Biomechanical Behavior of Tooth-Implant Supported Prostheses With Different Implant Connections: A Nonlinear Finite Element Analysis

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Purpose: *Biomechanical behavior of tooth-implant–supported prostheses (TISPs) with external and internal implants was compared.*

Materials and Methods: *Two 3-D models of TISP were designed by varying the implant: external (Model EH) and internal hexagons (Model IH). After loading, von Mises stresses were obtained in implants, abutments, and screws. Principal maximum (σmax) and minimum (σmin) stresses were analyzed in periodontal ligament (PL), alveolar bone, and periimplant bone.*

Results: *Model IH showed lower stress peaks in axial loading in the implant and in the screw but higher in abutment. In oblique loading, Model IH had lower stresses in the implant, but higher in the abutment and in the screw. In the σmax analysis for axial and oblique loads, stress peaks in Model IH were lower in PL, alveolar bone, and periimplant bone. In the σmin analysis for axial load, stress peaks in Model IH were lower in PL, but higher in alveolar bone and in periimplant bone. In oblique load, Model IH showed lower stress peaks in PL and alveolar bone, but higher stress peaks in periimplant bone.*

Conclusions: *TISPs with IH implants do present lower risk of biomechanical failure. (Implant Dent 2018;27:294–302)*

Key Words: dental implant, dental prosthesis, biomechanics

Marked improvements have been made in oral rehabilitation since the introduction of osseointegrated implants by Professor Per-Ingvar Brånemark.[1] Implants are used as independent support in single or multiple prostheses in the treatment of edentulous patients. However, in some situations because of anatomical limitations, dental implants can be used mutually with natural teeth in the same prosthesis, resulting in a tooth-implant supported prosthesis (TISP).[2–7]

Teeth and implants have biomechanical differences related to bone insertion and tactile sensitivity, resulting in different degrees of mobility and responses under occlusal loads.[5,8,9] These factors create concerns about joining implants and natural teeth in the same prosthetic structure. Nonetheless, longitudinal clinical studies have shown that prostheses containing both implants and teeth as abutments may present satisfactory outcomes because some factors are observed, such as periodontal health of the teeth, prosthesis design, and the absence of parafunctional habits.[10–13] Long-term clinical studies and a systematic review showed small failure rates in the first 5 years of function of TISP.[5,7,14,15] Regardless of the clinical success of TISP when certain criteria are strictly followed, improper fit between the implant and prosthetic abutment may still be a risk factor for mechanical and biological failures.[16–18] Macro- and micromovements in the implant-abutment interface can expose the joint to undesirable and concentrated stresses, which may cause loosening or fracture of the prosthetic screw[19] or even bone resorption.[20] Studies on the biomechanical behavior of TISP have mainly evaluated the

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influence of prosthetic design (number of elements, presence or absence of cantilever, rigid or semirigid connection, and material of the prosthesis) and the type of occlusal load.[1,2,10,14,21–33] To date, the influence of implant connection types on the stresses generated in TISP has not been evaluated.

The type of implant-prosthetic connection is an essential factor in the biomechanics of the prosthesis-implant-bone complex and may also influence the longevity of TISP. The intensity and nature of stresses in the marginal periimplant bone tissue,[34,35] as well as the stability of the prosthesis-implant joint, are dependent on the type of connection. Compared with the external hexagon (EH), the internal connections are more stable and more capable of reducing the stress generated in the neck of the implant, thus minimizing the risk of biomechanical problems.[36–40]

Therefore, the aim of the present study was to evaluate the biomechanical effect of the type of connection of the implants—(EH) and internal hexagon (IH)—in TISPs in the alveolar and periimplant bone, implant, and prosthetic components using the three-dimensional (3-D) finite element method (FEM).

Materials and Methods

Models' Construction

Models were designed as previously described, using computer-aided design software (SolidWorks; Dassault-Systèmes; SolidWorks Corp.).[41,42] A 3-D mandibular model was obtained through a computed tomography (i-CAT; Imaging Sciences International LCC) from a previously treated patient with complete dentition. The internal structure of the bone, teeth, and mandibular canal were manually segmented and reconstructed, resulting in a nonparametric model. To enable subsequent editing without significant distortion, the models were parameterized through a software supplement (Scan to 3-D; DassaultSystèmes; SolidWorks Corp.).

