Probability of survival and stress distribution of narrow diameter implants with different implant-abutment taper angles

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ABSTRACT

This study evaluated the probability of survival, failure mode and stress distribution of narrow diameter implants (NDIs) with internal implant-abutment conical connection comprised of different taper angles and thread designs. Sixty-three NDIs (Ø 3.5 × 8.5 mm) were divided according to the taper angle (TA), internal diameter (ID) and trapezoidal thread design (TD) (n=21/group), as follows: (i) 11.5°U (11.5° TA; ID: 2.5 mm; TD: dual threaded); (ii) 11.5°S (11.5° TA; ID: 2.5 mm; TD: single threaded); (iii) 16°S (16° TA; ID: 2.72 mm; TD: single threaded). They were subjected to step-stress accelerated life testing. The reliability and use-level probability Weibull curves were calculated at 50, 100, and 150N for a mission of 100,000 cycles and the failure mode was analyzed using a scanning electron microscope. For finite element analysis the von-Mises stress (σ_{vM}) was calculated for the abutment and implant. All groups showed high reliability (above 84%) and failures occurred predominantly in the abutment. In the FEA, 11.5°U showed higher σ_{vM} for the implant. All NDIs showed high reliability at clinically challenging loads. The system with greater taper angle showed lower σ_{Vm} in the implant, and dual threaded implants showed a higher stress concentration in the implant and cortical bone.

Keywords: dental implants, fatigue, finite element analysis, narrow diameter implants, probability of survival.

INTRODUCTION

According to the classification proposed by the ITI Consensus Report[1] narrow diameter implants (NDIs) have an external diameter ranging up to 3.5 mm, and are classified into three categories: Category 1: Implants with a diameter of <2.5 mm; Category 2: Implants with a diameter of 2.5 mm to <3.3 mm; and Category 3: Implants with a diameter of 3.3 mm to 3.5 mm[1]. These implants were developed to minimize surgical grafting procedures, facilitating the rehabilitation in areas where there is insufficient bone volume or limited interdental space.[2]

However, previous studies have shown that NDIs presented a significantly higher failure rate (25%) compared to larger diameter implants (13%) over the same follow-up periods.[3] Olate et al.[4] have demonstrated in a retrospective study that the largest loss of implants was observed in narrow implants (5.1%), followed by regular (3.8%) and wide (2.7%) implants.[4] Moreover, implant diameter appears to affect the failure rate, increasing the risk of abutment and/or implant fracture due to reduced hardware design and compromised mechanical stability of some specific abutment-implant connections, i.e. external hexagon connections.[5,6]

Hence, NDIs with an internal conical connection are commonly used in clinical practice. When presenting 3.5 mm in diameter, such implants typically present an inner diameter (ID) of approximately 2.5 mm and a taper angle (TA) of around 11.5 °. The study of Bozkaya et al.[7] has shown that the efficiency of the abutment-implant system, which is the ratio of the loosening torque to the tightening torque, depends on the taper angle of the implant, where an increase from 1 ° to 10 ° resulted in a decrease in efficiency from 0.97 to 0.9, depending on the value of the friction coefficient. Moreover, NDIs have thin cervical walls, and any increase in the taper angle, may impact the thickness of the wall, eventually influencing the survival of the system.

Furthermore, the impact of the taper angle on mechanics and stress distribution in NDIs are still scarcely addressed in the literature.

Besides the taper angle of the implants, the thread shape of each system can influence the stress distribution to the implant-bone interface, which may affect the balance between bone remodeling process since these factors are directly associated with stress.[8–11] Therefore, the design of the implant thread can influence the success or failure of implant-supported rehabilitation.[11–13] This study evaluated the probability of survival, failure modes, and stress distribution of NDIs with internal conical connection comprising different degrees of internal taper angle and different thread designs.

MATERIALS AND METHODS

For the mechanical testing, sixty-three narrow diameter implants (Ø 3.5 mm × 8.5 mm) fabricated with commercially pure grade IV titanium with internal conical connections (SIN Implant, São Paulo, Brazil), were assigned in three groups (n = 21 / group) according to the taper angle (TA), internal diameter (ID), and thread design (TD): Group 11,5°U: Unitite Prime (11.5° TA; ID: 2.5 mm; TD: dual trapezoidal threads); Group 11,5°S: Strong SWC (11.5° TA; ID: 2.5 mm; TD: single trapezoidal threads) and Group 16°S: Strong SW Plus (16° TA; ID: 2.72 mm; TD: single trapezoidal threads).

