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The Effect of Implant–Abutment Contact Area on the Stress Generation of Bone-Level and Tissue-Level Implants

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Abstract: Aims: This in silico study aimed to investigate the effect of implant–abutment contact surfaces on the stress generation of Morse taper implants under oblique loading. Materials and methods: Three-dimensional finite element models of Bone-Level and Tissue-Level implants were simulated with Standard and Partial contacts between the abutment and implant. The dimensional parameters followed the ISO 14801 guidelines, and an oblique load of 300 N was applied to the implants. The von Mises stress was acquired. Results: The Tissue-Level design showed a significant difference in the stress level when the connection with the implant, abutment, and screw was Partial. For the implant fixture, abutment, and screw, the Tissue-Level design showed 13% more stress in the implant, abutment, and screw when the connection was Partial. The Bone-Level design did not affect the connection and showed an overall 42% lower stress than the Tissue-Level design for the implant fixture. However, in the screw, there was a difference between the Bone-Level implants with a Standard and Partial connection. In contrast, for the Tissue-Level implant, this difference was less evident with higher stress peaks in the entire set. Conclusion: To achieve optimal outcomes, it is highly recommended to use original abutments, as they provide a more precise fit. The stress peaks were notably lower in Bone-Level implants compared to Tissue-Level implants. Furthermore, an implant–abutment connection with more contacting areas significantly reduced stress concentration, especially in Tissue-Level implant designs. By choosing well-fitting abutments, one can ensure more stable and durable implant performance with less stress.

Keywords: dental implants; finite-element analysis; prosthetic dentistry; implant–abutment design

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1. Introduction

Modern implants in dentistry, with proper connection, have a success rate of 97.0% throughout a 10-year interval [1]. This long-term success of dental implants in bone is achieved when osseointegration occurs [2]. This can be expected when following the correct

indications for dental implants [3]. Because of the high success rate, there is an increased demand for dental implants in daily practice. It is expected that the prevalence of dental implants will increase from 5.7% in 2015 to possibly 23% in 2026 [4].

Despite the high success rate, osseointegration can also be impeded by inflammation and infection surrounding the implant in a process named peri-implantitis [5]. This inflammation can lead to the loss of supporting bone and is successive and progressive [6]. Peri-implantitis can be caused by both biological and mechanical factors. Mechanical factors, such as a supra structure with partial marginal fit and overloads, can be risk factors for developing peri-implantitis [7]. They may be associated with a limitation in proper oral hygiene, which assists in developing peri-implantitis [8].

Due to several implant companies, there are a lot of designs and approaches to consider in the contact between abutments and implants. Previous literature showed that the implant–abutment joint has an impact on the mucosal integration around implants and affects the risk of developing peri-implantitis in the future [9], due to having a significant role in technical and biological complications [10].

Over time, different connection designs have been developed, each possessing distinct mechanical, biological, and esthetic attributes aimed at preserving the stability of the implant–prosthesis interface. However, fundamentally, two major groups of geometries exist: internal and external connections [11]. External connections feature an external hexagon on the implant platform. Internal connections can be commonly categorized into internal hexagon, internal octagon, and Morse taper connections, with the latter currently experiencing widespread utilization due to optimal sealing and better load distribution [12].

Despite the advantages of Morse taper implants, there is a rising occurrence of complications, including mechanical issues and peri-implantitis. Similarly to other mechanical structures, implants are susceptible to fracture with prolonged usage, and an improper partial implant–abutment joint is one of the many factors that can facilitate its occurrence [13]. Remarkably, this concern is rarely acknowledged or discussed in the literature [14]. This can lead to problems, like crown fractures, peri-implantitis, abutment–screw loosening, implant fracture, and other outcomes that could be avoided if not for the lack of data about the abutment joint in terms of stress [15].

One of the tools to study dental implants is finite-element analysis (FEA). Finite-element analysis stands as a research instrument for mechanical analyses within biological research. This methodology calculates mechanical responses with the aid of modeling intricate structures and their characteristics [16]. This method is crucial for predicting the performance of both implant materials and bone structures [17]. It is possible to investigate implant-supported restorations, facilitating the assessment of stress distribution and the identification of critical geometries that could lead to clinical failures [18].

Therefore, the purpose of this in silico study was to investigate the effect of implant–abutment contact surfaces on the stress generation of Morse taper implants under oblique loading. The hypothesis of this study was that there is no difference between Bone-Level and Tissue-Level implants in terms of stress distribution, regardless of the number of contact surfaces between implant and abutment.

2. Materials and Methods

For this study, Bone-Level (BL Ø 3.3 mm, NC, SLActive® 10 mm, Roxolid®, Guided) and Tissue-Level (SP Ø 3.3 mm, RN, SLActive® 10 mm, Roxolid®, Guided) implants were selected (Straumann Dental Implant System, Waldenburg, Switzerland). Figure 1 shows the selected implants for this study.

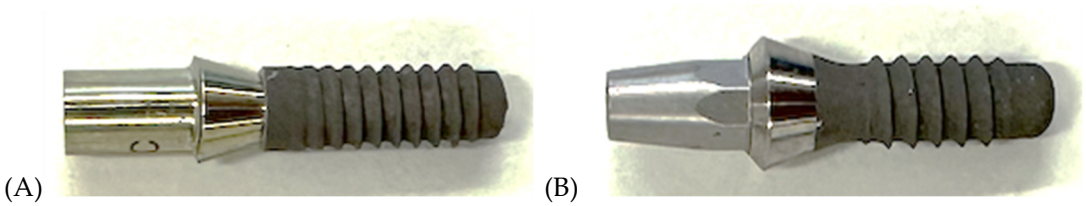


Figure 1. Selected implants and abutments for this study. (**A**) Bone-Level implant. (**B**) Tissue-Level implant.

