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Bivariate Evaluation of Cylinder Implant Diameter and Length: A Three-Dimensional Finite Element Analysis

Liang Kong, DDsc,[1] Yingying Sun, DDsc,[1] Kaijin Hu, DDsc,[1] Dehua Li, DDsc,[2] Rui Hou, DDsc,[1] Jin Yang, MSc,[3] & Baolin Liu, DDsc[2]

[1] Department of Oral and Maxillofacial Surgery, School of Stomatology, the Fourth Military Medical University, Xi'an, China
[2] Department of Implantology, School of Stomatology, the Fourth Military Medical University, Xi'an, China
[3] Department of Civil Engineering, Northwestern Polytechnical University, Xi'an, China

Keywords
Dental implant; three-dimensional finite element analysis; bivariate; stress; implant diameter; implant length.

Correspondence
Dr. Baolin Liu, Department of Implantology College of Stomatology, the Fourth Military Medical University, Changlexi Street 145, Xi'an 710032, People's Republic of China.
E-mail: Baolin@fmmu.edu.cn

Because of equal contributions to the paper, Dr. Liang Kong and Yingying Sun should be regarded as co-first authors and are listed alphabetically.

Accepted November 27, 2006.

doi: 10.1111/j.1532-849X.2007.00286.x

Abstract

Purpose: To evaluate continuous and simultaneous variations of implant diameter and length for an experimental cylinder implant.
Materials and Methods: A finite element model of a mandible segment with implant was created. The range of implant diameter (D) was set from 2.5 to 5.0 mm, and that of implant length (L) from 6.0 to 16.0 mm. The maximum Von Mises stresses in the mandible were evaluated, and the sensitivity of the stresses in the mandible to the variables was also evaluated.
Results: Under axial load, the maximum von Mises stresses in cortical and cancellous bones decreased by 73.3% and 69.4%, respectively, with D and L increasing. Under buccolingual load, those decreased 83.8% and 79.2%, respectively. When D exceeded 3.9 mm and L exceeded 10.0 mm, the tangent slope rate of the maximum von Mises stress response curve ranged from -1 to 0. The variation of the maximum von Mises stresses in the mandible was more sensitive to D than to L.
Conclusions: Buccolingual force is apt to be influenced by the two implant parameters; implant diameter and length favor stress distribution in cortical bone and cancellous bone, respectively. Implant diameter exceeding 3.9 mm and implant length exceeding 10.0 mm are the optimal choice for type B/2 bone in a cylinder implant. The implant diameter is more important than length in reducing bone stress.

A key factor for the success or failure of a dental implant is the manner in which stresses are transferred to the surrounding bone. The interface must tolerate the occlusal forces without adverse tissue response.[1] In natural teeth, the periodontal ligament acts as an intermediate cushion element to buffer the occlusal forces; however, in the osseointegrated dental implant, occlusal loads are transmitted directly to the surrounding bones. This could cause microfracture at the interface between the bones and implant, fracture of implant, loosening of components of the implant system, and unwanted bone resorption.

Several previous researchers have attempted to minimize the crestal bone resorption by increasing the contact area of bone-to-implant interface and therefore reducing stress at the cortical alveolar crest. Attempts to increase the contact area of bone-to-implant interface have focused on increasing the diameter and/or the length of the implant, the shape and characteristics of the implant surface, or altering the fixture design/shape.[2-9]

Therefore, it is necessary to understand how the stress concentration on jaw bones is affected by different types of loading, the diameter and length of the implants, the shapes and characteristics of the implant surface, and the prosthesis type. The use of the finite element method (FEM) in implant biomechanics analysis offers many advantages over other methods in simulating the complexity of clinical situations. That allows researchers to predict stress distribution in the contact area of the implants with cortical bone and around the apex of the implants in cancellous bone;[10,11] however, many of the previous finite element studies examined the effect of implant design parameters discretely and independently; therefore, the information about implant design parameters was not accurate, and some important information was lost.[11,12]

The main aim of this study was to find a new method of three-dimensional (3D) finite element analysis (FEA) on continuous and simultaneous variations of implant diameter and length, and to find optimal implant parameters under idealized axial and buccolingual loads, ensuring lower stress peaks in the jaw bone and enhancing further clinical success.

