Influence of Abutment Design on Stress Distribution in Narrow Implants with Marginal Bone Loss: A Finite Element Analysis

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Purpose: The aim of this study was to determine the effect of solid (one-piece) and two-piece abutments on the stress profile of narrow implants with marginal bone loss. *Materials and Methods:* Solid and two-piece abutments were connected to a conical internal octagon–connection implant (3.3 mm in diameter, 10 mm in length) and restored with a single crown. Three-dimensional finite element analysis was used to simulate the stress distribution in implant models with different levels of marginal bone resorption (0, 1, 2, and 3 mm). The effect of the design variables under increased bone resorption scenarios, including abutment screw length and diameter, were assessed. Static loading was applied to determine the mechanical response of the implant and cortical and trabecular bone. *Results:* Marginal bone resorption levels dominated the mechanical response under static loading conditions. A marginal bone loss of 3 mm significantly increased stress values in the implant vicinity and abutment screw. Both abutment designs displayed similar stress distribution in the surrounding bone, but lower stress values were observed in the implant body with two-piece abutments. The abutment screw length was more effective in the resultant stress, as the longer screws reduced the stress in the implants. *Conclusion:* Marginal bone resorption magnitude is the crucial parameter in biomechanics to determine the mechanical behavior. As bone loss increases, resultant stress around implants under mastication forces may lead to implant failure, regardless of abutment type. *Int J Oral Maxillofac Implants 2021;36:XXX–XXX. doi: 10.11607/jomi.8554*

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Standard dental implants can be defined as implants with diameters between 3.75 and 5 mm, and they have been used successfully for the rehabilitation of various forms of edentulism.[1] Tooth loss due to periodontal diseases, trauma, dental infections, or various bone pathologies leads to a decrease in residual alveolar bone. An insufficient volume of horizontal crestal bone might prevent the placement of a standard-diameter dental implant (SDI), as at least 1 mm of residual bone surrounding the implant is necessary for implant survival. Narrow-diameter implants (NDIs), which have a diameter of ≤ 3.5 mm, can be used in bone deficiencies as an alternative to bone grafting procedures.[2] NDIs of 3.3 to 3.5 mm are reported to be indicated in posterior single-tooth restorations, with results comparable to

SDIs.[3] However, a reduced implant diameter compromises the mechanical properties of the implant structure, as shown in the lower implant removal torque values and the reduced bone-to-implant contact surface.[4] Although NDIs might be more prone to marginal bone loss than SDIs,[5] comparable crestal bone levels and survival rates are reported in the literature.[6] However, the stress behavior of NDIs in reduced bone-to-implant contact conditions due to marginal bone loss remains unclear. **[AU: Editing okay?]**

Implant failure is not only limited to biologic factors, but can also be the result of mechanical problems, including fractures due to fatigue and overloading, which affect long-term implant survival. When the resistance limit of the material is exceeded, fractures can occur in the implants or abutments, with the abutment screw being the most common site of failure.[7] **[AU: Editing okay?]** Various factors contribute to the mechanical integrity of abutment screws, including the screw material, design and diameter, the implant-abutment interface connection, and preload. Depending on the system preferred, an abutment may be fixed to the implant by a separate screw (a two-piece abutment), or it may consist of a mechanism with threads connected directly on the abutment body itself (solid abutment). In order to allow passage for the abutment screw, material is removed from the abutment interior, creating thinner walls, which might suffer small changes in shape,

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resulting in increased frictional force under axial loads.[8] Abutments with a solid design tend to show greater resistance to deformation under oblique loads, due to their friction-fit rather than internal-hex connection.[9] In contrast to complications such as screw loosening or breaking in two-piece abutments, solid abutments provide better force transmission, which may decrease abutment loosening, as the abutment connection mostly relies on the frictional resistance of Morse taper and contact pressure.[10]

The screw joint of the abutment is vital for the integrity of the implant-abutment system under loading, as the loss of preload leads to instability and failure.[11] Only a few studies on screw design to prevent such failures have been carried out,[12–14] and the impact of screw design variables, including its length and width, on stress distribution in narrow implants has not been clarified.

Considering the masticatory load transfer to the implant body and surrounding bone through the abutment, the implant-abutment connection type and the design of its components might play a key role in the modification of this load, as well as the stability of the system. In particular, in NDIs with inferior mechanical strength and higher stress levels than SDIs,[15] the preference for one type of abutment over another can be critical in terms of long-term mechanical stability. The design parameters of an abutment, such as the length and the width of the abutment screw, might be important in the changes of the stress/strain distribution in the implant-abutment system. Therefore, the primary objective of this study was to assess the effect of abutment type and the marginal bone loss on the stress distribution of NDIs under static loading. The secondary aims were to evaluate the effect of the varying screw dimensions in the modeled implant-abutment systems with bone resorption.

