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Comparative Evaluation of the Stresses Applied to the Bone, fixtures, and Abutments of Implant-Supported Fixed-Partial-Dentures with Different Long-Spans, After Cyclic Loading Using 3 Dimensional Finite-Element-Method

Evaluation of implant-supported-FPDs using 3D FEM

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Abstract—Objectives: The pattern of forces transmission and the stress distribution are very important in success or failure of implant-supported fixed-partial-dentures (ISFPD). The exact number of pontics between two implant abutments has always been debatable. The reliability of Ante's law for ISFPDs is also questionable. The aim of this study was to evaluate the stresses applied to the bone and abutments of an ISFPD with two implants using 3D finite element method (3D FEM).
Materials and Methods: In this study, a model with type 2 bone and two implants (Diameter: 4.1, Length: 12 mm, solid abutment, solid cover, ITI, Straumann, Switzerland) were simulated by Solidworks 2007 software. Three-unit, 4-unit, and 5-unit ISFPD models were designed in the software. Osseointegration was assumed 100% between implants and bone. For all three models, forces equivalent to 50, 100, and 150 N were respectively applied to the first premolar, the second premolar, and the first molar dynamically. The maximum Vonmisses stresses (VMS) and strain values (SV) were recorded.
Results: The maximum VMS was seen in the bones around the crestal area of the cortical part in all three models. The maximum VMS applied in 5-unit model bone were higher than thoseof two other models. The maximum VMS in the abutments and fixtures of 5-unit model were higher than those of 3-unit and 4-unit ISFPDs.
Conclusion: The VMS imposed in 5-unit ISFPD in type II bone were comparable with thoseof 3-unit and 4-unit ISFPDs. Of course, all the strains were in the bone endurance range. VMS in the abutments and fixturesof all three models were in the permanent-durability-range of the bone.

Keywords-*Finite Element Analysis, cyclic loading, Dental prosthesis, implant-supported.*

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I.INTRODUCTION

One of the most important factors in implant-treatment-success is VMS. The force distribution on the surface unit is called "mechanical stress". The rate of mechanical stress depends on two different variables, including the magnitude of force and the area where force is distributed. Stress quantity was introduced for the first time by Cushi in 1822 within the elasticity theory [1]. Internal stresses in implant systems and their surrounding-biological-tissues during applying forces have an important effect on implant-long-term-durability. In order to increase the success rate of implants, the mechanical stress should be minimized and the force distribution should be uniform on the implants and their surrounding tissues. The bone-biological-response to mechanical loads can also affect the implant-durability in the patient's mouth. Since the force is transmitted through the prosthesis to the bone and implant, accurately-designed-prosthesis plays an important role in achieving proper stress distribution around the implants [2].
Short-term or long-term stress can cause complications such as failure of implant osseointegration, crestal-bone-resorption, porcelain fracture, loss of prosthesis retention, implant-components-failure, and screw loosening. So that the stress distribution is the most important factor that should be assessed and controlled before treatment planning in order to

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reduce its harmful effects [3]. During fabrication of an ISFPD, various changes may take place in impression making, pouring the master cast, fabricating metal framework, waxing, burn out, and porcelain application. All of these changes can increase the mechanical stresses [4-7]. Nowadays, ISFPDs are frequently used to reconstruct the partially-edentulous-patients. Different factors can influence the success or failure of ISFPDs such as strain, VMS, stress-distribution-patterns, prosthesis design, the edentulous-length-span, and biomechanical properties of the used alloy [8]. A leverage component called "pontic" has an important effect during loading an ISFPD. By increasing the pontic length, the forces applied to the abutments will increase, according to the leverages laws. The distance between two implants in ISFPDs has a significant effect on the stress-distribution-pattern in the bone surrounding implants [9]. In the tooth-supported FPDs, periodontal ligaments (pdl) acts as shock absorber, i.e. during force application the teeth abutments can move apically and laterally in the pdl. There is not such mechanism in ISFPDs, so that biomechanical problems increase [10].

