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RESEARCH AND EDUCATION

Influence of parafunctional loading and prosthetic connection on stress distribution: A 3D finite element analysis

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ABSTRACT

Statement of problem. Clinicians should consider parafunctional occlusal load when planning treatment. Prosthetic connections can reduce the stress distribution on an implant-supported prosthesis.

Purpose. The purpose of this 3-dimensional finite element study was to assess the influence of parafunctional loading and prosthetic connections on stress distribution.

Material and methods. Computer-aided design software was used to construct 3 models. Each model was composed of a bone and an implant (external hexagon, internal hexagon, or Morse taper) with a crown. Finite element analysis software was used to generate the finite element mesh and establish the loading and boundary conditions. A normal force (200-N axial load and 100-N oblique load) and parafunctional force (1000-N axial and 500-N oblique load) were applied. Results were visualized as the maximum principal stress. Three-way analysis of variance and Tukey test were performed, and the percentage of contribution of each variable to the stress concentration was calculated from sum-of squares-analysis.

Results. Stress was concentrated around the implant at the cortical bone, and models with the external hexagonal implant showed the highest stresses ($P<.001$). Oblique loads produced high tensile stress concentrations on the site opposite the load direction.

Conclusions. Internal connection implants presented the most favorable biomechanical situation, whereas the least favorable situation was the biomechanical behavior of external connection implants. Parafunctional loading increased the magnitude of stress by 3 to 4 times. (J Prosthet Dent 2015;114:644-651)

The load condition of implant-supported prostheses, an important factor in maintaining long-term osseointegration,[1] may predict the success and longevity of prosthodontic treatment.[2] Late implant failure appears to be related primarily to biomechanical complications of the system components,[3] such as porcelain fracture, screw relaxation, infrastructure fracture, and retention loss of the cemented crown and implant fracture.[4]

Unlike natural teeth, dental implants lack a periodontal ligament that dampens occlusal loads and provides proprioception.[5] When the periodontal ligament is absent, implants are rigidly connected to the bone tissue with low axial (3-5 μm) and lateral (10-50 μm) mobility,[6] providing limited proprioceptive feedback mechanisms of the jaw elevator muscles.

Because patients with implant-supported prostheses tend not to perceive small-magnitude occlusal forces, the occlusal overload increases, causing marginal bone loss and implant failure.[7] Physiological limits of the maxilla and mandible remain unknown.[8] However, excessive dynamic loading or overload by applying repeated loads can cause large strains (2000-3000 microstrain),[9]

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Clinical Implications

The internal connection may be appropriate for oral rehabilitation with an implant-supported prosthesis because of reduced distribution of stress on the surrounding tissues.

microfractures in the bone-implant interface (>4000 microstrain),[10] and a decrease in bone density around the implant platform, thereby leading to bone defects.[11]

Marginal bone loss may be affected by factors such as characteristics of the implant platform surface[12,13] or the prosthetic connection.[14-19] Thus, stability of the connections among different parts of the implant also contributes to the longevity of the implant-supported prosthesis, particularly in single posterior restorations where a resilient interface between the abutment and the implant is necessary.[20] Some researchers have found that the configuration of the platform connections would reduce the stress concentration in the peri-implant bone.[8,11,13-17]

The Brånemark system (Nobel Biocare), which is characterized by an external hexagonal configuration, was designed to facilitate implant insertion and to provide an antirotational feature.[21,22] Despite its reversibility and compatibility with different systems, its effectiveness is limited when subjected to nonaxial loads[23] and possible abutment micromovements that cause instability for the prosthetic interface, gap formation, screw loosening, or fatigue fracture because of reduced hexagonal height.[17,24]

To reduce or eliminate biomechanical problems inherent in external hexagonal configuration systems, alternative connections based on the opposition of the walls of the implants and abutments have been suggested: a conical connection with an angle of between 8 and 11 degrees[15,25-27] and an internal hexagonal connection.[28-30]

Few laboratory studies evaluating the influence of parafunctional loading in implant-supported prostheses are available, especially with regard to variation in prosthetic connection.[31,32] The 3-dimensional (3D) finite element analysis (FEA) method allows areas and/or points with a greater potential for failure after a load application to be predicted. This method may also predict failures resulting from clinical factors at the implant–abutment connection.[33,34]

The purpose of the present study was to assess the influences of parafunctional occlusal force and prosthetic connection on stress distribution by using 3D FEA. The hypotheses of the study were that parafunctional occlusal force would induce high stress concentrations in the cortical bone in all connections; that the external connection would produce a higher stress intensity at the bone level; that the oblique load would produce higher stresses in all prosthetic connections; that the external connection associated with parafunctional occlusal force would induce the highest stresses; and that the incidence of the oblique load and parafunctional occlusal force would result in high-intensity stress on all connections.