After the abutment installation (for BL: Straumann® NC [Narrow CrossFit®] Anatomic Abutment with 2 mm gingiva height (GH), abutment height of 6.0 mm, and abutment diameter of 4.0 mm; for TL: Straumann® RN synOcta® Cementable Abutment, designed for Regular-Neck (RN) Tissue-Level implants with abutment height of 5.5 mm and abutment diameter of 4.8 mm) following the manufacturer's recommendation, each specimen was embedded in self-curing red methyl-methacrylate (Acryfix, Struers GmbH, Willich, Germany) and cross-sectioned along the longitudinal axis using a precision cutting machine under constant water cooling.

The internal configuration was visually inspected and photographed under reflected-light scanning electron microscopy (XL20, Philips, Eindhoven, The Netherlands) at a 10× magnification, and each photomicrograph was joined to create the background for the modeling step. Figure 2 shows a microscopical photo of the implant–abutment interface for the respective Bone-Level implant and Tissue-Level implant.

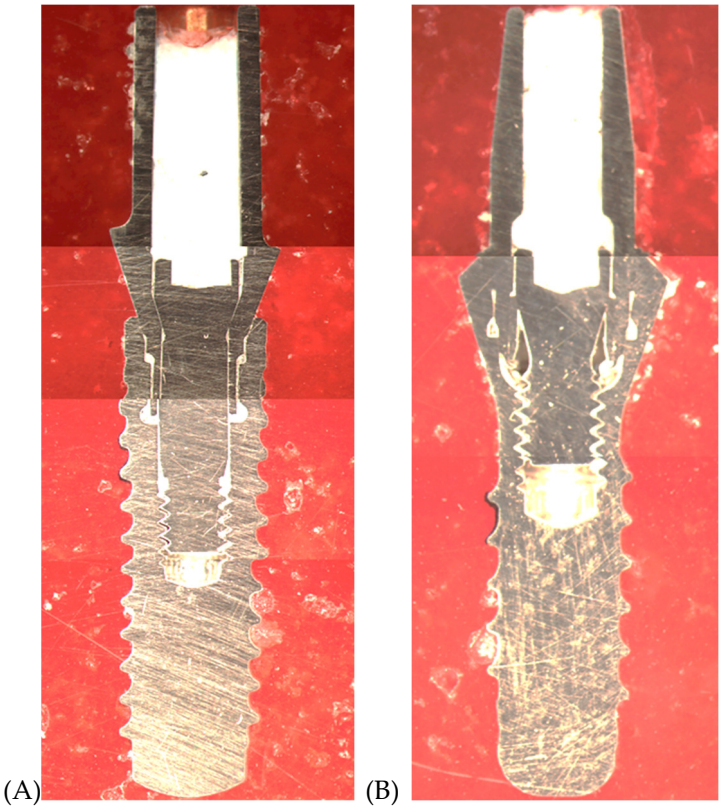
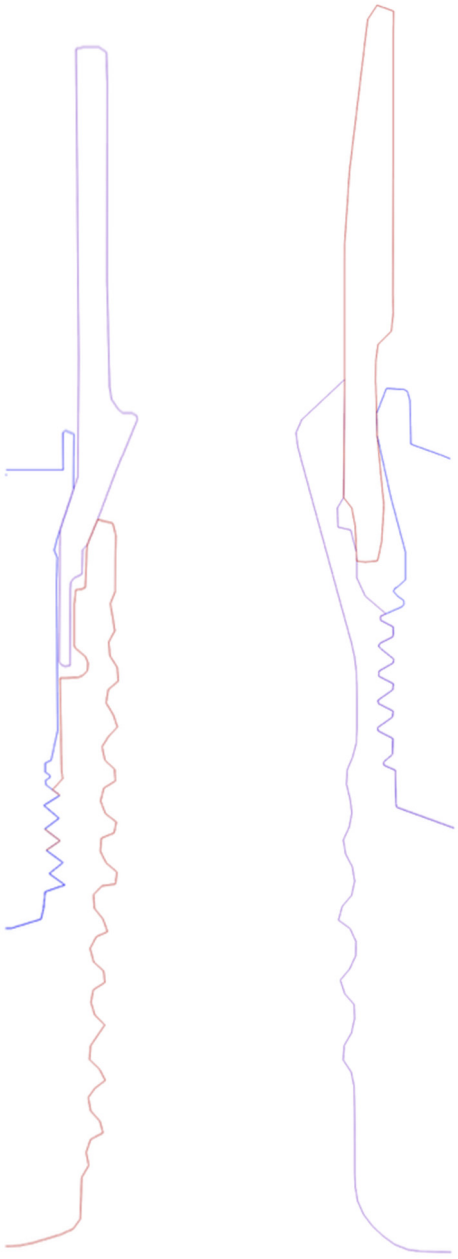


Figure 2. Microscopical photo of implant–abutment interface. (**A**) Bone-Level implant. (**B**) Tissue-Level implant.

To generate the 3D file in Standard for the Exchange of Product Data (STEP), the photomicrography figures were imported to computer-aided design software (Rhinoceros version 5.0 SR8, McNeel North America, Seattle, WA, USA) as background BMP files. Based

on the visible connection, the profiles of each implant consisting of a screw, abutment, and implant are connected by lines from the photomicrography. They show the three parts that are touching for the Bone-Level implant and the Tissue-Level implant in three different colors (Figure 3).