The implants were inserted in acrylic resin (Jet; Dental Articles Classic Ltda, Campo Limpo Paulista -SP, Brazil) using a silicone matrix as support (Zetalabor; Zhermack SpA, Badia Polesine, Italy). The platform of all implants was positioned at the same level as the matrix.[14–16] Then, the prefabricated universal abutments (4.5 x 2 x 6 mm) (SIN Implant System, São Paulo, Brazil) were torqued into the implants (Tohnichi BTG150CN-S; Tohnichi America, Buffalo Grove, IL, USA) following the manufacturer's instructions (20 N.cm). After this stage, the upper central incisors' crowns, made of cobalt-chromium alloy (Wirobond 280; BEGO, Bremen, Germany), were cemented onto the abutments with a resin cement (Rely X U200; 3M Oral Care, St Paul, MN, USA). Three samples from each group were submitted to a single load to failure test (SLF).[17] In SLF, an uniaxial compression load (Instron model 4411; Instron, Canton, MA, USA) was applied on the incisal edge of the superior central incisor, 30° off the axis (ISO 14801: 2016) at a speed of 1mm/min by a flat indenter made of tungsten carbide.

The remaining eighteen samples from each group were submitted to SSALT, divided into step-stress profiles following the ratio distribution of 3:2:1, which were: mild (n=9), moderate (n=6), and aggressive (n=3). These profiles were designed based on the mean values of the fracture load obtained in each group's SLF test.[17] The SSALT analysis was performed under water at 15 Hz using an all-electric dynamic test system (ElectroPuls E3000 Linear-Torsion; INSTRON, Norwood, MA, USA). A flat tungsten carbide indenter was used to apply the load at the crown's incisal edge, 30° off-axis, as previously mentioned (ISO 14801:2016).[17] All samples were tested until suspension (no failure up to the maximum load level of 1,500 N) or failure (bending or fracture of the implant or the abutment).

Based on the failure distribution, the use level probability Weibull curve (probability of failure versus number of cycles) using a load of 100 N was calculated and plotted using the Weibull distribution and the life-stress relationship by the inverse power law for damage accumulation (90% two-sided confidence interval, CI) (Synthesis 9, Alta Pro 9; Reliasoft Tucson, AZ, USA). The reliability was calculated at 50, 100, and 150 N for completing a mission of 100,000 cycles (90% two-sided CI). The use level probability Weibull analysis provided the beta value (β), which describes the behavior of the failure rate of the material over time.[17] If the calculated use level probability Weibull beta parameter of any group was <1, then a Weibull 2-parameter contour plot (Weibull modulus vs. characteristic strength) was calculated using the final load to failure data of all groups.[17]

The failure analysis was performed under a polarized light microscope (AxioZoom V16; Zeiss, Oberkochen, Germany) to examine the failed specimens, then scanning electron microscopy (SEM) (JSM-5600LV; Jeol, Boston, Massachusetts, USA) was used for fractographic analysis.

For finite element analysis, a CAD software (SolidWorks 2013; SolidWorks Corporation, Concord, MA, USA) created 3D virtual models of a crown, cement layer, abutment, implant, cortical and cancellous bone. The CADs of the implants and abutments were different according to TA, ID, and TD (trapezoidal or square), but the external diameter and the length were standardized (Ø 3.5 mm × 8.5 mm). A 70µm cement layer was simulated between the crown and the abutment. All models were exported to Ansys Workbench 15.0 (Ansys, Inc, Canonsburg, PA, USA) software for mathematical analysis, and a tetrahedral mesh (0.7 mm elements-size) was generated. The properties of each material were inserted in the program, being the elastic modulus and Poisson's ratio for each material, respectively: titanium for implant and abutment (104 GPa, 0.34),[18] resin cement (18.3 GPa, 0.33),[19] cortical bone (13.6 GPa, 0.26),[18] cancellous bone (1.36 GPa, 0.31),[18] and cobalt-chrome alloy (218 GPa, 0.33).[18] The models were considered linearly elastics, isotropic and homogeneous. The contact conditions between implant and abutment were assumed as "no separation," and the contacts between crown and abutment and between implant and bone, were assumed as "bonded." A load of 49 N was applied at the incisal edge of the crown, 30° off-axis. Von-Mises criteria (σ_{vM}) and deformation in millimeters were registered for the implant and the abutment, and the criterion of minimum principal stress (σ_{min}) and maximum shear stress (τ_{max}) were registered for the cortical and cancellous bone.

RESULTS

The use-level probability Weibull curves (90% 2-sided confidence bounds), showing the probability of failure versus number of cycles at a use load of 100 N were plotted in Figure 1. According to this analysis, the beta values (β) were provided for all groups, which were: β=2.35 for 11.5°U group, β = 0.44 for 11.5°S group and β = 2.33 for 16° S group, indicating that failure rate was chiefly controlled by fatigue damage accumulation for the 11.5°U and 16° S groups, and by material strength and egregious defects for the11.5°S group.