After parameterization of the models, a cortical bone layer 0.7 mm in thickness was determined around the periodontal ligament (PL) and around the mandibular canal. The superficial bone was set to be cortical with 2 mm in thickness, whereas its inner portion was set to be medullar, resulting in a type II bone.[43] The PL was modeled with 0.25 mm in thickness around the teeth. The area of the second left premolar, first molar, and second molar were

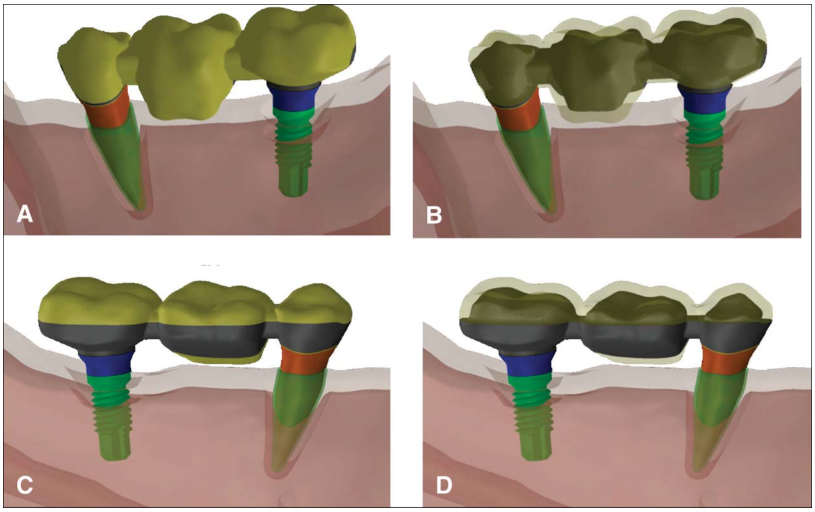

Fig. 1. Model EH: different views (A and B buccal; C and D lingual). 1,441,716 nodes and 895083 elements.

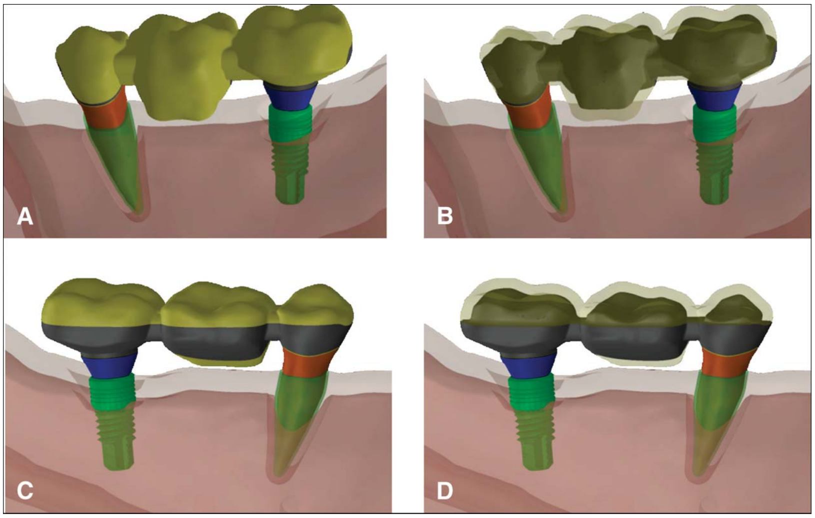

Fig. 2. Model IH: different views (A and B buccal; C and D lingual). 1,491,329 nodes and 926633 elements.

Table 1. Mechanical Properties of Materials

Material	Young Modulus (MPa)	Poisson Coefficient
Dentin[42]	18,600	0.31
Periodontal ligament[42]	68.9	0.45
Cortical bone[42]	13,700	0.30
Medullar bone[42]	1370	0.30
Nickel-chromium[43]	200000	0.33
Feldspathic porcelain[44]	69,000	0.30
Titanium[45]	110000	0.35
Zinc phosphate cement[46]	76,000	0.35

MPa indicates megapascal.

