



Comparison of different designs of implant-retained overdentures and fixed full-arch implant-supported prosthesis on stress distribution in edentulous mandible – A computed tomography-based three-dimensional finite element analysis



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ABSTRACT

A finite element analysis was used to compare the effect of different designs of implant-retained overdentures and fixed full-arch implant-supported prosthesis on stress distribution in edentulous mandible. Four models of an human mandible were constructed. In the OR (O'ring) group, the mandible was restored with an overdenture retained by four unsplinted implants with O'ring attachment; in the BC (bar-clip) -C and BC groups, the mandibles were restored with overdentures retained by four splinted implants with bar-clip anchor associated or not with two distally placed cantilevers, respectively; in the FD (fixed denture) group, the mandible was restored with a fixed full-arch four-implant-supported prosthesis. Models were supported by the masticatory muscles and temporomandibular joints. A 100-N oblique load was applied on the left first molar. Von Mises (σ_{VM}), maximum (σ_{max}) and minimum (σ_{min}) principal stresses (in MPa) analyses were obtained. BC-C group exhibited the highest stress values ($\sigma_{VM}=398.8$, $\sigma_{max}=580.5$ and $\sigma_{min}=-455.2$) while FD group showed the lowest one ($\sigma_{VM}=128.9$, $\sigma_{max}=185.9$ and $\sigma_{min}=-172.1$). Within overdenture groups, the use of unsplinted implants reduced the stress level in the implant/prosthetic components (59.4% for σ_{VM} , 66.2% for σ_{max} and 57.7% for σ_{min} versus BC-C group) and supporting tissues (maximum stress reduction of 72% and 79.5% for σ_{max} , and 15.7% and 85.7% for σ_{min} on the cortical and trabecular bones, respectively). Cortical bone exhibited greater stress concentration than the trabecular bone for all groups. The use of fixed implant dentures and removable dentures retained by unsplinted implants to rehabilitate edentulous mandible reduced the stresses in the periimplant bone tissue, mucosa and implant/prosthetic components.

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1. Introduction

The oral rehabilitation of edentulous patients has been improved by the use of dental implants (Branemark et al., 1977; Gallucci et al., 2009a) using fixed and removable prosthesis (Akca et al., 2007; Ekelund et al., 2003). The use of two or four implants to retain mandibular overdentures has been indicated with similar clinical and radiographic outcomes (Batenburg et al., 1998; Visser et al., 2005). In situations with short or narrow implants that require increased retention, more than two implants are inserted (Mericske-Stern et al., 2000). In order to

support mandibular fixed full-arch implant prosthesis, four to six implants are placed in the foramina area (Chee and Jivraj, 2006).

Several factors play a role on the decision between fixed and removable implant dentures as inter-foraminal space, inter-jaw relationship, oral hygiene, cost and patient's preference (Stellingsma et al., 2004). Overdentures are indicated when patients are not satisfied with stability and retention of the conventional removable denture but no complain about pain and discomfort of mucosa should exist (Zitzmann et al., 2005). Fixed full-arch implant-supported prosthesis is indicated in the presence of enough bone and inter-arch space (Chee and Jivraj, 2006). However, when there is loss of soft and hard tissues to support the facial tissue by the buccal denture flange, fixed prosthesis is contraindicated (Zitzmann and Marinello, 2002).

Overdentures are considered a simple, cost-effective, viable, less invasive and successful treatment option for edentulous

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patients (Assuncao et al., 2008; Barao et al., 2009). However, controversies toward the design of attachment systems exist (Bilhan et al., 2011). In vivo (Duyck et al., 1999; Menicucci et al., 1998) and biomechanical studies using finite element (FE) (Barao et al., 2009; Menicucci et al., 1998), strain gauge (Porter et al., 2002; Tokuhisa et al., 2003) and photoelastic (Celik and Uludag, 2007; Kenney and Richards, 1998) analyses displayed better stress distribution for overdentures retained by unsplinted implants while others showed superiority with the use of splinted implants (Assuncao et al., 2008; Cekic et al., 2007; Fanuscu and Caputo, 2004; Jofre et al., 2010).

Loading transmission during mastication influence the success of implant restorations (Tabata et al., 2011) and the loading distribution pattern in implant-retained overdentures would differ from those in implant-supported fixed restorations (Tokuhisa et al., 2003). However, no study has investigated the stress pattern of implant-retained overdentures versus fixed full-arch supported prosthesis.

This study aimed to compare the effect of different designs of implant-retained overdentures and fixed full-arch implant-supported prosthesis on stress distribution in edentulous mandible by using a 3D finite element analysis (FEA). The research hypotheses were: (i) fixed prosthesis would result in less stress level in the implant/prosthetic components and periimplant bone when compared to overdentures, (ii) the different designs of overdenture attachment systems would induce distinguished stress distribution and level in the implant/prosthetic components and periimplant bone.

2. Method

Four FE models of a complete edentulous human mandible were constructed and restored with different designs of implant-retained overdentures and fixed full-arch implant-supported prosthesis. In the OR group, mandible was restored with overdenture retained by four unsplinted implants with O'ring attachment system; in the BC-C and BC groups, mandibles were restored with overdentures

retained by four splinted implants with bar-clip attachment system associated or not with two distally placed cantilevers, respectively. Finally, in the FD group, mandible was restored with fixed full-arch four-implant-supported prosthesis.

