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RESEARCH ARTICLE

The evalution of stress distributions in 3 and 5 unit dental and implant supported fixed zirconia restorations: finite element analysis

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ABSTRACT

Purpose: In this study, it is aimed to compare the distribution of stress on teeth and implants in 3-and-5-unit-dental and implant supported zirconia restorations by using finite element analysis.

Material and Method: Stress distribution formed in teeth and implants as a result of chewing forces was analyzed in dental and implant (Astra Tech Microthread Osseo Speed, Sweeden) supported models of zirconia restoration with 5-unit placed on the numbers of 43, 44, 45, 46 and 47 and with 3-unit placed on the number of 45, 46 and 47. The study was performed through static nonlinear analysis with the three-dimensional finite element analysis method.

Results: The highest and the lowest stress were respectively found on the number of 45 and 47 in 3-unit tooth supported model. The highest and the lowest stress in 5-unit tooth supported model were respectively found on the tooth of number 45 and on the root apex of the implant of number 43. Stress accumulation was observed in the cervical portion of the implant in implant-supported models. Stress accumulation in the tooth-supported model was found less than in implant-supported models

Conclusion: The extreme forces on the dental and implant-supported restorations with increased units can reduce survival rate of restorations in mouth. In posterior restorations increased in the number of supported teeth and implant can reduce the destructive forces on teeth and implant and may allow longer period retention of the restorations in the mouth.

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INTRODUCTION

Since the osseointegration was defined as the directly structural and functional connection, without having a fibrous tissue between the living bone tissue and implant surface under loading in1960s, the dental implant-supported prosthesis have been scientifically accepted and a commen treatment choice in the case reconstructing of partial or total tooth loss.[1,2]

Full ceramic restorations have been developed instead of metal ceramic restorations owing to disadvantages aesthetically and biologically in prosthetic treatment of tooth loss.[3,4] Full ceramic restorations are also used in the construction of large restorations on posterior region with the development of high-resistant oxide ceramics. Especially zirconia (zirconiumdioksit) began to be commonly used nowadays among full ceramic restorations.[5,6]

Zirconia gathers almost all the advantages of dental materials in one single material. These excellent durability and elasticity values allow the upper structures to be done very precisely.[7,8] In addition, the surface of zirconia is very clean because zirconia (unlike metal alloys) does not have any static load. This enables the making of upper structures with an optimal entrance way, with less plaque accumulation and plaque retension force.[7,8] Furthermore, zirconia oxide ceramic is a material that has proven itself in medical technology.[7,8] Zirconia is similar to metals in terms of the mechanical properties and to the teeth in terms of color characteristics.[7,8] When used the appropriate connector in accordance with the studies, zirconia restorations are expected to be successful for a long time in the mouth.[7,8]

Although numerous advantages of zirconia oxyde, considerable amount of work has been devoted to the characterization of a less appealing characteristic of zirconia: its susceptibility to low temperature degradation.[9] This phenomenon was first reported by Kobayashi et. al.[10] and some other authors mentioned about this situation.[11-13] A high operating temperature of ≈1000◦C is necessary for zirconia to achieve the high level of ionic conductivity required for efficient operation.[14] Full stabilization is purposefully not achieved in the yitria-stabilized tetragonal zirconia polycrystals material.[9,15] For zirconia, spontaneous phase changes can occur in crystalline structure. This occurs at higher temperatures. This situation limits mass transport and thus raises the temperature when the critical particle size is reached.[16,17]

Finite Elements Analysis Method (FEAM) is a numerical method used to analyze the stress and deformations occuring in the structure of a geometric model. Not only is FEAM used to evaluate and analyze root-formed implants and the forces from the bone implant interface, but also it took its place in dental technology as a method for evaluation of the various clinical situations and prosthetic options. This method aims to solve the complicated problems through mathematical methods by seperating them into interrelated simpler and small structures.[18]

The aim of this study, in implant and teeth-supported designs placed into the mandibular region, is to evaluate stress values of fixed partial prosthesis with 3 and 5-unit zirconia occuring onto teeth and implants by the three dimensional FEAM.

MATERIAL AND METHOD

In separate tooth-supported and implant-supported designs in our study, we examined the stress distribution and values of chewing forces that occurred on teeth and implant in 3-unit zirconia restoration in areas 45-47 and in 5-unit zirconia restoration in areas 43-47.

The research was conducted by a three-dimensional (3D) finite elements stress analysis method and by static nonlinear analysis. By specifying boundary conditions for this method, four tooth-supported and implant-supported models involving areas 45 to 47 and only the areas 43–47 were used (Figure 1a-d).

