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Influence of Clinically Relevant Factors on the Immediate Biomechanical Surrounding for a Series of Dental Implant Designs

The objective of the present study was to assess the influence of various clinically relevant scenarios on the strain distribution in the biomechanical surrounding of five different dental implant macrogeometries. The biomechanical environment surrounding an implant, i.e., the cortical and trabecular bone, was modeled along with the implant. These models included two different values of the study parameters including loading conditions, trabecular bone elastic modulus, cortical/trabecular bone thickness ratio, and bone loss for five implant designs. Finite element analysis was conducted on the models and strain in the bones surrounding the implant was calculated. Bone volumes having strains in four different windows of 0–200 $\mu\epsilon$, 200–1000 $\mu\epsilon$, 1000–3000 $\mu\epsilon$, and >3000 $\mu\epsilon$ were measured and the effect of each biomechanical variable and their two-way interactions were statistically analyzed using the analysis of variance method. This study showed that all the parameters included in this study had an effect on the volume of bones in all strain windows, except the implant design, which affected only the 0–200 $\mu\epsilon$ and >3000 $\mu\epsilon$ windows. The two-way interaction results showed that interactions existed between implant design and bone loss, and loading condition, bone loss in the 200–1000 $\mu\epsilon$ window, and between implant design and loading condition in the 0–200 $\mu\epsilon$ window. Within the limitations of the present methodology, it can be concluded that although some unfavorable clinical scenarios demonstrated a higher volume of bone in deleterious strain levels, a tendency toward the biomechanical equilibrium was evidenced regardless of the implant design. [DOI: 10.1115/1.4003318]

Keywords: implant design, biomechanical surrounding, finite element method, factorial analysis

1 Introduction

It is general consensus that low amounts or the absence of biomechanical stimuli during the early stages of healing increase the likelihood of osseointegration of surgically placed endosseous implants (intimate contact between bone and implant at the optical microscopy level), enabling subsequent prosthetic loading of the system [1]. Thus, as this approach has resulted in high clinical survival rates, often above 90%, emphasis on better understanding the biomechanical environment surrounding an implant has gained significant attention in the recent years [2–4]. This biomechanical environment is a vital factor in promoting and maintaining osseointegration, allowing initial and long-term rehabilitation functional loading.

It has been demonstrated that excessive loading may result in treatment failure even for well osseointegrated implants [2–4]. Strain levels exceeding the physiological tolerance threshold of bone around the implant may cause microdamage accumulation and induce bone resorption [5,6]. It is accepted that bone can sense mechanical loading and modify its structure and morphology as a function of these loads [7,8]. One of the accepted theories for such mechanosensing behavior is the mechanostat theory [5], where changes in the bone density are related to the strain levels. The bone density decreases on underloading, where the strains are typically in the range of 0–200 $\mu\epsilon$ (disuse window), remains the

same in the strain range of 200–1000 $\mu\epsilon$ (adapted window), and increases through mild overload causing strains in the range of 1000–3000 $\mu\epsilon$ (mild overload window). Further loading, reaching pathological conditions, may lead to strains >3000 $\mu\epsilon$ (pathologic overload window), where bone may not be repaired by normal modeling/remodeling activity [9,10].

The occlusal loads acting on the implant are transmitted onto the surrounding bones. The dynamic occlusal loading and associated bone deformation result in modeling and remodeling of the supporting bones. Under normal loading conditions, bone microstructural alteration occurs temporally [9]. On the other hand, overloading during mastication can damage the bone in osseointegrated implant vicinities, substantially changing its mechanical load bearing capability. In addition to the occlusal condition (i.e., magnitude, direction, and frequency of the loading), numerous other patient dependent variations may result in substantially different implant-bone biomechanical behaviors [11] such as bone quantity (thickness and density), quality (mechanical properties), and inherent patient anatomy of the edentulous region. For example, the height and width of the alveolar bone ridge can be markedly reduced due to substantial bone resorption after tooth extraction [12]. Furthermore, the periimplant bone loss around the implant cervical region observed in some clinical situation may substantially affect the biomechanical environment of osseointegrated implants [13–15]. Finally, implant design may also shift strain levels, strain distributions, and bone volumes within each strain window [9].

Although a plethora of studies concerning mechanical simulations have been performed in the implant dentistry field, the vast majority of them have evaluated only a few variables in each

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Contributed by the Bioengineering Division of ASME for publication in the JOURNAL OF BIOMECHANICAL ENGINEERING. Manuscript received May 26, 2010; final manuscript received November 26, 2010; accepted manuscript posted December 22, 2010; published online February 7, 2011. Assoc. Editor: Avinash Patwardhan.

Table 1 Implant dimensions used in this study

Implant	Diameter (mm)	Length (mm)
3i	4	10
Unitite	4	10
Nobel MK-3	4	10
Nobel Replace	4.3	10
Straumann	4.1	10

study [16–19], allowing a narrow window for analyzing the effects of multiple variable combination on the system through statistical analysis. While a per model solution and analysis provides detailed information on one particular situation, multiple iterations with interplay of variables may allow for a more appropriate platform for better understanding the behavior of bone around implants [11,12]. Thus, the present study assessed the surrounding bone strain levels due to five different implant macrogeometries through a factorial study design including several clinically relevant factors and their combinations in order to determine their relative contribution to the strain distribution in bone.

