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Original Article

Effects of diameters of implant and abutment screw on stress distribution within dental implant and alveolar bone: A three-dimensional finite element analysis

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KEYWORDS
Abutment screw;
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Abstract *Background/purpose:* Few studies have investigated the effects of abutment screw diameter in the stress of dental implants and alveolar bones under occlusal forces. In this study, we investigated how variations in implant diameter, abutment screw diameter, and bone condition affect stresses in the abutment screw, implant, and surrounding bone.

Materials and methods: Three-dimensional finite element (FE) models were fabricated for dental implants with external hex-type abutments measuring 4 and 5 mm in diameter. The models also included abutment screws measuring 2.0 and 2.5 mm in diameter. Each implant model was integrated with the mandibular bone comprising the cortical bone and four types of cancellous bone. In total, 12 finite element models were generated, subjected to three different occlusal forces, and analyzed using FE software to investigate the stress distribution of dental implant and alveolar bone.

Results: Wider implants demonstrated lower stresses in implant and bone compared with standard-diameter implants. The quality of cancellous bone has a minimal impact on the stress values of the implant, abutment screw, and cortical bone. Regardless of occlusal arrangement or quality of cancellous bone, a consistent pattern emerged: larger abutment screw diameters led to increased stress levels on the screws, while the stress levels in both cortical and cancellous bone showed comparatively minor fluctuations.

Conclusion: Wider implants tend to have better stress distribution than standard-diameter implants. The potential advantage of augmenting the abutment screw diameter is unfavorable. It may result in elevated stresses in the implant system.

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Introduction

Brånemark System implants, first introduced in 1965, are extensively used in dental treatments. The average 5-year success rate for dental implant procedures exceeds 95 %, which confirms the high reliability of these prosthetic treatment modalities.[1] These procedures have been continually improved over the years in response to various clinical conditions. These improvements include the refinement of implant shapes, diameters, surface treatments, and prosthetic components.[2–5] Excessive bending moments have been associated with implant system failures, including abutment screw loosening, implant fractures, and other complications.[6–12] A substantial portion of implant fractures (90 %) occur in the premolar and molar regions,[13] implicating overloading as a factor contributing to implant fracture. Hence, biomechanics plays a pivotal role in the clinical application of implant prosthodontics.

The Brånemark System implants, demonstrated that the most effective strategy for enhancing implant strength involved increasing the implant diameter.[14] Increasing the diameter from 3.75 to 4.0 mm resulted in a 30 % increase in fatigue resistance.[15] However, loosening of abutment screws, which were made of titanium and could withstand loads of 10 and 20 N, became an issue.[16] Consequently, gold alloy abutment screws became popular. These screws could withstand a torque of 32 N·cm, addressing the problem related to abutment screw loosening.[17] However, increasing the diameter of the implant or adjusting the preload on the abutment screw did not completely resolve the problem; instead, it shifted the problem to a weaker part of the overall system.[13] In the OSSEOTITE Parallel Walled 3i implant system, grade 1 pure titanium and a cold-worked technique are used to enhance the material's yield strength without altering its elastic modulus, maintaining equivalence with pure titanium in terms of physical properties.[18,19] Only few studies investigated the effects of abutment screw diameter on the levels of stress within the implant system and surrounding alveolar bone. In 2005, Kwon et al.[20] conducted a finite element study revealing that narrow implants with standard abutment screws exhibited the highest levels of bone stress. Jeng et al.[21] found that implant neck wall thickness had a significant impact on stress values within the Morse taper design of implants and the surrounding bone, while the size of the abutment screw did not affect these outcomes. However, the type of attachment between dental implants and abutments is not limited to Morse taper design alone. For other systems, such as the external hexagon connection of dental implants, a comprehensive analysis and discussion are required to determine whether similar or different results are obtained. Additionally, understanding how abutment screw design interacts with the levels of stress within implants, abutment screws, and alveolar bone is crucial for improving clinical outcomes.[19a]

In this study, three-dimensional (3D) finite element analysis was performed to analyze the effects of abutment screw diameter on the distribution of stress within the alveolar bone surrounding the implants. We investigated how the structure behaves under force and hypothesized that in wide-diameter dental implant systems, increasing abutment screw diameter would lead to considerable differences in stress distribution across the implant, abutment screw, and adjacent alveolar bone under different occlusal forces.

