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ORIGINAL ARTICLE

Biomechanical evaluation of one-piece and two-piece small-diameter dental implants: *In-vitro* experimental and three-dimensional finite element analyses

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Background/Purpose: Small-diameter dental implants are associated with a higher risk of implant failure. This study used both three-dimensional finite-element (FE) simulations and *in-vitro* experimental tests to analyze the stresses and strains in both the implant and the surrounding bone when using one-piece (NobelDirect) and two-piece (NobelReplace) small-diameter implants, with the aim of understanding the underlying biomechanical mechanisms.
Methods: Six experimental artificial jawbone models and two FE models were prepared for one-piece and two-piece 3.5-mm diameter implants. Rosette strain gauges were used for *in-vitro* tests, with peak values of the principal bone strain recorded with a data acquisition system. Implant stability as quantified by Periotest values (PTV) were also recorded for both types of implants. Experimental data were analyzed statistically using Wilcoxon's rank-sum test. In FE simulations, the peak value and distribution of von-Mises stresses in the implant and bone were selected for evaluation.
Results: In *in-vitro* tests, the peak bone strain was 42% lower for two-piece implants than for one-piece implants. The PTV was slightly lower for one-piece implants (PTV = −6) than for two-piece implants (PTV = −5). In FE simulations, the stresses in the bone and implant were about 23% higher and 12% lower, respectively, for one-piece implants than those for two-piece implants.

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

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Conclusion: Due to the higher peri-implant bone stresses and strains, one-piece implants (NobelDirect) might be not suitable for use as small-diameter implants.

Introduction

The use of small-diameter dental implants has become more popular in specific clinical situations such as a thin alveolar crest, replacing a tooth with small dimensions, or limited inter-radicular space. In addition to small-diameter implants, bone grafting procedure is an accepted treatment for placing wider implants in insufficient width of alveolar bone. However, some patients still refuse this kind of treatment because of the additional surgery (including tissue harvesting and bone grafting), cost, and pain. Especially for autogenous bone grafting, many complications including paraesthesia and morbidity of the donor site have been reported.[1]

Nevertheless, the use of small-diameter implants has to be considered along with their potential limitations. From a biomechanical aspect, small-diameter implants are structurally weaker than standard-size implants (3.75–4 mm in diameter). An implant with a smaller diameter also has reduced surface area to accommodate bone to implant contact, which influences bone stress/strain transference and these high stress/strains may jeopardize the support provided by the bone surrounding the implant.[2–4] Additionally, implants with smaller diameters have a high risk of fatigue failure.[5] Nevertheless, some studies still report good results for small-diameter implants.[6,7] Where alveolar bone width is limited, the use of narrow-diameter implants may produce good survival rates.[8,9]

Many researchers are cautious about using small-diameter implants,[10,11] since different designs of small-diameter implants have recently been introduced into the market.[5] Among these, a one-piece small-diameter implant has been presented as stronger than a two piece design due to the absence of an abutment-fixture connection and retention screw which are features of a two-piece implant. Additionally, the one-piece implants are purported to exhibit minimal resorption of peri-implant bone due to the absence of the microgap, which is a result of the implant-abutment junction. These microgaps have been associated with microleakage and bacterial contamination.[12,13] In addition, two-piece small-diameter implants have demonstrated higher mechanical failure rates associated with small-diameter screws, screw loosening, and fracture.[13] However, high long-term clinical survival rates for two-piece small-diameter implants (up to 95%) have been reported.[8,14,15]

Many studies[16,17] have examined the influences of the small diameter of implants based on biomechanical factors. However, until now, there is no study investigating the effect of implants with both small-diameter designs and one-piece or two-piece concepts on biomechanical performance. Therefore, the present study used both three-dimensional finite element (FE) simulation and *in-vitro* experimental analysis to evaluate the difference of two design concepts (one piece or two pieces) of small-diameter implants on the stresses and strains of the implant and surrounding bone.

Materials and methods

In-vitro experiments

Implant design parameters and bone specimen preparation

Two kinds of implant systems were selected for analysis: (1) a one-piece small-diameter implant (NobelDirect Groovy NP, Nobel Biocare, Gothenburg, Sweden) and (2) a two-piece small-diameter implant (NobelReplace Tapered TiU NP, Nobel Biocare; Figure 1). In order to discriminate these two models easily, "G-NP" and "T-NP" are used henceforth to represent the one-piece and two-piece variants, respectively; their diameter and length were 3.5 mm and 13 mm, respectively.