Modeling

Teeth 43, 45, and 47 were used in our research. For this purpose, the front, side, upper, and lower images of the related tooth from the Wheeler atlas were obtained. The same atlas was also used for the tooth sizes.[19] The tooth was modeled, based on these images by using Rhinoceros software (USA), and scaled. In this way, an anatomically realistic tooth model was constructed (Figure 2a).

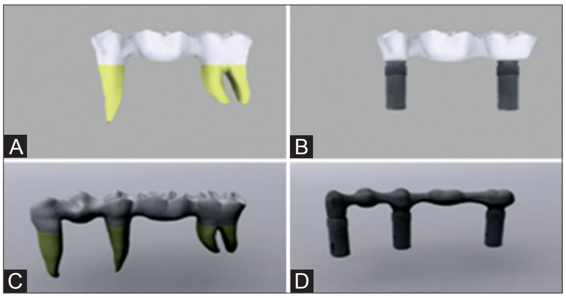

Figure 1. (a) 3 unit tooth-supported model (model 1). (b) 3 unit implant-supported model (Model 2). (c) unit tooth-supported model (model 3). (d) 3 unit implant-supported model (Model 4)

After modeling the teeth, the bone tissue surrounding the teeth models in at least 1 cm thickness with stress analysis was initiated. For this purpose, a 40 mm × 30 mm × 20 mm bone was modeled. A 2 mm-thick cortical bone was constructed in the bone by using the offset method. The interior surface in the cortical bone was defined as spongious bone. After characterizing cortical and spongious bone, the modeled teeth were extracted from the bone tissues by using a boolean method while the teeth were in their original positions. While these procedures were followed, the boolean method was not used for teeth 44 and 46. In this way, missing teeth were created for teeth 44 and 46. Because of this stage, the structure of the teeth in the bones was modeled (Figure 2b). The thickness of the periodontal ligament surrounding the teeth was defined as 100 μ. The the preparatory procedure was defined as 2 mm from the occlusal surface in teeth 43,45, and 47. It ended in the shape of a knife edge in the margins and was performed by using a computer program (Figure 2c). By using the upper structure of the accomplished digital preparation, a cement layer of 100 μ thickness was modeled with a shell method.[20]

The zirconia framework was modeled by using the upper part of the cement layer. The connector area was created in the size of 3 × 3 × 3 mm.

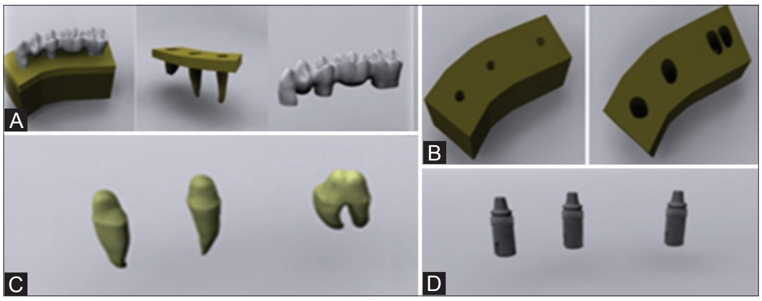

Figure 2. (a) The modeling of the bone and teeth. (b) Tooth and implant slots opened on computer aided designed bone models. (c) Preparations are modeled in computer. (d) Implants are modeled in computer

The veneer layer was modeled by extracting the zirconia lower model with the boolean method from the crown part that was created with the cutting of the tooth in the preparation border.

The implant–implant-supported model in the study was obtained by locating a 4.0 × 13-mm dental implant (Astra Tech Microthread OsseoSpeed 4.0,Sweden) to area 47 and locating a 4.0 × 11-mm dental implant (Astra Tech Microthread OsseoSpeed 4.0) to areas 43, 45, and 47 (Figure 2d).

Implants that were used in our study were scanned in 3D by the "Next Engine" scanner (USA, Santa Monica, CA) set in the macro mode. The point cloud that was acquired was saved in ".stl" format. The documents that were saved in this format were opened in Rhinoceros software and the adjustification of the implants with other sets were ensured. Modelings that were performed in Rhinoceros software were transferred to Fempro software (USA) by preserving the 3D coordinates.

The models that were obtained were converted to the concrete model as bricks and tetrahedra elements. Elements with 8 nodes were used in the bricks and tetrahedra concrete modeling system to the extent that they could be created in the Fempro model. The abutment and implant angle was determined as 0° in all created models.

Before analysis, the contact fixation was defined in the mergence sections of implant-abutment-screw mergence in the model. The friction coefficient in these areas was calculated as 0.5.[21]

The structure of the periodontal ligament in this study was accepted as nonlinear and was analyzed by using the following formula: $\sigma = 1.498246 \times 10^{-2}\varepsilon^3$ where, σ is stress and ε is strain.