Materials and methods

The external hex connection dental implant system (Biomet 3i; Palm Beach Gardens, FL, USA) was modeled using physical measurements and computer-aided design software (version 2017, SolidWorks, Concord, MA, USA). The simulation was performed focusing on the mandibular first molar region because of its suitability for wide-diameter implants and high edentulous rates. A mandibular first molar crown with dimensions of 10.5 mm buccolingually, 11 mm mesiodistally, and 7.5 mm occlusocervically and a cusp slope of 33° was integrated into the alveolar bone model (10 mm buccolingually × 16.3 mm mesiodistally × 24.2 mm occlusocervically). The alveolar bone model comprised a 2 mm thick cortical bone layer surrounding a solid cancellous bone interior. Various abutment screw sizes and implant diameters were incorporated into three different design modules (see Table 1), each simulated with four distinct qualities of cancellous bone, resulting in a total of 12 models. The dimensions of the metal coping of the prosthesis and the cement used for fixation were deemed negligible and thus excluded from this experiment. Structural analysis of the solid model was performed using Ansys software (version 19; Ansys, Canonsburg, PA, USA) considering 3D solid elements (Fig. 1). The model was meshed using SOLID187 elements with 10 nodes to establish a finite element model with a uniform mesh and well-shaped elements. Young's modulus and Poisson's ratio for each object are presented in Table 2.[19,22]

Three types of loading conditions and a boundary condition are described below and depicted in Fig. 2. A torque of 32 N·cm was applied to the abutment screws as the preload (Fig. 2a).

(1) Occlusal force under simulated chewing conditions[20]: a vertical force of 300 N was applied to the occlusal center.
(2) Occlusal force under simulated chewing conditions[20]: an oblique force of 300 N was applied to the surface

Table 1 Module conditions used in our analysis.

	Implant length (mm)	Implant diameter (mm)	Abutment platform diameter (mm)	Abutment screw diameter (mm)	Cortical bone thickness (mm)	Cancellous bone type[b]
Module 1[a]	13	4	4	2.0	2	I, II, III, IV
Module 2[a]	13	5	5	2.0	2	
Module 3[a]	13	5	5	2.5	2	

Note:

[a] The main difference between Module 1 and Module 2 is the implant diameter, which varies between the two models. The main difference between Module 2 and Module 3 is the abutment screw diameter, which varies between the two models. The differences between Module 1 and Module 3 includes variations in two design parameters — implant diameter and abutment screw diameter within the respective models.

[b] Cancellous bone is categorized into four grades based on varying strength levels, with Type I having the highest strength, followed by decreasing strength levels in sequential order, and Type IV exhibiting the lowest strength.

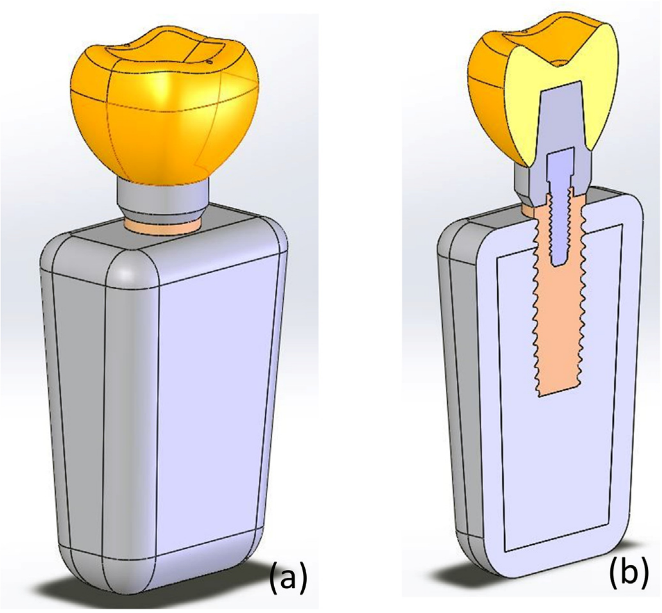

Figure 1 Three-dimensional model including cortical bone, cancellous bone, fixture, abutment, abutment screw, and crown in (a) combined view and (b) sectional view.

of the buccal cusp (30° to the vertical [long] axis of the implant).