2.1. Model design

The 3D geometry of the mandible and denture was reconstructed from cone beam computerized tomography (CBCT) (I-Cat Cone Beam Volumetric Tomography System, Imaging Sciences International, Hatfield, PA, USA) of a complete edentulous mandible of a 60-year-old man prepared with a radio-opaque (acrylic resin mixed with barium sulfate – 3:1 ratio) duplicate of his denture (Fig. 1). It was done an effort to provide the accurate relationship between denture and mandible. The mandibular section profiles were collected at 2 mm-increments. Patient signed an informed consent form according to our local Human Research Ethics Committee (#2008-00939).

The CBCT files were imported into the Simpleware 4.1 software (Simpleware Ltd, Rennes Drive, Exeter, UK) for construction of the 3D solid geometries of mandible and denture. Based on the actual position of mandible and denture, the precise geometry of mucosa in contact with the inner surface of the denture was deduced (Daas et al., 2008). In mandible, cortical and trabecular bones were delimited based on the CBCT data. The thickness of mucosa and cortical bone were approximately 3.0 mm and 1.5 mm in the interforamina area, respectively. The temporomandibular joint was modeled simulating the condyle resting in the glenoid fossa of the temporal bone (Devocht et al., 2001) with no articular disc.

Four cylindrical titanium implants with 11.5-mm length and 3.75-mm diameter were modeled (SolidWorks 2010, Dassault Systèmes SolidWorks Corp., Concord, MA, USA) and virtually inserted into each model. In all models, implants were placed in the center of the mandibular crest at 10 and 20 mm away from the midline on both sides of the mandible (Zarone et al., 2003) (Fig. 2).

The different attachment systems of the overdentures and the superstructure of the fixed full-arch implant-supported prosthesis were created into the same CAD software. The distal cantilever of the BC-C group had 3.5-mm height, 1.8-mm width and 7-mm length (Sadovsky and Caputo, 2000). In the FD group, the superstructure presented 4-mm height and 6-mm width (Sertgoz and Guvener, 1996) and a cantilever with 10-mm length (Zarone et al., 2003) (Fig. 2).

The implants and prosthetic components were imported into the Simpleware software and merged with the mandible and denture. For the FD group, denture bottom was reduced and planned. In the overdenture groups, the space for each attachment system in the inner surface of the denture was provided by a boolean operation (Fig. 2). Finally, models were meshed with parabolic tetrahedral interpolation solid elements. Mesh refinement was based on the analysis convergence (6%) (Pessoa et al., 2010). The models had 319,644 elements and 89,912 nodes in OR group,

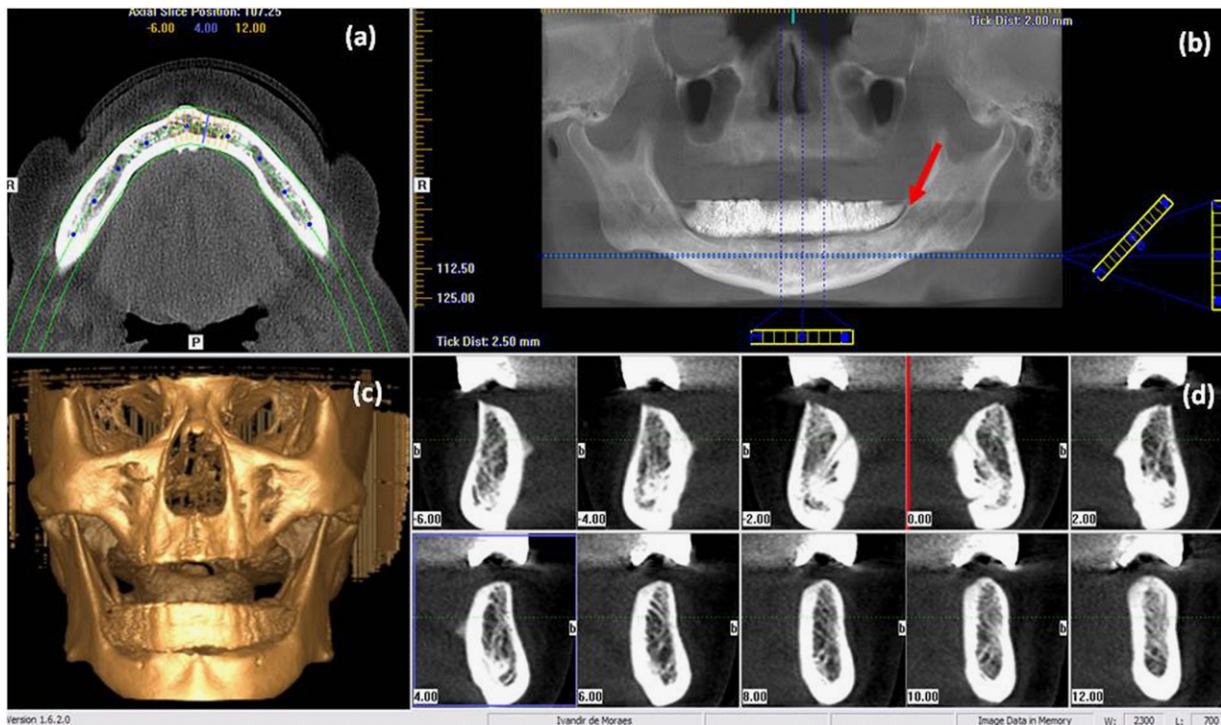


Fig. 1. Mandible CT scan of an edentulous patient with the radiopaque duplicated lower denture in position (a) Inferior view of the mandibular arch. (b) Panoramic view of the mandible with the lower denture in position (red arrow). (c) Instant 3D reconstruction of the CT scan. (d) Transverse section (2-mm thickness) of the mandible. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this article.)