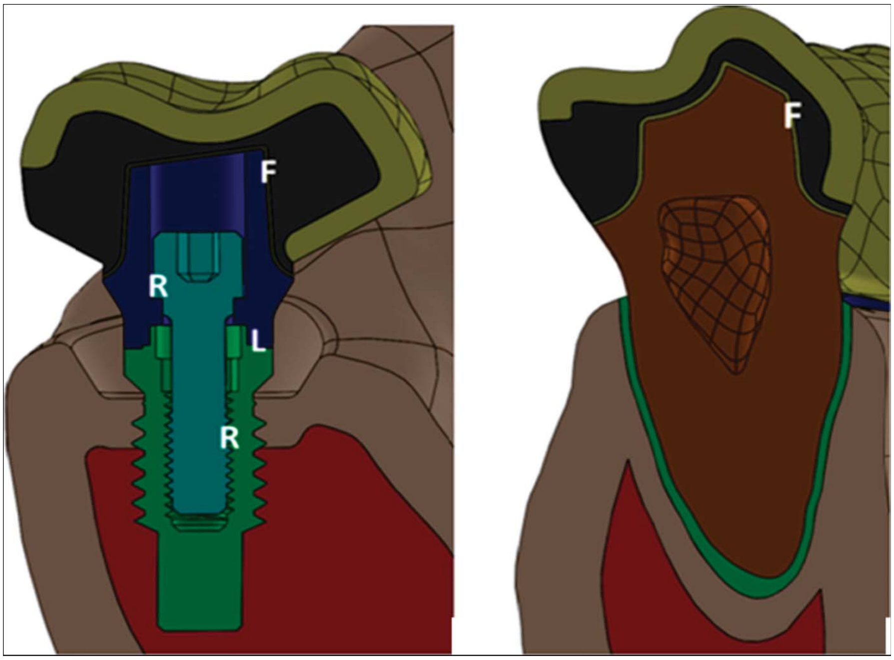

Fig. 3. Nonlinear contacts. R indicates infinite coefficient of friction; F, frictional 0.2 mm; L, frictionless.

reconstructed. An EH implant (Titamax Ti; Neodent), IH implant (Titamax II Plus; Neodent), cement-retained customized titanium abutments for EH and IH (Neodent), and prosthetic screws (Neodent) were modeled by reverse engineering using real parts as references.[42]

Two TISP models were created. Model EH represented a fixed partial prosthesis with the mandibular left second premolar as an abutment, an EH implant in the area of second molar, and the first molar as a pontic (Fig. 1). Model IH represented the same fixed partial prosthesis, but with an IH implant in the area of the second molar (Fig. 2). The position of the implants was the same in both models, as well as the external morphology of the prostheses. The prostheses presented nickel-chromium infrastructures with a minimum thickness of 0.3 mm, covered with feldspathic porcelain with a minimum thickness of 0.9 mm. A zinc phosphate cement layer approximately 0.1 mm thick was simulated between the infrastructure and the abutments.[44]

Simulation

The simulation was performed in finite element analysis (FEA) software (Ansys; Ansys Inc.). All materials and structures were considered homogeneous, linear elastic, and isotropic (Table 1).[45–49]

Nonlinear contacts were defined between different materials or structures.[19,42] Bone-implant union was considered bonded similar to an osseointegrated state. Contacts between infrastructure and zinc phosphate cement were defined with a friction coefficient of 0.2 mm because there is no cohesive adhesion between the zinc phosphate cement and other structures, only mechanical imbrications. The contacts between screw and implant, and between screw and abutment, were set to allow the formation of microspaces, but without slipping between the surfaces, being considered an infinite coefficient of friction. Between implant and abutment, it was considered a frictionless contact, allowing minor sliding between surfaces and formation of

Table 2. Peak Principal Maximum (σMax) and Minimum Stresses (σmin) found in Models EH and IH Under Axial and Oblique Loads. Values in MPa

	Periodontal Ligament				Alveolar Bone				Periimplant Bone			
	Axial Load		Oblique Load		Axial Load		Oblique Load		Axial Load		Oblique Load	
Models	σMax	σMin	σMax	σMin	σMax	σMin	σMax	σMin	σMax	σMin	σMax	σMin
Model EH	0.47	1.33	3.03	6.58	3.52	5.17	10.94	19.52	13.00	13.61	24.22	36.36
Model IH	0.44	1.30	2.75	5.99	3.49	5.20	10.65	18.00	11.25	17.83	21.90	48.11
Difference, %	7	3	9	9	1	–1	3	8	13	–31	10	–32

Difference in percentage between Models EH and IH.
MPa indicates megapascal.

Table 3. Maximum von Mises Equivalent Stresses found in Models EH and IH Under Axial and Oblique Loads. Values in MPa

	Implant		Abutment		Screw	
Models	Axial Load	Oblique Load	Axial Load	Oblique Load	Axial Load	Oblique Load
Model EH	32.64	168.05	38.21	85.78	12.18	78.11
Model IH	22.03	106.63	44.03	122.11	1.94	94.27
Difference, %	33	37	–15	–42	84	–20

Difference in percentage between Models EH and IH.
MPa indicates megapascal.