Table 1. Models used

Model	Description
1	External hexagon implant with screw-retained implant-supported crown
2	Internal hexagon implant with screw-retained implant-supported crown
3	Morse taper implant with cement-retained implant-supported crown

MATERIAL AND METHODS

The following 3 factors were studied: the connection type at 3 levels (external hexagon, internal hexagon, and Morse taper), the occlusal force at 2 levels (normal, parafunctional), and the load direction at 2 levels (axial and oblique). FEA was used to determine the maximum principal strain for the cortical bone.

Three 3D finite element models were constructed (Table 1). Each model represented a mandibular bone section of the second molar region that included an implant (5.0×10 mm) and a single implant-supported prosthesis. The external hexagon, internal hexagon, and Morse taper connections were tested. All models were loaded with normal and parafunctional occlusal forces and under axial and oblique loading. The bone block was obtained with a computerized tomography scan arrangement using medical imaging software (InVesalius; CTI). The bone block image was simplified by using 3D computer graphics and computer-aided design software (Rhinoceros v4.0, NURBS modeling for Windows; Robert McNeel and Associates).

The implant design was obtained by creating a styled layer descriptor format archive of a specimen of external hexagon, internal hexagon, and Morse taper implants (Conexão Sistemas de Prótese Ltda), which included their specific abutments. The implant and abutment geometries were simplified by 3D computer-aided design software (SolidWorks 2010; SolidWorks Corp).

The implant-supported crown was simulated with a screw connection for external and internal hexagon implant models and with cement connection for Morse taper implant models. The cement layer was simulated with a 0.03-mm thickness.[35]

After the modeling phase, all geometries were exported to finite-element software for pre- and postprocessing (FEMAP v10.2 software; Siemens PLM Software, Inc). The first step was to obtain the meshes by using tetrahedral parabolic solid elements for all structures that were involved. Mechanical properties were determined for values obtained from published reports

Table 2. Properties of materials modeled

Material	Elastic Modulus (GPa)	Poisson Ratio	References
Trabecular bone	1.37	0.30	Sevimay et al[37]
Cortical bone	13.7	0.30	Sertgoz et al[36]
Titanium	110.0	0.35	Sertgoz et al[36]
Ni-Cr alloy	206.0	0.33	Anusavice and Hojjatie[33]
Feldspathic porcelain	82.8	0.35	Eraslan et al[36]
Zinc phosphate cement	22.4	0.35	Anusavice and Hojjatie[36]

and are presented in Table 2.[35-38] All materials were homogeneous, isotropic, and linearly elastic.[33]

The crown-abutment and abutment-implant contacts were assumed to be symmetrical contacts. All other contacts were also assumed to be symmetrically welded. Constraints definitions were established as fixed in the *x*, *y*, and *z* axes at the mesial and distal boundary surfaces of the cortical and trabecular bone. All other model surfaces were unrestricted. A normal occlusal force (200-N axial load and 100-N oblique load)[39] and a parafunctional occlusal force (1000-N axial load and 500-N oblique load)[11] were applied on the occlusal surface of the crowns. The axial load was distributed on 4 points on the internal slope of the cusps, whereas the oblique load was divided into 2 points of loading.

The analysis was generated using preprocessing/postprocessing software (FEMAP v10.2; Siemens PLM Software, Inc) and exported to finite element software (NeiNastran v9.2; Noran Engineering, Inc). Results were then imported into the finite element software for plotting the maximum principal stress map.

The maximum principal stress value was used to evaluate cortical bone stress distribution analysis. This criterion makes it possible to distinguish between tensile and compressive stresses. Positive values represent tensile stress, and negative values represent compressive stresses (in megapascals [MPa]).[40]

Three-way analysis of variance (ANOVA) and the Tukey post hoc test were used to analyze interactions between the main results with statistical software (Sigma Plot 13; Systat Software, Inc) (α=.05). The sum-of-squares criteria was also used.[41] The power analysis was calculated with statistical software for each comparison (connection, load, and force) and analyzed with 3-way ANOVA.

RESULTS

Patterns of stress distribution were similar under normal axial occlusal force (Fig. 1) and parafunctional occlusal forces (Fig. 2) among the 3 structures were evaluated. Tensile and compression stresses were around the platform and at the level of the first implant thread. The parafunctional occlusal force was more damaging (P<.001) to bone tissue than normal occlusal force (Fig. 3). The power of the test was β=.983. The oblique load significantly (P<.001) influenced the stress intensity compared with that of the axial load (β=1) (Fig. 3).

The external connection significantly (P<.001) influenced the stress compared with that of the other connection types (Fig. 3), regardless of the load direction (axial or oblique) (Table 3). No statistically significant differences (P=.991) were found between the Morse taper and internal hexagon connections (Fig. 3). The power of the test was β=.983. The biomechanical behavior of the external connection concentrated the stress to a greater extent, regardless of the analyzed load direction (axial and oblique) (Table 3).

