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Original Article

Using optimization approach to design dental implant in three types of bone quality - A finite element analysis

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Abstract *Background/Purpose:* The use of finite element (FE) analysis in implant biomechanics offers many advantages over other approaches in simulating the complexity of clinical situations. The aim of this study was to perform an optimization analysis of dental implants with different thread designs in three types of bone quality.
Materials and methods: The three-dimensional FE model of a mandibular bone block with a screw-shaped dental implant and superstructure was simulated. In the optimization analysis, the design variables included the thread pitch and the thread depth of the implant. The objective was to minimize the displacement of the implant to the target value. Three FE models with different bone qualities (D2: better bone quality; D3: ordinary bone quality; D4: poor bone quality) were created.
Results: The FE results showed that the displacement of the implant and the stress of the cortical bone increased, while the Young's modulus of the cancellous bone decreased. In the D2 bone, changing the thread pitch and thread depth had little effect on cortical stress and implant displacement. However, in D3 and D4 bone, increasing thread depth reduced cortical stress by 40 % and implant displacement by at least 9 %.
Conclusion: Adjusted thread depth for D3 and D4 bone would reduce crestal bone stress and increase implant stability, but only a little alteration on crestal bone stress and implant stability for D2 bone.

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Introduction

Dental implants have gained popularity in dentistry as a common and successful procedure for restoring missing teeth or edentulism. Most published long-term implant studies have shown survival rates of over 95 %.[1—3] However, these survival rates tend to vary in different areas of the maxilla and mandible[4] and are highly dependent on bone quality, which allows for better initial stability and osseointegration in future restoration.[5] Bone quality is generally defined as the sum of all characteristics of bone that influence its resistance to fracture.[6] Poor bone quantity and quality have been recommended as the main risk factors for implant failure, as they may be closely related to excessive bone resorption and impaired healing in contrast to higher bone density.[7] In 1985, Lekholm and Zarb listed four bone qualities found in the anterior regions of the jaw and classified them based on the amount of cortical bone versus trabecular bone visible on pantograph film.[8] Subsequently, the term "bone quality" was introduced to refer to the different bone density types (D1 ∼D4).[9]

Type 1 consisted of homogeneous compact bone (D1). Type 2 consisted of a thick layer of compact bone surrounding a core of dense trabecular bone (D2). Type 3 consisted of a thin layer of cortical bone surrounding dense trabecular bone of favorable strength (D3). Type 4 consisted of a thin layer of cortical bone surrounding a core of trabecular bone of low density (D4). Engquist et al. reported that 78 % of all failures were in type 4 bone[10] and Weng et al. reported a failure rate of 20 % in the posterior maxilla.[11] Elements such as loading degree, surgical technique, implant surface design, and material biocompatibility have also been shown to influence the implant osseointegration and long term success. Bone loss around the implant neck and continued bone resorption activated by excessive implant loading can cause high stress concentrations at the implant—bone interface.[12] Markovic et al. recommended using the bone-condensing technique to increase implant stability in posterior maxilla for different bone qualities.[13] Turkyilmaz et al. reported that osteotomy using the last drill with a diameter slightly thinner than the installed implant in the maxillary posterior region with relatively low bone density.[14] It may be a viable option regarding alternative surgical protocols to increase primary implant stability.

There are two main categories of implant design: the macro design and the micro design. Microdesign consists of implant materials, surface morphology and coating.[15] When an implant is under optimal functional load, the surrounding bone is remodeled to form woven bone. However, under intense adverse loading, microfractures occur in the alveolar bone, inducing "osteoclastogenesis".[16] Bone loss is exacerbated because bone formation is not fast enough to fill the defect, ultimately leading to implant failure.[17] Some biomechanical researches were conducted and the findings indicated that reducing the thread pitch (P) could improve initial anchorage and primary stability in cancellous bone.[18,19] Chung et al. also found that implants with a pitch of 0.6 mm exhibited greater crestal bone resorption than those with a pitch of 0.5 mm in a beagle dog model.[20] Ma et al. employed three-dimensional finite element (FE) analysis to demonstrate that a pitch of 0.8 mm exhibited greater resistance to vertical load than those with a pitch of 1.6 mm and 2.4 mm.[21]

Thread depth (D) is defined as the distance between the major and minor diameter of the thread. In other words, it can also be calculated by subtracting the distance from the outermost tip to the innermost body of the thread.[22—24] Thread width is defined as the distance between the superior-most and inferior-most tip of a single thread, measured in the same axial plane.[25] The shallower the thread depth, the easier the implant insertion may be able to eliminate the need for tapping, particularly in the high-density bone. Conversely, the creation of deep threads increases the functional surface area at the bone-implant interface, which can enhance primary stability in low-density bone or regions with high occlusal force. Ao et al. demonstrated that thread depth exerts a more pronounced influence on stress distribution than width. Implants with a depth of greater than 0.44 mm and a width of 0.19—0.23 mm exhibited the most favorable biomechanical behavior for immediate loading.[26]

