A CT-based 3D-Finite element analysis of using zirconia prosthetic material as a full-arch hybrid fixed detachable mandibular prosthesis

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Abstract: The present study was designed to identify the stresses occurring on implants, abutments, and the surrounding bone using three-dimensional (3D) FEA. A CT-based 3D finite element model of the mandibular arch of an edentulous patient was created. In addition, four endosseous implants and their abutments were modeled using CAD designing followed by designing a prosthesis created from the studied type of zirconia material. The results showed maximum stress and maximum deformation values were presented in the Zirconia prosthetic material. Therefore, within the limitations of this study, for designing the implant-supported prosthesis, use of prosthetic material of high elastic modulus and high flexural strength like Zirconia optimized the stress distribution. Zirconia received the highest stresses and showed the highest deformation values among other components. Thus, Zirconia transmitted little stresses to the underlying components. All of stresses transmitted to the cortical and trabecular bone were less than the physiologic limit of the bony tissues.

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1. Introduction

Recently major dental laboratories reported that the percentage of use of full ceramic crown is higher than porcelain fused to metal crowns. Therefore, dental ceramics have been improved and become increasingly more popular in dentistry (Ghosh 2009). Among the dental ceramics, Zirconia has increased popularity in contemporary dentistry due to its high biocompatibility, low bacterial surface adhesion, high flexural strength, toughness due to a transformation toughening mechanism, and esthetic properties (Papaspyridakos and Lal 2013) (Guess et al 2010). These properties have led to the introduction of zirconia-based restorations as alternative to the traditional porcelain fused to metal (PFM) restorations. It is currently being used for the fabrication of implant abutments and all ceramic copings, multiple unit, and complete arch frameworks for both fixed prosthodontics and implant dentistry (Nakamura et al 2010).

Zirconia is a polymorphic material that displays four different crystalline structures. Different oxides are added to zirconia to stabilize the crystalline phases, Magnesia (MgO), Yttria (Y2O3), Calcia (CaO), and Ceria (CeO). This allows the generation of multiphase material known as Partially Stabilized Zirconia (PSZ), whose microstructure at room temperature generally consists of cubic zirconia as the major phase, with monoclinic and tetragonal zirconia precipitates as the minor phase. The addition of approximately 2%-3% of mol yttria (Y2O3) as a stabilizing agent in zirconia generates a multiphase structure, designated the metastable tetragonal phase, which has good mechanical properties (Piconi and Maccauro 1999). Owing to the metastable tetragonal phase, stabilized zirconia will display a stress-induced transformation toughening mechanism. The transformation from the tetragonal to the monoclinic phase is associated with a 3% to 4% localized volume expansion that induces counteracting compressive stresses in compromised areas (Christel et al 1989) (Roediger et al 2010) (Saridag et al 2013).

In 1975 the British physicist Ron Garvie published his research on the possibility of stabilizing the tetragonal structure of zirconium dioxide by adding about 5.5% yttrium oxide material helped to achieve the exceptional mechanical properties and high biological stability (Christel et al 1989).

Moreover yettria stabilized zirconia YSZ like BruxZir is an ideal material for dental restorations because of its physical properties. Recent studies showed that Bruxzir is less hard than feldspathic porcelain to the opposing natural dentition. PFM and full cast have dramatically decreased over the past couple years (Roediger et al 2010) (Saridag et al 2013).

Other studies conducted by many authors to investigate the optical properties of zirconia, they found that the most innovative property, is color and translucency. Zirconia has a natural opaque white hue, but some Laboratories have recorded advancements that allow zirconia to be changed into a more desirable translucent natural ivory shade. This shade is much more lifelike than typical snow-white zirconia (Ara Nazarian, 2012) (Mc Omie, 2012).

There are little clinical data for implant-supported zirconia prostheses. (Larsson et al. 2010) conducted clinical studies on zirconia as a supra structure prosthesis. They compared two different material systems for 2–5 unit zirconia fixed partial dentures. The study showed that one of the systems resulted in an unacceptable amount of porcelain fractures after 5 years in function. Moreover, another study carried out by (Larsson et al. 2010) over 3-year follow up. The study investigated the survival and complication rates of all-ceramic and metal-ceramic reconstructions after an observation period of at least 3 years. They resulted in high rate of ceramic chipping have been reported and demonstrated with cement retained zirconia complete denture prosthesis for edentulous mandibles.

However, (Heintze & Rousson 2010) conducted a clinical study with up to 5-year clinical follow up. They confirmed the high stability of zirconia as framework material for tooth supported fixed dental prostheses (FDPs) and crowns.

The introduction of stabilized zirconia created a real possibility and promise for the application of ceramics in dental reconstructions. However, (Kelly & Denry 2008) claimed that the mechanical properties of zirconia are not still great and a suitable processing and clinical application protocol is still not fully known and controlled.

On the other hand, (Denry and Holloway 2010) reviewed the properties of YSZ, they found that it is extremely useful in esthetically critical areas of the mouth, due to its high refractive index, low absorption coefficient and high opacity in the visible and infrared spectrum.