Materials and methods

The study was performed by means of 3D FEA.[13] The theory of elasticity was applied.[14]

3D model design

A posterior mandible segment with an implant and a superstructure were modeled on a personal computer, using a 3D program (Pro/E Wildfire, Parametric Technology Corporation, Needham, MA). A cross-section of a mandible in the first premolar region was used as the basis for a solid model, and then the cross-sectional image was extruded to create a 3D mandible segment. This section had a thick layer of cortical bone surrounding dense cancellous bone, that is, type B/2 bone according to the Lekholm and Zarb classification.[15] The thickness of cortical bone in the crestal region varied from 1.9 to 3.1 mm; the mesial and distal section planes were not covered by cortical bone. The dimensions of the bone segment are shown in Figure 1.

A cylinder implant and a 5-mm high solid abutment were modeled and simplified to one unit, as shown in Figure 1. A porcelain superstructure with 2-mm occlusal thickness was applied over the titanium abutment (Fig 1). The diameter of implant (D) and length of implant (L) were set as the input variables. D ranged from 2.5 to 5.0 mm, and L ranged from 6.0 to 16.0 mm. All models were meshed by Ansys Workbench 10.0 (SAS IP, Inc., Cary, NC).

Material properties

All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. The elastic properties were taken from the literature, as shown in Table 1.

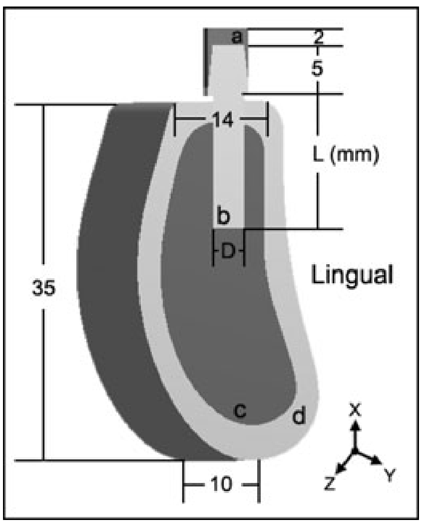

Figure 1 Cross-sectional view of the symmetry plane of one model. a = superstructure; b = implant and abutment; c = cancellous bone; d = cortical bone; D = diameter of implant (ranged from 2.5 to 5.0 mm); L = length of implant (ranged from 6.0 to 16.0 mm).

Table 1 Elastic properties of materials in the 3D FEM models

Materials[Reference]	Young's modulus (MPa)	Poisson ratio
Cortical bone[27]	14,000	0.30
Cancellous bone[28]	1370	0.30
Titanium[29]	110,000	0.35
Porcelain[30]	68,900	0.28

Interface conditions

The implant was rigidly anchored in the bone model along its entire interface. The same type of contact was provided at the prosthesis–abutment interface.

Elements and nodes

The models were meshed by 10-node-tetrahedron and 20-node-hexahedron elements. A refinement mesh was generated around the implant (Fig 2). Models were composed of an average of 33,000 elements and 56,000 nodes.

Constraints and loads

Models were constrained in all directions at the nodes on the mesial and distal bones. Because this study aimed at investigating bone effects to loads within the physiological limits, rather than to overloads, forces of 200 N and 100 N were applied axially (AX) and buccolingually (BL), respectively, to the middle point in the center of the superstructure.[16,17] The analysis of each load was performed by means of the Ansys Workbench software program. The maximum von Mises stresses (maximum equivalent stress, or "Max EQV stress") in the cortical and cancellous bones were set as output variables to evaluate the effect of different designs on the mandible. The sensitivity of the stresses in the mandible to the variables was also evaluated.

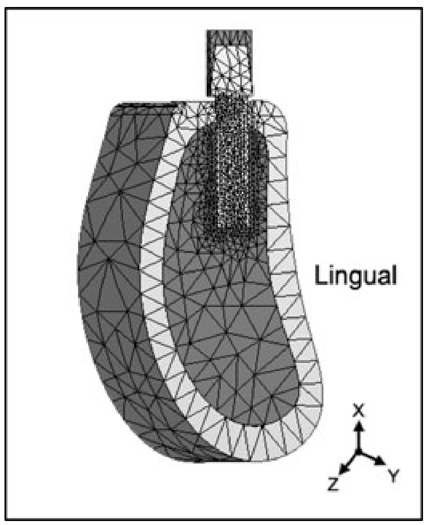

Figure 2 Cross-sectional view of the symmetry plane of one meshed model.