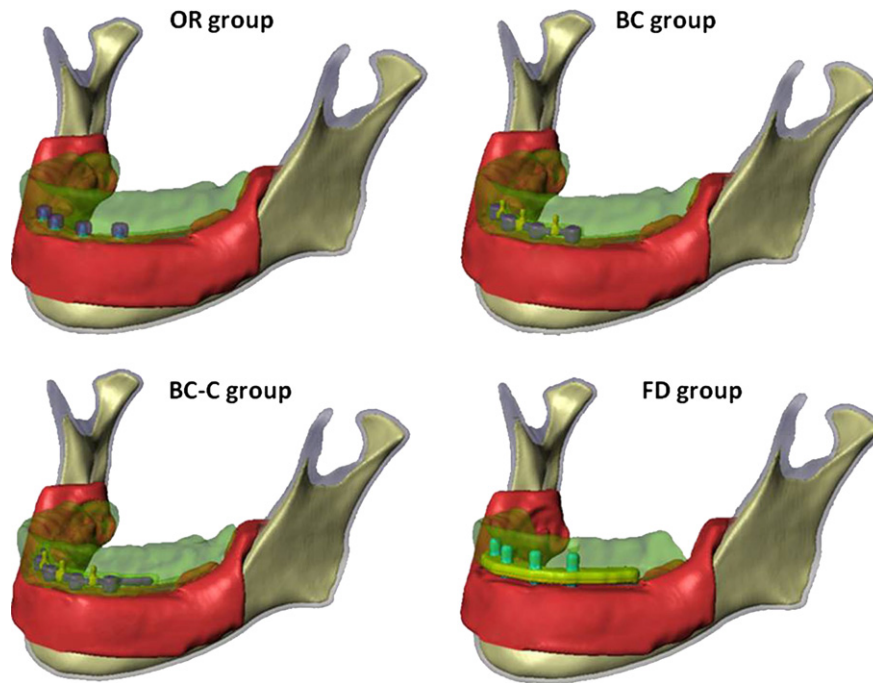


Fig. 2. 3D Complete models of all groups showing the cortical bone, trabecular bone, mucosa, denture, implants and prosthetic components.

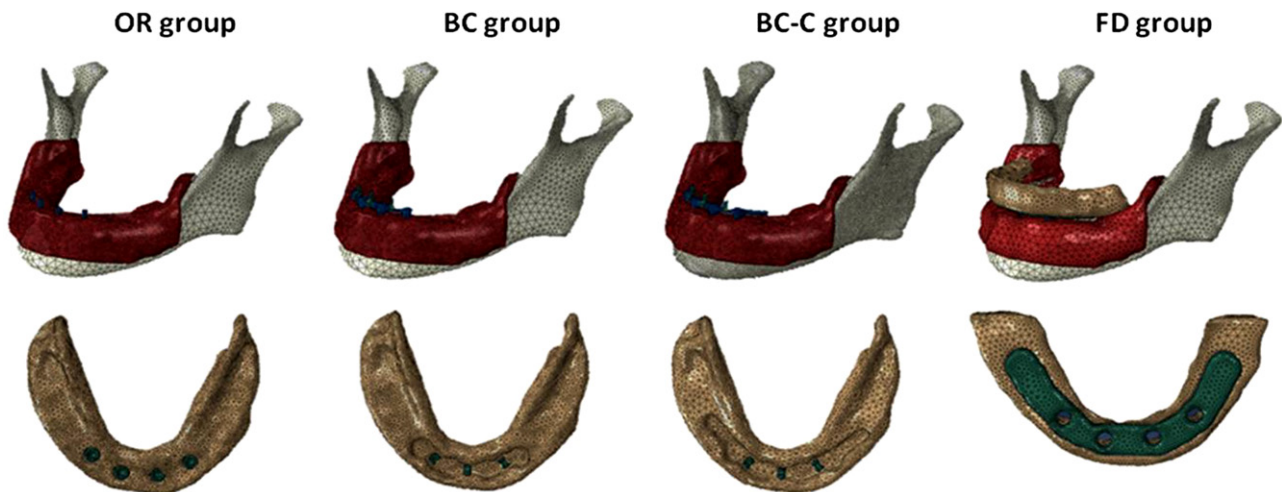


Fig. 3. Global view of meshing in each group.

318,909 elements and 89,616 nodes in BC group, 269,481 elements and 80,225 nodes in BC-C group, and 244,338 elements and 70,387 nodes in FD group (Fig. 3).

2.2. Material properties and interface conditions

Meshed models were imported into the FEA software (Abaqus 6.10-EF1, Dassault Systèmes Simulus Corp., Providence, RI, USA) to investigate stress distribution. All materials were considered isotropic, homogeneous and linearly elastic. The mechanical properties of the materials are presented in Table 1.

Total bonding between bone and implants was assumed to simulate a complete implant osseointegration so that no motion between the two structures occurs under applied loading (Cruz et al., 2003, 2006, 2009). To reproduce a clinical situation, contact was applied at the overdenture–mucosa interface and between parts of the attachments (Daas et al., 2008).