Clinical studies indicated that reconstructions could be fabricated with zirconia frameworks, either on teeth or on implants, with good clinical success. No fractures of zirconia abutment have been reported in studies of implant single crowns in anterior and premolar regions during a maximum of 4 years of function. (Chopra 2013) (Canullo 2007). (Abrahamsson et al. 1998) claimed that abutment material may play an important role in the prevention of crestal bone and soft tissue recession. Early clinical studies of zirconia abutments on implants reported good stability of implant-supported single reconstructions. Marginal bone levels around implants were stable, and healthy soft tissues were observed (Denry and Kelly. 2008) (Mawson et al.2006).

Custom implant abutments can be precisely designed to provide ideal support of the restoration and underlying soft tissue, while all-ceramic crowns & bridges milled from BruxZir Solid Zirconia, boasting a flexural strength far greater than layered porcelain, now exhibit sufficient esthetics for use in the anterior. The CAD/CAM technology inherent in the production of these restorations not only reduces their costs, but also limits the need for adjustments or remakes, resulting in less chair time per patient (McLaren and Giordano. 2005).

In the same manner, the use of finite element analysis in dental biomechanics has been submitted to a significant increase during the last decades (Ça˘glar et al. 2011) (Linkevicius and Apse 2008). Some authors (Geng et al 2001) (Ozen et al 2007) often carry out very simplified finite element models, pretexting comparative analyses. Consequently, the results obtained are quite different and non-reliable as they are very far from real loading conditions. Consequently, a great attention has to be given to the model quality if reliable results want to be obtained. Sufficiently fine meshes have to be used in areas where stress is expected as in peri-implant zones (Teixeira et al 1998). Boundary conditions need also to be modeled very accurately in order to get close to real interactions. Bone anisotropy and remodeling can also be introduced to take into account the directional variation of tissue properties and their evolution after implantation as a function of the applied load magnitude.

Therefore, the present study was designed to identify the stresses occurring on implants, abutments, and the surrounding bone using three-dimensional (3D) FEA.

2. Material and Methods

A CT-based 3D finite element model of the mandibular arch of an edentulous patient was created. In addition, four endosseous implants and their abutments were modeled using CAD designing followed by designing a prosthesis created from the studied type of zirconia material.

A 60-year old male patient with moderately resorbed residual ridge was CT-scanned using standardized CT scanning procedures (Somatom Definition Flash, Siemens Ag, Germany). Dicom images were generated with resolution 500 x 500 pixels and high contrast for bone algorithm. Dicom images were stored then imported to Mimics v10.01 software (Materialize Software, Belgium) where images were masked and 3d model of the edentulous mandible was created, (fig. l). The 3d model of the mandible was exported as STL file. Using 3-matic software v6; (Materialize Software, Belgium) STL file was refined and prepared for patching and be converting to solid cad file. The model was duplicated and shelled by 2 mm offset to simulate the cortical bone. The cad file then exported as IGS file format.

Table 1: Material properties of different components used in the study (including Young's Modulus and Poison's ratio)

	Young's Modulus(GPa)	*Poisson's ratio*	*References*
Implants and abutments[28]	110 GPa	0.33	Colling EW. The physical metallurgy of titanium alloys. Metals Park (OH): American Society for Metals; 1984.
Compact bone[29]	20 GPa	0.3	T. Takahashi et al. / Journal of Prosthodontic Research 54 (2010) 179–184
Cancellous bone[29]	2 GPa	0.4	T. Takahashi et al. / Journal of Prosthodontic Research 54 (2010) 179–184
Prosthesis[30]	200 GPa	0.3	Kohal et al. / Int J Prosthodont 2002;15:189–194.

A CAD Software (SolidWorks Corporation, Concord, MA, USA) was used for modeling implants, abutments, and Prosthesis. The implant fixture was designed by simulating the dimensions (4 mm diameter x 13.5 mm length) and configurations of the implant fixture (Brånemark MKIII Groovy, Nobel Biocare, Switzerland) and its abutment according to the manufacturer's data. Blueprint images of the selected fixture were used as a base for several building and subtracting operations to represent the unique characteristics of this implant type. The fixture tapering, threads and grooves were simulated. Additionally, the apical cutbacks and the coronal steps were formed. Two anterior implants were placed in the anterior area bilaterally and two angled (20°) were placed in the premolar area simulating (All-on-4) design. Finally, the prosthesis was designed to follow the shaped of the hybrid fixed-detachable prosthesis. The prosthesis design included both base and teeth. The models' assemblies were then transferred as IGS file format from the CAD software to the design modeler module of ANSYS v 14 finite element software (ANSYS Inc., Houston, PA, USA), (fig. 2).

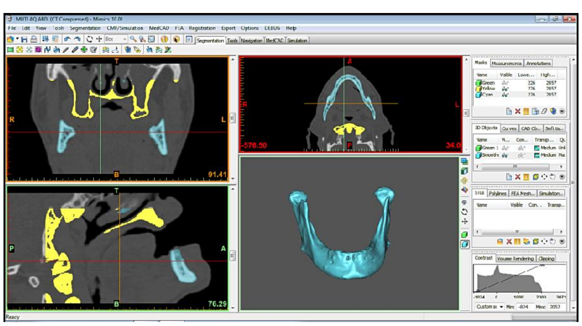

Fig. 1: Dicom images processed in Mimics software and converted both bone types to 3D STL file.