In ISFPDs, as the number of implants abutments increases, functional-contact-surface of bone-implant increases and the stress applied to ISFPD decreases, biomechanically [11]. However, inserting more number of implants is not always possible easily because of different factors, such as remaining bone quantity and/or quality and anatomical limitations [12]. In the literature, there is no consensus on the proper longspan between two implant abutments in order to obtain the ideal-biomechanical-support. According to Ante's law, the root surface of teeth abutments in tooth-supported FPDs should be equal or more than the root surface of the replaced teeth [13]. The major reasons of some FPDs failures were due to biomechanical overload, leverages, and torque forces. Of course, edentulous- long-span creates more leverage forces on FPD. Ante's law cannot be overgeneralized to ISFPDs, because the nature of implant abutments differ from the natural teeth which have pdl. The rate of FPD bending has direct correlation with the number and length of the pontics and reverse correlation with the occlusogingival thickness. This bending finally leads to tensile and shear forces applied to the bone and abutments which can cause more problems in implant abutments than teeth abutments due to lack of pdl [14]. According to Misch et al, the use of two implants to support a three-unit ISFPD is biomechanically appropriate for proper stress transmission. Replacing three or four pontics with two implants is normally safe, while Misch does not recommend replacement of five pontics using two implants due to increased risk of force application, probability of bone resorption, screw loosening or implant fracture [10].

There are few studies on the biomechanics of ISFPDs length span. The aim of this study was to examine and to evaluate the stresses and strains of bone and the implant components in ISFPDs with two implants abutments and one, two, and three pontics using 3D FEM. According to the null hypothesis, there was not any difference between the studied ISFPDs.

II. MATERIALS AND METHODS.

In this in-vitro study, two implants (diameter: 4.1, Length: 12, ITI, Straumann, Waldenburg, Switzerland) with two one-piece abutments (Solid abutments, length: 5.5 mm) were selected. They were measured using shadow graph and they were designed with SolidWorks-simulation-software 2009. The precise dimensions of various parts of the mentioned implants and abutments were determined with toolmakers microscope (OMT tool makers, Melton Mowbray, Leicestershire, UK). The microscope precision was 0.005mm and the precision of RPP-50 Floor-Profile-Projector (AMBALA CANTT, KOLKATA, INDIA) was 0.01mm. These data were matched with the information published by the manufacturer which ensured the accuracy of the measurements. After measuring the components, they were plotted in the form of two-dimensional-maps in Autocad software.

TABLE I.

Characteristics of the studied materials

Components	*Materials*	*Young's modulus*	*Poisson's ratio*	*Yield strength (MPa)*
Implant	Ti-6Al-4V	105	0.33	800
Abutment	Ti-6Al-4V	105	0.33	800
Framework	Cr-Co-alloy	220	0.30	720
Porcelain	Feldespatic	61.2	0.19	500

In order to bone modeling, the maxillary bone was designed in the form of a trapezoid-shaped-block. In the designed model, the occlusogingival length was 15mm and the buccopalatal width was 7.1mm and there was 1.5mm bone in the buccal and palatal sides of the implant. The bone was type II and it had both cortical bone and spongious bone. The thickness of cortical bone was 2mm which was covered with spongious bone.

In order to simulate the ISFPD model, scanned teeth (OrthoTac, Ivoclear, lichtenstine) were used. These teeth were placed on a spongy model similar to maxillary bone and a 3D CBCT was taken. The images and information resulted from this scan were used for modeling and simulating the metal-ceramic ISFPD. CT images were entered to Mimics software and a 3D model was created in this software. After performing the final operations on the model such as smoothing, the 3D model was entered to FEM software. In this 3D FEM study, three models of ISFPD frameworks were designed. In all three models, one-piece abutments (length: 5.5mm) were used. These Three models included: (1) a three-unit ISFPD model with the first premolar and the first molar as the abutments and the second premolar as the pontic, (2) a four-unit ISFPD model with the first premolar and the second molar as the abutments and the second premolar and first molar as the pontics, and (3)a five-unit ISFPD model with the canine and the second molar as the abutments and the first premolar, the second premolar, and the first molar as the pontics.

Then, the models were assembled with implants and abutments using the Abaqus software. After determining the mechanical properties of the studied components, they were assembled in the software. In this study, the fixture was placed in the bone in a way that the fixture collar was outside of the bone. Then, the one-piece abutments were tightened on the fixtures with 35Ncm torque. The superstructures were placed on the abutments. The fixture- abutment-attachment was determined in the form of Lagrangian-contact-attachment with friction coefficient equal to 0.3 based on the previous studies. The attachments between framework with abutment, framework with porcelain, and bone with fixture were

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considered fixed and bonded. The osseointegration between bone and implant was assumed 100%.

