FEA in Dentistry: A Useful Tool to Investigate the Biomechanical Behavior of Implant Supported Prosthesis

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1. Introduction

The use of dental implants is widespread and has been successfully applied to replace missing teeth (Amoroso et al., 2006). Although high success rate has been reported by several clinical studies, early or late dental implants failures are still inevitable. During mastication, overstress around dental implants may cause bone resorption, which leads to infection on the peri-implant region and failure of oral rehabilitation (Kopp, 1990). The way in which bone is loaded may influence its response (Koca et al., 2005). The results of cyclic loading into the bone differ from those of static loading (Papavasiliou et al., 1996). In case of repetitive cyclic load application, stress microfractures in bone may occur (Koca et al., 2005) and may induce osteoclastic activity to remove the damaged bone (Papavasiliou et al., 1996). So far, it is imperative to understand where the maximum stresses occur during mastication around the implants in order to avoid these complications (Nagasao et al., 2003).

Considering that stress/strain distribution at bone level is hard to be clinically assessed, the finite element analysis (FEA) has been extensively used in Dentistry to understand the biomechanical behavior of implant-supported prosthesis. To date, FEA was first used in the Implant Dentistry field by Weinstein et al. (1976) to evaluate the stress distribution of porous rooted dental implants. Nowadays, owing to the geometric complexity of implant-bone-prosthesis system, FEA has been viewed as a suitable tool for analyzing stress distribution into this system and to predict its performance clinically. Such analysis has the advantage of allowing several conditions to be changed easily and allows measurement of stress distribution around implants at optional points that are difficult to be clinically examined.

Therefore, this chapter provides the current status of using FEA to investigate the biomechanical behavior of implant-supported prosthesis. The modeling of complex structures that represents the oral cavity is described, and comparisons between two-dimensional (2D) and three-dimensional (3D) modeling techniques are discussed. Additionally, the application of microcomputer tomography to develop complex and more realistic FE models are assessed. Some sensitive cases are also illustrated.

2. Biomechanical behavior of implant-supported prosthesis

In order to enhance treatment longevity, it is important to understand the biomechanics of implant-supported prosthesis during masticatory loading. And the way that the stress/strain is transmitted and distributed to the bone tissue dictates whether the implant treatments will failure or succeed (Geng et al., 2001). Several variables affect the stress/strain distribution on the implant/bone complex such as prosthesis type, implant type, veneering and framework materials, bone quality, and presence of misfit.

2.1 Prosthesis and implant types

The implant-supported prosthesis can be classified as single- or multi- unit prosthesis. From a biomechanical point of view, the multi-unit prosthesis is subdivided into implant-supported overdentures and implant-supported fixed prosthesis (cantilevered design or not). The nature of FEA studies for these prosthesis designs is much more complex than for single-unit design (Geng et al., 2001).

Implant-retained overdentures are considered a simple, cost-effective, viable, less invasive and successful treatment option for edentulous patients (Assuncao et al., 2008; Barao et al., 2009). However, controversies toward the design of attachment systems for overdentures still exist (Bilhan et al., 2011). Our previous study (Barao et al., 2009) used a 2D FEA to investigate the effect of different designs of attachment systems on the stress distribution of implant-retained mandibular overdentures. The bar-clip attachment system showed the greatest stress values followed by bar-clip associated with two distally placed o'ring attachment systems, and o'ring attachment system (Fig. 1). Other 2D (Meijer et al., 1992) and 3D FEA studies (Menicucci et al., 1998) also showed stress optimization in overdenture with unsplinted implants (e.g. o'ring attachment system). The flexibility and resiliency provided by the o'ring rubber and the spacer in the o'ring system assembly may be the driven force toward the lower stress values with o'ring attachment system. Additionally, the stress breaking effect of the o'ring rubber can also decrease the stress in implants, prosthetics components and supporting tissues (Tokuhisa et al., 2003).

Tanino et al. (2007) evaluated the effect of stress-breaking attachments at the connections between maxillary palateless overdentures and implants using 3D models with two and four implants. Stress-breaking materials (with elastic modulus ranging from 1 to 3,000 MPa) connecting the implants and denture were included around each abutment. As the elastic modulus of the stress-breaking materials increased, the stress increased at the implant-bone interface and decreased at the cortical bone surface. Additionally, the 3-mm-thick stress-breaking material decreased the stress values at the implant-bone interface when compared to the 1-mm-thick material. Knowing that overdentures are retained by implants but are still supported by the mucosa, and facing the difference in displacement between implants (20-30 μm) and soft tissue (about 500 μm), our previous study (Barao et al., 2008) investigated the influence of different mucosa thickness and resiliency on stress distribution of implant-retained overdentures using a 2D FEA. Two models were designed: two-splinted-implants connected with bar-clip system and two-splinted-implants connected with bar-clip system associated with two-distally placed o'ring system. For each design, mucosa assumed three characteristics of thickness (1, 3 and 5 mm) varying its resiliencies (based on its Young's modulus) in hard (680 MPa), resilient (340 MPa) and soft (1 MPa), respectively. In general, the stress decreased at the supporting tissues as mucosa thickness and resiliency increased (Fig. 2).

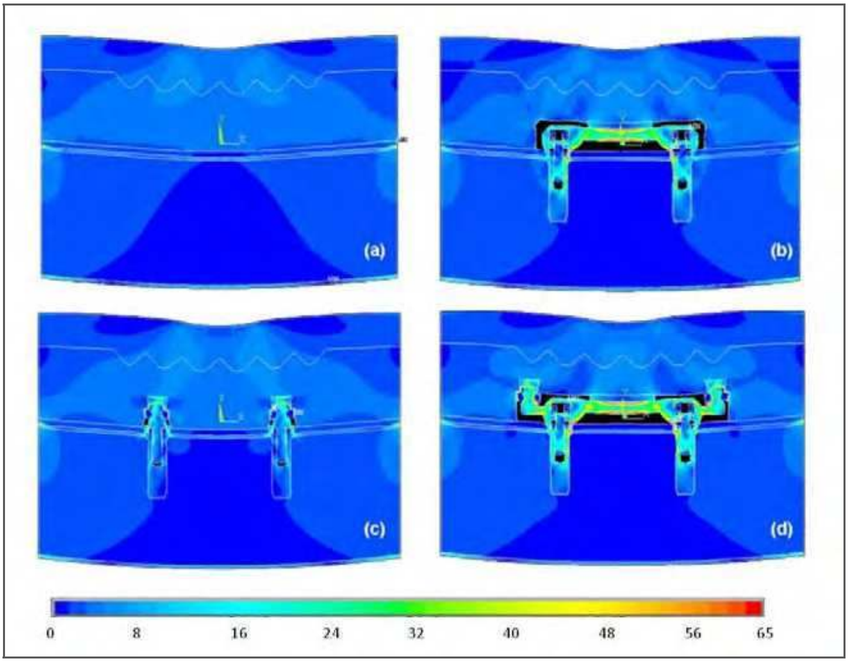

Fig. 1. First principal stress distribution (in MPa). (a) conventional complete denture. (b) overdenture – bar-clip system. (c) overdenture – o'ring system). (d) overdenture – bar-clip associated with distally placed o'ring system. Colors indicate level of stress from dark blue (lowest) to red (highest).

