A THREE-DIMENSIONAL FINITE ELEMENT ANALYSIS OF A PASSIVE AND FRICTION FIT IMPLANT ABUTMENT INTERFACE AND THE INFLUENCE OF IMPLANT DIAMETER ON THE STRESS DISTRIBUTION ON THE IMPLANT AND THE SURROUNDING BONE.

Kiran Kunwar Rathore[1*], Shreya Goenka[2], Priyanka Rai[3], Gattu Anusha[4], M Manoosha[5], Venkatesh Bejagam[6]

Abstract:

Background: Bone loss around implants that may lead to implant failures is mostly in unfavourable loading conditions. Thus this study aims to evaluate and compare the stress distribution pattern in the implant and surrounding bone for a passive and friction fit implant abutment interface, & the influence of implant diameter on the stress distribution pattern on the implant and surrounding bone.

Methodology: Six CAD models of implant–abutment interface were constructed with two different types of implant–abutment connections: NOBLE REPLACE TAPERED GROOVY, RP (4.3mm × 13mm) and NOBEL ACTIVE, INTERNAL RP (4.3mm × 13mm). The von misses stresses were analyzed on a total of six implant-abutment interfaces with three different implant dimensions by applying vertical and oblique load of 100N.The amount of stress and pattern of stress generated were recorded on a color scale using ANSYS 15.

Results: Kolmogorov- Smirnov test and Shapiro-Wilks test were employed to test the normality of data. Analysis of variance and Post hoc test was employed for the quantitative variables. It was found that stresses were mainly concentrated at the crestal bone covering the neck of implant and its surrounding bone.

Conclusion: The friction fit connection is superior to the passive fit connection, as the friction fit creates wedging effects to improve the implant abutment joint stability against the lateral force and helps to transfer the loading force along the conical surface to distribute the stress on the implant, ultimately reducing biological and biomechanical complications.

[1*,3]Senior Lecturer, Department of Prosthodontics and Crown & Bridges Army college of Dental sciences, Hyderabad.
[2]Clinical practitioner, Department of Prosthodontics and Crown & Bridges, Nagpur
[4]Reader, Department of Prosthodontics and Crown & Bridges, Army College of Dental sciences, Hyderabad.
[5]Senior Resident, Department of Prosthodontics ,Government Dental College And Hospital , Kadapa, Andhra Pradesh.
[6]Senior Lecturer, Department of Prosthodontics and Crown & Bridges, Meghna Institute of Dental Sciences, Nizamabad, Telangana, India

***Corresponding Author:** Kiran Rathore
*Senior Lecturer, Department of Prosthodontics and Crown & Bridges ,Army college of Dental sciences, Hyderabad.

DOI: 10.48047/ecb/2023.12.9.251

INTRODUCTION:

In the late 1970s, Branemark gave concept of modern implantology with the accidental discovery of osseointegration of titanium within the bone. Since then prosthodontic rehabilitation with osseointegrated implant turned out to be the therapeutic solution of choice for treating partially or completely edentulous arches. With the present advancements in the technology, osseointegration along with esthetics and function serves as important criteria to be fulfilled for a successful implant treatment.[1] The design of an implant system as characterized by its geometry and type of implant-abutment connection is an important factor in establishing the performance and maintenance of implant osseointegration and implant supported prosthesis since design determines load transmission at both the bone-implant and the implant-abutment interfaces and minimizes the micromovemnt.[3] Implant abutment connections could be external or internal depending on the distinct projection in implant body.

One of the most important aspects of the stability and long term prognosis of the implant is the type of implant connection that is friction fit implants versus passive fit implants. In a friction fit implant, the force used creates a wedging effect and better contact between the implant interface and the underlying bone. It has been shown to provide for better stability and stress distribution.[6] It is postulated that the increase in implant length and diameters cause a simultaneous increase in the surface area between the implant and the underlying bone. The larger surface area can thus distribute the load more evenly thereby causing less stress. Moreover, it gives a wider area for osseointegration. The diameter of an implant measures the outside dimension of the thread.[7]

When the implants are out of esthetic zone, the diameter of implant is more related to amount of force applied to the implant –prosthetic system. The posterior region should most often use 3.7 to 4.2 mm diameter implant in premolar region and 5 mm diameter implant in molar region. However the implant diameter is 5 mm to 6mm for maxillary molars but as titanium is 5-10 times more rigid than natural teeth.

More focus on increasing the number of implants should be given when diameter of implant do not provide sufficient surface area to compensate for soft bone types or unfavourable force factors example; parafunction. Due to difference in all these factors; all these factors must be assessed for better prognosis and longevity of implants. Thus this study aims to evaluate and compare the stress distribution pattern in the implant and surrounding bone for a passive and friction fit implant abutment interface, & the influence of implant diameter on the stress distribution pattern on the implant and surrounding bone.

METHEDOLOGY

After obtaining the institutional ethical clearance the present in vtro study was conducted using 3-D model of a mandibular premolar bone section and implant abutment-interface was designed with a computer aided design software (SOLIDWORKS 14). The model was made to undergo finite element analysis with software (ANSYS 15) and this determined the response and influence of implant –dimensions on the stress distribution pattern on the implant and surrounding bone.

