The σmin values in the cortical bone were quite similar among all the bone types. However, for the trabecular bone, bone types 3 and 4 (G3 and G4) exhibited lower stress values than bone types 1 and 2 (G1 and G2) in terms of the σmax and σmin values (decreases of 88.83% and 82.48% for G4 compared with G1, respectively) (Fig. 4).

In the superstructure, an increase in stress was observed as the bone quality decreased. G4 exhibited an increase of 9.04% in the σmax and 11.74% in the σmin (σmax = 1882.48 MPa and σmin = −1540.44 MPa) compared with G1 (σmax = 1712.14 MPa and σmin = −1506.98 MPa). For the implants, G4 exhibited the highest stress values (σmax = 789.91 MPa and σmin = −1086 MPa), whereas G1 showed the lowest values (σmax = 712.64 MPa and σmin = −1006.31 MPa). The stress increased 9.78% for the σmax and 7.33% for the σmin (Fig. 4).

Regarding the stress distribution in the cortical bone, both analysis criteria (σmax and σmin) exhibited the highest stress concentration in the peri-implant region for all the groups, primarily on the left side (loading region) of the distal implant (Fig. 5). In the trabecular bone, increased stress areas were noted around the implants (Fig. 6). For the implants and superstructures, the highest stress concentrations were observed in the lingual region of the implant, the distal portion of the bar, and the distal region of the left side; these results were similar to those for the cortical bone (Fig. 7).

4. Discussion

The results of this research confirmed our hypothesis that the stress in both the implant/superstructure assembly and the peri-implant bone tissue would be significantly lower in bone types 1 and 2. The stress distribution was reduced in the cortical bone, implants and superstructure with bone types 1 (G1) and 2 (G2). However, the stress level in the trabecular bone was lower for bone types 3 (G3) and 4 (G4).

Table 1
Mechanical properties of the materials used in this study.

Material	Elastic modulus (MPa)	Poisson coefficient	References
Type 1 bone	9500	0.3	Almeida et al. [23]
Type 2 bone	5500	0.3	Almeida et al. [23]
Type 3 bone	1600	0.3	Almeida et al. [23]
Type 4 bone	690	0.3	Almeida et al. [23]
Cortical bone	13,700	0.3	Barbier et al. [26]
Metallic bar (gold alloy)	120,000	0.25	Almeida et al. [23]
Mucosa	680	0.45	Barao et al. [27]
Titanium implant	103,400	0.35	Sertgoz and Gunever [28]
Acrylic resin	8300	0.28	Darbar et al. [29]

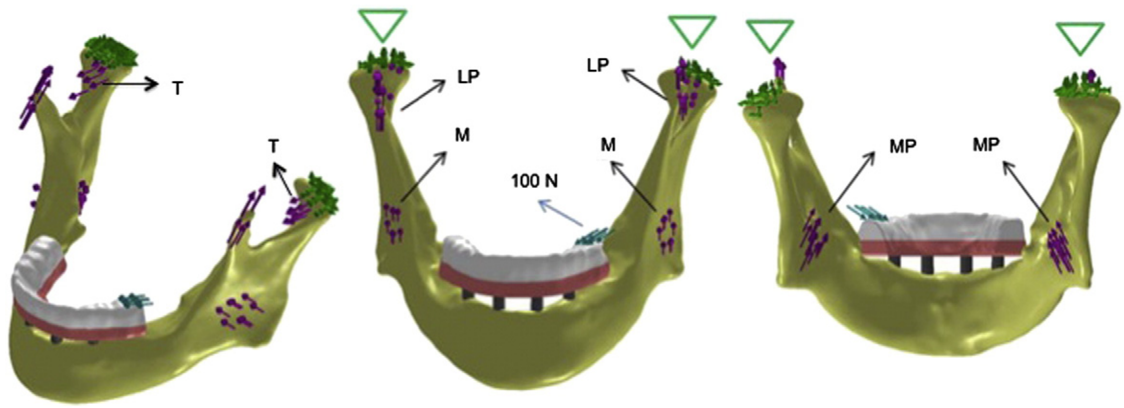

Fig. 3. Boundary conditions and model restrictions. The black arrows indicate the load application of each masticatory muscle (LP—lateral pterygoid, MP—medial pterygoid, M—masseter, T—temporal). The green triangles illustrate the restriction condition of the condyles. The blue arrows indicate the direction and area of the load application.