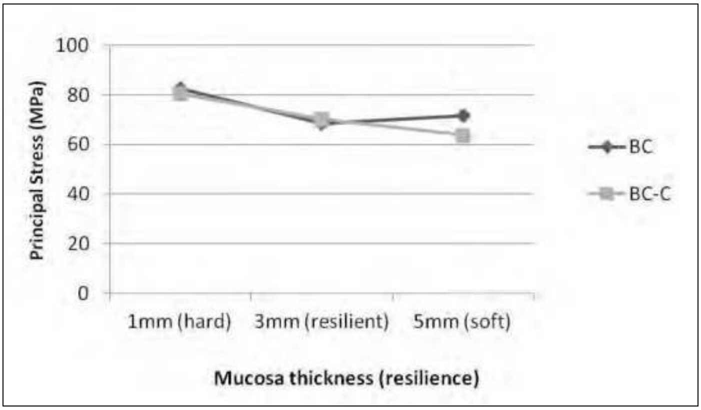

Fig. 2. Distribution of first principal stress (MPa) in supporting tissues for groups BC (bar-clip) and BC-C (bar-clip associated with two-distally placed o'rings) considering different mucosa thickness (1, 3 and 5mm) and resilience (hard, resilient and soft).

In relation to the implant-supported fixed prosthesis, the variety of factors that affect the stress distribution into the bone-implant complex comprise implant inclination, implant number and position, framework/veneering material properties, and cross-sectional design of the framework (Geng et al., 2001). The use of tilted implants mostly affected the stress concentration in the peri-implant bone tissue when compared to vertical implants (Canay et al., 1996). However, tilted implants have been used in case of atrophic jaw, to avoid maxillary sinus, and to reduce the cantilever extension (Silva et al., 2010). Caglar et al. (2006) investigated the effects of mesiodistal inclination of implants on the stress distribution of posterior maxillary implant-supported fixed prosthesis using a 3D FEA. Inclination of the implant in the molar region resulted in increased stress. Similar results were found by a Iplikcioglu & Akca (2002) who investigated the effect of buccolingual inclination in implant-supported fixed prosthesis applied to the posterior mandibular region using a 3D FEA. Bevilacqua et al. (2011) investigated the influence of cantilever length (13, 9, 5 and 0 mm) and implant inclination (0, 15, 30 and 45 degrees) on stress distribution in maxillary fixed dentures. This 3D FEA study showed that tilted implants, with consequent reduction of the posterior cantilevers, reduced the stress values in the peri-implant cortical bone.

Zarone et al. (2003) evaluated the relative deformations and stress distributions in six different designs of full-arch implant-supported fixed mandibular denture (six or four implants, cantilevered designed or not, cross-arch or midline-divided bar into two free-standing bridges) by means of 3D FEA. When the implants were rigidly connected by one-piece framework, the free bending of the mandible was hindered. The flexibility of the mandible was increased as the more distal implant supports were more mesially located. The use of two free-standing bars also reduced the overall stress on the bone/implant interface, fixtures and superstructure. Contradicting these findings, Yokoyama et al. (2005) observed that the use of single-unit superstructure was more effective in relining stress concentration in the edentulous mandibular bone than 3-unit superstructure. Other study (Silva et al., 2010), using a 3D FEA, assessed the biomechanical behavior of the “All-on-four” system with that of six-implant-supported maxillary prosthesis with tilted implants. The stress values were greater to the “All-on-four” concept, and the presence of cantilever increased the stress values about 100% in both models.

It is believed that loading distribution pattern in implant-retained overdentures differs from those in implant-supported fixed restorations (Tokuhisa et al., 2003). Our ongoing project has compared 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-FEA based on a computerized tomography (CT). Four 3D FE models of an edentulous human mandible with mucosa and four implants placed in the interforamina area were constructed and restored with different designs of dentures. In the OR group, the mandible was restored with an overdenture retained by four unsplinted implants with O’ring attachment; in the BC-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 group, the mandible was restored with a fixed full-arch four-implant-supported prosthesis. The masticatory muscles and temporomandibular joints supported the models. A 100-N oblique load (30 degrees) was applied on the left first molar of each denture in a buccolingual direction. Qualitative and quantitative analysis based on the von Mises stress (σ_{vM}), the maximum (σ_{max}) (tensile) and minimum (σ_{min}) (compressive) principal stresses (in MPa) were obtained. BC-C group exhibited the highest stress values (σ_{vM} = 398.8, σ_{max} = 580.5 and σ_{min} = -455.2) while FD group showed the lowest one (σ_{vM} = 128.9, σ_{max} = 185.9 and σ_{min} = -172.1) in the implant/prosthetic components. Within overdenture groups, the use of unsplinted implants (OR group) 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 bone and the trabecular bone, respectively). The cortical bone exhibited greater stress concentration than the trabecular bone for all groups. We concluded that the use of fixed implant dentures and removable dentures retained by unsplinted implants to rehabilitate completely edentulous mandible reduced the stresses in the peri-implant cortical bone tissue (Fig. 3), mucosa and implant/prosthetic components.

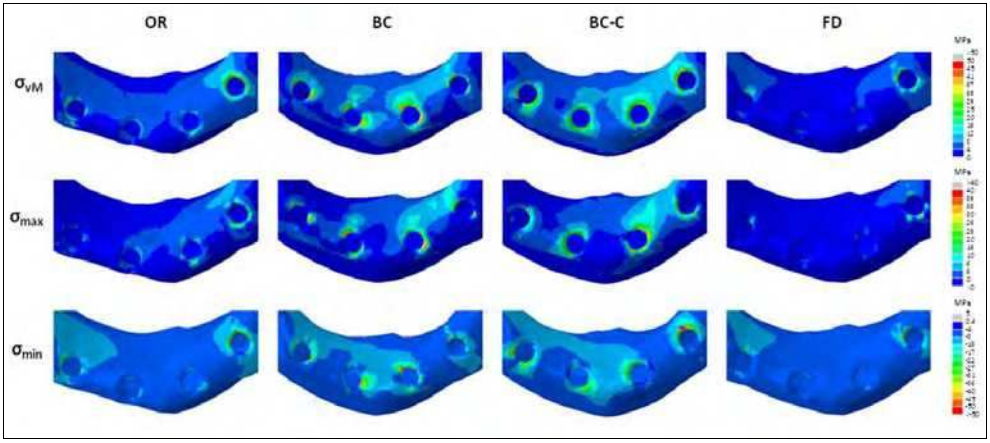

Fig. 3. von Mises stress (σ_{vM}), maximum (σ_{max}) and minimum (σ_{min}) principal stress distributions (in MPa) within cortical bone for o'ring (OR), bar-clip (BC), bar-clip with distally placed cantilever (BC-C) and fixed denture (FD) groups.