(3) Simulated lateral forces under a nonchewing condition, possibly arising from tongue, cheek, or molar forces: a horizontal force of 100 N was applied to the buccal cusp.

In an individual with intact dentition and a healthy temporomandibular joint, the process of chewing involves the closure of the maxilla and mandible. When an implant is subjected to occlusal forces, the upper and lower teeth come into contact, maintaining the mandible in a closed position and restricting arbitrary movement of the temporomandibular joint. To reflect this, boundary conditions were established around the entire alveolar bone in our analytical model.

After tightening the abutment and securing it with abutment screws, the interfaces remained immobilized

Table 2 Material properties in the finite element model.

Component		Young's modulus (GPa)	Poisson's ratio
Crown		80	0.3
Gold alloy		90	0.3
Titanium		110	0.35
Cortical bone		13	0.3
Cancellous bone[a]	Type I	9.5	0.3
	Type II	5.5	0.3
	Type III	1.6	0.3
	Type IV	0.69	0.3

Note:

[a] Cancellous bone is categorized into four grades based on varying strength levels, with Type I having the highest strength, followed by decreasing strength levels in sequential order, and Type IV exhibiting the lowest strength.

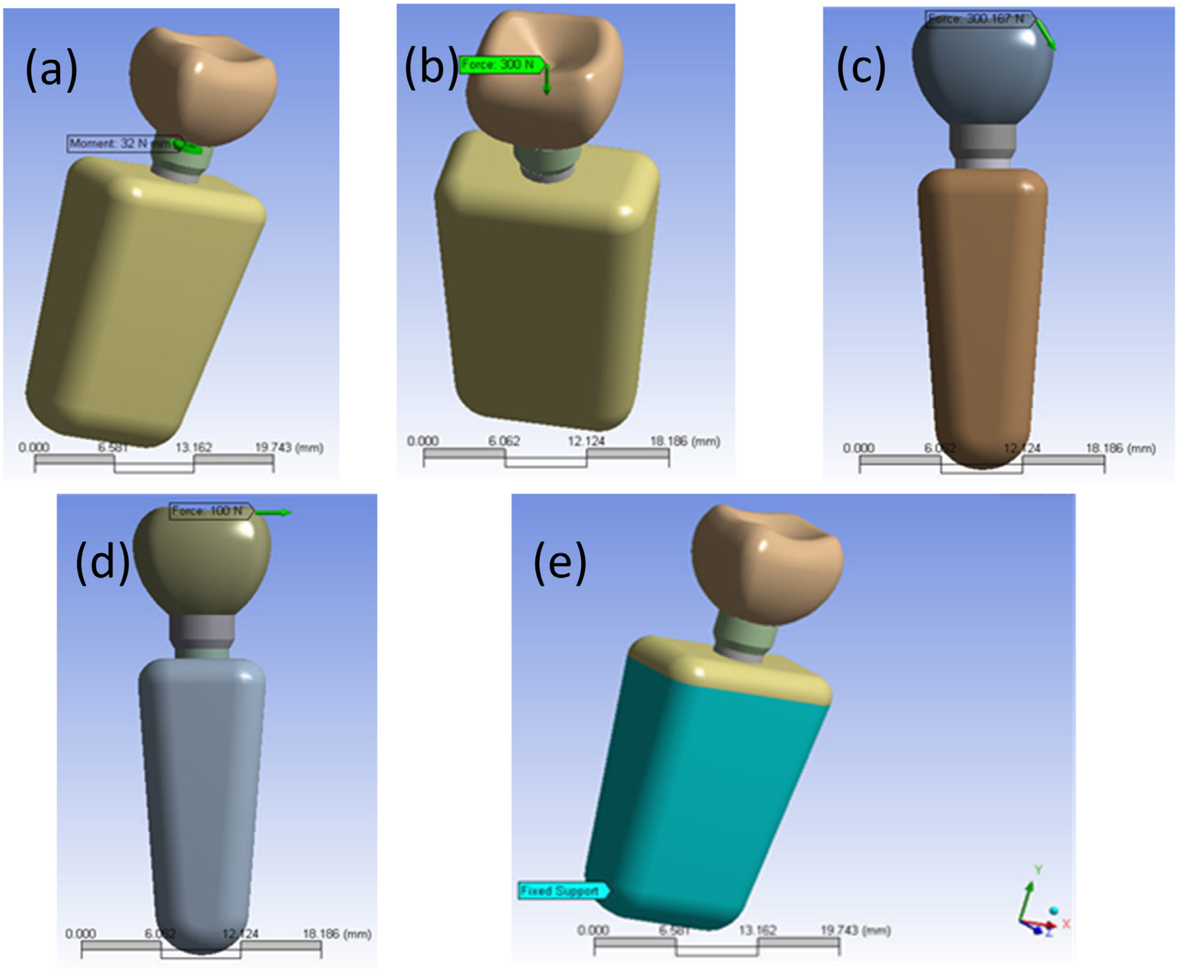

Figure 2 Setting loads and boundary conditions. (a) Preload (32 N·cm torque). (b) Vertical occlusal force. (c) Occlusal force at 30° to the vertical axis of the implant. (d) Horizontal occlusal force. (e) Boundary condition limits (blue—green part). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

under normal occlusal forces. Any movement or separation at these interfaces was considered to be either implant failure or a loose screw. Therefore, the contact surfaces of each component were bonded together to accurately simulate real-life conditions.

The initial setup for the analysis comprised 196,578 nodes and 128,443 elements. To improve the accuracy of the mesh, we reduced the element size, which resulted in an increase to 391,669 nodes and 264,551 elements. This alteration was made to test for convergence. The results demonstrated only a slight difference of 0.33 % between the two setups. In this study, the finite element analysis involved the assessment of the following parameters:

(1) Evaluation of stress distribution across implants, abutment screws, and alveolar bone.