In this study, the highest stress values for the cortical bone, superstructure and implants were observed with type 4 bone (G4). In type 1 bone, the elastic modulus of the trabecular bone is higher than that in bone types 2, 3 and 4. Therefore, a higher elastic modulus provides more resistance to deformation [34], which explains our results. These findings are in accordance with previous studies [22,23,25], and these data are in accordance with the increased failure rates of implants inserted into type 4 bone, which are due to the lack of stability at the bone–implant interface [38–40]. Although Tada et al. [22], Almeida et al. [23] and Pessoa et al. [25] observed higher stress values in low-quality trabecular bone (type 4 bone), the present study demonstrated lower stress values in trabecular bone of bone types 3 and 4 (decreases of 88.83% in the σmax and 82.48% in the σmin).

Almeida et al. [23] evaluated stress distribution in edentulous mandibles rehabilitated with full-arch implant-supported fixed prostheses with prefabricated superstructures. The highest stress in trabecular bone was observed in bone types 3 and 4 under all loading conditions, which is in disagreement with the present findings. In the study conducted by Almeida et al. [23], the models were fixed in their base; however, the present study fixed the models in the condyles, which allowed a mandible deflection similar to that observed under clinical conditions [22]. In addition, the different stress values reported by Almeida et al. may have resulted from variations in cortical bone thickness among the study groups [23]. Thus, the mandibular deflection (simulation of mandibular movements) associated with variations in the trabecular bone elastic modulus while the cortical bone thickness was maintained may explain why the lowest stress values occurred in the trabecular bone of type 4 bone (G4) in the current study.

Similarly, Tada et al. [22] assessed the performance of implants and peri-implant bone tissues by performing a finite element analysis that considered four bone types (types 1, 2, 3 and 4) and different elastic moduli of trabecular bone (9.5, 5.5, 1.6 and 0.69 GPa), but with the same cortical bone thickness in all the groups. Regardless of the loading direction, lower trabecular bone elastic moduli were associated with