Concerning the implant design, Ding et al. (2009) analyzed the stress distribution around immediately loaded implants of different diameters (3.3; 4., and 4.8 mm) using an accurate complete mandible model. The authors observed that with the increase of implant diameter, stress/strain on the implant-bone interface decreased, mainly when the diameter increased from 3.3 to 4.1 mm for both axial and oblique loading conditions. Other studies also showed more favorable stress distribution with the use of wide-diameter implants (Himmlova et al., 2004; Matsushita et al., 1990). Huang et al. (2008) analyzed the peri-implant bone stress and the implant-bone sliding as affected by different implant designs and implant sizes of immediately loaded implant with maxillary sinus augmentation. Twenty-four 3D FE models with four implant designs (cylindrical, threaded, stepped and step-thread implants) and three dimensions (standard, long and wide threaded implants) with a bonded and three levels of frictional contact of implant-bone interfaces were analyzed. The use of threaded implants decreased the bone stress and sliding distance about 30% as compared with nonthreaded (cylindrical and stepped) implants. With the increase of implant's length or diameter, the bone stress reduced around 13-26%. The immediately loaded implant with smooth machine surface increased the bone stress by 28-63% versus osseointegrated implants. The increase of implant's surface roughness did not reduce the bone stress but decrease the implant-bone interfacial sliding.

2.2 Veneering and framework material

The literature is scarce about the best material to fabricate superstructures of implant-supported prosthesis (Gomes et al., 2011). Originally, the protocol consisted of gold alloy framework and acrylic resin for denture base and acrylic resin or composite resin for artificial denture teeth (Zarb & Jansson, 1985). Rigid occlusal material such as porcelain on metal may increase the load transfer to the implant and surrounding bone tissue (Skalak, 1983). So far, the use of occlusal veneering based on resin material is indicated to absorb shock and consequently to reduce the stress on the implant-bone complex (Skalak, 1983). Gracis et al. (1991) stated that the use of harder and stiffer materials to fabricated implant-supported restorations increased the stress transmitted to the implant. On the other hand, some studies (Ciftci & Canay, 2001; Sertgoz, 1997) showed that the use of softer restorative materials lead to a higher stress on implants and supporting tissues.

Our previous studies (Delben et al., 2011; Gomes et al., 2011) evaluated the influence of different superstructures on preload maintenance of retention screw of single implant-supported crowns submitted to mechanical cycling and stress distribution through 3D FEA.

Twelve replicas for each group and 3D FEA models were created to simulate a single crown supported by external hexagon implant in premolar region. Five groups were obtained: gold abutment veneered with ceramic (GC) and resin (GR), titanium abutment veneered with ceramic (TC) and resin (TR), and zirconia abutment veneered with ceramic (ZC). During mechanical cycling, the replicas were submitted to dynamic vertical loading of 50 N at 2 Hz for detorque measurement after each period of $1x10^5$ cycles up to $1x10^6$ cycles. The FEA software generated the stress maps after vertical loading of 100 N on the contact points of the crowns. Significant difference (P<.05) between group TC (21.4 ± 1.78) and groups GC (23.9 ± 0,91), GR (24.1 ± 1.34) and TR (23.2 ± 1.33); and between group ZC (21,9 ± 2,68) and groups GC and GR for initial detorque mean (in N.cm) was noted. After mechanical cycling, there was significant difference (P<.05) between groups GR (23.8 ± 1.56) and TC (22.1 ± 1.86), and between group ZC (21.7 ± 2.02) and groups GR and TR (23.6 ± 1.30) (Fig. 4). The stress values and distribution in bone tissue were similar for groups GC, GR, TC and ZC (1574.3 MPa, 1574.3 MPa, 1574.3 MPa and 1574.2 MPa, respectively), except for group TR (1838.3 MPa) (Fig. 5). Group ZC transferred lower stress to the retention screw (785 MPa) than the other groups (939 MPa for GC, 961 MPa for GR, 1010 MPa for TC, and 1037 MPa for TR) We concluded that detorque reduction occurred for all superstructure materials but torque maintenance was enough to maintain joint stability in this study. The different materials did not affect stress distribution in bone. However, group ZC presented the best stress distribution for the retention screw. Previous study conducted by our research group also found similar stress distribution to single implant-supported prosthesis regardless of the type of veneering/framework material through a 2D FEA (Assuncao et al., 2010).

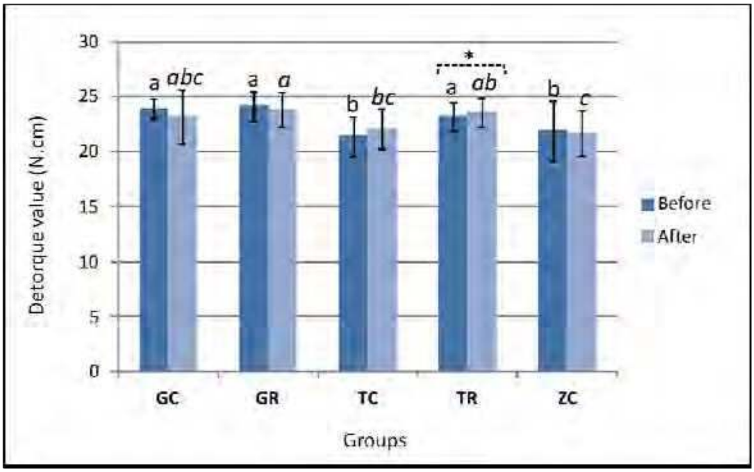

Fig. 4. Detorque mean value (N.cm) before and after mechanical cycling for all groups. Within each period, mean followed by different letters represent statistically significant difference (P<.05, Fisher's exact test).*denotes statistically significant difference within the same group (P<.05, Student's t-test).