2 Materials and Methods

Five different implant designs were selected for this study, which included 3i Certain (Biomet-3i, Palm Beach Gardens, FL), Unitite (SIN, Sao Paulo, Brazil), NobelBiocare MK-3 and Replace Select (NobelBiocare, Göteborg, Sweden), and Straumann Standard (Straumann, Basel, Switzerland). The dimensions of the implants are shown in Table 1. The computer-aided design (CAD) models comprising the implant, abutment, and screw of each implant were designed to follow their respective manufacturer's dimensions and are shown in Fig. 1. The CAD models were created using SOLIDWORKS 2009 (Dassault Systèmes SolidWorks Corp., Concord, MA). The cortical bone was modeled with two different thicknesses of 1 mm and 4 mm in an attempt to simulate variations in cortical to trabecular bone ratio [20] (Fig. 2). Two conditions of no cervical bone loss and 4 mm cervical bone loss were analyzed (Fig. 2).

The solids comprising the bones surrounding the implants were divided into two concentric cylindrical regions. The three-dimensional finite element models observed in Figs. 3(a) and 3(b) included a smaller cylinder of 3 mm radius (region of interest) adjacent to the implant and the outer cylinder surrounding the inner geometries because previous studies have shown that forces are rapidly dissipated outside this volume [21].

The models were imported into ANSYS WORKBENCH 11.0 (Ansys,

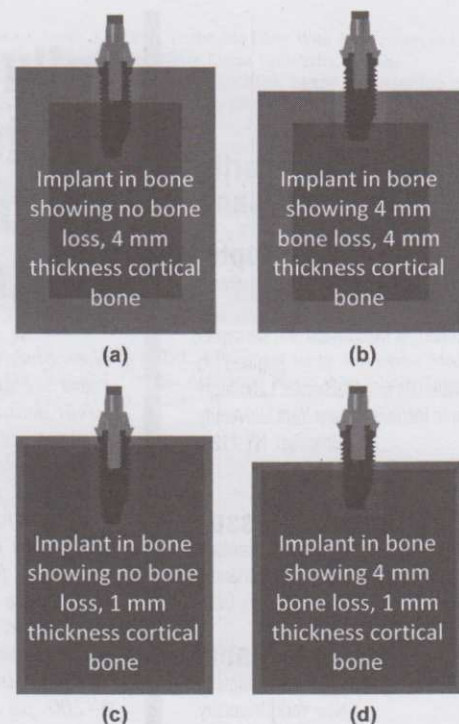


Fig. 2 Two-dimensional sections of the four different clinical scenarios investigated in this study for different loading conditions and implant design. All the models were evaluated for trabecular bone elastic modulus of 1 GPa and 4 GPa.

Inc., Canonsburg, PA) and were discretized using 3D 20-node structural solid element SOLID186 and 3D ten-node tetrahedral structural solid SOLID187. The cylindrical bone regions were divided into equal element sizes of 0.2 mm for all the models. A typical meshed model is shown in Fig. 3(c). An extensive convergence study was conducted. The difference in the resulting strain values was less than 4% between the selected mesh size and the finer mesh sizes. However, the difference increased sharply for larger mesh sizes. Therefore, the element size of 0.2 mm was selected to balance the accuracy of results and computational expense. The number of elements used in the analysis ranged from 80,000 to 130,000 depending on the geometry. A total of 16 models were simulated for each implant design to study various scenarios (the full study comprised 80 models).

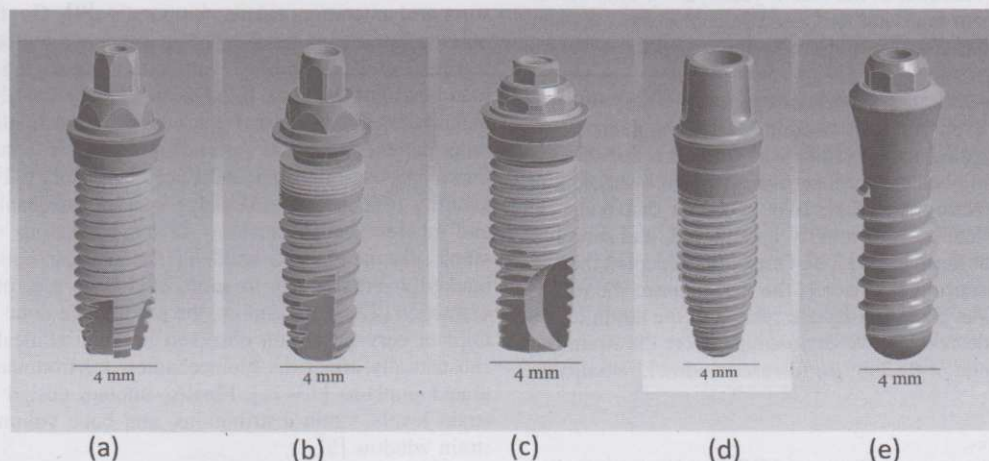


Fig. 1 CAD drawings of the different implant groups evaluated: (a) 3i Certain, (b) SIN Unitite, (c) Nobel MK-3, (d) Nobel Replace, and (e) Straumann

