Implant–Bone Interface Stress Distribution in Immediately Loaded Implants of Different Diameters: A Three-Dimensional Finite Element Analysis

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Abstract

Purpose: To establish a 3D finite element model of a mandible with dental implants for immediate loading and to analyze stress distribution in bone around implants of different diameters.
Materials and Methods: Three mandible models, embedded with thread implants (ITI, Straumann, Switzerland) with diameters of 3.3, 4.1, and 4.8 mm, respectively, were developed using CT scanning and self-developed Universal Surgical Integration System software. The von Mises stress and strain of the implant–bone interface were calculated with the ANSYS software when implants were loaded with 150 N vertical or buccolingual forces.
Results: When the implants were loaded with vertical force, the von Mises stress concentrated on the mesial and distal surfaces of cortical bone around the neck of implants, with peak values of 25.0, 17.6 and 11.6 MPa for 3.3, 4.1, and 4.8 mm diameters, respectively, while the maximum strains (5854, 4903, 4344 $\mu\varepsilon$) were located on the buccal cancellous bone around the implant bottom and threads of implants. The stress and strain were significantly lower ($p < 0.05$) with the increased diameter of implant. When the implants were loaded with buccolingual force, the peak von Mises stress values occurred on the buccal surface of cortical bone around the implant neck, with values of 131.1, 78.7, and 68.1 MPa for 3.3, 4.1, and 4.8 mm diameters, respectively, while the maximum strains occurred on the buccal surface of cancellous bone adjacent to the implant neck, with peak values of 14,218, 12,706, and 11,504 μm, respectively. The stress of the 4.1-mm diameter implants was significantly lower ($p < 0.05$) than those of 3.3-mm diameter implants, but not statistically different from that of the 4.8 mm implant.
Conclusions: With an increase of implant diameter, stress and strain on the implant–bone interfaces significantly decreased, especially when the diameter increased from 3.3 to 4.1 mm. It appears that dental implants of 10 mm in length for immediate loading should be at least 4.1 mm in diameter, and uniaxial loading to dental implants should be avoided or minimized.

Since the introduction and redefinition of osseointegration (close bone contact with biomaterial) were made by a Swedish group over several decades, dental implants have become a successful restorative modality in clinical dentistry, with a report of over 90% success rate.[1] A successful dental implant procedure largely depends on the presence of osseointegration. Recently, treatments with implant-supported fixed and/or removable partial prostheses have been widely recommended for partially edentulous patients.[2] The science of implantology is quite consolidated, although patients need further advances, especially in terms of reducing the traditional "waiting to-load" time. Two interventions are required to load the implant in the classical loading protocol proposed by Branemark.[3] During 3 to 6 months of healing, the patient needs to wear a denture or other removable partial prosthesis, or may require several soft relines, which result in additional visits and cost to the final rehabilitation. This also leads to increasing discomfort for the patient because of the additional time required to complete the prosthetic treatment and the difficulties of wearing a temporary prosthesis. This led to a desire for a shortened healing period, and an immediate loading protocol has been introduced;[4] however, the delayed loading protocol for osseointegrated oral implants has a rationale based on the consideration that premature loading of dental implants may lead to fibrous encapsulation instead of a direct bone–implant interface.[5] At this time, it is safe to say that completely undisturbed healing of the implant–bone interface is not necessary for successful osseointegration to occur, and in some situations, implants placed into immediate, full functional load can be expected to survive and function well.[6-9]

Recently, promising results have been observed when implants were subjected to immediate functional loads. Whether an implant is placed in function following a period of undisturbed healing or immediately after placement, the likelihood of osseous integration thereafter is greatly influenced by the biomechanical environment.[10] In the past three decades, finite element analysis (FEA) has been used extensively to predict the biomechanical performance of various dental implant designs as well as the effect of clinical factors on the success of implantation.[11] FEA allows researchers to predict stress distribution in the contact area of the implants with cortical bone and around the apex of the implants in trabecular bone. A key factor for the success or failure of a dental implant is the manner in which stresses are transferred to the surrounding bone.[12] This method offers the advantage of solving complex structural problems by dividing them into smaller and simpler interrelated sections by using mathematical techniques.[10]

The aim of this study was to analyze stress distribution around immediately loaded implants of different diameters using accurate modeling capable of obtaining more precise data.

Materials and methods

An accurate model of an edentulous mandible, which was essential for obtaining more precise results, was developed at the beginning of the work.

Mandible geometric modeling

Initially, a woman with an edentulous mandible in which three Straumann implants had been embedded between the mental foramen was selected for computerized tomography (CT) (Light Speed Pro 16, GE Healthcare, Chalfont St. Giles, UK) scanning according to the Frankfort horizontal plane (Fig 1). To

Figure 1 CT image of the mandible with three Straumann implants embedded between mental foramen.