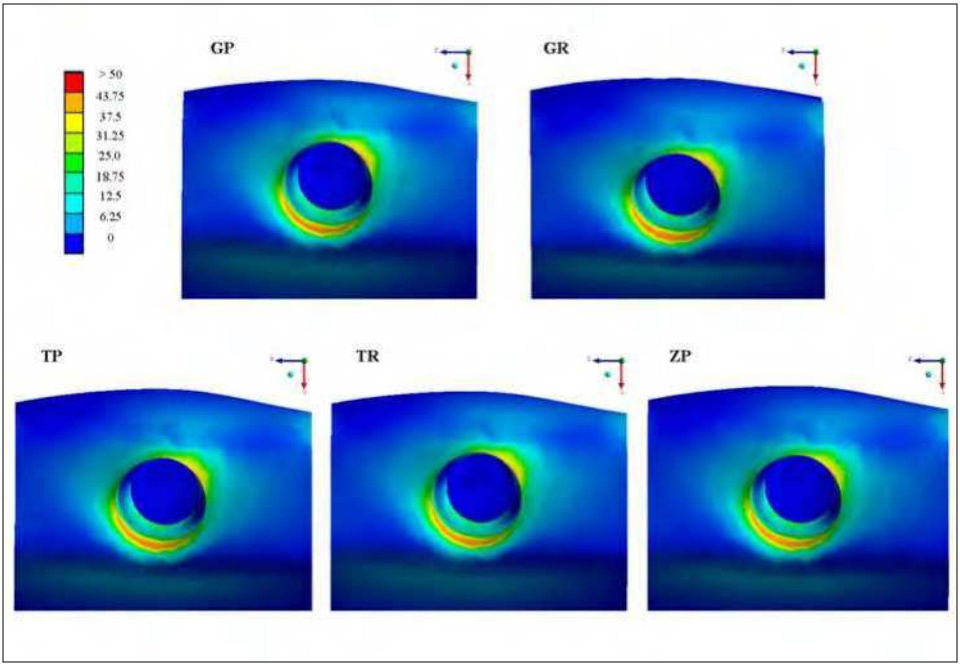

Fig. 5. von Mises stress distribution (MPa) within supporting bone for GP, GR, TP, TR and ZP groups. Colors indicate level of stress from dark blue (lowest) to red (highest).

2.3 Bone quality

The bone quality is strongly correlated with the implant success as pointed out by several longitudinal clinical studies (Friberg et al., 1991; Jemt & Lekholm, 1995; van Steenberghe et al., 1990). The biomechanical behavior among the different types of bone (I, II, III or IV) differs substantially, which affect the ability of bone to support physiological loads (de Almeida et al., 2010). The poor quality bone type 4 has promoted greater failures of dental implants (Jaffin et al., 2004) owing to its reduced capability to bond the implant to the bone (Drage et al., 2007; Shapurian et al., 2006).

de Almeida et al. (2010) investigated the influence of different types of bone (types I to IV) on the stress distribution on the supporting tissue of a fixed full-arch implant-supported mandibular prosthesis based on a prefabricated bar by using a 3D FEA. Three unilateral posterior loads of 150 N were applied on the prosthesis: L1 – axial loading; L2 – oblique loading (buccolingual direction, 30 degrees); L3 – oblique loading (linguobuccal direction, 30 degrees). Type III and IV bones displayed the greatest stress values in the axial and buccolingual loading conditions, while stiffer bones (type I and II) exhibited the lowest. For the linguobuccal loading condition, the poorest quality cortical bone (type IV) had the highest stress concentration followed by types III, II and I.

Tada et al. (2003) evaluated whether bone quality affect the stress/strain distribution of single-unit implant-supported mandibular prosthesis with different implant type and length. Screw and cylinder implants with 9.2, 10.8, 12.4 and 14.0 mm length were used and virtually placed in 4 types of bone. Two different loads (axial and buccolingual forces) were applied to the occlusal surface at the center of the abutment. As the bone density decreased, the stress/strain into the bone increased. Under axial loading, the stress in the cancellous bone was lower with the screw-type implant when compared with cylinder-type implant. Additionally, longer implants displayed lower stress values. The bone quality also influences the stress distribution under buccolingual load. According to the authors, low-density bone presents reduced stiffness, which increases implant displacement. Under greater displacement, the bone is deformed and consequently higher stresses in the cortical and cancellous bone are expected.

Another 3D FEA study (Sevimay et al., 2005) examined the effect of the bone quality on stress distribution for an implant-supported mandibular crown. A 3D FE model of a mandibular section of bone with a missing second premolar and an implant to receive a crown restoration was used. A total vertical force of 300 N was applied from the buccal cusp (150 N) and distal fossa (150 N) in centric occlusion. Low-density bone (types III and IV) displayed the greatest stress values (163 and 180 MPa, respectively) mainly at the peri-implant cortical bone. On the other hand, type I and II bones exhibited the lowest levels of stresses (150 and 152 MPa, respectively). Other similar study (Holmes & Loftus, 1997) found that the placement of implants in type I bone resulted in less micromotion and reduced stress concentration.

2.4 Presence of misfit

An increase of clinical failures has been correlated with misfit of implant-supported prostheses (Klineberg & Murray, 1985; Skalak, 1983). In order to prevent mechanical (i.e. retention screw and abutment screw loosening and fracture, superstructure mobility, and implant fracture) (Carlson & Carlsson, 1994; Dellinges & Tebrock, 1993) and biological (i.e. sensorial disturbances, soft tissue injuries, peri-implantitis, and bone loss) (Berglundh et al., 2002) complications of the implant treatment, a passive fit between the crown and implant should be achieved (Sahin & Cehreli, 2001). Previous studies (Assuncao et al., 2011; Kunavisarut et al., 2002; Natali et al., 2006) have showed an increase of stress on peri-implant bone tissue under the presence of misfit in implant-supported prostheses.

Our previous study (Assuncao et al., 2011) used a 3D FEA to investigate the effect of vertical and angular misfit in three-piece implant-supported screwed crown on the biomechanical behavior of peri-implant bone, implants, and prosthetic components. A total of four 3D models were fabricated to represent a posterior mandibular section with one implant in the region of the second premolar and another in the region of the second molar. The implants were splinted by a three-piece implant-supported metal-ceramic prosthesis and differed according to the type of misfit, as represented by four different models: control - prosthesis with complete fit to the implants; unilateral angular misfit - prosthesis presenting unilateral angular misfit of 100 µm in the mesial region of the second molar; unilateral vertical misfit - prosthesis presenting unilateral vertical misfit of 100 µm in the mesial region of the second molar; and total vertical misfit - prosthesis presenting total vertical misfit of 100 µm in the platform of the framework in the second molar (Fig. 6). A vertical load of 400 N was distributed and applied on 12 centric points (a vertical load of 150 N was applied to each molar in the prosthesis and a vertical load of 100 N was applied at the second premolar). We observed that stress on the peri-implant cortical bone was slightly affected by the presence of misfit. Each type of misfit overloaded a specific region of the implant-supported system. The unilateral angular misfit was most harmful for the implant body and retention screw, the unilateral vertical misfit placed the most stress on the framework, and the total vertical misfit added stress to the implant hexagon.