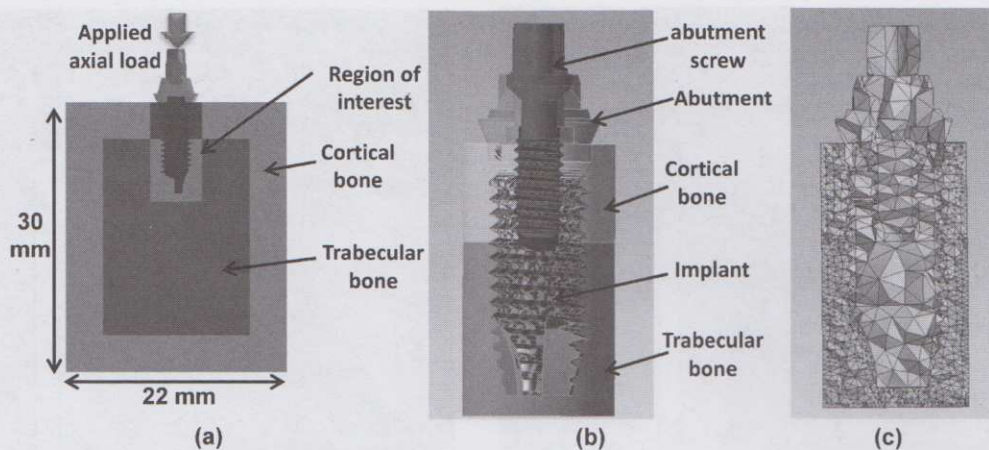


Fig. 3 (a) Schematic FEA model containing 3i implant under axial loading condition, (b) expanded view of the region of interest, and (c) mesh setup of the implant and the immediate surroundings

The load applied on the model was either a 200 N axial force or a 200 N axial force combined with a 100 N transverse force [18]. The combination of axial and transverse loading, termed as mixed loading, simulates practical conditions where the actual applied force may be inclined with respect to the implant axis and components can be resolved in the longitudinal and transverse directions. The load was applied on a small cap modeled on top of the screw connecting implant to abutment, as it has been previously shown that abutment configuration does affect the overall system mechanics and should be considered during analysis [15]. The base and the outer sides of the cortical bone were rigidly fixed. Considering that the strains dissipated sharply in a small volume around the implant, these remote boundary conditions did not have an impact on the analysis results [21].

The following assumptions were made: (1) all solids were linear elastic, homogeneous, and isotropic, (2) complete osseointegration between the bone and the implant, (3) no flaw in any of the components, and (4) no friction between the components. The assigned Young modulus (E) and Poisson ratio (ν) are given in Table 2. The trabecular bone modulus of 1 GPa and 4 GPa were taken to cover a wide range and study the effect of trabecular bone modulus on the strain levels in the immediate surroundings of the implant.

The post-processing step included identifying the number of nodes within the desired strain windows (0–200 $\mu\epsilon$, 200–1000 $\mu\epsilon$, 1000–3000 $\mu\epsilon$, and >3000 $\mu\epsilon$) [10] within the cylinder of analysis and reporting them as percentage volume of the bone in these strain windows. A typical set of results displaying strain distributions in two different study conditions is shown in Fig. 4. The numerical values of strains in the entire region of interest were identified and subjected to statistical analysis.

For statistical evaluation, five study parameters including implant design, cortical bone thickness, trabecular bone modulus of elasticity, crestal bone loss, and load condition were utilized as independent variables. The dependent variables considered in this study were the four bone strain windows, namely, 0–200 $\mu\epsilon$,

200–1000 $\mu\epsilon$, 1000–3000 $\mu\epsilon$, and >3000 $\mu\epsilon$. The obtained results were analyzed and appraised using a general linear model (GLM) procedure (SPSS v. 17.0, SPSS, Inc., Chicago, IL) to evaluate the one-way and two-way interactions of variables. The higher order interactions were summated as a grouped error term. Significant parameters are reported with type 1 error rates less than 5%.

3 Results

While the descriptions of effects of each variable and their two-way interactions are presented as follows, detailed information is presented in Tables 3–8. The significant two-way interactions with p -values <0.05 can be identified from the tables. Table 3 shows the GLM model results for implant type, load position, and cortical bone thicknesses, while Table 4 presents the results for cervical bone loss and trabecular bone elastic modulus.

Besides implant design, which did not significantly affect bone volume percent in two of the strain windows (200–1000 $\mu\epsilon$ and 1000–3000 $\mu\epsilon$), all other variables presented a significant effect on all strain windows. Figure 5 represents the varying percentages of the affected bone volume as a function of the implant design. Relative to other systems, the Straumann implant presented higher and lower volume percents of bone affected in the 0–200 $\mu\epsilon$ and in the >3000 $\mu\epsilon$ window, respectively (Fig. 5). When taking into consideration the other analyzed variables (i.e., cortical bone thickness, trabecular bone modulus of elasticity, crestal bone loss, and load condition), a significant effect was observed for all strain window levels (Table 3). However, depending on the strain level, the influence of the different conditions presented different trends. Axial loading condition presented significantly lower percentage of bone volumes in 0–200 $\mu\epsilon$, 1000–3000 $\mu\epsilon$, and >3000 $\mu\epsilon$ windows compared with the mixed loading but a higher volume in the 200–1000 $\mu\epsilon$ window.