(2) Identification of maximum stress locations and analysis of correlations between models.

Results

The results of applying a torque of 32 N·cm to the abutment screws when the three modules were in Type I bone are presented in Table 3 and Fig. 3. The equivalent von Mises stress on the implant and abutment screw was the smallest in Module 2. Additionally, the principal stresses on the cortical and cancellous bones of Modules 2 and 3 were almost equivalent for wide-diameter implants.

Table 4 presents the variation in stress ratios associated with the differences in implant and abutment screw diameters for each part, and Fig. 4 shows the stress distributions of implant and type I bone under a 300 N of vertical load. Regardless of the direction of occlusal forces, for Type III and Type IV bones, the stress levels on cancellous bone for 5-mm implants were 1.51 to 1.34 times and 1.36 to 2.62 times greater than that for 4-mm implants, respectively. However, relatively low stress levels were observed in other parts for wide-diameter implants. When vertical occlusal forces were applied, the stress on wide-diameter implants was reduced by 0.87 to 0.75 times that on standard-diameter implants. Conversely, when the occlusal force was at 30° to the vertical axis of the implant or horizontal, this reduction increased further to 0.56—0.58 times that noted for standard-diameter implants. When considering vertical occlusal forces on abutment screws, approximately 50 % (0.57—0.58 times) reduction was noted in the levels of stress on wide-diameter implants compared

Table 3 Stress distribution when a preload of 32 N·cm torque was applied to the abutment screws in Type I bone.

Module[a]	Diameter		Equivalent (von-Mises) stress (MPa)		Principal stresses (MPa)	
	Implant (mm)	Abutment screw (mm)	Implant	Abutment screw	Cortical bone	Cancellous bone
Module 1	4	2.0	6.66	3.62	1.44	0.21
Module 2	5	2.0	4.02	1.79	1.05	0.17
Module 3	5	2.5	6.82	3.69	1.04	0.16

Note:

[a] The main difference between Module 1 and Module 2 is the implant diameter, which varies between the two models. The main difference between Module 2 and Module 3 is the abutment screw diameter, which varies between the two models. The differences between Module 1 and Module 3 includes variations in two design parameters — implant diameter and abutment screw diameter within the respective models.