Models designed by this software were:

a) Implant abutment connection : 1.Passive fit (NOBLE REPLACE TRILOBE),2. Friction fit (NOBEL ACTIVE CONICAL CONNECTION)
b) Six model of implant –abutment connection with three different dimensions each respectively; 3.5 mm,4.3mm, 5.0 mm.
c) Geometric model of cortical and cancellous bone modeled using computed tomography scan

CONSTRUCTION OF GEOMETRIC MODELS:

Six CAD models of implant–abutment interface were constructed with two different types of implant–abutment connections: NOBLE REPLACE TAPERED GROOVY, RP (4.3mm × 13mm) and NOBEL ACTIVE, INTERNAL RP (4.3mm × 13mm). The average height and width of the mandibular section model was 25 mm and 12mm with thickness of 10mm surrounded by 2mm cortical bone. Three different implant dimensions: narrow, regular and wide were constructed and total of six models were then placed in cancellous and cortical bone.

APPLICATION OF MATERIAL PROPERTIES:

All materials used in the model were considered as homogenous, isotropic and linear elastic. The cortical and cancellous bone were treated as anisotropic and assumed that there was osseointegration between the implant and surrounding bone.

APPLICATION OF BOUNDARY CONDITIONS

A cross sectional image of human mandible in the premolar region surrounded by 2mm cortical and cancellous bone was prepared and was subjected to axial and oblique forces. a) Vertical force of 100 N was applied onto the top surface of crown vertically along the long axis b) Oblique force of 100N at 45 degree to the implant A systematic comparison with the same loading condition, boundary condition and constraints was applied in all the models.

ANALYSIS OF STRESS PATTERN

The von misses stresses were analyzed on a total of six implant-abutment interfaces with three different implant dimensions by applying vertical and oblique load of 100N.The amount of stress and pattern of stress generated were recorded on a color scale using ANSYS 15. The von misses stress values are defined as the beginning of deformation for ductile materials such as metallic implant. Failures occur when the von misses stress values exceed the yield strength of an implant material. Therefore they are important for interpreting the stress occurring within the implant material.

The data obtained from ANSYS calculation was presented in a stress distribution map with a colour scale, which made it possible to compare directly the stress level in various component structures of all models. The amount of stress and the pattern of stress generated after applying a load of 100 N on each model in vertical and oblique direction was recorded on a colour scale. Each colour band represents a particular range of stress value, which is given in Newton per meter square (N/m^2). Blue and red colour represent minimum and maximum stress respectively

Results:

The data collected was entered in Microsoft Excel and subjected to statistical analysis using Statistical Package for Social Sciences (SPSS, IBM version 20.0). The level of significance was fixed at 5% and $p \leq 0.05$ was considered statistically significant. Kolmogorov- Smirnov test and Shapiro-Wilks test were employed to test the normality of data. Analysis of variance and Post hoc test was employed for the quantitative variables.

Post hoc analysis for vertical loading using both Friction Fit and Passive Fit connection revealed significant differences between all the parameter except between narrower and ideal for friction fit and ideal and wider for passive fit connection (Table 1a&1b).

Stress distribution		**Mean Difference**	**Narrow**	**Mean Difference**	**Regular**	**Mean Difference**	**Wide**
On Implant	Narrow	-	-	-20.06	**.001***	-9.85	**.001***
	Regular	20.06	**.001***	-	-	10.21	**.001***
	Wide	9.85	**.001***	-10.21	**.001***	-	-
On Cortical Bone	Narrow	-	-	-.525	**.001***	-.369	**.001***
	Regular	.525	**.001***	-	-	.155	**.001***
	Wide	.369	**.001***	-.155	**.001***	-	-
On Cancellous Bone	Narrow	-	-	-.00432	.995	-4.209	**.001***
	Regular	-.00432	.995	-	-	4.205	**.001***
	Wide	4.209	**.001***	4.205	**.001***	-	-

Table 1a: Post hoc analysis of stress distribution using vertical Loading **-** Friction Fit Connection

Stress distribution		**Mean Difference**	**Narrow**	**Mean Difference**	**Regular**	**Mean Difference**	**Wide**
On Implant	Narrow	-	-	-14.54	**.001***	-13.83	**.001***
	Regular	14.54	**.001***	-	-	.71	**.001***
	Wide	13.83	**.001***	-.71	**.001***	-	-
On Cortical Bone	Narrow	-	-	**-**.911	**.001***	-.22	**.001***
	Regular	**.**911	**.001***	-	-	.690	**.001***
	Wide	.22	**.001***	-.690	**.001***	-	-
On Cancellous Bone	Narrow	-	-	-.136	**.001***	-.148	**.001***
	Regular	.136	**.001***	-	-	-.012	.133
	Wide	.148	**.001***	.012	.133	=-	-

Table 1b: Post hoc analysis of stress distribution using Vertical Loading – Passive Fit Connection

Post hoc analysis for oblique loading using both Friction Fit and Passive Fit connection revealed significant differences between all the parameter except between ideal and wider on implant surface and ideal and narrower on Cancellous bone for friction fit and ideal and between ideal and wider on cortical bone for passive fit connection (Table 2a&2b).