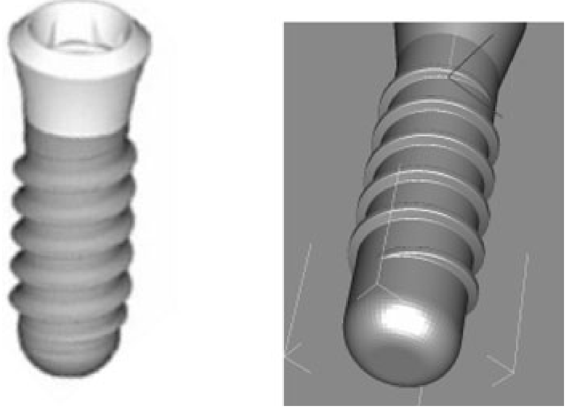

Figure 2 The standard Straumann implant (left: actual implant picture; right: solid model picture).

model the bone geometry more realistically, we reconstructed an individual geometry with a complete range of mandibles, including the separation between cortical and cancellous bones. The images obtained were converted into data and transferred to be processed by a powerful 3D segmentation modular software tool in a self-developed surgical assisted system, USIS (Universal Surgical Integration System), where the coordinates of contouring points were extracted from these plots.

Implant system

A single-piece, screw endosteal dental implant system (solid implant, ITI, Institute Straumann AG, Waldenburg, Switzerland) was selected for this study. The length of applied implants was 10 mm, and the diameters were 3.3, 4.1, and 4.8 mm, respectively. Computer-aided design (CAD) implant models of various diameters were constructed with USIS. Figure 2 displays the standard implant (4.1 mm diameter × 10 mm length). Superstructure was simplified by a solid titanium abutment with 5.5 mm height.

Construction of the 3D finite element model

Three Straumann thread implant models of the same diameter were implanted in the anterior zone of the mandible model according to the dental implant contour shown in the CT image. Three models simulating implants with 10 mm length and diameters of 3.3, 4.1, and 4.8 mm were developed to investigate the influence of diameter. Then, these geometric models were input into an automatic mesh generator (AMG) designed especially for the human mandible and implant in USIS. In the present study, finite element models of implant and bone were meshed with tetrahedron elements. As a tradeoff, linear elements require a finer mesh to provide an acceptable level of accuracy. Hence, the mandible comprised 116,645 tetrahedron elements with 171,488 nodes. The different diameter implant models were divided as such: 8731 nodes and 4708 elements (3.3 mm diameter), 10,950 nodes and 6095 elements (4.1 mm diameter), and 12,938 nodes and 7422 elements (4.8 mm diameter) (Fig 3). The regions on the implant–bone interfaces were divided to refine smaller elements. In addition, all of the linear tetrahedral elements (four nodes) were converted to parabolic elements (ten nodes), to ensure numerical accuracy. The implant–bone interface was assumed as a frictional interface (before osseous integration). To obtain initial stability for the situation of immediate loading after implantation, it was modeled using nonlinear frictional contact elements, which allowed minor displacements between implant and bone. Under these conditions, the contact zone transfers pressure and tangential forces (i.e., friction), but no tension. The friction coefficient was set to 0.3.[13] Figures 4 and 5 show the 3D finite element

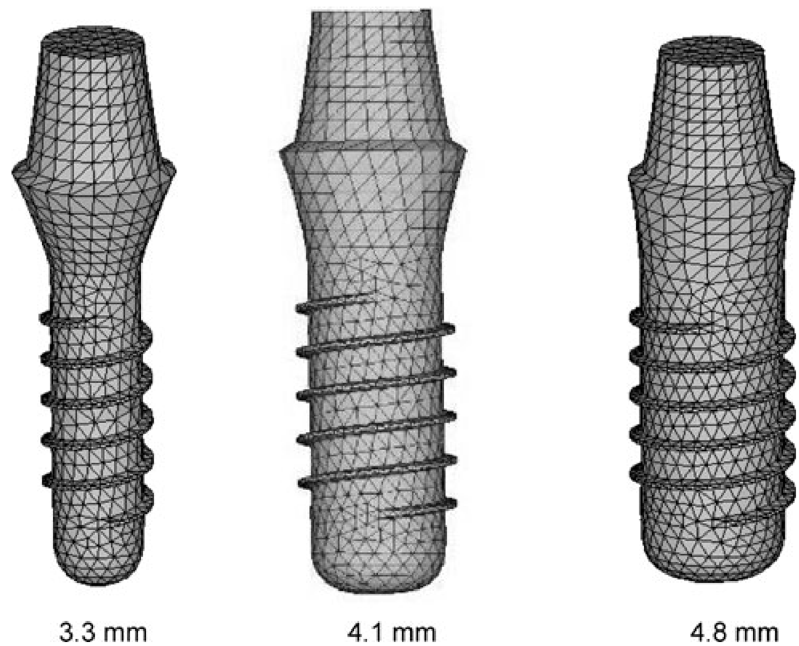

Figure 3 Mesh models of different diameter implants with abutment.