The application of FE analysis in implant biomechanics offers a number of advantages over other approaches in simulating the complexity of clinical situations. Currently, available implants with different thread designs provide a high level of primary stability to the prosthetic connection, which allows for immediate loading.[27] Various attempts have been made to optimize the dental implant designs, resulting in a range of potential optimization opportunities for the components of the implant.[28] Based on FE analysis, topology optimization may be employed to identify redundant material distribution on a dental threaded implant and to redesign a new implant macro-geometry while evaluating biomechanical capacity. The pitch, depth and width of the implant thread of a dental implant with a buttress thread profile were found to be optimal for reducing stress at the bone-implant interface using a Taguchi approach.[29] Prior to this study, there had been no report of evaluating the influence of both thread pitch and depth on stress distribution and micro-motion in different types of bone quality. Consequently, the specific objective of the present study was to perform a biomechanical analysis of dental implants with different thread designs around different types of bones by means of design optimization of three-dimensional FE analysis.

Materials and methods

FE model of the bone-implant system

A three-dimensional model of a mandibular bone block with a screw-shaped dental implant and superstructure was constructed using FE software (ANSYS Workbench 14.0, Swanson Analysis System Inc., Huston, PA, USA). The solid bone block was approximately 14 mm in width and 13 mm in height, with a thickness of 1 mm. It contained cortical and cancellous bone tissue. The mesial and distal section planes were not covered by cortical bone. The geometry of the 10-mm OSSEOTITE® Certain® implant (3i Implant Innovations, Inc., Palm Beach Gardens, FL, USA) was employed as a reference to model a V-shaped (0.5 mm pitch, 0.4 mm depth) threaded implant. A 3.4 mm diameter and a 5-mm-high abutment connection, assuming a platform switching configuration, was simulated. The implant and abutment were simplified to one unit and a full zirconia crown with 2-mm occlusal thickness over the abutment (Fig. 1). All materials used in the models were considered to be isotropic, homogeneous and linearly elastic. Since the bone type D1 seldom presented problems regarding the stability of implantation, three different bone types (D2, D3, D4) with their different material properties[30–33] were analyzed (Table 1).

The convergence test was set at a change of displacement variation of less than 1 % for the model with different element sizes. The element size was controlled to approximately 0.5 mm, and the FE model was constructed with 82,334 elements and 141,498 nodes as an analytic model (Fig. 2).

Table 1 Mechanical properties of materials used in the FE models.

Material	Elastic modulus (MPa)	Poisson's ratio	References
Zirconia	220,000	0.3	Kohal et al.[33]
Titanium	110,000	0.35	Benzing et al.[32]
Cortical bone	13,700	0.3	Barbier et al.[30]
Cancellous bone			
Type 2 (D2)	5500	0.3	Almeida et al.[31]
Type 3 (D3)	1600	0.3	Almeida et al.[31]
Type 4 (D4)	690	0.3	Almeida et al.[31]

Boundary and loading conditions

In boundary condition, constraints should be placed on nodes located far from the region of interest to prevent the overlap of stress or strain fields associated with reaction forces at the bone-implant interface. Therefore, zero-displacement constraints must be applied to some boundaries of the model to ensure an equilibrium solution. In our FE analysis, the area of interest consisted of crestal bone stress and implant stability. Considering the real clinical situation and mechanical equilibrium, the models were constrained in all directions at the nodes of the three faces (mesial, distal, and bottom faces). These three faces fit the criteria of non-interest areas and represent the actual immobilized region.

In loading condition, bite force can be defined as a combination of compressive and shear forces. For realistic simulations, combined oblique loads are generally used. Additionally, clenching intensity was assessed by masseteric EMG activity, with general intensity levels including 30 % and 60 % of the maximum voluntary contraction. Therefore, in our study, the loading condition was considered under physiological status (50 % of the maximum voluntary contraction). As a result, the loading was simulated by applying an oblique load (vertical load of 100 N and horizontal load of 20 N) to the middle point at the center of the crown (Fig. 3).

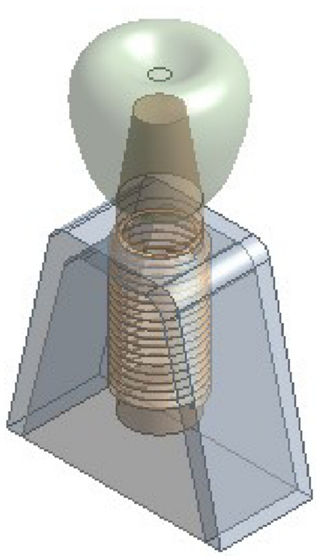

Figure 1 FE model of the bone-implant system.