The thickness of the cortical bone (from which the modeling was made) was prepared at 2 mm. It was homogenous throughout. As with all other materials, the cortical and trabecular bone linear structures were regarded as homogenous and isotropic materials. In our study, we acknowledge that the implant was fully combined with bone and no other material on the surface of the bone implant was defined.[22]

Material Features

The features of the material significantly affect the stress and strain distribution inside the structure that will be used. Homogenous, linear, and elastic forms of the materials that are used in finite elements analysis are characterized with two material fixations: the elasticity module (i.e., Young's module) and the Poisson ratio. Table 1 shows the Poisson ratio and elasticity module values of the materials that were used in our study.

Boundary Conditions

The models we obtained were fixed in such a way that it would have 0 movement ability in every degree of freedom (DOF) from the lower and side areas of the cortical bone and the trabecular bone.

Table 1. Poisson ratio and elasticity modulus values of materials used in the study

Material	Poisson Ratio	Elasticity Modulus (MPa)
Cortical Bone	0,3	13700
Spongious Bone	0,3	1850
Titanium	0,35	117000
Dentin	0,31	14700
Periodontal ligament	0,45	171
Zirconia	0,35	200000
Ceramic	0,19	60000
Zincphosphate cement	0,35	13700

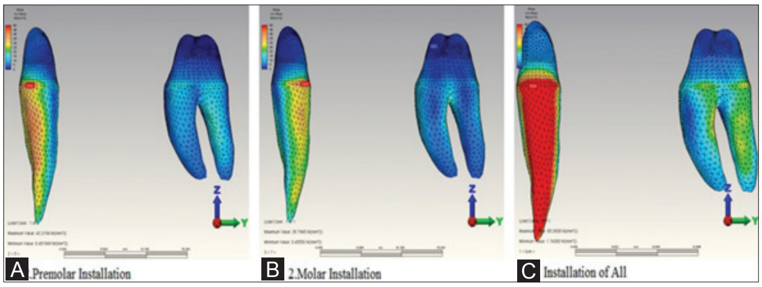

Figure 3. (a-c) Respectively 2. premolar installation, 2. molar installation and installation of all

Loading Conditions

Non-linear static analysis in 4 different loading conditions was used on three dimensional concrete models that we prepared.

- Loading condition 1:
  150 N force was applied in 30 degree from buccal inclination of number 43 restorations' buccal cusp to far axis.[23,24,25]
- Loading condition 2:
  200 N force was applied in 30 degree from buccal inclination of number 45 restorations' buccal cusp to far axis.[23,24,25]
- Loading condition 3:
  300 N force was applied in 30 degree from buccal inclination of number 47 restorations' buccal cusp to far axis.[23,24,25]
- Loading condition 4:
  150 N force over number 43 restoration, 200 N force over number 44 restoration, and 200 N from the each of the rest of the units, as a result, 950 N force was applied in 30 degree from buccal inclination of numbers 43, 44, 45, 46, and 47 restorations' buccal cusp to far axis.[23,24,25]

Implementation of Measurements:

In our study, loading was applied to 3-unit and 5-unit teeth and implant-supported zirconium-based fixed bridge prostheses. Only the highest von Mises stress values from the teeth and implants as part of the findings resulting from loading were compared by and among themselves. The examination was performed to determine which condition and which location had the greatest stress values. The "X" coordinate system in the Figure 9 is expressed as "lingual"; the "Y" coordinate system, as "distal"; and the "Z" coordinate system, as the "maxillary upper structure."

Statistical analyses could not be performed because the values obtained by using finite elements stress analysis resulted from nonvariational mathematical calculations. The purpose was to carefully examine and interpret the values and stress distributions obtained from the analyses.[18]

RESULTS

In Model 1, when the loading was done; through the 2nd premolar restoration, the highest stress value was measured in mesio lingual section of 2nd premolar tooth's neck, through the 2nd molar restoration, the highest stress value was measured in distio lingual section of 2nd premolar tooth's neck, through the all parts of restoration, the highest stress value was measured again in lingual section of 2nd premolar tooth's neck. Minimum stress values are observed in 2nd molar tooth (Figure 3 a,b,c).

As seen in Table 2, the highest stress, with regard to the stress applied to the teeth, was on the second premolar tooth. The other remarkable point is that the second molar tooth had the minimum stress values, as indicated by the pictures. (Table 2).

