Effect of Diameter and Length on Stress Distribution of the Alveolar Crest around Immediate Loading Implants

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ABSTRACT

Background: Many clinical observations have shown that immediate loading is indicated when the stabilization of the bone/implant is optimal and when the estimated loads are not excessively high. Nonetheless, more experimental studies are needed to consider the immediate loading protocol as a safe procedure. Mechanical analysis using the finite element (FE) method analysis has been employed by many authors to understand the biomechanical behavior around dental implants.

Purpose: This study was to evaluate the effect of the diameter and length on the stress and strain distribution of the crestal bone around implants under immediate loading.

Materials and Methods: By an ad hoc automatic mesh generator, high-quality FE models of complete range mandible was constructed from computer tomography, with three Straumann (Straumann Institute, Waldenburg, Switzerland) implants of various sizes embedded in the anterior zone. The implant diameter ranged from 3.3 to 4.8 mm, and length ranged from 6 to 14 mm, resulting in seven designs. The implant–bone interface was simulated by nonlinear frictional contact algorithm. For each design, vertical and oblique loadings of 150 N were applied, respectively, to each implant, and stresses and strains in the surrounding cortical bone were evaluated.

Results: The biomechanics analysis provided results that the oblique loading would induce significantly higher interfacial stresses and strains than the vertical loading, while the intergroup stress difference significant levels was evaluated using *t*-tests method and the level of significance (.05) that was accepted for significance. Under both loadings, the maximal values were recorded in the 3.3 (diameter) $\times$ 10 (length) mm implant configuration, whose mean and peak values were both higher than that of others with significant statistical differences. The second maximal one is 4.1 $\times$ 6 mm configuration, and the minimal stresses were recorded in 4.8 $\times$ 10 mm configuration, whose strains were also near to lowest.

Conclusions: Increasing the diameter and length of the implant decreased the stress and strain on the alveolar crest, and the stress and strain values notably increased under buccolingual loading as compared with vertical loading, but diameter had a more significant effect than length to relieve the crestal stress and strain concentration.

KEY WORDS: automatic mesh generator, biomechanics, complete range mandible, dental implant, diameter, finite element analyses, immediate loading, length, stress

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.[1]

Different studies agree that biomechanical behavior plays an important role in the survival of an implant,

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and that FEM can be a reliable method for studying the biomechanical behavior of implants. The degree of accuracy of the FEM is related to the knowledge of real load and supporting conditions.[2] Several authors have found that the highest risk of bone resorption occurs in the neck region of an implant by using FEA.[3–5] Bone loss usually begins at the crestal area of the cortical bone and can progress toward the apical region, jeopardizing the longevity of the implant and prosthesis.[6] In the literature, crestal bone loss has primarily been attributed to two factors: plaque-induced peri-implantitis, which is plaque-induced inflammation in the peri-implant tissues with subsequent bone loss,[7] and occlusal overload, in which excessive loads applied to the implant may cause pathologic stresses and strains in the crestal bone stimulating resorption.[8] Animal experiments and clinical studies have shown that in the absence of plaque-related gingivitis, bone loss around implants is associated with unfavorable loading conditions.[5] As load is transferred to the bone through the implant, the diameter and length of implants are important factors in achieving appropriate stress distribution in the bone.

Several investigators have attempted to minimize crestal bone loss by increasing the contact area of bone–implant interface and therefore reducing stress at the cortical alveolar crest. Attempts to increase the contact area of bone–implant interface have focused on increasing the diameter and/or the length of the implant, or altering the implant design/shape.[9–12]

Of these studies, finite element (FE) analysis has been widely used to evaluate the effect of those design variables on stress and strain distribution in the peri-implant region. But it should be noted that the validity of simulation depends on assumptions made in modeling geometry, material properties, boundary conditions, and the bone–implant interface.[13]

The purpose of this study was to investigate the influence of clinically feasible implant diameter and implant length on the stress and strain distribution of the crestal bone around multiple Straumann (Straumann Institute, Waldenburg, Switzerland) threaded implants embedded in the anterior zone of the mandible.