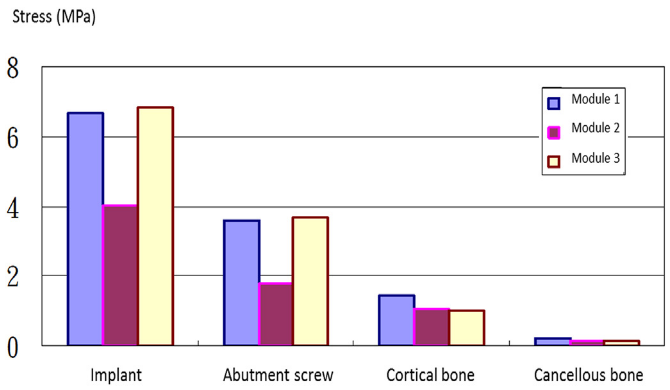

Figure 3 Equivalent von Mises stress on each module applying a torque of 32 N·cm to abutment screws in Type I bone.

with the levels of stress on standard-diameter implants; by contrast, under occlusal forces at 30° to the horizontal axis of the implant or horizontal occlusal force, this reduction increased to 0.34—0.36 times that of the stress on standard-diameter implants. For cortical bone, when vertical occlusal forces were applied, approximately 50 % (0.58—0.67 times) reduction was noted in the levels of stress on wide-diameter implants compared with the levels of stress on standard-diameter implants; by contrast, with occlusal forces at 30° to the horizontal axis of the implant or horizontal occlusal force, this reduction was smaller, 0.9—0.93 times that of the stress on standard-diameter implants. For the cancellous bone, the reduction in the levels of stress on wide-diameter implants under any occlusal force did not vary considerably; this reduction was 0.6—0.85 times that of the stress on standard-diameter implants.

The stress levels of implants of different diameters in Type I—IV cancellous bones under different occlusal forces are shown in Fig. 5. When the occlusal force passed through the vertical axis of the implant, the stress on each part was minimal. Increasing implant diameter generally resulted in the reduction of the levels of stress on the implants, abutment screws, cortical bone, and cancellous bone, except for Type III and Type IV cancellous bones. Under similar forces, stress on the implants and abutment screws remained consistent regardless of bone condition. When vertical occlusal forces were applied, stress on the cortical bone was found to be correlated with bone quality. The levels of stress in cortical bone were higher on Type III and Type IV cancellous bones than on Type I and Type II cancellous bones.

The results of comparison of different abutment screw diameters for wide-diameter implants are presented in Fig. 6. Regardless of occlusal force direction or bone condition, increasing the diameter of the abutment screw led to increases the stresses of the abutment screw and implant while the stress level within the cortical bone remained unchanged. The effect on the level of stress within cancellous bone was negligible.

Discussion

The risk of abutment screw loosening is influenced by several factors, including screw type, screw size, screw material,[23,24] screw length, implant diameter, abutment screw retightening, abutment screw head shape, lateral screws, repeated opening and closing of abutment screws, new screws, torque values, and prosthetic screw types.[4] The coating of abutment screws to reduce friction and allow for increased rotational angles and preload values has also been explored.[25] This approach involves ensuring the abutment screws do not fully engage with the implant.[25] However, few studies have examined the effects of abutment screw diameter on the levels of stress within dental implants and adjacent bones. The present study highlights that increased diameters of abutment screws correlate with increased levels of implant stress, which may negatively affect the implant performance.

In a biological context, forces can vary in magnitude, direction, and location. Mericske-Stern et al. conducted a study to assess occlusal forces in implant-supported prostheses and found that these forces ranged from 210 to 400 N in the second premolar region and from 130 to 395 N in the molar region.[26] In the present study, a vertical force of 300 N was applied at the central point of the occlusal surface. Additionally, a force of 100 N was applied horizontally and at 30° to the buccal slope.

The present study found that wide-diameter implants have better stress distribution outcomes under occlusal forces than do standard-diameter implants. The maximum level of stress within the alveolar bone was found to be consistent with findings from another study.[22] However, increasing the diameters of the abutment screw and the implant resulted in elevated levels of stress for both components. This occurred because the implant wall surrounding the abutment screw became thinner. This effect was particularly noticeable when occlusal forces were not aligned with the implant's axis or when the quality of the alveolar bone was poor.

The stress patterns for off-axis occlusal forces were similar to those reported by Geramy and Morgano,[10] who compared various implant designs for single molars. This finding highlighted the importance of positioning the occlusal force on the central axis of the implant to minimize stress on that component. However, when noncentral occlusal forces were present, relatively low levels of stress were experienced by cancellous bone. Because the influence of abutment screw diameter on the distribution of stress within the implant system remains to be explored, direct comparisons cannot be made. Increasing the diameter of the abutment screw may also increase its volume, potentially providing sufficient strength to withstand noncentral occlusal forces.