2.3. Constrains and loading conditions

The models were supported by the masticatory muscles and temporomandibular joints. The forces created by the elevator masticatory muscles (temporal, masseter, medial pterygoid and lateral pterygoid) and their positioning on the mandibular body were based on previous studies (Cruz et al., 2003, 2006, 2009)

(Fig. 4). The temporal muscle was inserted onto the coronoid process of the mandible. The masseter was inserted into the angle and lower half of the lateral surface of the mandible ramus. The medial pterygoid was inserted into the lower and back part of the medial surface of the ramus and mandible angle, while the lateral pterygoid was inserted onto the neck of mandible condyle (Gray et al., 1974). A total of 10 nodes of the elements were used to define each muscular area action. Forces directions were established by the cosines α , β and γ (Cruz et al., 2003), which represents the x , y and z axes of each force, respectively (Table 2). The applied muscular forces (in N) were 59.23 for masseter, 39.60 for medial pterygoid, 34.44 for lateral pterygoid and 34.09 for temporal. The temporomandibular joints were fixed in all three degrees of freedom at each node located in the front bevel face of the condyles (Ding et al., 2009) simulating a rigid contact between the condyle and glenoid fossae. Rigid movements of rotation and translation of the condyles were not allowed (Fazi et al., 2011).

To simulate the mean value of posterior bite force in humans, a 100-N oblique load (30° in relation to the long axis of the implant) was applied on the left first molar of each denture in a buccolingual direction (Tabata et al., 2011) (Fig. 4). The applied load was distributed dispersedly (4-mm in diameter) under the lingual cusps of the first molar to avoid false stress concentration in the loading area.

Considering that the implant–prosthetic components complex consists of a ductile material, the von Mises equivalent stress (σ_{vM}) was obtained. Nevertheless, as the cortical and trabecular bones, and mucosa are friable materials (nonductile materials), the maximum (σ_{max}) (tensile) and minimum (σ_{min}) (compressive)

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

Material	Structure	Elastic modulus (MPa)	Coefficient of Poisson	References
–	Cortical bone	13,700	0.3	Barbier et al. (1998)
–	Trabecular bone	1370	0.3	Barbier et al. (1998)
–	Mucosa	680	0.45	Barao et al. (2008)
Titanium (Ti-6Al-4 V)	Implant	103,400	0.35	Sertgoz and Guvener (1996)
Gold alloy	Fixed full-arch implant-supported prosthesis superstructure and overdenture bar	120,000	0.25	Zarone et al. (2003)
Acrylic resin	Artificial denture teeth and denture base	8300	0.28	Darbar et al. (1995)
Plastic	Overdenture clip	3000	0.28	Barao et al. (2009)
Stainless steel	O'ring capsule	19,000	0.31	Barao et al. (2009)
Rubber	O'ring rubber	5	0.45	Chun et al. (2005)

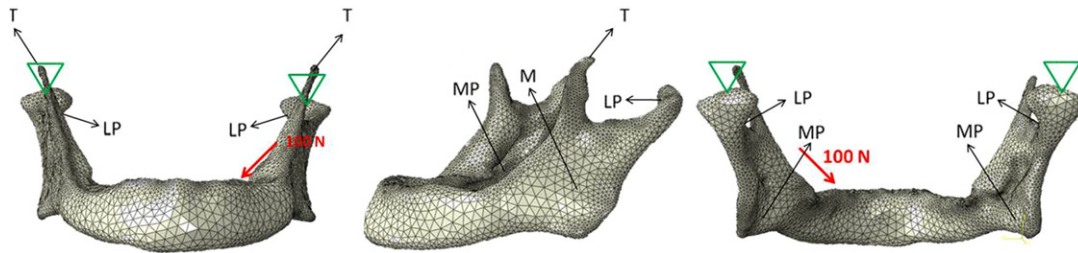


Fig. 4. Boundary conditions of the mandible FE model used in all groups based on Cruz et al. (2003, 2006, 2009). Black arrows indicate the direction of the applied masticatory muscle forces (T – temporal; LP – Lateral Pterygoid; MP – Medial Pterygoid and M – Masseter). Green triangles show the fixation of the temporomandibular joints. Red arrow shows the applied masticatory force on the left first molar region. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this article.)

Table 2
Directional cosines of the resultant muscular forces based on Cruz et al. (2003).

	Muscles							
	Masseter		Medial pterygoid		Lateral pterygoid		Temporal	
	Right	Left	Right	Left	Right	Left	Right	Left
Cos (α)	–0.043	0.043	0.587	–0.587	0.714	–0.714	–0.325	0.325
Cos (β)	–0.011	0.011	–0.165	0.165	–0.692	0.692	0.219	–0.219
Cos (γ)	0.999	–0.999	0.792	–0.792	0.106	–0.106	0.920	–0.920

principal stresses were also obtained to better understand the influence of different prosthesis designs on stress distribution in the periimplant bone tissue.

3. Results

The stress values (in MPa) within implant/prosthetic components and supporting tissues for all groups are displayed in Fig. 5. BC-C group exhibited the highest stress values which was located into the implant/prosthetic components ($\sigma_{VM}=398.8$, $\sigma_{max}=580.5$ and $\sigma_{min}=-455.2$) while FD group showed the lowest one ($\sigma_{VM}=128.9$, $\sigma_{max}=185.9$ and $\sigma_{min}=-172.1$). Comparing the overdenture groups, the use of unsplinted implants associated to the O'ring anchor reduced the stress level in the implant/prosthetic components (reduction of 59.4% for σ_{VM} , 66.2% for σ_{max} and 57.7% for σ_{min} versus BC-C group).

The highest stress levels were located under the applied loading; i.e. in the bar portion between the two left implants for splinted groups and in the distal ball abutment in the left side for O'ring group (Fig. 6a–d). In the overdenture groups, the σ_{VM} seems to be distributed to all implants while, in the FD group, the stress was mostly located at the implants and cantilever on the loaded side. On the implants, stress was mainly placed from the neck to the middle third region (Fig. 6a–d).