In Model 2, when the loading was done through the 2nd premolar restoration,

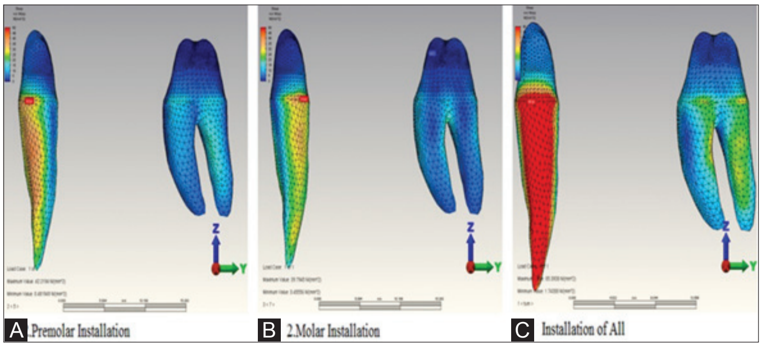

Figure 4. (a-c) Respectively and implatnt of 2.premolar installation, the implant of 2. molar installation and installation of all

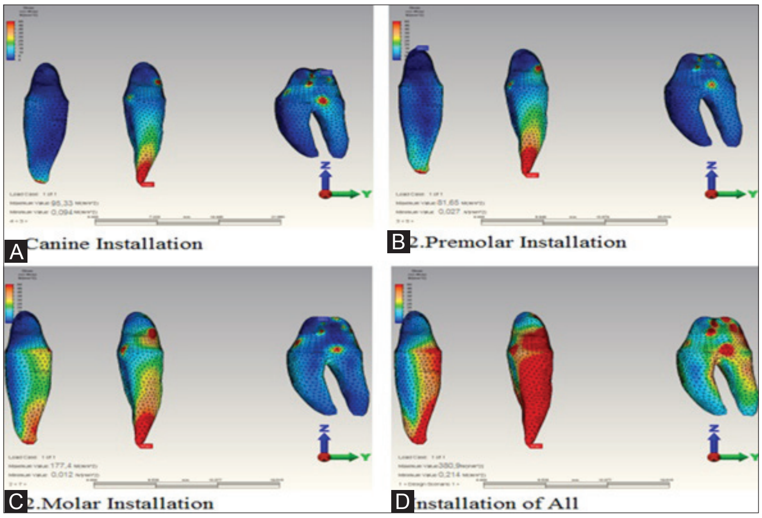

Figure 5. (a-d) Respectively canine installation, 2.premolar installation, 2. molar installation installation of all.

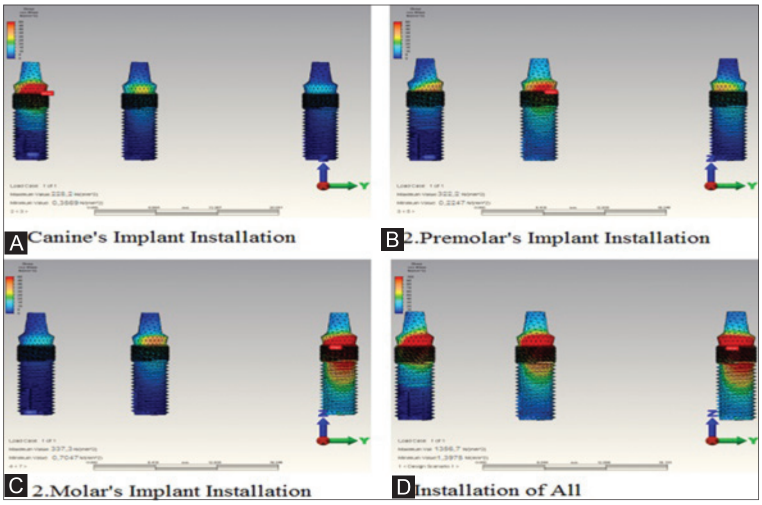

Figure 6. (a-d) Respectively the implatnt canine installation, the implant of 2. premolar installation, the implant of 2.molar installation and installation of all.

the highest stress value was measured in 2nd premolar implant's neck, and the lowest stress value was observed in root apex of 2nd molar implant,through the 2nd premolar

Table 2. Von Mises stress value in model 1

Model 1	Maximum value	Minimum value
2.premolar installation	42.21 MPa	0.48 MPa
	On the neck of the root of 2.premolar	On the root of 2.molar
2.molar installation	39,79 MPa	0.45 MPa
	On the neck of the root of 2.premolar	On the 2.molar
Installation of all	85,39 MPa	1,74 MPa
	On the neck of the root of 2.premolar	On the 2.molar

Table 3. Von Mises stress value in model 2

Model 2	Maximum value	Minimum value
2.premolar's implant installation	156,20 MPa	0.31 MPa
	On the neck of the implant of 2.premolar	On the root apex of the implant of 2.molar
2.molar' implant installation	198,10 MPa	0.29 MPa
	On the neck of the implant of 2.molar	On the root apex of the implant of 2.premolar
Installation of all	389,66 MPa	4,96 MPa
	On the neck of the implant of 2.molar	On the root apex of the implant of 2.premolar

restoration, the highest stress value was measured in 2nd molar implant's neck, and the lowest stress value was observed in root apex of 2nd premolar implant, through the all parts of restoration, the highest stress value was measured in 2nd molar implant's neck, and the lowest stress value was observed in root apex of 2nd premolar implant (Figure 4 a,b,c).