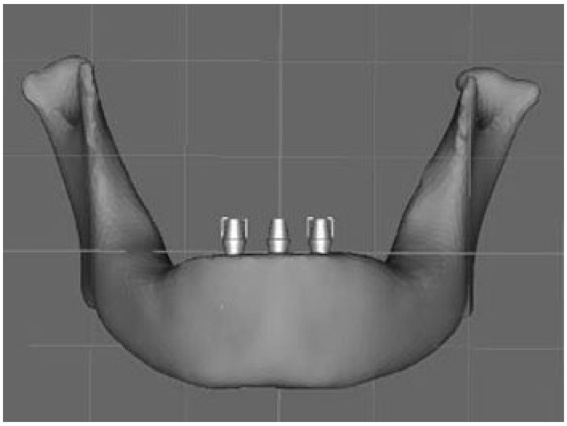

Figure 4 Three-dimensional finite element solid model of the mandible with thread implants for immediate loading.

solid and mesh models of the mandible with thread implants for immediate loading.

Material properties

The material properties of cortical and cancellous bones as well as implants in the models were assumed to be homogeneous, isotropic, and linearly elastic. Young's modulus and Poisson's ratio of materials used in the analysis are listed in Table 1.

Boundary and load

The models were constrained from the attachment regions of masticatory muscles (masseter, temporalis, medial pterygoid, and lateral pterygoid) to prevent rotation of the model around the condyles. Boundary conditions included constraining all three degrees of freedom at each of the nodes located in the front bevel face of the condyles (Fig 6). The applied forces were static. An average occlusal force of 150 N was used. A vertical and a buccolingual oblique of 45-degree forces were loaded dispersedly on the top of the implant abutment.

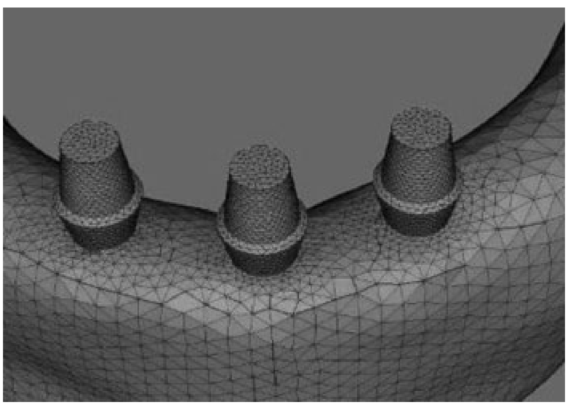

Figure 5 Three-dimensional finite element mesh model of the mandible with thread implants for immediate loading.

Table 1 Mechanical properties ascribed to materials in the model[14,15]

Material	Young's modulus (GPa)	Poisson's ratio
Titanium	103.4	0.35
Cortical bone	13.7	0.30
Cancellous bone	1.37	0.30

Operating conditions and calculation analysis

The construction of FE models was based on USIS, which was developed by the authors. Ultimately, these models were converted into the ANSYS 9.0 software (ANSYS, Inc., Canonsburg, PA) and were analyzed with it. All these were processed by a personal computer (P4 3.0 PCU, 2G EMS memory 240G HD, Windows XP Professional Edition, Microsoft Corp., Redmond, WA). Von Mises stress and strain distributions along selected zones of the bone–implant interface were calculated. Statistical analysis was performed using JMP (SAS Institute Inc., Cary, NC). The intergroup stress was evaluated using *t*-tests (level of significance = 0.05).

Results

Stress and strain levels were calculated using von Mises stress and strain values. The von Mises stresses have been reported in FEA studies to summarize the overall stress state at a point.[16] The stress and strain values around the implant–bone interface around the central implants were analyzed in this study.