A Sawbones model of trabecular bone with a density of 0.4 g/cm^3 and an elastic modulus of 759 MPa (number 1522-05, Pacific Research Laboratories, Vashon Island, WA, USA) was prepared for attachment to 3-mm thick commercially available synthetic cortical shell (model 3401-02, Pacific Research Laboratories) with an elastic modulus of 16.7 GPa. The density of trabecular bone used in this study was simulated as Type 2 bone according to the bone-density classification of Misch.[18,19] The thickness of the cortical bone was consistent with that used by Hahn,[20] whereby Type 2 bone was associated with a cortical bone height of 2.5–4 mm. The

Figure 1 Two-piece (left) and one-piece (right) small-diameter implants.

synthetic bone had a rectangular shape with dimensions of 41 mm × 30 mm × 43.5 mm. Three specimens of artificial foam bone were prepared for each implant system.

Implant stability measurement. After an implant was placed into the Sawbones block, the mobility of the implant was measured using the Periotest device (Periotest Classic, Medizintechnik Gulden, Modautal, Germany). The tip of the measurement device was positioned perpendicularly at 2 mm from the abutment, and it impacted the implant four times per second for a 4-second period.[21] Periotest values (PTVs) were similarly measured four times in the four orthogonal directions for each model.

Strain gauge measurements. Rectangular rosette strain gauges (KFG-1-120-D17-11L3M3S, Kyowa Electronic Instruments, Tokyo, Japan) were attached to the buccal and lingual sides of the crestal cortical region of the bone model around the implant using cyanoacrylate cement (CC-33A, Kyowa Electronic Instruments; Figure 2A). A self-developed jig was designed with an adjustable rotational screwing device so that a 30° lateral force could be applied to the top surface of the implant in the experiments. Each loading procedure involved applying a force of 190 N to the cylindrical abutment using a universal testing machine (JSV-H1000, Japan Instrumentation System, Nara, Japan) with a head speed of 1 mm/min (Figure 2B).[22] When the forces were applied, signals corresponding to the three independent strains, ε_a, ε_b, and ε_c measured by the rosette strain gauge, were sent to a data acquisition system (NI CompackDAQ, National Instruments, Austin, TX, USA) and analyzed with the associated software (LabVIEW SignalExpress 3.0, National Instruments). After each measurement had been repeated three times for each specimen, the maximum (ε_{max}) and minimum (ε_{min}) principal strains were obtained as follows:

(1) $\varepsilon_{max} = 1/2(\varepsilon_a + \varepsilon_c) + 1/2\sqrt{[(\varepsilon_a - \varepsilon_c)^2 + (2\varepsilon_b - \varepsilon_a - \varepsilon_c)^2]}$

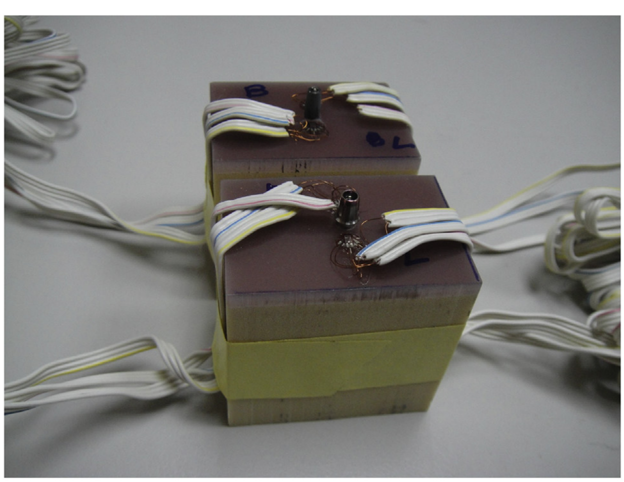
(2) $\varepsilon_{min} = 1/2(\varepsilon_a + \varepsilon_c) - 1/2\sqrt{[(\varepsilon_a - \varepsilon_c)^2 + (2\varepsilon_b - \varepsilon_a - \varepsilon_c)^2]}$

Statistical analysis. The measured primary implant stability and the peak minimum principal strains under loading both types of implants were summarized as median and interquartile range values. Comparisons of the measured data between the two implant systems were analyzed with Wilcoxon's rank-sum test. All analyses were performed using commercial statistical software (SAS version 9.1, SAS Institute, Cary, NC, USA) with an α value of 0.05.