It is seen that the highest stress values in Table 3 are much more compared to Table 2. While the highest stress values in Table 2 are observed in 2nd premolar tooth, in Table 3 it is observed in the support where force was applied. Minimum stress values have been observed in 2nd molar tooth (Table 3).

After all the loadings that are done in Model 3, the highest stress values are measured in root apex section of 2nd premolar tooth.

Minimum stress values are observed in crown sections of canine and 2nd molar teeth (Figure 5 a,b,c,d).

As seen in Table 4, in terms of the stress on support teeth, the most stress significantly comes to 2nd premolar tooth (Table 4).

The highest stress values in Model 4 was measured in neck sections of the implant which belongs to the restoration that the loading was done.

Table 4. Von Mises stress value in model 3

Model 3	Maximum value	Minimum value
Canine installation	95,33 MPa	0,09 MPa
	On the root apex of 2.premolar	On the 2.molar
2.premolar installation	81,65 MPa	0,02 MPa
	On the root apex of 2.premolar	On the canine
2.molar installation	177,47 MPa	0,01 MPa
	On the root apex of 2.premolar	On the 2.molar
Installation of all	380,97 MPa	0,21 MPa
	On the root apex of 2.premolar	On the 2.molar

The lowest stress values have been seen in root apex part of canine implant (Figure 6 a,b,c,d).

It is seen that the highest stress values in Table 4 are much more compared to Table 5. While the highest stress values in Table 4 are observed in 2nd premolar implant, in Table 5 it is observed in the support implant where force was applied. Minimum stress values have been observed in the implant which was applied in canine area (Table 5).

DISCUSSION

Clinical and histomorphometric studies have shown that bone loss from the

Table 5. Von Mises stress value in model 4

Model 4	Maximum value	Minimum value
Canine's implant installation	228,24 MPa	0,35 MPa
	On the neck of the implant of canine	On the root apex of the implant of canine
2.premolar's implant installation	322,23 MPa	0,22 MPa
	On the neck of the implant of 2.premolar	On the root apex of the implant of canine
2.molar's implant installation	337,36 MPa	0,70 MPa
	On the neck of the implant of 2.molar	On the root apex of the implant of canine
Installation of all	1356,7 MPa	1,39 MPa
	On the neck of the implant of 2.molar	On the root apex of the implant of canine

implant neck is an important cause of implant loss in the post-loading period.[26-28] One possible reason is uneven stress distribution in the bone socket, thus exposing the bone implant neck to maximum stress.[28] In the present study, we used FEAM to compare the stress distributions on the teeth and implants of dental- and implant-supported fixed zirconia restorations. We laser-scanned four models and transferred the data to a computer to calculate the highest and lowest stresses on teeth, implants, and surrounding bone, and to identify where the stresses occurred. When a zirconia infrastructure was employed, more stress was evident in implant-supported than dental-supported models. In addition, the literature indicates that more stress accumulated in the neck regions of implants in all tested implant-supported models. In tooth-supported models, stress accumulation was generally greater in the root apex.[26-30]

Stress analysis of living tissues such as bones, teeth, and the periodontium (either in vivo or in vitro) is difficult, and sometimes impossible. Thus, stress analysis of living tissues is preferably conducted via computer modelling. Finite element analysis is suitable for stress analysis of structures with complex geometries.[26] In the present study, we used FEAM, in line with literature indications. However, mathematical models have certain drawbacks, being unable to simulate individual variations in living tissues. Such models can be used only to explain experimental results; any predictive power is comparative only.[31,32]

In many works on dental implants and prosthetic options, changes in implants and surrounding bones are generally studied by calculation of the Von Misses stress values; thus, we calculated such values in this work.[26,33] Ismail et al.[34] performed two- and three-dimensional finite element analyses of blade implants, and showed that two-dimensional analysis failed to reflect normal stress distributions in detail, being adequate only to explore principal stress distributions. Thus, we used a three-dimensional finite element method to obtain more realistic results and to create more realistic models.

