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Finite Element Comparison of Dental Implants: Static and Dynamic Analysis

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Abstract

The purpose of this study is to analyze and compare the implant-bone interface stresses using three-dimensional finite element models of a partially osseointegrated implant with platform and a conventional matching-diameter implant platform and abutment in the anterior mandible. In order to reach this purpose the mechanical properties of both peri-implantar regions will be modelled as isotropic and resonance frequency analysis will be used to validate their values. Material properties of compact and cancellous bone are modelled as transversely isotropic and isotropic, respectively. The osseointegration process is simulated by changing the Young`s modulus of the implant-bone interfacial tissue. The numerical resonance frequency is validated by means of clinical measurement assessed using a *resonance frequency analysis* (RFA) device that uses magnetic technology (Ostell). For static analysis one model is simulated using a 4.3 mm diameter abutment connection and the other uses a narrower 3.7 mm diameter abutment connection, simulating a platform-switching configuration. The loading scenario consists of an oblique (200 N vertical and 40 N horizontal) occlusal load and, perfect bonding is assumed at all interfaces. The results suggest that the use of resonance frequency analysis using an electromagnetic pulse is sensitive and hence can be used to qualify the implant stability during the osseointegration process.

Keywords: platform switching, dental implants, finite element.

1 Introduction

Histologic and radiographic observations suggest that a biological dimension of hard and soft tissues around dental implants extends apically from the implant-abutment interface [1]. The stress and strain distribution developed on this surrounding bone is the main mechanical reason for failure of many implants. Thus, the establishment and maintenance of osseointegration requires implant stability [2, 3] and, therefore clinical measurements of implant stability and osseointegration are the input to assess the clinical success. The implant stability is detrmined by the implant boundary condition and, osseointegration is characterized by clinical stability of a functionally loaded implant [4]. Resonance frequency analysis (RFA) has been established as a non-invasive and non-destructive quantitative measurement of implant integration by assessing changes in implant stability over time [5, 6]. The changes on RFA measured values, during a healing process, are quantitatively related with the increase/decrease in the stiffness of implant-bone interface. Two commercial RFA devices are available to detect implant stability: the first RFA device is an electronic device that uses a direct connection between the transducer and the resonance frequency analyzer [7], and the second RFA device uses magnetic frequencies between a magnetic peg and the resonance frequency analyzer [8, 9]. The results are expressed as an implant stability quotient (ISQ), which represents a standardized unit of stability [5, 6]. Several computational studies have been performed on RFA for dental implants [10-16]. From all these woks, the main conclusion is that computational resonance frequencies are dependent on several parameters, such as dental implant anchorage, implant diameter, bone quality, length of the embedded implant, boundary conditions and orientation of the transducer. In the work now presented, clinical measurements are obtained by using a magnetic RFA device (Osstell) and, the clinical results, are compared with the computational RFA, obtained from 3D finite element models of a dental implant placed at a lower Jaw and, assuming an interface with some level of osseointegration. Using this procedure it will be possible to select mechanical properties of peri-implantar regions to perform finite element studies.

In 1991, Implant Innovations introduced wide-diameter implants with matching wide-diameter platforms. When introduced, however, matching-diameter prosthetic components were not available, and many of the early 5.0- and 6.0-mm-wide implants received "standard"-diameter (4.1-mm) healing abutments and were restored with "standard"-diameter (4.1-mm) prosthetic components [1]. Long-term radiographic follow-up of these "platform-switched" (PS) restored wide-diameter dental implants have demonstrated a smaller than expected vertical change in the crestal bone height around these implants than is typically observed around implants restored conventionally with prosthetic components of matching diameters. The biomechanical rationale for this phenomena is addressed by Yashinobu et al. [17] in a 3D finite element analysis, were they have concluded that PS abutment configuration has the biomechanical advantage of shifting the stress concentration area away from the cervical bone–implant interface. These models however were very simplified and did not consider the clinical situation on how marginal bone stability can be influenced by different bone qualities around implants restored with Standard or PS abutments. In this paper a special attention will be given to these clinical situations and the finite element models produced have the advantage of being validated with clinical data.

2 Finite Element Model

2.1 Mandible geometry

In this work, a virtual model of a lower jaw on the region of tooth 36 (FDI) is developed from a computerized tomography (CT) data. With a help of new technologies it was possible to create a 3D geometric model of a human mandible and, in order to account for average dimensions of a human jaw, the virtual model is simplified according the data from craniometric measurements of human edentulous mandibles [18]. However, because the bone quality of a human mandible is usually classified into four distinct classes [19], the geometry of a human mandible that is adopted is presented on Figure 1.

