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Computational Biomechanical Analysis of Engaging and Nonengaging Abutments for Implant Screw-Retained Fixed Dental Prostheses

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Abstract

Purpose: To evaluate the stress distribution, using 3-dimensional finite element analysis (FEA), on different implant components of a mandibular screw-retained fixed dental prosthesis (FDP) situation when using different combinations of engaging and nonengaging abutments.
Material and Methods: A model of artificial bone was digitally designed. Dental implants were positioned in the lower right posterior area of teeth #'s 28 (premolar – pm) and 30 (molar – m). Restorative implant components were digitally designed and placed into the implant model. Four different implant abutment situations were simulated through FEA: (1) Both engaging abutments (mE-pmE), (2) both nonengaging (mNE-pmNE), (3) molar nonengaging and premolar engaging (mNE-pmE), and (4) molar engaging and premolar nonengaging (mE-pmNE). Thirty-five (35) Ncm preload to the abutment screws and 160 N static load at 45° angle to the occlusal plane were applied in each group.
Results: The equivalent Von Mises stress was measured on each component. Stress distribution changed among the different configurations and ranged from 516.0 to 1304.6 MPa in the implants, and from 554.6 to 994.5 MPa with the abutments. Higher stress was found for the mNE-pmNE designs (1078.6-1106.9 MPa). Engaging and nonengaging abutments had different stress distributions on the screw (698.8-902.5 MPa). Peak stress areas were located on the upper part of the screws for the nonengaging configuration, and on the lower areas for the engaging abutments. The sum of the stress on both implants decreased in the following order: mNE-pmNE > mE-pmNE > mNE-pmE > mE-pmE.
Conclusion: Under conditions of this study, abutment design produced different stress patterns to the implant components. The lowest and most balanced stress distribution was found for the mE-pmE configuration followed by the mNE-pmE configuration.

Fixed implant-supported dental prostheses (FDPs) have become a routine treatment for restoring partially or fully edentulous patients. A stable implant-abutment interface (IAI) is vital to long-term clinical success and depends on the stress and displacement of this system.[1] Mechanical complications such as screw loosening and fracturing, and abutment fractures occur when high stress is applied onto the prosthetic restoration and implant components.[2,3,4–7] The patterns of stress in the implant-prosthesis-bone complex may be influenced by the macrogeometric shape of the implants (external vs internal indexed), abutment types (engaging vs nonengaging), implant component materials, position of the dental implants, masticatory forces, and fit of the prosthesis.[8,9]

Although screw-retained FDPs have several advantages over cement-retained, achieving passive fit can be more complex and difficult when restoring multiple, nonparallel, internally indexed implants.[17] Retention failures have been reported and have been largely due to misfit of the prosthesis and inadequate torque placed on the prosthetic screws.[10,2,3,11,12] Nevertheless, flexible options of using engaging and nonengaging abutments are available, and good survival and success rates have been reported in the literature.[13]

The design at the IAI can significantly reduce stress and strain on the abutment screw by increasing the contact area for engaging abutments. This results in better load distribution and decreases micromovement that in turn decreases the chance of screw loosening.[14] Although these features of internal indexed implants and engaging abutments enhance the stability of the prosthesis, the path of insertion or draw for multiunit prosthesis on nonparallel implants may be challenging.

Nonengaging abutments is another option that allows up to a certain degree of correction with achieving a good fit of the prosthesis to multiple, nonparallel internally indexed implants. The caveat with nonengaging abutments is that greater stress has been reported around the abutment screws and the IAI when eccentric forces are applied. The choice for using engaging or nonengaging largely depends on the implant positions and angulations.[8,15,16] Therefore, different combinations of engaging and nonengaging abutments have been evaluated and proposed as an alternative to regain some advantages of utilizing the internal connection with an engaging abutment, while achieving the required passive fit with a nonengaging abutment.[8]