Because of the particularities of the parafunctional occlusal force, it was analyzed separately. Thus, 3-way ANOVA and the Tukey post hoc test were used to analyze connections, regions, and loading. Results indicated no significant difference between the internal hexagon and Morse taper implants (P=1.0). The external hexagon implant significantly influenced stress intensity compared with the internal hexagon and Morse taper connections (P<.001) (Fig. 4).

Specific analysis of the different regions and connections showed statistically significant differences among the distal versus lingual (P<.001), distal versus mesial (P<.001), buccal versus lingual (P<.001), buccal versus mesial (P<.001), and mesial versus lingual (P<.001) regions. However, no statistically significant results were found for the distal versus buccal region (Fig. 5).

Figure 5 also shows that the external hexagon implant exhibited the highest stress compared with that of the other groups (P<.001). However, no statistically significant differences were found between the internal connection implants (P=1.00). Specific analysis of load direction showed that oblique loading significantly influenced the stress compared with the axial force (P<.001), particularly in the external hexagon model (Fig. 6).

DISCUSSION

The FEA method has been used successfully to evaluate the biomechanical performance of implant-supported prostheses. However, it has certain limitations such as simplification and the impossibility of simulating the biological response of the structures analyzed. Thus, simplifications should be performed so that the final configuration reflects the clinical situation and best reveals the characteristics to be analyzed. Simplifications performed on all models, therefore, did not generate large changes in the final results. The simplification level was controlled to achieve an acceptable margin of error.[33]