MATERIALS AND METHODS

The study consisted of 3D models of three different bone levels for implants defined as peri-implant bone resorption of 0, 1, 2, and 3 mm. Two types of abutments were selected, a solid abutment or a two-piece abutment. CBCT images of an edentulous human mandible were used to construct a 3D finite element bone model via 3D-Doctor software (Able Software). Bone quality D2 was simulated for bone models with cortical bone thickness of 1.5 mm in the crestal and 1.0 mm in the buccal and lingual aspect, to avoid inconsistent strains. A titanium implant of 3.3-mm diameter and 10-mm length (Xxx, XXX) [AU: What should these be?], with a conical internal octagon design with an 11-degree Morse taper, was simulated within the premolar area. A mandibular second premolar cement-retained porcelain-fused-to-metal crown restoration model was generated and connected with abutments (Fig 1). The metal framework was determined as 0.8 mm thick, and porcelain was prepared as at least 2 mm, which was then modeled accordingly with the premolar anatomy. [AU: Editing okay?] Cement thickness was ignored, due to its low volume and its minimal impact on mechanical properties. All the models were connected to abutments with screw sizes of 1.2, 1.5, and 1.8 mm in diameter and 2.0, 2.8, and 3.6 mm in length (Fig 2). Details for each model are given in Table 1.

All materials were considered to be homogenous, isotropic, and linearly elastic. Implants were assumed to have 100% osseointegration. The finite element meshes consisted of 10-node tetrahedral structures, and the exterior nodes at the mesial, distal, and inferior surfaces of the mandible segment were fixed at each edge of the mandibular corpus to be set as the boundary condition for all models (Fig 3). Cortical bone and trabecular bone, implant and bone, implant and abutment, and abutment and implant-supported crown were considered to be in full contact. The convergence test was used to refine meshes. Regarding the convergence monitoring, the maximum von Mises stress in the bone was used with a tolerance of 5%. If a change was < 5% in bone tissue, including the cortical and the cancellous bone, it was considered convergent. When two subsequent mesh-refinements did not change the result considerably, an adaptive convergence was achieved. The number of elements and nodes used for the models are given in Table 1. The mechanical properties of the bone and materials used in the study are shown in Table 2.

Static loading was used to determine the behaviors of the bone and implant under static conditions. Static loading conditions were applied to each model, simulating the occlusal force with axial (114.6 N), mesiodistal (23.4 N), and lingual (17.1 N) components.[16,17] However, these forces did not affect the implant components significantly (data not shown); therefore, the components of the occlusal force were increased by 50% to establish critical conditions and applied with forces of 171.9, 35.1, and 25.65 N in the axial, mesiodistal, and lingual directions, respectively. Maximum and minimum principal stress values for cortical and trabecular bone and maximum von Mises stress were determined for each implant-abutment system. The stress was evaluated at reference points set in the mesial, distal, buccal, and lingual aspects of the implant body.