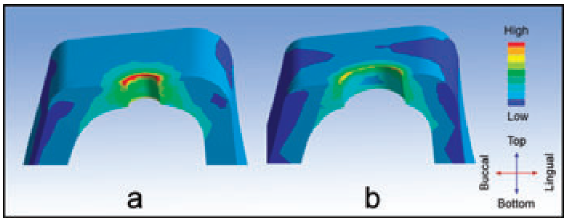

Figure 3 Cross-sectional view of EQV stress distribution in the cortical bone under AX loads. For comparison, the same scale was used in all models. (a: D = 2.5 mm, L = 6.0 mm; b: D = 5.0 mm, L = 16.0 mm).

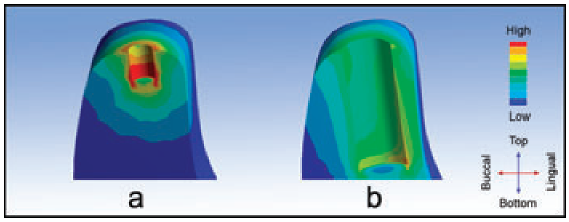

Figure 5 Cross-sectional view of EQV stress distribution in the cancellous bone under AX loads. For comparison, the same scale was used in all models: (a) D = 2.5 mm, L = 6.0 mm; (b) D = 5.0 mm, L = 16.0 mm).

Results

The stress distributions of the cortical and cancellous bone are shown in Figures 3–6. The bivariates to Max EQV stress in the mandible are shown as response surface charts with different colors between certain ranges (Figs 7–10). When one variable is equal to the median, the response curves of the other variable to the Max EQV stress are shown in Figure 11. Because the sensitivities of the mandible to the variables are similar in full range, the sensitivity chart (D = 3.75 mm, L = 11 mm) is shown in Figure 12. All figures were drawn by Ansys Workbench DesignXplorer module.

It is known that when a straight line is tangent to a curve, the slope rate of the straight line shows the changing frequency of the curve. When the slope rate ranges from −1 to 1, it indicates the slight changing of the Max EQV stress to variables (Fig 13); therefore, the optimum cylinder implant parameters should be selected in this range.

Stress distribution

In all loading situations, the highest stress in the bone, as a whole, was concentrated in the cortical bone, around the implant. Because of a great difference between the stress values in the cortical and cancellous bone, the stress distributions in these bone regions are shown separately for better visualization.

In all models under AX load, the highest EQV stress of the cortical bone was observed around the implant neck. High stress surrounded the implant neck like a ring. The distribution of the EQV stress was similar for all models (Fig 3). Under BL load, the highest EQV stress was observed buccally and lingually near the implant neck in all models. The distribution of the EQV stress was similar for all models (Fig 4).

In all models under AX load, the highest EQV stress of the cancellous bone was observed at the implant bottom in all the models, but its value was much lower than in the cortical bone (Fig 5). Under BL load, the highest EQV stress occurred near the cortical plates on the buccal and lingual sides. The EQV stress on the buccal sides showed much higher than that on the lingual sides for all models (Fig 6).

Under AX load in cortical bone

As D and L increased, Max EQV stress in cortical bone decreased, ranging from 47.5 to 12.7 MPa, and reduced by 73.3% (Fig 7). The tangent slope rate of response curve ranged from −1 to 0 when D exceeded 3.7 mm or L exceeded 10.0 mm (Fig 11). From sensitivity analysis, D affected the Max EQV stress of cortical bone more than L did (Fig 12).

Under AX load in cancellous bone

Max EQV stress in cancellous bone decreased by 69.4% with D and L increased, and ranged from 4.81 to 1.47 MPa (Fig 8). When L exceeded 10.0 mm, the tangent slope rate of response curve ranged from −1 to 0. The tangent slope rate of response curve reached about −1 when D ranged from 2.5 to 5.0 mm (Fig 11). D and L affected Max EQV stress in cancellous bone similarly (Fig 12).