The 4 mm cortical bone thickness showed higher bone volume in 0–200 $\mu\epsilon$, 200–1000 $\mu\epsilon$, and >3000 $\mu\epsilon$ windows. On the contrary, 1 mm cortical bone thickness presented a greater influence on the 1000–3000 $\mu\epsilon$ window. Comparing the 4 mm and no bone loss situation, a higher percentage of bone volume was only observed for 4 mm bone loss in the >3000 $\mu\epsilon$ window. For the 1 GPa and 4 GPa trabecular region elastic moduli, a higher volume of elements was observed for the 1 GPa in all strain windows, except for >3000 $\mu\epsilon$.

Presented in Tables 5–8 are the two-way variable interactions. In addition, Figs. 6–8 show the variables that presented significant effects on the different strain windows. In general, regardless of the bone loss amount, a cortical bone thickness of 4 mm presented

Table 2 Material properties used in the finite element analysis

Material	E (GPa)	ν
Trabecular bone	1 or 4	0.3
Cortical bone	13.7 ^a	0.3 ^a
Titanium	110 ^a	0.33 ^a

^aProperties taken from Ref. [17].

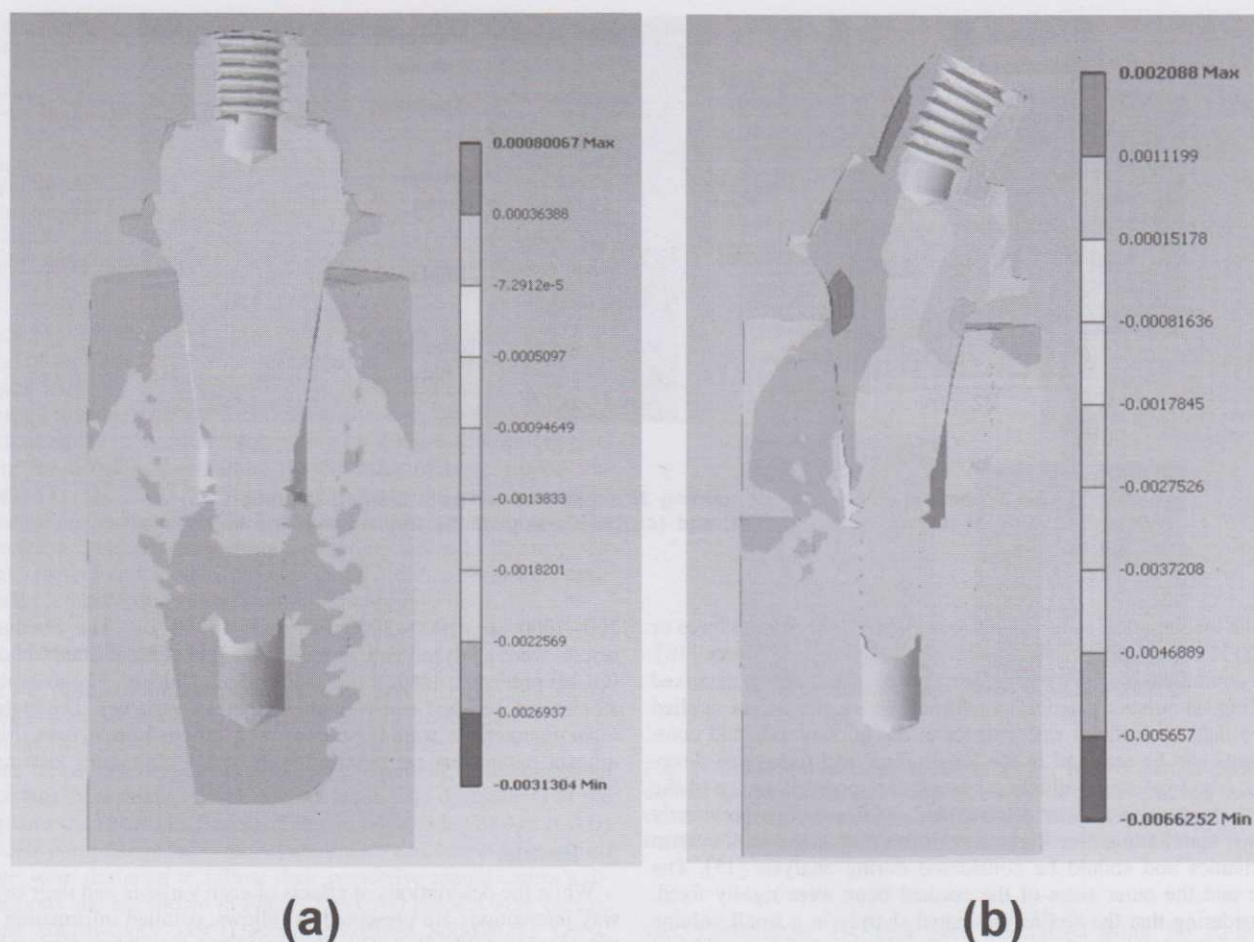


Fig. 4 A typical set of FEA results for the SIN implant system subjected to (a) axial loading and (b) mixed loading