Model validation

Lin et al. conducted an in vitro study to replicate the ABS (P400 ABS, styrene terpolymer, Stratasys Inc., Minneapolis, MN, USA) plastic bone model as an experimental sample to validate the corresponding FE simulation results.[34] The microstrain measured from the validated experiment (mean $\pm$ standard deviation) was 75.2 $\pm$ 9.6 $\mu\varepsilon$ in the bucal–lingual direction. The calculated microstrain of the FE model was 69.58 $\mu\varepsilon$ under the same loading condition designed by Lin et al. The corresponding error between the experimentally measured and numerically calculated microstrain was 7.5 %, indicating a reasonable model validation.

Optimization analysis of screw designs around different types of bones

A static structural analysis was conducted on the von Mises stress and displacement within the implant and surrounding bone. Two types of FE modes were compared, each with three types of bone quality, including the original model and an optimized model. In the optimization analysis, two initial design variables were considered: thread pitch ($P = 0.5$ mm) and thread depth ($D = 0.4$ mm) (Fig. 4). The range of the thread pitch was 0.25 mm $< P <$ 1.0 mm, while the depth was 0.2 mm $< D <$ 0.8 mm. In the constraint function, the restriction was that the von Mises stress surrounding the cortical bone in the optimization model was lower than that in the original model.

To achieve greater implant stability, the objective function was set to the minimum displacement. In the optimization analysis, the subproblem approximation method was employed, with a total of 30 iterations and a convergence tolerance of 0.01. The subproblem approximation analysis was initially performed to locate an approximate optimum in

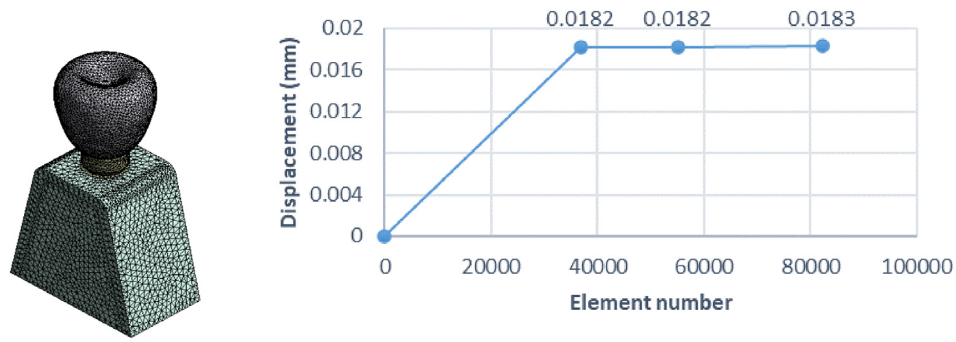

Figure 2 Convergence test for the FE model with different element sizes from 0.3 to 0.5 mm.

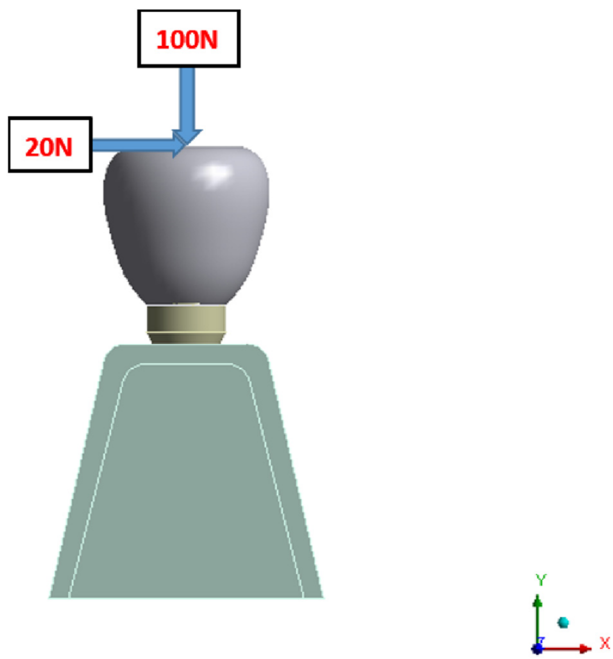

Figure 3 An oblique load (vertical load of 100 N and horizontal load of 20 N) to the middle point at the center of the crown.

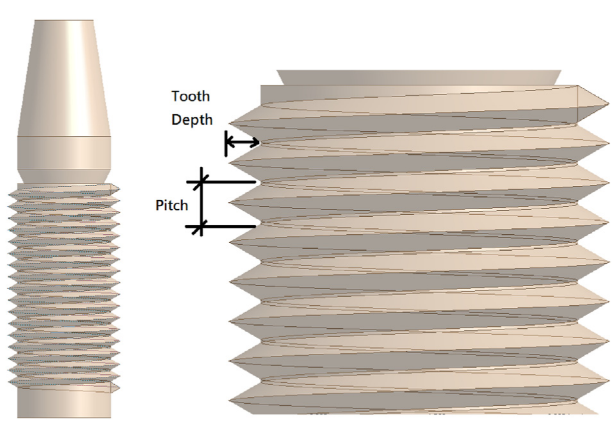

Figure 4 Design parameters of thread pitch and thread depth.

the feasible design space, and then the first order method was used to perform the final search.