MATERIALS AND METHODS

FE Modeling

To model the bone geometry more realistically, an individual geometry of the complete range of mandible was reconstructed, including the separation between cortical and cancellous bone. This data basis originally stems from computer tomography (CT) data, which was processed by a three-dimensional segmentation modular software tool in the self-developed surgical assisted system Universal Surgical Integration System (USIS; SH Liao, Hangzhou, Zhejiang, China).

In addition, three Straumann threaded implant CAD models of various sizes were embedded in the anterior zone of the mandible: one group of models simulating implants with a length of 10 mm and diameters of 3.3, 4.1, and 4.8 mm were developed to investigate the influence of the diameter factor, and another group with a diameter of 4.1 mm and lengths of 6, 8, 10, 12, and 14 mm to investigate the influence of the length factor, as shown in Figure 1A. A simplified cone was used for the representation of the restoration for all models and modeled the restoration and abutment as a continuous unit.

Then, these geometry models were input into an automatic mesh generator (AMG), designed specially for human mandible and implant in USIS, to create patient-specific tetrahedral FE models of high quality, as demonstrated in Figure 1, B and C. The design of this ad hoc AMG has a combined constructive mechanism, allowing the generation of new mandible/implants complex models of various sizes in a very efficient way and enabling similar conditions for the models except for the implant diameter and length.

Finally, an interface procedure in USIS converted these models into the FE software system ANSYS 9.0 (ANSYS Inc., Houston, PA, USA) for numerical simulations, by an ANSYS Parametric Design Language script. In addition, all of the linear tetrahedral elements (four nodes) are converted to parabolic elements (10 nodes), to ensure numerical accuracy.

Properties of the Materials

The mechanical properties of all the materials were assumed to be isotropic and linearly elastic. To describe the mechanical behavior, knowledge of the value of two parameters is sufficient: Young's elastic modulus (E) and Poisson's ratio (ν). The values of these parameters were $E = 13.7$ GPa and $\nu = 0.3$ for the cortical bone, and $E = 1.37$ GPa and $\nu = 0.3$ for the cancellous bone. The elastic properties of the titanium implant were $E = 103.4$ GPa and $\nu = 0.35$.[13]