Three-dimensional FE modeling

Computer-aided design software (SolidWorks 2009, SolidWorks Corporation, Concord, MA, USA) was used to construct the models of the G-NP and T-NP implant systems (Figure 3). A Sawbones models of size 41 cm × 30 cm × 43.5 cm were also constructed with a 3-mm thick synthetic cortical bone. All models were combined by Boolean operations (Figure 4), and the solid model was then exported in initial graphics exchange specification format to ANSYS Workbench 10.0 (ANSYS, Inc., Canonsburg, PA, USA) to generate the FE model using 10-node tetrahedral h-elements (ANSYS solid 187). The interface between the abutment and implant was set as contact with a frictional coefficient of 0.3.[23] The interface between the implant and bone was also set as contact, with frictional coefficients for the surface contacts of the rough implant surface with the cortical bone and trabecular bone assumed to be 0.4 and 0.8, respectively.

The material properties are listed in Table 1. The implants and abutment were assigned the material properties of titanium with homogeneous and isotropic elastic properties.[24] The material properties of bone blocks were determined from the data provided by Pacific Research Laboratories, and were also considered to be homogeneous and isotropic. The mesial and distal surfaces of the bone block were constrained as the boundary condition. A lateral



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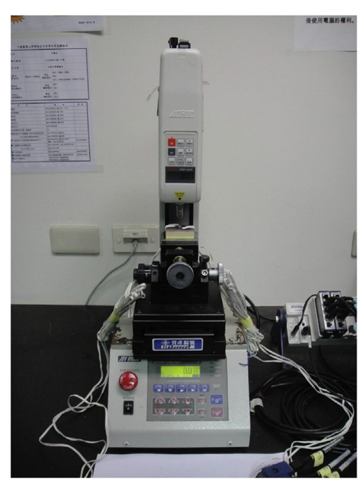

B

Figure 2 Strain gauges used and the universal testing machine: (A) two rosette strain gauges were attached to the top surface of the Sawbones block buccolingually near the implant; and (B) a 30° lingual lateral force of 190 N was applied to the top of the abutment by the universal testing machine.

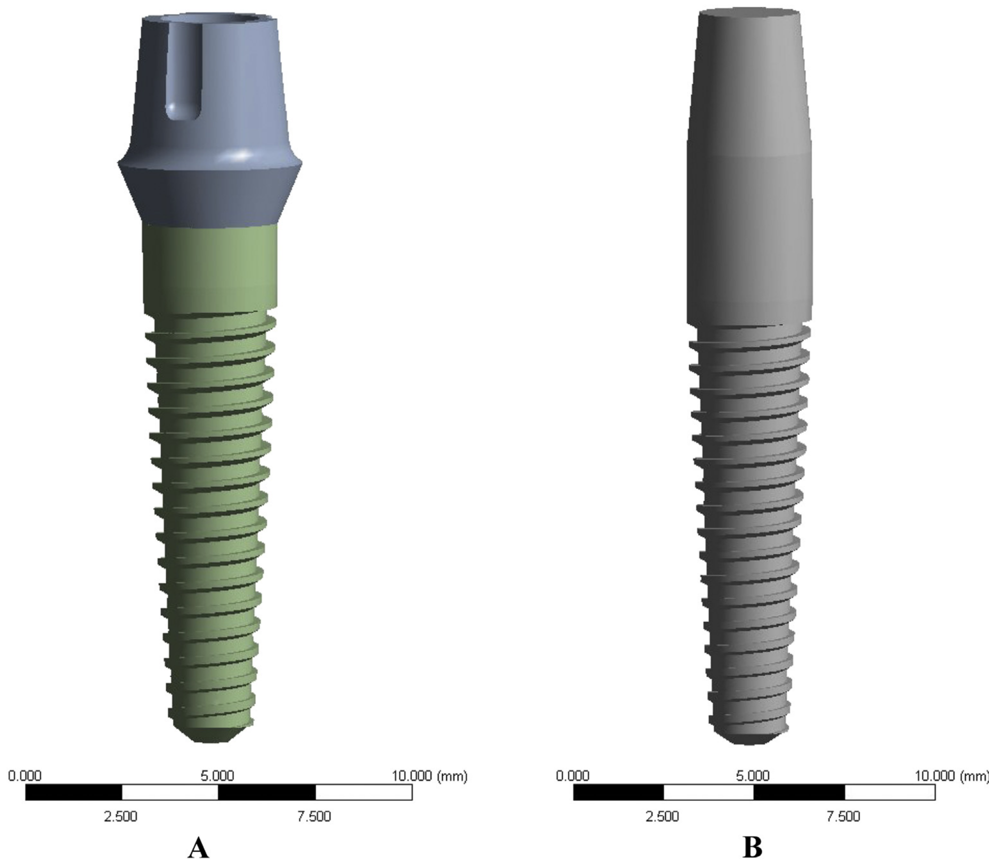

Figure 3 Computer-aided design models of a: (A) two-piece implant and (B) one-piece implant.