Results

Thread of optmization model

The optimal implant thread pitch and depth among the three bone types are presented in Table 2 and Fig. 5. The D2 bone exhibited a pitch of 0.39 mm, which was smaller than the original 0.5 mm, and a depth of D2 = 0.387 mm, which was almost equal to the original 0.4 mm. The pitch of the other bone types (D3 = 0.484 mm, D4 = 0.506 mm) was found to be slightly closer to the original model. However, the depth of D3 and D4 (0.763 mm, 0.791 mm) was observed to have increased by 91 % and 98 %, respectively.

Biomechanical performance of the original model

The displacement of the implant among the three bone types (D2 to D4) is presented in Table 3. The D2 bone exhibited the lowest displacement, with an average value of approximately 14 μm. The D3 and D4 model demonstrated higher displacement values than D2 bone, with increases of 35.7 % and 64.3 %, respectively. In D4 cortical bone surrounding the implant platform, the highest von Mises stress was observed at 21.14 MPa.

Biomechanical performance of the optimized model

In the optimization design, the displacement of D3 and D4 bone was observed to decrease by 9 % and 14 %, respectively, in comparison to the original model (Table 3). The

Table 2 Optimization results of implant thread pitch and thread depth when compared to the original thread pitch (P = 0.5 mm) and thread depth (D = 0.4 mm).

	Pitch (mm)	Depth (mm)
D2	0.390 (−22 %)	0.387 (−3%)
D3	0.484 (−3%)	0.763 (91 %)
D4	0.506 (1 %)	0.791 (98 %)

Note: percentages mean the three groups were normalized to the control group (P = 0.5 mm and D = 0.4 mm).

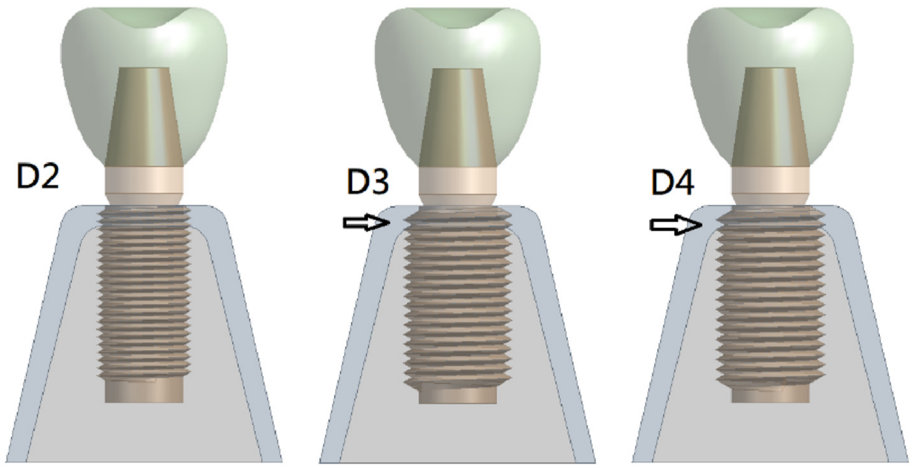

Figure 5 Change of pitch and thread depth in optimized models under same diameter of implant. Note: Arrows indicated the increase of thread depth in D3 and D4 model.

Table 3 Displacement of implant and von Mises stress of crestal bone in the original and optimization models.

	Original	Optimization	Decrease
Displacement (mm)			
D2	0.0127	0.0124	2 %
D3	0.017	0.0155	9 %
D4	0.0215	0.0186	14 %
von Mises stress (MPa)			
D2	6.67	6.67	0 %
D3	12.68	7.61	40 %
D4	21.14	12.15	43 %

Note: The percentage was defined as follows ((original − optimization)/(original))*100 %.

top of implant exhibited more instability in D3 and D4, but the displacements were reduced after optimizing design as shown in Fig. 6.

Additionally, the apparent decline in the von Mises stress value of D3 and D4 cortical bone (Fig. 7) was observed to be 40 % and 43 % lower, respectively. The maximum von Mises stress in the entire FE model was concentrated in different locations among the three bone types (Fig. 8). In D2 bone, a higher stress of 45.3 MPa was observed in the in situ region, which is the area between the implant abutment and the crown margin. The cortical bone surrounding the platform of the implant exhibited a higher stress value of 47.8 MPa in D3 bone, while a higher stress value of 58.4 MPa was observed in adjacent to the first thread of the implant near the junction of D4 cortical and cancellous bone.