Biomechanical aspects of implant-supported FDPs and stress distribution in the components have been investigated in a number of in vitro, animal and clinical studies to predict the clinical behavior of implant-supported restorations.[17–20] Photoelastic analysis and finite element analysis (FEA) have been utilized in dentistry to investigate the biomechanical behavior of teeth, biomaterials, orthodontic appliances and dental implants.[21,22]

The wide application of FEA is due to its cost effectiveness, predictability, and continuously improved accuracy. The response of an implant-prosthesis-bone complex to mechanical loading can be analyzed through FEA. Development of FEA models that are able to predict in vitro and clinical outcomes are still lacking.[23,24] These models are needed to compare a large number of designs and configurations that would otherwise not be feasible with in vitro testing. To date, no study has evaluated different implant abutment configurations for screw-retained implant FDPs using 3D FEA. Therefore, the aim of this study was to use numerical analysis to compare the mechanical response of different engaging and nonengaging abutment configurations for FDPs in the region of the posterior mandible. The primary focus was to evaluate the stress values and distributions of the IAI.

Materials and methods

An artificial bone block (45 mm in length × 20 mm in width × 20 mm in height) and a cortical shell that had a homogeneous thickness of 1.5 mm was designed using Ansys 19 software (Ansys Inc., PA).[25] Digital models of 2 bone level internally indexed endosteal dental implants (Ø 4.1 mm × 10 mm) were created and designed according to an implant manufacturer (SLActive®, RC – Regular CrossFit®, Straumann, MA). The finite element model (FEM) of the teeth was used to position the implants at the level of the bone block in teeth #'s 28 (premolar – pm) and 30 (molar – m) locations. The implant components (screw and abutments) were designed using Geomagic Wrap (3D systems, SC) and Ansys 19 (Ansys Inc.).

The implant screw-retained FDP was created from a digital typodont model through CAD modeling using Geomagic Wrap and exported as an IGES (Initial Graphics Exchange Specification) model (Fig 1).

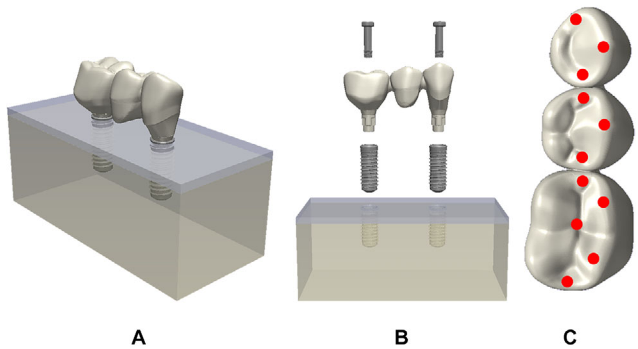

Figure 1 Mesial view of the digital finite element model (A); deconstructed view of all components (B); occlusal contacts used to distribute the load (C).

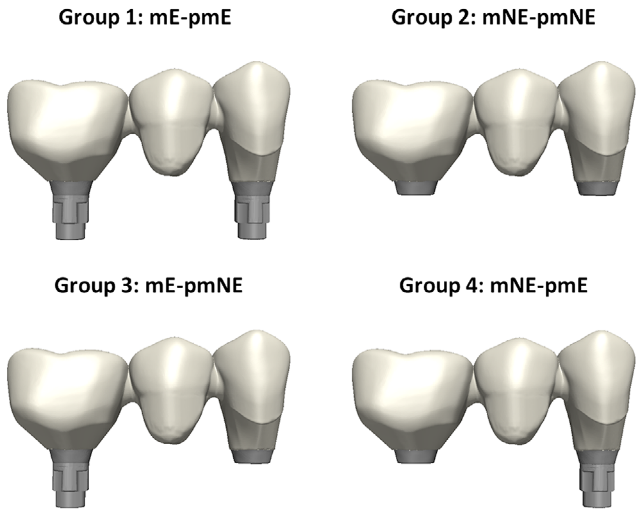

Figure 2 Representation of the four different engaging and nonengaging abutment configurations.