Relationship between von Mises stress and implant diameter

When implants were loaded with vertical force, von Mises stress mainly concentrated on the mesial and distal surface of cortical bone around the implant neck for all models (Figs 7 and 8) with peak values of 25.0, 17.6, and 11.6 MPa for 3.3, 4.1, and 4.8 mm diameters, respectively. The values of stress were significantly lower ($p < 0.05$) with an increase in implant diameter. When implants were loaded with buccolingual force, maximum von Mises stress occurred on the buccal surface of cortical bone around the implant neck for all models (Figs 8 and 9), with peak values of 131.1, 78.7, and 68.1 MPa for 3.3, 4.1, and 4.8 mm diameters, respectively. The stress of 4.1-mm diameter implants was significantly lower ($p < 0.05$) than that of 3.3-mm diameter implants but not statistically different from that of 4.8-mm diameter implants (Table 2). As shown in Table 2, it appears that an increase in the implant diameter resulted in a reduction of stress on the implant–bone interface. Meanwhile, the stresses were strikingly increased when implants were loaded with buccolingual force compared to vertical force and well distributed for wider diameter implants (Figs 7 and 9).

Figure 10 represents the maximum von Mises stress on cortical bone around the implant neck with variant diameter. When the diameter increased from 3.3 mm to 4.1 mm and from 4.1 mm to 4.8 mm, the maximum stress decreased by 29.6% and 34.1%, respectively, for vertical force. Matching data for buccolingual force were 40% and 13.5%, respectively. This indicated that maximum stress reduced as the diameter increased, especially when implants were loaded with buccolingual force and the diameter increased from 3.3 to 4.1 mm. The maximum stress increased when implants were loaded with buccolingual force compared to vertical force (Figs 8 and 10).

Relationship between von Mises strain and implant diameter

When implants were loaded with vertical force, von Mises strain mainly concentrated on the buccal side of cancellous bone around the implant bottom and threads of implants for all models, with peak values of 5854, 4903, and 4344 $\mu\varepsilon$ for 3.3, 4.1, and 4.8 mm diameters, respectively. The values of strain were significantly lower ($p < 0.05$) with the increase of implant diameter. When implants were loaded with buccolingual force, von Mises strains increased and occurred on the buccal surface of cancellous bone adjacent to the implant neck for all models, with peak values of 14,218, 12,706, and 11,504 $\mu\varepsilon$ for 3.3, 4.1, and 4.8 mm diameters, respectively. The strain of 4.1-mm diameter implants was not significantly lower ($p > 0.05$) than those of 3.3-mm diameter implants, nor was the strain of 4.8-mm implants compared with the 4.1-mm implants; however, the stress of 4.8-mm diameter implants was significantly lower ($p < 0.05$) than those of 3.3 mm diameter (Table 3). As shown in Table 3, strain decreased when diameter increased, whether implants were loaded with vertical force or buccolingual force. Moreover, the strains had a smooth distribution for wider diameter implants; however, the maximum strain enhanced with the increased diameter when implants were loaded with buccolingual force. This was probably due to residual cancellous bone adjacent to the implant neck, which was very weak when larger diameter implants were embedded.

Figure 11 represents the maximum von Mises strain on cortical bone around the implant neck with variable diameters. When the diameter increased from 3.3 mm to 4.1 mm and from 4.1 mm to 4.8 mm, the maximum strain decreased by 29.7% and 0.8%, respectively, for vertical force. Matching data for buccolingual force were 38.8% and 1.3%, respectively. This indicates that maximum strain reduced with increased diameter, especially

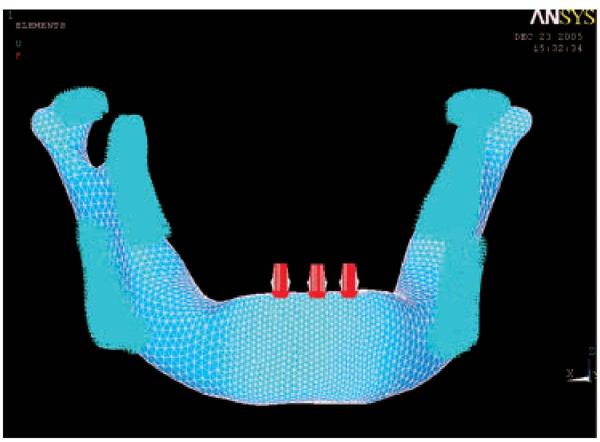

Figure 6 Boundary condition of the mandible FE model with dental implants.