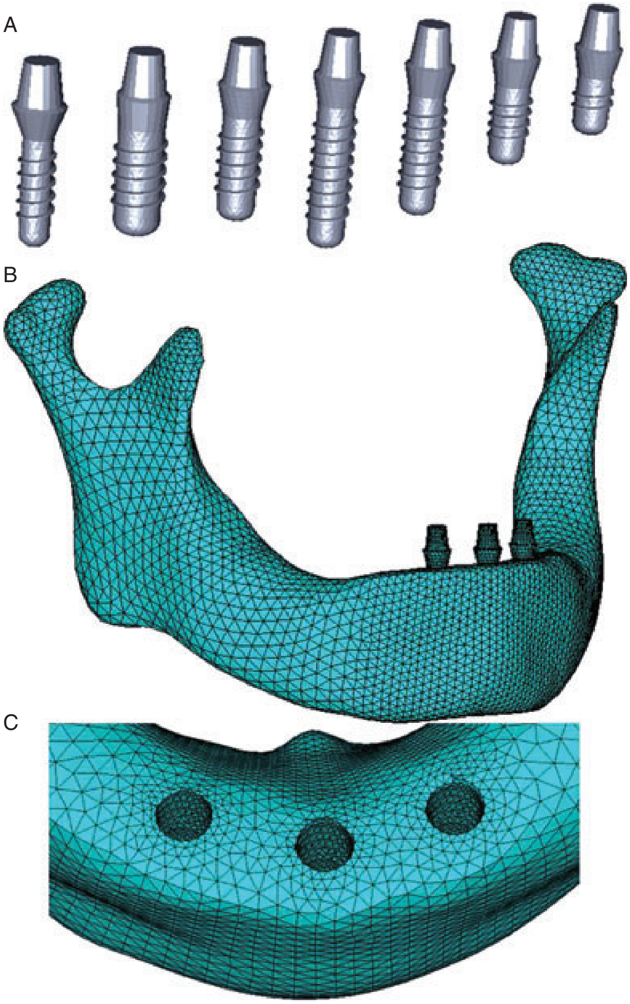

Figure 1 *A,* Depicts seven solid-screw ITI dental implants with different sizes (3.3 × 10, 4.8 × 10, 4.1 × 10, 4.1 × 14, 4.1 × 12, 4.1 × 8, 4.1 × 6 mm); *B,* illustrates the finite element meshes of the complete mandible and three implants embedded in the anterior zone; *C,* demonstrates zoomed in view of the embedding region.

Implant–Bone Interface Design

To investigate initial stability for the situation immediately after implantation, the implant–bone interface was assumed as before the occurrence of osseointegration and simulated by nonlinear contact zones with friction. The coefficient of friction was set to 0.3.[14] This means that the contact zones transfer only pressure and tangential frictional forces, whereas tension is not transferred.

Boundary and Loading Conditions

For each model, boundary conditions included constraining all three degrees of freedom at each of the nodes located at the joint surface of the condyles and the attachment regions of the masticatory muscles (masseter, temporalis, medial pterygoid, and lateral pterygoid). Loading was simulated by applying vertical and buccolingual oblique (45 degree angle) loadings of 150 N, respectively, at the top of each implant.

Statistical Analysis

Statistical analysis was performed using JMP (SAS Institute Inc., Cary, NC, USA). The intergroup stress difference significant levels were evaluated using the *t*-test method and the level of significance (.05) that was accepted for significance.

RESULTS

The simulation results showed that the occlusal forces are distributed primarily to the crestal bone, rather than evenly throughout the entire surface area of the implant interface, as demonstrated in Figures 2 and 3.

Stress Variation by Changing Implant Size

For each size configuration, the mean and peak values of von Mises stress on the implant–bone interface

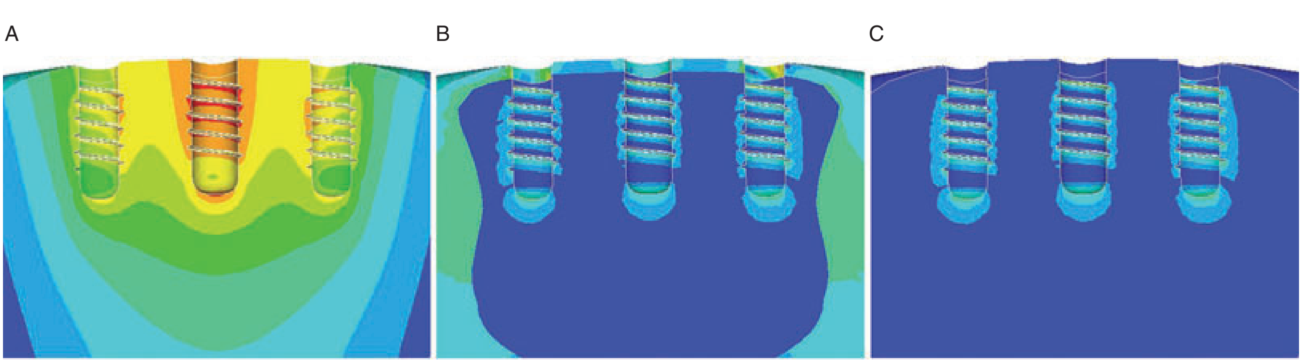

Figure 2 Displacement (A), von Mises stress (B), and von Mises strain (C) fields distribution with stand implant size (4.1 × 10 mm) under vertical loading.

