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Application of Finite Element Analysis in Implant Dentistry

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1. Introduction

Since Brånemark's discovery, dental implants have become the most common restorative technique for the rehabilitation of edentulism. Many factors can impact the survival of implant-supported restorations. The most important factor for determining the long-term success of osseointegration is the state of the peri-implant bone [1-3]. Ideal biomechanical conditions directly affect bone remodeling and help to maintain the integrity of non-living structures such as the implant, abutment, and superstructures (Figures 1-7). Oral dental implant interventions involving surgical and restorative procedures for the rehabilitation of various causes of edentulism are associated with several risks. In particular, mechanical and technical risks plays a major role in implant dentistry, resulting in increased rates of repairs, unnecessary costs and lost time, and even complications that may not be easily corrected (Figures 8-10) [4-7]. Therefore, the potential mechanical and technical risks of failure or associated complications need to be evaluated before undertaking such interventions, since the application of necessary precautions may improve the survival of implant-supported restorations. Consequently, the number of biomechanical studies in the field of implant dentistry has dramatically increased in an effort to reduce failure rates.

Several methods based on photoelastic, strain-gauge, and finite element analysis (FEA)-based studies have been used to investigate stress in the peri-implant region and in the components of implant-supported restorations [8-11]. FEA is a numerical stress analysis technique that is widely used to assess engineering and biomechanical problems before they occur [12,13]. A finite element model is constructed by dividing solid objects into several elements that are connected at a common nodal point. Each element is assigned appropriate material properties corresponding to the properties of the object being modeled. The first step is to subdivide the complex object geometry into a suitable set of smaller 'elements' of 'finite' dimensions. When combined with the 'mesh' model of the investigated structures,

each element can adopt a specific geometric shape (i.e., triangle, square, tetrahedron, etc.) with a specific internal strain function. Using these functions and the actual geometry of the element, the equilibrium equations between the external forces acting on the element and the displacement occurring at each node can be determined [9].



Figure 1. Missing molar in the mandible, to be treated with a dental implant-supported restoration

In implant dentistry literature, commonly used materials in FEA studies can be classified as either implant, peri-implant bone (cortical and cancellous bone), and restoration (Figure 11). This method allows application of simulated forces at specific points in the system and stress analysis in the peri-implant region and surrounding structures. 2-D and 3-D models can be created and models for every treatment alternative can be explored. However, 2-D models cannot simulate the behavior of 3-D structures as realistically as 3-D models, so most recent studies have focused on 3-D modeling [14-17].



Figure 2. After flap elevation, the cortical bone is visible



Figure 3. Dental implant with an abutment to be placed in the ridge created by the missing molar



Figure 4. Implant is placed in the ridge

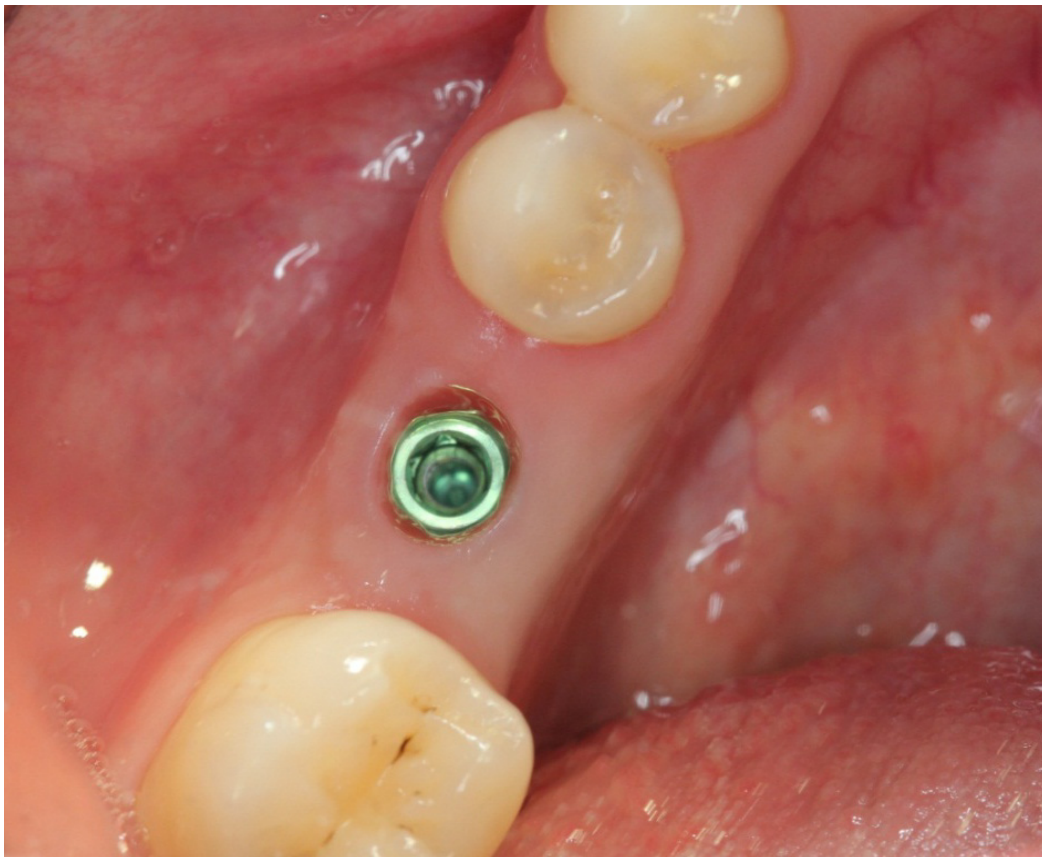


Figure 5. Occlusal view of the implant after 2 months of healing



Figure 6. Abutment is prepared and attached to the implant



Figure 7. Porcelain-fused metal implant-supported restoration in use with optimum treatment planning

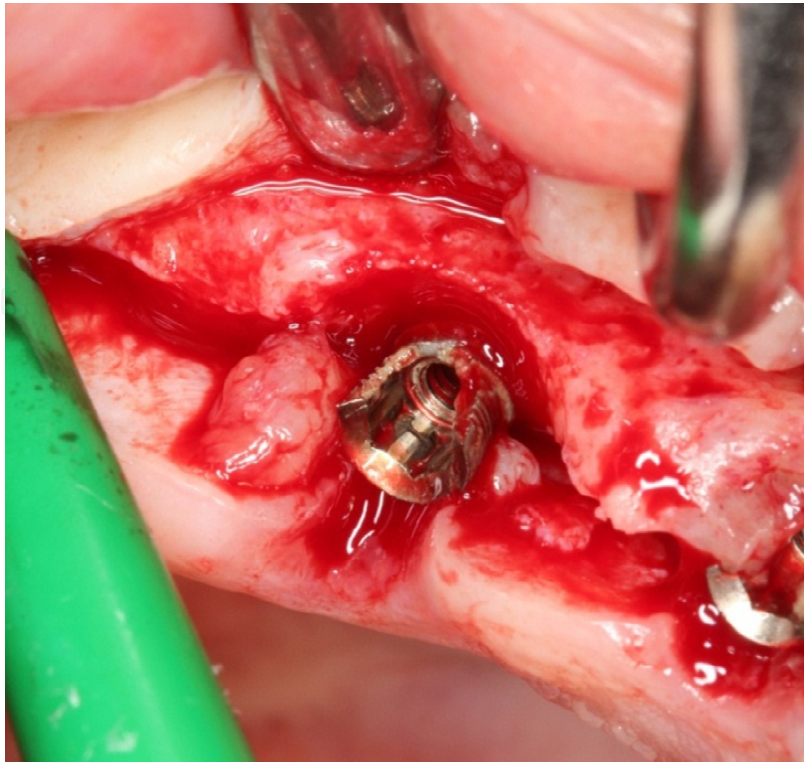


Figure 8. Intraoral picture of a broken implant due to excessive loading after 1 year of use

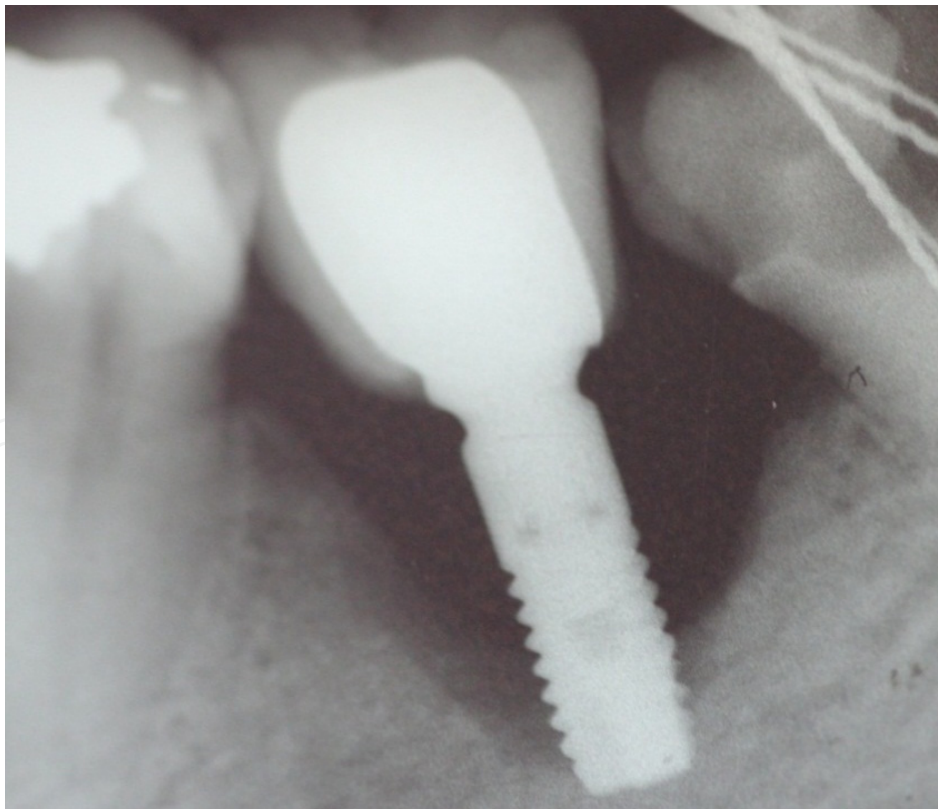


Figure 9. Severe bone resorption after 2 years of loading; implant and superstructure have no mechanical failure, but peri-implant bone could not resist excessive loading (biomechanical failure because of improper occlusal adjustment)