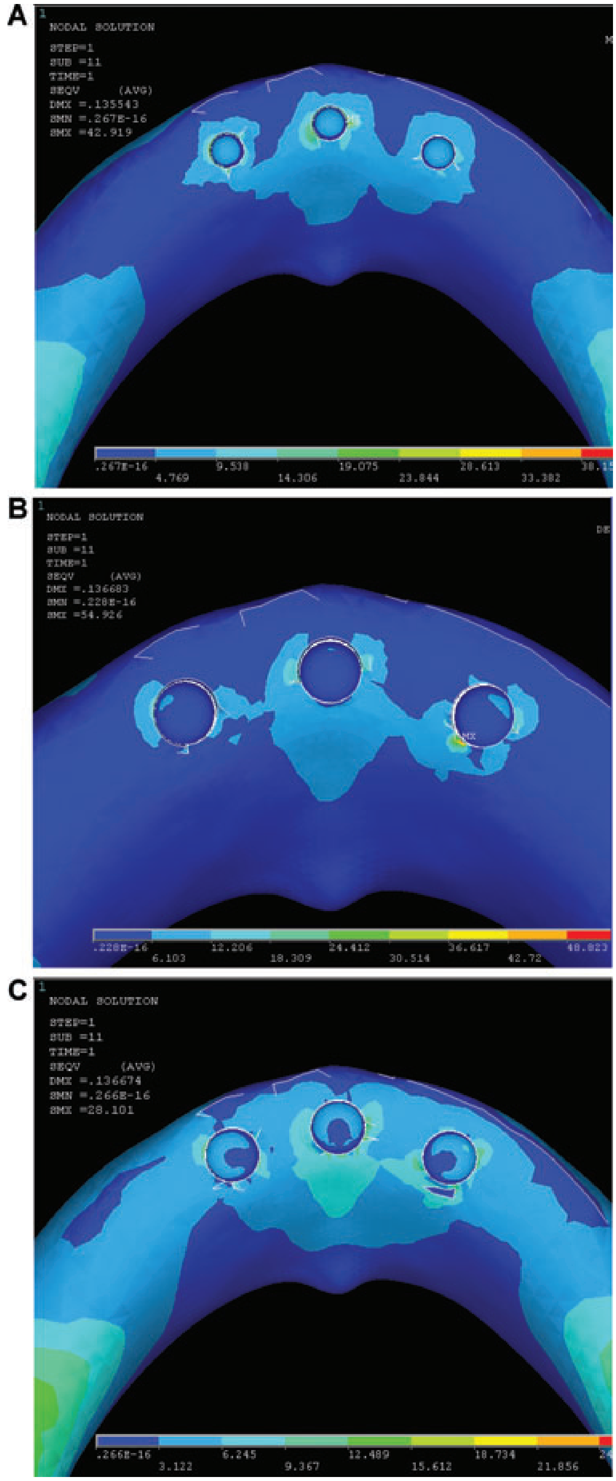

Figure 7 (A) Von Mises stress distribution of dental implants with 3.3 mm diameter loaded with vertical force. (B) Von Mises stress distribution of dental implants with 4.1 mm diameter loaded with vertical force. (C) Von Mises stress distribution of dental implants with 4.8 mm diameter loaded with vertical force.

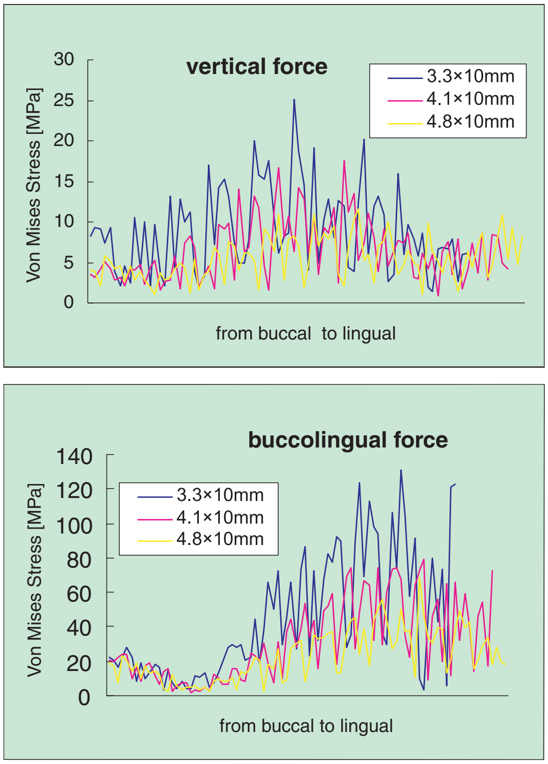

Figure 8 Von Mises stress on the cortical bone around the implant neck with variant diameter (vertical and buccolingual load cases).

when implants were loaded with buccolingual force and diameter increased from 3.3 to 4.1 mm. The maximum strain increased when implants were loaded with buccolingual force compared to vertical force.