This study has several limitations because of certain assumptions that were made. These included assuming that the alveolar bone was isotropic, homogeneous, and elastic; assuming a fully osseointegrated interface between the implant and the alveolar bone; assuming a uniform thickness of the cortical bone; and not accounting for variations in biomaterial properties as would be encountered in clinical settings. Additionally, the complexities of chewing patterns, occlusal forces, and the alveolar bone's responses to force were simplified and did not fully reflect real-world scenarios.[19] Furthermore, we simulated only static loads on the surface rather than alternating stresses experienced by structures under continuous compressive and tensile forces. To validate the clinical relevance of this research, future investigations should explore dynamic fatigue failure of materials.

Wide-diameter implants have lower stress distribution to that of standard-diameter implants. However, the prospect of increasing the abutment screw diameter does not yield favorable benefits. This modification could lead to heightened stresses on both the abutment screw and the implant. As a result, it is advisable to employ standard abutment screw sizes when dealing with wide-diameter implants in instances of robust bone quality. This strategy enhances a

Table 4 Variations in stress ratios for different diameters of dental implants and abutment screws.

Type of cancellous bone[a]	Ratio results[b]	A vertical occlusal force of 300 N				An occlusal force of 300 N at 30° of the implant axis				A horizontal occlusal force of 100 N			
		Equivalent (von-Mises) stress (MPa)		Principal stresses (MPa)		Equivalent (von-Mises) stress (MPa)		Principal stresses (MPa)		Equivalent (von-Mises) stress (MPa)		Principal stresses (MPa)	
		Implant	Abutment screw	Cortical bone	Cancellous bone	Implant	Abutment screw	Cortical bone	Cancellous bone	Implant	Abutment screw	Cortical bone	Cancellous bone
Type I	A	0.57	0.34	0.91	0.60	0.57	0.34	0.91	0.60	0.58	0.36	0.92	0.72
	B	1.15	2.21	1.00	1.00	1.15	2.21	1.00	1.00	1.09	2.07	1.00	1.00
Type II	A	0.56	0.34	0.91	0.80	0.56	0.34	0.91	0.80	0.58	0.35	0.93	0.77
	B	1.17	2.18	1.00	1.00	1.17	2.18	1.00	1.00	1.12	2.13	1.00	0.98
Type III	A	0.56	0.34	0.90	1.30	0.56	0.34	0.90	1.30	0.58	0.35	0.93	1.34
	B	1.23	2.16	1.00	1.00	1.23	2.16	1.00	1.00	1.17	2.16	1.00	0.86
Type IV	A	0.56	0.34	0.90	2.20	0.56	0.34	0.90	2.20	0.58	0.34	0.93	2.62
	B	1.27	2.17	1.00	1.00	1.27	2.17	1.00	1.00	1.21	2.17	1.00	0.88

Note:

[a] Cancellous bone is categorized into four grades based on varying strength levels, with Type I having the highest strength, followed by decreasing strength levels in sequential order, and Type IV exhibiting the lowest strength.

[b] A represents the ratio of the maximum outcomes achieved with a 5 mm diameter implant to those achieved with a 4 mm diameter implant, both utilizing the same 2 mm diameter abutment screw. B indicates the ratio of the maximum outcomes achieved with a 2.5 mm diameter abutment screw to those achieved with a 2.5 mm diameter abutment screw, both utilizing the same 5 mm diameter implant.