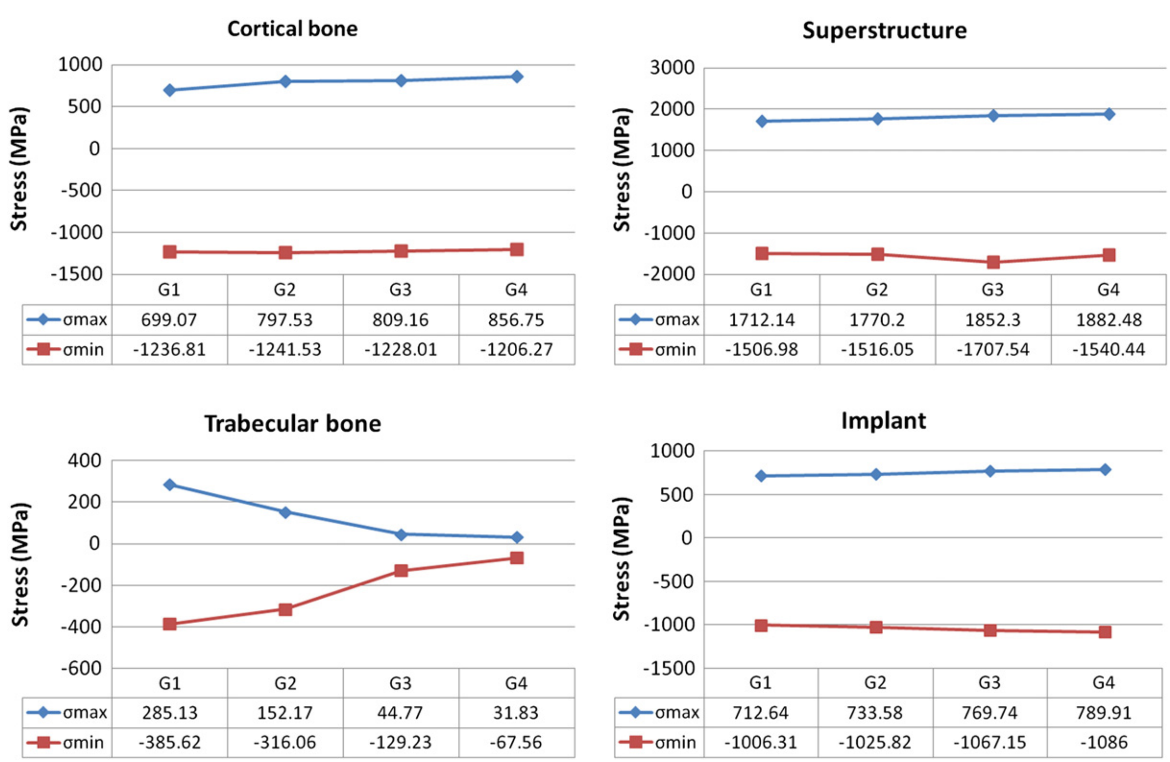

Fig. 4. Maximum (σ_{max}) and minimum (σ_{min}) principal stress (MPa) values determined for cortical bone, trabecular bone, superstructure and implants. In the cortical bone, superstructure and implants, the stress values tended to increase with decreases in the bone elasticity modulus (from G1 to G4). In contrast, in trabecular bone, the stress level decreased with reductions in the bone elasticity modulus.

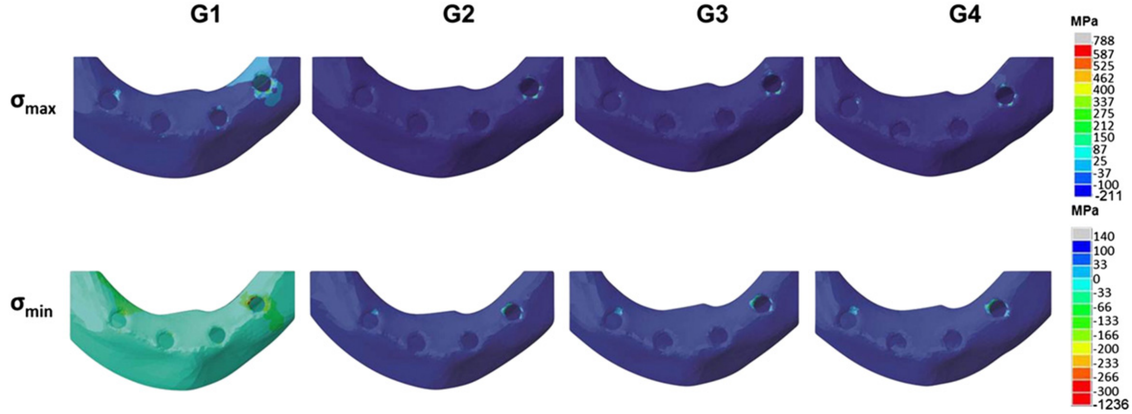

Fig. 5. Maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions determined for cortical bone in all the groups (G1, G2, G3 and G4). In all the groups, the peri-implant bone region showed the highest stress concentration, mainly in the distal implant on the left side (loading region).

higher stress values. Similar to Almeida et al. [23], the study by Tada [22] fixed the models laterally, which avoided bone deflection and resulted in higher stress in low-quality bone.