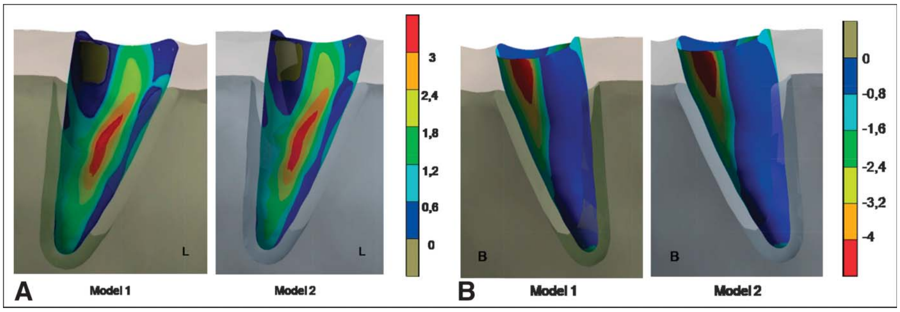

Fig. 4. **A**, Maximum principal stresses in alveolar bone in models EH and IH, under axial load. Lingual view. **B**, Minimum principal stresses in alveolar bone in models EH and IH, under axial load. Buccal view.

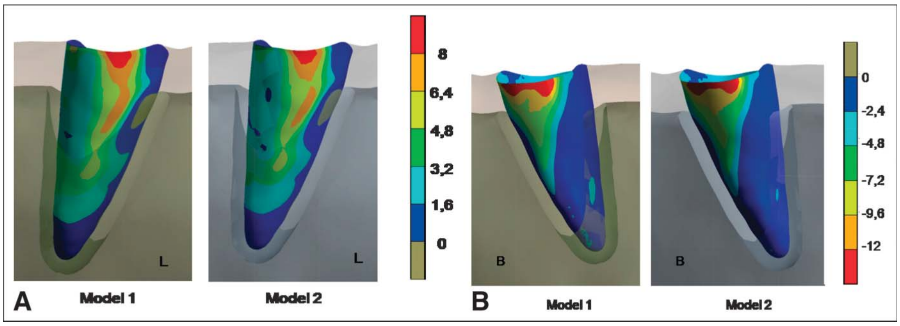

Fig. 5. **A**, Maximum principal stresses in alveolar bone in models EH and IH, under oblique load. Lingual view. **B**, Minimum principal stresses in alveolar bone in models EH and IH, under oblique load. Buccal view.

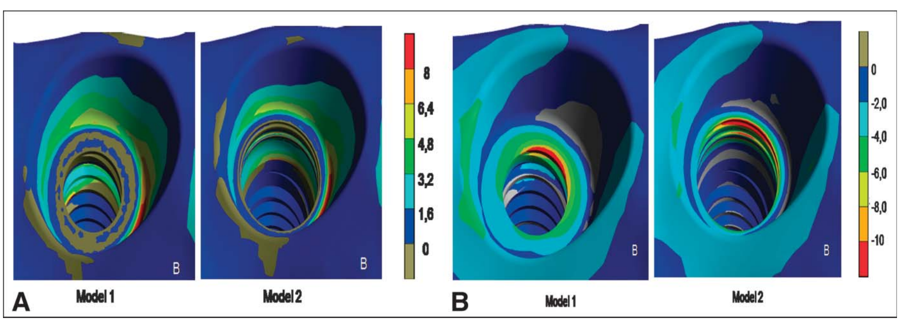

Fig. 6. **A**, Maximum principal stresses in periimplant bone in models EH and IH, under axial load. Occlusal view. B = buccal. **B**, Minimum principal stresses in periimplant bone in models EH and IH, under axial load. Occlusal view. B = buccal.

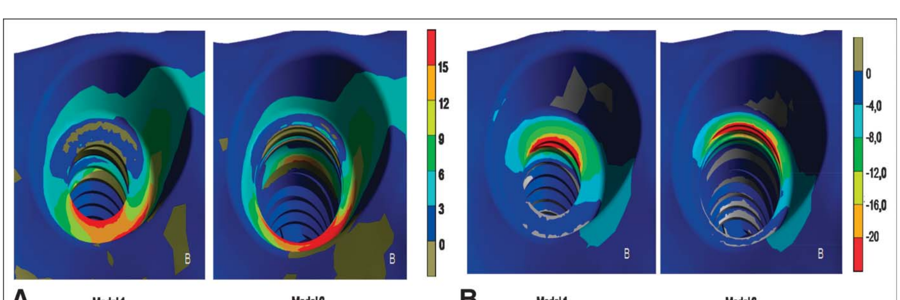

Fig. 7. **A**, Maximum principal stresses in periimplant bone in models EH and IH, under oblique load. Occlusal view. B = buccal. **B**, Minimum principal stresses in periimplant bone in models EH and IH, under oblique load. Occlusal view. B = buccal.

microgaps because of the action of masticatory loads. All other contacts, except for the antagonist in axial loads, were simulated as perfectly bonded.[19,42,50] (Fig. 3).