We used non-linear contact analysis to reflect implant-abutment connections in real life.[35,36] A non-linear FEA was employed to predict the behaviours of structures loaded beyond the elastic limit of the material of interest. Each such structure experiences plastic deformation and does not subsequently regain its original status or shape. Stress is the development of resistance against an external force. Stress occurring as an "inside" reaction to such force is equal in intensity to the applied force, but the vector lies in the opposite direction. Both the force applied and the resistance developed spread over the entire volume of the object. Thus, stress is defined as the force applied per unit area.[37]

Meijer et al.[38] found that three-dimensional modelling of a particular region of interest was adequate; it was not necessary to model the entire lower jaw. Thus, we did not model the entire mandible.

The stresses on implant-supported models were considerably higher than those on tooth-supported models (Model 1,2,3,4).

The highest stresses in implant-supported models were typically located in the cervical portions of implants, in agreement with the literature, which also indicates that forces on implants are compensated by cortical bone.[26,27,29] Also, dental implants are more likely to be subjected to excessive occlusal forces than are natural teeth, caused by the lack (in the former reconstructions) of important regions of the periodontium, such as the shock-absorbing periodontal ligament, which is sensitive to touch and affords proprioceptive feedback, as do the natural teeth of dental implants.[29] We consider that this explains the lower accumulation of stress in dental-supported than implant-supported prostheses, as indicated in the literature.

Finite element analysis studies on titanium implants showed that stress developed in implant neck regions. The stresses on implant-supported models were considerably higher than those on tooth-supported models, consistent with the literature.[26,29] High stresses were evident around the root tips of tooth-supported models but rather in the cervical regions of implants in implant-supported models. Thus, stress spread through the root end because the periodontal ligament was present, consistent with literature data on the differences between teeth and implants.[29,30] In our present study, stress from the cortical bone was transferred to the neck of the implant. Stress on implants under vertical load is both high and variable; is delivered to the head and neck of an implant; and is compressive in nature. Compared to forces generated in cancellous and cortical bone, forces triggered by loading were higher in the cortical bone of all tested implants. Thus, cortical bone absorbs most of the force because the bone has a high elastic modulus. Cancellous bone compensates for the force because such bone has a low elastic modulus.[28,39] This emphasises the need for some cancellous bone within cortical bone.

Rangert et al.[40] studied implant fractures and found that 90% were in the posterior regions of prostheses supported by one or two implants. High occlusal force may both cause implant loss (leading to crestal bone loss) and abutment and/or implant loss (leading to loosening of the abutment and/or a screw connection).[22,41] Implant neck fractures should also be considered if implant loss occurs after loading. Nobel Biocare implant-supported single molars suffered an implant fracture rate of 14% in the posterior mandibular region.[14] The long-term success of dental implants is negatively affected by bacterial infections. Also, it is necessary to protect the quality of surrounding bone and to not overload the bone/ biomaterial interface.[2,22,42] In our present work, the highest loading stresses on all implant-supported models occurred in the cervical regions of 47 implants, consistent with the literature. Our data support those of Rangert et al., who found high fracture rates in implants in the molar region.[40] We found that the stress on abutted teeth in five-unit fixed partial dentures was significantly greater than that on three-unit dentures in our tooth-supported prosthetic model. This may indicate that a higher number of abutted teeth can increase stress accumulation from masticatory forces, caused by inclusion of the canine tooth in the restoration (in addition to posterior teeth). For example, the use of a tooth that has received endodontic treatment as a fixed partial denture in the posterior region may reduce the survival rate of a prosthesis.[43] In addition, in a prosthesis featuring an implant-supported fixed partial denture, the stress on abutted implants increases as the number of such units increases, as seen when prostheses with tooth-supported fixed partial dentures are placed. Our data are consistent with those of studies indicating that increased implant support decreases the stress on such implants.[44,45,46] Many researchers have suggested that a reduction in the number of implants supporting a prosthesis increases the distribution of forces developing in the implant and surrounding tissues.[44,45,46] From this viewpoint, then, the numbers of abutment implants may affect the long-term life of an implant-supported prosthesis in the posterior region.

This study had several limitations:

We evaluated zirconia only.

Only one of our implant models was analysed at two different lengths. Implants differing in length and diameter were generally not evaluated.

Only two prosthetic options were evaluated (in four different ways); inclusion prosthetic options would have been better.

The finite element model that we used does not reproduce all of the important features of living tissue. In vivo studies should yield more accurate results. Magnetic resonance imaging to obtain geometric information is desirable; tissue force conductivities could be calculated.[47]

CONCLUSION

Within the limitations of the present study, stress on dental implants reduces long-term implant survival and that stress is concentrated in the implant neck. In addition, increases in the numbers of supported teeth, and implants, can compensate for such destructive forces and may allow longer-term survival of dental restorations.