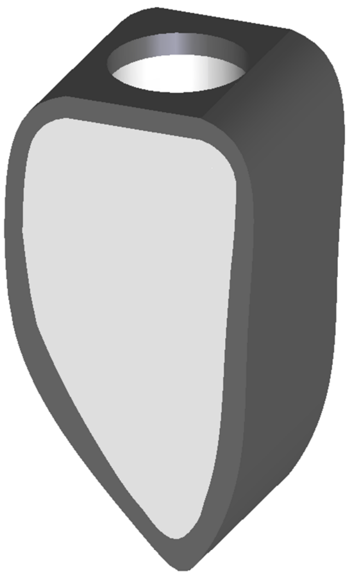

Figure 1. Virtual solid model of lower jaw, third class mandible.

On the third class mandible the thick layer of cortical bone (1 mm) surrounds less dense cancellous bone.

2.2 Implant and abutments geometries

The implant geometry adopted in this work is from the Camelog brand. The implant geometric model is imported to the CAD software and virtually placed in the lower jaw. Figure 2 shows the Camelog implant and abutment geometries used.

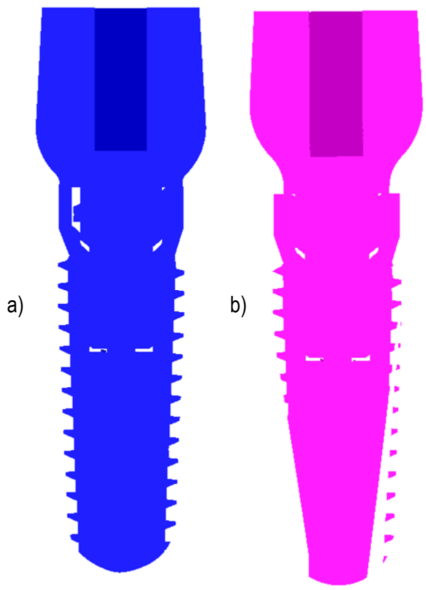

Figure 2. Virtual solid model of a Camelog implant and abutment model: (a) standard geometry; (b) platform switching geometry.

The abutment is placed on the implant geometry and an artificial crown is reconstructed over it. Standard abutment geometries are those in which implant and abutment diameters have coincident values, see Figure 2 a). The concept of platform switching is associated with an implant/abutment geometry in which a smaller-diameter abutment is mounted on a larger-diameter implant collar, as presented on Figure 2 b).

2.2 Material properties, load and boundary conditions

The material properties of titanium are listed in Table 1 and, are used to FEA of Camelog implants. The osseointegration is related with the long-term behaviour of the interface between the implant and the surrounding bone. In fact, the term osseointegration has been used to evaluate the structural and functional connection between living bone and the surface of a load bearing implant [16]. The implant stability and, therefore, osseointegration may increase by bone formation and by bone remodelling at the implant-bone [14]. Thus, the mechanical properties of the bone that surrounds the implant may be modelled with different values depending on the implant stability. In this work, the virtual model of the lower jaw comprises a bone-implant circular cross-section with a 1mm width that surrounds the implant. This region includes compact and cancellous bone with the mechanical properties presented on Table 1. The bone-implant interface of type f show mechanical properties that, typically, are associated with an interface showing some degree of osseointegration during the healing process.

Material	E [GPa]	ν	G [GPa]	ρ [g/cm^3]
Titanium	110	0.3	-	4.5
aluminium	70	0.33	-	2.7
Niobio	210	0.3	-	7
Compact bone	E_{11}= 15 $E_{22} = E_{33}$ = 11.1	$\nu_{12} = \nu_{13}$ = ν_{23} = 0.3	G_{23}= 7.8 $G_{12} = G_{13}$ = 4.2	2
Cancellous bone	0.6	0.3	-	1
bone-implant Interface				
compact cancellous	2 0.1	0.3	-	

Table 1. Material properties used on the static and dynamic anaysis

Resonance frequency analysis (RFA) provides a non-invasive assessment of implant stability. Some commercially RFA devices are available to dentist for detecting implant stability. The Ostell Mentor device has a transducer, a metallic rod with a magnet on top, which is screwed onto an implant or an abutment. The magnet is excited by a magnetic pulse from a wireless probe. After excitation, the peg vibrates freely and the magnet induces an electric voltage in the probe coil. That voltage is the measurement signal sampled by the resonance frequency analyzer. The results are expressed as an implant stability quotient (ISQ), which represents a standardized unit of stability [6]. The mechanical properties of peg and magnet materials are listed on Table 1 and, are denoted by aluminium and niobio, respectively.