Considering that all the forces were applied unilaterally, higher stress concentrations in the cortical bone, superstructure, implants and trabecular bone were observed in the distal region of the left side in all the models. Clinically, forces on the implants and prosthetic components are applied bilaterally by the masticatory muscles. However, this study was based on previous research that also applied unilateral forces to the left inferior first molar [14,23,34]. The authors of the present study believe that similar results would be observed if the forces were applied bilaterally. Although a different stress distribution might be observed in the posterior region, the highest stress would remain in the distal area of the prosthesis/implant assembly.

Herein, an asymmetric load was used to create a propensity for denture sloping, as reported by Daas et al. [41] and Barao et al. [34]. The loaded side of the denture moves down, whereas the non-loaded side moves up [41]. This situation leads to an increased displacement of the denture, so that the supported structures (framework, implants and bone) are critically tested. Also, during the early phase of mastication, only one side of the denture is loaded [34]. It has been shown that the position of the food being masticated can affect the bending plane of the denture [41]. Future studies should evaluate the effects of different loading positions on stress distribution in implant–denture systems.

Additionally, this study used a constant load application; nevertheless, the biting load and application time are not static because chewing can vary depending on the type of food. FEA studies using a dynamic load application are warranted. Herein, a glue contact between the superstructure and acrylic resin prosthesis was simulated to provide a continuity of the displacement and traction vectors [35]. To our knowledge, no studies have determined the friction value between the structure material and an acrylic resin denture base in implant-supported prostheses.

Clinically, poor bone quality (a low elastic modulus) affects the success of the implant treatment, mainly osseointegration in the immediate loading scenario [42]. In a study of this issue, Vanden Bogaerde et al. [43] evaluated the effects of immediate loading of implants inserted in a maxilla and posterior mandible with poor bone density during 18 months of follow-up. Those authors [43] reported a minimum torque of 40 N/cm during insertion and a short-term success rate of 98%. However, the immediate prostheses were fabricated with acrylic resin without a cantilever, and they remained out of occlusion during the osseointegration period (3 to 6 months), which limited their immediate function [43]. Several studies (clinical and *in vivo*) have demonstrated higher failure rates for poor bone tissue; the causes were a lack of primary stability during implant insertion, premature loading, smoking and diabetes [38–40]. In general, these studies revealed a higher frequency of premature failure (during the osseointegration period) of implants placed in low-quality bone tissue. Accordingly, higher stress values should have been observed with bone types 3 and 4 in the present study. The results for the cortical bone confirmed this expectation; however, a decrease in stress of 78% to 88% was observed in the trabecular bone of G1 compared with G4. In this study, the implants

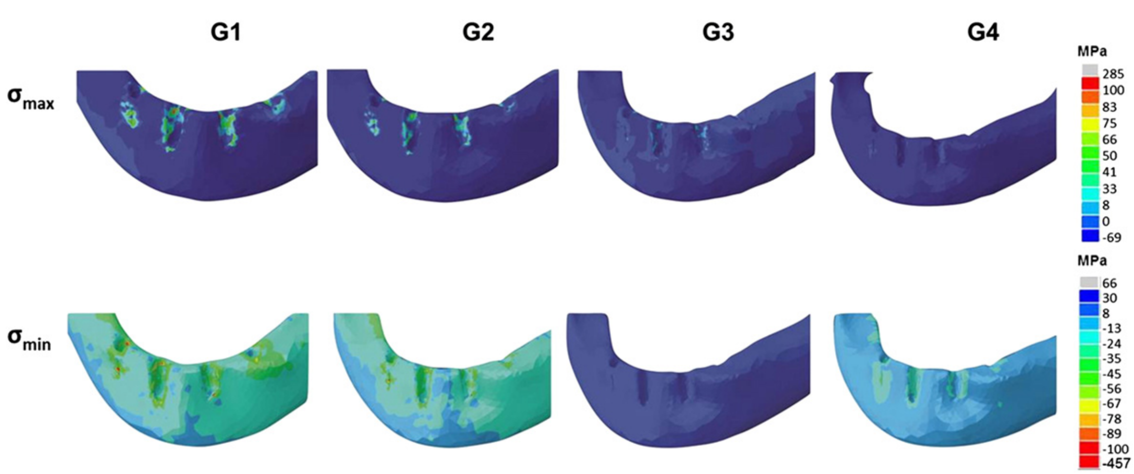

Fig. 6. Maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions measured for trabecular bone in all the groups (G1, G2, G3 and G4). Increased stress areas were noted around the implants, mainly on the non-loading side.