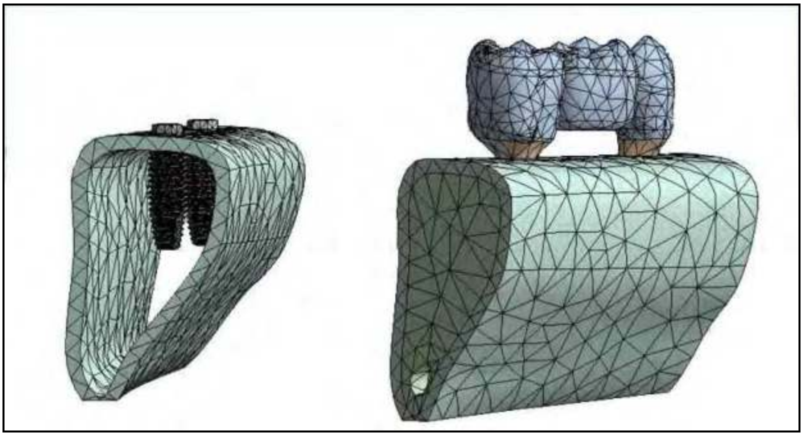

Fig. 6. Mesh of the main model: cortical bone, trabecular bone, implant and crown (framework, veneering material and retention screw).

Another study conducted by our group (Gomes et al., 2009) assessed the influence misfit on the displacement and stress distribution in the bone-implant-prosthesis complex using a 2D FEA. A single-unit mandibular implant-supported prosthesis was fabricated. Different unilateral angular misfit (0 µm – control, 50 µm, 100 µm, and 200 µm) was represented on the contact region between the implant and the crown. An oblique (30 degrees) load of 133 N was applied at the opposite direction of misfit on the models. The greater the angular misfit, the higher the stress and displacement values in the bone-implant-prosthesis assembly. On the other hand, Spazzin et al. (2011) investigated the effect of different levels of vertical misfit (5, 25, 50, 100, 200 and 300 µm) between implant and bar framework in overdenture and showed that the presence of misfit did not influence the stress level at the peri-implant bone tissue, but stress on the prosthetic components (bar framework, retention screw) and implants increased with greater misfit levels.

Spazzin et al. (2011) also assessed the influence of horizontal misfit (10, 50, 100 and 200 µm) and bar framework material (gold alloy, silver-palladium alloy, commercially pure titanium and cobalt-chromium alloy) on the stress distribution in implant-retained mandibular overdenture associated with bar attachment system using a 3D FEA. The increase in horizontal misfit promoted an enhancement of stress levels in the inferior region of the bar, retention screw neck, cervical and medium third of the implant, and peri-implant cortical bone. The stiffer the bar material was, the greater the stress on the framework.

3. Modeling complex structures

As the oral cavity is very complex in nature, it is very hard to represent this structure with high accuracy by means FE models. For this reason, several simplifications are necessary to our reality. Most of the FEA studies in Dental field consider the materials as isotropic, homogeneous and linearly elastic. The modeling of biological tissues (e.g. bone) is a very difficult task because of their inherent heterogeneous and anisotropic character (Cowin & editor, 2001). The use of isotropic properties instead of anisotropic properties for bone tissue may affect the overall results of stress distribution (O'Mahony et al., 2001). In addition, the ultimate strain and Young modulus of bone under compression is different than those under tension (Geng et al., 2001). A previous study (Liao et al., 2008) investigated in what extend the anisotropic elastic properties affect the stress and strain distribution around implants under physiologic load in a complete mandible model based on CT. Models were loaded obliquely, and the principal stress and strain values in the peri-implant bone tissue were recorded. The authors observed an increase of up to 70% of stress/strain values for the anisotropic model versus isotropic model. O'Mahony et al. (2001) compared implant-bone interface stresses and peri-implant principal strains in anisotropic versus isotropic 3D models of the posterior mandible. Anisotropy increased by 20 to 30% the stress and strain in the cortical bone when compared with isotropic case. In the trabecular bone, the anisotropy enhanced by 3- to 4-fold the stress level versus isotropic condition. So far, anisotropy has significant effects on peri-implant stress and strain; therefore, careful consideration should be given to its use in biomechanical FE studies (Liao et al., 2008; O'Mahony et al., 2001).

The bone-implant contact in most of FEA studies is considered at a 100%; however, in the clinical scenario the implant-bone contact ranges from 30% to 70% (Geng et al., 2001). Additionally, the bone is not homogeneous and porosities are presented. Nowadays, it is possible to insert contact algorithms to simulate contacts (friction coefficients). In Dentistry, this factor is very important because it allows the representation of different degrees of osseointegration and the scenario of immediate loading. Huang et al. (2008) compared two conditions of implant-bone contact (bonded interface and non-bonded interface – friction coefficient of 0.3) on peri-implant bone stress distribution and implant-bone interfacial sliding of different implant designs and implant sizes. The use of friction coefficient to represent the immediate loading condition of the implants increased the bone stress by 28-63% when compared with the osseointegrated condition (bonded contact) for all implant designs and sizes. The interfacial sliding between bone and implant decreased with the presence of friction coefficient.

Thread configuration is an important objective in biomechanical optimization of dental implants (Valen & Locante, 2000). However, several 2D and 3D FEA studies have not considered the threads modeling of implants and retention screws or have modeled them using only concentric rings owing to the difficulty in modeling the thread helix (Sertgoz, 1997). Additionally, some justify the use of simplified model due to the difficulty in constructing a 3D complex model and the enormous increase in element numbers (Assuncao et al., 2009). As the oversimplication of implant complex geometry may affect the results of several FEA studies (Al-Sukhun et al., 2007), some authors (Lang et al., 2003; Sakaguchi & Borgersen, 1995) believe that the modeling of the perfect geometry of the implant, including the thread helix of the screw and the screw bore, is essential to finite element analysis simulation. Therefore, our previous study (Assuncao et al., 2009) investigated whether or not the representation of implant's threads would affect the outcome of a 2D FEA. Two models reproducing a frontal section of edentulous mandibular posterior bone were constructed. In the first models, the implants threads were accurately simulated (precise model) and, on the other implants with a smooth surface (press-fit implant) were used (simplified model). A load of 133 N was obliquely (30 degrees) applied with on the models. Precise model (1,45 MPa) showed higher maximum stress values than simplified model (1,2 MPa). Stress distribution and stress values in the cortical bone (292.95 MPa for precise model and 401.14 MPa for simplified model) and trabecular bone (19.35 MPa for precise model and 20.35 MPa for simplified) were similar, and the stresses were mostly located around implant neck and implant apex. We concluded that considering implant and screw analysis, remarkable differences in stress values were found between the models. Although models have showed differences on absolute stress values, the stress distribution was similar.

4. Two-dimensional (2D) versus three-dimensional (3D) analysis

The biomechanical performance of mandibular and maxillary bones associated to other natural (teeth, periodontal ligament, gingiva, etc.) and artificial (prostheses, dental implants, etc.) structures has been considered clinically and biologically relevant for modern Dentistry. In this sense, several tools have been developed for biomechanical evaluations, such as the finite element method.

The first studies reported in the literature using the Finite Element Method in Dentistry presented simple 2D models (Takahashi et al., 1978; Weinstein et al., 1979; Yettram et al., 1976). However, the complexity of the oral environment required the development of 3D models. Khera et al. (1988) were pioneers on 3D modeling for human mandible.