Boundary conditions for static and dynamic analysis are prescribed at the nodes placed on the mesial and distal bone surfaces. Models are constrained in such away that the three nodal degrees of freedom of each node placed on the mesial and distal bone interfaces are restrained.

Some numerical results related with the platform switching geometries in implant dentistry have been presented using specific load conditions [20], thus the loading is simulated by applying an oblique load from buccal to palatal direction, which is represented by a load of 200 N in the axial direction and a load of 40 N in the buccal-palatal direction.

3 Finite Element Analysis

3.1 Modal analysis

Dynamic analysis is carried out by imposing the boundary conditions defined on section 2.2. To use the finite element method for assessing the ability of frequency analysis in detecting the degree of oral implant osseointegration, the smartpeg geometric model needs to be created and placed on the implant, Figure 3 shows the smartpeg geometry.

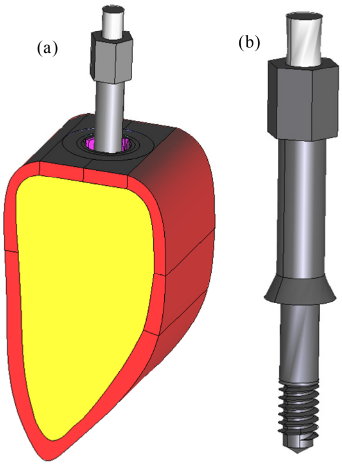

Figure 3. Dynamic model: (a) smartpeg placed on a third class mandible; (b) smartpeg geometry.

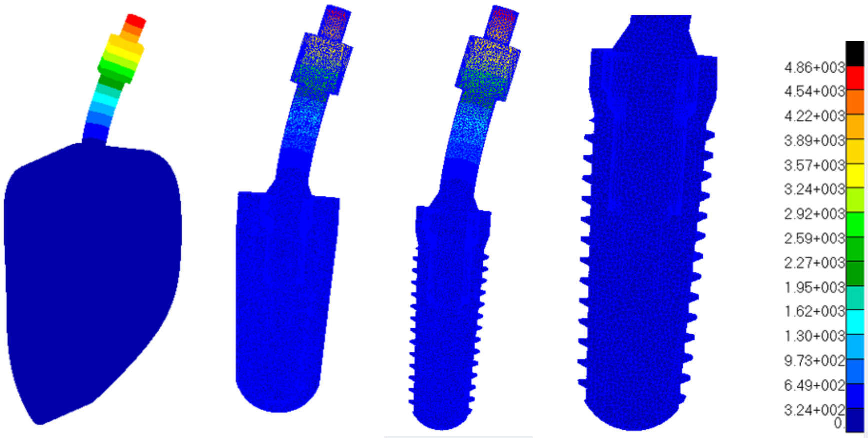

Figure 4. first vibration mode of the smartpeg placed at the implant.

The smartpeg is constituted by an aluminium base that is screwed on the implant like a prosthetic abutment and by a niobio cylinder that is fixed in the aluminium base. The first resonance frequency obtained by the finite element model is of 6964.8 (78), in which unity dimension are Hz and (ISQ), respectively. The first vibration mode is always of the same type and, is depicted on Figure 4. Clinical measurement of the first frequency is of 79 ISQ.

3.2 Static analysis

Static analysis is carried out by imposing loading and boundary conditions defined on section 2.2. Figure 5 shows the von Mises stress distribution in compact bone. The apparent stress concentration, on both models, is observed in a region between buccal and palatal sides, but closest to the buccal side than to the palatal side. The maximum von Mises stress in the conventional model is of 51.5 MPa and in the platform switching model is of 40.8 MPa. The maximum stress in both models is founded on the crestal bone region. Figure 6 shows the von Mises stress distribution at cancellous bone. The maximum von Mises stress in the conventional model is of 49.2 MPa and in the platform switching model is of 45.9 MPa. The maximum stress in both models is founded on the bone that is in contact with the root head of the implant.

Comparing Figures 5 (a) and 6 (a) is possible to conclude that in the conventional model the value of the maximum von Mises stress at cancellous bone (49.2 MPa) is smaller than that observed on the cortical bone (51.5 MPa) while comparing Figures 5 (b) and 6 (b) is possible to conclude that in the platform switching model the value of the maximum von Mises stress at the cancellous bone (45.9 MPa) is higher than that on the cortical bone (40.8 MPa). Moreover, comparing figures 6 (a) and 6 (b), the maxiumum von Mises stress in cancellous bone is smaller in the platform switching model than in the conventional model.