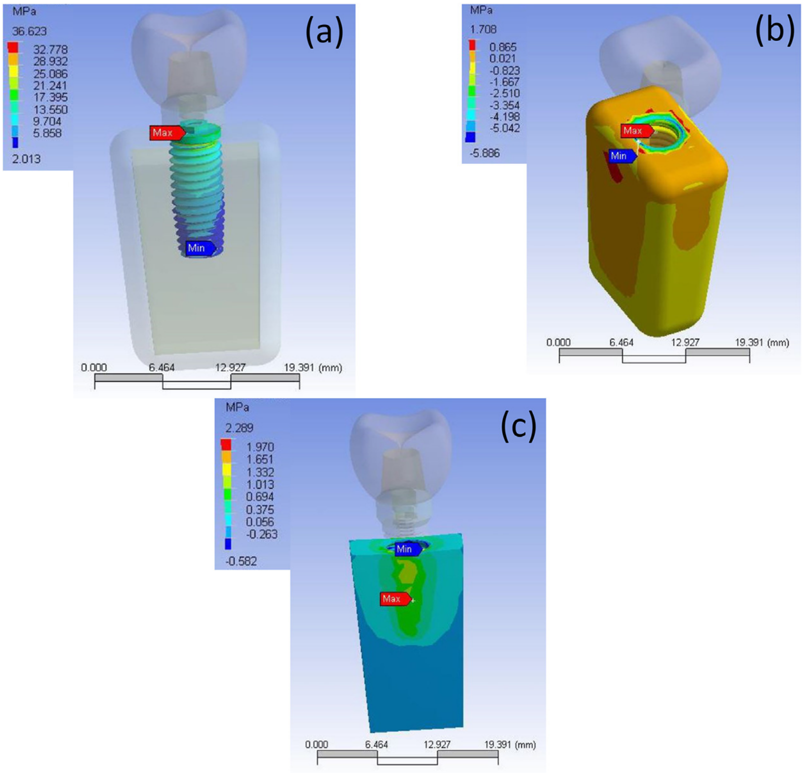

Figure 4 Stress distributions of (a) implant, (b) cortical bone, and (c) cancellous bone under vertical occlusal force of 300 N in Type I bone condition.

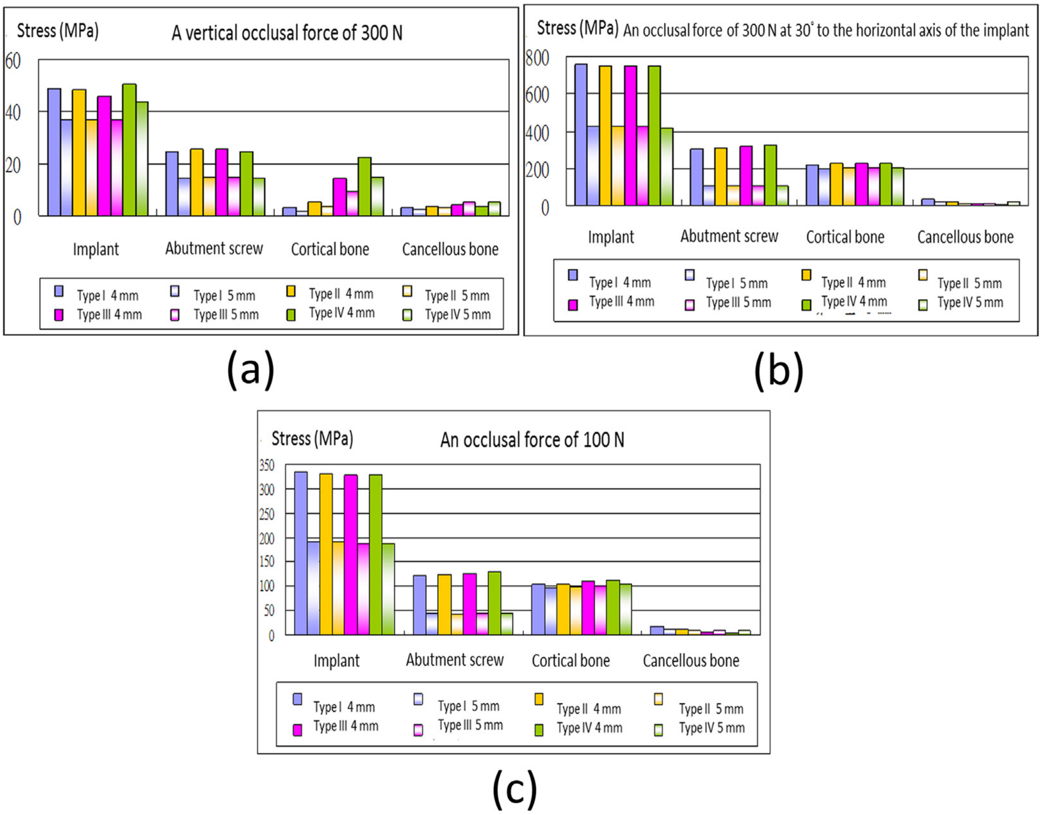

Figure 5 Stress values of implants of different diameters under different occlusal forces in Type I—IV cancellous bones. (a) Vertical occlusal force of 300 N. (b) Occlusal force of 300 N at 30° to the horizontal axis of the implant. (c) Occlusal force of 100 N.