Under BL load in cortical bone

Max EQV stress in cortical bone ranged from 169.0 to 27.4 MPa and decreased by 83.8% with D and L increased (Fig 9). The tangent slope rate of response curve ranged from −1 to 0 when D exceeded 3.5 mm or L exceeded 10.0 mm (Fig 11). In comparison, D affected the Max EQV stress of cortical bone more than L did (Fig 12).

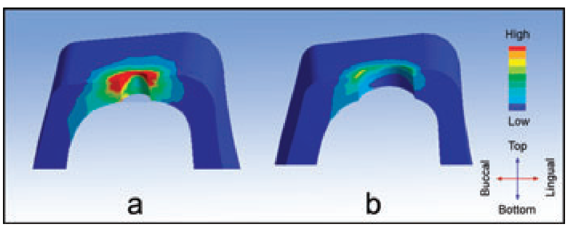

Figure 4 Cross-sectional view of EQV stress distribution in the cortical bone under BL loads. For comparison, the same scale was used in all models: (a) D = 2.5 mm, L = 6.0 mm; (b) D = 5.0 mm, L = 16.0 mm.

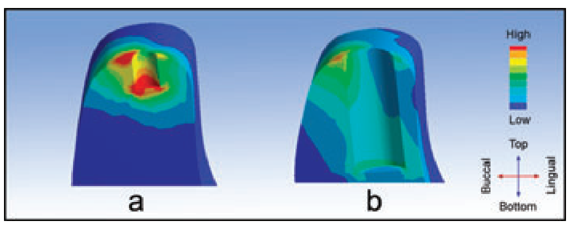

Figure 6 Cross-sectional view of EQV stress distribution in the cancellous bone under BL loads. For comparison, the same scale was used in all models. (a) D = 2.5 mm, L = 6.0 mm; (b) D = 5.0 mm, L = 16.0 mm.

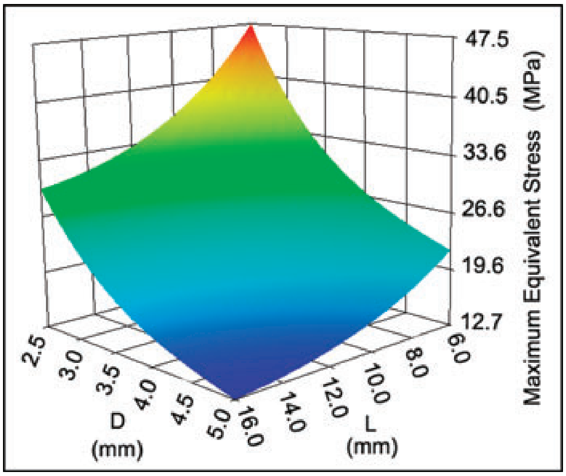

Figure 7 Under AX load, response surface nephogram of variable D and L to Max EQV stresses in cortical bone.

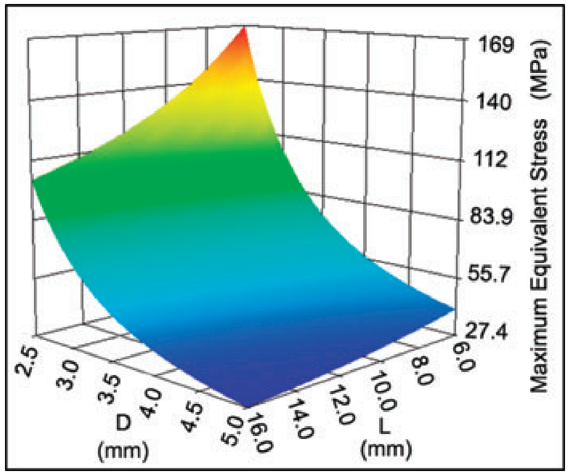

Figure 9 Under BL load, response surface nephogram of variable D and L to Max EQV stresses in cortical bone.

Under BL load in cancellous bone

Max EQV stress in cancellous bone ranged from 6.15 to 1.28 MPa and decreased by 79.2% with D and L increased (Fig 10). The tangent slope rate of response curve ranged from −1 to 0 when D exceeded 3.9 mm or L exceeded 9.0 mm (Fig 11). D and L affected Max EQV stress in cancellous bone similarly (Fig 12).