The influence of bone resorption and abutment type on the stress profile was determined by comparing models T1, T2, T3, T4, T9, T10, T11, and T12 with the same screw size. The effect of screw diameter and length was evaluated in 3-mm bone resorption scenarios by comparing models T4, T7, T8, T12, T15, and T16; T4, T5, T6, T12, T13, and T14, respectively.
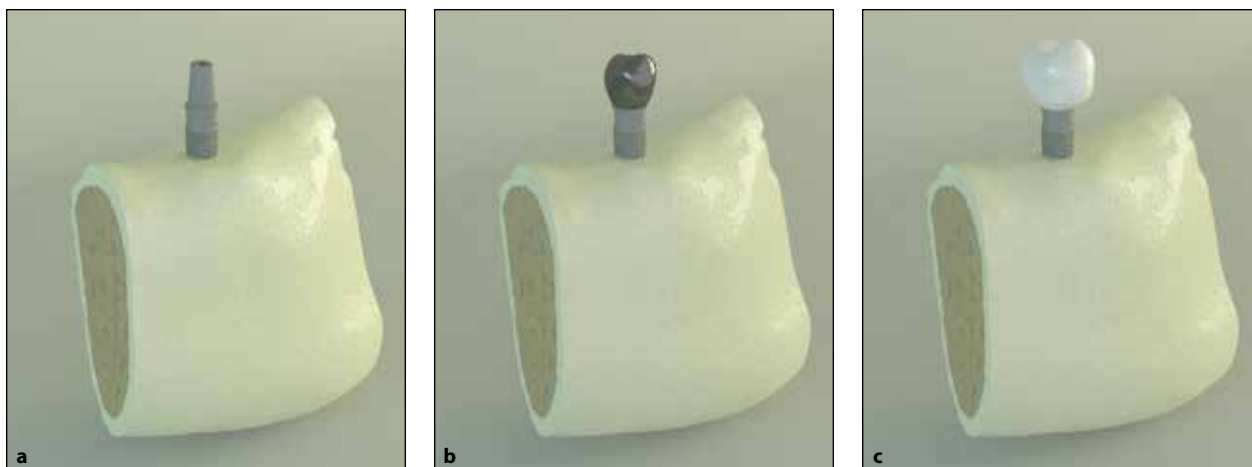
Fig 1 Constructed 3D images of the mandible section. *(a)* Implant inserted with 3-mm resorption at the implant neck with the abutment. *(b)* Implant-supported crown with the framework. *(c)* Final constructed model with the porcelain-fused-to-metal crown. *(d)* Abutment with the screw.
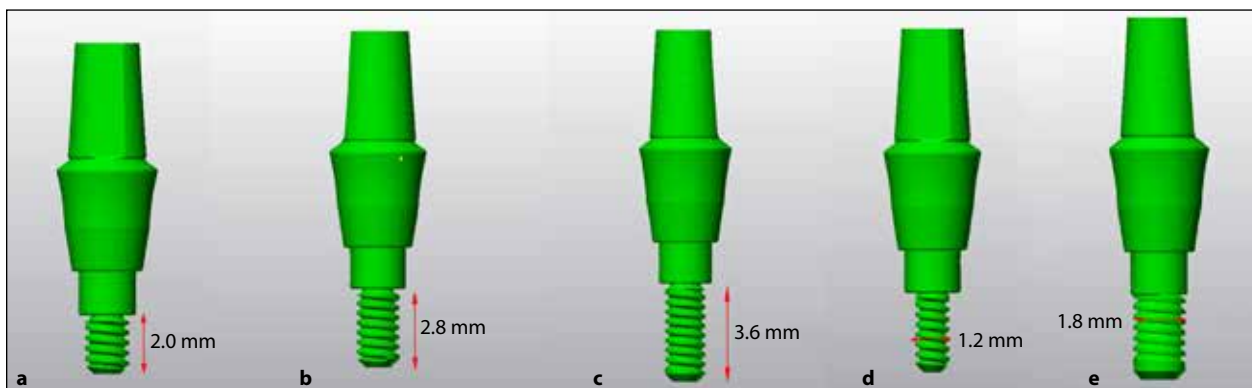

Fig 2 Schematic images of abutment designs, solid abutments *(a)* with 1.5-mm width and 2.0-mm length (model T5); *(b)* with 1.5-mm width and 2.8-mm length (models T1, T2, T3, and T4) *(c)* with 1.5-mm width and 3.6-mm length (model T6); *(d)* with 1.2-mm width and 2.8-mm length (model T7); and *(e)* with 1.8-mm width and 2.8-mm length (model T8).

Table 1 Implant Design Models with Parameters Used in Analysis

Model	Abutment type	Abutment screw length (mm)	Abutment screw diameter (mm)	Marginal bone resorption (mm)	No. of elements	No. of nodes
T1	Solid	2.8	1.5	0	478,955	112,070
T2	Solid	2.8	1.5	1	466,281	109,149
T3	Solid	2.8	1.5	2	455,018	106,565
T4	Solid	2.8	1.5	3	442,303	103,454
T5	Solid	2.0	1.5	3	436,056	106,737
T6	Solid	3.6	1.5	3	472,342	110,347
T7	Solid	2.8	1.2	3	455,714	110,604
T8	Solid	2.8	1.8	3	438,688	102,847
T9	Two-piece	2.8	1.5	0	478,955	112,070
T10	Two-piece	2.8	1.5	1	466,281	109,149
T11	Two-piece	2.8	1.5	2	455,018	106,565
T12	Two-piece	2.8	1.5	3	446,883	109,316
T13	Two-piece	2.0	1.5	3	433,118	101,343
T14	Two-piece	3.6	1.5	3	487,427	119,440
T15	Two-piece	2.8	1.2	3	455,714	110,604
T16	Two-piece	2.8	1.8	3	416,255	104,190

Fig 3 Representative mesh modeling (model T4) with boundary conditions.
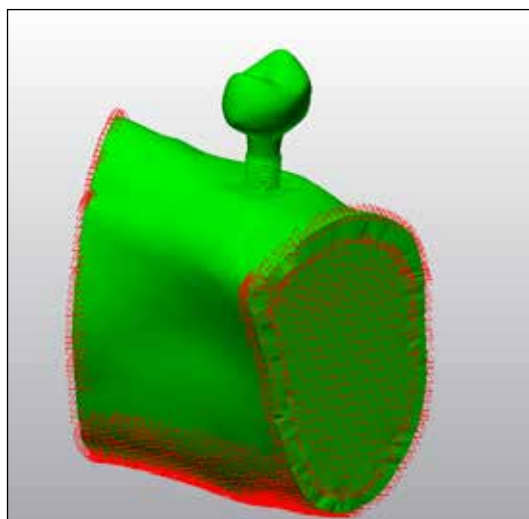
Table 2 Mechanical Properties of Material Used in Models

Material	Elastic modulus (GPa)	Poisson ratio
Feldspathic porcelain	82.8	0.35
Cr-Co metal framework	210	0.35
Ti–6Al–4V	110	0.32
Cortical bone	13.7	0.30
Trabecular bone	1.37	0.30

Three-dimensional modeling was achieved with Rhinoceros 4.0 (McNeel &Assoc); software and models were analyzed with Algor Fempro software (Algor) on a computer (Intel Xeon CPU 3.30 GHz processor, and 14 GB RAM).