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REFERENCES

1. Branemark PI, Adell R, Breine U, Hansson BO, Lindstrom J, Ohlsson A. Intra-osseous anchorage of dental prostheses. I. Experimental studies. Scand J Plast Reconstr Surg 1969;3:81-100.

2. Buser D, Mericske-Stern R, Bernard JP, Behneke A, Behneke N, Hirt HP, et al. Long-term evaluation of non-submerged ITI implants. Part 1: 8-year life table analysis of a prospective multi-center study with 2359 implants. Clin Oral Implants Res 1997;8:161-172.

3. Christensen GJ. Ceramics vs. porcelain fused to metal crowns: give your patients a choise. J Am Dent Assoc 1994;125:311-312,314.

4. Hansen PA, West LA. Allergic reaction following insertion of a Pd-Cu-Au fixed partial denture: a clinical report. J Prosthodont 1997;6:144-148.

5. Vult von Steyern P, Carlson P, Nilner K. All-ceramic fixed partial dentures designed according to the DC-Zirkon technique. A 2-year clinical study. J Oral Rehabil 2005;32:180-187.

6. Piwowarczyk A, Ottl P, Lauer HC, Kuretzky T. A clinical report and overview of scientific studies and clinical procedures conducted on the 3M ESPE Lava All-Ceramic System. J Prosthodont 2005;14:39-45.

7. Studart AR, Filser F, Kocher P, Luthy H, Gauckler LJ. Cyclic fatigue in water of veneer-framework composites for all-ceramic dental bridges. Dent Mater 2007;23:177-185.

8. Rismanchian M, Shafiei S, Nourbakhshian F, Davoudi A. Flexural strengths of implant-supported zirconia based bridges in posterior regions. J Adv Prosthodont 2014;6:346-350

9. Kelly JR, Denry I. Stabilized zirconia as a structural ceramic: an overview. Dent Mater 2008; 24(3): 289-298.

10. Kobayashi K, Kuwajima H, Masaki T. Phase change and mechanical properties of ZrO_2-Y_2O_3 solid electrolyte after aging. Solid State Ionics 1981; 3(4):489–495.

11. Sato TMS. Transformation of yttria-doped tetragonal ZrO2 polycrystals by annealing in water. J Am Ceram Soc 1985; 68(6): 356–359.

12. Sato T, Shimada M. Crystalline phase-change in yttria-partially stabilized zirconia by low-temperature annealing. J Am Ceram Soc 1984; 67(10): 212–223.

13. Cales B, Stefani Y, Lilley E. Long-term in vivo and in vitro aging of a zirconia ceramic used in orthopaedy. J Biomed Mater Res 1994; 28: 619–624.

14. Luo J, Ball Richard J, Stevens R. Gadolinia doped ceria/yttria stabilised zirconia electrolytes for solid oxide fuel cell applications. Journal of materials science 2004; 39(1): 235-240.

15. Heuer AH. Transformation toughening in ZrO2-containing ceramics. J Am Ceram Soc 1987; 70(10): 689–698.

16. Reidy DJ, Holmes JD, Morris MA. Preparation of a highly thermally stable titania anatase phase by addition of mixed zirconia and silica dopants. Ceramics international 2006; 32(3): 235-239.

17. Kingery WD, Bowen HK, Uhlmann DR. Introduction to Ceramics, second ed., Wiley & Sons, New York, 1976, p. 58.

18. Petrie CS, Williams JL. Comparative evaluation of implant designs: influence of diameter, length, and taper on strains in the alveolar crest. A three-dimensional finite-element analysis. Clin Oral Implants Res 2005;16:486-494.

19. Wheeler RC. An atlas of tooth form. Toronto: Harcourt Canada; 1969.

20. Heravi F, Salari S, Tanbakuchi B, Loh S, Amiri M. Effects of crown-root angle on stress distribution in the maxillary central incisors' PDL during application of intrusive and retraction forces: a three-dimensional finite element analysis. Progress in Orthodontics 2013; 14:26.

21. Covani U, Ricci M, Tonelli P, Barone A. An Evaluation of New Designs in Implant-Abutment Connections: A Finite Element Method Assessment. Implant Dentistry 2013; 22(3):263-267.

22. Baggi L, Cappelloni I, Di Girolamo M, Maceri F, Vairo G.The influence of implant diameter and length on stress distribution of osseointegrated implants related to crestal bone geometry: a three-dimensional finite element analysis. J Prosthet Dent 2008;100(6):422-431.

23. Petrie CS, Williams JL. Comparative evaluation of implant designs: influence of diameter, length, and taper on strains in the alveolar crest. A three-dimensional finite-element analysis. Clin Oral Implants Res 2005;16:486-494.