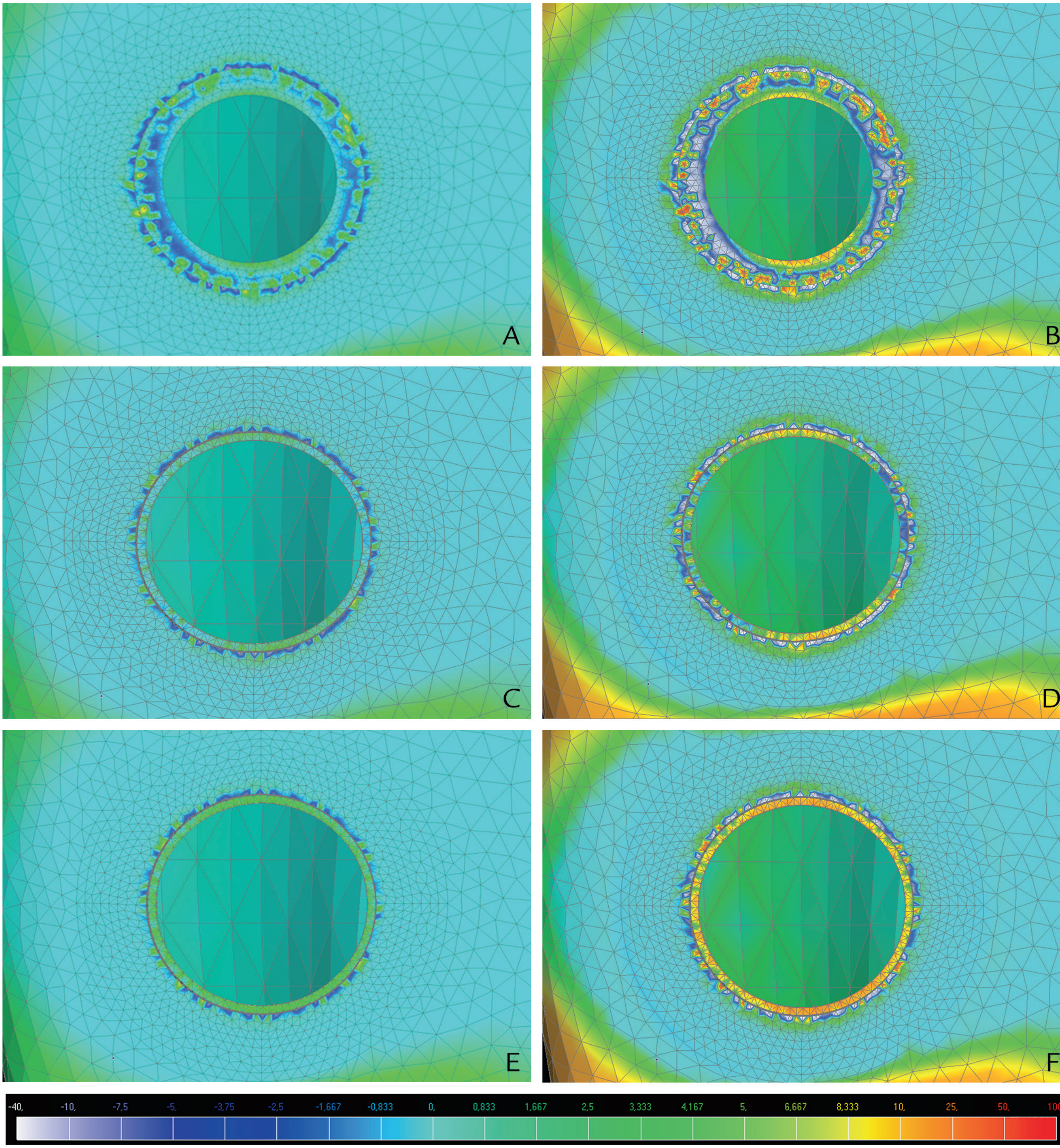

Figure 1. Maximum principal stress of cortical bone under axial load. A, Normal occlusal force, external hexagon. B, Normal occlusal force, internal hexagon. C, Normal occlusal force, Morse taper. D, Parafunctional occlusal force, external hexagon. E, Parafunctional occlusal force, internal hexagon. F, Parafunctional occlusal force, Morse taper.

Results of this study demonstrated that parafunctional loads and oblique loads produced high stress levels in bone tissue, especially in the external hexagonal connection. Study hypotheses, therefore, were confirmed in that the maximum principal stress values based on statistical analyses showed that these factors greatly influenced stress concentration.

The FEA of this study showed that the characteristics of a prosthetic connection are closely associated with the stress distribution pattern, which has also been observed in previous studies.[15,16] Within this context, implants that have an external connection configuration do not allow positive locking. The loads are thus absorbed by the abutment screw, which can