In this 3D FEM study, dynamic loadings were used in all three models. In dynamic loading 50, 100, and 150N forces were applied on the first premolar, the second premolar, and the first molar, respectively. The forces were vertically applied on the teeth based on the cusp-fossa-occlusal-plane. The dynamic forces began from zero and reached to the maximum value. After 300000 cycles of periodic loading, the stresses applied to bone, implant, and abutments were examined in all three groups. The forces were surveyed using Abaqus software. Then the stress and strain patterns applied to the bone, implant, and abutment were evaluated.

The VMS accumulation places and stress patterns in the implants, abutments, and bone of three-unit ISFPD after dynamic loading in the frontal section are shown in figures 1 to 4.

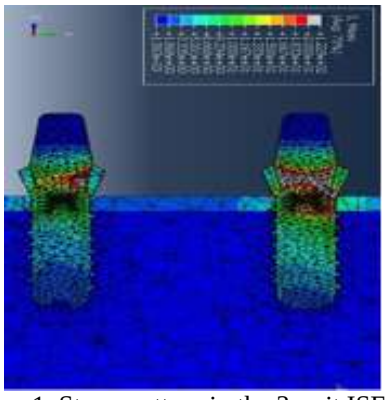

Figure 1. Stress pattern in the 3-unit ISFPD.

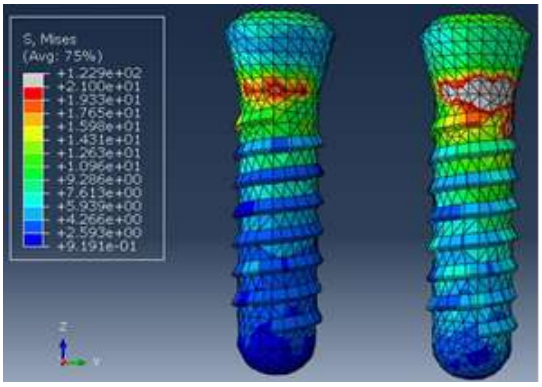

Figure 2. Stress pattern in the implants of 3-unit ISFPD.

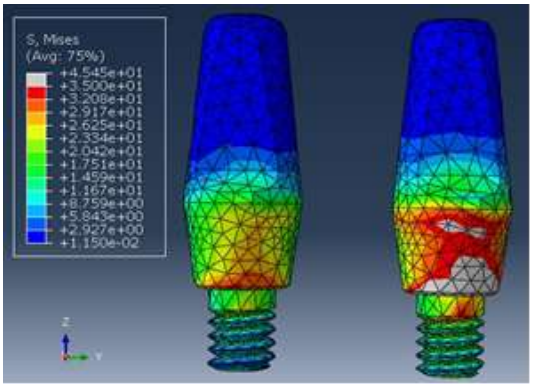

Figure 3. Stress pattern in the abutments of 3-unit ISFPD.

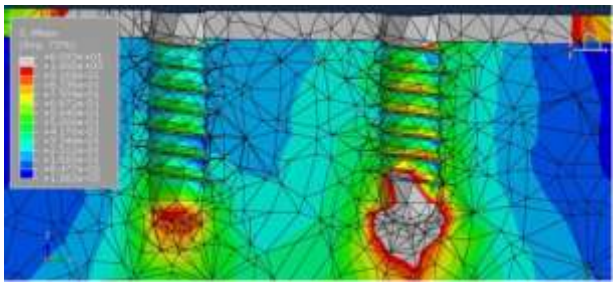

Figure 4. Stress-accumulation-places in the bone around the implants of 3-unit ISFPD.

The VMS accumulation places and stress patterns in the implants, abutments, and bone of 4-unit ISFPD after dynamic loading in the frontal section are shown in figures 5 to 8b.