(A) (B)

Figure 3. Profile lines were generated based on photomicroscope background. (**A**) Bone-Level implant. (**B**) Tissue-Level implant.

In the sequence, an axis-symmetric 3D solid structure was creatred according to each model. Figure 4 shows the geometries finished in CAD software (Rhinoceros software version 5.0 SR14 (McNeel North America, Seattle, WA, USA)) for the Bone-Level implant and the Tissue-Level implant.

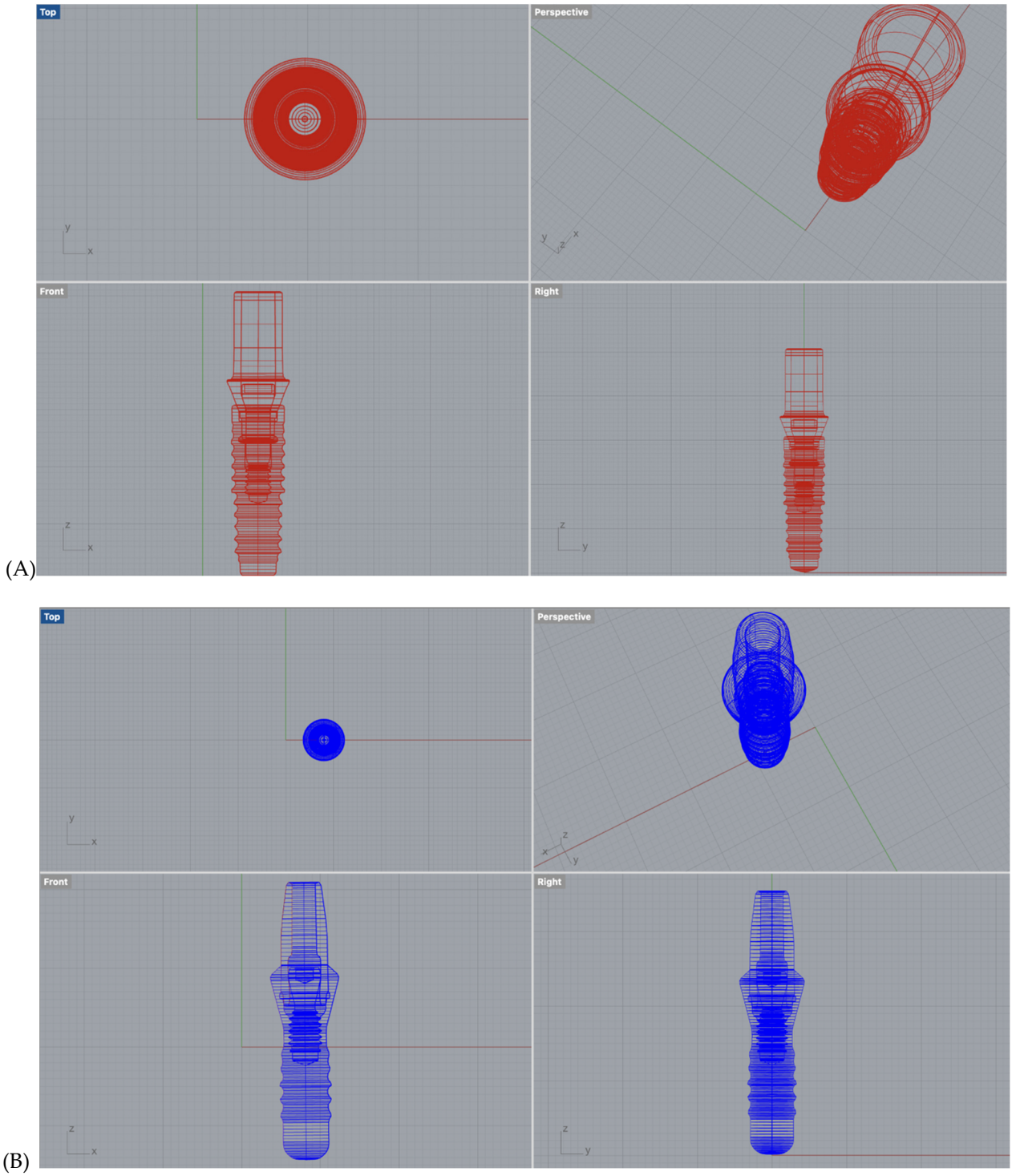


Figure 4. Geometries finished in CAD software. (**A**) Bone-Level implant. (**B**) Tissue-Level implant.

In sequence, the 3D file formatted in Standard for the Exchange of Product Data (STEP) was finished with computer-aided design software (Rhinoceros version 5.0 SR8, McNeel Noth America, Seattle, WA, USA). The model consisted of the following components: an implant fixture (4.1 × 11 mm), prosthetic screw, and abutment. The dimensional parameters and geometrical relationships were based on ISO 14801 guidelines [19] for the loading test in endosseous dental implants.

Following the completion of the modeling phase, the three-dimensional model was subsequently imported into analysis software (ANSYS 23.0, ANSYS Inc., Houston, TX,

USA). To perform the static analysis, the titanium alloy was designated as homogeneous, linear, and isotropic. The values for Young's modulus and Poisson's ratio were 110 GPa and 0.3, respectively. To replicate the absence of joint defects, all contact interactions were assumed to be rough. Rough contact is typically modeled when two surfaces are in physical contact, and there is friction or resistance to sliding between them. It simulates a condition where the surfaces are not perfectly smooth but instead have microscopic asperities (roughness) that prevent relative motion.