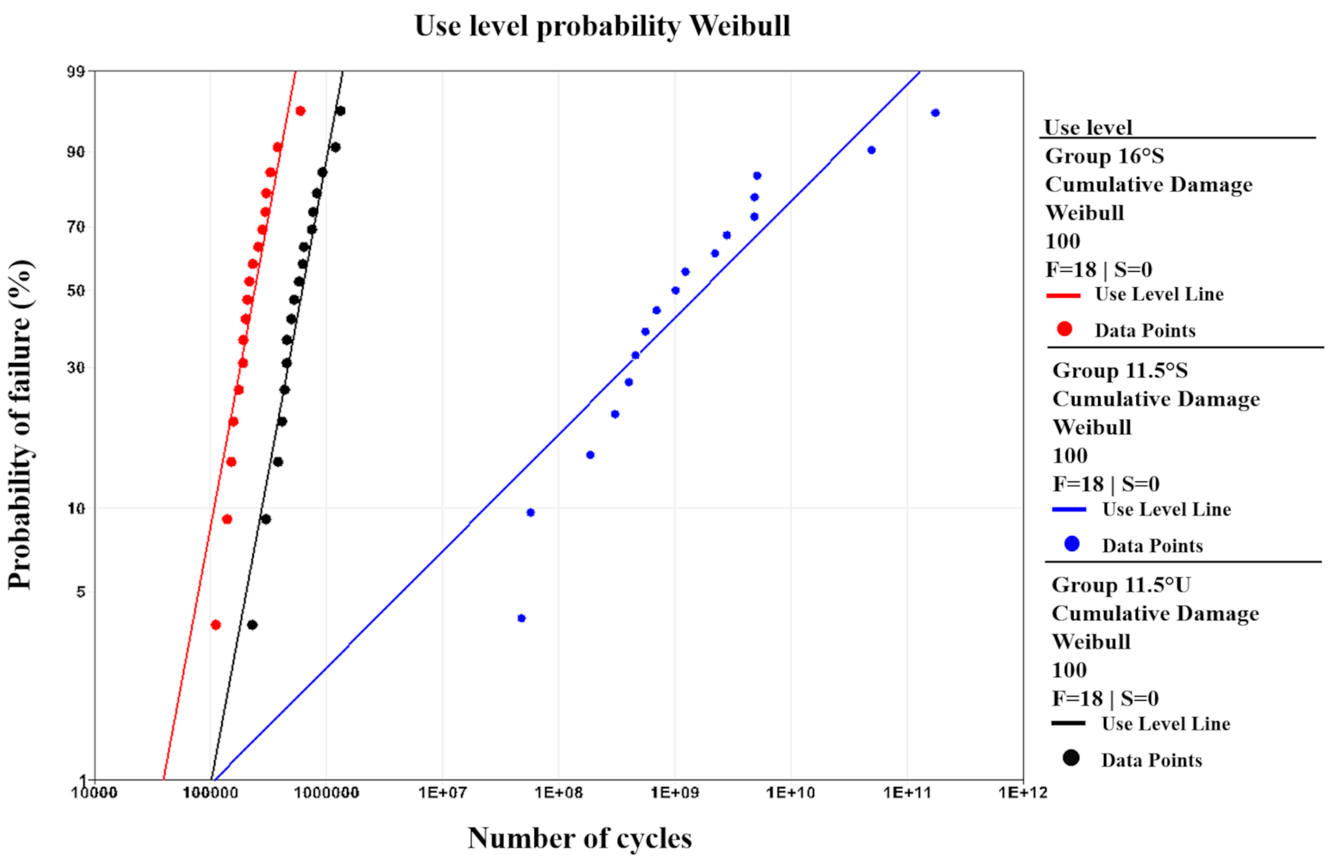

Figure 1. Use level probability Weibull (90% confidence bound) showing the probability of failure versus number of cycles for tested groups (set load of 100 N).

The contour plot showing the Weibull modulus (m) and the characteristic strength (η) are represented in Figure 2, showing that the Weibull modulus was not significantly different between groups (5.21–8.74). Furthermore, the characteristic strength values were not significantly different between 16° S (317 N) and 11.5°S (336 N) considering the overlap of the contours, and the 11.5°U group (244 N) was statistically different among all groups.