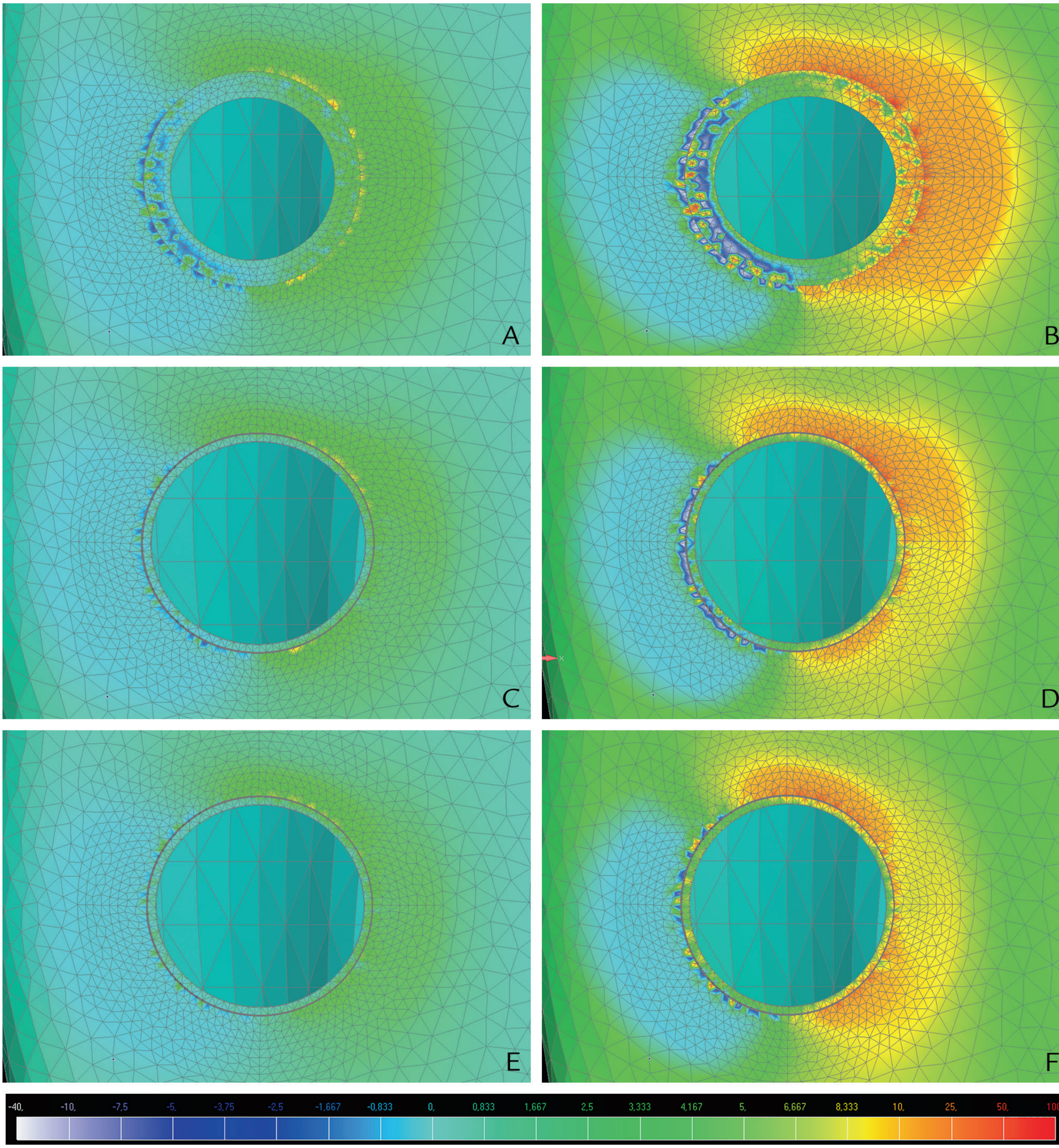

Figure 2. Maximum principal stress of cortical bone under oblique load. A, Normal occlusal force, external hexagon. B, Normal occlusal force, internal hexagon. C, Normal occlusal force, Morse taper. D, Parafunctional occlusal force, external hexagon. E, Parafunctional occlusal force, internal hexagon. F, Parafunctional occlusal force, Morse taper.

produce increased micromovement because of the hexagonal size and the rotation center and can reduce the resistance to lateral and rotational movements.[17] This may negatively influence the concentration of stresses in periimplant bone, which has been observed in other studies.[17-19]

Internal connections exhibited the most favorable situation from a biomechanical point of view, probably because of their geometric configuration. With regard to the Morse taper connections, stress centralization, micromovements reduction,[13,14,17] connection depth, manner of locking, and friction of the internal conical

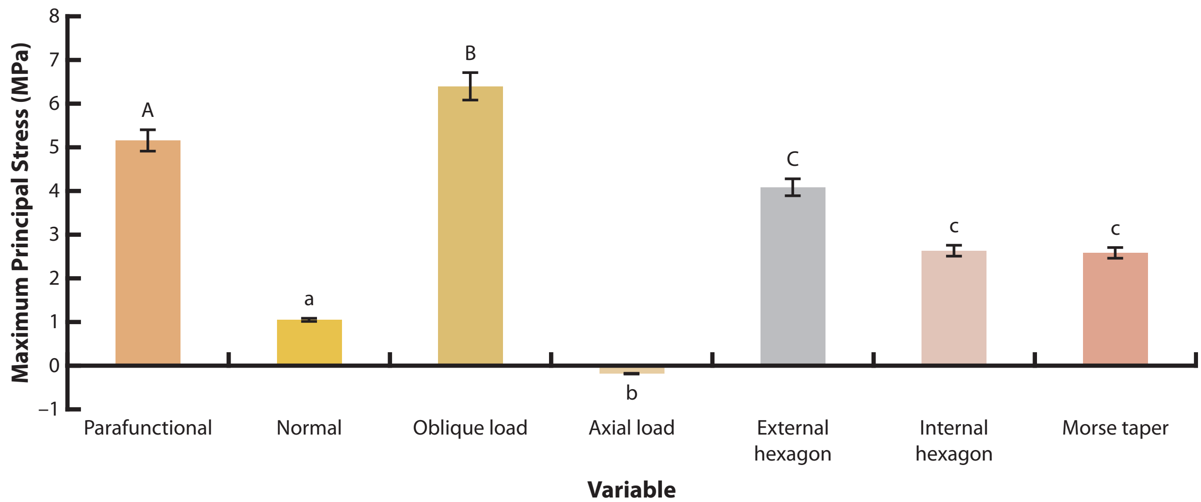

Figure 3. Maximum principal stress for different variables analyzed. Items with the same uppercase and lowercase letters indicate a statistically significant difference ($P<.05$). Items with the same lowercase letters indicate no statistically significant difference ($P>.05$).

Table 3. Maximum principal stress to prosthetic connections

Connections	Force	Load	Mean	±SD
External hexagon	Normal	Axial	-0.0374	0.27
Internal hexagon	Normal	Axial	-0.0942	0.15
Morse taper	Normal	Axial	-0.0523	0.12
External hexagon	Normal	Oblique	2.869	1.83
Internal hexagon	Normal	Oblique	1.866	1.20
Morse taper	Normal	Oblique	1.733	1.29
External hexagon	Parafunctional	Axial	-0.215	1.33
Internal hexagon	Parafunctional	Axial	-0.481	0.72
Morse taper	Parafunctional	Axial	-0.24	0.58
External hexagon	Parafunctional	Oblique	13.743	9.32
Internal hexagon	Parafunctional	Oblique	9.265	6.15
Morse taper	Parafunctional	Oblique	8.916	6.67
*$P<.001$ SS: 301.289	$P<.001$ SS: 2612.734	$P<.001$ SS:6683.694		

Data show means, standard deviations (SD), and P values for maximum principal stress to prosthetic connections (external hexagon, internal hexagon, and Morse taper), force (normal and parafunctional), and load (axial and oblique loading). Sample size: 12.
SS, sum of square.
*Three-way analysis of variance ($P<.05$).