Axial loads of 100 N were applied to simulate occlusal contact, and loads of 100 N were applied in the buccolingual direction with 45 degrees angulation to simulate the resultant vectors of oblique loading. The meshes were validated by means of a refinement process, verifying the convergence of results. The number of nodes and elements was gradually increased in the areas of peak stress until the difference in peak results between 1 mesh refinement and the other was 5% or less. The mesh was generated with quadratic tetrahedral elements of 10 nodes allowing the simulation of irregular structures such as the present work. The analysis was nonlinear in relation to the contact.

Analysis of Results

Implants, abutments, and screws were analyzed by the von Mises criterion. PL, alveolar, and periimplant bone were analyzed by the criterion of maximum principal stress (σmax, predominantly tensile stresses) and minimum principal stress (σmin, predominantly compressive stresses).

Results

Tables 2 and 3 and Figures 4–12 refer to the results of stress values in both load simulations found in PL, alveolar and periimplant bone, implants, and their prosthetic components, as well as their distribution in these structures.

Discussion

FEM is used for this comparative study because it is neither invasive nor destructive; as a numerical computational analysis, it allows the identification of distinct types of internal or external stresses and displacements in any area of the studied structure. It was, therefore, possible to identify tensile, compressive, and equivalent stresses in areas of the prosthesis-implant-bone complex that are inaccessible by other biomechanical study methods.

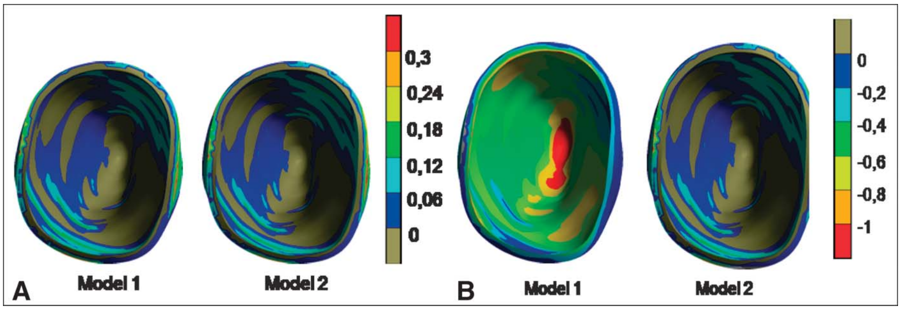

Fig. 8. A, Maximum principal stresses in periodontal ligament in models EH and IH, under axial load. Occlusal view. **B**, Minimum principal stresses in periodontal ligament in models EH and IH, under axial load. Occlusal view.

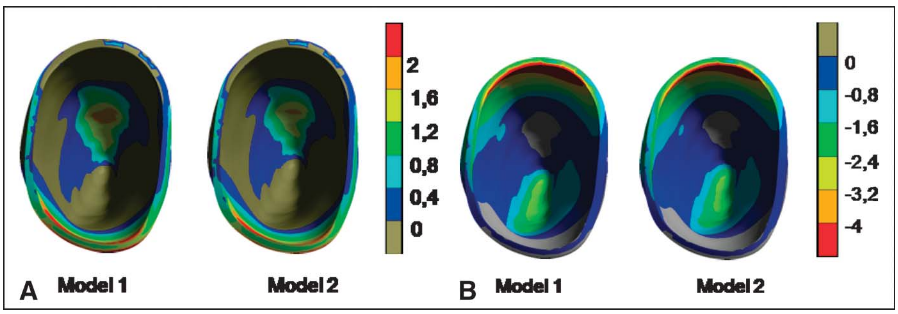

Fig. 9. A, Maximum principal stresses in periodontal ligament in models EH and IH, under oblique load. Occlusal view. **B**, Minimum principal stresses in periodontal ligament in models EH and IH, under oblique load. Occlusal view.

Commonly used in implant dentistry, FEM is applied in various simulations.[18–20,31,37–39,42,46,47] However, similar to any research method, it has its limitations, particularly when used to extrapolate results to the clinical field. To minimize these limitations, modeling must be as close as possible to the real structure, as well as to the surface interactions between different materials. In this study, implants and prosthetic components were carefully modeled using reverse engineering techniques.[42] Bone was designed using the 3-D reconstruction of a real tomography, aiming to replicate faithfully the anatomy within the actual structure.[41] Furthermore, nonlinear contacts were simulated between components that did not present cohesive union, also representing real conditions. With these precautions and refinements, an FEA allows identification of the most probable site for mechanical failure or, as in the present situation, determination of the prosthesis type displaying better biomechanical behavior.