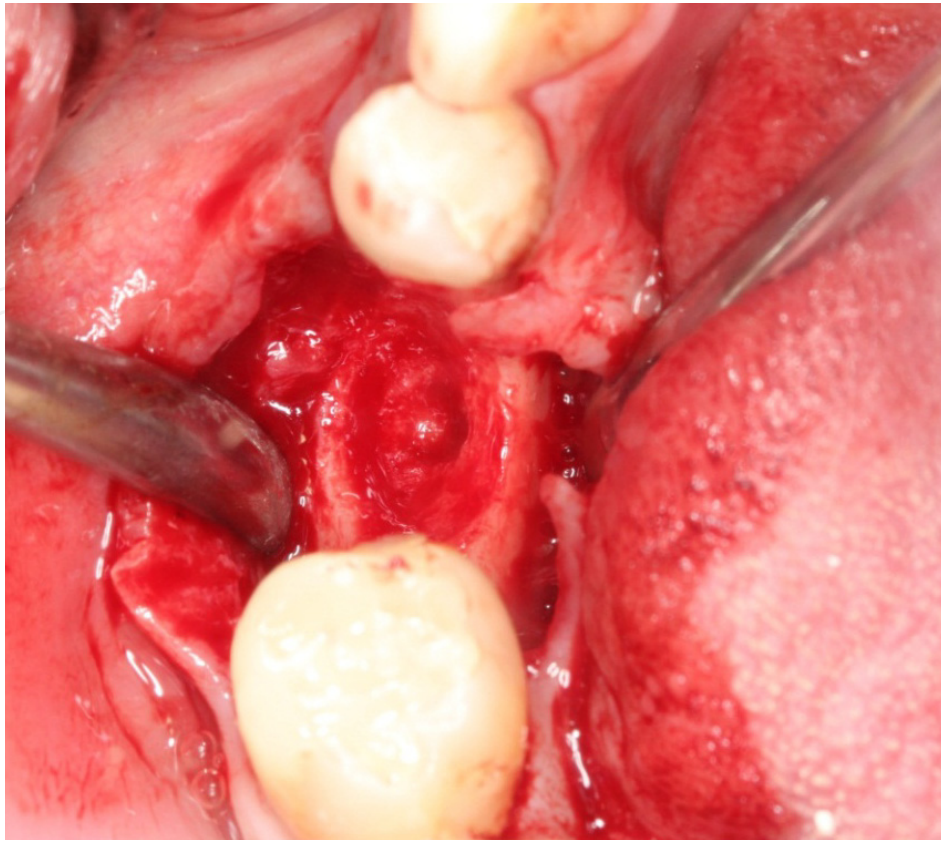


Figure 10. Severe bone defect is seen after implant removal; advanced bone regeneration techniques are needed to replace the implant



Figure 11. Modeling of bone, implant, abutment, and restoration

2. Modelization of living structure (bone)

To improve the quality of FEA research, strict attention should be paid to the modelization procedure as one of the most important part of FEA studies. The features of the model should resemble the physical properties of the actual structure as closely as possible, with respect to dimension and material properties. The most difficult and complex part of the modelization process involves capturing the detailed properties of living structures. Therefore, in general, specifications drawn from chapters of a detailed anatomy book or from tomographic scans of a jaw from a cadaveric human specimen can be used for the modeling procedure. Volumetric data obtained from tomography devices or magnetic resonance imaging are digitally reconstructed [18,19]. Then, the material properties applied to the elements can be varied according to the modeling requirements of a particular situation. Computed tomography offers another advantage for realistic modeling in not only the development of anatomic structures, but also the inclusion of material properties according to different bone density values [20,21]. In some studies, the bone is totally or partially modeled as a simple rectangle, ellipsoid, or U-shape [18]. In detailed studies, especially with data obtained from scanners, bone can be modeled in a very realistical manner; however, this increased level of geometric detail will result in increased working and computing time. According to the treatment alternatives being investigated, cortical bone can be layered in millimeters or can be neglected altogether in order to simulate weak bone properties similar to those found in the posterior maxilla (Figure 12). Bone properties related to density can be calibrated to range from very soft to dense bone, according to the individual research protocol. If only a specific area and/or condition of the mandible or maxilla is being investigated, there is no need to visualize and construct a model of entire jaw. Limiting the scope or features of the model will distinctly decrease the working time and costs, as previously discussed. A region of interest can be extracted using a number of techniques, such as a Boolean process (Figures 13-15), and any implant design can be adopted for the study. Regions of interest may change according to the study protocol. Portions of the mandible or maxilla, maxillary sinus region, and temporomandibular joint are the most common anatomical areas used in studies related to implantology. In the existing literature 2-D FEA bone models are generally simplified as a rectangular shape [14]. However, recent studies have used 3-D bone modeling to better represent the realistic anatomy of these complex structures [22-24].

In a previous FEA study, the human mandible model was based on a cadaveric mandible obtained from the anatomy department [25]. The edentulous cadaver mandible was scanned using a dental volumetric computed tomography device (ILUMA, Orthocad, CBCT scanner, 3M ESPE, St. Paul, MN, USA) (Figure 16). Volumetric data were reconstructed in 0.2 mm thick sections. The mandibular height and width were at least 10 mm and 5 mm, respectively. More detailed anatomic representations could be created in future studies through the use of computed tomography scanners that can slice objects into thinner sections, but this may increase the working time and development cost of the final finite

element model (FEM). In the study mentioned above, sections were digitized into the DICOM 3.0 format and visualized using 3-D Doctor software (Able Software Corp., Lexington, MA, USA). Cortical bone of 2 mm uniform thickness, and cancellous bone were also modeled (Figure 17). In this study, cortical and cancellous bone model components were considered homogenous. However, in fact, cancellous bone in particular has widely variable density properties. The non-uniform nature of the density of this anatomic structure may affect the magnitude and distribution of stress concentration after loading. These simplifications are common in studies that employ FEA and are aimed at limiting the computing difficulties associated with performance of these studies [18,26,27]. To develop more realistic models of living structures, future studies may include variable density properties obtained from bone density values measured in Hounsfield Units or from other advanced data obtained from computed tomography scans performed with individual patients (Figure 18) [28-30].

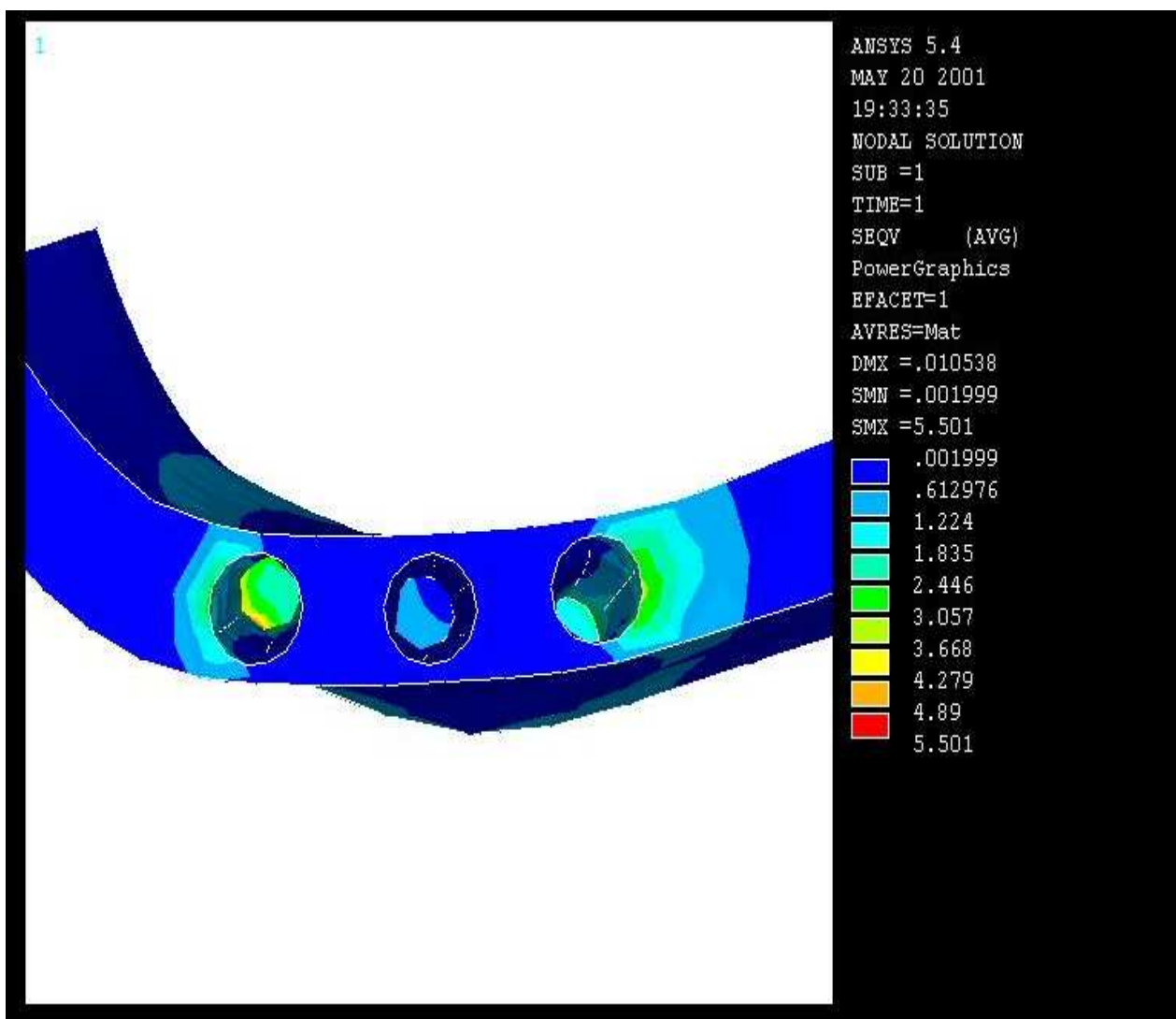


Figure 12. Cortical thickness of the posterior maxilla is neglected; only cancellous bone properties are modeled