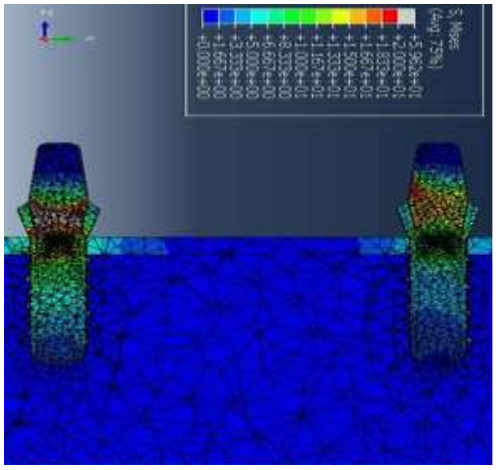

Figure 5. Stress pattern in 4-unit ISFPD

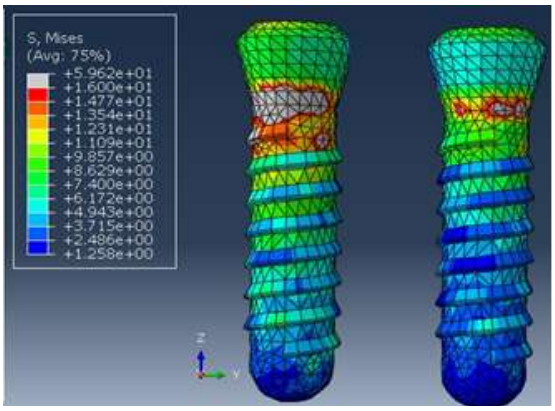

Figure 6.. Stress-accumulation-places in implants of 4-unit ISFPD.

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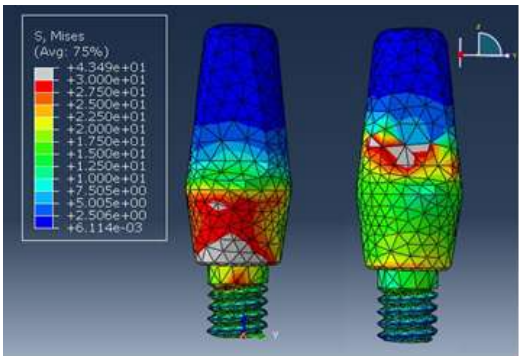
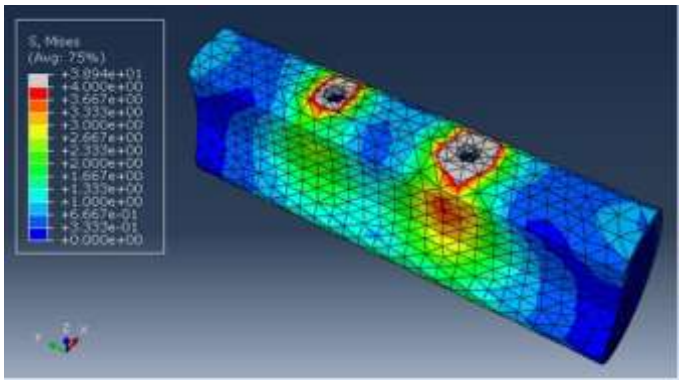
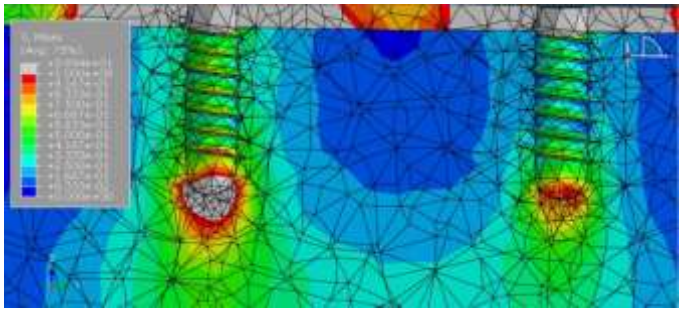


Figure 7.. Stress-accumulation-places in the abutments of 4-unit ISFPD.



(8 a)



(8 b)

Figure 8 (a,b)- Stress-accumulation-places in the bone of 4-unit ISFPD

The VMS accumulation places and stress patterns in the implants, abutments, and bone of 4-unit ISFPD after dynamic loading in the frontal section are shown in figures 9 to 12b.

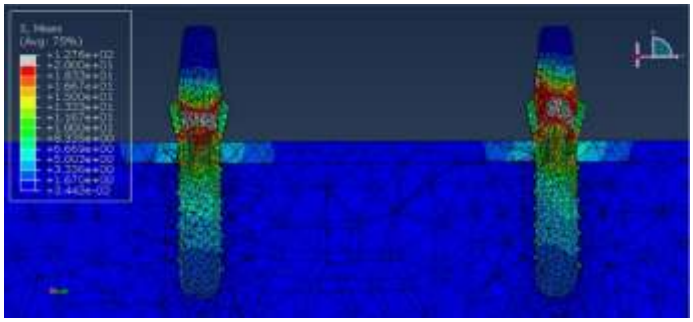


Figure 9.- Stress-accumulation-places in five-unit ISFPD after dynamic loading.