Using the predefined material library, material properties were assigned to each solid component of the assembly according to Table 1. All material properties were assumed linear, homogenous and isotropic to facilitate calculation process and reduce solving time. Accordingly, two values (Young's modulus and Poisson's ratio) were assigned for each part, Table 1.

After adding material data to the assembly, the model was meshed using tetrahedron (four-sided) elements. Mesh refinement were applied at the areas of contacts and fine model details, (fig. 3). The total elements number used for meshing the model assembly was (190049 elements) with total nodes equal (337436 nodes).Boundary condition was designed using bonded-contact type at all areas of contact of the model components, especially between implants and bone to simulate full osseointegration. The boundary conditions included constraining all three degrees of freedom at each of the nodes located at the joint surface of the condyles and the attachment regions of the masticatory muscles (masseter, temporalis, medial pterygoid, and lateral pterygoid). A unilateral load was simulated by applying an oblique load (vertical load of 200 N and horizontal load of 40 N) from buccal to lingual direction of lower first molar (Liao et al 2008). The load applied was sparsely distributed (4 mm in diameter) on the lingual cusps of the first molar in order to avoid false stress concentration in the area of load application.

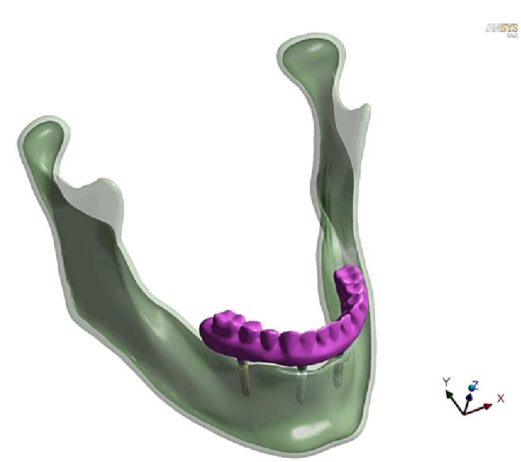

Fig. 2: The full design assembly imported to design modeler module of the finite element software.

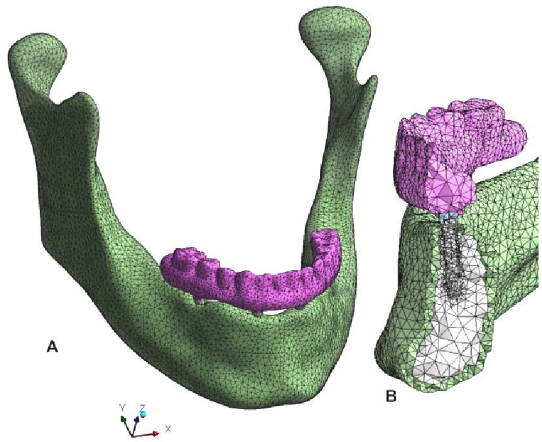

Fig. 3: Meshing of the model assembly using tetrahedron element type,A. A representative cross-section at molar areas showing elements in 3D seen in B.

Considering that implants, abutments, prosthetic component and bones are varied from ductile to brittle materials, the VonMises or equivalent stresses(σvM), the main maximum (σmax) (tensile) and minimum (σmin) (compression) stresses were obtained (Ferreira et al.2014). In addition, total deformation of the studied components was also observed.

3. Results

Stresses Transmitted to Cortical Bone

Stresses and deformation for the cortical bone were measured in MPa and in mm respectively. (fig. 4)

Qualitatively, σvM (18. 584 – 5.8827e-1) MPa were observed in the bone tissue surrounding the implants in the same side and on the opposite side of load application. The highest concentrations of σvM were more distant from the midline, near the angle of the mandible and in the peri-implant region. Stresses in the peri-implant region of same side of load application are more than that of the implant crossing the midline.(Fig. 4).

Compressive stresses (principal stress minimum) (1.5173 – -23.621) MPa occurring in the cortical bone concentrated in the palatal and proximal regions of the bone surrounding the implants. Some stresses represented near the angle of the mandible of same side of load application. Stresses in the peri-implant region of same side of load application are more than that of the implant crossing the midline.(Fig. 4).

Tensile stresses (principal stress maximum) (12.942 – -4.9712) MPa were concentrated near the angle of the mandible of same side of load application. More stresses represented in the buccal region of the mesial implant socket than the distal implant socket of the same side of load application (fig. 4).

- Deformation of the cortical bone

Deformation (0.014352 – 0) mm were presented all over the cortical bone. The highest values were concentrated in the implant distant from the midline. deformation also observed in the inferior border of the mandible in the same side of load application. (fig.4)

- Stresses Transmitted to Trabecular Bone

Stresses and deformation for the trabicular bone were measured in MPa and in mm respectively. (fig. 5)

The σvM stress (2.5018 – 8.2966e-7) MPa in trabecular bone was concentrated on the palatal of the implant socket. The stresses in the trabecular bone of the implant socket of the same side of the load application were more than the implant crossing the midline. (Fig. 5)

Compressive stresses (principal stress minimum) (0.33732 – -2.1184) MPa occurring in the trabicular bone were concentrated in the palatal and proximal regions of the bone, whereas the stresses were decreased in the buccal region. Stresses in the peri-implant region of same side of load application are more than that of the implant crossing the midline, (Fig. 5).