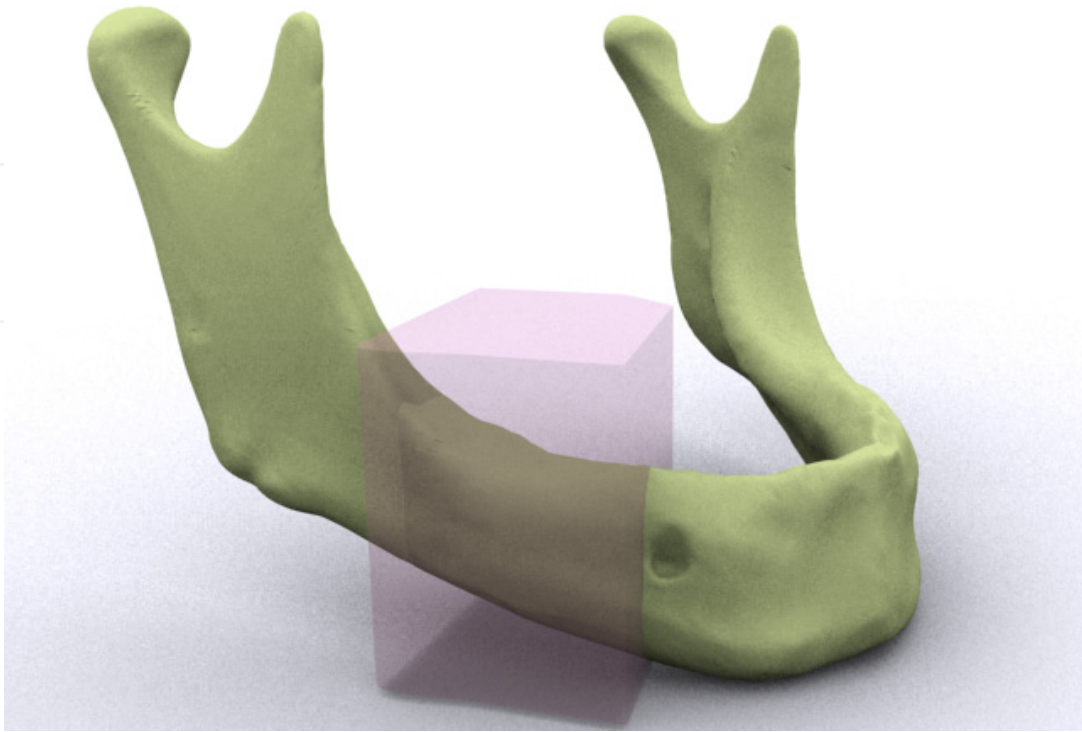


Figure 13. Mandible is modeled and region of interest is selected

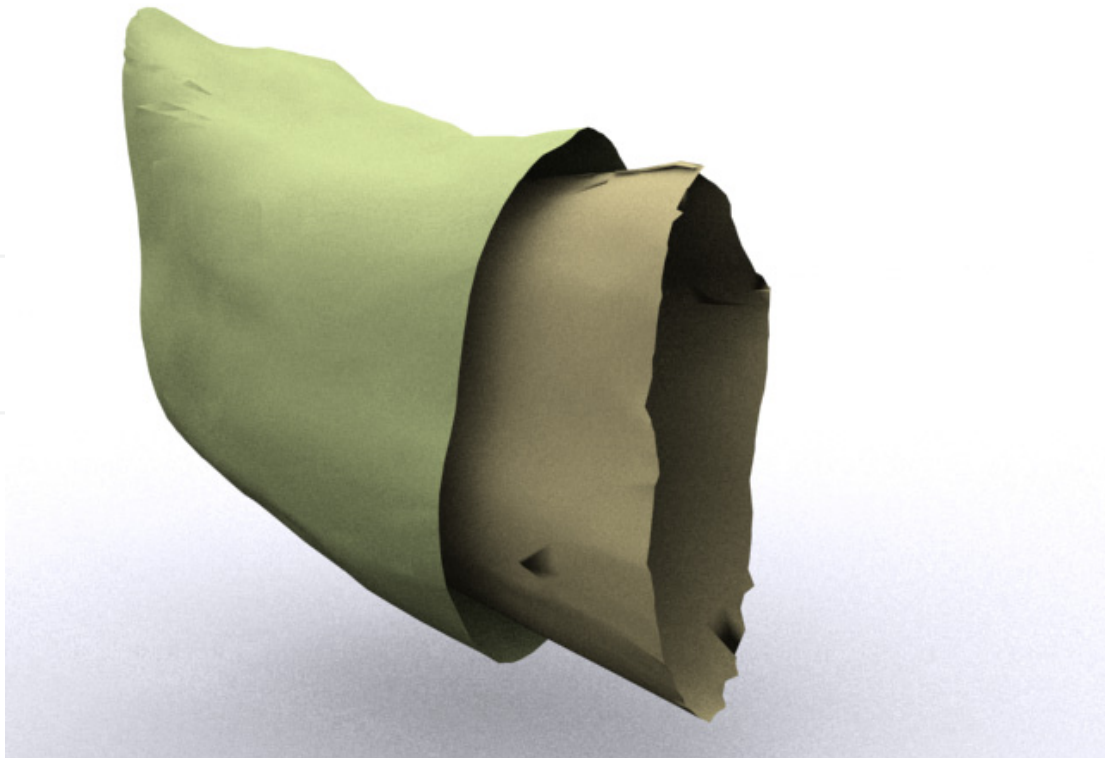


Figure 14. Region of interest is extracted by Boolean process

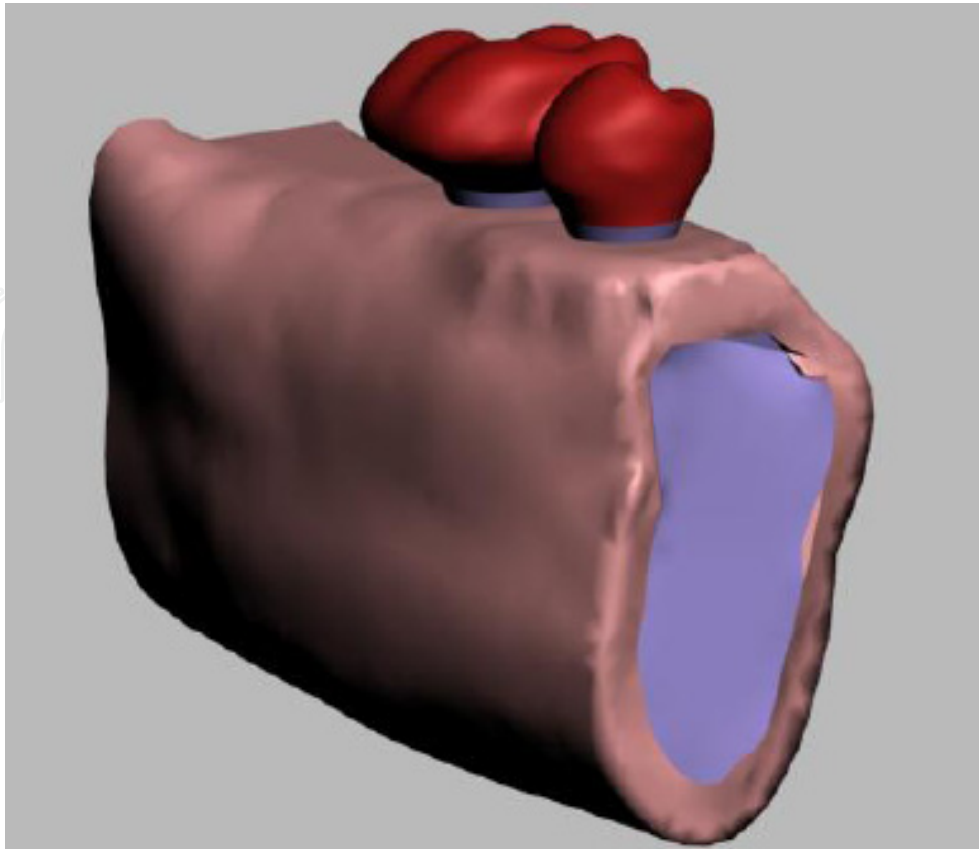


Figure 15. Part of the mandible modeled with superstructure, implant, and surrounding bone



Figure 16. The edentulous mandible obtained from a cadaver was scanned using a dental volumetric tomography device