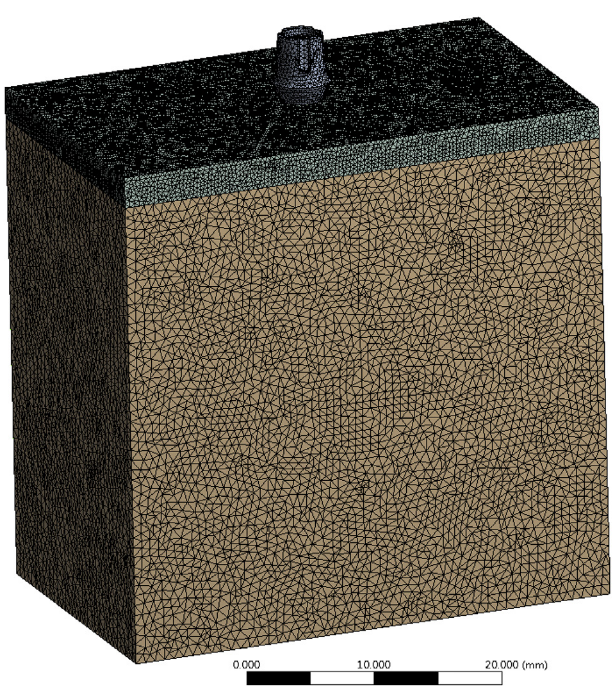

Figure 4 A three-dimensional finite element model of an implant and the surrounding bone.

Table 1 Material properties used in the finite element model.

Material	Young's modulus E (MPa)	Poisson's ratio ν
Cortical bone	16,700	0.3
Trabecular bone	759	0.3
Titanium	104,000	0.3

force of 190 N was applied to the whole top surface of the abutment at 30° relative to the long axis of the implant.[22] Based on the results of convergence testing, all of the elements were smaller than 0.4 mm in all of the models.

Results

In-vitro experiments

The peak bone strain on the lingual side did not differ significantly between the one-piece and two-piece implants, but it was 42% lower in the T-NP model than in the G-NP model on the buccal side (Table 2). The primary implant stability was slightly better in the G-NP model (median PTV = −6) than in the T-NP model (median PTV = −5; Table 3).

Three-dimensional FE analysis

The stress in the implant was highest at its neck and in the area near the first thread (Figure 5), with the peak value being about 12% lower in the G-NP model than in the T-NP model (Table 4).

The stresses in the alveolar cortical bone and trabecular bone were highest at the crestal region around the implant (Figure 6A), and they were also high near the apex of the implant (Figure 6B). The peak von-Mises stress in cortical bone was more than 23% higher in the G-NP model than in the T-NP model (Table 4), whereas that in trabecular bone was lower in the T-NP model (7.0 MPa vs. 8.9 MPa; Table 4).

Discussion

The clinical use of small-diameter implants is becoming more popular due to the increasing demand for oral implant therapy among the elderly as this is the population that also resists any invasive surgical procedures. Both one-piece and two-piece small-diameter implants have been developed, and their differing structures result in specific biomechanical characteristics, such as differences in the stresses and/or strains in both the implant itself and the surrounding bone. However, only a few researchers have investigated these biomechanical differences quantitatively.[25] The present study could be the first to have used both experimental strain-gauge measurements and an FE method of nonlinear contact analysis to investigate the biomechanical performances of one-piece and two-piece small-diameter implants. Sensors (e.g., strain gauges) can be used in experimental tests to accurately measure certain parameters on the surface, but the data are only obtained at specific locations. Moreover, a strain gauge attached close to the bone around an implant is unable to measure the peak value of the bone strain when it occurs inside the bone. In contrast, FE simulations allow the peak values and the distribution of the internal bone stresses and/or strains to be determined easily. However, the usefulness of an FE approach depends on how accurately it represents the real situation. Therefore, in this study three-dimensional FE simulations with nonlinear contact analysis and *in-vitro* experimental test were both used to facilitate the understanding of the biomechanical mechanisms relevant to one-piece and two-piece small-diameter implants and their surrounding bone.