Concerning the supporting tissues, whereas the BC group displayed the greatest stress levels within cortical bone ($\sigma_{VM}=97.1$,

$\sigma_{max}=100.7$ and $\sigma_{min}=-93.5$), trabecular bone ($\sigma_{VM}=28.9$, $\sigma_{max}=24.5$ and $\sigma_{min}=-38.6$) and mucosa ($\sigma_{VM}=1.93$, $\sigma_{max}=4.15$ and $\sigma_{min}=-5.33$), the FD group showed the lowest stress levels (cortical bone – $\sigma_{VM}=51.0$, $\sigma_{max}=60.4$ and $\sigma_{min}=-51.9$; trabecular bone – $\sigma_{VM}=6.43$, $\sigma_{max}=4.21$ and $\sigma_{min}=-4.6$; mucosa – $\sigma_{VM}=0.8$, $\sigma_{max}=1.2$ and $\sigma_{min}=-0.7$), except for the σ_{max} on the cortical bone in which the O'ring group took place (28.1). In the overdenture group, the use of unsplinted implants decreased the stress concentration in all supporting tissues (maximum stress reduction of 72% and 79.5% for σ_{max} , and 15.7% and 85.7% for σ_{min} on the cortical and the trabecular bones, respectively) (Fig. 5).

Cortical bone exhibited greater stress concentration than trabecular bone (Fig. 5). Periimplant area exhibited the maximum stress concentration (Figs. 7 and 8). In the cortical bone, the highest tensile stress was observed on the buccal side of the periimplant area while the greatest compressive stress was noted on the lingual side (Fig. 7). In OR and FD groups, the stress on periimplant cortical bone was mainly limited to the loaded side while in the BC and BC-C groups, the stress was distributed to all periimplant cortical bone areas.

4. Discussion

Both research hypotheses were confirmed. Concerning the implant/prosthetic components, implant-retained overdenture

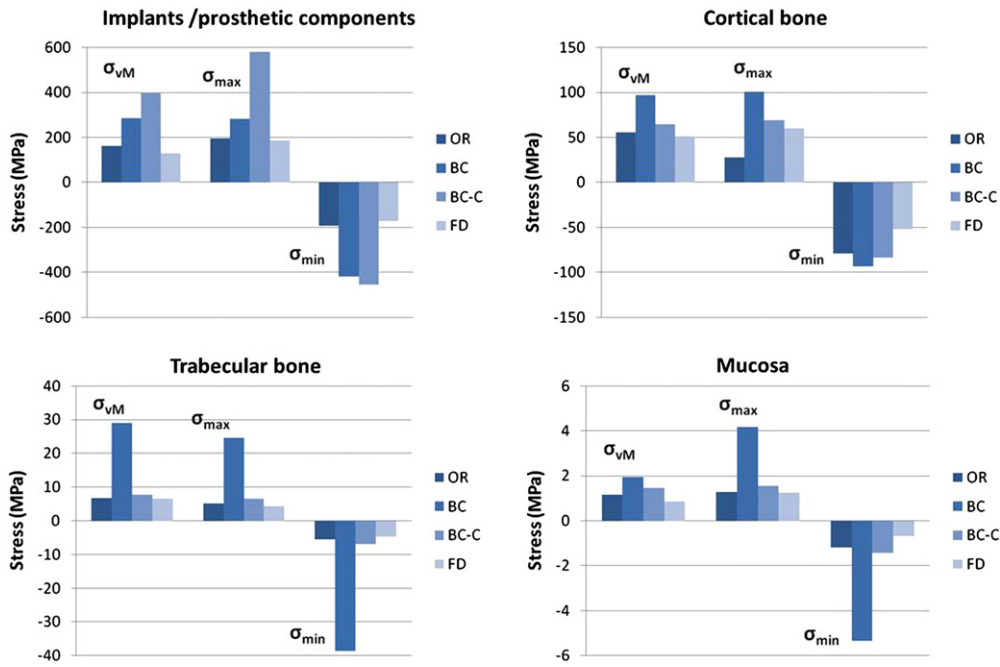


Fig. 5. von Mises stress (σ_{vM}), maximum (σ_{max}) and minimum (σ_{min}) principal stress values (in MPa) within implant/prosthetic components, cortical bone, trabecular bone and mucosa for all groups.