The formation of four distinct groups was conditional on the combination of implant design and contacting surfaces at the abutment–implant joint. In this case, the Standard connection was based on the contacting surfaces observed in the cross-section plane while the Partial connection was simulated considering only cervical contact at the implant platform level (without contact at the connection level). To simulate the difference between the Standard and Partial abutment–implant connections, we modified the contact definitions in the computer-aided engineering (CAE) software rather than altering the CAD geometry. The Partial configuration represented a "third-party" abutment with similar overall geometry but insufficient contact with the implant walls. To replicate this condition, we reduced the number of contacting faces in the contact definition, ensuring that the abutment did not fully engage with the implant walls as observed in microscopy. This approach represented the mechanical behavior of an improperly fitting abutment without artificially changing its dimensions. For the Bone-Level implant, the Partial contact in the cervical portion, calculated using the Area command in CAD, was 31 square mm, while the Standard connection covered 73 square mm, representing 42.47% of the total area. In contrast, at the Tissue Level, the cervical area was 64 square mm, and the total area was 74 square mm, representing 86.49% of the total area.

An equal number of tangent faces between solids was established to facilitate analysis convergence. An initial mesh consisting of tetrahedral elements was automatically generated, and a thorough check ensured the absence of any mesh marked as obsolete by the software before proceeding to the final mesh refinement (Figure 5). For meshing, the convergence test was based on 150,320 nodes and 72,430 elements. The mesh used tetrahedral elements with 10 nodes (Tet-10). The mesh quality parameters were as follows: an element size of 0.4 mm, quality of 0.78 ± 0.12, aspect ratio of 1.56 ± 0.45, average maximum corner angle of 93.50°, and average skewness of 0.25 ± 0.09. The inflation option was applied to ensure a smooth transition between the geometries. Rigid body behavior was standardized with dimensionally reduced settings to optimize the model's accuracy. Mesh control was determined using a 10% convergence test to minimize its impact on the results of the mathematical calculation [18].

The load was based on the trigonometric identity to calculate the resulting force. In this study, the resultant force was set at 300 N (Figure 6). The oblique loading and exposed threads were used to simulate the ISO14801 worst-case scenario, positioning the implant's long axis at a 30° angle in the loading direction. The current boundary conditions considered a fixation at the neck of the implant, where it interfaced with the cortical bone, the primary region that bears the majority of the load, especially during axial loading scenarios.

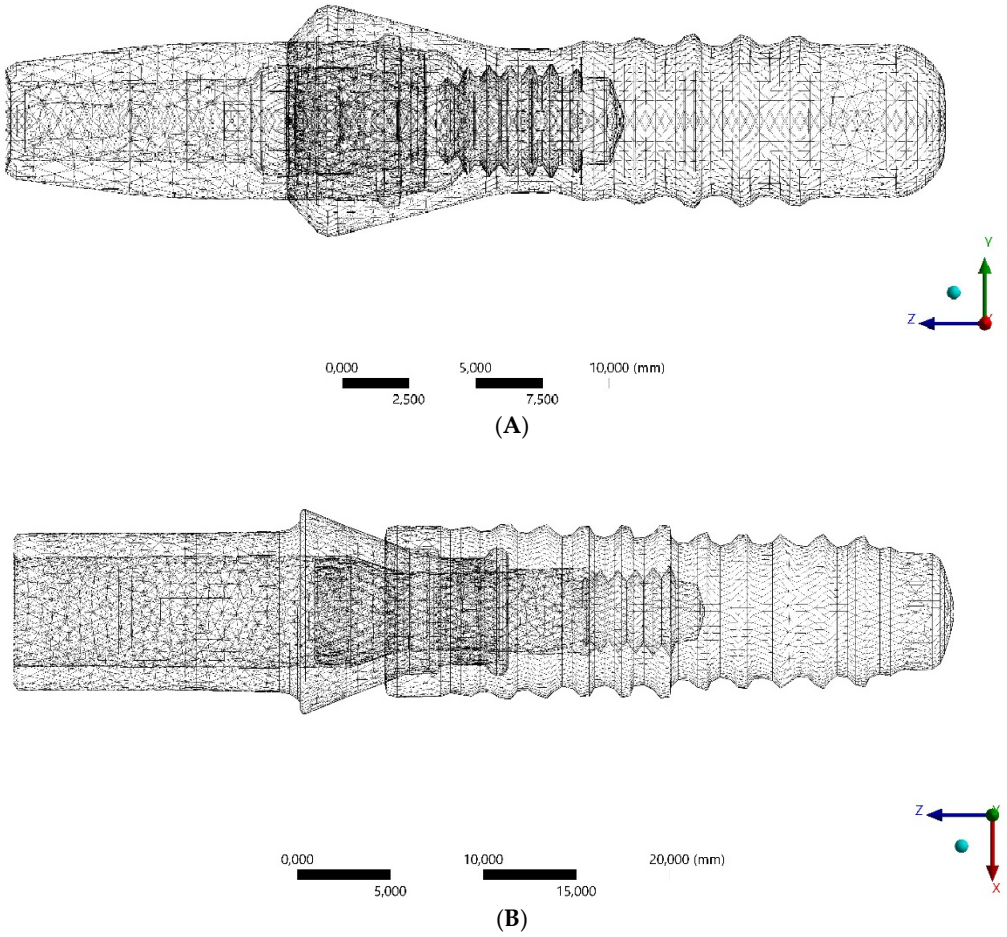


Figure 5. (**A**) Mesh generated for Tissue-Level model and (**B**) Bone-Level model.