RESULTS

In order to determine the effect of bone resorption and abutment type on the surrounding bone and implant system components, T1, T2, T3, and T4 (a solid abutment group with a marginal resorption of 0, 1, 2, and 3 mm, respectively) models were compared with T9, T10, T11, and T12 (a two-piece abutment group with a marginal resorption of 0, 1, 2, and 3 mm, respectively), as these models were designed with abutment screws of the same size, 2.8 mm long and 1.5 mm in diameter. Under static loading conditions, among the mentioned models, the maximum principal stress was obtained at the lingual region of the cortical bone, with the highest values in models T4 (32.3 MPa) and T12 (32.5 MPa), whereas the minimum principal stress values indicated increased stress at the distal aspect of the cortical bone,

Table 3 Von Mises Stress Values in Implants, Abutments, and Screws, and Highest Maximum and Minimum Principal Stress Values in Cortical and Trabecular Bone

	Maximum von Mises (MPa)			Cortical bone (MPa)		Trabecular bone (MPa)	
	Implant	Abutment	Screw	Pmax	Pmin	Pmax	Pmin
T1	157.8	155.9	–	11.1 (L)	–18.6 (D)	2.8 (B)	–0.4 (A)
T2	162.6	160.8	–	22.1 (L)	–18.9 (D)	4.0 (B)	–0.5 (A)
T3	167.2	158.1	–	29.7 (L)	–20.6 (D)	4.6 (B)	–0.9 (A)
T4	171.9	184.5	–	32.3 (L)	–28.2 (D)	5.7 (B)	–2.2 (A)
T5	202.1	216.5	–	38.3 (L)	–30.2 (D)	6.4 (B)	–2.6 (A)
T6	157.3	156.0	–	30.8 (L)	–24.5 (D)	4.8 (B)	–2.0 (A)
T7	198.8	198.6	–	34.9 (L)	–30.1 (D)	5.8 (B)	–2.4 (A)
T8	163.7	172.1	–	30.7 (L)	–25.6 (D)	5.0 (B)	–2.1 (A)
T9	158.9	150.4	74.7	11.5 (L)	–18.5 (D)	3.0 (B)	–0.4 (A)
T10	161.5	158.3	75.9	21.4 (L)	–19.9 (D)	4.1 (B)	–0.5 (A)
T11	167.5	169.0	84.3	29.4 (L)	–20.6 (D)	4.4 (B)	–0.8 (A)
T12	184.3	185.9	86.7	32.5 (L)	–27.7 (D)	5.5 (B)	–2.3 (A)
T13	198.5	221.0	89.4	37.5 (L)	–29.7 (D)	6.4 (B)	–2.6 (A)
T14	152.1	167.7	77.2	31.2 (L)	–24.7 (D)	5.1 (B)	–2.0 (A)
T15	141.7	207.3	67.7	36.9 (L)	–30.6 (D)	6.0 (B)	–2.4 (A)
T16	156.9	176.4	80.7	31.5 (L)	–26.0 (D)	4.8 (B)	–2.0 (A)

L = lingual, D = distal, B = buccal, A = apical. **[AU: Editing okay?]**

with the highest values in models T4 (–28.2 MPa) and T12 (–27.7 MPa), as shown in Table 3 and Fig 4. Maximum principal stress values were found to be lower in trabecular bone, possibly due to its high damping and stress distribution capacity. The highest tensile and compressive stress values were observed at the buccal region of the trabecular bone in all models except for the T4 and T12 models, in which the highest compressive stress was found to be concentrated in the apical vicinity of the peri-implant trabecular bone.

The maximum von Mises equivalent strain values of the solid and two-piece implant-abutment models and stress distribution are described in Table 3 and Fig 5. Among the models with the same screw dimensions but different abutment types (T1, T2, T3, T4, T9, T10, T11, and T12), the highest von Mises stress value was recorded in the T4 model for the solid abutment models, wherein greater stress was observed on the abutment compared with the implant. However, the stress values were lower in models T1, T2, and T3, in which the overall stress values were similar between solid abutments and implants, with a minor increase in the implant body rather than the abutment. In the model with a progressed defect, T4, the maximum stress magnitude increased, and the deformation risk was present at the connection between the first thread of the abutment and the abutment shank (Fig 6). In the two-piece

abutment models, the increased von Mises stress values were seen at the abutment, compared to the implant with the highest values in model T12. The maximum stress concentration in the abutment was similar to that of the solid abutment model, T4.