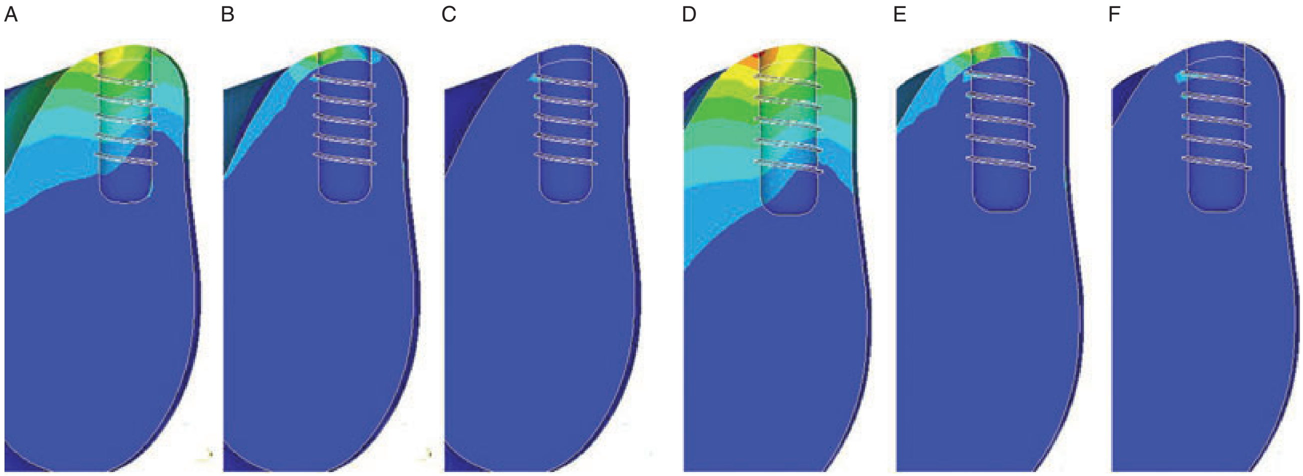

Figure 3 *A–C,* Illustrates the displacement, von Mises stress, and von Mises strain fields distribution at the first implant location with stand implant size (4.1 × 10 mm) under buccolingual oblique loading; *D–F,* illustrates those at the second (center) implant location.

surrounding the cervical regions of implants were evaluated and listed in Table 1 and Figure 4. For convenience, group A to group G for these configurations were denoted and were arranged in a descending sort by stress values: group A, group D, group E, group B, group F, group G, and group C. The intergroup stress difference significant levels were evaluated using the *t*-test method, as shown in Table 2.

Under vertical loading, the maximal stress values were recorded in group A, whose mean and peak values were both higher than that of the other groups with significant statistical differences (significance level $p = .000$ for B, C, F, and G, .001 for E, and .030 for D). The second maximal one is group D, whose mean and peak values were both higher than that of the other five groups, while it only had significant statistical differences with groups C, F, and G ($p = .002$, .036, .009). Under oblique loading, the maximal values were also in group A, whose mean and peak values were both higher than that of the other groups with significant statistical differences (with the exception of group D). The second maximal one is also group D, whose mean and peak values were both higher than that of the other five groups, while it only had significant statistical differences with groups C, F, and G ($p = .000$ for C, .022 for F, .008 for G). There was no significant statistical difference of stress values between group B and group E, neither between group F and group G. Under both vertical and oblique loadings, the minimal stresses were recorded in group C. Group B (stand implants) only had a significant statistical difference of stress values with group A and group C.

These results suggest that increasing the implant diameter and implant length both resulted in the reduction of crestal stress, as shown in Figure 4. The effect was more significant for the diameter factor and under the buccolingual oblique loading.

Strain Variation by Changing Implant Size

For each group, the mean and peak values of von Mises strain on the implant–bone interface surrounding the cervical regions of the implants are listed in Table 3, and the intergroup strain difference significant levels using the *t*-test method are demonstrated in Table 4.