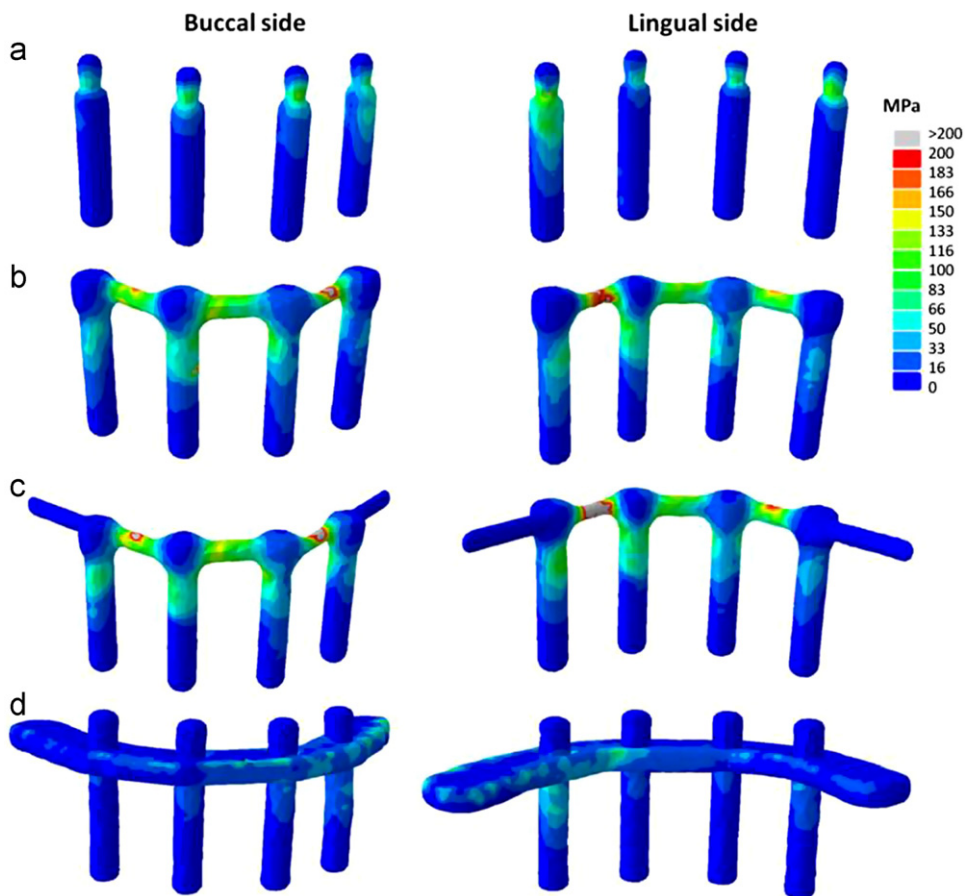


Fig. 6. von Mises stress (σ_{vM}) distribution (in MPa) within implant/prosthetic components for (a) OR, (b) BC, (c) BC-C and (d) FD groups.

associated with bar-clip anchor with two distally placed cantilevers displayed the greatest stress level. Although cantilever extensions of bar attachments have been recommended for mandibular

implant-retained overdentures to increase denture stability against non-axial loading (Mericske-Stern et al., 2000; Naert et al., 1997), higher rate of bar fracture associated with distal cantilevers than

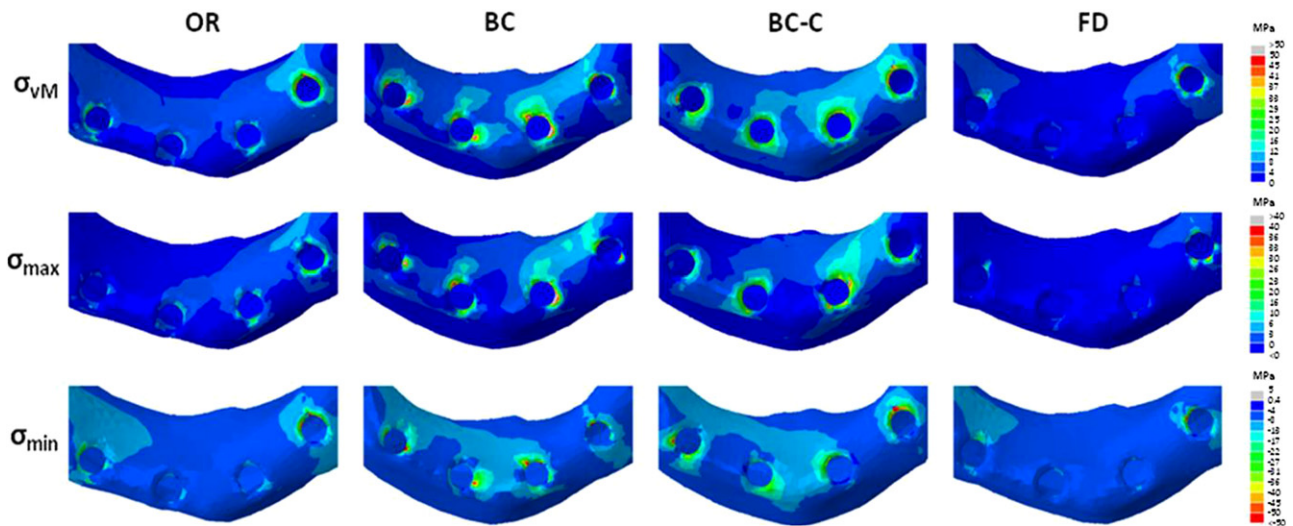


Fig. 7. von Mises stress (σ_{vM}), maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions (in MPa) within cortical bone for all groups.