Various abutment screw sizes were designed with 3-mm bone loss. A minor increase in maximum von Mises stress values was noted for the abutments in two-piece abutment models compared with solid abutment models. Regarding the abutment screw length, in the 1.5-mm-diameter abutment screw groups (T4, T5, T6, T12, T13, and T14), the highest von Mises stress values were recorded in screws of 2.0-mm length for both abutment types, T5 (216.5 MPa) and T13 (221.0 MPa), whereas the lowest values were found in the T6 (156.0 MPa) and T14 (167.7 MPa) models with a screw length of 3.6 mm. In models with 2.8-mm-long screws (T4, T7, T8, T12, T15, and T16), the highest von Mises stress values were noted in models with a screw diameter of 1.2 mm: T7 (198 MPa) with a solid abutment and T15 (207.3) with a two-piece abutment. In two-piece abutments, the maximum von Mises stress was concentrated at the fourth thread of the screw in the models with the same diameter (T12, T13, and T14), as well as in model T16, which had the thickest diameter of 1.8 mm. On the other hand, in the model with a narrow diameter (T15), the critical point shifted more apically (Fig 7).
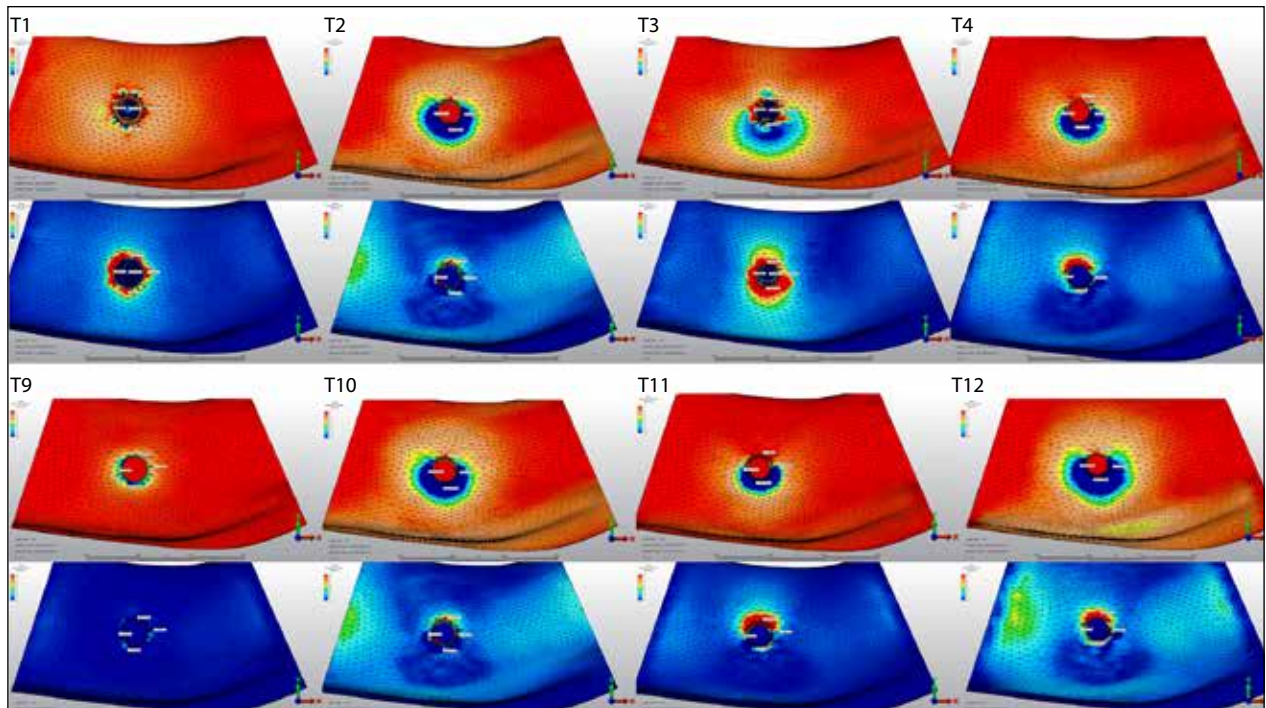
Fig 4 Finite element method images of the minimum *(red)* and maximum *(blue)* principal stress distribution of bone in models with solid abutments (T1, T2, T3, and T4) and two-piece abutments (T9, T10, T11, and T12).
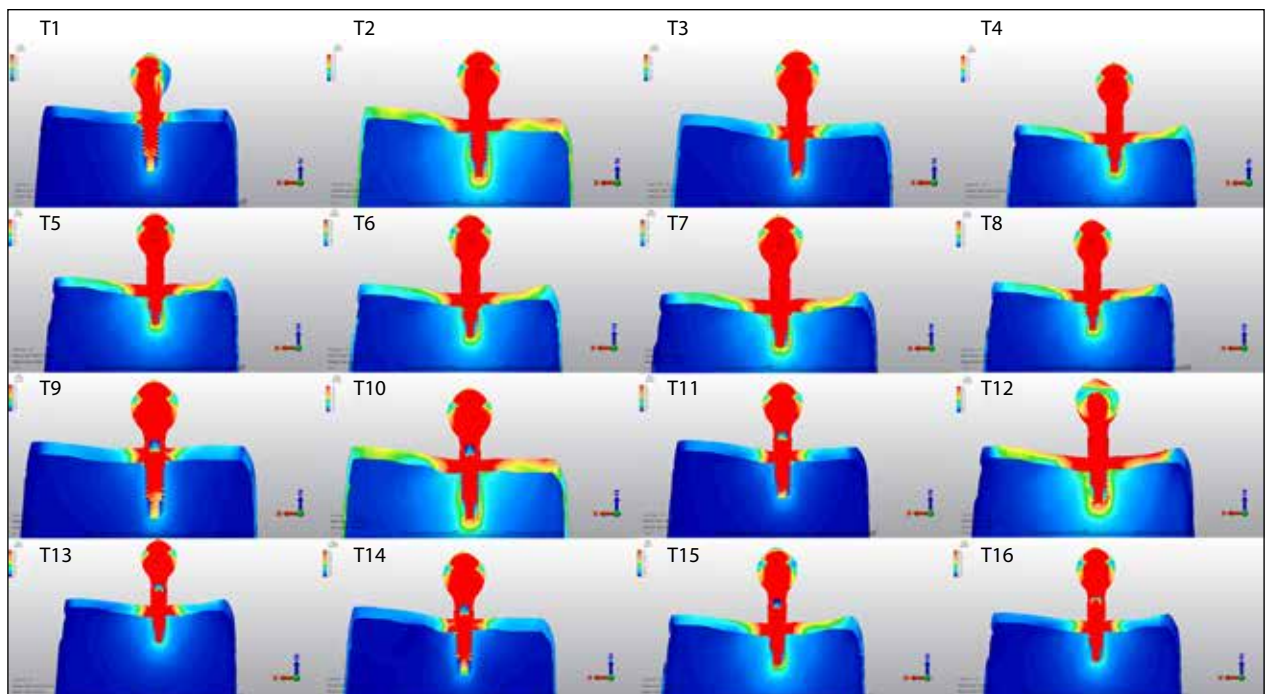