Tensile stresses (principal stress maximum) (1.5922 - 4.7362) MPa were concentrated in the socket of the implant of the same loading side. Whereas some stresses showed buccally around the implant and crossing the midline. Stresses in the peri-implant region of same side of load application are more than that of the implant crossing the midline, (Fig. 5).

Generally, the stresses occurring in the cortical bone were higher than those seen in the trabecular bone

- Deformation of the trabecular bone

Deformation (0.014052 – 1.0399e-8) mm was presented all over the trabicular bone. More deformation is observed in the peri-implants region in the same side of load application. The highest values were concentrated in the implant distant from the midline, Palatally more than buccally and proximally. Some deformation showed in the inferior border of the mandible of the same loading side (fig.5)

- Stress in the implants

Stresses and deformation for implants were measured in MPa and in mm respectively. (fig. 6)

The points of greatest von Misses stresses (32.4-0.069404) MPa were located in the first implants (on the side of loading). Stresses on the distal implant are higher than on the mesial implant. However, the stresses were dissipated to the last implant opposite to loading and distributed throughout the set implant, cylinder, and screw. The highest stresses were concentrated at the buccal neck region of the implant in contact with the compact bone (Fig. 6).

Compressive stresses (principal stress minimum) (5.1934— -42.608) MPa were located in the first implants (on the side of loading). Stresses on the mesial implant are higher than on the distal implant. However, the stresses were dissipated to the last implant opposite to loading and distributed throughout the set implant, cylinder, and screw. The highest stresses were concentrated at the buccal neck region of the implant in contact with the compact bone (Fig.6).

Tensile stresses (principal stress maximum) (21.377 – -19.376) MPa were concentrated in the the first implants (on the side of loading). Stresses on the mesial implant are higher than on the distal implant. However, the stresses were dissipated to the implants crossing the midline. stresses distributed throughout the set implant, cylinder, and screw. The highest tensile stresses were concentrated at the buccal neck region of the implant in contact with the compact bone. (Fig. 6).

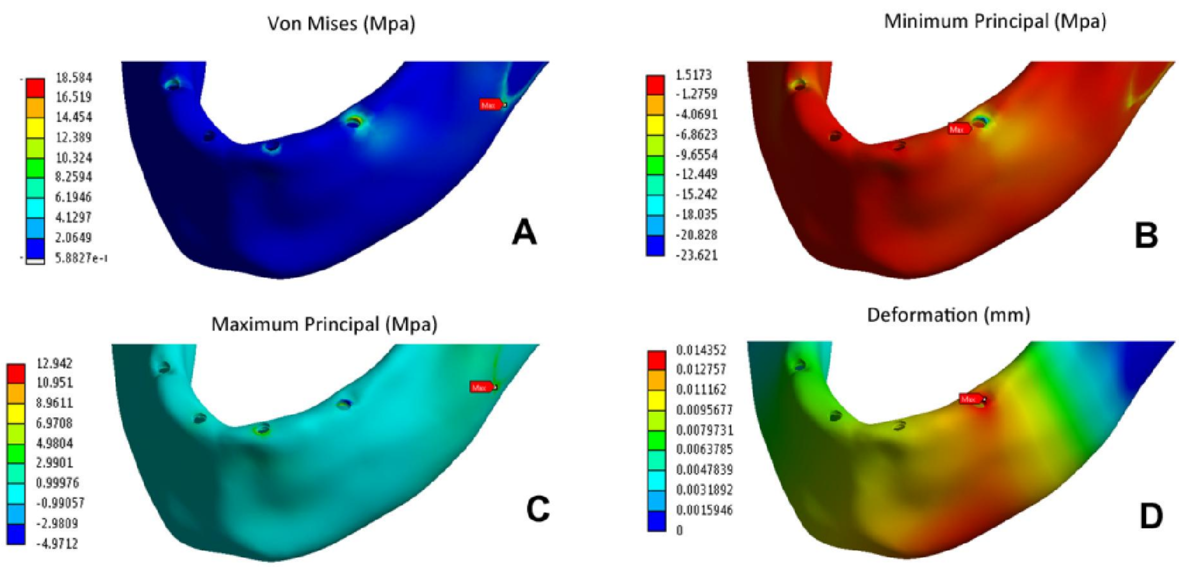

Fig. 4: Values of equivalent stresses, maximum and minimum principal stresses in MPa and deformation in mm for the cortical bone