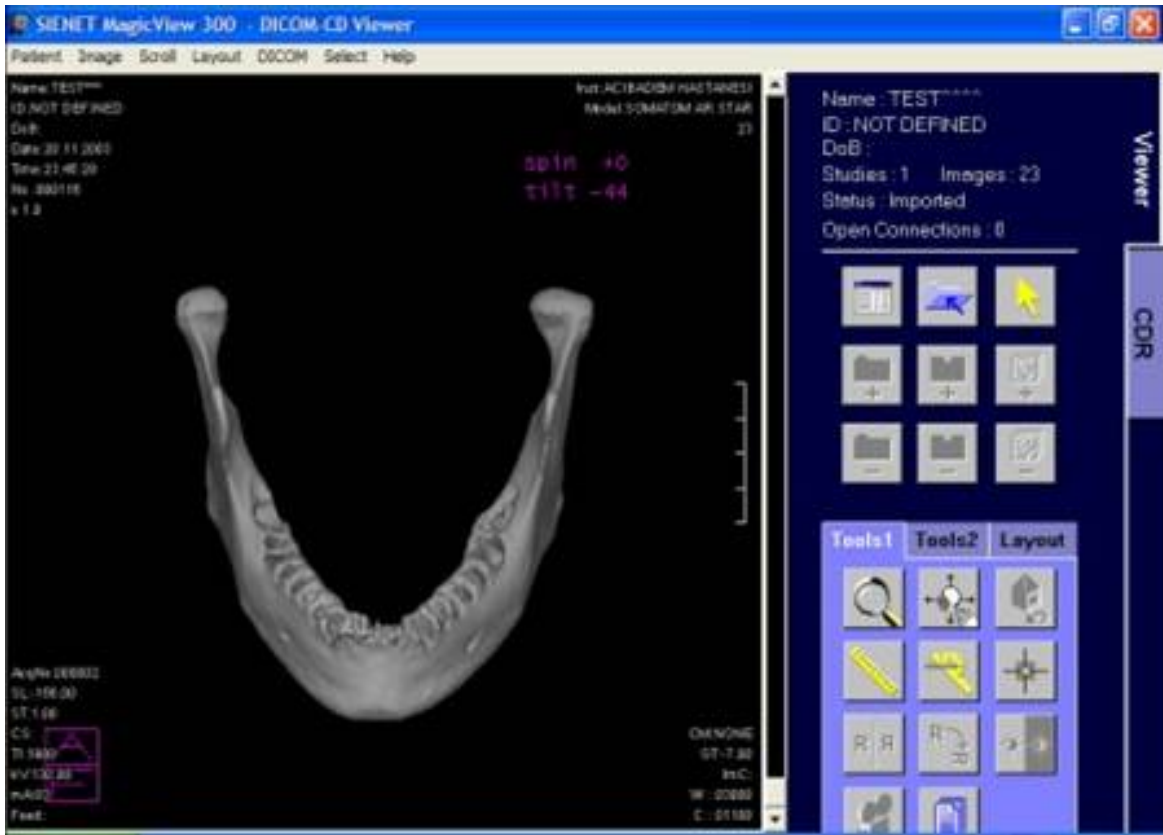


Figure 17. Volumetric data were reconstructed in 0.2 mm thick sections

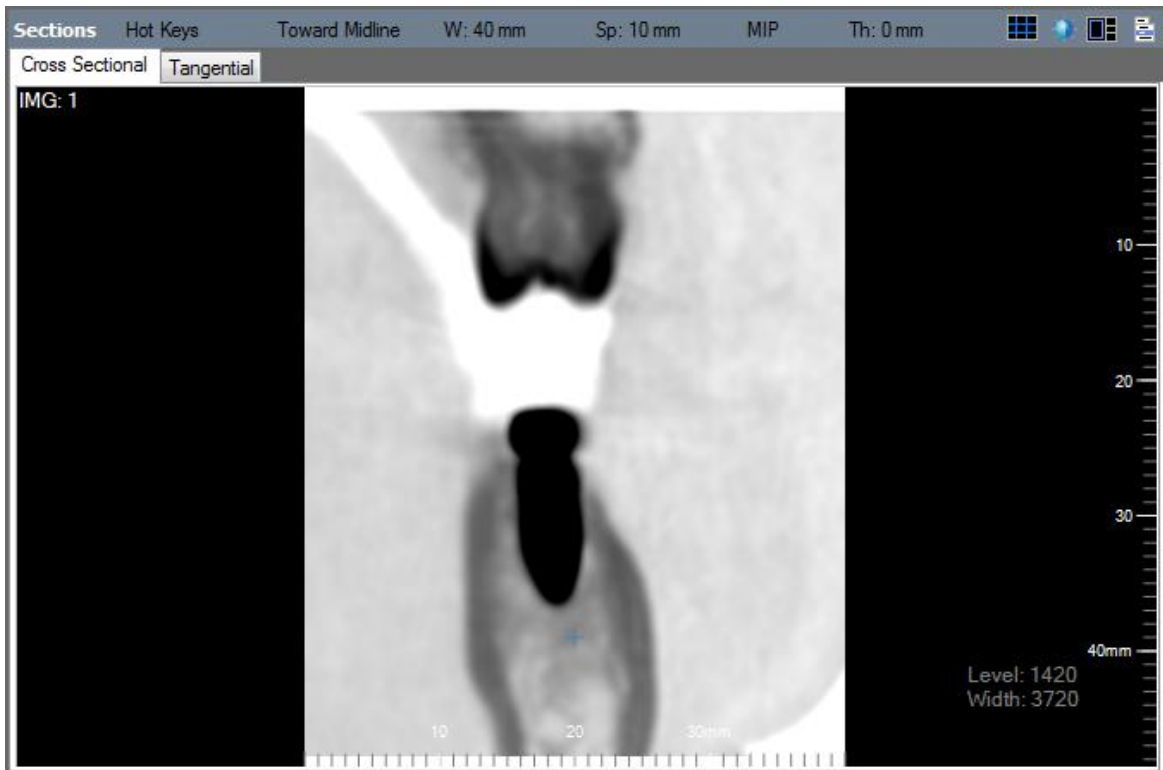


Figure 18. Bone density values can be measured according to gray scale using advanced 3-D radiographic techniques

3. Modelization of non-living structure (materials)

Non-living mechanical structures such as implants, abutments, and restorations can be simulated in detail and can substantially influence the calculated stress and strain values, similar to living structures. These materials can be digitally modeled in FEA studies using previously determined isotropic, transversely isotropic, orthotropic, and/or anisotropic properties [31]. In an isotropic material, the relevant material properties are the same in all directions, resulting in only 2 independent material constants, such as Young's modulus and Poisson's ratio [9,13,31]. Young's modulus (MPa), also known as the tensile modulus, is a quantity used to characterize materials and is a measure of the stiffness of an elastic material. Young's modulus is also called the elastic modulus or modulus of elasticity, because Young's modulus is the most commonly used elastic modulus [9,13,32,33]. When a sample object is stretched, Poisson's ratio is the ratio of the contraction or transverse strain (perpendicular to the applied load), to the extension or axial strain (in the direction of the applied load). When a material is compressed in 1 direction, it tends to expand in the other 2 directions perpendicular to the direction of compression. This phenomenon is called the Poisson effect. Poisson's ratio is a measure of the Poisson effect [9,13,32,33].

An anisotropic material has material properties that vary by direction [31]. Isotropic material properties are used in most FEA studies related to implant dentistry [18,25,34]. For instance, the material properties of living bone are anisotropic, and inhomogeneous. These properties of real bone greatly affect stress and strain patterns. In addition, bone density may differ among various regions of the same jaw and areas of differing densities may only be separated by millimeters. For simplification and to overcome computing difficulties, in most cases, the materials are modeled as homogenous, isotropic, and linearly elastic [35-39]. However, some studies have modeled the bone block using anisotropic properties (i.e., the material properties differ with respect to direction) [26]. The material properties of both living and non-living structures are chosen in accordance with the goal of the modeling exercise.

In some studies, implants are modeled using a screw design but without threads (Figure 19). This may simplify the computing process, but does not reflect the reality of implant geometry. If one or more study parameters are related to implant dimensions, there is little doubt that inclusion of implant threads in the model is quite important to the quality of the research. Most clinicians are interested in the magnitude and distribution of stress that may induce microdamage to the bone and result in crestal bone resorption; therefore, macro and micro threads are crucial in the modeling stage of an implant study. The implant thread design influences the induced bone stress around the implant, which contributes to crestal bone loss, and can jeopardize the maintenance of osseointegration [40-43]. In recent FEA studies, implant threads are modeled in detail (Figure 20,21). There are 2 ways to model implant and abutment materials. One way is to obtain all of the geometric information (e.g., length, diameter, macro-micro thread configuration) in

millimeters from the manufacturer. The second option is to scan implants and abutment materials and digitally reconstruct them. Efficient and realistic models can be obtained by using either option. In general, for the digital preparation of crown models, an anatomy atlas of the tooth can be used as a reference to calculate the form and both mesiodistal and buccolingual dimensions [44]. The prosthetic superstructure can be simulated according to various treatment protocols. Superstructure can also be modeled as a geometric figure, such as a simple rectangular shape, but this may interfere with the realism of the model (Figure 22).

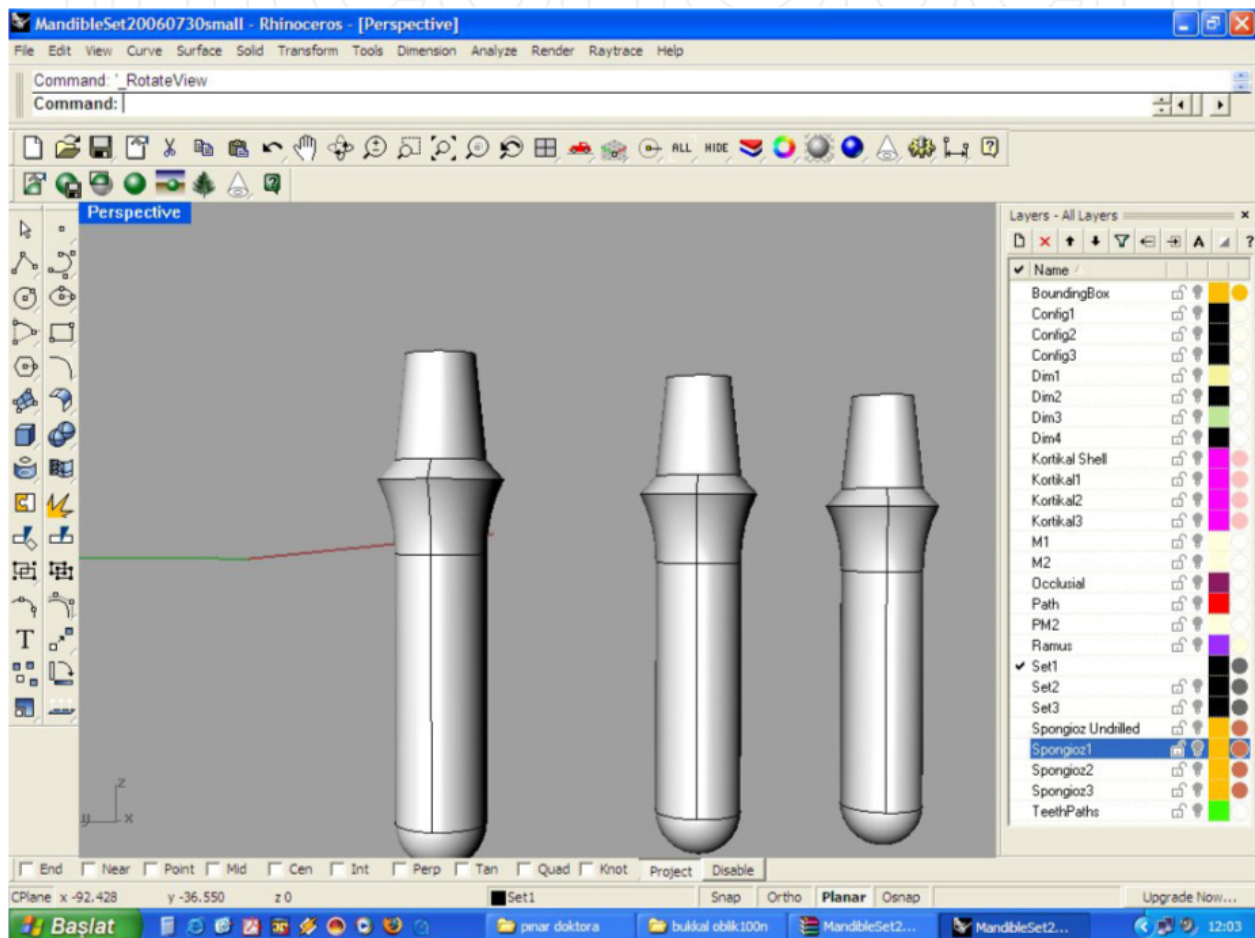


Figure 19. Implants are modeled without threads

In a previous study, the crown model was simulated as porcelain fused to metal restoration. To calculate the mesiodistal width of the second premolar and first molar, Wheeler's Atlas of Anatomical Natural Tooth Morphology was used (Figure 23) [44]. The atlas was used again for digital preparation of the crown models. Properties of chromium-cobalt alloy were used for the framework and feldspatic porcelain as used to simulate the second premolar and the first molar of a mandibular model. The metal thickness of the framework was 0.8 mm and the porcelain thickness was at least 2.0 mm. The thickness of porcelain changes with the creation of pits and trabeculae of the tooth surface. In most FEA studies, not only the cement thickness but also the interface between the materials is assumed to be 100%

bonded [9,18,25,31,34]. Implant, abutment, abutment screw, framework, and porcelain structures are considered to be a single unit (Figure 24). In contrast, there are some studies that use a contact condition between the abutment and implant set as a frictional coefficient [26]. In these studies, the corresponding material properties are used and modeled separately. Most studies also model the implant as rigidly anchored in the bone model along its entire interface and with total osseointegration. It is impossible to visualize these interface conditions in real life, but simplifications in interface conditions will inevitably result in considerable inaccuracy. The most common drawback of FEA from the clinical perspective is that many features that directly affect model accuracy, such as loading conditions, material properties, and interface conditions are neglected or ignored. In most cases, researchers neglect one or more features in their studies. Moreover, bias may result from interpretation of data obtained from an FEA study to that obtained from another. Within a single study, these simplifications are consistent for all the simulated models; therefore, the accuracy of the analysis from the stress distribution viewpoint is not affected, as long as the models are compared with each other in the same study [9,18,25,31,34].