Under vertical loading, the maximal strain values were recorded in group A, whose mean and peak values were both higher than that of the other groups with significant statistical differences (with the exception of group D). The second maximal one is group D, whose mean and peak values were both higher than that of the other five groups with significant statistical differences. Under oblique loading, the maximal values were also in group A, whose mean and peak values were both higher than that of the other groups with significant statistical differences (with the exception of group D and group E). The second maximal one is also group D, whose mean and peak values were both higher than that of the other five groups with significant statistical differences (with the exception of group E). There was no significant statistical difference of strain values between group B and group E, neither between group F and group G. The minimal strains were recorded in group C and group G. Group B (stand implants) only had a significant statistical difference of strain values with group A

TABLE 1 Von Mises Stresses on the Implant–Bone Interface Surrounding the Cervical Region of the Implants (MPa)

	Vertical Loading			Oblique Loading		
Group	Mean	SD	Peak	Mean	SD	Peak
A (3.3 × 10)	8.88	5.20	25.05	44.69	36.10	131.13
B (4.1 × 10)	6.38	3.50	15.22	31.45	23.07	78.66
C (4.8 × 10)	5.42	2.61	11.63	22.82	14.86	68.05
D (4.1 × 6)	7.19	4.48	20.69	38.75	32.75	105.6
E (4.1 × 8)	6.40	3.74	16.13	32.31	25.51	85.29
F (4.1 × 12)	5.94	3.12	12.71	28.78	22.00	74.56
G (4.1 × 14)	5.65	2.84	12.03	27.28	21.38	72.35

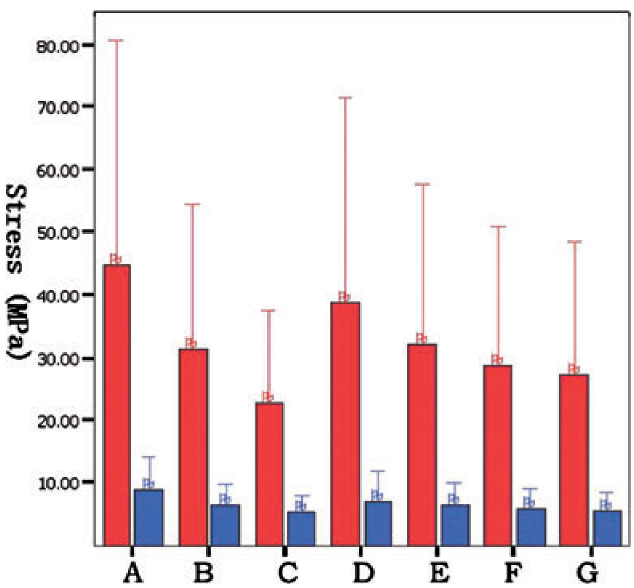

Figure 4 Stress variation on the cervical implant–bone interface by changing implant size.

and group D under vertical loading, and only had a significant statistical difference with group A, group D, and group C under oblique loading.

Increasing the implant diameter and implant length both resulted in the reduction of crestal strain, and the effect was greater for the diameter factor.

DISCUSSION

Load transfer from implants to surrounding bone depends on the type of loading, the bone–implant interface, the diameter and length of the implants, the shape and characteristics of the implant surface, the prosthesis type, and the quantity and quality of the surrounding bone. This study focused on the effect of the two most common implant design variables of clinical interest: implant diameter and implant length, by two groups of systemic FE simulations. In contrast to many previous studies cited in the literature, our investigation paid particular attention to the condition of immediate loading and improve the quality of mathematic modeling.

TABLE 2 Intergroup Stress Difference Significant Level (*p* Value) Using the *t*-Test Method*

	A	B	C	D	E	F	G
A (3.3 × 10)	/	.000	.000	.030	.001	.000	.000
B (4.1 × 10)	**.007**	/	.040	.193	.980	.380	.140
C (4.8 × 10)	**.000**	**.004**	/	.002	.048	.243	.595
D (4.1 × 6)	**.277**	**.097**	**.000**	/	.214	.036	.009
E (4.1 × 8)	**.014**	**.820**	**.004**	**.157**	/	.385	.150
F (4.1 × 12)	**.001**	**.440**	**.039**	**.022**	**.337**	/	.538
G (4.1 × 14)	**.000**	**.226**	**.117**	**.008**	**.169**	**.655**	/

*The upper-right *p* values are for vertical loading, and the bottom-left *p* values are for oblique loading.