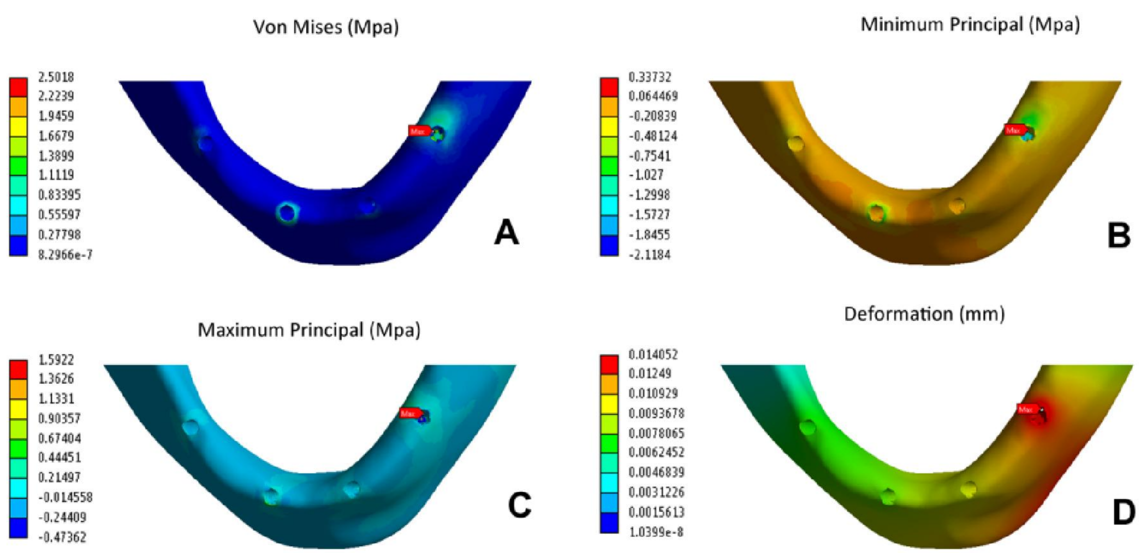

Fig. 5: Values of equivalent stresses, maximum and minimum principal stresses in MPa and deformation in mm for the trabicular bone

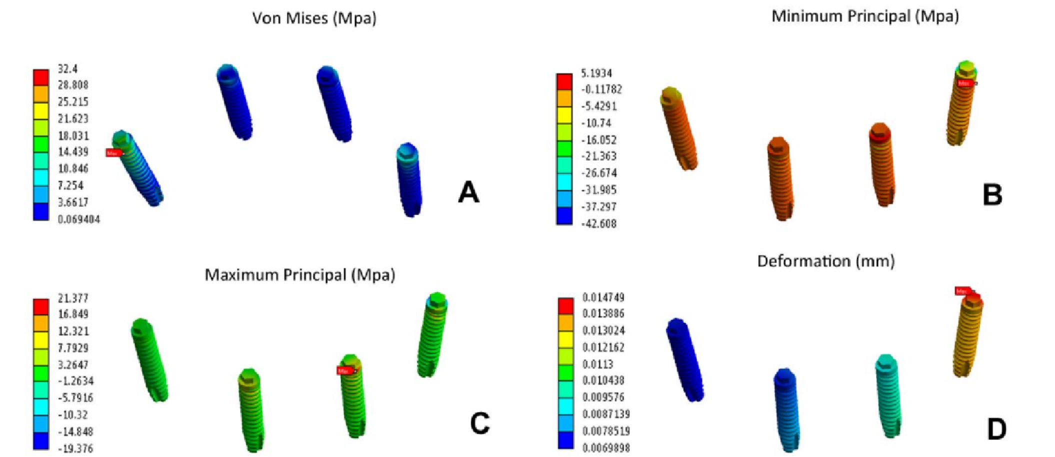

Fig. 6: Values of equivalent stresses, maximum and minimum principal stresses in MPa and deformation in mm for implants.

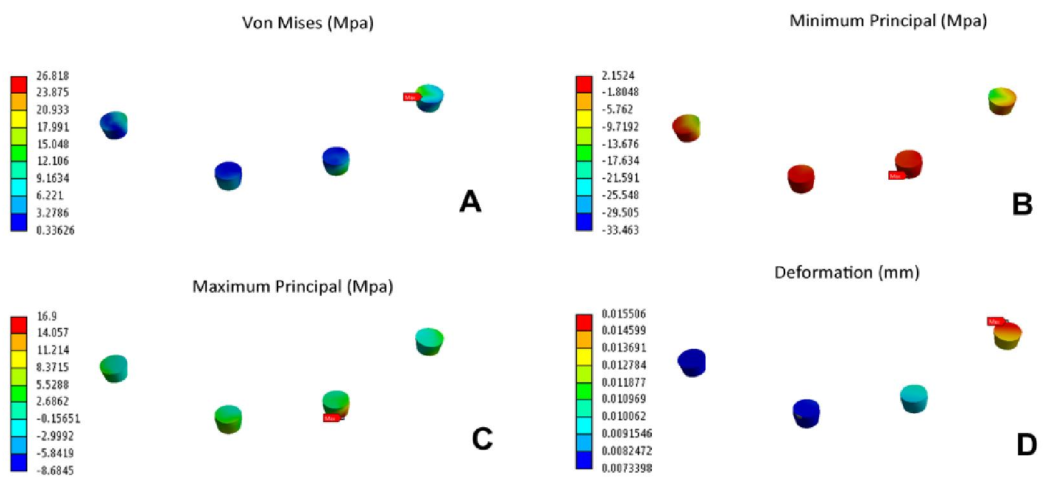

Fig. 7: Values of equivalent stresses, maximum and minimum principal stresses in MPa and deformation in mm for the abutments.