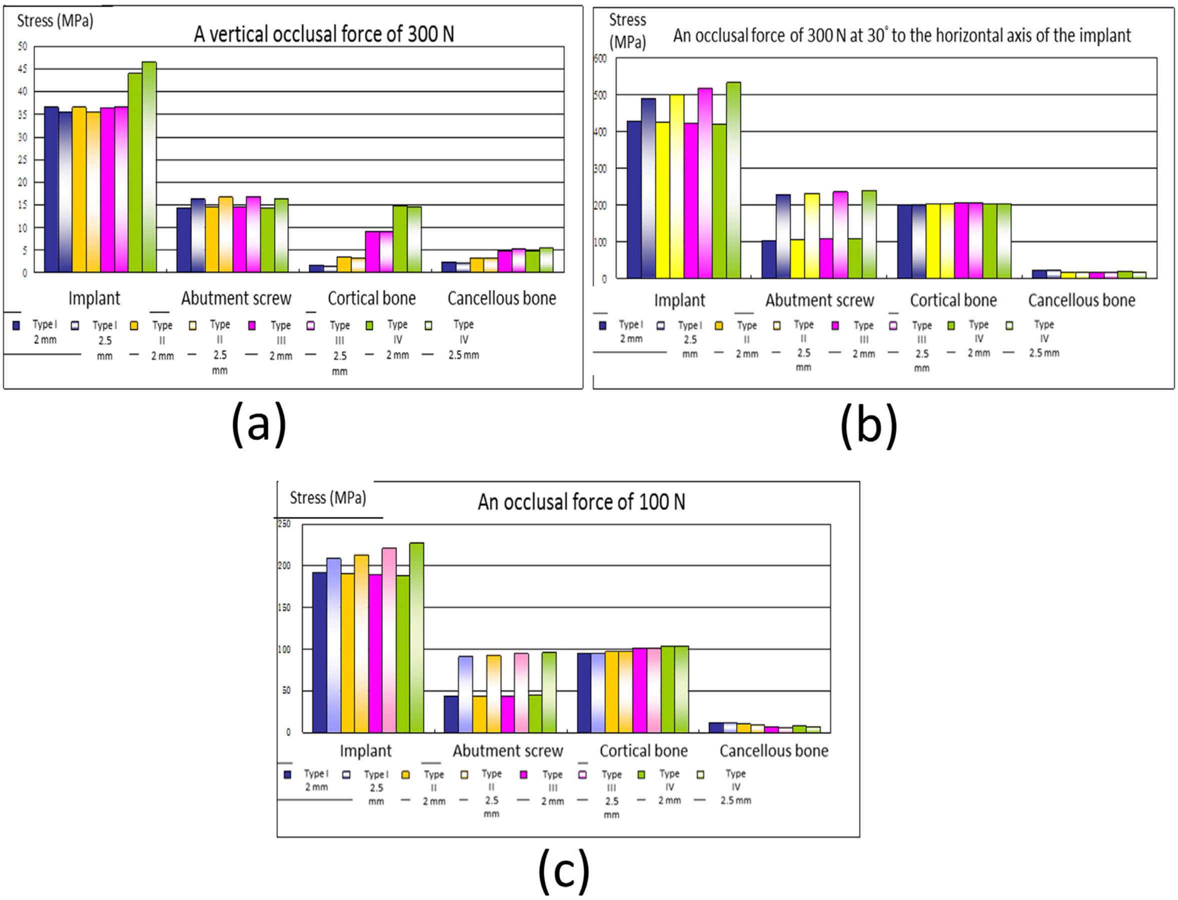

Figure 6 Stress levels within abutment screws of different diameters in a wide-diameter implant system in Type I—IV cancellous bones under different occlusal forces. (a) Vertical occlusal force of 300 N. (b) Occlusal force of 300 N at 30° to the horizontal axis of the implant. (c) Occlusal force of 100 N.

more favorable distribution of stress throughout the implant system.

Declaration of competing interest

The authors have no conflicts of interest relevant to this article.

Acknowledgments

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