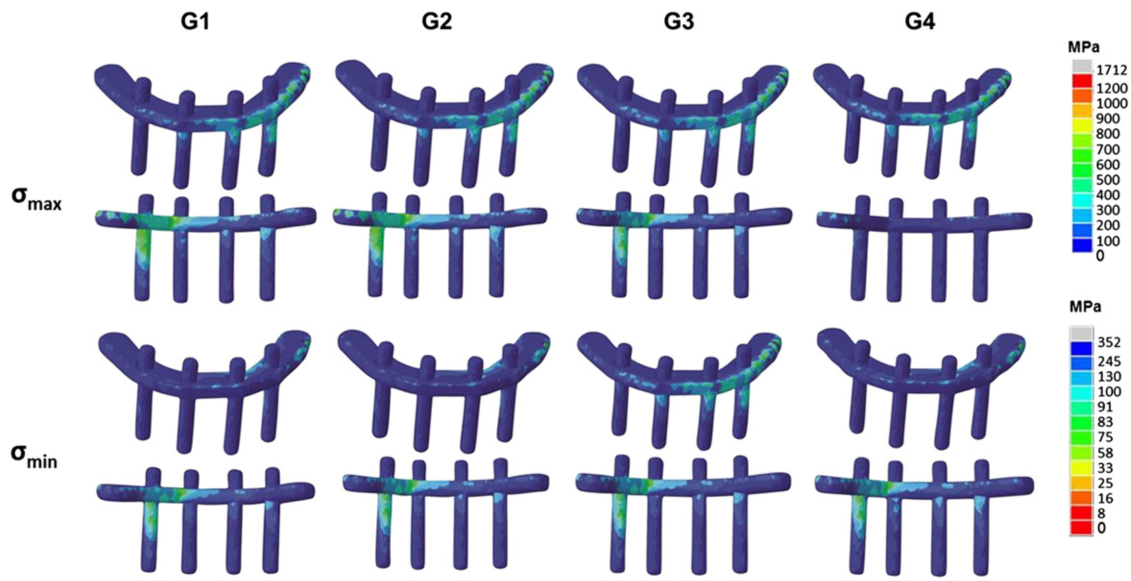

Fig. 7. Maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions measured for the implant/superstructure assembly in all the groups (G1, G2, G3 and G4). Higher stress concentrations were observed in the lingual region of the implant, the distal portion of the bar, and the distal region on the left side.

were fully osseointegrated; the insertion of contact elements at the implant/bone interface to simulate different degrees of osseointegration should be conducted in future studies with different bone quality conditions.

Considering the present finding that poor bone quality increased the stress levels in the cortical bone, implants and superstructure, higher rates of biological (i.e., bone resorption and bone fracture) and mechanical failures (i.e., screw loosening and bar fracture) should be expected in such cases. Additional factors may influence the long-term success of implant treatment, such as harmful habits (bruxism and teeth clenching) and secondary factors that cause peri-implantitis (poor oral hygiene, smoking and diabetes).

5. Conclusion

Within the limitations of the present study, it can be concluded that:

1. The highest and lowest stress values in the cortical bone, superstructure and implants were exhibited in low-quality bone (type 4 — G4) and compact bone (type 1 — G1), respectively. Poor bone quality may increase biological and mechanical failures.
2. The more compact and dense bones (types 1 and 2) are the most suitable for supporting full-arch implant-supported fixed prostheses.
3. The highest stress concentration in trabecular bone was observed in type 1 bone (G1), followed by types 2 (G2), 3 (G3) and 4 (G4).

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