Although the technological advancement allowed the use of accurate and fast software and hardware to obtain the images, the decision about 2D or 3D modeling for FEA remains uncertain. It is important to understand that the biomechanical performance of complex oral structures depends on several factors. Therefore, the accuracy of the results may be influenced by: 1- complex geometry of the model; 2 – materials properties, such as isotropy or anisotropy; 3 – type, size and quantity of elements in the mesh; 4 – boundary and loading conditions similar to clinical scenario; and 5 – analysis mode (static or dynamic). In addition, the choice between 2D or 3D models should be guided by the expectances and applicability of the results. Thus, the researcher should understand the advantages and limitations of both modeling types (Romeed et al., 2006).

The 2D modeling has been continually applied in Dentistry since it is a simple, fast and low-cost approach. On the other hand, the 3D model is more accurate and may represent the details of a real condition (Fig. 7). However, a complex model is not worthy if it is misinterpreted considering that the higher the complexity of the model, the higher the density of the elements mesh. Thus, the following question should be answered before starting a FEA study "How can I represent the model accurately to obtain results within the ideal parameters?"

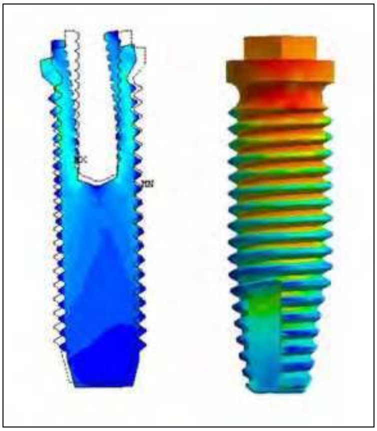

Fig. 7. Different types of FE implant models: Two-dimensional model and Three-dimensional model from left to right, respectively.

Several studies (Poiate et al., 2011; Romeed et al., 2006) were conducted to compare the different models applied from Engineering to Dentistry to verify the effect of 2D or 3D modeling on the analysis of the biomechanical performance of complex structures. Romeed et al. (2006) evaluated the mechanical behavior of a second maxillary premolar restored with full-coverage crown under different occlusal schemes and observed similar stress distribution and minor differences for the stress values. On the other hand, Poiate et al. (2011) compared the biomechanical performance of a maxillary central incisor between 2D and 3D modeling and concluded that 2D models can be safely applied only for qualitative analysis since these models showed overestimated values for the quantitative analysis of the stress. The differences between 2D and 3D models have been attributed to the geometric representation of the model. Although 2D models are simplified and easier to be obtained than 3D models, the biaxial state may influence the reliability of the results since some important biomechanical aspects may be not reproduced (Gao et al., 2006).

4.1 Case sensitive

Two- and 3- D models were constructed to evaluate the behavior of different veneering materials of single implant-supported prostheses (Assuncao et al., 2010; Gomes et al., 2011). For both models, five types of finite element (FE) models were simulated according to the different framework (gold alloy, titanium, and zirconia) and veneering (porcelain and modified composite resin) materials. However, several differences between the models representation were introduced according to the limitations of each modeling process. First of all, it was observed differences about the geometry, mainly for the superstructure (veneering material and framework) and bone tissue (Fig. 8). In the 2D model, the superstructure was modeled with 8-mm in height and 8-mm in diameter and the surrounding bone model assumed the characteristic of a block. In the 3D model, the superstructure was modeled based on the characteristics of a left first premolar tooth and the bone tissue reproduced a segment of the maxilla with a missing left first premolar tooth. In relation to the loading, the load was applied at a 30-degree inclination and 2-mm off-axis in 2D simulation while a 100-N vertical force was applied to the contact points of the crowns in the 3D model. Additionally, the 3D model represented a contact element between the abutment and the implant to simulate the clinical situation. At the end, it was observed difference between the 2D and 3D models for both qualitative and quantitative analyses.

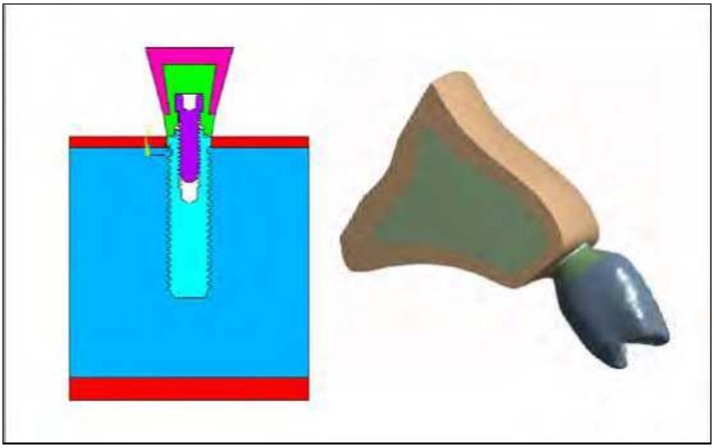

Fig. 8. Differences about the geometry representation of the superstructure and bone tissue for single implant-supported prostheses: 2D and 3D model from left to right, respectively.

Thus, the selection between 2D or 3D modeling should be guided by the researcher knowledge about both methods. The 3D models are similar to real structures but are time-consuming for modeling and data processing even when powerful computers are used. The higher the model complexity, the higher the number of elements and the complexity of the analysis. Thus, a simple geometry may generate an accurate mesh with satisfactory results in a faster way.

Two-dimensional models offer excellent access for pre- and post-processing, and because of the reduced dimensions, computational capacity can be preserved for improvements in element and simulation quality. On the other hand, 3D models, although more realistic with respect to the dimensional properties, are generally more coarse, with elements that are far from their ideal shapes. Moreover, examination of the model is more difficult. Depending on the investigated structure and boundary conditions, 2D modeling may be justified as a reasonable or even sensible simplification (Korioth & Versluis, 1997). Additionally, combinations of 2D or 3D FEA may offer the best understanding of the biomechanical behavior of complex dental structures in certain situations (Romeed et al., 2006).

5. Microcomputer tomography in FE models

In the beginning, the FE models were obtained from sectional images of bone tissue, tooth, surrounding structures and other elements related with the study that would be executed. Khera et al. (1988) constructed a 3D human mandible based on a 2D model. Initially, a 2D model was obtained and, using the projection of several pictures in a magnifying monitor, a 3D model was generated and an axial *z*-axis was defined.

Another widely used technique to obtain 2D and 3D models is the embedment of structures in acrylic resin. Gomes et al. (2009) embedded a prosthesis/retention screw/implant system in resin and sectioned it longitudinally to investigate the effect of different levels of unilateral angular misfit prostheses in the assembly and surrounding bone using a 2D FEA. The embedded model was scanned to produce digitalized images that were imported into CAD image analysis software and placed within the supporting tissue based on literature data. The outline of the images was manually quoted and each point was converted into x and y coordinates. At the end, the coordinates were finally imported into the FE software as key points of the final images (Fig. 9).