Discussion

Alveolar bone is the most active bone in the body and is particularly sensitive to stress.[35] In order to achieve stable osseointegration for implant restoration, it is important to avoid the generation of high stress concentrations or distributions in bone, as these can induce severe resorption in the surrounding bone, leading to gradual loosening and, ultimately, complete loss of the implant.[36] The results of the present study indicate that displacement increased from 14 to 23 μm and maximum von Mises stress in cortical bone increased from 6.67 to 21.4 MPa, accompanied by a reduction in bone density from D2 to D4 bone. The bone types investigated in this study demonstrated the capacity to amplify the influence of thread pattern on displacement of the implant and stress of cortical bone, in accordance with the findings of previous studies.[37–39] Consequently, the objective function was set at the minimal displacement of the implant.

The thread pitch was found to determine the surface area available for load transfer to the peri-implant tissue, with a more critical role in enhancing primary stability in low-density bone than in high-density bone.[40] Chun et al. reported that maximum effective stress reduced with pitch decreased, however, when the pitch decreased to a certain value, the maximum effective stress at the bone-implant interface would not change significantly.[22] This indicates that there was an appropriate pitch value. Nevertheless, the optimal value was affected by the shape of the thread and the bone density.[41–43] A variety of bone qualities in patients made it difficult to adapt the thread configuration completely. Therefore, it might be considered that a personalized thread pitch design could improve the stress distribution at the bone-implant interface for patients with inadequate bone conditions. Thus, the constraint function was that it could lower the von Mises stress value surrounding the cortical bone. In a clinical context, the thread depth affects the ease of implant installation and surface area. A shallower thread depth facilitates implant insertion, particularly in high-density bone. Conversely, a deeper thread depth increases the functional surface area at the bone-implant interface, potentially enhancing primary stability in low-density bone. The majority of previous studies have indicated that a thread pitch of 0.8 mm and a depth of 0.34–0.5 mm is optimal for the biomechanical properties of D2 bone. Following the optimization of the implant thread profile, a reduction of 22 % in the thread pitch (0.39 mm) and 3 % in the thread depth (0.387 mm) of D2 bone was observed to have no significant impact on displacement and stress values, due to the higher elastic

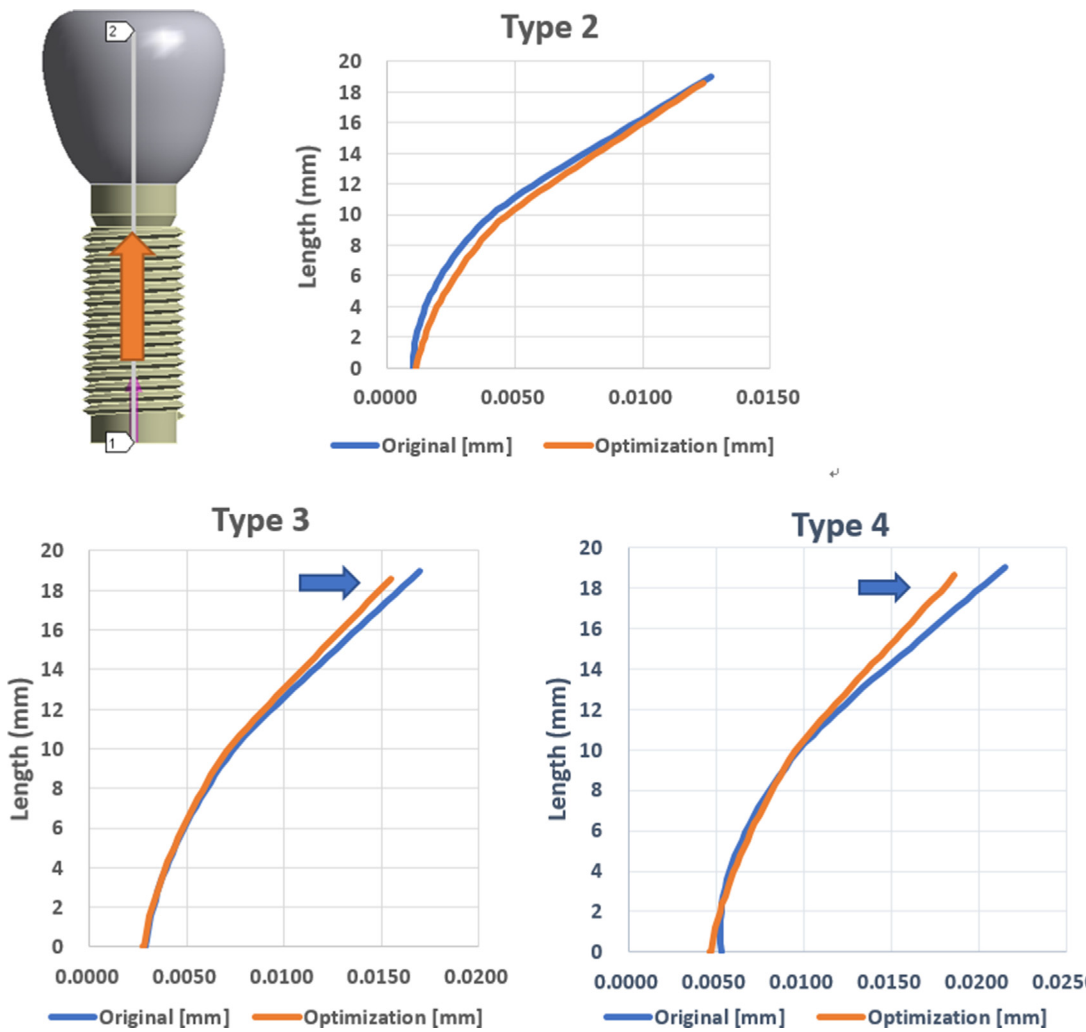

Figure 6 Comparison of displacement from bottom of implant to top of crown in the three groups. Note: blue arrows indicated a greater difference.