Discussion

The aim of the present study was to find the pure effect of the variations of implant diameter and length upon bone stresses. For this reason, it was assumed that all parameters of the models were identical except implant diameter and length. This made it possible to make a comparison between implants of different diameters and lengths. In this study, the Workbench Simulation module was used to define the environmental loading conditions of the model. An optimized implant parameter design was selected by the DesignXplorer module, which uses parameter as its basic language.

There are three key points in this new FEM: *self-adapting 3D models assembling, bidirectional parameters transmitting*, and *variable settings*. In this study, *self-adapting 3D models assembling* means all the models were rebuilt based on implant parameters. In other words, the parameters of other models (cortical and cancellous bone) changed, with the parameters of implant varying automatically. *Bidirectional parameters transmitting* means CAD and CAE software (Pro/E and Ansys Workbench in this study) could transmit a model's parameters mutually and seamlessly. *Variable settings* include input variables (diameter and length in this study) and output variables (Max EQV stresses in cortical and cancellous bones in this study). Therefore, only one assembled model was needed, and the time of model regeneration and solving process were shortened.

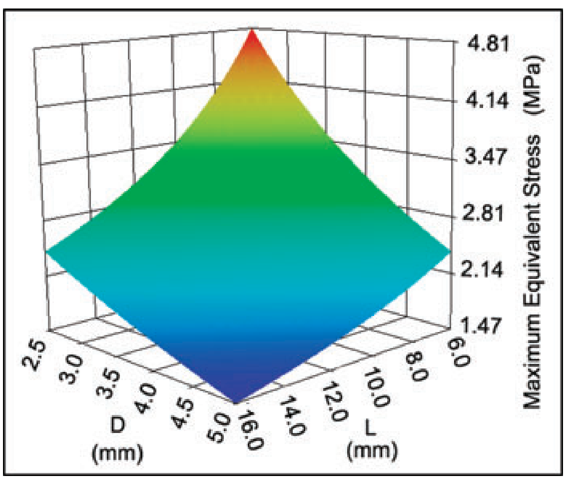

Figure 8 Under AX load, response surface nephogram of variable D and L to Max EQV stresses in cancellous bone.

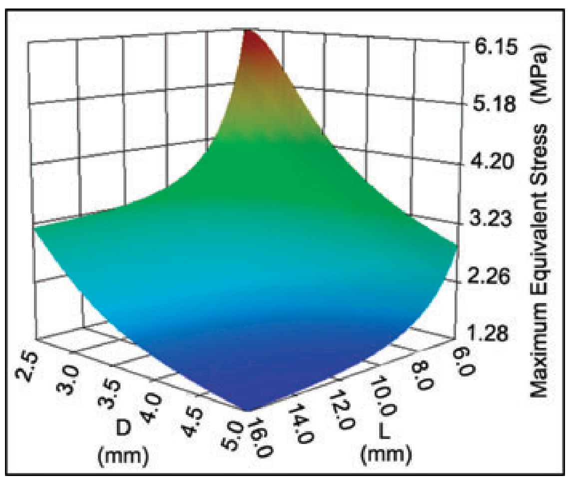

Figure 10 Under BL load, response surface nephogram of variable D and L to Max EQV stresses in cancellous bone.