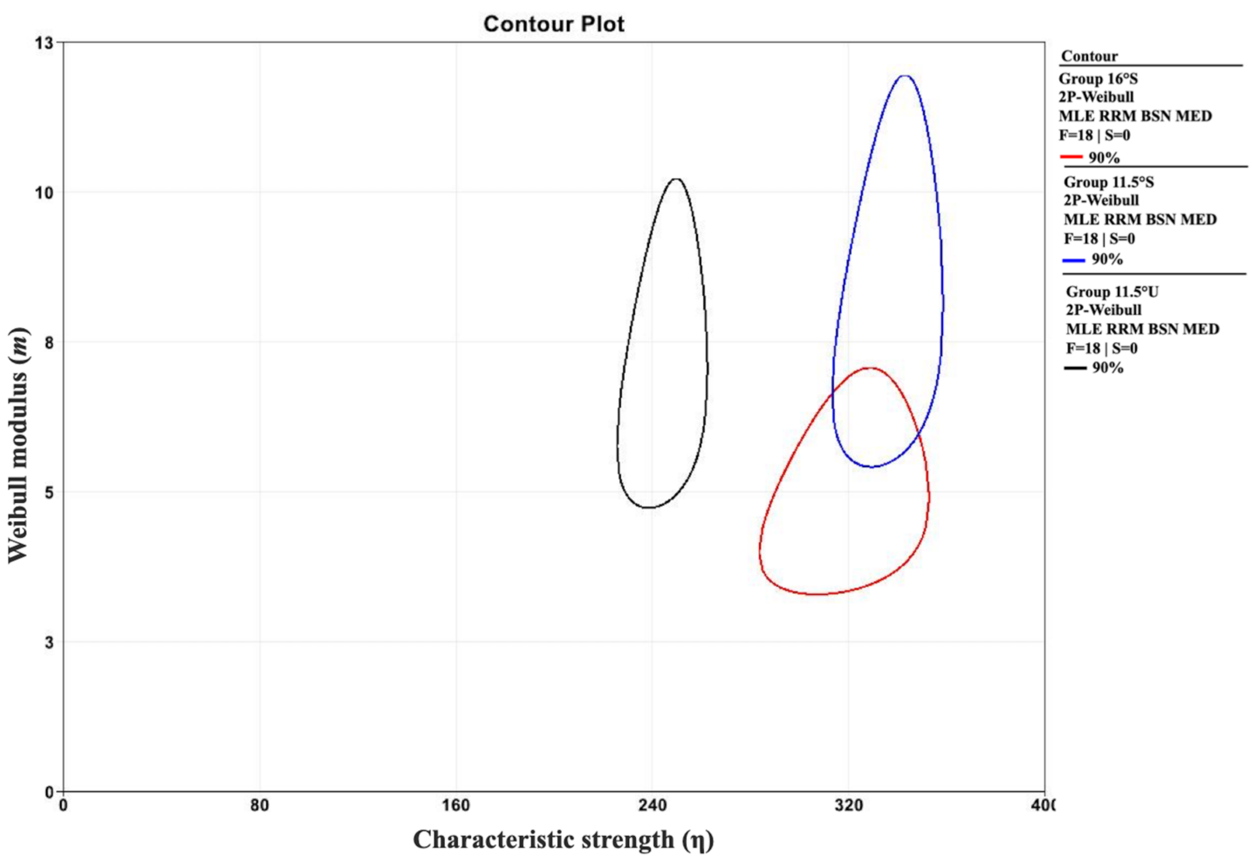

Figure 2. Contour plot showing Weibull modulus (m) and characteristic strength (η).

The probability of survival for completion of a mission of 100,000 cycles at 50, 100 and 150 N was presented in Table 1. In this analysis, it was possible to observe that the reliability was higher than 84% for all groups in all loads. Besides that, all samples failed after SSALT, and the failure was restricted to fractures in the abutment. The SEM micrographs demonstrate the fractured samples' failure mode, shown in Figure 3, in which fracture initiated at the lingual side, where the loading condition caused a local tensile stress, and propagated to the opposite buccal side.

Table 1. Calculated reliability (%) for a given mission of 100,000 cycles at a set load of 50, 100 and 150 N.