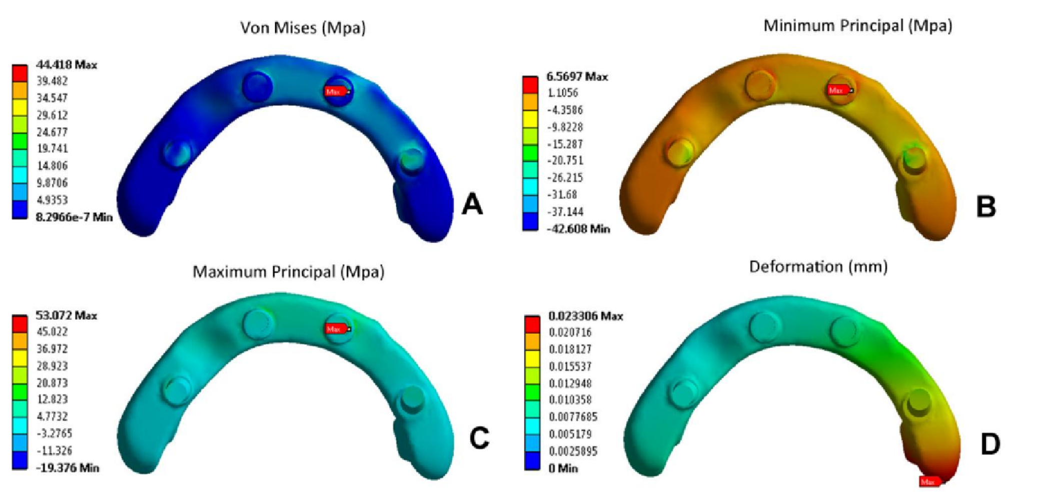

Fig. 8: Values of equivalent stresses, maximum and minimum principal stresses in MPa and deformation in mm for the prosthesis.

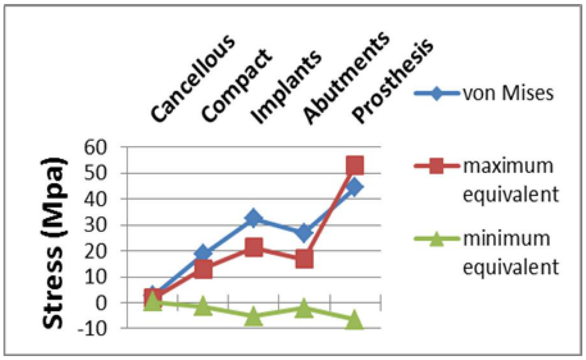

Fig. 9: Values of stresses in MPa for bone, implants, abutments and the prosthesis.

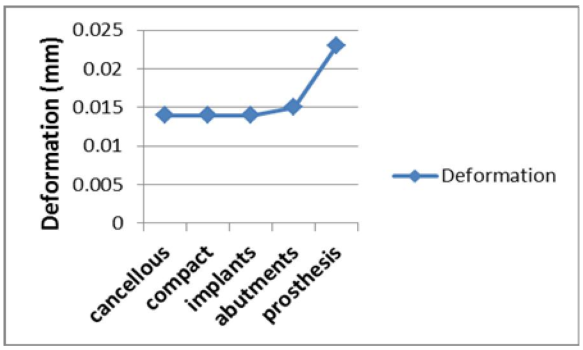

Fig. 10: Values of deformation in mm for bone, implants, abutments and the prosthesis.

- Deformation of the implants

Deformation (0.014749 – 0.0069898) mm were presented maximally on the distal implant of the same loading side. Deformation at the neck showed the maximum value all over this implant. Less deformation observed in the implants crossing the midline than on the implant in the same side of load application. The highest deformation values observed at the buccal region of the implant.

- Stresses Transmitted to Abutments

Stresses and deformation for the abutments were measured in MPa and in mm respectively. (fig. 7)

The highest σvM (26.818 - 0.33626) MPa stresses were observed palatally on the distal abutment and bucally on the mesial abutment. Some stresses showed on the abutments crossing the midline. (Figs.7).

The highest compressive (2.1524 - -33.463) MPa stresses were observed all over the four implants. These stresses were concentrated on the buccal finish line of the distal abutment of the same load application. Some stresses showed on the distal abutment crossing the midline, (Fig.7).

The highest tensile (16.9 - -8.6845) MPa stresses were observed all over the four implants. These stresses were concentrated on the buccal finish line of the abutments of the same load application. (Fig.7).

- Deformation of the abutments

The highest deformation (0.015506 - 0.0073398) mm was observed palatally on the distal abutment of the same loading side. Stresses on the abutment crossing the midline were less than those on the abutments of the same loading side (Fig.7).

- Stress in the prosthesis

Stresses and deformation for the prosthesis were measured in MPa and in mm respectively (fig. 8).

The σvM stresses were (39.482 - 8.2966e-7) Qualitatively, the stress areas were located in the region of load application; however stress was a spread palatalyl and buccally on the implant prosthesis junction areas (Fig. 8).