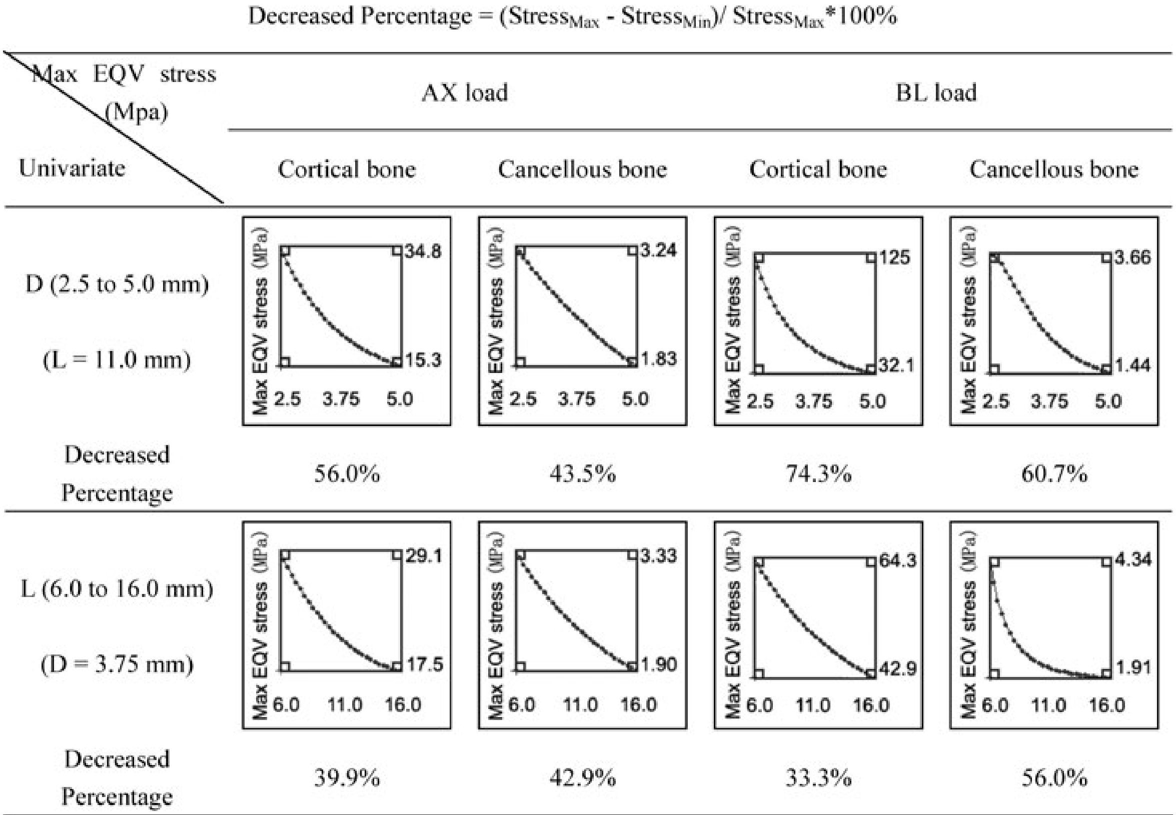

Figure 11 Response curve of univariate to Max EQV stresses in the mandible.

Furthermore, the result could be shown as response surface (Figs 7–10), response curve (Fig 11), and sensitivity chart (Fig 12). Other input and output parameters, such as thread height, thread pitch, superstructure thickness, elastic properties, strains in mandible, shear strains in mandible, loading forces, etc., could be evaluated simultaneously or respectively in future studies.

In this study, nine analysis results were performed to construct the response surfaces (Fig 7–10). The samplings of this research are listed in Table 2. In a DesignXplorer environment, the sample generation is based on the Latin hypercube sampling (LHS) technique. The LHS technique is a more advanced and efficient form of Monte Carlo simulation methods. The only difference between LHS and the direct Monte Carlo sampling technique is that the LHS has a sample "memory," meaning it avoids repeating samples that have been evaluated before (it avoids clustering samples). It also forces the tails of a distribution to participate in the sampling process. Generally, the LHS technique requires 20–40% fewer simulation loops than the direct Monte Carlo simulation technique to deliver the same results with the same accuracy.

Ansys DesignXplorer can also provide sensitivity charts to allow a user to see the impact of the input parameters on the response and derived parameters. The sensitivity charts are "single parameter sensitivities." This means that DesignXplorer calculates the change of the output based on the change of each