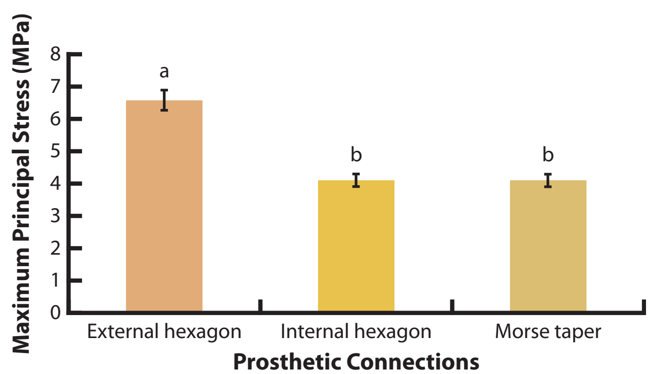

Figure 4. Maximum principal stress for parafunctional occlusal force versus prosthetic connections. Items with different letters indicate a statistically significant difference ($P<.05$). Items with the same lowercase letters indicate no statistically significant difference ($P>.05$).

connection are involved in the resistance to nonaxial loads[15,27] and, therefore, in stress reduction in bone tissue. Favoring the stress distribution in the connection area among the inner walls of the internal hexagon implant produced the same effect, as verified by mathematical models[28] and by in vitro studies.[29,30]

According to Misch,[22] external hexagon implants represent an option for rehabilitating patients with bruxism because the increase in the inner diameter of internal hexagon implants and consequent reduction of its walls compared with that of the external connection implant can reduce its resistance by approximately 40%. However, evidence from the current study indicates that areas of tensile stress (ranging from 50 to 100 MPa) in cortical bone around the platform, particularly under parafunctional oblique loading (which was also verified by statistical analysis), could predispose the bone tissue to a limiting situation, given that, according to Bozkaya et al[11] the limit of cortical bone under tensile stress is 100 MPa.

Parafunctional loading increased the magnitude of the stresses 3 to 4 times and created a greater area of distribution in the implant system and in the bone tissue, although the tensile strength of the structures was not exceeded. According to Misch,[42] bruxism modifies the normal masticatory force duration (in hours rather than in minutes), its direction (lateral rather than axial), type (shear rather than compression), and magnitude (4-7 times). These factors, when combined, could have contributed to the results observed in the 3D models of this study.

Bruxism and other oral parafunctions have been listed as risk factors with regard to dental implant treatment.[31,32] Researchers have used this parafunctional habit as an exclusion criterion for clinical studies.[43] For

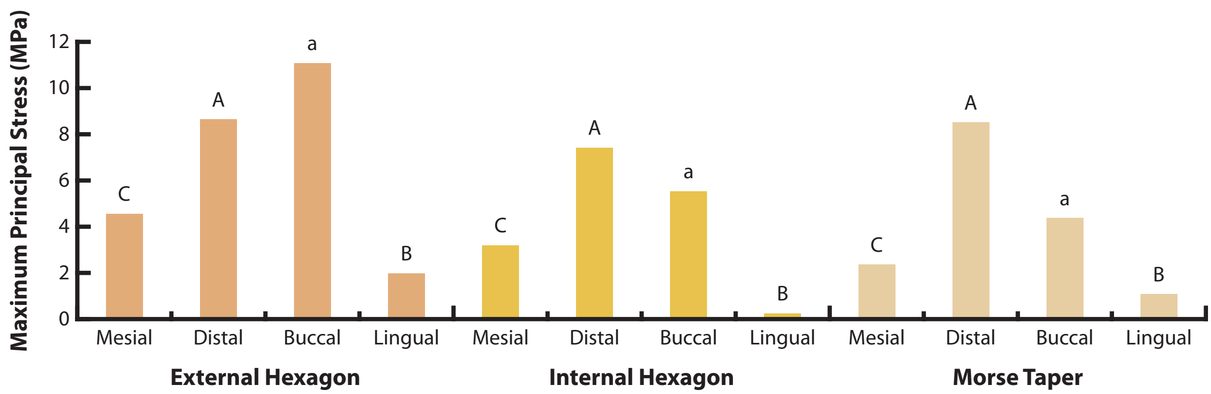

Figure 5. Maximum principal stress for prosthetic connections versus different areas. Items with different letters indicate a statistically significant difference ($P<.05$). Items with same lowercase letters indicate no statistically significant difference ($P>.05$).