TABLE 3 Von Mises Strains on the Implant–Bone Interface Surrounding the Cervical Region of the Implants (μ ε)

	Vertical Loading			Oblique Loading		
Group	Mean	SD	Peak	Mean	SD	Peak
A (3.3 × 10)	767.44	316.55	1,850	3,693.21	2,470.66	9,620
B (4.1 × 10)	608.85	229.09	1,220	2,868.57	1,711.36	5,890
C (4.8 × 10)	546.00	196.52	1,050	2,374.10	1,458.14	5,820
D (4.1 × 6)	704.90	260.68	1,520	3,576.79	2,265.58	7,750
E (4.1 × 8)	625.80	236.53	1,210	3,049.12	1,860.10	6,580
F (4.1 × 12)	575.60	194.39	1,070	2,641.91	1,548.90	5,450
G (4.1 × 14)	536.57	174.86	1,050	2,503.07	1,564.92	5,300

The principal difficulty in simulating the mechanical behavior of dental implants is the modeling of human bone tissue. A limited number of FEA studies have been used to build the three-dimensional FEA model database of the mandible.[15] However, many studies have been conducted by modeling only the region under investigation with FEA.[16–19] The common procedure in these studies has modeled the section of interest instead of constructing the whole mandible. In contrast, this study employed an ad hoc AMG to construct high-quality FE models of complete range mandible from CT, which remarkably improved the approximation of reality. It was a well reproduced individual geometry of a complete range of mandible, including the separation between cortical and cancellous bone. This data basis originally stemmed from CT data, which was processed by the discretized marching cube algorithm and saved in a "tiled surface" form. Meanwhile, 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 well. Furthermore, most FEA models assumed a state of optimal osseointegration for the interface between the bone and implant, meaning that the cortical and cancellous bones are perfectly bonded to the implant,[16,17,20,21] which does not occur exactly in clinical situations of immediate loading. Thus, this study incorporated frictional contact area for the bone–implant interface to simulate immediate stability clinically after implantation, which allows minor displacements between the implant and bone to keep the stabilization of the implant, an excellent simulation of the bone–implant interface with immediate load.

As most bone loss/resorption was initiated at the alveolar crest around the implant's neck,[6,22] this study focused on the values of von Mises stress and von Mises strain on the surrounding cortical bone interface for all variations. The simulation results agreed with other FE investigations, that the occlusal forces are distributed primarily to the crestal bone, rather than evenly throughout the entire surface area of the implant

TABLE 4 Intergroup Strain Difference Significant Level (*p* Value) Using the *t*-Test Method*

	A	B	C	D	E	F	G
A (3.3 × 10)	/	.000	.000	.175	.002	.000	.000
B (4.1 × 10)	**.014**	/	.055	.012	.637	.309	.023
C (4.8 × 10)	**.000**	**.043**	/	.000	.018	.321	.740
D (4.1 × 6)	**.756**	**.023**	**.000**	/	.041	.000	.000
E (4.1 × 8)	**.066**	**.511**	**.009**	**.101**	/	.134	.006
F (4.1 × 12)	**.002**	**.367**	**.244**	**.002**	**.124**	/	.173
G (4.1 × 14)	**.000**	**.150**	**.578**	**.000**	**.042**	**.564**	/

*The upper-right *p* values are for vertical loading, and the bottom-left *p* values are for oblique loading.