		50 N			**100 N**			**150 N**		
		11,5°U	16°S	11,5°S	11,5°U	16°S	11,5°S	11,5°U	16°S	11,5°S
	upper bound	1	0.98	1	1	0.97	1	0.93	0.95	0.99
100,000 cycles	**Reliability**	**1**	**0.94**	**1**	**0.99**	**0.91**	**0.99**	**0.84**	**0.87**	**0.97**
	lower bound	0.99	0.85	0.98	0.95	0.78	0.95	0.66	0.69	0.88

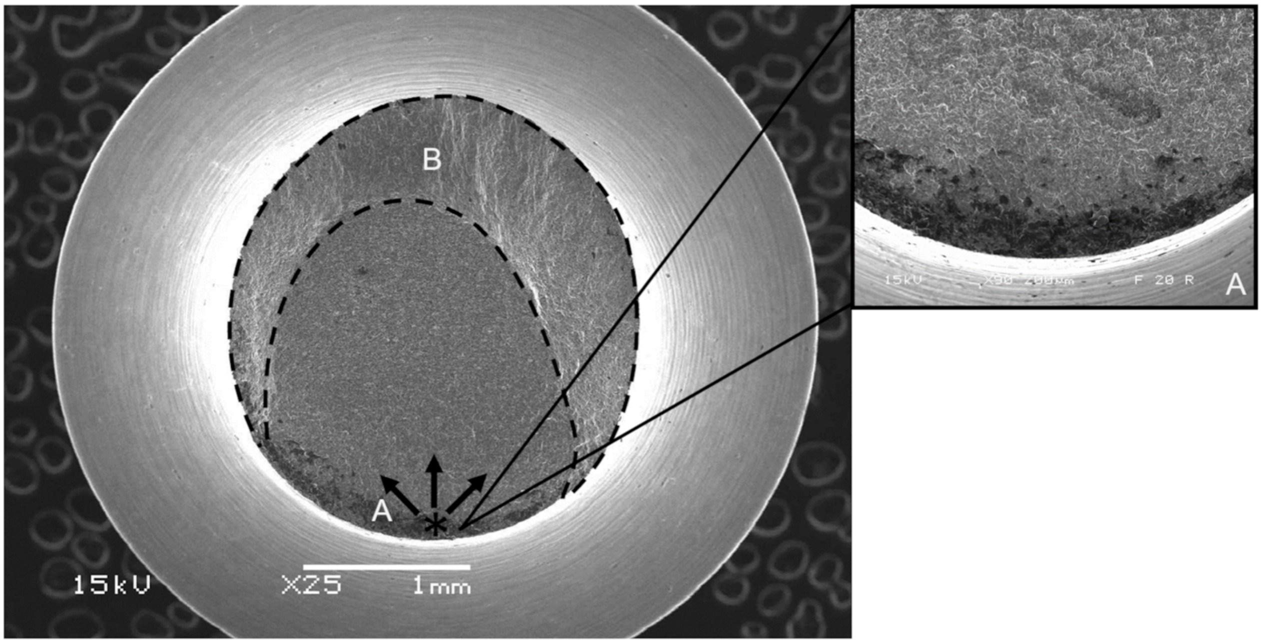

Figure 3. Abutment fractured during SSALT (25x magnification). Fracture origin (black asterisk) where the surface was subjected to cyclic tensile stress, magnified on the right side (A). The black arrows indicate the direction of crack propagation. The rupture zone at the area submitted to compression stress was represented on the opposite side (B).

Figure 4 shows the place where there was a higher stress concentration, as evidenced in the FEA and the view of the fracture in the abutment by the SEM. In all models, the abutment and the internal walls of the implant were surfaces with the highest stress concentration (Figures 5 and 6). Data on the von-Mises criteria (σ_{vM}), deformation in millimeters, minimum principal stress (σ_{min}), and shear stress (τ_{max}) of all groups are presented in Table 2. These results demonstrated that 11.5°U group showed higher σ_{vM} for the implant when compared to 16°S and

11.5°S groups (36,99% and 30,29% higher, respectively). In addition, the stress distribution in the abutment resulted in lower σ_{vM} for the 16°S group compared to 11.5°U and 11.5°S groups (23,39% and 29,97% lower, respectively). The σ_{min} and τ_{max} in the cortical bone demonstrated a higher stress concentration for the 11.5°U group when compared to 11.5°S (31.49% and 28.16% higher, respectively) and 16°S groups (24.83% and 48.34% higher, respectively).