Table 3 Significance levels of one-way interactions including implant design, loading condition, and cortical bone thickness

Interaction parameter	p-value for strain windows				Observations
	0–200 ($\mu\epsilon$)	200–1000 ($\mu\epsilon$)	1000–3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Implant design	<0.001	0.399	0.307	0.008	Only 0–200 $\mu\epsilon$ and >3000 $\mu\epsilon$ windows were significantly affected by implant design, presented in Fig. 5
Loading condition (axial or mixed)	<0.001	<0.001	<0.001	<0.001	0–200 $\mu\epsilon$: axial significantly lower compared with mixed 200–1000 $\mu\epsilon$: axial significantly higher compared with mixed 1000–3000 $\mu\epsilon$: axial significantly lower compared with mixed >3000 $\mu\epsilon$: axial significantly lower compared with mixed
Cortical bone thickness (1 mm or 4 mm)	<0.001	<0.001	<0.001	<0.001	0–200 $\mu\epsilon$: 4 mm cortical thickness presented significantly higher values than 1 mm cortical thickness 200–1000 $\mu\epsilon$: 4 mm cortical thickness presented significantly higher values than 1 mm cortical thickness 1000–3000 $\mu\epsilon$: 4 mm cortical thickness presented significantly lower values than 1 mm cortical thickness >3000 $\mu\epsilon$: 4 mm cortical thickness presented significantly higher values than 1 mm cortical thickness

Table 4 Significance levels of one-way interactions including cervical bone loss and trabecular bone elastic modulus

Interaction parameter	p-value for strain windows				Observations
	0–200 ($\mu\epsilon$)	200–1000 ($\mu\epsilon$)	1000–3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Bone loss (no cervical loss vs 4 mm bone loss)	<0.001	0.001	<0.001	<0.001	0–200 $\mu\epsilon$: significantly higher values were observed for no bone loss vs 4 mm bone loss 200–1000 $\mu\epsilon$: significantly higher values were observed for no bone loss vs 4 mm bone loss 1000–3000 $\mu\epsilon$: significantly higher values were observed for no bone loss vs 4 mm bone loss >3000 $\mu\epsilon$: significantly lower values were observed for no bone loss vs 4 mm bone loss
Trabecular bone elastic modulus (1 GPa or 4 GPa)	<0.001	<0.001	<0.001	<0.001	0–200 $\mu\epsilon$: significantly higher values were observed for 4 GPa vs 1 GPa modulus 200–1000 $\mu\epsilon$: significantly higher values were observed for 4 GPa vs 1 GPa modulus 1000–3000 $\mu\epsilon$: significantly higher values were observed for 4 GPa vs 1 GPa modulus >3000 $\mu\epsilon$: significantly lower values were observed for 4 GPa vs 1 GPa modulus

a higher bone volume in the 0–3000 $\mu\epsilon$ window while higher bone volumes in the >3000 $\mu\epsilon$ window were found for a cortical bone thickness of 1 mm. In addition, when the 4 mm bone loss is considered, a significantly higher percentage of bone volume in the >3000 $\mu\epsilon$ window for the 1 GPa trabecular bone elastic modulus was observed. The cortical bone thickness of 1 mm also have higher bone volumes in the >3000 $\mu\epsilon$ window regardless of the implant design.

4 Discussion

Adverse forces over the implant-supported prostheses can be directly transmitted to the periimplant bone, leading to microdam-

age accumulation and eventually the loss of the implant osseointegration [2–4]. On the other hand, a small stimulus has been demonstrated to improve the implant osseointegration [22,23]. Although precise determination of the loading level that separates mechanical loading into acceptable, osteogenic, or failure-inducing levels is difficult to determine and, until now, unresolved, some published studies focused on the strain amplitudes as a measure of the mechanical stimulus for the bone adaptive process [6,24,25]. Several additional biomechanical factors are recognized to influence the implant to bone load transfer, including bone quality in the insertion area, the nature of the bone-implant interface, the material properties of the implants and prosthesis,

Table 5 Significance levels for two-way interactions involving implant design with loading condition, cortical bone thickness, bone loss, and trabecular bone elastic modulus. Note that the gray cells represent nonsignificant differences.