Inflammation and occlusal overload have been the main mechanisms proposed for explaining marginal bone loss.[3,4] In order to maintain the bone level or improve the long-term success rates, for dentists an important goal is to minimize the stress and/or strain to the crestal bone around the implant. However, the implant surface area is

Table 2 Peak values of the principal strain of cortical bone around the implants of NobelDirect Groovy NP (G-NP) and NobelReplace Tapered TiU NP (T-NP) by *in-vitro* experimental tests.

Location	Model	Microstrain, median (IQR)
Buccal side	G-NP	−1119.28 (337.02)
	T-NP	−693.91 (264.35)
p[a]		<0.001
Lingual side	G-NP	148.08 (191.73)
	T-NP	108.65 (246.13)
p[a]		0.825

IQR = interquartile range.
[a] Multiple comparisons with Wilcoxon'srank-sum test.

Table 3 Periotest values (PTV) of implants for the models of NobelDirect Groovy NP (G-NP) and NobelReplace Tapered TiU NP (T-NP).

Model	PTV, median (IQR)
G-NP	−6.00 (0.00)
T-NP	−5.00 (0.00)
p[a]	0.005

IQR = interquartile range.
[a] Multiple comparisons with Wilcoxon'srank-sum test.

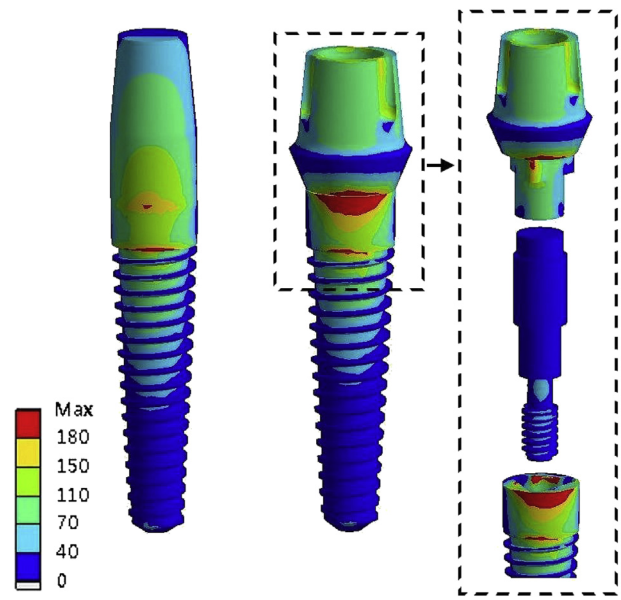

Figure 5 Von-Mises stress distributions in one-piece (left) and two-piece (right) small-diameter implants. Max = maximum.

Table 4 Von-Mises stresses of implants, cortical bone, and trabecular bone of NobelDirect Groovy NP (G-NP) and NobelReplace Tapered TiU NP (T-NP).

Model	Area	Von-Mises stress (MPa)
G-NP	Implant	220.5
	Cortical bone	151.7
	Trabecular bone	8.9
T-NP	Implant	250.9
	Cortical bone	115.3
	Trabecular bone	7.0