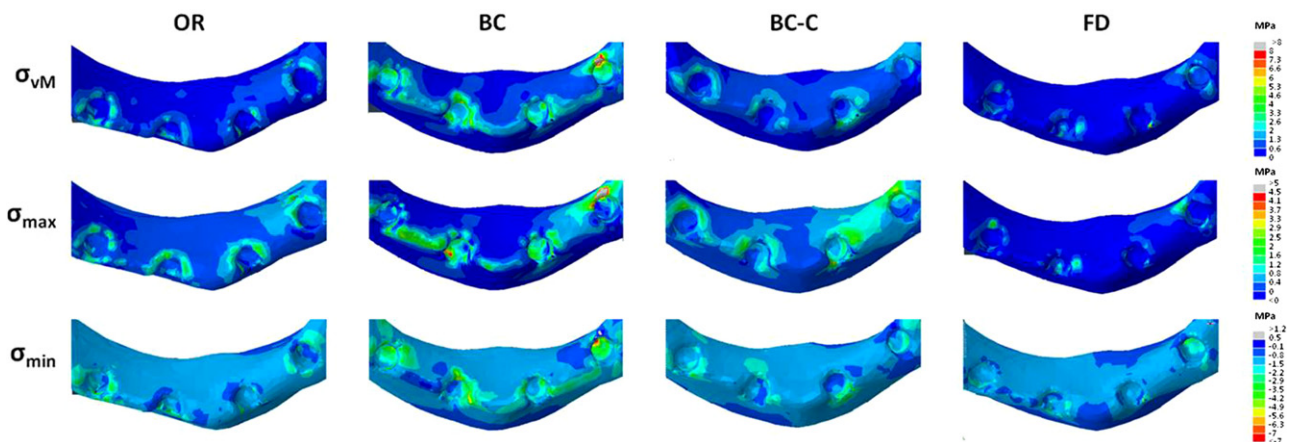


Fig. 8. von Mises stress (σ_{vM}), maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions (in MPa) within trabecular bone for all groups.

without cantilever was shown (den Dunnen et al., 1998). In the present study, it may be explained by the 29.7% increase in the σ_{vM} in the bar system associated with distal cantilevers than without cantilevers. Additionally, great load on the distal implants was observed in cantilevered implant-retained maxillary overdentures during in vivo masticatory cycles (Benzing et al., 1995). Even though the presence of cantilevers, the fixed full-arch implant-supported prosthesis exhibited the lowest stress levels on the superstructure, probably due to the bar size and shape creating a more stable and robust system. No study compared technical complication rates between removable and fixed dentures with bar systems; however, the current results presumably indicate less bar fracture rate to the fixed design. Gallucci et al. (2009b) showed that among 45 completely edentulous patients treated with implant-supported fixed mandibular prosthesis, 4.4% of patients ($n=2$) had fracture of metal framework after 5 months and 3.3 years of delivery. In relation to the bar design of overdentures, among 31 patients, 6.5% ($n=2$) exhibited bar fracture and 9.7% ($n=3$) exhibited other failures (i.e. loosening of the retention screw) after 3.5 years (Behr et al., 1998).

Within overdenture groups, the O'ring system reduced the stress level on the implant/prosthetic components. The flexibility and resiliency provided by the O'ring rubber may work as a stress-breaking system explaining these results (Barao et al., 2009; Tokuhisa et al., 2003). It is supported by clinical outcome

in which the survival rate of mandibular overdenture was higher for ball attachment (98.8%) versus bar (97.7%) after 10-year follow-up (Ferrigno et al., 2002). Nevertheless, the postinsertion maintenance of bar-clip attachment in mandibular overdenture was lower than for ball attachment (Stoker et al., 2007).

Regarding the supporting tissues, the BC group promoted the highest stress levels whereas the FD group the lowest one. In the FD group, the denture is completely supported by implants with no mucosa contact. Additionally, the strong framework increases the system stability which decreases the stress mainly on the periimplant area. Similarly, clinical data showed higher cumulative success rate of implants (96.2%) for full-arch implant-supported mandibular restoration than for implant-retained mandibular overdentures (93.7% and 93.9% for ball and bar anchors, respectively) (Ferrigno et al., 2002).

In the overdenture groups, the use of unsplinted implant optimized the stress level on the supporting tissues when compared to splinted implants, similarly to several studies (Barao et al., 2009; Celik and Uludag, 2007; Menicucci et al., 1998; Tokuhisa et al., 2003). Solitary anchors allow mandible's flexure and the O'ring rubber promoted a stress-breaking effect onto the bone (Barao et al., 2009). In vivo study showed higher implant loss for overdenture retained by two splinted implants (6.5%) than unsplinted implants (5.2%).

The cantilever on the bar-clip system decreased the stress level on the supporting tissues when compared with the bar-clip without cantilever. It may be explained by higher stress

concentration in the implant/prosthetic components of the BC-C group away from the periimplant bone and reduction of the overdenture contact in the denture bearing area, which relieves the supporting tissues. Therefore, distal extensions may provide higher stability against lateral forces and seclude the denture bearing area from loading forces (Mericske-Stern, 2003). Nevertheless, Sadowsky and Caputo (2000) advocated that cantilevered bar overdenture exhibited higher stress level to the distal ipsilateral implants than non-cantilevered bar overdenture. However, the overdenture cantilever was 7-mm long in the current study while it was 11.5-mm long in Sadowsky and Caputo study (2000).

Within cortical bone, peak of stress was observed on the periimplant region. When the implant is loaded, the stress is transferred to its first material contact (i.e. periimplant cortical bone), which explains the clinical marginal bone loss around implants (Isidor, 2006). Correlation between high occlusal stress onto the implants and marginal bone loss has been showed (Naert et al., 2001). The higher stress values observed in cortical bone may rely on its higher elastic modulus when compared with trabecular bone (Yokoyama et al., 2005).

The stress distribution in the cortical bone surrounding the implants differed between the removable and fixed groups. In the former, the stress spread to the ipsilateral implants whereas it was concentrated at the loaded side in the fixed group. This may be attributed to the denture bearing area contact of the overdenture group and the greater stability of the fixed group.