The highest compressive (6.5697 - -42.608) MPa stresses were observed all over the prosthesis. These stresses were concentrated on the buccal finish line of the implant prosthesis junction areas of the same loading side. And on the palatal surface of the prosthesis on the same load application. (Fig.8).

The highest tensile (53.072- -19.376) MPa stresses were observed all over the prosthesis. These stresses were concentrated on the buccal finish line of the implant prosthesis junction areas of the same load application. (Fig.8).

- Deformation of the prosthesis

The highest deformation (0.03306 - 0) mm were observed palatally and distally on the prosthesis on the same loading side. (Fig.8).

Values of stresses in MPa for bone, implants, abutments and the prosthesis are shown in fig.9

The highest stresses were presented in the prosthesis.

Values of deformation in mm for bone, implants, abutments and the prosthesis are shown in fig.10

The highest values were presented in the prosthesis.

4. Discussions

The successful osseointegration and long-term survival of oral implants depend on several biomechanical factors. They are influenced by the way the mechanical stresses are transferred from the implant to the surrounding bone. Several factors may influence the stress transfer from implant prosthesis and bones, such as loading type, bone quality and volume and selected implant position. In addition, stress-strain distribution influenced by prosthesis material properties and design. (Kregzde, 1993)

Accordingly, manufacturers have been trying to develop implant designs and materials that reduce the stresses generated around implants during loading. Superstructures of the implants may also affect the stress distribution. It has been suggested that stress-absorbing systems might be incorporated into the superstructures to compensate deficient viscoelasticity of the bone-implant interface and so reduce implant loading (Ciftçi and Canay 2000).

The results of the present study showed that zirconium prosthetic material received the highest σvM, compressive and tensile stress values among all other components. It also presented the highest deformation value compared to the other components. The σvM, compressive and tensile stresses transmitted to the cortical bone were higher than those seen in the trabecular bone. However, under studied loading condition, all of the stress values transmitted to the cortical and trabecular bone were less than the physiologic limit of the bony tissues. These results might be due to the high rigidity of the zirconia prosthesis. Zirconia is used as a prosthesis material. The choice of this material was because of its high mechanical properties. Zirconium is a rigid material and has twice the young's modulus of titanium material. Zirconia also has a high flexural strength (900 to 1,200 MPa) (Stevens 1986 and Piconi et al. 1998.) Consequently, stresses were absorbed by the prosthetic material, into which they distributed thoroughly. Thus, zirconia might inhibit stress transmition, to the underlying components, to a minimal measurable values and it seems to act as a protective shield, which allowed favorable distribution of stresses to underlying components. So that the use of highly rigid Zirconia prosthetic material possibly might be a reason behind the lesser stress values transmitted to the implants and bones respectively.

High-rigidity prostheses were recommended by (Brenzing et al 1995). He clarified that the use of low elastic moduli alloys for the superstructure predicts larger stresses at the bone–implant interface on the loading side than the use of a rigid alloy for a superstructure with the same geometry.

The result of the current study was also in agreement with (Eskitascıoglu et al.1996) they reported that porcelain crowns absorbed and redistributed stresses with minimal stress transfer to the implant and bones in single implant replacement. Moreover, (Jacques et al.2009) stated that a material with a lower elasticity modulus offers lower flexural strength, and structures made of rigid basic alloys undergo less deformation and hence do not overload the screws and other prosthesis components.

The current study also agreed with (Ferreira et al. 2014) findings. They conducted a study to investigate the influence of the type of esthetic veneering material as a determining factor to differentiate the stress distribution in the prosthesis, implants and bone. They found that resin teeth showed 50% increase of stress values in infrastructures when compared with porcelain teeth. They believed that the high rigidity; the high elasticity modulus, and consequently the high flexural strength of porcelain, favor the stress dissipation and reduce the risk of mechanical overload of other structures. While the acrylic resin with a low elasticity modulus can result in greater deflection, especially in the loading side, transmitting more stresses to infrastructures.

Therefore, acrylic resin has a great flexibility and better to be exchanged with porcelain teeth; particularly in the cases of prostheses with larger cantilevers, in order to distribute the stresses favorably and increase the full-arch implant-supported fixed prosthesis rigidity and thus avoiding fractures.

Moreover, (Assunção et al. 2010) believed that according to the biomechanical point of view, materials with high elasticity modulus are more suitable for the superstructure of implant-supported prosthesis.

On the other hand, Some researchers reported that a more resilient superstructure material would be useful to reduce stresses around the implant by the materials' elastic deformation behaviors.(Stegaroiu et al.1998). Porcelain is another material option, presents greater wear resistance, and provides more favorable esthetics results than acrylic resin. However, some authors reported that porcelain is a more rigid material and does not absorb the stresses, thus the forces developed in the occlusal surface are transmitted directly to the prosthesis, implant and bone/implant interface, unless they are interrupted somehow (Geng et al.2001 and). Moreover, (Skalak 1983) was proponents to the use of acrylic resin teeth and attributed that to the shock absorption effect and so protection of implants. They recommended the use of acrylic resin as the material of choice for the occlusal surfaces of implant-retained prostheses. The resiliency of this material was suggested as a safeguard against the negative effects of impact forces and microfracture of the bone-implant interface. In fact, acrylic resins are burdened with technical and subjective disadvantages, due to their low wear resistances, premature contacts often occur after several months of prosthesis delivery. On the other hand, gold and porcelain surfaces are believed not to provide force absorption, but they are also frequently used. The choice of prosthesis material remains as a topic of controversy and argument. (Cehreli 2004).