24. İplikçioğlu H, Akça K. Comperative evaluation of the effect of diameter, length and number of implants supporting three unit fixed partial protheses on stress distribution in the bone. Journal of Dentistry 2002;30:41-46.

25. Chou I-C , Lee S-Y, Wu M-C , Sun C-W, Jiang C-P. Finite element modelling of implant designs and cortical bone thickness on stress distribution in maxillary type IV bone. Computer Methods in Biomechanics and Biomedical Engineering 2014;17(5):516-526.

26. Geng JP, Tan KB, Liu GR. Application of finite element analysis in implant dentistry: a review of the literature. J Prosthet Dent 2001;85:585-598.

27. Wiskott HW, Belser UC. Lack of integration of smooth titanium surfaces:a working hypothesis based on strains generated in the surrounding bone. Clin Oral Implants Res 1999;10:429-44.

28. Himmlova L, Dostalova T, Kacovsky A, Konvickova S. Influence of implant length and diameter on stress distribution: a finite element analysis. J Prosthet Dent 2004;91:20-25.

29. Koyano K, Esaki D. Occlusion on oral implants: current clinical guidelines. J Oral Rehabil 2014 Oct. 6 doi: 10.1111/joor.12239 (Epub ehead of print).

30. Svensson KG, Trulsson M.Impaired force control during food holding and biting in subjects with tooth- or implant-supported fixed prostheses. J Clin Periodontol. 2011;38(12):1137 1146.

31. Adams MA. Spine Update Mechanical Testing of the Spine An Appraisal of Methodology, Results, and Conclusions. Spine 1995; 20(19): 2151-2156.

32. Arkın H, Xu LX, Holmes KR. Recent developments in modeling heat transfer in blood perfused tissues. Biomedical Engineering, 1994; 41(2): 97-107.

33. Teixeira ER, Sato Y, Akagawa Y, Shindoi N. A comparative evaluation of mandibular finite element models with different lengths and elements for implant biomechanics. J Oral Rehabil 1998;25(4):299-303.

34. İsmail YH, Pahountis LN, Fleming JF. Comparison of two-dimensional and three-dimensional finite element analysis of a blade implant. Int J Oral Implantol 1987;4(2), 25-31.

35. Wakabayashi N, Ona M, Suzuki T, Igarashi Y. Nonlinear finite element analyses: advances and challenges in dental applications. J Dent 2008;36:463-471.

36. Lin CL, Wang JC. Nonlinear finite element analysis of a splinted implant with various connectors and occlusal forces. Int J Oral Maxillofac Implants 2003;18:331-340.

37. O'Brien W J. Dental materials and their selection, Quintessence Pub Co Inc 2nd Edition,1997.

38. Meijer HJ, Starmans FJ, Bosman F, Steen WH. A comparison of three finite element models of an edentulous mandible provided with implants. Journal of Oral Rehabilitation 1993;20(2):147-157.

39. 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 Mater 2012 Nov;28(11):1126-1133.

40. Rangert B, Krogh PH, Langer B, Van Roekel N. Bending overload and implant fracture: a retrospective clinical analysis. Int J Oral Maxillofac Implants 1995;10:326-334.

41. Misch CE. Contemporaray Implant Dentisrty. Mosby Elsevier 2008;Third edition:337

42. Winkler S, Morris HF, Ochi S. Implant survival to 36 months as related to length and diameter. Ann Periodontol 2000;5:22-31.

43. De Baat C, Van Loveren C, Van Der Maarel-Wierink CD, Witter DJ, Creugers NH. After care for durability and profitability of single-unit and multi-unit fixed dental prostheses. Ned Tijdschr Tandheelkd 2013;120(7-8):411-420.

44. Ogawa T, Dhaiwall S, Naert I, Mine A, Konstrom M, Sasaki K, Duyck J. Impact of implant number, distribution and prothesis material on loading on implants supporting fixed prothesis. J Oral Rehabil 2010;37:525-531

45. Sahin S, Cehreli Z, Yalcin E. The influence of functional forces on the biomechanics of implant-supported prothesis: A rewiev. J Dent 2002;30(7-8):271-282.

46. Duyck J, Van Oosterwyck H, Vander Sloten, De Cooman M, Puers R, Naert I. Magnitude and distribution of occlusal forces on oral implants supporting fixed prostheses: an in vivo study. Clin Oral Implants Res 2000;11(5):465-475.

47. Yan Y, Nunez PL, Hart RT. Finite-element model of the human head: scalp potentials due to dipole sources. Medical and Biological Engineering and Computing 1991;29(5):475-481.

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