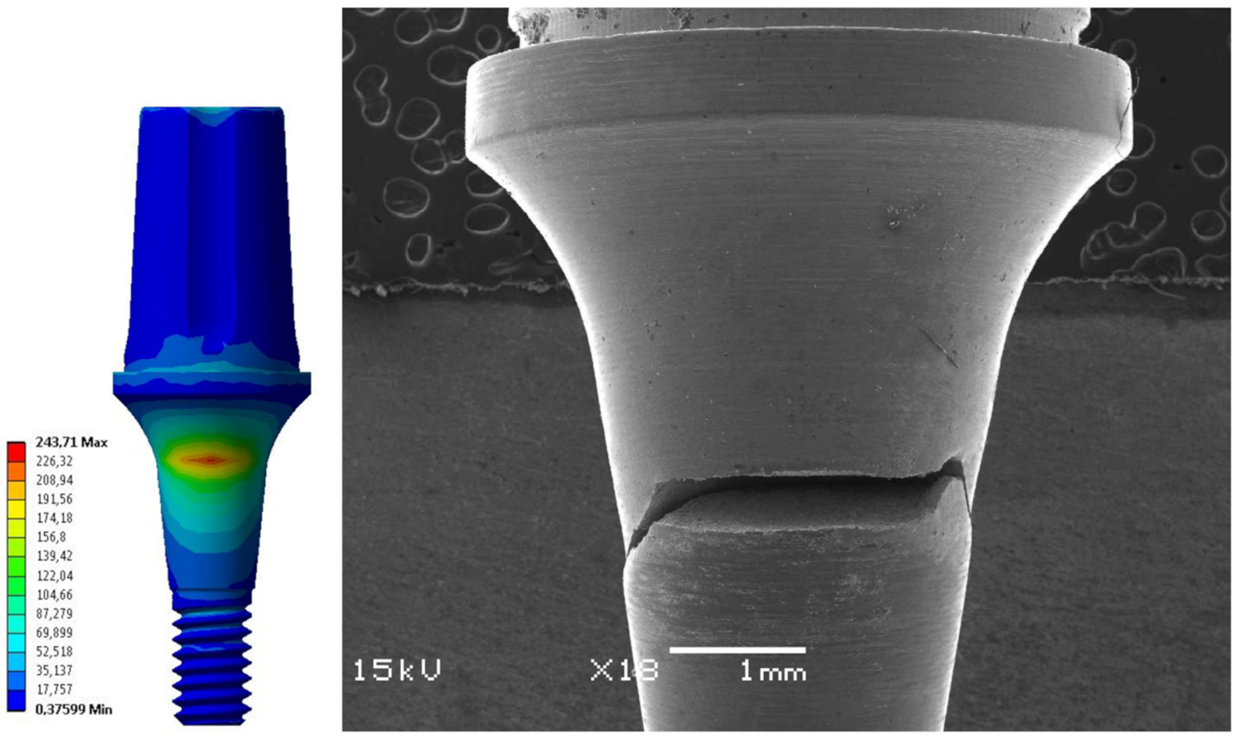

Figure 4. Von-Mises stress peak concentration in fractured abutment comparable at SEM images after the SSALT test.

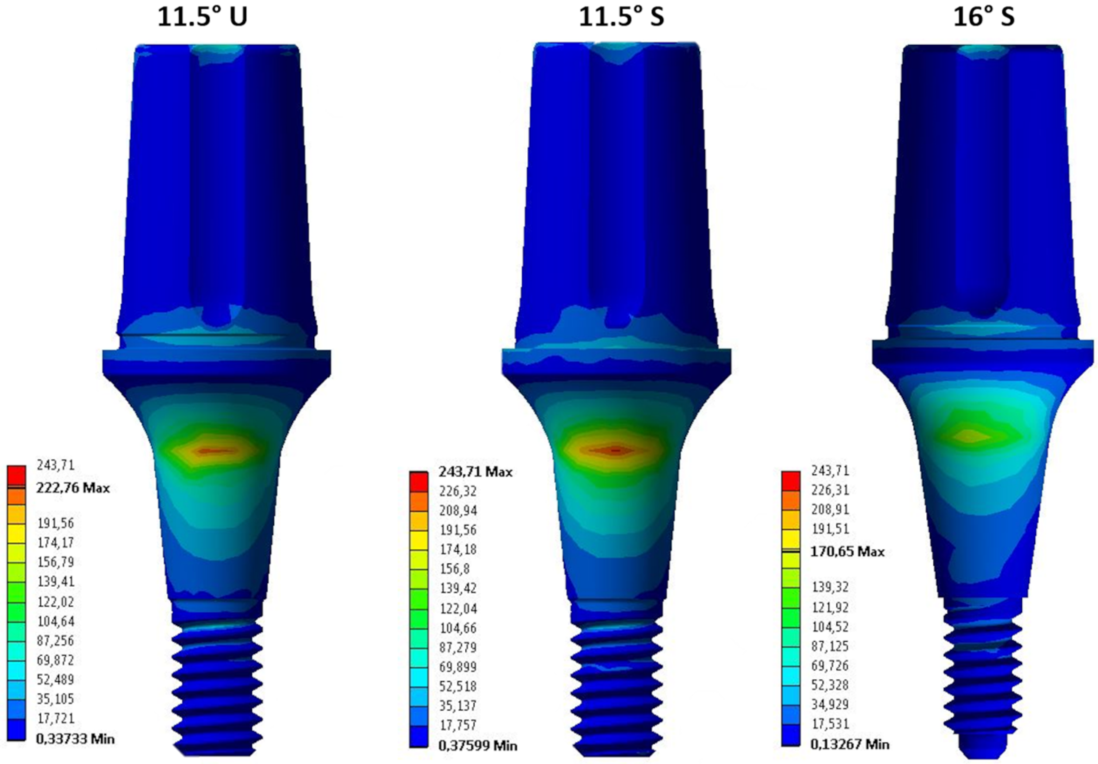

Figure 5. Stress peak concentration at the abutment for all groups.