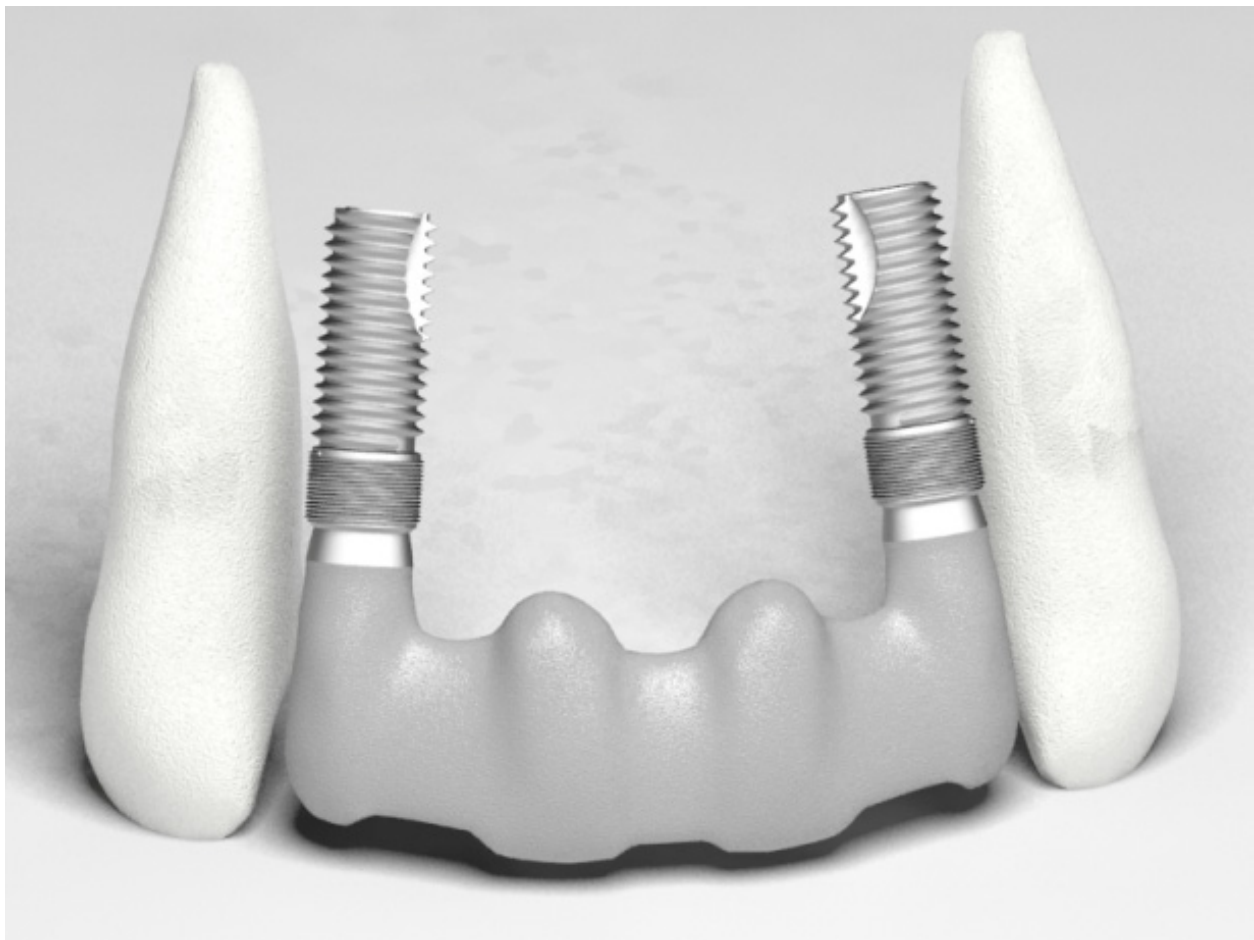


Figure 20. Implants are modeled with micro and macro threads



Figure 21. Implants are modeled with threads and abutments

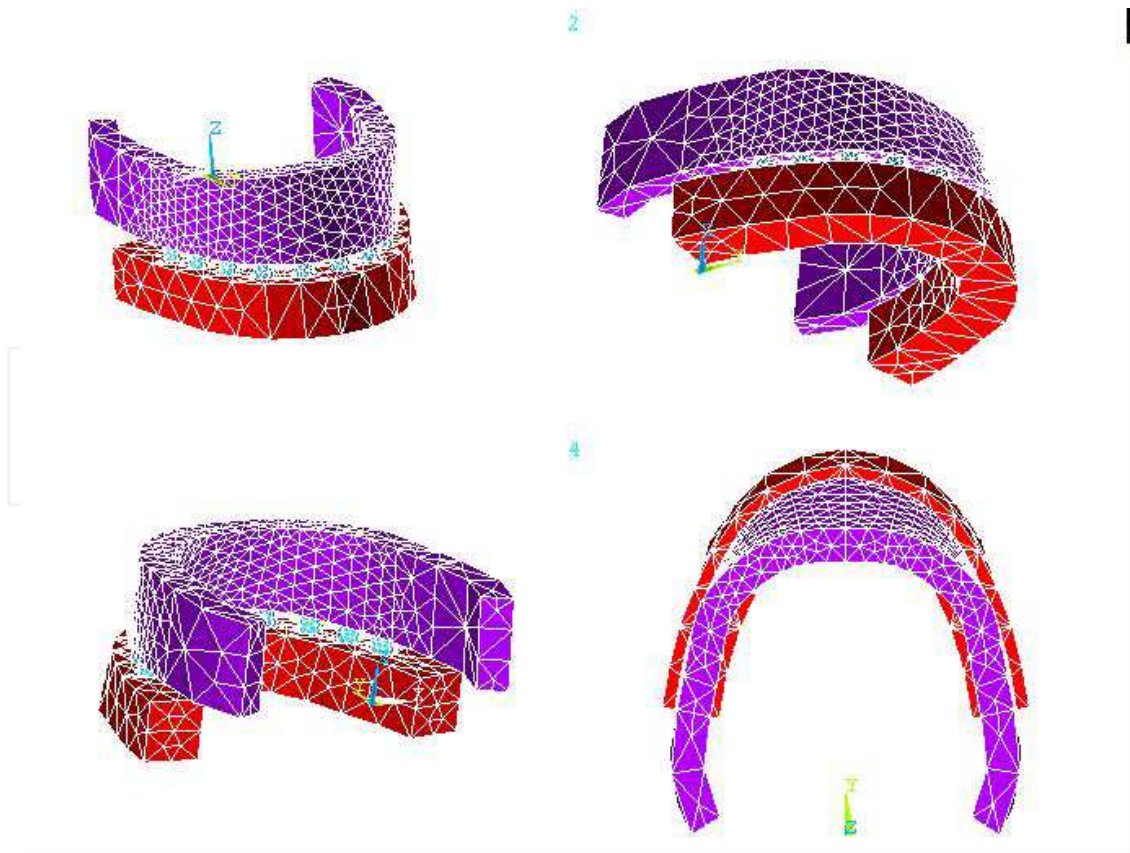


Figure 22. Superstructure modeled into a rectangular shape

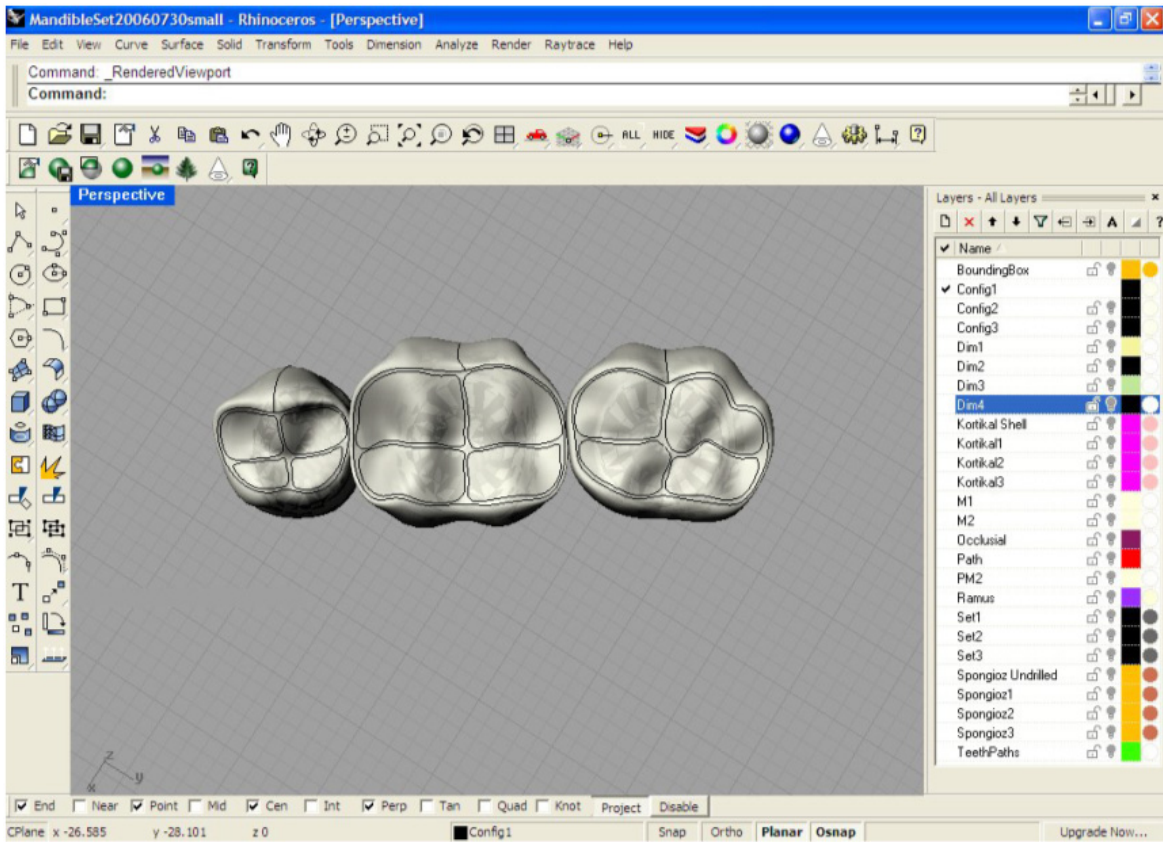


Figure 23. Digital preparation of crown models

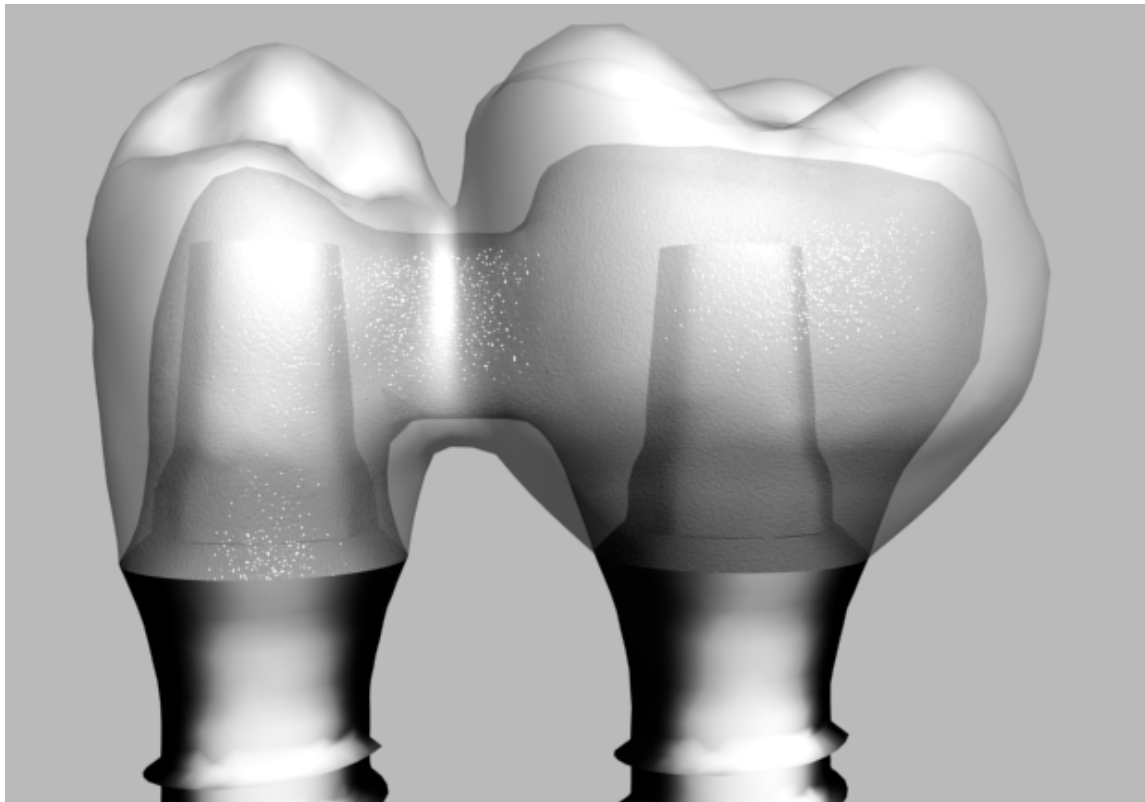


Figure 24. Implant, abutment, abutment screw, framework, and porcelain structure are modeled as 1 unit