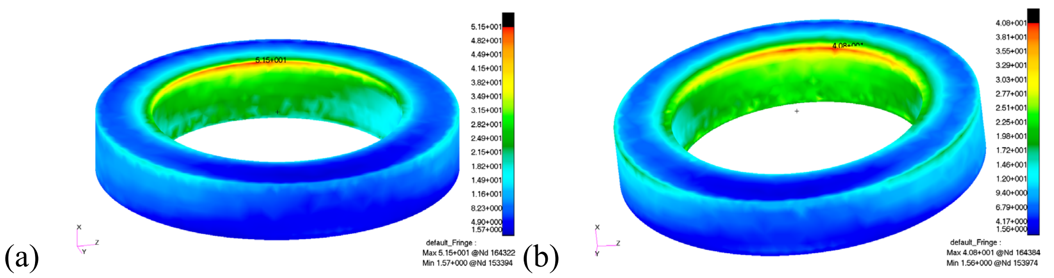

Figure 5. Von Mises stress distribution in compact bone.

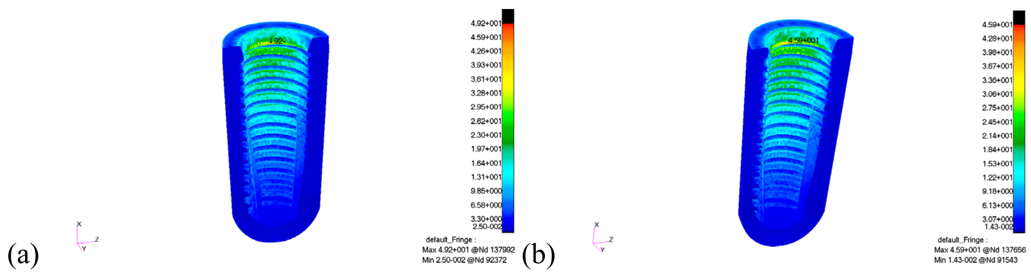

Figure 6. Von Mises stress distribution in cancellous bone.

Figures 7 and 8 show the bone-implant principal stresses at compact and cancellous bone, for conventional and platform switching models, respectively. The maximum decrease on the stress levels occurred on the minimum value of first principal stress in which a variation of 51.5% is observed. Nevertheless, the peri-implant cancellous bone shows similar levels for compression and tension stresses on both models, with the platform switching geometry showing a maximum increase of 1.6% on the maximum value of second principal stress and a maximum decrease of 8.8% on its maximum value.

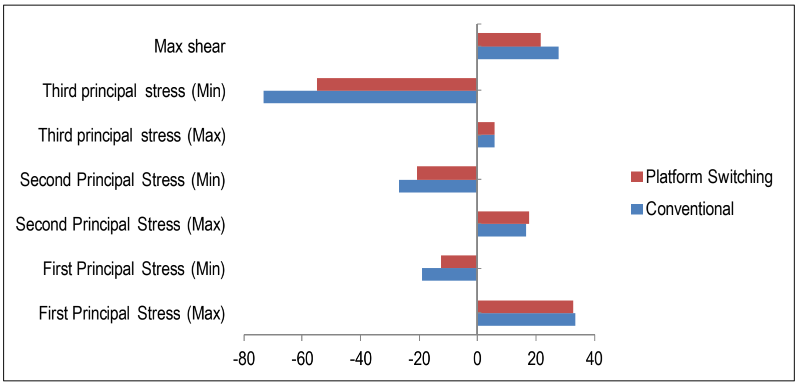

Figure 7. Peri-implant maximum shear stress and principal stresses in compact bone for conventional and platform switching models.

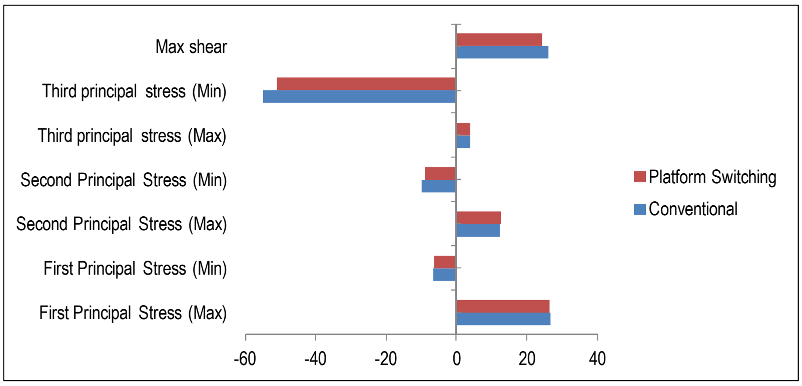

Figure 8. Peri-implant maximum shear stress and principal stresses in cancellous bone for conventional and platform switching models.