Fig 5 Von Mises stress distribution in the implant-abutment systems modeled in the study.

Fig 6 Von Mises stress values of solid and two-piece abutment models with the same abutment screw size in the 3 mm of marginal bone resorption.
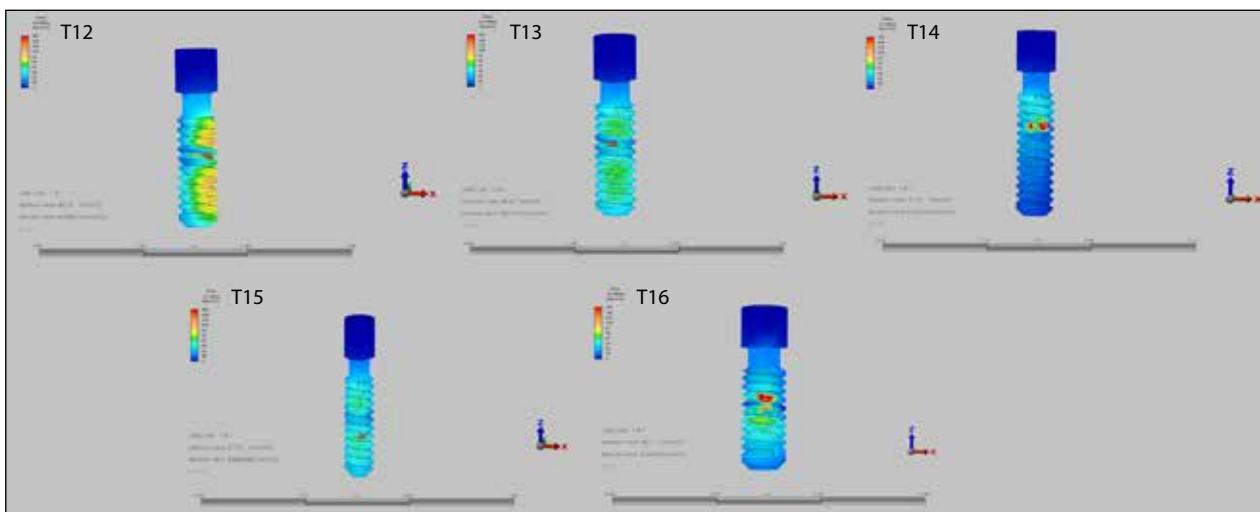

Fig 7 Maximum von Mises stress distribution in abutment screws of two-piece abutments with varying width (T14: 1.5 mm; T15: 1.2 mm; T16: 1.8 mm) and varying length (T12: 2.8 mm; T13: 2.0 mm; T14: 3.6 mm).