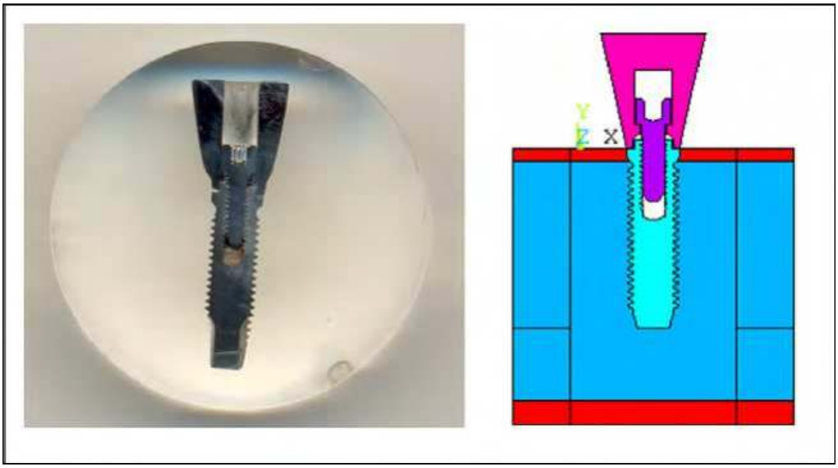

Fig. 9. Scanned embedded model and finite element model.

Recently, microcomputer tomography (CT) images have been obtained as a useful tool to model the bone complex in FE models and have gained general consensus among researches. The 3D model simulated from CT images provides high fidelity to the anatomical dimensions and configuration of all oral structures because it is possible to define the geometry and the local tissue properties of the bone segment to be modeled.

However, some caution should be taken when these data are used. The X-ray images in grey scale must be recorded by a CT in DICOM (Digital Imaging Communications in Medicine) format. Thus, the procedure should be carefully indicated for the patients due to the exposition to radiation and the research project should be approved by the ethics committee ensuring that physical and geometrical parameters within safety limits will be adopted. Furthermore, the patient should sign an informed consent form to authorize the reproduction of images and results.

The processing techniques used to extract this information from the CT data may be also frequently affected by no negligible errors that propagate in an unknown way through the various steps of the model generation, affecting the accuracy of the model (Taddei et al., 2006). The first source of geometric error and distortion is the resolution of the dataset that depends on the scan parameters setting (Taddei et al., 2006). The ones that yielded the best results for image quality were obtained in the regime of 120 kV, 150 mA, 512 × 512 matrix, 14 cm × 14 cm field of view, and slice thickness of 0.5 mm (Poiate et al., 2011). The second error may result from the segmentation process of the region of interest. Several segmentation algorithms have been proposed, with various level of automation, starting from complete manual contours extraction to complex fully automatic algorithms (Taddei et al., 2006). Considering that several softwares are currently available in the market, the professional should be trained to accurately use the tools to convert CT images into FE models.

5.1 Converting CT images into FEA models

Some steps to convert CT images into solid models will be presented in this section considering the availability of softwares, such as Mimics (Materialise, Leuven, Belgium), Simpleware (Simpleware Ltd, United Kingdom), InVesalius (Brazilian Public Software, Brazil), and ITK Snap (General Public License, USA).

The use of Simpleware software to convert a maxillary CT images into a maxillary FEA model will be discussed. After submission by the Ethic Committee in Research of Dental School of Ribeirão Preto, University of São Paulo (Process CAAE – 0038.0.138.000-11 and SISNEP – FR – 430209), the CT images obtained from patient imaging with 219 cross sections were imported into the Simpleware 4.1 software (Simpleware Ltd, United Kingdom) (Fig. 10) at the *ScanIP* segment.

In the *ScanIP*, the segmentation tool was used to identify the pixels value of the tomographic image. Then, it was possible to separate an object from other adjacent anatomical structures in different masks, such as cortical bone and trabecular bone (Fig. 11). According to its radio-density, expressed in Hounsfield unities, the program picked the values up and automatically created different masks. After that, fine adjustments were executed to further improve the quality of the model masks (Fig. 12). The program also provides tools to eliminate any interference of the tomographic image. By examining the image, a certain level of noise in the data could be corrected by filters. After obtaining the masks for the cortical and trabecular bone, the soft tissue was constructed with 2.0 mm in thickness for the whole model using a morphological filter tool in structuring element. Then, the step to convert the CT image into a solid model was completed (Figs. 13 and 14).

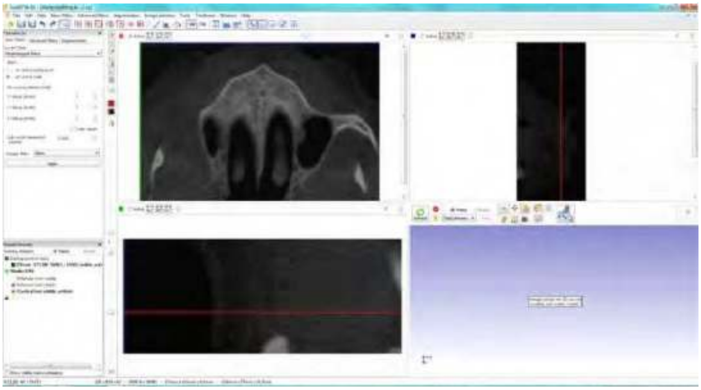

Fig. 10. Tomographic image of an edentulous maxilla imported into *ScanIP*.

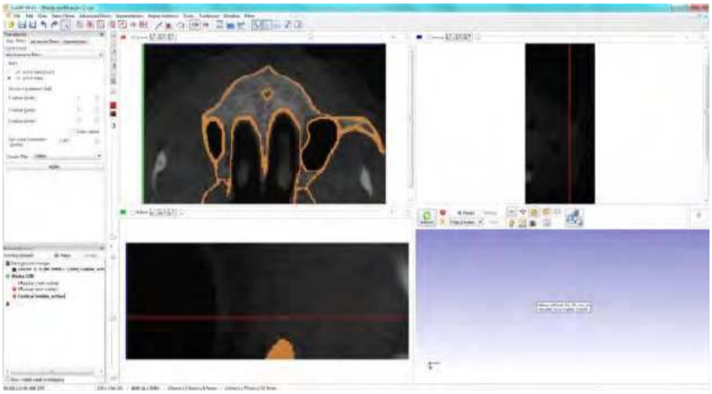

Fig. 11. Tomographic image of an edentulous maxilla processed in *ScanIP*. Determination of the cortical bone tissue.