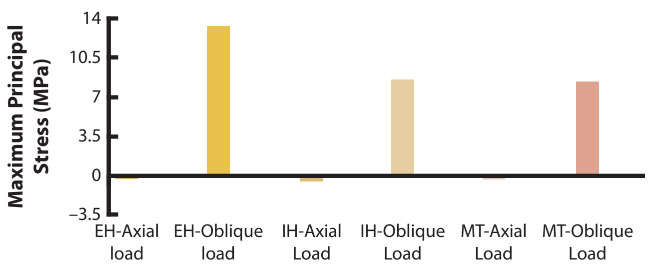

Figure 6. Maximum principal stress values for prosthetic connections versus load direction.

this reason, published reports have provided limited information about the influence of bruxism on dental implants. Only 1 recommendation, based on the clinical experience of experts, is available as a guideline.[7]

CONCLUSIONS

The following conclusions were reached based on the methodology and results: the internal connection implants presented the most favorable biomechanical situation under different loading types and directions; the biomechanical behavior of the external connection implant was least favorable in relation to bone tissue; and the parafunctional loading induced an increase of 3 to 4 times the magnitude of the stresses in bone tissue compared to that with functional loading.

REFERENCES

1. Duyck J, Naert I, Van Oosterwyck H, Ronold HJ, Naert I, Vander Sloten J, et al. The influence of static and dynamic loading on marginal bone reactions around osseointegrated implants: an animal experimental study. Clin Oral Implants Res 2001;12:207-18.
2. Attard NJ, Zarb G. Long-term treatment outcomes in edentulous patients with implant-fixed prostheses: the Toronto study. Int J Prosthodont 2004;17:417-24.
3. Taylor TD, Agar JR, Vogiatzi T. Implant prosthodontics: current perspective and future directions. Int J Oral Maxillofac Implants 2000;15:66-75.
4. Brägger U, Aeschlimann S, Bürgin W, Hämmerle CH, Lang NP. Biological and technical complications and failures with fixed partial dentures (FPD) on implants and teeth after four to five years of function. Clin Oral Implants Res 2001;12:26-34.
5. Chaichanasiri E, Nanakorn P, Tharanon W, Sloten JV. Finite element analysis of bone around a dental implant supporting a crown with a premature contact. J Med Assoc Thai 2009;92:1336-44.
6. Schulte W. Implants and the periodontium. Int Dent J 1995;45:16-26.
7. Kim Y, Oh TJ, Misch CE, Wang HL. Occlusal considerations in implant therapy: clinical guidelines with biomechanical rationale. Clin Oral Implants Res 2005;16:26-35.
8. Sahin S, Cehreli MC, Yalçin E. The influence of functional forces on the biomechanics of implant-supported prostheses—a review. J Dent 2002;30:271-82.
9. Stanford CM, Brand RA. Toward an understanding of implant occlusion and strain adaptive bone modeling and remodeling. J Prosthet Dent 1999;81:553-61.
10. Roberts WE. Fundamental principles of bone physiology, metabolism and loading. In: Baert I, van Steenberghe D, Worthington P, editors. Osseointegration in oral rehabilitation. An introductory textbook. London: Quintessence; 1993. p. 163-4.
11. Bozkaya D, Muftu S, Muftu A. Evaluation of load transfer characteristics of five different implants in compact bone at different load levels by finite elements analysis. J Prosthet Dent 2004;92:523-30.
12. Penarrocha M, Palomar M, Sanchis JM, Guarinos J, Balaguer J. Radiologic study of marginal bone loss around 108 dental implants and its relationship to smoking, implant location, and morphology. Int J Oral Maxillofac Implants 2004;19:861-7.
13. Liu S, Tang C, Yu J, Dai W, Bao Y, Hun D. The effect of platform switching on stress distribution in implants and periimplant bone studied by nonlinear finite element analysis. J Prosthet Dent 2014;112:1111-8.
14. Carvalho MA, Sotto-Maior BS, Del Del Cury AA, Pessanha Henriques GE. Effect of platform connection and abutment material on stress distribution in single anterior implant-supported restorations: a nonlinear 3-dimensional finite element analysis. J Prosthet Dent 2014;112:1096-102.
15. Balik A, Karatas MO, Keskin H. Effects of different abutment connection designs on the stress distribution around five different implants: a 3-dimensional finite element analysis. J Oral Implantol 2012;38:491-6.
16. Yamanishi Y, Yamaguchi S, Imazato S, Nakano T, Yatani H. Influences of implant neck design and implant-abutment joint type on peri-implant bone stress and abutment micromovement: three-dimensional finite element analysis. Dent Mat 2012;28:1126-33.
17. Maeda Y, Satoh T, Sogo M. In vitro differences of stress concentrations for internal and external hex implant-abutment connections: a short communication. J Oral Rehabil 2006;33:75-8.
18. Pessoa RS, Muraru L, Júnior EM, Vaz LG, Sloten JV, Duyck J, et al. Influence of implant connection type on the biomechanical environment of immediately placed implants—CT-based nonlinear, three-dimensional finite element analysis. Clin Implant Dent Relat Res 2010;12:219-34.
19. Rangert B, Jemt T, Jorneus L. Forces and moments on Branemark implants. Int J Oral Maxillofac Implants 1989;4:241-7.
20. Theoharidou A, Petridis HP, Tzannas K, Garefis P. Abutment screw loosening in single-implant restorations: a systematic review. Int J Oral Maxillofac Implants 2008;23:681-90.
21. Brånemark P-I, Zarb G, Albrektsson T. Tissue-integrated prostheses. Chicago: Quintessence; 1985. p. 156-62.
22. Misch CE. The effect of bruxism on treatment planning for dental implants. Dent Today 2002;1:76-81.
23. Weinberg LA. The biomechanics of force distribution in implant-supported prostheses. Int J Oral Maxillofac Implants 1993;8:19-31.
24. Gracis S, Michalakis K, Vigolo P, Vult von Steyern P, Zwahlen M, et al. Internal vs. external connections for abutments/reconstructions: a systematic review. Clin Oral Implants Res 2012;23:202-16.