In a posterior edentulous mandibular area, if the amount and quality of the bone tissue are adequate, the most frequently indicated therapy is rehabilitation with implant-supported prostheses, which is a predictable treatment with high survival rates.[14] However, in cases where there is extensive bone loss, especially in height, this anatomical limitation influences the therapeutic decision. A viable alternative is rehabilitation with TISP,[7,10] which eliminates the need for inferior alveolar nerve transposition, the risk of bone graft complications, long cantilevers, or the use of removable partial prostheses. However, tooth vitality and caries activity, periodontal conditions, and biomechanical long-term risks should be considered.[9,13] Clinical and experimental studies present heterogeneous methodologies, which leads to a lack of consensus, creating uncertainty around the clinical decision to use TISP.[13,29] Moreover, TISP can increase stress at the bone-implant-abutment interface because of the cantilever effect caused by physiological tooth movement given by PL, creating a bending moment in the region.[16–18,29] Thus, it is essential to have a thorough understanding of the biomechanics of this type of rehabilitation before recommending its clinical use.

Clinical studies have shown that the success rate and survival of implants in TISP ranged from 90% to 98% in 5 years from 89% to 94.9% in 10 years.[7,10,14,15,24,25,27,30]

However, none of these studies considered the type of implant connection as a parameter for analysis and discussion. This study aimed to elucidate possible biomechanical differences when implants of different connections are used as TISP pillars.

The analyses in this study were performed in the alveolar and periimplant bone, PL, implants, abutments, and retention screws because previous studies showed different TISP failures, including fracture of the prosthesis, loosening and fracture of the screw, loss of retention due to failure of the cement, infrastructure fracture, loss of osseointegration, implant fracture, and intrusion of the tooth.[1,2,5,9,12,22,28] As these studies were clinical, it is not possible to compare their results directly with those of the present study. However, a link designed to analyze the mechanical and biological risks of the structures could be performed.

Bone was evaluated for biological risk around tooth and implants. The mechanical adaptation of bone to masticatory forces acts in its remodeling process, with compressive stresses (σmin) possibly promoting bone growth, whereas tensile stresses (σmax) may cause resorption.[35] Clinical studies of 2, 3, and 15 years of follow-up showed low rates of bone loss in TISP.[10,24,25,27] In this study, under axial and oblique loads, the stresses σmax and σmin in the alveolar bone presented similar stress values between the 2 models. In the PL, the pattern and values of stresses were also similar in both models. PL is a physiologically modifiable tissue when exposed to different loads, acting as a shock absorber.[22] In addition, the PL's presence around the teeth acts to dissipate stress in the alveolar bone.[21,31] The different implant connection resulted in minor differences in the tooth in TISP, regarding principal stress, because of the presence of PL.