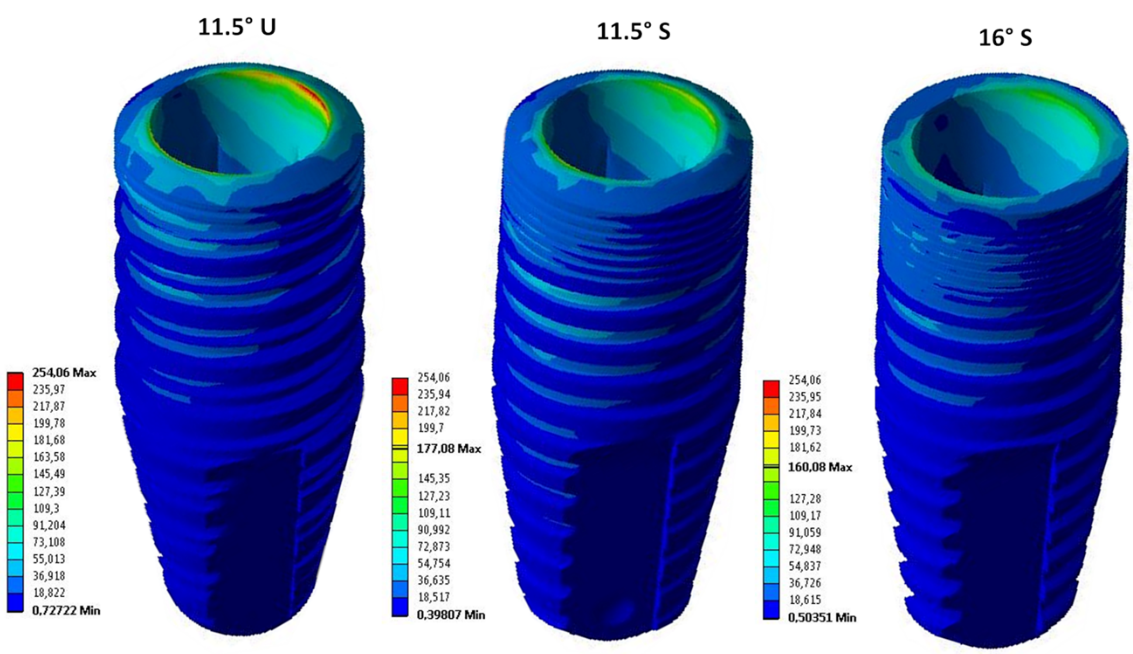

Figure 6. Stress peak concentration at the implant for all groups.

Table 2. Von-Mises criteria and deformation (mm) of implants and abutment. Minimum principal stress and shear stress for cortical and cancellous bone (MPa).

	11.5° U	16° S	11.5° S
σ_{vM} implant (MPa)	254.06	160.08	177.08
σ_{vM} abutment (MPa)	222.76	170.65	243.71
Deformation for implant (mm)	0.0053704	0.0050817	0.0050461
Deformation for abutment (mm)	0.032954	0.026581	0.031584
σ_{min} Cortical bone (MPa)	33.109	26.522	25.179
σ_{min} Cancellous bone (MPa)	1.9665	2.0305	2.6
τ_{max} Cortical bone (MPa)	18.247	12.3	14.237
τ_{max} Cancellous bone (MPa)	1.6009	1.4137	2.0467

DISCUSSION

Understanding the impact of the variations in the design of the implant-abutment system in fatigue and stress is essential to predict potential clinical complications.[11,20] In this context, this study evaluated the probability of survival and stress distribution of narrow diameter implants (NDIs) with an internal conical connection comprising different internal taper angles and different thread designs. Although implants with single trapezoidal threads and greater taper angle (TA= 16°) presented better stress distribution in the implant, abutment, and cortical bone relative to single and dual trapezoidal threads and lower taper angle (11.5°), the three designs tested showed high reliability at clinically relevant loads.