modulus. Nevertheless, a slight adjustment to the thread pitch and a doubling of the thread depth, up to 0.76 and 0.79 mm, respectively, was observed in the D3 and D4 bones. This was intended to enhance stability (9—13 %) and reduce the von Mises stress of the cortical bone surrounding the implant, down to 40—43 %.

In a recent investigation into dental implant,[44] the objective function of optimal results according to bone quality; but the study was not involved to the implant thread design. Shen et al. examined how the diameter of abutment screws affects the stress on dental implants and alveolar bones when subjected to occlusal forces.[45] Their study found that wider implants resulted in reduced stress levels in both the implants and the bone compared to standard-diameter implants. Jin et al. conducted a study to explore the impact of microthreads on stress distribution in peri-implant bone at varying bone levels using FE analysis.[46] Their findings indicated that the microthread design in the implant neck can help reduce marginal bone loss by minimizing shear stress in the peri-implant bone. Sakar et al. investigated how engaging (hexagonal) and non-engaging (non-hexagonal) abutments affect stress distribution and loading in the implant neck, implant abutment, and surrounding bone across various six-unit fixed prostheses.[47] Their findings indicate that different combinations of abutments, whether engaging or non-engaging, exert a comparable influence on stress distribution within the implant system. From these studies, the stress distribution surrounding crest bone and implant system were related to design of implant and screw. While numerous studies have demonstrated the biomechanical effects of dental implants, no research was focused on the biomechanical analysis of implant thread pitch and depth simultaneously across three distinct bone types.

In implant macro-design, implant prosthetic connections have been broadly categorized into external and internal hex design. The connection between the implant and abutment involved stress distribution to implant members and crestal bone near the implant.[48] These researches demonstrated that Morse taper internal connection inflicted least amount of marginal bone loss compared with external and normal internal hex design.[49,50] Morse taper internal connections resulted in better biomechanical performance owing to lower stress value of the cortical bone around implant neck portion and improved stress transfer along the entire contact surface efficiently.[51] Another