Almost all of the elastic properties of selected living and non-living materials are available in the literature [9,25,31,34]. Young's modulus and Poisson's ratio are used in models to simulate reality as closely as possible. For example, alveolar bone (both cortical and cancellous portions), implant, abutment, metal framework, and porcelain can be included in the model properties.

4. Boundary conditions

A boundary condition is the application of force and constraint. The different ways to apply force and moment include a concentrated load (at a point or single node), force on a line or edge, a distributed load (force varying as an equation), bending moments, and torque [45]. In structural analysis, boundary conditions are applied to those regions of the model where the displacements and/or rotations are known. Such regions may be constrained to remain fixed (have zero displacement and/or rotation) during the simulation or may have specified, non-zero displacements and/or rotations. The directions in which motion is possible are called degrees of freedom (DOF). Zero-displacement constraints must be placed on some boundaries of the model to ensure an equilibrium solution. The constraints should be placed on nodes that are located far from the region of interest to prevent overlap of the stress or strain fields associated with reaction forces with the bone-implant interface. In maxillary FEA models, the nodes along the external lines of the cortical bone of the oral and nasopharyngeal cavities were fixed in all directions [46].

In most FEA studies that include models of the mandible, the boundary conditions are set as a fixed boundary [9]. Zhou et al. developed a more realistic 3-D mandibular FEA model from transversely scanned computed tomography imaging data. The functions of the muscles of mastication and the ligamentous and functional movements of the temporomandibular joints (TMJs) were simulated by means of cable elements and compressive gap elements, respectively. Using this mandibular FEA model, it was concluded that cable and gap elements could be used to set boundary conditions, improving the model mimicry and accuracy [47]. Chang et al. used a technique in which only half of the model was meshed, thus symmetry boundary conditions were prescribed at the nodes on the symmetry plane. Models were constrained in all directions at the nodes on the mesial and symmetrical distal bone surfaces [48]. Expanding the domain of the model can reduce the influence of inaccurate modeling of the boundary conditions. This, however, will be at the expense of computing and modeling time. Teixeira et al. concluded that in a 3-D mandibular model, modeling the mandible at distances greater than 4.2 mm mesial or distal from the implant did not result in any significant increase in FEA accuracy [49]. Use of infinite elements is another potential method for modeling boundary conditions [9].

5. Loading conditions

Marginal bone loss in the peri-implant region may be the result of excessive occlusal force [50]. Extensive investigations are needed to establish and understand the correlation

between marginal bone loss and occlusal forces; including the engineering principles, biomechanical relationships to living tissues, and the mechanical properties of bone surrounding implants [50]. In recent years, a greater amount of materials used for oral implantology are fabricated from titanium and titanium alloy. The Young's modulus of titanium is 5-10 times greater than that of cortical ridge bone surrounding implants [51]. The fundamental engineering principle, composite beam analysis, expresses the concept that when 2 materials of different Young's modulus are placed in direct contact with no intervening material and 1 material loaded, a stress contour will be described at the point where the 2 materials come into contact [52]. For oral implantology, these stress contours are of greater magnitude at the crestal bone. Therefore, the loading condition is another important part of FEA studies. Each component modelization stage contributes to the final analysis after loading. In other words, from the beginning to the end, all procedures and FEA stages add to the ability to extrapolate the results of bite forces surrounding the peri-implant region and prosthetic structures.

Bite forces may be defined as compressive, tensile, or shear forces. Compressive forces attempt to push materials toward each other. Tensile forces pull objects apart. Shear forces on implants cause sliding. The most detrimental forces that can increase the stress around the implant-bone interface and prosthetic assembly are tensile and shear forces. These forces tend to harm material integrity and cause stress build-up. In general, the implant-prosthetic unit can adapt to compressive forces [51]. In actual mastication, the repeated pattern of cyclic forces transmits loading via the restoration and dental implants to peri-implant bone. This generates different amounts of stress around the ridge and also in the prosthetic structure. However, randomized cyclic forces are not easily simulated. Therefore, most FEA studies use static axial and/or non-axial forces. Non-axial loads generate distinctive stress in the ridge especially in the cortical bone. The main remodeling differences between axial and non-axial loading are affected mostly by the horizontal component of the resultant stresses [53]. Therefore, for realistic simulation, combined oblique loads (axial and non-axial) are generally used. One study, comparing dynamic with static loading, revealed that dynamic loading resulted in greater stress levels than static loading [54]. Dynamic loading has consistently been found to have more osteogenic potential than static loading [55]. Sagat et al. investigated the influence of static force on peri-implant stress. In varied models, 100 N static forces were applied vertically and separately to the anterior and posterior parts of a bridge [18]. In another study, static forces of 100 N were applied at 30 degrees obliquely and separately to the lingual inclination of the buccal cusps of a crown (Figures 25,26) [25]. In another study, loading was simulated by applying an oblique load (vertical load of 100 N and horizontal load of 20 N) from buccal to palatal region at 4 different locations. An equivalent load of 200 N was applied in the vertical direction and 40 N in the buccal-palatal direction. The application point of the force was on the central and distal fossae of the crown [48]. Eskitascioglu et al. used an average occlusal force of 300 N applied to a missing second premolar implant-supported crown. Three-point vertical loads were applied to the tip of the buccal cusp (150 N) and distal fossa (150 N); the tip of the buccal cusp (100 N), distal fossa (100 N), and mesial fossa (100 N) [56].

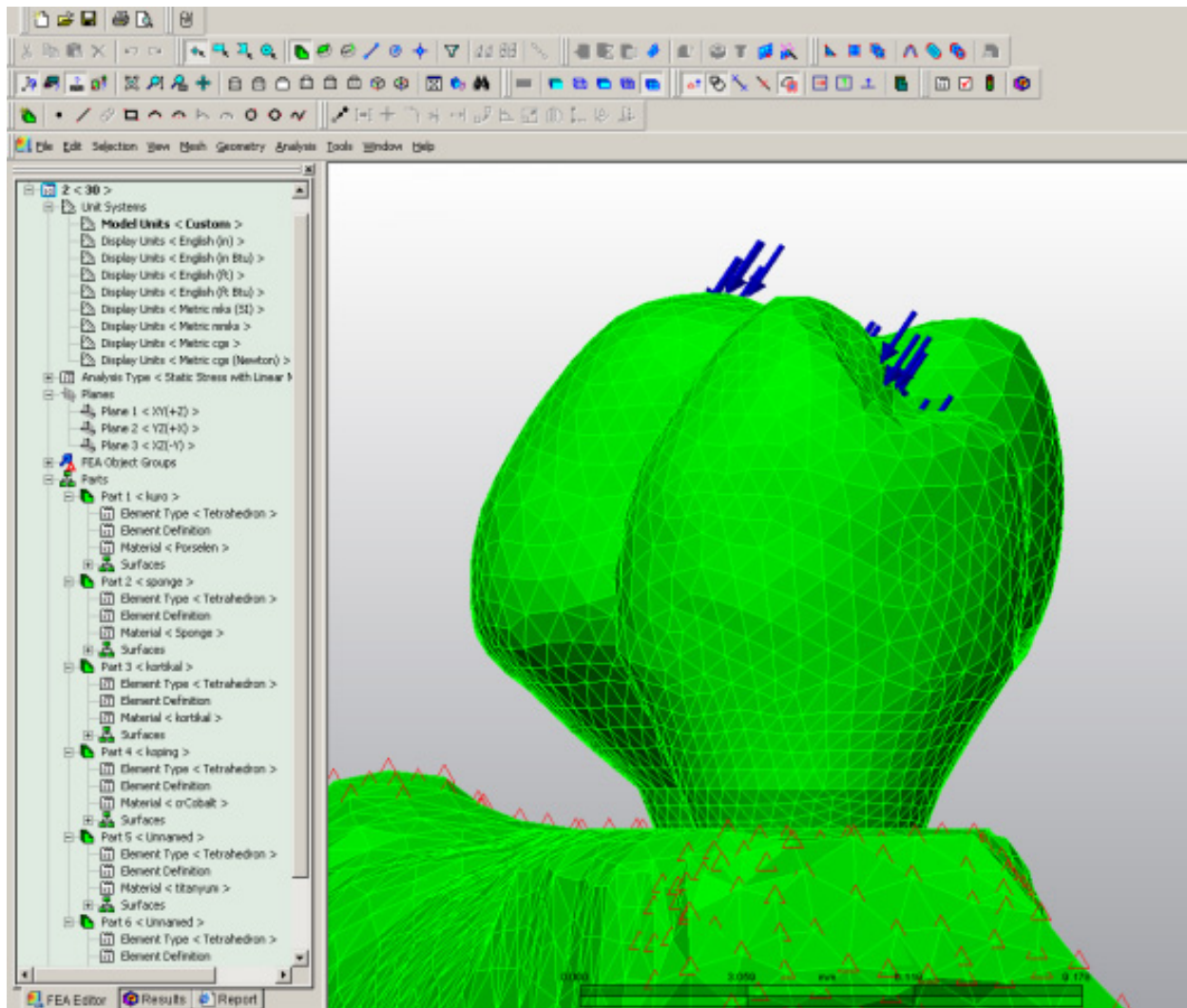


Figure 25. Static forces were applied at 30 degrees obliquely and separately to the lingual inclination of the buccal cusps of the crown

As mentioned before, oblique loads are more destructive to the peri-implant bone region and clinically disruptive to prosthetic structures. The magnitude of bite force may change according to age, sex, edentulism, parafunctional habits, and may differ from anterior to posterior in the same mouth [9,31]. In FEA literature, the locations for the application of bite force change according to the modeling of the restoration [9,31]. In advanced modeling studies, more realistic force application could be described including ridges of the cusp, labial or lingual surfaces of crown, occlusal surface, distal, and mesial fossa [9,27,31,57]. For realistic simulation of biting, loading forces should be applied to the restoration first, and then transmitted by the abutment to the implant and surrounding bone. Stress concentrations will then be generated, evaluated, and proper risk assessment will be considered.