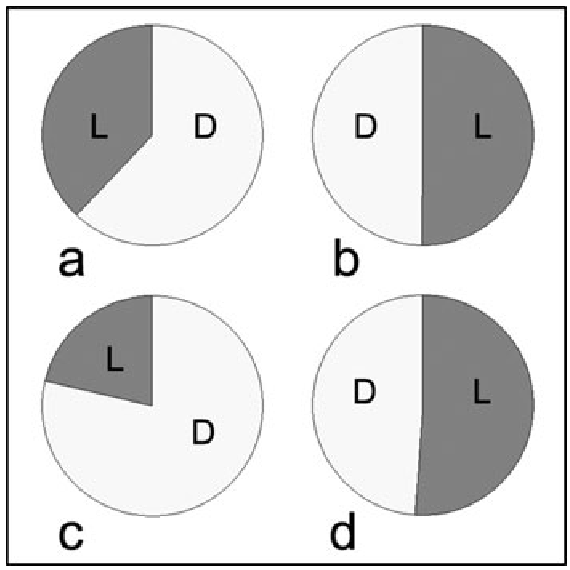

Figure 12 The sensitivity analysis of Max EQV stresses in mandible to variable D and L: D = 3.75 mm, L = 11 mm (a = under AX load, Max EQV stresses in cortical bone to variable; b = under AX load, Max EQV stresses in cancellous bone to variable; c = under BL load, Max EQV stresses in cortical bone to variable; d = under BL load, Max EQV stresses in cancellous bone to variable).

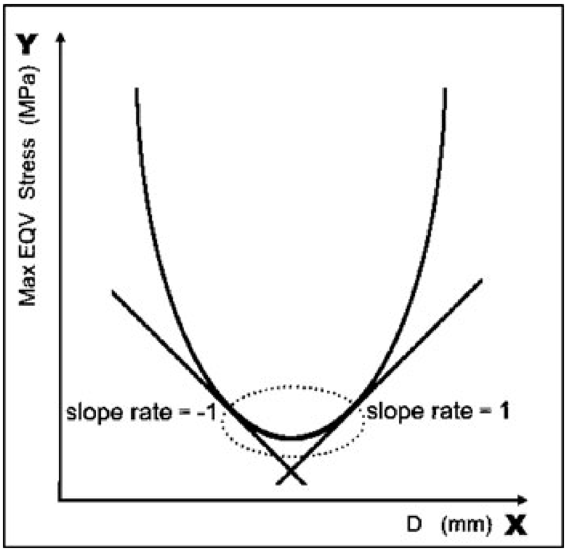

Figure 13 Chart of the optimum selection of the curve: slight changing and minimal value of the curve.

input independently at the current value of each input parameter. The larger the change of the output, the more significant is the input parameter that was varied. As such, single parameter sensitivities are local sensitivities. Changing the input parameter values will update the sensitivities. When the input parameter (D and L) changed, sensitivities charts changed little in this study. So when one of the input parameters was set (D = 3.75 mm; L = 11.0 mm), the corresponding sensitivity chart was selected to show the output sensitivity to input in the full range (Fig 12).

The use of FEM in implant biomechanics analysis offers many advantages over other methods in simulating the complexity of clinical situations; however, because of simplifications intrinsic to FEM, it is advisable to focus on qualitative rather than quantitative data from these analyses.[12] Different from the discrete variations of previous finite element studies, continuous variations of the two investigated factors were shown as response surface and curve in this study. More accurate and visualized results and more qualitative information about the design parameters were achieved.[11] Furthermore, the results of this study show that the effects of the two investigated factors (D and L) on Max EQV stress in jaw bone are likely to be interrelated. The effect of each of the two variables on Max EQV stress in jaw bone cannot be analyzed independently. This is another important finding in this study, because many of the previous finite element studies examined the effect of only one implant design parameter.[12] The conclusions of previous finite element studies should be reconsidered in light of these two findings.

van Eijden reported that in normal dentition without implants, mean maximal vertical (axial) bite force magnitudes in humans could be 469 ± 85 N at the region of the canines, 583 ± 99 N at the second premolar region, and 723 ± 138 N at the second molar. In general, maximal bite force in medial and posterior directions was larger than that in corresponding lateral and anterior directions, respectively.[18] As this study aimed

Table 2 Max EQV stresses in the mandible of the samplings (MPa)