Four implant screw-retained FDP (teeth #'s 28-×-30) groups with different abutment combinations (Straumann, MA) were created (Fig 2): dual engaging (mE-pmE), dual nonengaging (mNE-pmNE), molar nonengaging and premolar engaging (mNE-pmE), and molar engaging and premolar nonengaging (mE-pmNE) abutments.

The implant components, implants and bone block were aligned to the screw-retained FDPs using rigid roto-translatory movements. Using Boolean operations, the bone block was remodeled to create the space for the implants. Constant frictional contacts (0.3)[26] were applied between all implant FDP components except for the abutment-FDP contact surfaces, which were bonded. To simulate complete osseointegration, the implant-bone block contacts were also set as bonded.[27] The nonlinear contact surfaces were solved using the "Pure penalty" settings with a maximum allowed penetration of 0.005 mm. The choice of simulating complete osseointegration was to control the variable and to focus on the different abutment designs on the IAI. All bodies were meshed with solid

Table 1 Mechanical properties assigned to each component

Component	Young's modulus (GPa)	Poisson's ratio
Abutment/screw/implant (Ti6Al4V)	110	0.33
FDP (Zirconia Y-TZP)	209.3	0.32
Cancellous bone	13.7	0.3
Cortical bone	1.37	0.3

elements and were assigned linear elastic mechanical properties (Table 1).[28] According to manufacturer's guidelines, 35 Ncm preload were applied to the screws, and the mesial and distal extremities of the bone block were fixed in all directions.[29,30]

Using Ansys 19 (Ansys Inc.), the simulation was divided into two steps: preload application and occlusal load application. The 35 Ncm preload was simulated applying a 583 N preload onto the screws using the "Bolt Pretension" function. To simulate a more realistic condition, a 160 N load at a 45° angle to the occlusal plane[31] was applied in each group and distributed on the occlusal points (Fig 1C).[32] A convergence test was performed on the mNE-pmNE model to ensure that the mesh size was appropriate for the specific analysis. The Von-Mises stress on the molar implant was used as a reference and a variation of <5% of its value was considered acceptable. The mesh size of 0.075 mm edge length was found to be appropriate and the total number of elements and nodes for the four simulated designs ranged from 501,617 and 858,088 for the mNE-pmNE model to 665,830 and 1,138,859 for the mE-pmE model. Von-Mises stresses were measured for all FDP components with a particular focus on the implant screws and the IAI. The maximum principal stresses were measured for the cortical and cancellous bone, as it is a more effective parameter to predict failure for brittle materials.

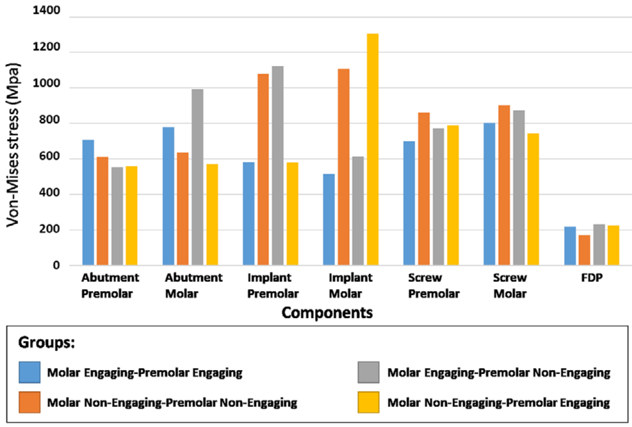

Figure 3 Graphical representation of the maximum Von-Mises stress for each implant FDP component and configuration.

Results

The peak Von-Mises stress values were calculated for each of the FDP components (Fig 3) and Table 2 presents the peak Von-Mises stress values calculated for each of the implant FDP components and the peak maximum principal stress for cancellous and cortical bone. Von-Mises stress values ranged from 173.5 MPa with the mNE-pmNE configuration, to 231.2 MPA with the mNE-pmE configuration. The areas with the highest stress were located at the connection with the abutment of tooth #28 in the mE-pmE configuration, and in the area of the prosthetic connector between pontic tooth #29 and abutment tooth #30 for the other configurations (Fig 4).