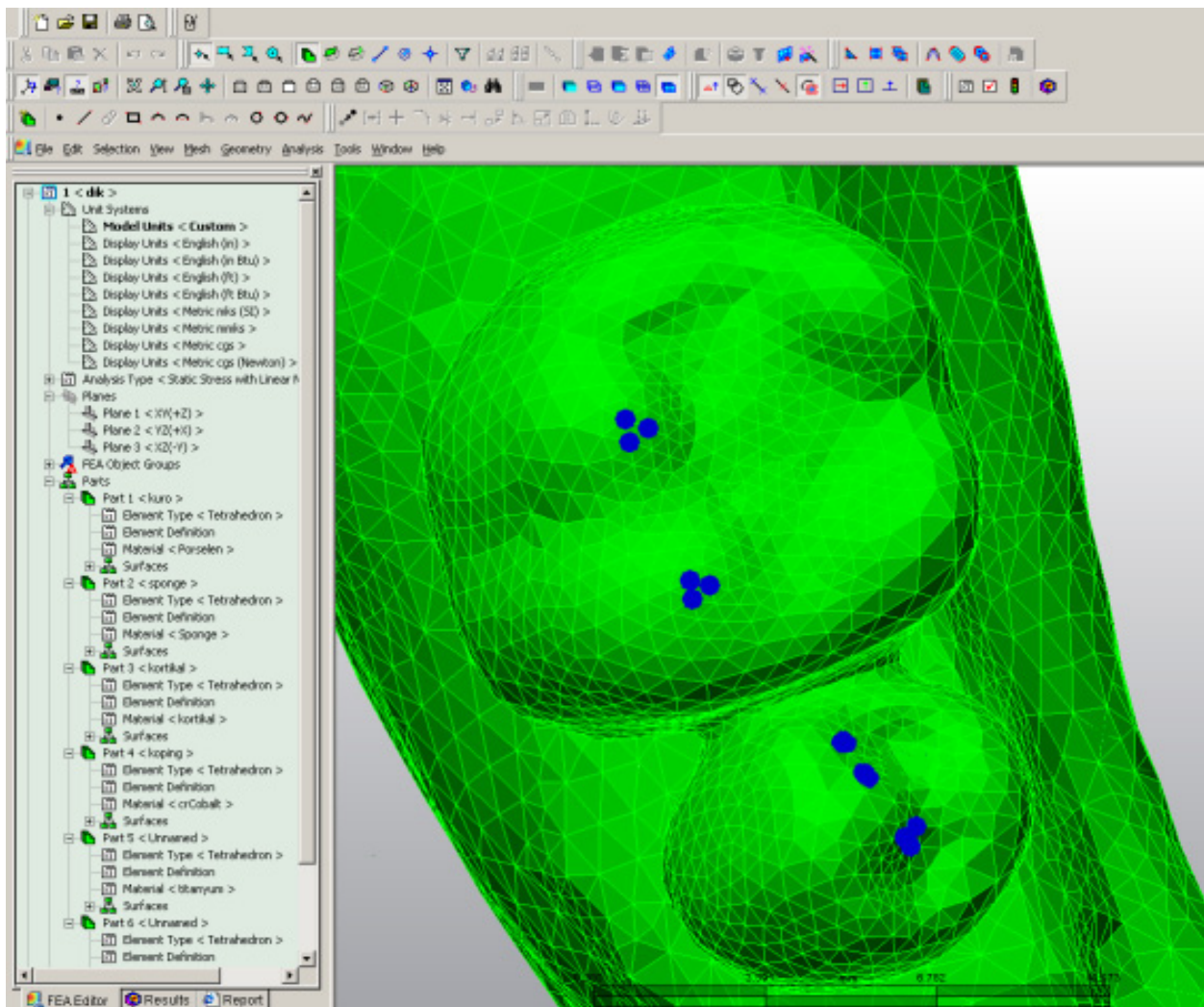


Figure 26. Force application to the region of restoration

6. Bone-implant interface

The 'osseointegration' concept was described as the direct contact between living bone and a loaded dental implant surface by Brånemark et al. [58]. The most widely used material for dental implant manufacture is pure titanium (Grade 4), titanium alloy (Grade 5), and rarely zirconia [59-62]. These materials have good biocompatibility with surrounding tissues, are resistant to deformation, and are easily manipulated for shaping as a natural tooth root forms by Computer Numerical Control (CNC) machines [59-62]. Titanium alloy has mechanical advantages over pure titanium in implant manufacture. With increases in grade number, the alloy becomes much stronger and more resistant to fractures or wearing of the components [59-62]. However biocompatibility may be reduced in inverse proportion the increase in grade number. Implant companies use Grade 4 or Grade 5 titanium for the implant body and generally choose Grade 5 titanium for implant abutment manufacture. Recently, to increase the strength of implant bodies, new materials have also been introduced into the market, such as roxolid (a zirconium and titanium combination) [63]. The use of zirconium and titanium combination material as an implant body has limited

scientific data and requires long-term investigations. Therefore, most FEA studies in the literature involve titanium and titanium alloys [9,18,24,31].

The most commonly used surfaces for implant bodies are rough surfaces. Different implant surface modifications (sandblasted, acid-etched, sandblasted and acid-etched, anodized, hydroxyapatite coatings, and plasma-sprayed) are proposed to change the characteristics of the surface from machined to rough, to increase the osteoblastic cell attachment level and also bone-implant contact (BIC) [64-68]. The influence of these surface modifications on BIC and cell attachment are still being investigated for a stronger osseointegration level between implant body and bone. Comparative studies show different BIC levels changing from 13% to 80% percent [69-79]. BIC values may change according to the jaw, placement of the region of the implant, healing time, implant design, and surface structure [64,69,70,72-74].

In most FEA studies, the bone-implant interface was assumed to be 100% bonded or completely osseointegrated [9,16,18,23,25]. As mentioned before, this is not proper modeling from a clinically realistic point of view. Cortical and cancellous bone also have different levels of BIC because of density and availability. Therefore, most studies use cortical bone of uniform thickness surrounding cancellous bone and proper material properties are chosen while modeling [9,16,18,23,25]. The degree of BIC distinctly affects the stress concentration value and distribution. In denser bone, there is less strain under loading compared with softer bone [80]. In some studies, BIC levels were assumed to be $\leq 100\%$ for simulation of soft bone or immediate loading scenarios [9,81]. Evaluation of peri-implant stress in FEA studies is important for obtaining accurate treatment methods in implant dentistry. Implant and surrounding bone should be stressed within a certain range for dynamic physiologic remodeling. If ideal functional forces are placed on a restoration, the surrounding bone can adapt to the stresses and increase its density [82]. Overload may cause high stresses at the crest of the ridge and result in bone resorption. The direct opposite of this result is disuse atrophy of bone due to too little stress in the peri-implant region. Maintenance of bone density and stabilization is a direct result of the ideal stress distribution [80]. According to Frost studies, strains in the range of 50-1500 microstrain stimulates cortical bone mass and represents the physiological range. Strain beyond this range may cause overload and strain less than this range may not stimulate bone enough [80,83-85]. Most FEA studies, evaluate the risk assessment according to high stress values [9,16,18,23,25]. In other words, the most favorable modeling has the lowest stress values, and in contrast, the most deleterious modeling has the highest stress values [9,16,18,23,25]. However intensely lower stress values may also cause bone resorption because of inadequate bone stimulation.

7. Evaluation of stress

Under bite force, localized stress occur at the prosthesis structure and bone. Stress is the magnitude of the internal forces acting within a deformable body. It is a measure of the average force per unit area of a surface within the body on which internal forces act. These internal forces appear as a response to external forces directed on the body [86-88]. Internal resistance after the application of the force applied on the body is not practically measurable. Therefore an easier process is to measure the applied force to a cross-sectional area. The dimension of stress is that of pressure, the Pascal (Pa), which is equivalent to 1

Newton (force) per square meter (unit area), that is N/m^2 . Stress is often reported in scientific publications as MPa. Stress is directly proportional to the force and inversely proportional to the area across which the force is applied. It is important to determine the area across which any force is applied. For example, the surface area of the occlusal pit restoration less than 4 mm. For this reason, the magnitude of stress in many restorations reaches hundreds of MPa [9,16,18,23,25,51].

When the force is applied to mass, a deformation occurs as a result of this force. A strain is a normalized measure of deformation representing the displacement between particles in the body relative to a reference length [9,16,18,23,25,51,86,87,88]. There is no measurement unit of strain. Strain can be defined as the deformation ratio of the original length.

In FEA studies related to implant dentistry, frequently von Mises stress (equivalent tensile stress), minimum principal, and maximum principal are used to evaluate the effect of loading forces on the peri-implant region or prosthesis structure [9,16,18,23,25,89]. When a specific force is applied to the body, von Mises stress is the criterion used to determine the strain energy principles. Loading forces affecting the object can be evaluated 2 or 3 dimensionally. There are 3 "Principal Stresses" that can be calculated at any point, acting in the x, y, and z directions. The von Mises criteria refer to a formula for combining these 3 stresses into an equivalent stress, which is then compared to the yield stress of the material [25,90]. The major stress values are formed when all the components of the shear are zero. When an element is in this position, the normal stresses are called principal stresses. Principal stresses are classified as maximum, intermediate, and minimum principal stresses. The maximum principal stress is a positive value indicating the highest tension. The intermediate principal stress represents intermediate values. The minimum principal stress is a negative value indicating the highest compression [9,16,18,23,25,89]. If the data obtained from the analysis are positive values, then they are considered tensile stresses, negative values indicate compression-type strains.

Frequently, different color figures are used according to the amount of stress around peri-implant regions and prosthetic structures (Figure 27). Stresses on each model are evaluated according to the stress values from low to high. In other words, the most favorable model has the lowest stress values, and in contrast, the most deleterious model has the highest stress values (Figure 28).