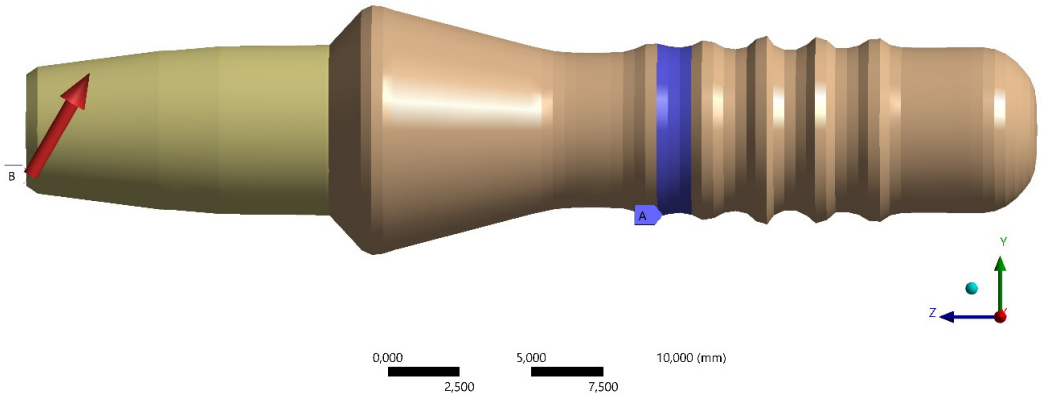


Figure 6. Boundary conditions simulated in the present study. The blue area shows the fixation and the red arrow shows the loading.

Two models, the Tissue-Level (TL) and Bone-Level (BL) ones, were created, each featuring scenarios with both Partial and Standard contacts between the abutment and implant (Figure 7).

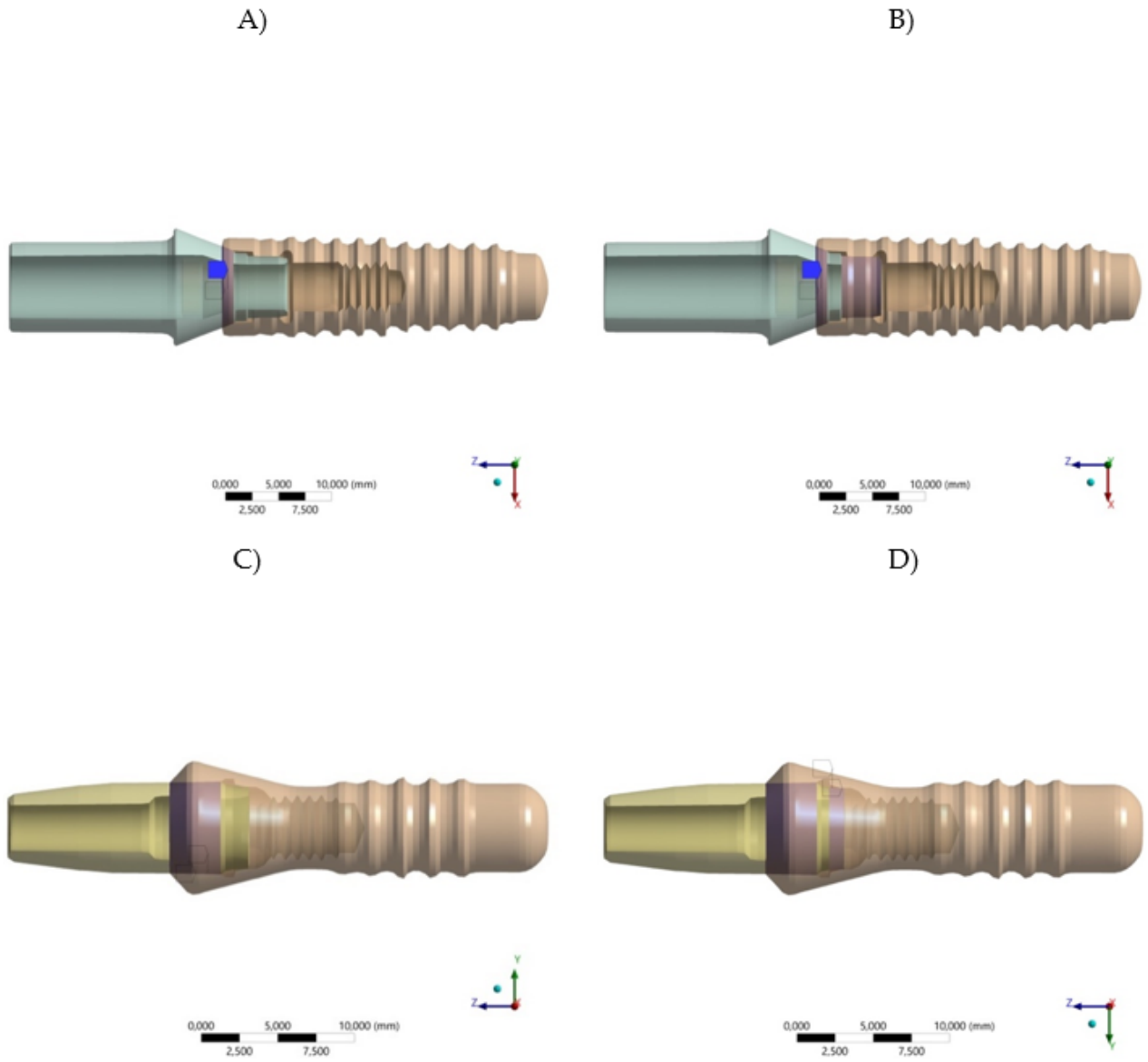


Figure 7. BL design with (**A**) Partial abutment joint and (**B**) Standard abutment joint. TL design with (**C**) Partial abutment joint and (**D**) Standard abutment joint.