The equivalent stress on the abutments ranged from 554.62 MPa for abutment #28 in the mE-pmNE configuration to 994.5 MPa for abutment #30 in the same configuration. Abutment #30 had the highest stress values for all scenarios and for both abutments, higher stress was found for surfaces that contacted the screw.

Table 2 Von-Mises stress values for each implant FDP component and the maximum principal stress for the bone

		Abutment configurations			
Stress	Components	Molar engaging-premolar engaging	Molar nonengaging-premolar nonengaging	Molar engaging-premolar nonengaging	Molar nonengaging-premolar engaging
Von-Mises equivalent (MPa)	Screw 28	698.8	860.4	772.2	790.9
	Screw 30	804.5	902.5	874.4	745.3
	FDP	220.3	173.5	231.2	226.9
	Abutment #28	708.1	612.3	554.62	560.7
	Abutment #30	779.3	637.1	994.5	573.1
	Implant #28	581.9	1078.6	1123.4	580
	Implant #30	516	1106.9	615	1304.6
Maximum principal (MPa)	Cancellous	3.4	2.5	4.1	3.7
	Cortical	22.2	51.5	44.1	32.3

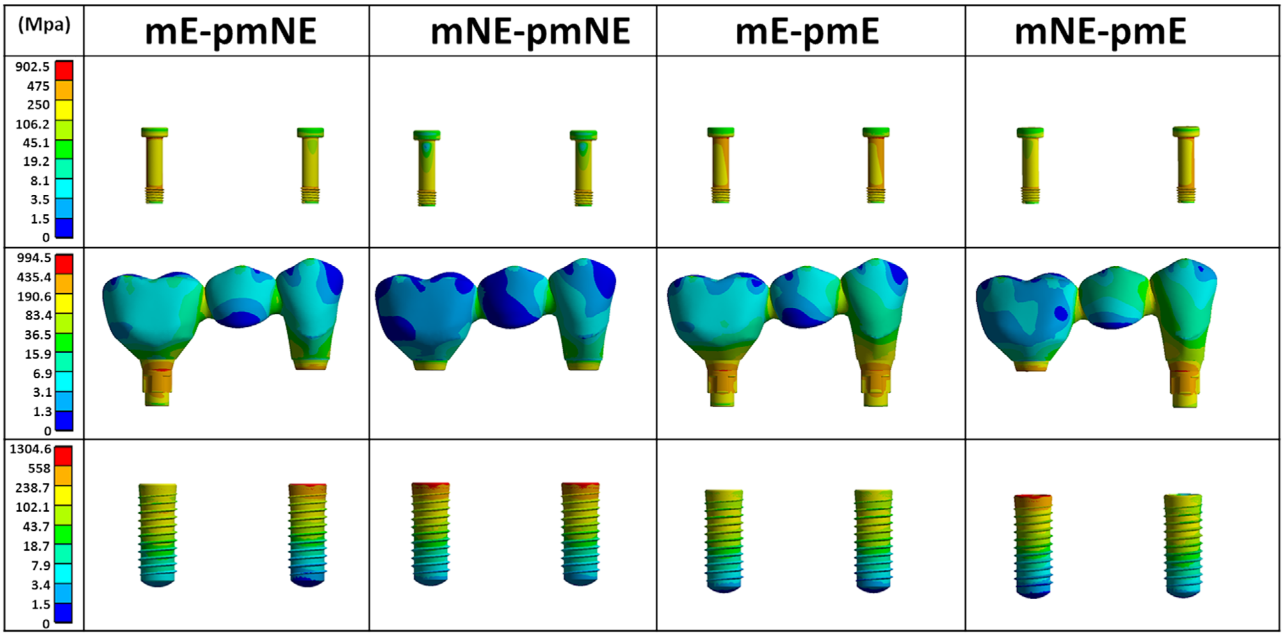

Figure 4 Color map of implant fixed dental prosthesis (FDP) components for each simulation.