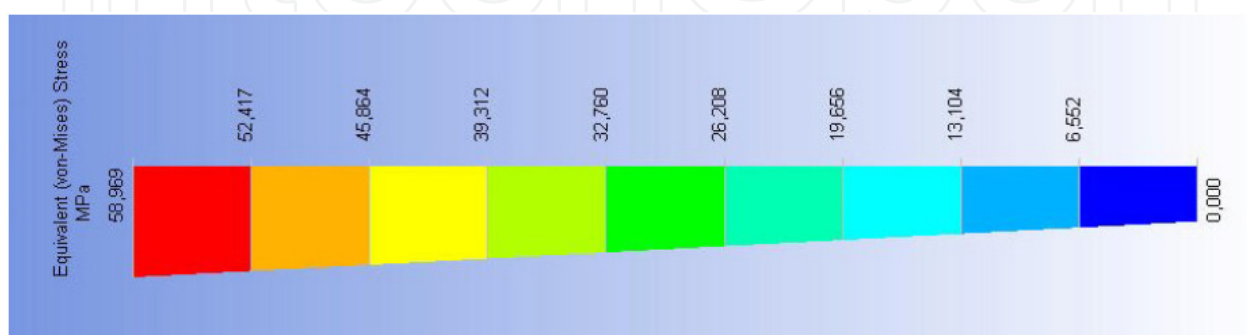


Figure 27. Different colors indicate the amount of stress around the peri-implant region and prosthetic structure

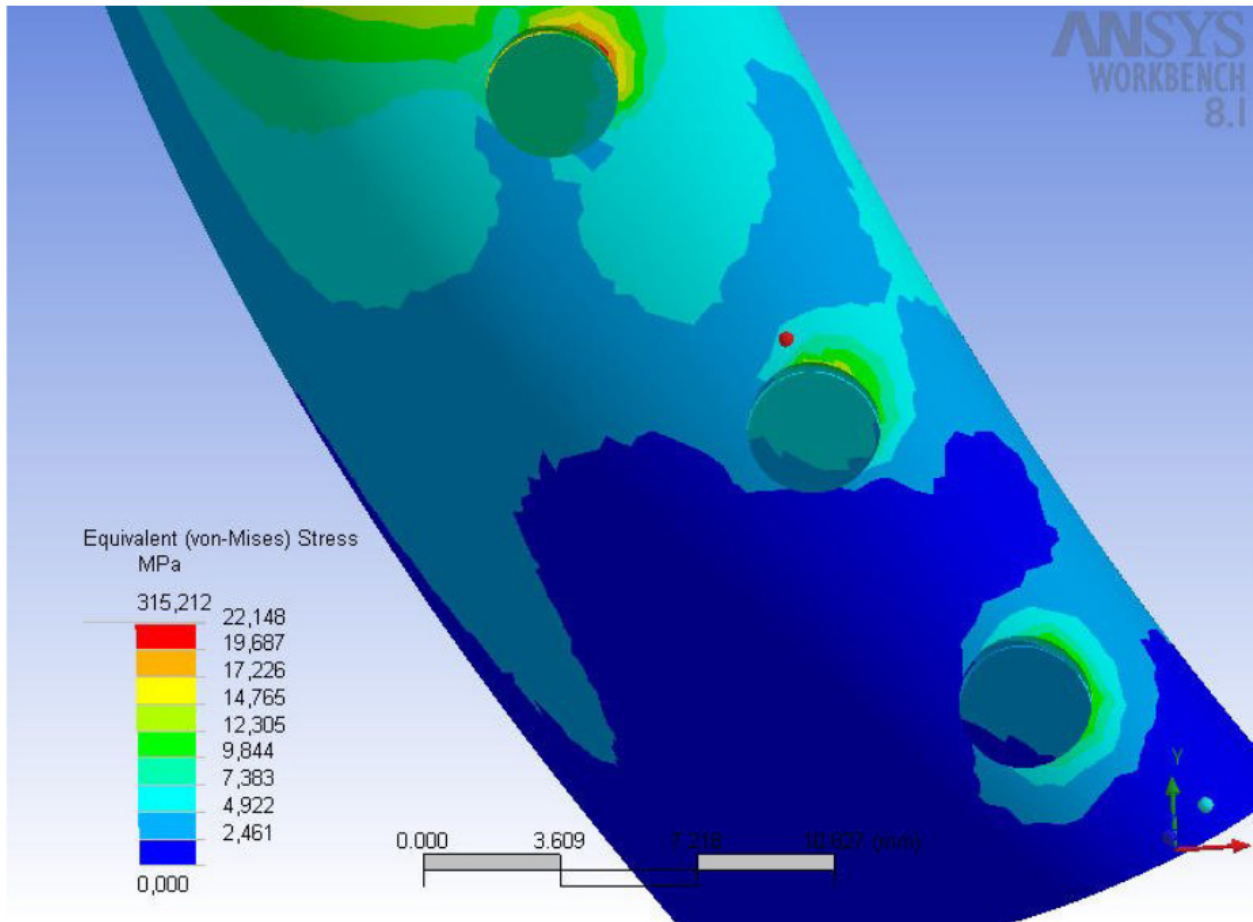


Figure 28. High and low stress values depicted in different colors in the models

In a previous study evaluating stress distribution, maximum von Mises (equivalent) stresses on each model are depicted around peri-implant region [18]. Eskitascioglu et al. evaluated maximum stresses (maximum von Mises) within the cortical bone surrounding the implant, framework of restoration, and occlusal surface material [56]. In a previous study, the FE model was used to calculate not only von Mises stress but also the principal stress. Authors explained their approach for this debate as follows: bone can sometimes be classified as brittle material; therefore, the principal stress was also implemented to evaluate the situation of cortical bone around implants [48].

8. Good FEA research development in implant dentistry

This section is provided for clinicians and researchers who want to plan FEA studies related to implant dentistry and to provide a brief summary of research methodology.

1. *Planning a scenario:* The most important part of an FEA study is planning a unique model of treatment. There are countless FEA studies in the implantology literature; therefore, at the beginning of the study, it is highly recommended that you evaluate the available literature on your subject. Implant technology is improving rapidly. There is currently no perfect dental implant design or implant-abutment connection.

Implant manufacturers change their macro design and connections according to perceived clinical benefits. The aim of these improvements are less bone resorption around peri-implant regions, less micromotion at abutments, better loading distributions at dental implant structures, and good conical sealing. These properties are commonly related to biomechanics and should be investigated not only with clinical studies but also with FEA studies. All novel designs of implants or materials can be subject to investigation and can be compared with traditional structures. Another way of instituting FEA study is investigating treatment alternatives. New and old treatment modeling can be compared, limitations, and application areas can be better understood.

2. *Computer stage*: This is the second part of FEA study. Generally clinicians have limited knowledge about modeling in computers and need help from computer engineers. It will be very wise to collaborate with friends at that field. Without a collaborator in computer engineering, too much time will be spent learning how to prepare models and developing the appropriate knowledge for the computational techniques necessary for model implementation. The clinician should manage the study and provide direction to the engineer. If the engineer does not have knowledge of the field of implant dentistry, seminars can be given to introduce the basic concepts of implantology. The seminars can include concepts such as indications for dental implants, dental implant parts, bone physiology, biting forces, connections of implants with bone, and the logic of implantology. As mentioned before, the shape of the materials can be scanned and converted digitally. Dental volumetric or computed tomography are good alternatives to scan and build bone structures. Devices used for routine treatments, can be found easily and are not expensive. For modeling of implant parts and superstructure, there are many sources, including manufacturers guidelines, scanning (advanced engineering 3-D scanning needed), and tooth atlas. The clinician should make every effort to maintain contact with their colleagues to allow frequent and efficient model evaluation and adaptation. The number of elements and nodes, can be increased to achieve more detailed modeling. However, this may be quiet time-consuming and may implicate computing complications. Therefore, the engineer should clearly understand the aim of the research. Boundaries, limitations can be applied at modeling and element numbers can be increased only at the region of interest. These applications should not directly affect the results achieved. In the literature there are many software packages available for FEA study. The computer engineer can aid clinicians in choosing the appropriate software package for the specific application. In general, von Mises (equivalent stress), minimum, and maximum principal stress values are being used in FEA studies related to implant dentistry. These stress values are evaluated from low to high, and assessments are made according to these values. Higher values are considered more destructive and involve greater risk than low values. The most common material properties used in FEA studies of implant dentistry are listed in Table 1 [9,27,48,56,57,91-109].

Material	Young Modulus (MPa)	Poisson Ratio	Ref. No.
<i>Ti-6Al-4V</i>	110,000	0.35	27, 48, 57, 91
	110,000	0.33	
	100,000	0.35	
<i>Pure titanium</i>	117,000	0.3	9, 92, 93
<i>Type 3 gold alloy</i>	90,000	0.3	48, 94, 95
	100,000	0.3	
	80,000	0.33	
<i>Cortical bone</i>	13,700	0.3	27, 56, 57, 96, 97
	13,400	0.3	
	10,000	0.3	
	15,000	0.3	
<i>Trabecular bone</i>	1,370	0.3	27, 56, 57, 98, 99
	1,500	0.3	
	1,370	0.31	
	150,000	0.3	
	250,000	0.3	
	790,000	0.3	
<i>Periodontal ligament</i>	170	0.45	108
<i>Ni-Cr alloy</i>	204,000	0.3	108
<i>Dentin</i>	18,600	0.31	108
<i>Porcelain</i>	66,900	0.29	31, 48, 109
	67,700	0.28	
<i>Co-Cr alloy</i>	218,000	0.33	56
<i>Feldspathic porcelain</i>	82,800	0.35	56
<i>Enamel</i>	41,400	0.3	97, 100-102
	46,890	0.3	
	82,500	0.33	
	84,000	0.33	
<i>Mucosa</i>	10	0.40	103
<i>Ag-Pd alloy</i>	95,000	0.33	109
	80,000	0.33	
<i>Resin</i>	2,700	0.35	31
<i>Resin composite</i>	7,000	0.2	31
<i>Gold alloy screw</i>	100,000	0.3	93
<i>Titanium abutment</i>	110,000	0.28	109
<i>Titanium abutment screw</i>	110,000	0.28	109
<i>Zirconia implant</i>	200,000	0.31	105, 107
<i>Zirconia abutment</i>	200,000	0.31	105, 107
<i>Zirconia core</i>	200,000	0.31	105, 107
<i>Zirconia veneer</i>	80,000	0.265	106, 107

Table 1. Material properties used in finite element analysis studies of implant dentistry

3. *Interpretation of results:* FEA studies have several advantages over clinical, pre-clinical, and in vitro studies. Most importantly, patients will not be harmed by the application of new materials and treatment modalities that have not been previously tested. Animals will not suffer from these biomechanical studies. However, clinicians should be aware that all of these applications are being performed on a computer, with critical limitations and assumptions that will clearly affect the applicability of the results to a real scenario. In the application of FEA studies, the most common drawback is overemphasis of the results. Simplifications are made for all simulated models; therefore, the models should be compared with each other within the same study. Other studies may use varied material properties and different planning scenarios. Confirming the FEA results with mechanical tests, conventional clinical model analysis, and preclinical tests are essential. It should not be forgotten that FEA studies are helpful for clinical trials but the results achieved from these studies are not valuable as clinical study results. However, before beginning biomechanical clinical trials, it will be wise to refer to FEA studies.

9. Conclusion

FEA is a numerical stress analysis technique and is extensively used in implant dentistry to evaluate the risk factors from a biomechanical point of view. Simplifications and assumptions are the limitations of FEA studies. Although advanced computer technology is used to obtain results from simulated models, many factors affecting clinical features such as implant macro and micro design, material properties, loading conditions, and boundary conditions are neglected or ignored. Therefore, correlating FEA results with preclinical and long-term clinical studies may help to validate research models.

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