Statistical Analysis

Following the mechanical static structural analysis, von Misses stress peaks were exported onto spreadsheets for each structure (implant, screw, and abutment), corresponding to the element number from the mesh. The highest values for each structure were selected, according to the maximum stress peak as reported by the mechanical APDL software. Descriptive statistics (mean and standard deviation) were employed for data analysis, alongside one-way ANOVA for each factor studied, followed by Tukey's test to assess differences between groups. All tests were considered significant at a 10% level due to the mesh convergence test. This consistency ensured that the statistical analysis aligned with the precision and variability observed in the stress convergence results.

3. Results

The summarized data can be seen in Tables 1 and 2. The stress peaks ranged from 364 MPa in the screw of the Bone-Level implant with a Standard connection to 681 MPa in the abutment, implant, and screw of the Tissue-Level implant with a Partial connection. The highest average was observed in the implant of the Tissue-Level implant with a Partial

connection, while the lowest average was observed in the screw of the Bone-Level implant with a Standard connection. In general, the Bone-Level implants had lower average stress peaks than the Tissue-Level implants. The standard deviation was not bigger than 9 MPa, showing that the stress was homogenous.

Table 1. Descriptive statistics from stress peaks according to implants' components.

Connection	Design	Region	Average	Std. Dev.	Min.	Max.
Standard	Bone Level	abutment	389.96	1.61	388	393
	Bone Level	implant	387.34	2.55	383	393
	Bone Level	screw	376.26	8.52	364	393
	Tissue Level	abutment	584.39	1.82	581	588
	Tissue Level	implant	584.91	1.67	582	588
	Tissue Level	screw	583.08	2.72	578	588
Partial	Bone Level	abutment	389.81	1.70	387	393
	Bone Level	implant	387.99	2.25	384	393
	Bone Level	screw	382.05	6.43	371	393
	Tissue Level	abutment	674.02	3.16	669	681
	Tissue Level	implant	676.74	1.57	673	681
	Tissue Level	screw	673.06	3.96	667	681

Considering the factors of the design, connection, and region, 3-way ANOVA was performed. All the factors were significant, including their interaction, which was also a factor.

Table 2. Analysis of variance.

Analysis of Variance Source	DF	F-Value	*p*-Value
connection	1	45,032.50	0
region	2	312.28	0
design	1	1,249,304.51	0
connection*region	2	17.38	0
connection*design	1	41,047.01	0
region*design	2	163.38	0
connection*region*design	2	23.99	0

To observe significant differences between groups, Tukey's test was performed. Observing the implant fixture, the Tissue-Level design showed a significant difference in the stress level when the connection with the implant, abutment, and screw was Partial.

The most affected structure for the Tissue-Level design was the implant, and for the Bone-Level design, it was the abutment structure.

For the implant, abutment, and screw, the Tissue-Level design showed a significant difference in terms of stress. It showed 13% more stress in the implant, abutment, and screw when the connection was Partial. However, for the Bone-Level design, not only was the stress 42% lower than that for the Tissue-Level design, but this model was also not affected by the contacting area at the connection.

Figure 8 summarizes the stress maps for each group. In the Bone-Level implant, the stress was higher in the abutment neck near the screw head, which is displayed by the color red. The Bone-Level implant has colder colors in the abutment and in the implant, which illustrate lower stress magnitudes.

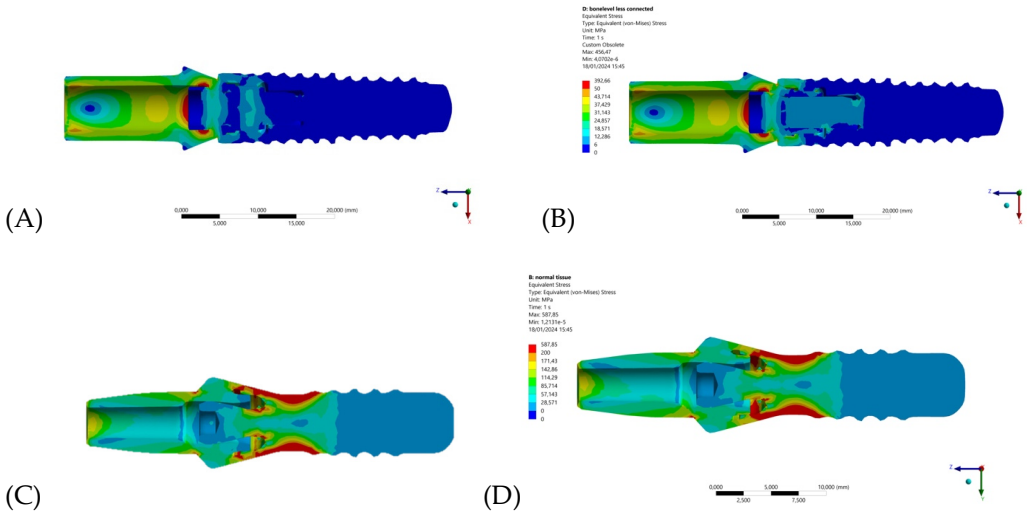


Figure 8. von Mises stress maps for BL design with (**A**) Standard abutment joint and (**B**) Partial abutment joint. TL design with (**C**) Standard abutment joint and (**D**) Partial abutment joint.