Discussion

Branemark's protocol of delayed loading includes two separate procedures. First, implants are placed and submerged under a hermetically sutured mucosa to permit proper healing without risk of bacteremia in the absence of any functional solicitation. Second, implants are uncovered with abutment attached, and if osseointegration has occurred, a restoration can be placed on the abutment.[17] To eliminate the psychological and functional handicap during the healing period of 6 to 12 months,[18] a one-step surgical technique was proposed by the International Team for Oral Implantology (ITI) (Waldenburg, Switzerland) and has achieved a comparable success rate.[19-21] The technique includes nonsubmerged implants, and loading starts earlier than that of the Branemark technique. According to the experimental work with macaques,[22] implants that underwent immediate loading had a higher level of calcification and higher percentages of bone–implant appropriate contacts. The authors concluded that the healing of peri-implant bone under load seemed to be beneficial. Similarly, in another experimental study on animals, Henry et al[23] confirmed good predictability of the immediate loading protocol for bone healing. More recently, Jaffin et al[24] reported a success rate of 95%, comparable to that observed in delayed loaded oral implants. Piattelli et al[25] reported histological findings from two immediately loaded oral implants retrieved from two patients due to abutment fracture and psychological reasons. The authors found a very high percentage of bone–implant contact (about 60 to 70%) and an absence of fibrous encapsulation at the implant–bone interface.

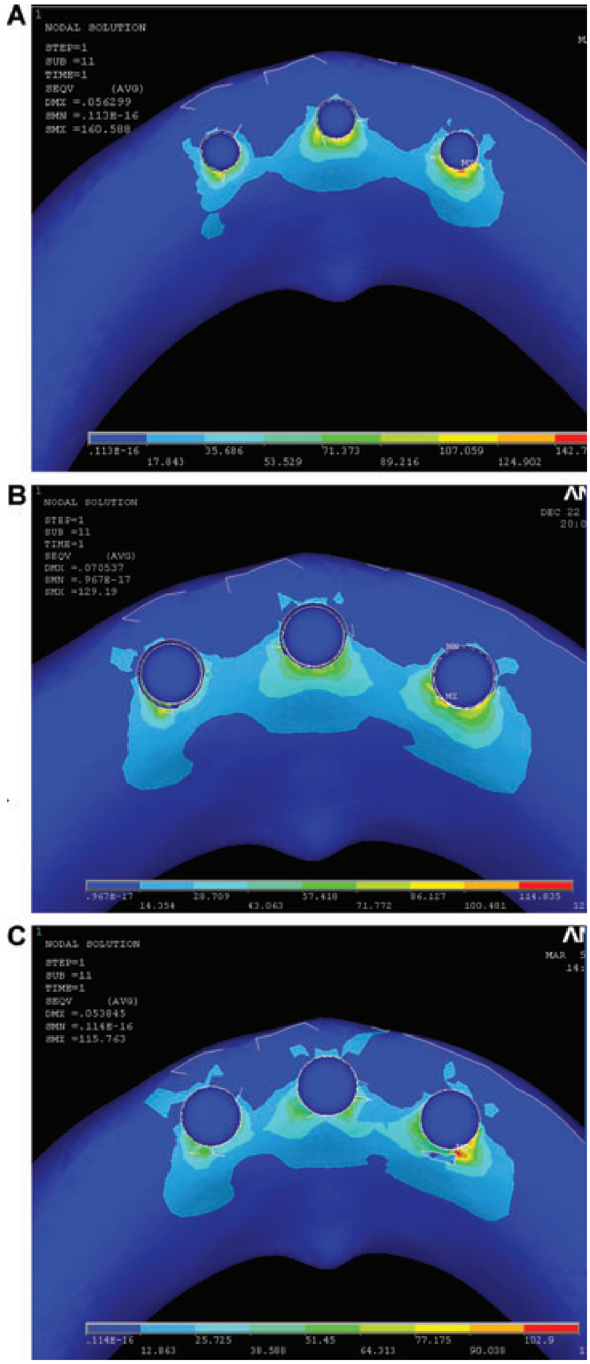

Figure 9 (A) Von Mises stress distribution of dental implants with 3.3 mm diameter loaded with buccolingual force. (B) Von Mises stress distribution of dental implants with 4.1 mm diameter loaded with buccolingual force. (C) Von Mises stress distribution of dental implants with 4.8 mm diameter loaded with buccolingual force.