25. Çehreli MC, Şahin S, Akça K. Role of mechanical environment and implant design on bone tissue differentiation: current knowledge and future contexts. J Dent 2004;32:123-32.
26. Lang AL, Kang B, Wang R, Lang RB. Finite element analysis to determine implant preload. J Prosthet Dent 2003;90:539-46.
27. Merz RB, Hunenbart S, Belser CU. Mechanics of the implant-abutment connection: an 8-degree taper compared to a butt joint connection. Int J Oral Maxillofac Implants 2000;15:519-26.
28. Norton MR. An in vitro evaluation of the strength of an internal conical interface compared to a butt joint interface in implant design. Clin Oral Implants Res 1997;8:290-8.
29. Sailer I, Sailer T, Stawarczyk B, Jung RE, Hämmerle CH. *In vitro* study of the influence of the type of connection on the fracture load of zirconia abutments with internal and external implant-abutment connections. Int J Oral Maxillofac Implants 2009;24:850-8.
30. Seetoh YL, Tan KB, Chua EK, Quek HC, Nicholls JI. Load fatigue performance of conical implant-abutment connections. Int J Oral Maxillofac Implants 2011;6:797-806.
31. Manfredini D, Lobbezoo F. Relationship between bruxism and temporomandibular disorders: a systematic review of literature from 1998 to 2008. Oral Surg Oral Med Oral Pathol 2010;109:26-50.
32. Hsu YT, Fu JH, Al-Hezaimi K, Wang HL. Biomechanical implant treatment complications: a systematic review of clinical studies of implant with at least 1 year of functional loading. Int J Oral Maxillofac Implants 2012;27:894-904.
33. Geng JP, Tan KBC, Liu GR. Application of finite element analysis in implant dentistry: a review of the literature. J Prosthet Dent 2001;85:585-98.
34. Akour SN, Fayyad MA, Nayfeh JF. Finite element analyses of two antirotational designs of implant fixtures. Implant Dent 2005;14:77-81.
35. Anusavice KJ, Hojjatie B. Stress distribution in metal-ceramic crowns with a facial porcelain margin. J Dent Res 1987;66:1493-8.
36. Sertgöz A. Finite element analysis study of the effect of superstructure material on stress distribution in an implant-supported fixed prosthesis. Int J Prosthodont 1997;10:19-27.
37. Sevimay M, Usumez A, Eskitascioglu G. The influence of various occlusal materials on stresses transferred to implant-supported prostheses and supporting bone: a three-dimensional finite-element study. J Biomed Mater Res B Appl Biomater 2005;73:140-7.
38. Eraslan O, Sevimay M, Usumez A, Eskitascioglu G. Effects of cantilever design and material on stress distribution in fixed partial dentures—a finite element analysis. J Oral Rehabil 2005;32:273-8.
39. Morneburg TR, Pröschel PA. Measurement of masticatory forces and implant loads: a methodologic clinical study. Int J Prosthodont 2002;15:20-7.
40. Lin C-H, Wang J-C, Ramp LC, Liu P-R. Biomechanical response of implant systems placed in the maxillary posterior region under various conditions of angulation, bone density, and loading. Int J Oral Maxillofac Implants 2008;23:57-64.
41. Sotto-Maior BS, Senna PM, da Silva WJ, Rocha EP. Del Bel Cury AA. Influence of crown-to-implant ratio, retention system, restorative material, and occlusal loading on stress concentrations in single short implants. Int J Oral Maxillofac Implants 2012;27:e13-8.
42. Misch CE. Treatment planning: force factors related to patient conditions. In: Misch CE, editor. Contemporary implant dentistry. 3rd ed. St. Louis: Mosby; 2008. p. 105-29.
43. Lobbezoo F, Brouwers JE, Cune MS, Naeije M. Dental implants in patients with bruxing habits. J Oral Rehabil 2006;33:152-9.

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