DISCUSSION

Dental implants with a reduced diameter can be employed to reduce necessary or complex bone augmentation procedures to prevent morbidity and potential related complications, such as nerve damage, infection, bone fracture, hemorrhage, and wound dehiscence.[18] However, a reduction in implant diameter results in less bone-to-implant contact, compromised mechanical strength, and higher stress levels.[2] Mechanical complications resulting from the reduced diameter suggest an increased risk of fracture in implant components and marginal bone loss.[19] Therefore, the aim of this study was to evaluate the performance of NDIs with different abutment types under static loading and **CL [AU: Please spell out CL]** to determine whether the implant-abutment connection design altered the stress distribution in potential marginal bone loss scenarios.

Along with the bone volume and density of the peri-implant bone, constituents of the implant system can be decisive in the progression of bone loss and implant stability.[20] Various modifications of the implant design are suggested to prevent marginal bone loss, including platform switching, the use of microthreads at the implant neck, and the internal or external connection type of the implant body to abutments.[21,22] Another efficient connection system is the Morse taper connection, which exhibits high mechanical stability and lower bacterial leakage and bone loss, due to its self-locking ability by friction fit.[23] The Morse taper connection presents favorable results in terms of less bacteria, which may be an advantage, as they lower the risk of harboring oral microorganisms.[24] The implant model used in the present study has an internal octagonal design, which can be connected to either solid or two-piece abutments, combining the features of a Morse taper and an internal octagon implant. This connection can be more stable

mechanically and establish an enduring link between the solid abutment and the implant.

The implant-abutment complex is affected by compression forces that establish the integrity of the bone-implant interface, whereas tensile and shear forces on implants result in disruption, which risks the integrity of the implant system.[25] The compressive strength of cortical bone is the greatest and can stand a longitudinal load of 193 MPa.[26] The axial and lateral components of the occlusal load are concentrated near the superior region of the cortical bone: up to 300 N for occlusal loads; hence, cortical bone is not overloaded in various implant systems.[27,28] The highest compressive and tensile stress values demonstrated in the present study were within the resistance range of cortical bone. As the bone defect advanced, tensile stress almost tripled in the lingual aspect of the cortical bone and doubled in the buccal trabecular bone where it was concentrated. However, as the yield point for cortical bone is 69 MPa,[16] the highest maximum stress value range was 11.1 to 38.3 MPa for all models, indicating that the yield strength of the cortical bone was not reached. Although overloading conditions were not demonstrated in the present study, it is possible to suggest that progression of a bone defect strongly affects the tensile stress on cortical and trabecular bone in the implant neck vicinity. Peri-implant bone resorption starting from 3 mm can jeopardize the integrity of the implant, as suggested by the findings of the present study.

Vertical bone loss around an implant may put it at risk, especially in the presence of high lateral loads, in terms of mechanical complications. Under static loading conditions, stress in the implant increased linearly with the resorption depth. In the present study, in models with no resorption and vertical bone resorption of 1 and 2 mm, similar stress distribution with no risk of failure was observed, regardless of abutment type. The relatively homogenous stress distribution demonstrated in the models mainly relates to the assumption of uniform vertical bone resorption around the peri-implant bone used in this study. However, a vertical bone resorption of 3 mm significantly increased the stress building on the implant-abutment complex, showing anisotropic behavior in both abutment types. Similar studies on gradual bone loss revealed higher stress values in bone resorption, especially in pure vertical bone resorption models, in which stress on cortical and cancellous bone increases with resorption depth. In implants with a vertical peri-implant bone loss of 3.0 mm, higher maximum and minimum principal stresses were recorded than in those with a loss of 1.3 mm, and the maximum equivalent stresses ranged between 130 and 208 MPa,[28] in parallel with the results of the present study. According to Wang et al,[29] the maximum deformation forces are lower for implants connected to titanium-alloy abutments, with a marginal bone resorption of 3 mm compared with 1.5 mm, as they can resist forces up to 540.6 N and 1,070.9 N, respectively. **[AU: Editing okay?]** Although the induced stresses in the present study were below the deformation of the implant, progression in bone resorption may potentially lead to mechanical failure. However, the morphology of bone defects around dental implants needs to be taken into consideration. The modified designs of bone defects can lead to different tendencies of stress behavior around peri-implant tissues. Sánchez-Pérez et al[30] suggested that in conical defects, the oblique cortical bone lowers the bending tendency of the cortical bone, and the compressive stress accumulating around the implant neck by decomposing axial forces, whereas axial loads contribute to the bending of the bone in vertical bone defects. The presence of the cortical bone layer is shown to improve the biomechanical performance in progressive marginal bone loss.[31] Therefore, in contrast to a vertical bone defect, as simulated in the present study, the presence of a conical defect within biologic limits in functioning implants does not necessarily raise the risk of implant failure.