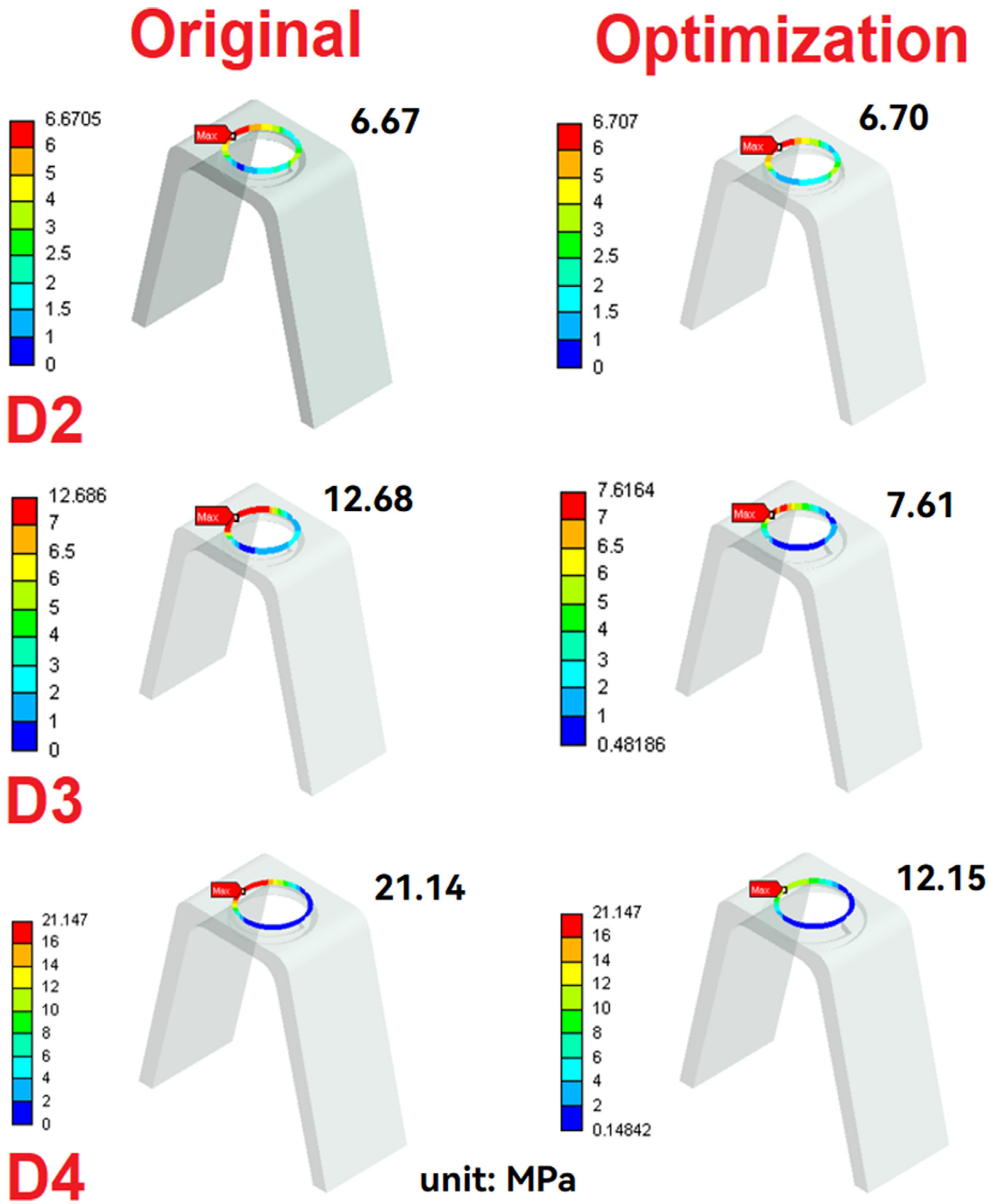

Figure 7 Comparison of von Mises stress of cortical bone surrounding the platform of the implant between original and optimization model.

important design of implant-abutment connection are platform switching concept and the following bone-level implants. The crestal bone preservation included the alteration of the location of the implant–abutment junction toward the central axis due to the smaller abutment diameter and reduced the stress concentration area from cortical bone shifting to cancellous bone under oblique load.[52] This enabled cancellous bone with more compensative ability and elasticity to withstand more occlusal load and reduce the amount of crestal bone resorption to around 0.22 mm.[53,54] Additionally, Antonelli et al. reported that implant micro-design would affect primary stability in low-density bone.[55] Especially in the magnetodynamic preparation technique, it could get better primary implant