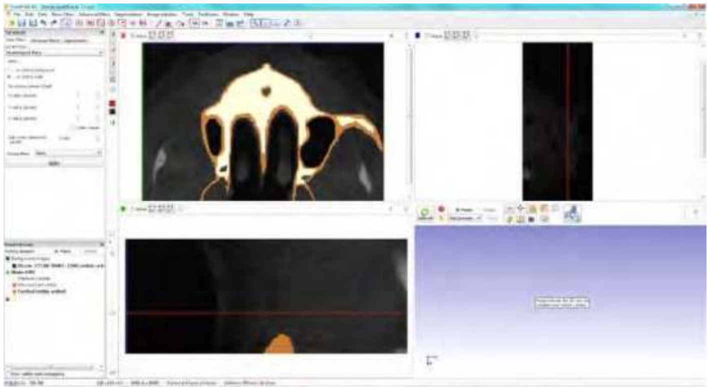

Fig. 12. Tomographic image of an edentulous maxilla processed in *ScanIP*. Determination of the trabecular bone tissue.

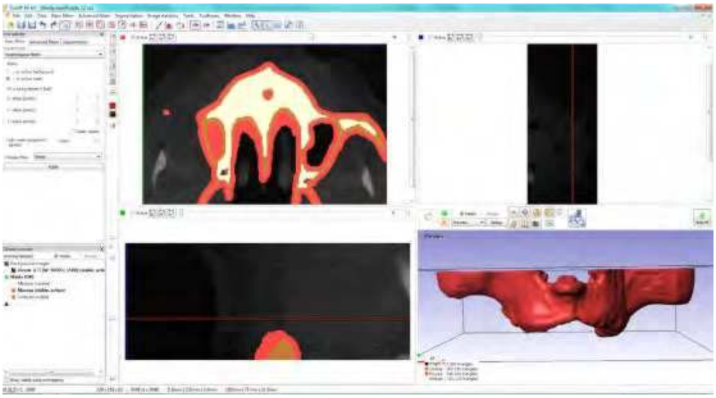

Fig. 13. Tomographic image of an edentulous maxilla processed in *ScanIP*. Determination of the soft tissue with 2.0mm in thickness in the whole extension.

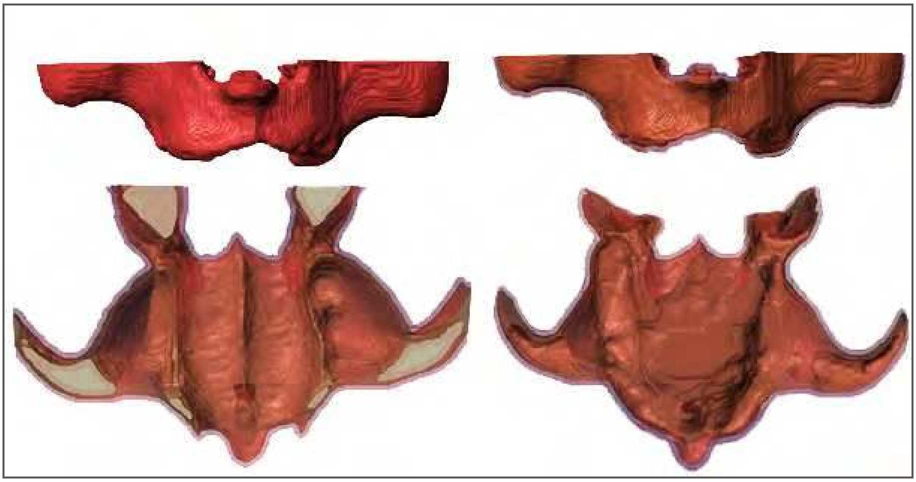

Fig. 14. Edentulous maxilla generated in the *ScanIP* representing the cortical and medullary bone and the soft tissue.

After obtaining the solid model, the finite elements mesh was generated. The mesh can be created either in the software for image conversion or in the FE software. In this study, the Simpleware software generated the mesh. A mix of tetrahedral and hexahedral elements was obtained using gallery elements + FE Free (Fig. 15). Afterwards, the meshed model is ready to be exported to a FEA software in order to conduct stress and displacement analysis.

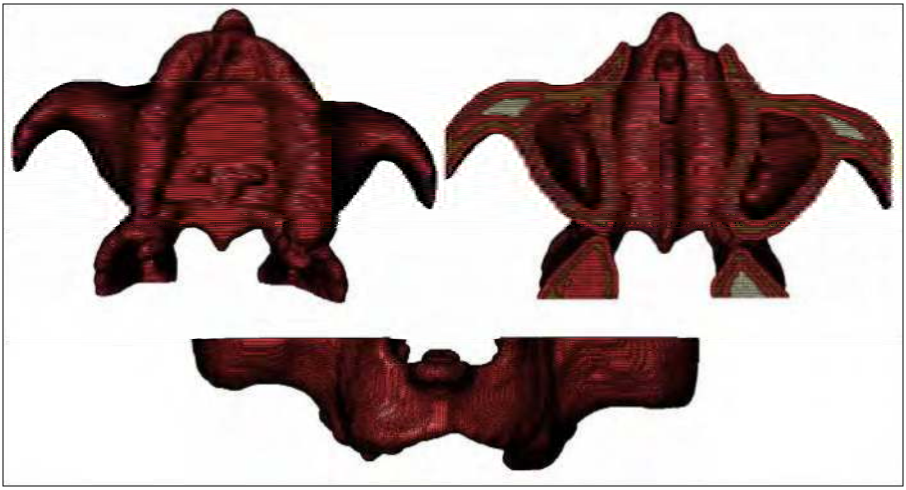

Fig. 15. Generated mesh with parabolic tetrahedral interpolation solid elements by the Simpleware software. The meshed model is ready to be imported by the finite element analysis software to investigate de stress distribution into the bone tissue.

6. Future perspectives

Considering that computational power is exhibiting rapid progress and hardware costs are decreasing, the numerical techniques probably will increase its application over time. Thus, the finite element method will be increasingly applied in Dentistry to generate reliable results for the biomechanical investigation of dental and supporting structures at lower cost than other *in vitro* and *in vivo* approaches. In addition, this technique can be associated to clinical evaluations as a further tool for diagnosis and/or treatment planning. For instance, the numerical techniques of the finite element method are increasingly indicated to simulate dental movements induced by orthodontic force systems. Thus, this method may provide information to the orthodontist about the choice of individual therapy (Clement et al., 2004).

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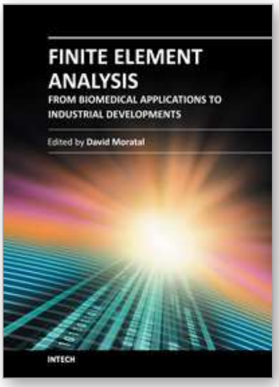

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Finite Element Analysis represents a numerical technique for finding approximate solutions to partial differential equations as well as integral equations, permitting the numerical analysis of complex structures based on their material properties. This book presents 20 different chapters in the application of Finite Elements, ranging from Biomedical Engineering to Manufacturing Industry and Industrial Developments. It has been written at a level suitable for use in a graduate course on applications of finite element modelling and analysis (mechanical, civil and biomedical engineering studies, for instance), without excluding its use by researchers or professional engineers interested in the field, seeking to gain a deeper understanding concerning Finite Element Analysis.