Two-way interaction parameters	p-value for strain windows				Observations
	0–200 ($\mu\epsilon$)	200–1000 ($\mu\epsilon$)	1000–3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Implant design/loading condition	<0.001	0.085	0.108	0.169	0–200 $\mu\epsilon$: significantly higher values were observed for all systems, except the Straumann, when mixed loading condition was applied
Implant design/cortical bone thickness	<0.001	0.346	0.186	0.038	0–200 $\mu\epsilon$: all implant systems showed higher values for the 4 mm cortical thickness >3000 $\mu\epsilon$: significantly higher results for implants placed in a 1 mm cortical thickness
Implant design/bone loss	<0.001	0.178	0.495	0.064	0–200 $\mu\epsilon$: all implant systems presented significantly higher values for the 4 mm bone loss condition relative to no bone loss
Implant design/trabecular bone elastic modulus	<0.001	0.403	0.061	0.057	0–200 $\mu\epsilon$: significantly higher values were observed for all systems when trabecular bone modulus was 4 GPa compared with 1 GPa

Table 6 Significance levels for two-way interactions involving bone loss with cortical bone thickness and loading condition. Note that the gray cells represent nonsignificant differences.

Two-way interaction parameters	p-value for strain windows				Observations
	0–200 ($\mu\epsilon$)	200–1000 ($\mu\epsilon$)	1000–3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Cortical bone thickness/bone loss	<0.001	<0.001	0.914	<0.001	0–200 $\mu\epsilon$: regardless of bone loss amount, significantly higher values were observed for 4 mm cortical bone thickness 200–1000 $\mu\epsilon$: regardless of bone loss amount, significantly higher values were observed for 4 mm cortical bone thickness >3000 $\mu\epsilon$: regardless of bone loss amount, significantly higher values were observed for 1 mm cortical bone thickness
Loading condition/bone loss	0.178	<0.001	0.002	<0.001	200–1000 $\mu\epsilon$: values were not affected for the no bone loss condition, significantly higher values were observed for the axial loading compared with the mixed loading when 4 mm bone loss was considered 1000–3000 $\mu\epsilon$: for both bone loss conditions, significantly higher values were observed for the mixed vs axial loading condition >3000 $\mu\epsilon$: for both bone loss conditions, mixed loading showed higher values compared to axial loading

the surface roughness of the implant material, the occlusal condition (i.e., magnitude, direction, and frequency of loading), and the design of the implant [25–28].

The results of the present study showed that when the system variations other than implant design were evaluated separately, significant differences were observed in the bone volume occur-

ring in all strain windows due to load type, cortical bone thickness, cervical bone loss, and trabecular bone elastic modulus. For less favorable biomechanical scenarios, such as addition of a transverse load, decreased cortical bone thickness, increased cervical bone loss, or decreased trabecular bone elastic modulus, the results demonstrated complex shifts in the volumes in different

Table 7 Significance levels for two-way interactions involving cortical bone thickness with loading condition and trabecular bone elastic modulus. Note that the gray cells represent nonsignificant differences.

Two-way interaction parameters	p-value for strain windows				Observations
	0–200 ($\mu\epsilon$)	200–1000 ($\mu\epsilon$)	1000–3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Loading condition/cortical bone thickness	<0.001	0.438	0.114	<0.001	0–200 $\mu\epsilon$: significantly higher values were observed without the horizontal component for both 1 mm and 4 mm cortical thicknesses >3000 $\mu\epsilon$: significantly higher values were observed with the horizontal component for both 1 mm and 4 mm cortical thicknesses
Trabecular bone elastic modulus/cortical bone thickness	<0.001	<0.001	<0.001	<0.001	0–200 $\mu\epsilon$: significantly higher values were observed for both cortical bone thicknesses when trabecular bone modulus was 4 GPa 200–1000 $\mu\epsilon$: for both trabecular bone elastic moduli, 4 mm cortical bone thickness showed higher values 1000–3000 $\mu\epsilon$: significantly higher values for both trabecular bone elastic moduli were observed for 1 mm cortical bone thickness >3000 $\mu\epsilon$: for both trabecular bone elastic moduli, the 1 mm thick cortical bone showed significantly higher values

Table 8 Significance levels for two-way interactions involving trabecular bone elastic modulus with loading condition and bone loss. Note that the gray cells represent nonsignificant differences.

Two-way interaction - parameters	p-value for strain windows				Observations
	0-200 ($\mu\epsilon$)	200-1000 ($\mu\epsilon$)	1000-3000 ($\mu\epsilon$)	>3000 ($\mu\epsilon$)	
Loading condition/trabecular bone elastic modulus	<0.001	0.276	0.003	0.001	0-200 $\mu\epsilon$: regardless of loading condition, higher values were observed for the 4 GPa compared with the 1 GPa trabecular bone elastic modulus 1000-3000 $\mu\epsilon$: regardless of loading condition, higher values were observed for the 1 GPa compared with the 4 GPa trabecular bone elastic modulus >3000 $\mu\epsilon$: regardless of loading condition, higher values were observed for the 1 GPa compared with the 4 GPa trabecular bone elastic modulus
Bone loss/trabecular bone elastic modulus	<0.001	0.001	0.320	<0.001	0-200 $\mu\epsilon$: when no bone loss is considered, significantly higher values were observed for the 4 GPa vs the 1 GPa trabecular bone elastic modulus, and when 4 mm crestal bone loss is considered, no difference in values between both trabecular bone elastic moduli 200-1000 $\mu\epsilon$: for both bone loss situations, significantly higher values were observed for the 4 GPa vs the 1 GPa trabecular bone elastic modulus >3000 $\mu\epsilon$: when no bone loss is considered, no difference was observed between trabecular bone elastic moduli, when 4 mm crestal bone loss is considered, significantly higher values were observed for the 1 GPa vs the 4 GPa trabecular bone elastic modulus