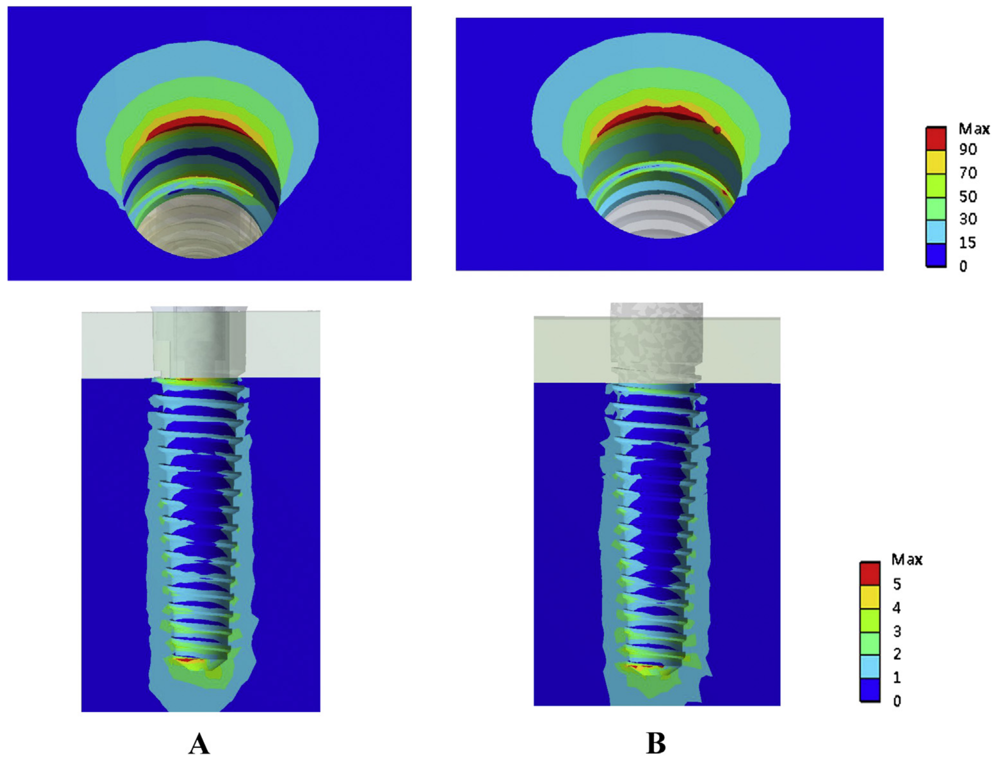

Figure 6 Stress patterns in cortical bone (upper figures in top view) and trabecular bone (lower figures in cross-sectional view) for an: (A) two-piece small-diameter implant; and (B) one-piece small-diameter implant. Max = maximum.

always smaller for a small-diameter implant than for a standard-size implant, and it is reasonable to assume that the bone stress and/or strain will be higher around the small-diameter implant. Changing the design of the implant could be possible to increase or decrease the alveolar bone stress around the implant. In the present study, it was found that the stresses and strains in the peri-implant bone for small-diameter implants were higher for one-piece design (G-NP) than for two-piece design (T-NP). Because overloading the bone has been considered as one of the reasons for leading peri-implant bone loss,[3,4] a small-diameter of one-piece G-NP implant may be associated with an increased risk of the bone loss around the implant.

One stated benefit of the one-piece G-NP implant is preservation of the marginal bone level due to the absence of an implant-abutment junction, but some short-term retrospective clinical studies have found these implants to be associated with poor outcomes.[6,12,26,27] The radiographic results indicated the presence of extensive marginal bone loss (>3 mm) around the G-NP implants. The success rate of G-NP implants has previously been reported to be lower for 3-mm diameter implants subject to immediate/early loading.[6] However, the results of the present study did not indicate that high stress will always result in overloading-induced bone loss when one-piece implants are used. Another design of one-piece implants showed no significant difference for the stress in the surrounding bone compared to two-piece implants.[25] Further investigation of detailed information related to the transmission of bone stress and/or strain around different designs of one-piece small-diameter implant is still required in the future.

The results of the FE simulations indicated that high stresses were induced in the two-piece small-diameter implant. This is consistent with Cehreli et al[25] finding that two-piece implants experience higher mechanical stresses under lateral loading. It may therefore be that a one-piece implant has a greater mechanical strength in clinical applications than a two-piece implant due to the inherent characteristics of a one-piece structure. Nevertheless, Allum et al[5] reported that the diameter seems to be the main factor influencing the fatigue strength of the small-diameter implant. Therefore, irrespective of whether one-piece or two-piece implants are used in clinics, caution is necessary for implants with diameters smaller than 3 mm due to the increased risk of implant fracture.

One of the limitations of this study was the simplified bone shape employed. Although the strength of the bone block used was similar to that of the jaw bone (based on the ASTM F1839 certification), the strain patterns are likely to vary with the bone geometry. Additionally, bone is a porous material with complex material characteristics (e.g., inhomogeneous, anisotropic, and viscoelastic properties). Future FE studies could employ more sophisticated simulations of the shape and material properties of bone, which might reduce the inconsistencies between the simulated and experimentally measured surface strains. Furthermore, the present study only applied a static occlusal force in both the experiments and FE simulations. Even though lateral force has been suggested to represent a realistic occlusal direction,[28] chewing simulation—especially for tooth-to-tooth contact—needs to be considered in future investigations.[29]

Within the limitations of this study, the following conclusions can be drawn: (1) using a one-piece small-diameter G-NP implant might increase the stress and/or strain in peri-implant bones and increases the risk of overloading-induced bone loss; and (2) the mechanical stress in the implant itself is higher in a two-piece small-diameter implant.

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