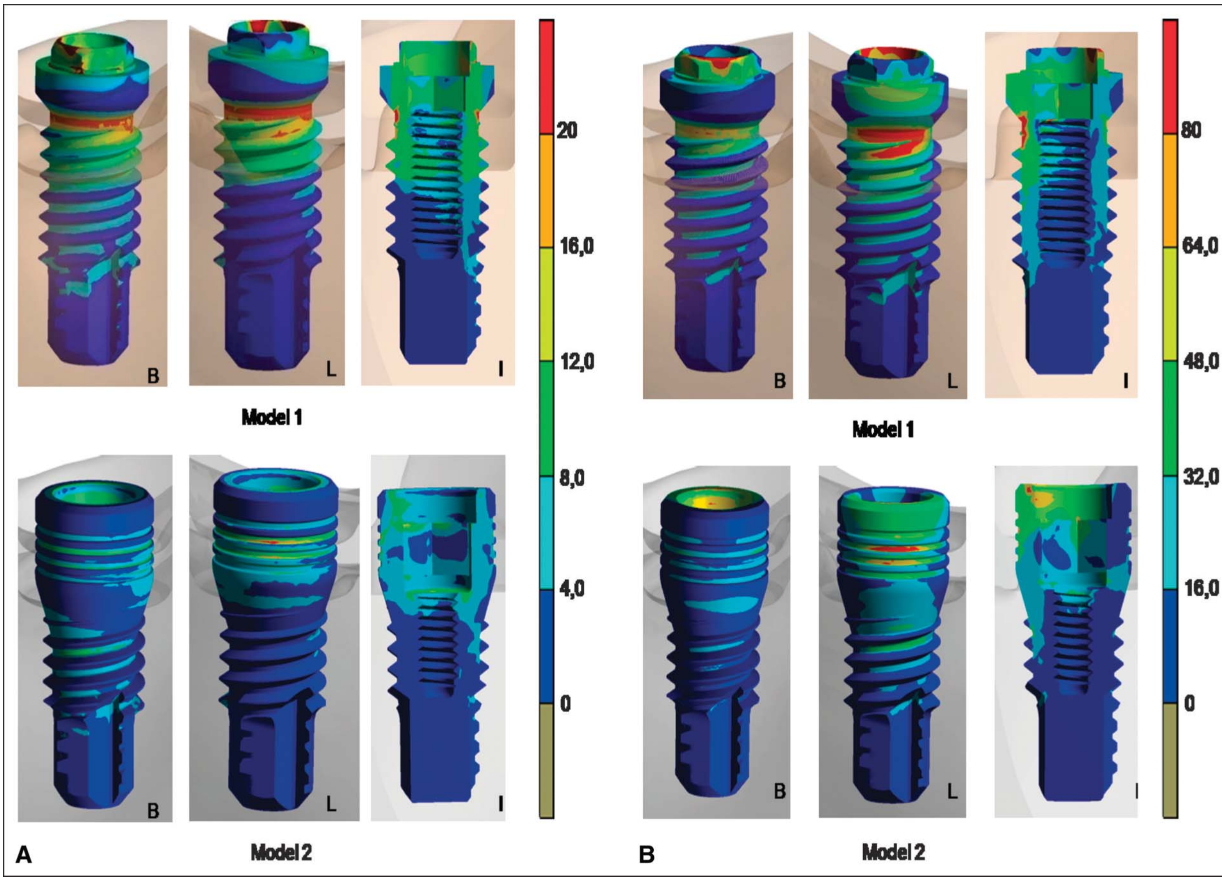

Fig. 10. **A**, von Mises stresses in external hexagon (Model EH) and internal hexagon (Model IH), under axial load. B indicates buccal; L, lingual; I, interior. **B**, von Mises stresses in Model EH and Model IH, under oblique load. B indicates buccal; L, lingual; I, interior.

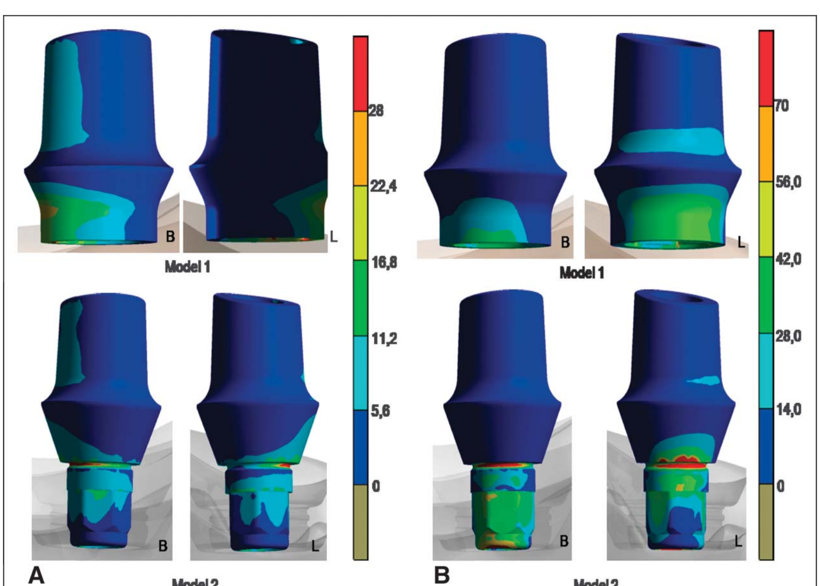

Fig. 11. **A**, von Mises stresses in external hexagon abutment (Model EH) and internal hexagon abutment (Model IH), under axial load. B indicates buccal; L, lingual. **B**, von Mises stresses in Model EH and Model IH abutments, under oblique load. B indicates buccal; L, lingual.

In the analysis of periimplant bone, stresses σmax and σmin were concentrated in the cervical area. Under axial and oblique loads, Model IH presented lower values of σmax (13% axial and 9% oblique) and higher values of σmin (31% axial and 32% oblique). σmin stresses are less deleterious to bone tissue and sometimes may even promote bone growth,[34,35] so it is important to focus on tensile stresses. It is possible that higher σmax stresses found in Model EH may be relevant in the long term, leading to a risk of bone loss. Repeated loads, even at low intensity, can lead to material fatigue, often causing irreversible damage.[34,35]

In implants, the results showed that von Mises stresses were concentrated in the neck, under both axial and oblique loads, with Model IH presenting 33% (axial loading) and 37% (oblique loading) lower stress values. The concentration of stress in the cervical area of