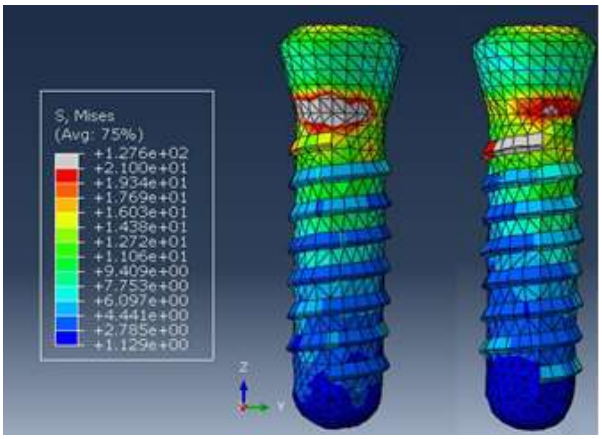


Figure 10- Stress-accumulation-places on the abutments of five-unit ISFPD

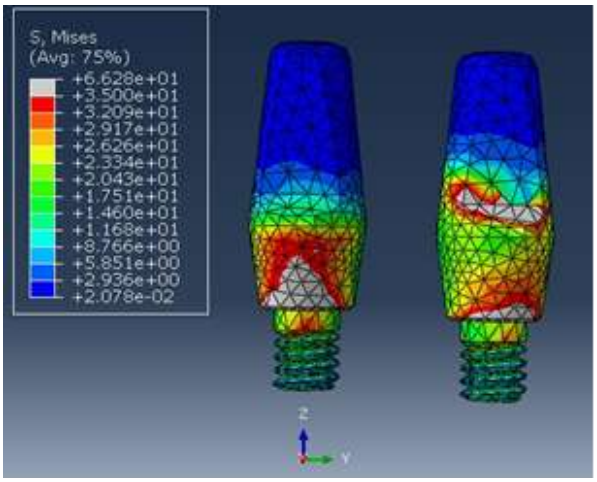


Figure 11- Stress-accumulation-places on the abutments of five-unit ISFPD

The location of maximum VMS concentration in the studied models

	Three-unit model	*Four-unit model*	*Five-unit model*
Abutment	In the shank and lower cone of the posterior abutment	In the shank and cone of the anterior abutment	In the shank and the cone of anterior abutment
Implant	On buccal surface of the first thread of posterior-abutment-implant-connection	On the first thread of the anterior-abutment-implant-connection	On the first thread of the anterior-abutment-implant-connection
Bone	The crestal area of cortical bone of posterior implant	The crestal area of cortical bone of anterior implant	The crestal area of cortical bone of anterior implant

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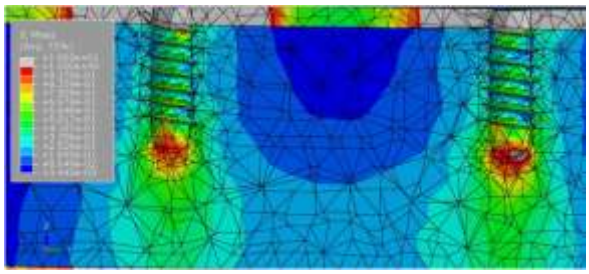
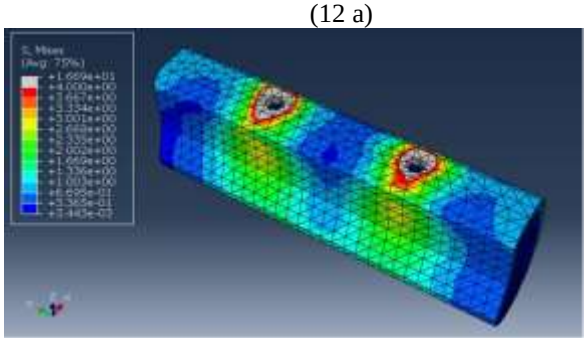


(12 a)



(12 b)

Figure 12 (a,b)- Stress-accumulation-places on the bone of ISFPD.

III. RESULTS

The results of this 3D FEM study are expressed in the following self-explanatory tables.
The fatigue analysis was performed using Goodman, Soderberg, and Gerber fatigue-theories according to the following formulae shown in table II.

TABLE II.

TABLE II.