Table 2 Von Mises stress (MPa) for the three tested diameter implants under different load directions

	Vertical loading			Oblique loading		
Diameter	Mean	SD	Peak	Mean	SD	Peak
3.3 mm	3.51	2.89	25.0	7.05	16.13	131.1
4.1 mm	2.75	2.03	17.6	5.50	10.14	78.7
4.8 mm	2.30	1.66	11.6	4.79	7.03	68.1

Table 3 Von Mises strain ($\mu\varepsilon$) for the three tested diameter implants under different load directions

	Vertical loading			Oblique loading		
Diameter	Mean	SD	Peak	Mean	SD	Peak
3.3 mm	2147	1389	5854	2624	1812	14,218
4.1 mm	1758	1093	4903	2531	1764	12,706
4.8 mm	1468	933	4344	2448	1676	11,504

All these findings support the idea that immediately loaded implants could be used to rehabilitate edentulous or partially edentulous jaws. Many other clinical observations[26,27] have also shown that immediate loading is indicated when the stabilization of the bone/implant is optimal (with enough volume and density) and when the estimated loads are not excessively high. Nonetheless, more experimental studies are needed to consider immediate loading protocol as a safe procedure.

Mechanical analysis using the finite element method (FEM) has been employed by many authors to understand the biomechanical behavior around dental implants with a suitable degree of reliability and accuracy, but without the risk or expense of implantation.[28] FEM is a numerical method of analysis for stresses and deformations in structures of any given geometry. A structure is broken down into many small simple blocks or elements that can be described with a relatively simple set of equations. Just as the set of elements would be joined together to build the whole structure, the equations describing the behaviors of the individual elements are joined into an extremely large set of equations that describe the behavior of the whole structure.[17] The type, arrangement, and total number of elements affect the accuracy of the results. The jawbone and implants are very complicated structures. It is difficult to establish an accurate and valid 3D FEM using conventional modeling techniques. Accurate and efficient modeling can provide an insight and understanding of the complicated nature of a dental implant surrounded by the jawbone. Successful modeling depends on accuracy in simulating the geometry and surface structure of the implant, material characteristics of implant and jawbone, loading and support conditions, and the biomechanical implant–jawbone interface.[11] The principal difficulties in simulating the mechanical behaviors of dental implants are modeling of living human bone tissue and its response to applied mechanical forces.[29,30]

This study employed an ad hoc AMG to construct high-quality FE models of a complete range of mandibles from CT, including the separation between cortical and cancellous bones. This originally stemmed from CT data, which remarkably improved the approximation of reality. Furthermore, the models were constrained from the attachment regions of the masticatory muscles and the front bevel face of the condyles, simulating the actual situation of human biting and chewing. Meanwhile, this study was modeled using nonlinear frictional contact elements, which allow minor displacements between implant and bone to keep the implant stable and provide an excellent simulation of the implant–bone interface with immediate load; however, many studies have been conducted by modeling with the region under investigation only using FEA and assuming 100% osseointegration on the interface of implant and bone mostly.[31–35]

In this study, the von Mises stress was chosen to display the computational results for comparison. The result of FE simulations revealed that the von Mises stress mainly concentrated on the mesial and distal surfaces of cortical bone around the implant neck with vertical loading, while on the buccal surface of cortical bone around the implant neck with buccolingual loading. Furthermore, the stress values notably increased under buccolingual loading as compared with vertical loading. Results reported by some literature based on complete osseointegration on the implant–bone interface are very similar to our data. Several authors using FEA have found that the highest risk of bone resorption occurs in the neck region of an implant.[36–40] The stresses were concentrated in the cortical bone around the implant neck, which is probably due to the fact that

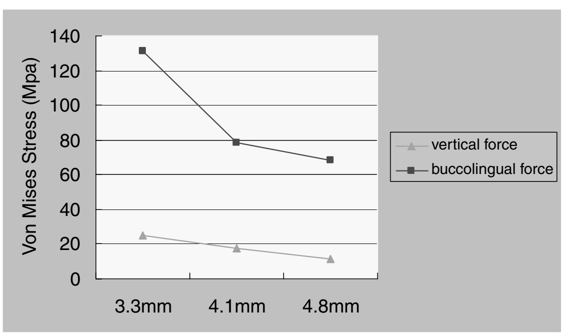

Figure 10 The maximum von Mises stress on the cortical bone around the implant neck with variant diameter.

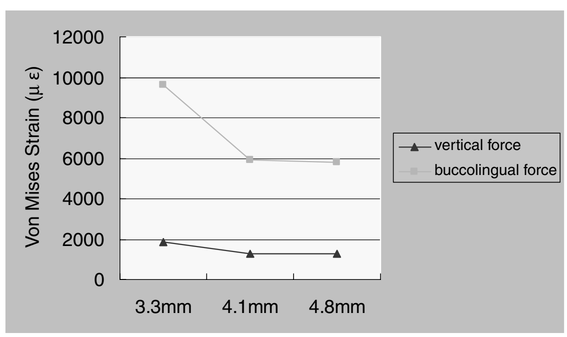

Figure 11 The maximum von Mises strain on the cortical bone around the implant neck with variant diameter.