The resultant stress values under loading are important, along with the coordinates of stress distribution, as these factors can be altered with modified implant and abutment designs to prevent mechanical failures. Abutment type is found to have significant influence on the stress distribution in bone due to different load transfer mechanisms and different contact areas between the implant and abutment.[8,32] Under dynamic and static loading conditions, maximum stress is concentrated at the connection between the abutment shank and the first thread of the abutment screw.[33] The present study indicates that the maximum stress location of the abutment is at the interface between the first thread of the screw and the abutment shank in both systems, when the resorption depth is highest, which might indicate a potential mechanical failure at this point. The reduced thickness of the implant neck surrounding the screwless abutment is demonstrated to be the weak region of the system, which is indicated by the mechanical failure presenting as an implant fracture instead of abutment bending or fractures.[34] In particular, in NDIs, which are more susceptible to mechanical failure than wide implants, the risk of implant fracture must be considered in both types of abutment designs, as there was a minor difference in the von Mises stress values in the implant body. **[AU: Editing okay?]** However, when the screw diameter was reduced to 1.2 mm, the maximum von Mises stress was increased to 198.8 MPa in the implant body for the solid abutment system compared with 141.7 MPa in the two-piece abutment system, which might render the solid abutments unfavorable in such critical scenarios. Cehreli et al[19] showed that in

reduced-diameter Morse-taper implants connected to solid abutments, subjection to high bending jeopardizes the integrity of the implant, especially at the implant neck region and at the stem of the abutment screw. Considering the lower stress levels in the implant body with two-piece abutments, this preference over solid abutments might be beneficial, as the breaking of the abutment screw might be a warning that could prevent more damage, including the bending of the implant neck, which is difficult to compensate for in clinical situations. However, due to the difficulty of retrieving fractured screws, especially in NDIs with limited space to maneuver, the risk of potential damage to the internal implant design and material should not be neglected, as this might necessitate the removal of the implant body as unfit for new restorations.

Current studies on screw design favor long screws with an increased thread number due to a decrease in the risk of screw loosening, regardless of the implant-abutment connection.[14] Under static forces, longer abutment screws are more resistant to fracture than short abutment screws.[35] The maximum stress values within the tested screw dimensions did not exceed yield strength, although the highest stress values were noted in the short abutment screw, in accordance with previous studies.[14] However, the present study demonstrates lower stress distribution in implant models with longer abutment screws under static loading; the location of the critical point in the screw did not change according to the screw length. **[AU: Editing okay?]** Cehreli et al[19] indicated the possibility of the undesired bending of the implant body due to the reduced metal thickness in narrow implants around the abutment screw, suggesting the use of an abutment screw with a smaller diameter to prevent such potential failures. In the present study, the reduced screw diameter resulted in lower stress in the implant body and the abutment screw by increasing the stress on the abutment. The diameter of the screw was more effective in determining the critical point of the screw, which was found to be in the same location for screws of 2.0 to 3.6 mm in length and 1.5 to 1.8 mm in width. When the screw diameter was reduced to 1.2 mm, the maximum von Mises stress focus shifted toward more apical threads. As the critical point of the screw surpasses the susceptible implant neck region, the implant body is not jeopardized.

With complex systems in which many variables need to be considered, the finite element method enables the manipulation of each parameter to determine its single effect on the investigated model. **[AU: Editing okay?]** In this regard, it is possible to observe each biomechanical response that is difficult to evaluate in clinical conditions with many repetitions of the tests as desired. This eliminates the use of animal studies and enables determination of the applications of varying

materials in possible scenarios.[36] However, there are limitations to finite element studies related to simplifications and assumptions. Since the bone-implant interface in the systems simulated by the finite element method is defined as perfectly bonded, a major limitation concerning the finite element method is the lack of proper modeling of the overload state and the bone-implant interface–induced load transduction. Hence, these transmission losses cannot be followed in studies using the finite element method. Furthermore, all materials used in the present study are assumed to be isotropic, although bone tissue behavior is subject to change depending on the direction of the applied load. Although this method does not represent in vivo systems entirely as numerical values, it enables the analysis of stress coordinates and design elements that occur in the implant. This is the most likely limitation of the present study, like other finite element studies, as these assumptions do not entirely reflect clinical situations.

CONCLUSIONS

Marginal bone loss around the neck of the implant increases stress in the implant-abutment complex, which could be critical in terms of the mechanical stability of narrow implants. Within the limitations of this study, two-piece and solid abutments display similar stress behavior in the surrounding bone, although the use of two-piece abutments decreases strain in the implant body. The bone loss levels dominate the mechanical response under static loading, although design variables such as abutment screw length and diameter are also effective in resultant stress values.

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