There are also more variations in the colors for the Bone-Level implant than for the Tissue-Level implant. However, the Tissue-Level implant shows a high stress magnitude in the implant neck and a lower magnitude near the screw head compared to the Bone-Level implant. In addition, the Tissue-Level implant shows a larger surface with the color red, which shows the highest stress peaks. The difference between groups with similar implant designs is not qualitatively visible. However, at the screw, it is possible to observe that there is a difference for the Bone-Level implants with Standard and Partial connections. In contrast, for the Tissue-Level implants, this difference is less evident. This is also in agreement with Table 3, which shows no significant difference between Standard and Partial connections for the Bone-Level implant.

Table 3. Tukey pairwise comparisons according to the connection, implant design, and evaluated region.

Tukey Pairwise Comparisons Using the Tukey Method and 95% Confidence:									
Connection*Region*Design	**Mean ***	**Group**							
Partial*implant*Tissue Level	676.74	A							
Partial*abutment*Tissue Level	674.02		B						
Partial*screw*Tissue Level	673.06		B						
Standard*implant*Tissue Level	584.91			C					
Standard*abutment*Tissue Level	584.39			C	D				
Standard*screw*Tissue Level	583.08				D				
Standard*abutment*Bone Level	389.96					E			
Partial*abutment*Bone Level	389.81					E			
Partial*implant*Bone Level	387.99						F		
Standard*implant*Bone Level	387.34						F		
Partial*screw*Bone Level	382.05							G	
Standard*screw*Bone Level	376.26								H

* Means that do not share a letter are significantly different.

The Tissue-Level implants with Standard and Partial connections show, according to Table 3, no significant difference between the abutment and the screw, whereas the map shows fewer stress peaks in the screw.

4. Discussion

Earlier research reported that Morse tapers can be susceptible to the occurrence of complications. This is justified because implants are vulnerable to fracture with extended usage, and an insufficient implant–abutment joint is one of the many reasons that can facilitate this [12]. To understand fracture mechanics and stress magnitude, several studies have applied finite element analysis (FEA) to evaluate the biomechanical response. Regardless of the limitations of this method, FEA is fundamental to providing information concerning mechanical failures and stress–strain distribution.

A former finite element study investigated the effect of different prostheses with misfits on the distribution of stress, implant structures, and adjacent bone [20]. They found that occlusal loads near the gap increased the stress in implant structures and peri-implant bone tissue. This corroborates the present study, which showed higher stress levels when the connection was Partial. However, the present study did not evaluate adjacent bone. Additionally, their model had a different implant design, which was from the company Nobel Biocare. It is important to mention that they also used other boundary conditions, which were different in magnitude and direction from the oblique load simulated in this research.

The results of this study corroborate another previous article which showed that Tissue-Level implants displayed more stress in the surrounding bone than Bone-Level implants [21]. This can be explained because Tissue-Level implants have a shorter screw nearby the cervical area and a thinner axial wall than the Bone-Level implant, which can lead to higher stress levels in the surrounding bone. This previous article [21] also showed that Tissue-Level implants were associated with more bone remodeling in the cortical bone. Complementary to [20]'s findings, this research investigated stress in the implants with different implant–abutment contact surfaces, reinforcing the importance of a good implant–abutment connection.

A previous study aimed to predict time-dependent bone remodeling around Tissue-Level and Bone-Level implants used in patients with partial bone width. They found that the Tissue-Level implants had higher stresses than the Bone-Level implants [22]. According to the authors, the Bone-Level implants had lower displacement than the Tissue-Level implants, showing that the force distribution was better in the second one [22]. The present study showed a similar biomechanical response when comparing both implant designs. However, it is important to mention that the present study did not measure the stress in the bone tissue. Instead, the stress was calculated between the components. Based on these previous reports [21,22] and the results presented in this study, we can expect more mechanical failures for Tissue-Level implants in comparison to Bone-Level implants.

A prior finite-element study analyzed the effect of discrepancy in implant-level fixed partial dentures and marginal bone support on the generation of implant fractures [23]. The misfit groups had higher coronal body stress levels than the no-misfit groups, which supports the present study's findings, which showed higher stress when the connection was partial, especially with the Tissue-Level implant. In addition, the authors reported that the effective stress was higher in the conical connection in the group with bone loss. It is essential to note that the prior study used an Astra Tech Implant System Osseospeed TX from the company Dentsply Sirona. They considered a misfit when there was a certain distance between structures, while the present study considered the connection partial when the internal implant structures were not ideally touching. It is also important to mention that the present study did not measure marginal bone support and did not use a framework on top of the abutment. In summary, the importance of the internal connection besides the external fit for reducing stress levels is evident.

The results of the present study indicate that the stress distribution in Bone-Level and Tissue-Level implants is strongly influenced by the implant–abutment contact area. However, direct comparisons with previous finite-element analyses (FEAs) should be made cautiously due to differences in implant design, boundary conditions, and modeling approaches. For instance, a prior FEA study that simulated a 200 N occlusal load in various directions reported higher stress levels in Bone-Level implants compared to Tissue-Level implants [24]. However, several key differences exist between that study and the present one. First, their implant–abutment connection was designed using an idealized CAD model, whereas our study utilized real measurements obtained through microscopy. This distinction is critical, as real-world connections often introduce microgaps and irregularities that affect stress distribution. Additionally, their loading setup differed from our oblique force application, which may account for the discrepancy in the results.