strain windows, and that these shifts did not necessarily occur toward higher strains. Such observation supports the possibility of dynamic process of bone remodeling over time as bone resorption and apposition is taking place to pursue a potential biomechanical equilibrium. These findings are in agreement with the clinical studies reporting predictable outcomes and long-term success for

osseointegrated implants independently on the clinical scenarios [29,30]. The results from the factorial analysis showed that all the parameter variations considered in the present study significantly affect the volume percentage of bone in at least one strain window. However, all the strain windows were not affected in a similar manner for all scenarios. The mixed loading presented a 4-10% higher bone volume in 0-200 $\mu\epsilon$, 1000-3000 $\mu\epsilon$, and >3000 $\mu\epsilon$ windows for comparable models under axial loading.

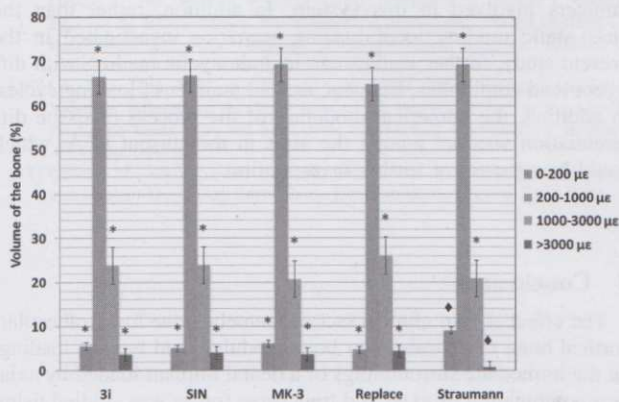


Fig. 5 Statistical summary concerning the effect of implant design in the four different strain windows. Note that significant differences occurred only for the 0-200 $\mu\epsilon$ and >3000 $\mu\epsilon$ windows for the different systems. Statistically homogeneous groups are identified within each strain window and marked with * and ♦ for the two different groups. Mean $\pm$ 95% confidence interval.

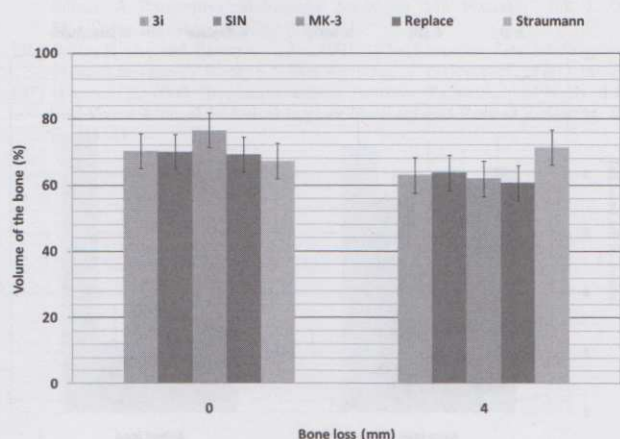


Fig. 6 Variation of volume percent of bone affected by implant design-bone loss two-way interaction in the 200-1000 $\mu\epsilon$ window

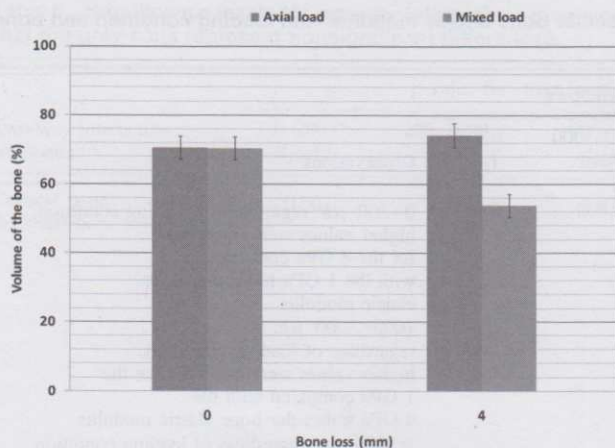


Fig. 7 Variation of volume percent of bone affected by loading condition-bone loss two-way interaction in the 200–1000 $\mu\epsilon$ window

This finding is in agreement with previous studies [31], where in, as demonstrated by finite element analysis (FEA), oblique occlusal forces produce a lateral bending moment that significantly increases the periimplant stresses when compared with axial occlusal forces. In addition, offset loading was found to increase the magnitude of strains compared with the axial loading in previous studies [32]. In the present study, vertical loading condition presented significantly higher volume only in the 200–1000 $\mu\epsilon$ window. Thus, prudent control of the biomechanical load magnitude and direction on dental implants is imperative to achieve long-term clinical success [33]. In the presence of a marginal bone loss, the lever arm of force will be increased, so that the moment with respect to the marginal bone level will also be increased [14]. Correspondingly, the present study also demonstrated a higher percentage of bone volume in the >3000 $\mu\epsilon$ window for 4 mm bone loss, compared with the no bone loss situation.