The difference between the results of the present study and those studies that recommended the use of low modulus prosthetic material could be claimed to the difference of the type of prosthesis and experimental design conditions (attachment, loading, mesh generation, and material properties). In case of full-arch implant-supported fixed dentures, multiple implants splinted through a metallic infrastructure, which may influence the stress distribution. These multiple prosthesis may also include other controlling factors such as cantilevers, different implant positions, distribution, and angulations; as All-on-4 design. Additionally, the unilateral loading of the prosthesis may promote a stress concentration in all implants and surrounding bone in different degrees, which may also cause a significant incidence of failures in various parts of the prosthesis, such as screws, abutments, infrastructure and implant.

On the contrary, there are studies that reported changing the superstructure material did not influence the stress levels. (Stegaroiu et al.1998), (Bassit et al.2002).

Contradicting results were noted by (Hulterstrom and Nilsson 1994). They investigated clinically the longevity of implant-supported fixed dentures of the Branemark type. Gold cylinders and cobalt–chromium alloy frameworks were used. The superstructure materials did not affect the peri-implant bone loss. Similarly, Gomes et al.2011 observed no influence of the material superstructure of single implant supported prosthesis on the stress distribution in the bone tissue.

(Stegaroiu et al.2004) showed same findings regarding the influence of the prosthetic material. They conducted a study to investigate the effect of 3 superstructure materials (highly filled composite resin, acrylic resin, and gold alloy) on the strain around an implant under static and non-impact dynamic loading. Although the strain values differed significantly between the static and dynamic loading, they found no significant difference among the superstructure materials under either loading condition. These findings are in agreement with in vivo measurements that recommend using cyclic rather than static loading during investigating occlusal material behavior. Finally, they concluded that under static and dynamic loading, the 3 superstructure materials tested had the same influence on the strain transmitted to the bone-implant interface surrounding single implant.

This finding may be related to the selected loading type in the present study. It is stated that under static loading, changing the resilience properties of different superstructure materials does not result in significant differences in stress concentrations and distributions around the implants (Gracis et al.1991). So that, the advantages of using resilient materials might be apparent under dynamic loads and impact forces.

An oblique load was used in the current study. This was similar to load applied in previous study (Barão et al.2013). The occlusal characteristics and cusp inclination of implant-supported prostheses have important influence on the stress/ strain transmission to the implant surrounding bone tissue. It has been showed that the cusp inclination would enhance the lateral forces when axial loads are induced on occlusal area (Weinberg 1998). In this sense, the oblique load provides a more realistic approach to the implant-supported system than axial or horizontal forces applied isolated. Oblique load promotes greater stress development in the cortical bone versus vertical or horizontal forces. This was agreed by many authors (Petrie et al.2005).

The location of stress concentration in bone, implants, abutments and prosthesis was in the area of load application and in the most distal region, on the same side of the load application. These areas were directly subjected to the applied load, so they received the highest stresses. The most distal region and the area of mandibular angle also received high stresses. These areas might be located as a fulcrum for the applied load. Some stresses showed on the abutments crossing the midline. Stresses presented on the implants were concentrated buccally at the collar areas were implants meet the bone. These parts of the implants presumably represent areas of the highest torque and stress concentrations, caused by levering effects. These results were in agreement with the findings of (Aramouni et al.2008).

This study was based on previous studies, which also applied forces unilaterally, in the mandibular first molar (Barão et al.2013). Even if the forces were applied bilaterally, obviously, there would be a higher compensation in posterior stress distribution, however, it would not alter the profile of results presented in this study, and the greatest stresses would remain located in the distal portion of the prosthetic/implant complex.

The design of the implants, their distribution and their angulation may be another favorable factor that allowed less stresses transmitted to the bone on the expenses of the prosthesis.

The use of all-on-4 design was selected for the present study as a popular design used nowadays for rehabilitating mandibular edentulous patients. It is also selected to maximize the benefit of the current study by using a mechanically questionable but clinically successful design.

There are some simplifications rather than limitations used in this study. Static loading behaviour was used rather than dynamic loading and the bone tissues treated as linear materials and all contacts as a bonded contact.

It is recommended to conduct more studies similar to the present study using zirconia implant material instead of using titanium implant alloy followed by analysis of the amount and pattern of stresses transmitted to the bone.

5. Conclusion

Within the limitations of this study, it is wise to use of prosthetic material of high elastic modulus and high flexural strength like Zirconia for designing the implant-supported prosthesis to optimize stress distribution. Zirconia absorbed the highest stresses and showed the highest deformation values among other components. Thus, Zirconia transmitted little stresses to the underlying components. All of stresses transmitted to the cortical and trabecular bone were less than the physiologic limit of the bony tissues.

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