The implants were subjected to stress values that ranged from 516.0 to 1304.6 MPa. Differences in stress patterns were found between engaging and nonengaging abutments (Fig 4). For nonengaging abutments, higher stress was found around the implant collars. For the prosthetic screws, different stress and distributions (range of 698.8-902.5 MPa) were found between engaging and nonengaging configurations (Fig 4).

The cancellous bone showed the lowest principal stress value (2.5 MPa) with the mNE-pmNE configuration and the highest (4.1 MPa) with the mE-pmNE configuration. For the cortical bone, the lowest principal stress value (22.2 MPa) was found with the mE-pmE configuration, while the highest (51.5 MPa) was found with the mNE-pmNE configuration.

Discussion

The results of this study showed that abutment design (engaging and nonengaging) and location can affect the stress distribution with implants and restorative components. Out of all tested abutment combinations, the dual nonengaging (mNE-pmNE) abutment configuration had higher stress in the area of the IAI and could be attributed to the decreased extension of the abutment into the internal connection of the implant. This increase in localized stress could possibly cause material wear and loss, which could lead to the breakdown of the implant-prosthetic assembly.[33]

This study uses 3D FEA to evaluate the stress distribution in different implant components. The dual engaging (mE-pmE) abutment configuration had the lowest overall stress distributions while dual nonengaging (mNE-pmNE) abutment configuration had the highest stress distributions when an oblique static load was applied to the model. When comparing both hemi-engaging (mE-pmNE and mNE-pmE) configurations, a more balanced stress distribution among the components was found when the engaging abutment was in the more anterior position.

Previous reports have used in vitro testing to analyze the mechanical behavior and the fatigue response of different implant restorations. Dogus et al[8] presented an in vitro study analyzing the effect of internal engaging abutment position in a cantilevered fixed-splinted 3 unit prosthesis. Their analysis showed that the engaging abutment should be placed further away from the heaviest expected load for the best fatigue response. Although no direct comparisons can be made, this study also found that for hemi-engaging abutments, the mNE-pmE had the best stress distribution when compared to the dual nonengaging and mE-pmNE abutment configurations.

Different stress distributions for the screws were found with the various abutment combinations (Fig 4). For nonengaging abutments, higher stress was found near the neck of the screw, close to the abutment. For engaging abutments, two regions of high stress were found: on the lower part of the screws, near the screw threads and near the neck of the screw. It is expected that this different stress distribution can affect the failure mode, because for the nonengaging abutment screws the stress was more distributed all over the screw, instead of being concentrated around the screw neck.

There appears to be a relationship between abutment design and stress around the bone (Table 2). The cortical bone showed a stress of 22.2 MPa with both engaging abutments but increased to 51.5 MPa with both nonengaging abutments. A plausible explanation is that concentrated areas of stress were found in the area of the implant necks, which may be transferred through the implant surface and to the alveolar bone. This warrants further investigation.

The IAI consists of the implant platform, the prosthetic screws, and the restorative abutments. The higher peak stresses found with engaging abutments could be due to the more restricted configuration (i.e., Extending into the internal aspect of the implant). However, studies have described more favorable dissipation of forces with engaging abutments.[9,29] It was interesting to observe that the most distal prosthetic screw (implant #30) had higher stress values than the more anterior screw for all configurations except for the mNE-pmE configuration. The most distal abutment also had higher stress values for all configurations. These observations could be partially explained by the occlusion that was designed for this study in that higher occlusal forces, depicted as stress values, are generated in the posterior area.