In fact, the current results suggest that all implants, regardless of TA, presented high reliability (up to 84%) simulating maximum bite forces in the anterior region (50-150 N).[21–23]

Previous studies have also indicated high reliability for NDIs, exhibiting similar probability of survival at 100 N in a mission of 50,000 and 100,000 cycles (97% and 96%, respectively)[24]. Moreover, the current findings are in accordance with a previous study of Bordin et al. (2018)[15], which investigated NDIs with different prosthetic connections and presented higher reliability for a given mission of 50,000 cycles at load of 75 N for internal conical implants relative to external hexagon implants.

The beta value provided by the SSALT data analysis is important to understand the failure rate over time. Therefore, the groups 11.5° U and 16° S showed a beta value >1, indicating that, irrespective of the difference in the taper angle, internal diameter, and macrogeometry, the fatigue damage accumulation was an acceleration factor for failure and the failure rate increased over time.[17] Consequently, it is possible to assume that the longer the abutment/implant system remains in the mouth, the likelihood of failure due to the accumulation of damage will increase once this system receives constant and repetitive loads. However, for the 11.5 ° S group, the beta value was <1, indicating that, for this group, the implant-abutment connection failure was controlled by the strength of the material.[5]

Although the SSALT analysis has shown high reliability for all groups, the FEA demonstrated higher von Mises stress criteria for the implants in the 11.5°U group. Similarly, for the minimum principal stress and shear stress for the cortical bone, 11.5°U group showed higher stress concentration than the other groups. These results demonstrated that the dual trapezoidal thread design of the implant threads results in a higher stress concentration in the cortical bone and implant when compared to implants that have single trapezoidal threads. Our results corroborate with studies of Chowdhary et al. (2013)[25], De Andrade et al. (2017)[8] and Hansson and Werke (2003)[26], which indicated a low stress concentration in the cortical and cancellous bone in implants with single trapezoidal threads that was attributed to its smaller flank angle at the bottom of the thread. Despite the differences observed in the minimum principal stress levels among groups, a previous study indicated that approximately 50 MPa is the threshold compressive stress to trigger bone resorption, which is notably higher than the values herein presented for all implant systems (25-33 MPa)[25].

From a mechanical perspective, it could be expected that the increase in the ID and TA and decrease in the implant wall of group 16°S would cause an increase in the stress concentration on the implant surface [26]. This pattern was observed in the study of van Staden et al. (2008)[26], which, using the FEA, observed a progressive increase in the stress concentration at the implant-abutment interface when reducing the external diameter of implants, where 3.5 mm implants showed a higher stress concentration compared to implants with external diameters of 4.0, 4.5, and 5.5 mm, respectively. These results are in contrast with the current findings, where the 16°S group presented lower stress concentration at the implant and abutment than other groups. A possible explanation for this fact is the increase in the internal diameter and the conical interface of the implant, providing a proportional increase in the prosthetic component, and ensuring a greater contact area between the prosthetic component and the implant, which may have improved the stress distribution and provided better results for the implant-abutment device.

The FEA results allowed the identification of the area of higher stress concentration in the system, the abutment collar level, which is the area where all fractures occurred during SSALT.[14,15,27] The origin of the fracture and the direction of crack propagation of the prosthetic component were from the lingual to the buccal sides, where the forces naturally occur in the buccal environment, as depicted in the fractographic analysis .[14,28–30] The predominant abutment fracture may be associated with the contact of the prosthetic component with the internal walls of the implant that promotes the implant-abutment friction locking and protects the implant from fractures and deformations.[15,28,31]

One of the main advantages of SSALT is that the test reproduces failures quickly by increasing stress levels while reproduces, *in vitro*, the fracture modes observed in clinic.[17] In addition, it is important to highlight that the SSALT failure mode corroborated with the areas of higher stress concentration in the *in silico* analyses, increasing the validity of the results of the present study. However, a limitation of the current fatigue analysis was the use of metallic crown instead of an all-ceramic one. Although the reason was to confine failures within the desired variables (implant and/or abutment), the assembly fails to represent a clinical scenario. Also, under clinical conditions, masticatory function occurs at variable and low stress rates.[17] Therefore, it is essential to acknowledge that future clinical trials on narrow implant systems with varying internal taper should be conducted.

CONCLUSION

Based on the results of this study, the following conclusions can be drawn:

1. All narrow diameter implants tested showed high reliability at clinically relevant loads of anterior teeth, and failure was restricted to abutment fracture.

2. Implant systems with greater taper angle, 16° TA, and single trapezoidal threads showed lower σ_{Vm} for the abutment, implant, and cortical bone relative to systems 11.5° TA with single and dual trapezoidal threads.

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CONFLICT OF INTEREST

There are no conflicts to declare.

DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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