Numerical values in three fatigue theories of Saderberg, Goodman, and Gerber

		Gerber	Goodman	Soderberg
Three-unit ISFPD	implant	5.96	5.3	5.01
	Abutment	4.23	3.76	3.55
Four-unit ISFPD	implant	4.53	4.02	3.8
	Abutment	4.7	4.18	3.95
Five-unit ISFPD	implant	3.48	3.41	3.23
	Abutment	4.07	3.62	3.42

Where Sy, Su, and Se represent for Yield stress, ultimate stress, and endurance stress, respectively. Finally, the results were
extracted from the software in the form of the maximum VMS, SV.
The maximum values of VMS in three studied models after dynamic loading are shown in Table III.

Table III.

TABLE IV.

The mean and maximum values of strain in the bone of three studied models.

	Three-unit model	*Four-unit model*	*Five-unit model*
	Maximum strain in the bone		
Strain ×10-3	1.58	1.66	1.94
	Strain values in the ISFPD		
Strain ×10-3	1.58	1.66	1.94

TABLE V.

Maximum values of VMS after dynamic loading.

		Three-unit ISFPD model	*Four-unit ISFPD model*	*Five–unit ISFPD model*
Abutment	***Anterior implant***	63.78	57.38	66.28
	posterior implant	44.77	37.97	61.83
Fixture	***Anterior implant***	68.33	59.61	70.51
	posterior implant	50.78	45.30	66.88
Bone	***Anterior implant***	38.71	20.78	16.68
	posterior implant	29.57	15.52	12.40

Table 6 shows the numerical values obtained in each of three fatigue theories of Saderberg, Goodman, and Gerber for implants and abutments in the three-unit, four-unit, and five-unit ISFPDs. As shown in the table, numerical values obtained in each of the three models were above 1, which indicates the abutments and implants will never fracture, if there is not any structural problem such as cracking, etc.

Table VI.

Fatigue formulae	**Fatigue theories**
$\left(\frac{\sigma a}{Se}\right)+\left(\frac{\sigma m}{Su}\right)=\frac{1}{N}$	***Goodman***
$\left(\frac{\sigma a}{Se}\right)+\left(\frac{\sigma m}{Sy}\right)=\frac{1}{N}$	***Soderberg***
$\left(\frac{N\sigma a}{Se}\right)+\left(\frac{N\sigma m}{Sy}\right)=1$	***Gerber***

IV. DISCUSSION

Although many studies have been performed in the field of dental implants, but all parameters and complications in the failure and success have not been completely specified so far. Several studies have examined the influence of bone and abutments stresses on the success of the implant-supported prosthesis treatment. All implant systems are designed to tolerate the mechanical forces and to have appropriate

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endurance. In addition, complete fitting of the dental-implant-system is too important, because it reduces the stress on the framework, implant, and the surrounding bone [1, 2]. Abduo J. et al have shown that the bone-implant-interface plays a key role in the amount of bone and surrounding tissues stresses. Osseointegration is one of the most important factors in implant success [15]. If the stress at the bone-implant-interface minimizes, the implant-survival-rate will increase. Christensen GL. et al have shown that the surface conditions, implant structure, and implant design may affect osseointegration. Because the force applied to implant system is finally transmitted to the bone and it has significant impact on bone remodeling [16]. Ponkaj N. et al expressed that if the stress applied to the bone is resulted from functional and non-functional forces, inflammation, misfitting, and implant design, the osseointegration may improve in short-term which is due to the disruption in physiologic activity of bone tissues. However, in long-term the mentioned stress may lead to the increased stress and inflammation in bone tissues which may cause bone resorption and may stop bone-healing-process [17]. The importance of biomechanical factors cannot be overemphasized in implant success [16]. Among these biomechanical factors, the stress and stress distribution is too important. During load application, the in-vivo internal-stresses resulted from the implant system and its surrounding-biological-tissues have great influences on the long-term-implants-survival-rate. In order to increase implant success, the goal of treatment planning should be to minimize the mechanical stress and to uniformly distribute the mechanical stresses on the implant and the osseointegrated bone. In addition, the bone-biological-responses to mechanical loads can affect the in-vivo-implant-survival-rate [1,16]. Naert I. et al in their animal study have shown that in the absence of plaque-related-gingivitis, bone resorption around the implant may threaten the implant-system-status even after functional forces [18]. As the force is transmitted from the prosthesis and implant to the bone, accurately designing and treatment planning of ISFPDs and implant-supported overdentures (ISO) play an important role in proper-stress-distribution around the implant (2). According to El-sheikh AM et al, the proper stress distribution around the inserted implant in many ISOs is too important in preventing the damaging effects of poor stress distribution [19].