This study showed how the abutment design combination and location can strongly affect the stress in the implants and restorative components. When the stress distribution on the FDP components becomes higher than the yield strength of the material, the components incur a plastic deformation which can consequently lead to screw loosening. Moreover, high stress concentrations can result in deformation and wear between the components.[34] Stress in the area of the implant platforms for nonengaging abutments exceeded the yield strength (880 Mpa) of titanium alloy.[35] This was also observed for the implant abutment #30 (994.5 MPa) for the mE-pmNE configuration.

Considering all tested abutment combinations, the dual nonengaging (mNE-pmNE) configuration had the most stress on the implants while the dual engaging (mE-pmE) configuration was the most balanced for all components. Restoring a screw-retained FDP with dual engaging abutments requires meticulous planning, adequate bone to allow precise parallel placement of the dental implants, and achieving passive fit with the prosthesis; all of which are difficult to achieve for a number of reasons.[14,36] Based on the results of this study, the dual engaging configuration was the best in stress distribution and is recommended if all criteria can be achieved. However, to allow a certain degree of divergence between dental implants and achieve passive fit, an alternative is to use either dual nonengaging or hemi-engaging combinations of abutments.

To control the variables that may affect the results of our study, a more simplified model was designed. A realistic model should account for noncomplete osseointegration and a real bone geometry simulating friction instead of bonded contacts. In a real clinical situation, the actual bone model should be reconstructed possibly by using microcomputed tomography to digitize the actual complex trabecular and cortical bone structure.[32] Additionally, the actual bone should be characterized by anisotropic mechanical behavior, which is often simplified as homogenous in FEA studies.[37] In this study, a comparative biomechanical analysis was performed, and according to previous studies, the comparison is not affected by the simplified modeling.[27]

A study by Wang et al[38] demonstrated through electronic strain measurements that FEA is an accurate tool to measure the mechanical response of FDP. For this reason, this study focused on the comparison of all possible engaging and nonengaging abutment configurations for screw-retained FDPs in the mandibular posterior region, which is a common area for performing biomechanical testing due to reported high success and survival of dental implants, occlusal loads, and is a common area for missing teeth.[39]

An advantage of FEA allows for preliminary analysis that could provide hints for future in vitro experiments. For clinical and in vitro testing, it is important to apply the appropriate preload to the implant screws to prevent loosening or fracturing and for proper study design. For 3D FEA analysis, it is essential to include a preload condition to the screws in order to replicate the aforementioned. Modeling the preload will help achieve a more realistic biomechanical model, creating an initial stress between the FDP components.[29] The preload is responsible for the stability of the prosthetic system. The optimum preload should not exceed the yield strength of the screws and was reported to be ideally 75% of the yield strength. When the external forces overcome the preload, the deformation and loosening of the screw begin, ultimately resulting in prosthetic complications and failure.[30]

Long-term success depends on the stress distribution over the entire implant-restorative assembly. It is difficult to evaluate these stress distributions clinically. In vitro biomechanical tests have provided evidence of material fatigue, wear, and deformation. However, both clinical and in vitro tests are costly, time consuming, and have limitations in controlling certain variables. The use of 3D FEA is most suitable to evaluate the complex geometries of dental implants and restorative components, and the alveolar bone. It can provide additional information that can support and enhance our understanding of the mechanisms of implant complications.

This study is based on numerical settings under ideal testing conditions. Additional studies are needed for comparisons, as well as validation with in vitro mechanical testing. Future studies should evaluate both static and cyclic forces in occlusal and oblique directions. Another limitation is that this study evaluated only one implant design and components. Expanding this to include other designs and components would be beneficial and could provide further insights into the mechanisms of implant complications and failures.

Conclusion

Different combinations of engaging and nonengaging abutment designs resulted in different stress patterns with the implants and restorative components. The dual engaging abutment design had the best stress distribution, followed by the hemi-engaging (mNE-pmE) abutment design where the engaging abutment was placed in the more anterior implant position. Nonengaging abutments resulted in higher stress areas at the implant platform and the prosthetic screws, which exceeded the yield strength of titanium alloy.

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