Similarly, another FEA study [25] reported that Bone-Level implants exhibited higher stress peaks than Tissue-Level implants. However, their analysis focused on Osstem implants, which differ significantly from the Straumann implants used in our study. Notably, the Osstem Tissue-Level implant features an abutment that partially rests on the implant platform, whereas the Straumann design does not. Additionally, the Bone-Level implant in their study had an abutment wider than the implant diameter, altering the load transmission characteristics. Furthermore, differences in the prosthetic screw design across implant brands may also influence force distribution and stress concentration.

Given these variations in implant geometry, connection design, and modeling methodologies, it is important to interpret comparisons with caution. Our findings emphasize the importance of implant–abutment fit in reducing stress levels, particularly in Tissue-Level implants. Future research should explore how these factors influence long-term mechanical stability, incorporating experimental validation to complement FEA findings.

In addition to computational analysis, in vitro studies are also important. A previous in vitro study evaluated the internal fit and cyclic fatigue life of three implant–abutment configurations, comparing original and alternative replicated cast-to-gold abutments [26]. They found that original abutments transferred occlusal loads more homogeneously because of better fitting with internal components. The present study corroborates this, showing that a Standard connection leads to 13% lower stress than a Partially connected abutment, especially when Tissue-Level implants are used.

A prior finite-element study analyzed the strain and stress distribution surrounding Standard and short posterior implants [27]. They investigated the vertical and oblique loading of 365 N for the molars and 200 N for the premolars at 30° in abutments and implants of three brands. They found that the stress and strain in trabecular and cortical bone were improved when the implant had an internal hexagon design, showing that a perfect fit between the implant and abutment reduces stress. This corroborates the present results which showed that the stress was lower when the internal connection was standard.

Comparing BL and TL implants, however, it seems that the height of the implant can affect the outcome. According to a previous FEA study, short implants led to BL implants generating higher overall stress values compared to TL implants, though these values were still below the threshold for fracture, suggesting that short implants can be a viable treatment option for patients unable to undergo more invasive surgeries. However, quantitative results for the prosthetic screw cannot compared with the present investigation [28].

Despite differences in the mechanical response, a clinical trial assessed peri-implant mucositis (PM) in Tissue-Level (TL) and Bone-Level (BL) implants after non-surgical debridement. Fifty-four patients with 74 implants (39 TL, 35 BL) underwent subgingival debridement. After 6 months, both groups showed similar prevalence and significant improvements in clinical parameters. However, the complete resolution of PM was not

achieved in either group, with comparable changes in bleeding upon probing observed for both implant types [29]. This may have been due to similarities in manufacturing control and surface treatment, which likely influenced the biological response more than mechanical failure.

The present investigation supports the findings of Anniwaer et al., 2024 [30], which suggested that abutments in a TL group were particularly sensitive to preload changes compared to those in a BL group. Their study showed that variations in the preload significantly affected stress distribution in both implants and screws. It is important to note that the present analysis focused on single-unit implant restoration. Other studies [31] have also simulated the replacement of a premolar with an implant-supported cantilever prosthesis, showing that implant type influences fracture load. Their results indicate that Bone-Level implants limit catastrophic failures to restoration, whereas Tissue-Level implants are more prone to deformation in the transmucosal portion, complicating repairs. This aligns with our findings that stress is higher at the neck of the implant in TL implants.

This outcome is also reflected in a case report [32], where a patient with a Tissue-Level implant (right maxillary first molar) placed ten years prior experienced crown movement. A panoramic radiograph revealed a fractured abutment screw. After removing the screw fragment, a new restoration was placed, but two years later, both the screw and implant fractured, requiring surgical removal. The authors emphasized that prevention is key to avoiding implant failure, recommending the use of wide-diameter implants placed centrally in the occlusion, particularly in the posterior area. The present authors echo and support this recommendation, suggesting that implants with thicker walls may help reduce stress during loading.

In the present simulation, while rough contact models were used for simplicity, they can also be considered a limitation because they oversimplify the physical interaction between implant–abutment surfaces. They typically assume a perfect, frictional connection between the surfaces without considering real-world complexities like sliding, varying friction, microscopic surface deformations, contaminants, saliva, and other aspects present in implant-supported restoration. Another study limitation is that the residual stress due to tightening was not directly simulated. Additionally, the absence of dynamic loading and fatigue considerations limits the understanding of the implant's behavior under real-world conditions.

In summary, the simulated model showed that the Tissue-Level implants had a shorter screw nearby the cervical area than the Bone-Level implants. As it bent during force incidence, higher stress levels closer to the cervical area could be observed. Additionally, it is important to observe that the axial wall from the Tissue-Level implants was thinner than that from the Bone-Level implants. The combination of the thinner wall and smaller screw probably increased the stress on the Tissue-Level implants during oblique load incidence, an effect that can be exacerbated when an abutment–implant connection is not ideal.

5. Conclusions

Based on the present results, the stress peaks were lower in the Bone-Level implants compared to those in the Tissue-Level implants. Additionally, implant–abutment connections with larger contact areas reduced the stress concentration compared to those with partial contact area. This effect was more present in the Tissue-Level implant design.

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