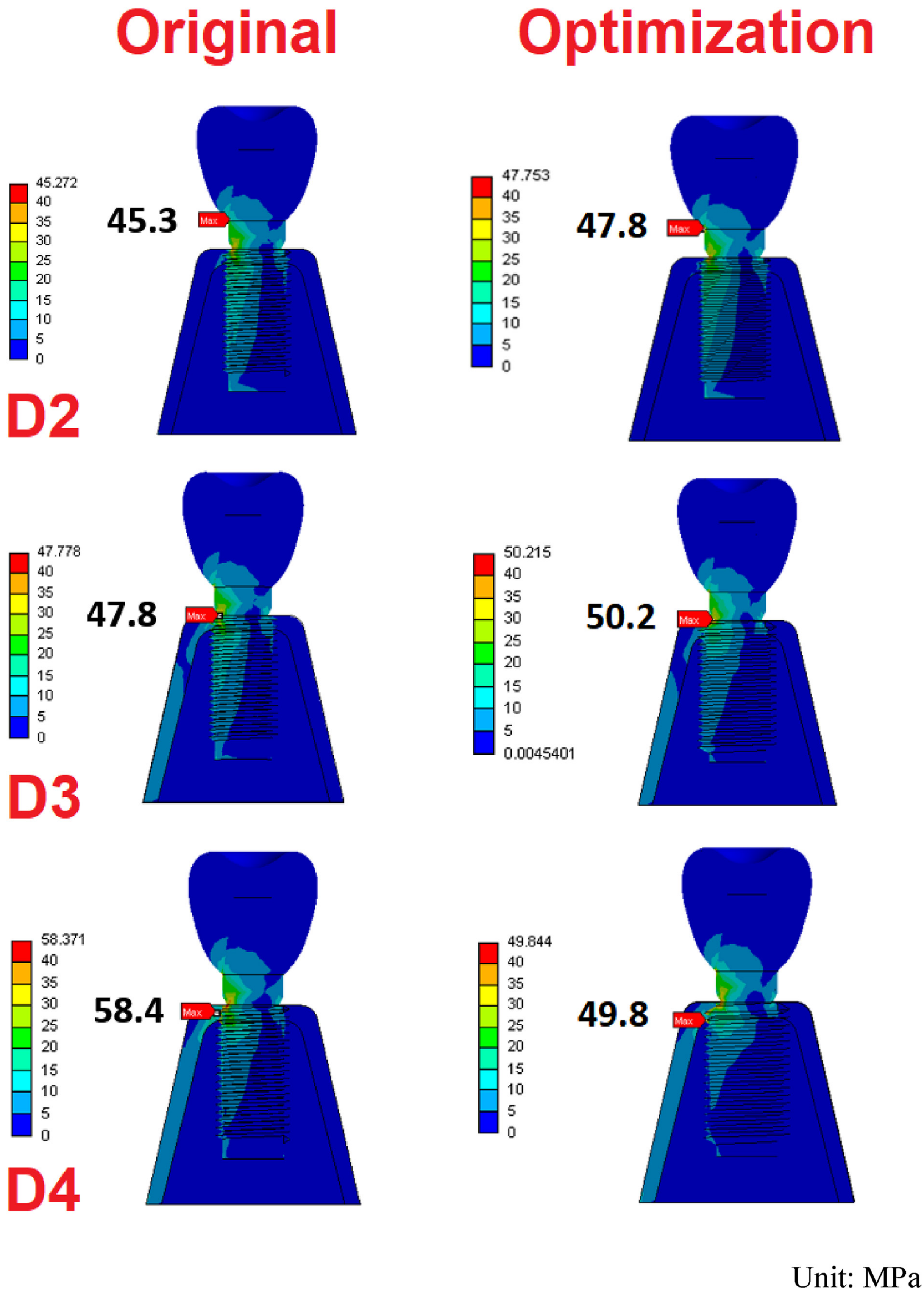

Figure 8 Comparison of the maximum von Mises stress in the FE models among three bone types.

stability in the vitro study. They attributed a better primary implant stability to the relationship between the bone walls and the implant surface. As a result, this was why we wanted to change implant surface by means of optimization design.

In implant micro-design, there are four types of implants with thread helix available commercially, including single, double, triple, and asymmetric thread. The double- and triple-thread implants are primarily indicated in type IV cancellous bone and assumed to install implant faster into the osteotomy site.[56] Asymmetric thread type implants with micro-threads in the coronal portion have been claimed to increase the bone implant contact, preserve peri-implant soft tissue, and improve stress distribution in cancellous bone could reduce cortical bone overload and loss.[57]

It should be noted that the results of this study may be limited by the assumptions made regarding the properties of the materials and tissues represented in the FE models. (1) The different thicknesses of real cortical bone and densities of real cancellous bone are neither homogenous nor isotropic. This assumption could result in some deviations in numerical calculations. However, to eliminate these deviations, the study performed normalization between the control and experimental groups. Therefore, the results showed a similar trend when compared. Additionally, in our previous study,[58] we conducted an FE simulation to compare the differences between isotropic and orthotropic materials and reported a similar trend after normalizing the results for these two different materials. (2) The simplification of the interface between the jawbone and implant, the cylindrical shape of the implant, the level of osseointegration, and the magnitude and direction of applied static forces could affect the FE results. Specifically, a cylindrical implant shape would generate better stability and reduce stress at the crestal bone while compared to conical implant shape. A real implant exhibits a conical shape due to considerations of self-drilling. Because of this, the actual crestal bone stress in a conical shape should be higher than that in a cylindrical shape. These assumptions should be reconsidered in future studies. (3) The study didn't consider different implant surface morphologies and materials. In fact, there are a few difference in FE calculation under while having a little change in terms of the two factors. For example, the material instead of stainless steel from titanium, the implant stress would follow increase because elastic modulus of implant increase. (4) Although the optimized thread pitch and depth values obtained within these limitations provided for choices in different bone densities, it should be noted that the results of FE analysis may not show the same outcome in a clinical environment due to the inability to completely replicate the human clinical situation. Basically, the current FE results aimed to pave the way to design implant with different thread shapes for different bone qualities.

Nevertheless, these studies represent a relatively simple, fast and inexpensive approach to predict the biomechanical behavior of implants in an oral cavity compared to human studies. The FE model was also experienced by model validation to confirm its reality and convergence test to confirm FE accuracy. Additionally, the study made two suggestions for clinical application: (1) the shape of the implant should be specially manufactured with different thread pitches and depths based on the patient's bone quality. This will reduce crestal bone stress and minimize bone resorption as much as possible; (2) In D3 bone, the design of the implant can effectively improve implant stability. However, in D4 bone, improving implant stability through design revisions is challenging. In such cases, the addition of bone grafts or other strengthening materials is needed to enhance implant stability. Anyway, the FE results can be an initial guide for further clinical research.

In conclusion, the displacement of the implant and the stress of the cortical bone increased, whereas the Young's modulus of the cancellous bone decreased. Following alterations to the material properties of cancellous bone, the locations of the maximum stress in the implantation model were changed. In the D2 bone, alterations to thread pitch and thread depth resulted in only a minor change in stress of cortical bone and displacement of implant. However, in the D3 and D4 bone, an increase in thread depth was found to reduce stress of cortical bone by 40 % and displacement of implant by 9 % at least.

Declaration of competing interest

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

Acknowledgments

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