			Max EQV stress in cortical bone		Max EQV stress in cancellous bone	
	D (mm)	L (mm)	AX load	BL load	AX load	BL load
1	3.75	11.0	20.555	46.252	2.2986	2.0753
2	3.75	6.0	29.910	68.914	3.5104	4.3461
3	3.75	11.0	16.564	43.952	1.7817	1.9444
4	2.50	11.0	34.534	129.96	3.3160	3.6641
5	5.00	11.0	15.583	31.102	1.9541	1.4678
6	2.50	6.0	47.570	168.06	4.7986	5.3440
7	2.50	16.0	30.506	99.562	2.5042	3.6358
8	5.00	6.0	21.334	39.387	2.2689	2.8574
9	5.00	16.0	13.212	28.973	1.5014	1.2659

at investigating bone effects to loads within the physiological limits, rather than to overloads, half of maximal bite force of 200 N was applied axially to the middle point in the center of the superstructure. In this study, we also assumed the buccolingual force was half of the axial force referred to in Kitamura et al's research.[16]

In general, the use of short implants has not been recommended, because it is believed that the occlusal forces must be dissipated over a large implant area for preservation of the bone. Some clinical studies demonstrated that the success rate of an implant was proportional to the implant length.[19] At the same time, other studies showed that large implant diameters provided more favorable stress distributions,[20,21] and stresses in cortical bone decreased in inverse proportion to an increase in implant diameter with both vertical and lateral loads.[20] Several clinical studies reported higher survival rates and reduced crestal bone loss for wide-diameter implants.[22,23] Petrie and Williams[24] and Meijer et al[25] observed that the length of implant had less influence on the amount of stress levels than diameter did. Rangert et al also reported that patients with fractured implants were diagnosed to have parafunctional activities, and all implants were 3.75 mm in diameter.[26] On the other hand, there were no reports of International Team for Implantology (ITI) standard 4.1-mm diameter solid screw implant fractures. Thus, in addition to well-defined factors leading to implant fracture, the diameter of implant could also be a principal factor.

Bivariate analyses

Based on the bivariate response surface of the current study, with the increasing of the implant diameter and length, Max EQV stress in cortical bone decreased by 73.3% and 83.8% under AX and BL loads, respectively. The value is much higher than in cancellous bone. It indicates that the effect of implant diameter and length on Max EQV stress in cortical bone is more significant than on that in cancellous bone. Under BL load, the value of Max EQV stress in cortical and cancellous bones decreased by 83.8% and 79.2%, respectively. The value is much higher than under AX load, indicating that the buccolingual force is apt to be influenced by the two parameters.

Univariable analyses

By the analysis of univariable to Max EQV stress from Figure 11, with the increase of implant diameter, Max EQV stress in cortical bone decreased by 56.0% and 74.3% under AX and BL loads, respectively. Max EQV stress in cancellous bone decreased by 43.5% and 60.7% under AX and BL loads, respectively. All the decreased percentages were higher with increase of implant diameter than that with an increase of implant length. The results demonstrate that implant diameter affects the Max EQV stress of the mandible more than length does under AX or BL load. Also, implant diameter favors stress distribution in cortical bone more than it does in cancellous bone.

On the other hand, the value of Max EQV stress in cancellous bone decreased by 42.9% and 56.0% under AX and BL loads, respectively, with the increasing of the implant length. Percentage decreased was higher than that in cortical bone, indicating that implant length favors stress distribution in cancellous bone more than it does in cortical bone.

By the analysis of tangent slope rate of the univariable response curve, when implant diameter exceeded 3.7 mm and length exceeded 10.0 mm, the most stable stress in cortical bone could be achieved and the stress value reached about the minimal level. When implant diameter exceeded 3.9 mm, and length exceeded 10.0 mm, the most stable stress in cancellous bone could be achieved, and the minimal stress value was found.

Sensitivity analyses

Similar to the univariable analysis, implant diameter affects the Max EQV stress of jaw bone much more than length does by the sensitivity analysis of Max EQV stress to variables. It indicates that in reducing bone stress, the implant diameter is more important than length, and improvement of horizontal bone quality may be more effective than improvement of vertical bone quality.

Conclusions

Based on the results from numerical analyses, the following conclusions are obtained from the effects of implant diameter and length of osseointegrated implant on stress distributions in the mandible:

(1) Buccolingual force is apt to be influenced by the two implant parameters.
(2) Implant diameter and length favor stress distribution in cortical bone and in cancellous bone, respectively.
(3) Implant diameter exceeding 3.9 mm and implant length exceeding 10.0 mm are the optimal choices for type B/2 bone in a cylinder implant.
(4) Implant diameter is more important than length in reducing bone stress.

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