the elastic modulus of cortical bone is higher than cancellous bone and that cortical bone is much stronger and more resistant to deformation.[41,42] The von Mises strain mainly concentrated on the buccal side of cancellous bone around the implant bottom and threads of implants with vertical loading, while it strikingly increased and occurred on the buccal surface of cancellous bone adjacent to the implant neck with buccolingual loading. Nonaxial loading has often been related to marginal bone loss, failure of osseointegration, failure of the implant and/or the prosthetic components, and failure of the cement seal on the natural tooth if connected to natural teeth.[43,44] It has long been recognized that both implant and bone should be stressed within a certain range for physiological homeostasis. Overload will cause bone resorption or failure of the implant, whereas bone underloading may lead to disuse atrophy and subsequent bone loss.[45] Usually the stress levels that actually cause biological response, such as resorption and remodeling of the bone, are not comprehensively known. Therefore, the data of stress provided from the FEA need substantiation by clinical research.[30] Although osseointegration does exist between bone and dental implants with immediate loading, overloading or implant fractures still can affect its integrity. Thus, the importance of avoiding or minimizing horizontal loads was emphasized.[46]

The structures in the model were assumed to be homogeneous isotropic and to possess linear elasticity. The properties of the materials modeled in this study, particularly the living tissues, are different, however. For instance, it is well described that the actual cortical bone of the mandible is transversely isotropic and inhomogeneous.[32] Therefore, the absolute stress and strain values cannot be related to results computed under different conditions.[43] In this study, the stress and strain were significantly lower ($p < 0.05$) with the increased implant diameter under vertical loading. The stress of 4.1-mm diameter implants was significantly lower ($p < 0.05$) than those of 3.3 mm diameter but not statistically different from that of the 4.8 mm diameter under buccolingual loading. When the diameter increased from 3.3 to 4.1 mm and from 4.1 to 4.8 mm, the maximum stress on cortical bone around the implant neck decreased by 29.6% and 34.1%, respectively, for vertical force. Matching data for buccolingual force were 40% and 13.5%, respectively. When the diameter increased from 3.3 to 4.1 mm and from 4.1 to 4.8 mm, the maximum strain around cortical bone decreased by 29.7% and 0.8%, respectively, for vertical force. Matching data for buccolingual force were 38.8% and 1.3% (76 $\mu\varepsilon$), respectively. These data showed that the maximum stress and strain would be reduced with an increased diameter, especially when implants are loaded with buccolingual force and diameter increased from 3.3 to 4.1 mm. The data from Himmlova et al,[43] who computed values of von Mises stress at the implant–bone interface for all variations in implant diameter using FEA, were the same as ours except for where supposed implant and bone bonded completely. Matsushita et al[47] also showed that stresses in cortical bone decreased in proportion to an increase in implant diameter with both vertical and lateral loads by using FEA, which was consistent with our result.

Stress distribution along the implant should be even and minimal to avoid possible complications.[48] In the present study, stress and strain levels were lower with more favorable distribution for wider implants. A wider implant was recommended in clinical use to increase contacted surfaces between bone and implant to reinforce implant stability.[49] Also, the wider area in the cervical portion of an implant may better dissipate the masticatory forces. Since occlusal loading is complicated in crestal bone loss around the implant, it is postulated that wider implants may reduce the stress around the crestal bone and potential bone loss. No reports of ITI standard 4.1-mm diameter solid-screw implant fractures can be found in the literature so far. Thus, in addition to well-defined factors leading to implant fracture, the diameter of implant could also be a principal factor;[32] however, the increase in implant diameter is restricted by finite thickness of alveolar bone. Therefore, from a biomechanical perspective, an implant with a maximum possible diameter allowed anatomically is the optimum choice. In addition, Holmgren et al showed that the widest diameter implant is not necessarily the best choice when considering stress distribution to surrounding bone within certain morphologic limits, and that an optimum dental implant size exists for decreasing stress magnitudes at the bone–implant interface.[50] Implants less than 5.0 mm have been proposed to reduce heat generation in the drilling process and to reduce subsequent bone damage. Studies have also shown that 5.0 mm diameter implants have a higher failure rate than 3.75 or 4.0 mm diameter implants.[51]

Conclusions

This finite element study on immediately loaded implants showed that increased implant diameter better dissipated the simulated masticatory force and decreased the stress and strain around the implant neck, especially when the diameter increased from 3.3 to 4.1 mm. It appears that dental implants of 10 mm in length for immediate loading should be at least 4.1 mm in diameter, and uniaxial loading to dental implants should be avoided or minimized. Further research concerning human bone response to stress and strain is needed.

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