3D FEM is one of the most important and practical softwares available to evaluate the stresses applied to geometrically-complex-objects. By dividing the object to smaller elements, 3D FEM can evaluate the VMS and strain applied to the whole object and all of its elements. 3D FEM is helpful in better understanding of biomechanical aspects of the objects with complex geometry such as implant system and its surrounding bone. 3D FEM is very helpful to evaluate the stress pattern and stress distribution in the bone and prosthetic elements.

Field C. et al preferred 3D FEM studies to in-vivo studies considering repeatability and the ability to control factors effective in mechanical phenomena [12]. Geng JP et al explained that FEA technique can predict stress distribution in the bone, implant-cortical-bone-interface, and implant-trabecular-bone-interface during different loading conditions [20]. Barbier L. et al in an animal-FEM-study, could change the conditions of FEA to a more realistic situation by differently dividing the forces, increasing the accuracy of elements, and using the trabecular structure [21].

As shown in the results, the maximum VMS in and dynamic loadings were observed in the crestal area of the cortical bone. The maximum VMS in the bone of five-unit ISFPD was lower than the bone of the other two models. However; the maximum VMS in the abutments and fixtures of five-unit ISFPD was more than the other two models. Of course, the VMS amount in 5-unit ISFPD in type II bone was almost comparable with those in 3-unit and 4-unit ISFPDs. The VMS of the bone of all models were in the bone-endurance-range.

Vaillancourt H. et al [22] showed that the amount of functional force in natural dentition on the premolar teeth is 50 N. On the other hand, Scwartz et al [23] reported that in the molar areas, the force is three times more than that in the premolar areas. The pattern and type of loading in the present models were consistent with the models of Scwartz et al [23] and Feild et al [24]. In their FEM study, they assessed the bone-biomechanical-responses in a three-unit FPDs. They applied 50, 100, and 150 N forces on the first premolar, the second premolar, and first molar, respectively which were similar to the present study.

Dynamic loading is generally preferred to static loading, because dental implants are always influenced with cyclical loading during different functional and parafunctional actions such as chewing, swallowing, clenching, and bruxism. So that in this study, dynamic loadings were used. According to Javier G et al, using cyclical loading to assess fatigue endurance of implant-system-materials plays an important role in estimating the long-term success of dental implant [25]. Chen et al [26], examined the stresses applied to two implant systems after static loading and dynamic loading. They expressed that VMS values of bone in dynamic loading were more than those in static loading. In their study, the maximum VMS in both types of static and dynamic loading were seen in the crestal area of bone.

In the current study, the maximum VMS in all three models were accumulated in approximately 2 mm of crestal area of cortical bone which was consistent with some other studies [27-29]. Bone density is directly related to bone strength. In the present study, D2 bone was used which is approximately 50% stronger than D3 bone (5). Bone type is important as it may cause the difference between the elastic modulus of bone with that of implant. When a stress is applied to D1 bone, less microstrains are created in this bone compared to D4 bone. In D4 bone the differences between microstrains of titanium with microstrains of bone may be categorized as the pathologic-overload-area, according to Frost [30]. In this study, D2 bone was preferred as it has higher elastic modulus compared to D3 and D4 bones. The difference between microstrains of D2 bone and microstrains of titanium implant is less than those of D3 and D4 bones, so that D2 bone is better in terms of stresses distribution.

The maximum strain amount among the models were seen in five-unit ISFPD model (1.94×10^{-3}), followed by four-unit ISFPD model (1.66×10^{-3}), and three-unit ISFPD model (1.58×10^{-3}). According to these results by increasing the number of pontics, the bone strain will increase. According to Frost et al, if the amount of microstrains is less than 50, the bone is categorized in disused atrophy zone. If it is between 50 to 1500 microstrains, it is in the adapted-window-zone. If the

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microstrains amount of bone is between 1500 to 3000, it is in mild-overload-zone, and when it is over 3000 microstrains, it should be considered in pathologic-overload-zone [30]. In this study, the mean strain in all three models were obtained between 1500 and 1950 microstrains, so that they were in mild-overload-zone. In the mild-overload-zone, bone density increases [30]. In all three models, the place of maximum VMS accumulation in the abutments were in the shank area (Morse Taper area). This area has the greatest contact area between the abutment and the fixture, so that it showed the maximum VMS. Increased VMS accumulation in the shank area is more desirable clinically compared to the VMS accumulation in the abutment screw. Because all the components fit together properly, the applied stress to the abutment screw will decrease. So that the possibility of the abutment-screw-loosening or fracture will decrease.