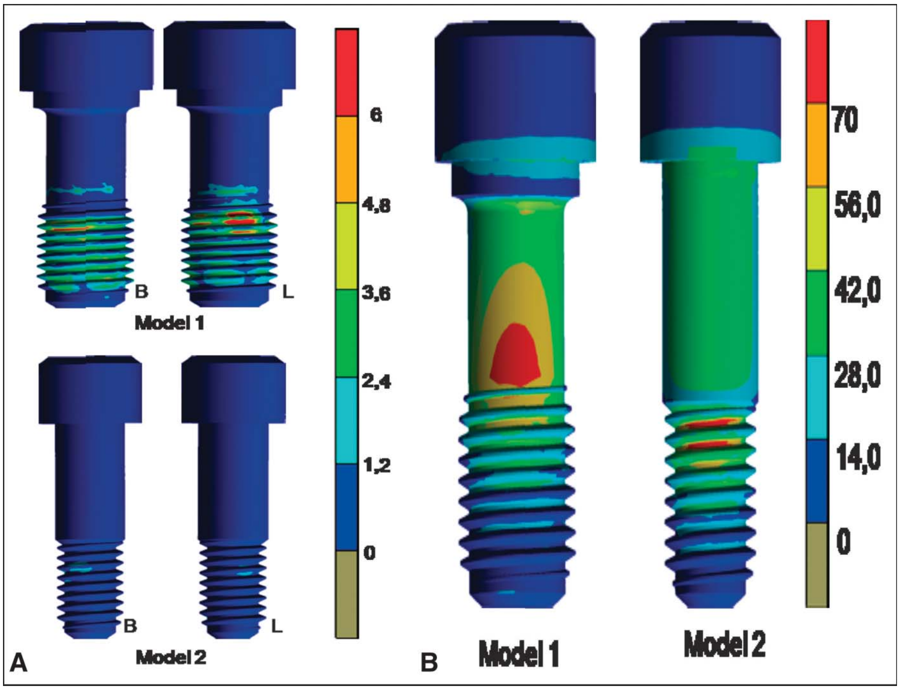

Fig. 12. **A**, von Mises stresses in external hexagon screw (Model EH) and internal hexagon screw (Model IH), under axial load. B indicates buccal; L, lingual. **B**, von Mises stresses in Model EH screw and Model IH screw, under oblique load. Buccal view.

the implant was expected; in accordance with the classic principle of Saint-Venant, the greatest stress occurs in the area where 1 material, the implant, initially meets another, bone. The stress values were of greater magnitude under oblique load, demonstrating the damaging effect of this type of occlusal loading. Stresses were more favorably distributed in IH implants as previously described,[38–40] probably across the larger area of contact between the abutment and implant found in internal connection implants, which decrease points of stress concentrations in implants.

Regarding the abutments, Model IH presented higher values of von Mises stresses under axial and oblique loads, with the stresses concentrated in the hexagons. The higher mechanical bonding and locking of the internal connection inside the implant can lead to the concentration of stresses in this region because the displacement is reduced. On the other hand, analysis of the screws showed that Model IH presented more favorable results, with better stress distribution. Stresses were concentrated on the screw threads of both models under axial load. Under oblique loading, the stresses on the screws of Model EH implants were concentrated on the screw shank. High stresses in the threads of screws can lead to screw loosening, and peak stress on the shank can cause a fracture.[17,18] Quantitatively, under axial load, stress on the screw of Model IH was 84% lower, whereas under oblique load Model IH showed stresses 20% higher on the screw. This may be justified by the greater mechanical imbrication of the internal connection and low EH height (0.7 mm), which allows greater displacement in the abutment-screw interface of EH under oblique load, relieving stress on the screws.[36] This displacement may eventually decrease stress on the screw; however, it brings great instability to the prosthetic joint.

Biomechanical advantages reported in clinical studies and laboratory and numerical simulations found for internal connection implants in single crowns or multiple prostheses[37–40] seem to also be observed in TISPs. The different connections did not result in different stresses in the alveolar bone and in the PL; however, the more favorable distribution of stresses in the periimplant bone, implant, and prosthetic components may justify the choice of internal connection implants when a TISP is planned.

Conclusions

Under axial loading, TISP with IH had a lower mechanical risk for implants and screws, but higher risk for abutments. IH prosthesis also showed lower biological risk for periimplant bone. Under oblique loading, TISP with IH presented a lower mechanical risk for the tooth and implant and greater risk for the abutment and screw. The biological risk for bone was similar in both prostheses. In general, TISP with IH presented a lower biomechanical risk.

Disclosure

The authors claim to have no financial interest, either directly or indirectly, in the products or information listed in the article.

Roles/Contributions by Authors

Gustavo Assis de Paula: study design, study conduction, data compilation, results interpretation, and initial writing. Guilherme Carvalho Silva: writing and final review. Ênio Lacerda Vilaça: data compilation. Tulimar Machado Cornacchia: study design and results interpretation. Cláudia Silami de Magalhães: study design and results interpretation. Allyson Nogueira Moreira: study design and review.

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