The maximum von Mises stress at the compact bone in the standard model is about 26.2% higher than in the platform switching model although at the cancellous bone the difference between the maximum von Mises stress values at both platforms is of 7.2%, being higher for the conventional platform model than for the switching platform model. The location of maximum von Mises stresses in the peri-implant cortical bone and in the peri-implant cancellous bone is the same on both models: in the peri-implant cortical bone the maximum von Mises stress occurred in the crestal bone around the implant neck and, in the peri-implant cancellous bone the maximum von Mises stress appears at the base of implant cone. The principal stresses on the peri-implant cortical bone to the platform switching model showed smaller compressive stress values than the conventional model. The tensile principal stresses show some tendency to be higher for the platform switching model than to the conventional model. Nevertheless, maximum shear stress is about 28.3% smaller on the swithching platform than in the conventional model. In the peri-implant cancellous bone the platform switching model have smaller maximum shear and principal stresses than the conventional model.

4 Discussion

In this paper, the RFA technique is used to evaluate the influence of the bone-implant interface and bone quality on the resonance frequency values. Therefore, the first aim of this study is to simulate from a mechanical point of view the evolution of the osseointegration process for a dental implant by means of a computational RFA and compare the computed results with the clinical results. With that purpose, one level of adhesion between the implant and surrounding bone is modelled using specific values of mechanical properties to the bone that surrounds the implant. This idea is based on the mechanical principle that states as the bone anchorage of an implant increase the resonance frequency of the dental system would also increase, due to an increase in the overall stiffness of the implant-bone interface. Nevertheless, because the initial implant stability is mainly affected by the mechanical interlocking between the implant and bone [15], the compact bone quality (thickness) of mandible is a morphological parameter that affects the resonance frequency values [16] and, perhaps, the trend of time evolution. Another aspect that can affect the primary stability is the contact between the bone and the thread of the implant and, therefore, the geometrical shapes of the thread and implant must be accounted for. The present geometrical model of a dental implant is provided by Camelog and all its geometric features are maintained on the related finite element models.

A virtual model of a lower jaw on the region of tooth 36 (FDI) is developed from a computerized tomography (CT) data and, in order to account for average dimensions of a human jaw, the virtual model is simplified according the data from craniometric measurements of human edentulous mandibles [18]. The bone quality of a human mandible is usually classified into four distinct classes [19], nevertheless only one geometry of a human mandible is adopted in this work. The peri-impant region is defined as an alveolar region that surrounds the implant and has a 1 mm thickness. This region includes compact and cancellous bone with different mechanical properties. Several experimental studies have been performed in order to determine the mechanical characteristics of the bone-implant interface along the shear direction [21, 22] and normal directions [23]. Thus, the mechanical properties present a wide range of variation. Moreover, Moreo et al. [24] performed a sensitivity analysis on the mechanical properties of the interface with respect the patient activity, stem surface finishing and other model parameters, concluding that high differences were found regarding time to achieve stability. The mechanical properties of bone-implant interfaces with some degree of osseointegration are chosen in such away, that a good fit between experimental and computed resonance frequencies is achieved. The computational model is able to predict the evolution of the resonance frequency values obtaining a good agreement with the clinical results.

The interfacial bonding between the implant and bone can be affected by factors such as, surgical technique, host bed, implant design, implant surface, material biocompatibility and loading conditions [25]. Understanding this factors and how they influence each other is the first step to improve osseointegration capabilities and, therefore, minimize the risk of implant failure. Thus, biological and biomechanical hypothesis are factors that alongside contribute for attainment and maintenance of osseointegration. The biomechanical behaviour is associated with the implant load and his interaction with the supporting bone by developing compressive, tensile and shear forces that may affect the osseointegration process. To achieve stable osseointegration for implant restoration, the generation of high stress concentration in bone should be avoided, since can induce severe resoprtion of the surrounding bone, [20, 26-28] leading to gradual loosening and, finally, complete loss of the implant.

From the comparison of the results presented it's clear that the platform switching technique allows reducing the stress concentration on the peri-implantar cortical region and, it seems that the stress distribution area does not change significantly between models.

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