Herein the applied load had 100 N of amplitude as also used previously (Akca et al., 2007; Cehreli et al., 2005; Cruz et al., 2003; 2006; 2009; Teixeira et al. 2010). However, variations in load application have been observed (Bonnet et al., 2009; de Almeida et al., 2010; 2011; Fazi et al. 2011; Naini et al., 2011). Depending on the oral cavity area and patient's characteristics (ethnic components, feeding habits), the bite force can vary considerably (Hagberg, 1987; Teixeira et al., 2010). It is noteworthy to highlight that this FE study is linear; therefore, the stress values will increase proportionally with the applied load (O'Mahony et al., 2001).

An asymmetric load was used to simulate a tendency of denture tilting as observed in Daas et al. (2008) study. Denture working side moves down under the applied load while its nonworking side moves up (Daas et al., 2008). This great displacement requests the attachment systems so that they are tested in a critical situation. In addition, this asymmetric load is clinically relevant during the early phase of mastication, in which the foodstuff is placed in one side of the denture (working side) and no occlusal contact occurs on the nonworking side. Although the position of the foodstuff can alter the denture bending plane, no qualitative variation would occur among the groups herein (Daas et al., 2008). Molar area was chosen for load application since it is considered the greatest loading force support (Bonnet et al., 2009). Moreover, oblique load instead of axial one is preferable to simulate clinical reality with greater stress levels (Tabata et al., 2011).

This study has several limitations and some assumptions were made. All materials were homogeneous, isotropic and linearly elastic in nature. However, it is known that the cortical bone is transversely isotropic (anisotropic) and heterogeneous (Cochran, 2000). The use of anisotropic properties instead of isotropic properties of the mandible bone in 3D -FE models increased up to 20–30% the stress levels at the implant–bone interface and periimplant crestal bone (O'Mahony et al., 2001). Same trend was shared by Bonnet et al. (2009) and Clelland et al. (1993). Despite the difference in stress values between models created with isotropic or anisotropic bone, the stress distribution (qualitative analysis) seems to be similar. Thus, only minor consequences would occur for the qualitative stress analysis if the models were assumed as anisotropic. Additionally, modeling bone with anisotropic property is complex, problematic and time-consuming in nature.

The condition of the bone–implant interface has a burly influence on the stress intensities over the peri-implant area (O'Mahony et al., 2001). The stress at the periimplant bone and implant apex are greater for a free contact when compared to a fixed bond (Weinans et al., 1993). Before osseointegration, the implant–bone interface presents a frictional component which allows minor displacement. In this situation, the contact interface transfers pressure and frictional/tangential forces, with no tension (Ding et al., 2009). Herein, the implant–bone interface was assumed to be glued in order to simulate the bone full maturation and a 100% of osseointegration condition of the dental implant. This type of interface provides the continuity of displacement and traction vectors (Bonnet et al., 2009). In-situ bone–implant contact percentage varies from 30% to 70% (Schrotenboer et al., 2008). Indeed, the elastic modulus of the bone tissue would be lower and more isotropic at an earlier phase of bone–implant interface development (O'Mahony et al., 2001). For this reason, inferior stress and strain in the interface would be expected (Huja et al., 1998). Baldassarri et al. (2012) reported an enhancement in elastic modulus and hardness of human cortical bone around retrieved plateau root form implants in the first 5 years of implantation and stable mechanical properties after this period. The simplification in the bone–implant contact may affect our results in somehow. The stiffness dissimilarity among groups (e.g. strong framework of the fixed group versus thin (BC, BC-C) or missing crossbars (OR)), may yield different stress distribution under a slightly increased movement of the implant superstructures. In this sense, neighboring areas may exhibit higher load while the local stress magnitudes may reduce; additionally, increased deformations may promote greater local stress levels.

Herein, no thread was represented in the implants. Previous FEA studies (Barao et al., 2008, 2009; de Almeida et al., 2010; 2011; Fazi et al., 2011) did not model the implant threads owing to (i) the complexity in precisely determine the thread helix and (ii) the increased number in elements into the models which may be time-consuming to generate the stress results and, in some cases, make the resolution of the problem impossible to be solved (Assuncao et al., 2009). Similar stress distribution (qualitative analysis) for threaded and non-threaded implants in FE models has been showed, although differences on absolute stress values (quantitative analysis) were noted (Assuncao et al., 2009).

Additionally, temporomandibular joint was not precisely represented. The absence of articular disc may affect the overall stress values into the models. In order to validate such results, further FEA studies considering the anisotropic property of bone and coefficient of friction at implant/bone interface are warranted. Furthermore, animal and clinical studies are necessary to prove such biomechanical outcomes.

5. Conclusion

Based on the outcomes of the present study and within the limitations of the methodology used the following conclusions can be drawn

1. Cantilevered bar-clip overdenture (BC-C group) displayed the highest von Mises stress, maximum and minimum principal stresses values within implant/prosthetic components whereas the fixed full-arch implant-supported prosthesis (FD group) presented the lowest ones.
2. In the supporting tissues (cortical bone, trabecular bone and mucosa), bar-clip overdenture (BC group) showed the greatest stress values followed by cantilevered bar-clip overdenture (BC-C group), O'ring overdenture (OR group) and fixed full-arch implant-supported prosthesis (FD group).

3. Within overdenture groups, the use of unsplinted implants to retain the overdentures (OR group) reduced the von Mises stress, maximum and minimum principal stresses levels in both implant/prosthetic components and supporting tissues.
4. Within supporting tissues, periimplant cortical bone exhibited the highest stress values for all groups. In the overdenture groups, the stress was transferred to the ipsilateral periimplant cortical bone.

Conflict of interest statement

All authors declare no financial and personal conflict of interest in this study.

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