Ante's law should not be overgeneralized to ISFPDs. Because implant abutments do not have pdl. So that they may have more complications and problems than natural teeth. By increasing the number of pontics, bending of FPD will increase which may cause more problems in ISFPDs than tooth-supported FPDs. In the three-unit ISFPD, the maximum VMS among the bone, abutments, and implants was seen in the posterior abutment; while in the 4-unit and 5-unit ISFPDs, the max VMS was observed in the anterior implant. These results seem to be explainable considering the magnitude and location of applied loadings. In the three-unit model, a 150N axial force was applied to the posterior abutment (i.e. the first molar) with a subsequent-bending-force from the adjacent pontic. However, in two other models, the posterior abutments were the second molars to which the axial forces were not directly applied, but merely bending forces resulting from loading of pontics were applied to them.

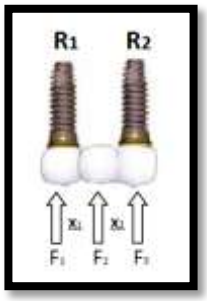

Figure 13. $2R_1= F_1\times(X_1+X_2) +F_2 \times (X_2)$ and $R_2= F_3 \times (X_1 + X_2) + F_2 + (X_2)$

Where R_1 is the anterior-implant-torque-force and R_2 is the posterior-implant-torque-force. In the 3-unit ISFPD, R_1= 100 and R_2 = 200.

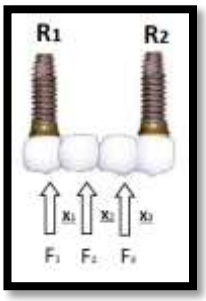

Figure 14. $3R_1=F_1\times(X_1+X_2+X_3)+F_2\times(X_2+X_3)$ and $3R_2=F_3\times(X_1+X_2)+F_2\times(X_2)$

Where R_1 is the anterior-implant-torque-force and R_2 is the posterior-implant-torque-force. In the 4-unit ISFPD, R_1= 166 and R_2 = 133.

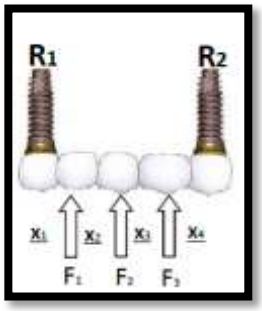

Figure 15..$4R_1=F_1\times(X_1+X_{2+}X_3)+F_2\times(X_3+X_4)+F_3\times X_4$ and $4R_2=F_3\times(X_1+X_2)+F_2\times(X_2)$

Where R_1 is the anterior-implant-torque-force and R_2 is the posterior-implant-torque-force. In the 5-unit ISFPD, R_1= 175 and R_2 = 125.

The increased VMS in the abutments and the fixtures of the five-unit model may be due to more number of pontics which may increase the bending of the FPD and may create tensile and shear stresses on the abutments. The impact forces were applied on the pontics of 5-unit model which may be another reason for the increased VMS. In a FEM study, Guven et al studied the stress distribution in pre-implant tissues and periodontal tissues and bones of 3-unit and 5-unit zirconia ISFPDs. They stated that the stress concentration in the bone of tooth-supported model was less than that of implant-supported model. In implant-supported model, more stress concentration was seen in the cervical area of the implant, while in the tooth-supported model, it was seen in the root-surrounding-bones. The highest stress was also seen in the 5-unit implant-supported model during applying 200 and 850N forces [31].

Generally, it is concluded that the stress amount in bone type II in 5-unit ISFPD is comparable to those in 3-unit and 4-unit ISFPDs. All of these VMS in bone were in the bone-endurance-range (i.e. mild-overload-zone according to Frost).

Always, there are some limitations in the FEM studies. Because of non-linear characteristics of biological tissues, FEM cannot accurately reconstruct and simulate biological-tissues-behaviors. So that overgeneralization of the results of FEM studies to clinical situations should be always done cautiously. FEM studies provide a general view on the biomechanical aspects, in the normal conditions. These biomechanical results should be examined along with the clinical results. Correct hypotheses, the intermediate and boundary conditions can affect the accuracy of FEM studies [32].

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