Regarding the influence of the bone quality, the bone volume in the higher strain windows increased with the decrease in cortical bone thickness and elastic modulus. Low-density bone has low stiffness, causing higher implant displacement, sinking, and tilting under axial and oblique loads [34]. The greater implant displacement led to increased bone deformation and caused higher stresses and strains in the cortical and cancellous bones. This result corroborates the clinical findings in which higher failure rates were observed for type 4 bone than for types 1–3 bones [35,36].

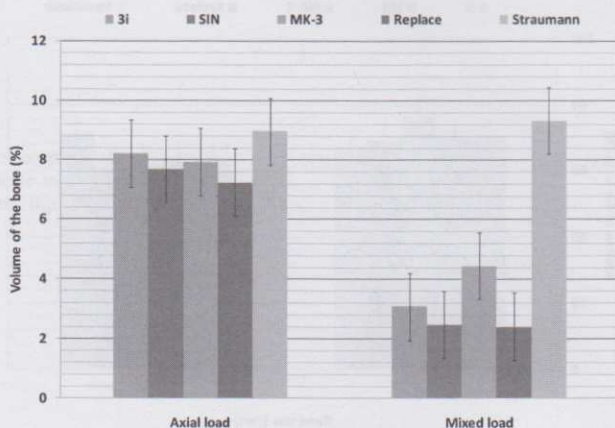


Fig. 8 Variation of volume percent of bone affected by implant design-loading condition two-way interaction in the 0–200 $\mu\epsilon$ window

The results on the selected representative implants showed that the implant design only affected the 0–200 $\mu\epsilon$ and >3000 $\mu\epsilon$ windows, which are ranges where bone loss occurs due to disuse and overloading, respectively. While all other systems presented comparable numbers in both windows, the Straumann Standard implant presented the highest and lowest percentage bone volumes for the 0–200 $\mu\epsilon$ and >3000 $\mu\epsilon$ windows, respectively. This likely results from the fact that strain distribution for osseointegrated implants is uneven and a large part of the bone volumes that appears in the >3000 $\mu\epsilon$ window is located around the implant neck. Hence, the characteristic of the cervical portion of the implants, especially the connection type, may have a strong influence on the strain values. These results are in agreement with other FEA studies [15,37], which showed that a conical implant-abutment interface at the level of the bone crest decreases the peak bone-implant interfacial stress as compared with the flat top surface. However, although having a low bone volume in the >3000 $\mu\epsilon$ window is favorable, it may be counteracted by having a significantly high bone volume in the 0–200 $\mu\epsilon$ window.

Compared with the existing studies that consider one variable at a time, the present study focused on understanding the two-way interactions between numbers of biomechanical parameters [17,18]. The two-way interactions from the factorial analysis showed that the combinations that have been suggested to be clinically deleterious, such as cervical bone loss, lower cortical bone thickness, and presence of a horizontal loading component, all associated with low trabecular bone modulus, did not necessarily result in detrimental strain levels. Moreover, the results showed that for some two-way interactions, shifts occurred toward lower strain levels. Interestingly, a lower number of significant differences were observed when implant type was present in two-way interactions, suggesting that anatomic conditions may have a stronger effect on the system biomechanics than the implant design itself. On the other hand, the assumptions made during the process of developing the finite element model, principally regarding the material properties, the interface conditions, and the simplification of bone anatomy, limit the validity of the absolute values of the stress/strain and displacement calculated in the analysis. Nevertheless, applying the FEA method along with statistical analysis has provided a qualitative insight into the effect of various parameters and their two-way interactions on the strain levels in the bone immediately surrounding the implant. It is possible to extend the analysis to include higher order interactions and include more parameters such as implant-bone interfacial bonding, implant surface roughness, and other loading conditions. However, the number of simulations required and the computational cost involved may be prohibitive, given the number of parameters involved in this system. In addition, rather than the quasi-static unidirectional loading scenarios investigated in the present study, further studies can include cyclic loading with different load amplitudes, frequencies, and number of loading cycles. In addition, the numerical modeling of the process of tissue differentiation was not among the aims in the current FEA, which could be a matter of further investigation.

5 Conclusions

The effect of four clinical factors, namely, bone loss, trabecular/cortical bone ratio, trabecular bone modulus, and type of loading, on the immediate surroundings of a dental implant loaded by axial or a combination of axial and transverse forces was studied using FEA and factorial analysis.

Within the limitations of the present methodology, where experimental validation of the results was not part of this study, it was concluded that although some unfavorable clinical scenarios demonstrated a higher volume of bone in deleterious strain levels, a tendency toward the biomechanical equilibrium was evidenced regardless of the implant design.

Acknowledgment

This research is supported by the National Science Foundation through Grant No. CMMI-0726723 and the NYU-Poly-NYU joint seed grant. The Department of MAE at NYU-Poly is acknowledged for the support and facilities provided. Dr. Nguyen Q. Nguyen and Mr. Dung Dinh Luong are thanked for their valuable help. The authors acknowledge the support of Ansys, Inc.

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