interface.[10,23] So this study focused on the values of von Mises stress and von Mises strain on the surrounding cortical bone interface for all implant size variations. The result of FE simulations revealed that under both vertical and oblique loadings, increasing implant diameter has a very significant effect to decrease the stress and strain in the surrounding crestal bone and to relieve the stress concentration phenomenon, which agrees with other FE investigations.[10,12] Our study incorporated frictional contact area for the bone–implant interface to simulate immediate stability clinically after implantation, whereas other investigations assumed a state of optimal osseointegration for the interface between the bone and implant, meaning that the cortical and cancellous bones are perfectly bonded to the implant. The maximal stresses and strains were both recorded in the model with implant size 3.3×10 mm, whose mean and peak values were both higher than that of the other groups with significant statistical differences ($p < .05$). And the minimal stresses were recorded in the 4.8×10 mm implant, whose strains were also near to lowest. These were also confirmed by several clinical studies that reported higher survival rates and reduced crestal bone loss for wide-diameter implants.[24–26] From a biomechanical standpoint, the use of wider implants increases stiffness of the implant and bone–implant contact surfaces, allowing the engagement of a maximal amount of bone and a theoretical improvement on the distribution of stress in the surrounding bone. The use of wider components also allows for the application of higher torque in the placement of prosthetic components.[27]

The use of wide implants, however, is limited by the width of the residual ridge and aesthetic requirements for a natural emergence profile; the use of wide-diameter implants could lead to bone loss when narrow posterior ridges exist.[10] Higher failure rates for wide implants have been found in clinical reports.[28] Increasing implant length had a moderate-to-large influence on reducing the stress and strain in the surrounding crestal bone, which agrees with other FEM studies.[11,12,29] Our study incorporated frictional contact area for the bone–implant interface to simulate immediate stability clinically after implantation, whereas those FEM studies assumed a state of optimal osseointegration for the interface between the bone and implant, meaning that cortical and cancellous bones are perfectly bonded to the implant. In this study, the second maximal stresses and strains were recorded in the model with implant size 4.1×6 mm, whose mean and peak values were both higher than that of the other five groups, and most of the differences were statistically significant ($p < .05$). Clinical investigations also have shown statistically better survival rates for implants with greater length than shorter ones. The 7-mm implant, among the shortest implant length produced by most implant companies, exhibits greater failure than all other implant lengths.[30]

The relationship between implant length and survival, however, is limited.[3,17] These studies indicated that the use of longer implants did not necessarily relieve the stress concentration in the bone around the implants.[31,32] The relatively weaker influence of implant length on stress distribution was also seen in the results of the present study. There was no significant statistical difference of stresses and strains between the model with implant size 4.1×10 and 4.1×8 mm, neither between implant size 4.1×14 and 4.1×12 mm, and standard implant size 4.1×10 mm only had significant statistical differences with implant size 3.3×10 and 4.8×10 mm. Other mechanical analyses also supported the view that increasing the implant length may only increase the success rate to a certain extent.[33] In conclusion, increasing both the implant diameter and implant length resulted in the reduction of crestal stress. The effect was more significant for the diameter factor and under the buccolingual oblique loading.

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, however, are different. For instance, it is well described that the actual cortical bone of the mandible is transversely isotropic and inhomogeneous.[18] Therefore, the absolute stress and strain values cannot be related to results computed under different conditions.[34] The result of FE simulations revealed that the stress and strain values notably increased under buccolingual loading as compared with vertical loading. Results reported by some literature concerning the localization of stresses on implant are very similar to our data. Several authors have found that the highest risk of bone resorption occurs in the neck region of an implant by using FEA.[4,5,35] 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 the natural teeth.[34] It has long been recognized that both implant and bone should be stressed with a certain range for physiologic homeostasis. Overload will cause bone resorption or failure of the implant, whereas bone underloading may lead to disuse atrophy and subsequent bone loss.[22] Usually, the stress levels that actually cause biologic response, such as resorption and remodeling of the bone, are not comprehensively known. Therefore, the data of stress provided from the FE analysis need substantiation by clinical research.[16]

CONCLUSIONS

This FE study on immediate loading implants revealed that the occlusal forces are distributed primarily to the alveolar crest, and the stress and strain values notably increased under buccolingual loading as compared with vertical loading. In light of these findings, increasing both the diameter and length of the implant resulted in the reduction of crestal stress and strain, but diameter had a more significant effect than length to relieve the crestal stress and strain concentration, and the effect was greater for the diameter factor as well.

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