Stress distribution		Mean Difference	Narrow	Mean Difference	Regular	Mean Difference	Wide
On Implant	Narrow	-	-	-26.12	**.001***	-30.91	**.001***
	Regular	26.12	**.001***	-	-	-4.78	.199
	Wide	30.91	**.001***	4.78	.199	-	-
On Cortical Bone	Narrow	-	-	-2.09	**.001***	-.61	**.001***
	Regular	2.09	**.001***	-	-	1.47	**.001***
	Wide	.61	**.001***	-1.47	**.001***	-	-
On Cancellous Bone	Narrow	-	-	-.021	.262	-14.28	**.001***
	Regular	.021	.262	-	-	-14.26	**.001***
	Wide	14.28	**.001***	14.26	**.001***	-	-

Table 2a: Post hoc analysis of stress distribution using Oblique Loading **-** Friction Fit Connection

Stress distribution		Mean Difference	Narrow	Mean Difference	Regular	Mean Difference	Wide
On Implant	Narrow	-	-	-101.9	**.001***	-166.6	**.001***
	Regular	101.91	**.001***	-	-	-64.73	**.001***
	Wide	166.6	**.001***	64.73	**.001***	-	-
On Cortical Bone	Narrow	-	-	-.57	**.001***	-.59	**.001***
	Regular	.57	**.001***	-	-	-.02	.892
	Wide	.59	**.001***	.02	.892	-	-
On Cancellous Bone	Narrow	-	-	-.83	**.001***	-.25	**.006***
	Regular	.83	**.001***	-	-	.58	**.001***
	Wide	.25	**.006***	-.58	**.001***	-	-

Table 2b: Post hoc analysis of stress distribution using Oblique Loading – Passive Fit Connection

DISCUSSION:

A dental implant abutment is formally defined as "that portion of a dental implant that serves to support and/or retain a prosthesis". It functions to physically connect the clinical crown(i.e. prosthesis) to the implant. There are at least three ways this occurs among different implant systems. One is a modular design in which the endosseous implant and the transmucosal abutments are separate components. Alternatively, the endosseous implant and transmucosal aspect of the system may be one component and, in such cases, the crown margin is part of this integrated implant system. The two key features that are critical to use and understanding of the modular versus integrated design systems are that the integrated design system lacks an implant-abutment interface approximating the implant-bone interface, and that the crown margin for integrated implant designs is established by implant placement and cannot be modified with preparation of the implant itself. A modular system, while presenting an Implant-abutment interface at the implant-bone interface, permits the crown margin location to be modified in relation to implant position. A third design has emerged that is unitary in which the endosseous, transmucosal and restorative aspects of the implant system are a single component. [6]

The aim of the study was to analyze the influence of two different types of implant abutment connection on the stress distribution pattern in the implant and the surrounding bone. The implant abutment interface that have been analyzed in the study represent two broad categories of implant abutment connection currently available in the market, the passive fit or the slip‑ fit represented by the Nobel Replace Tri‑ lobe connection and the friction fit or active fit represented by the Nobel active conical connection. The stress distribution pattern was studied at different diameters dimension. Six models were constructed in SOLIDWORKS 15 of the two implant abutment connection for three different diameter dimensions each. The implant and abutment complex was placed in cortical and cancellous bone modeled using a CT scan.

This complex was subjected to a force of 100 N in the axial and oblique direction. The amount of stress and the pattern of stress generated were recorded on ANSYS colour scale. The mean values of the overall stresses on the implant and the bone shows that the friction fit connection absorbs more stress and dissipates less stress to the surrounding bone. The larger contact area and deeper position inside the implant for friction fit connection allowed for better stability and broader stress distribution which as has been observed in several other studies. Conical connections were developed to achieve friction based fit of the implant components. This frictional fit creates wedging effects to improve the implant abutment joint stability against the lateral force and helps to transfer the loading force along the conical surface to distribute the stress on the implant, ultimately reducing biological and biomechanical complications. The internal conical connections help the abutment screw retain greater preload after repeated loads since the loading stress is not entirely concentrated on the screw as in the external hex butt joint implant systems. The friction locking mechanics and the solid design of the friction fit connections provided greater resistance to deformation and fracture under oblique compressive loading when compared to the passive fit connection.

In passive fit connection, the cold welding does not occur when the abutments are tightened thus an inevitable gap between the implant and abutment may still exist. This can cause micro‑motion at the interface during clinical loading, which in turn may contribute to stress on the screw and therefore loss of preload and loosening of abutment thereby leading to bacterial colonization of the micro gap. The threshold of deleterious micro‑motion level asserted by various researchers' lies within the range of 50–150um. Beyond these levels of micro‑motion, stress concentration may occur around inserted dental implants leading to crestal bone loss. The highest stress occurs in the implant's most cervical region when an occlusal load is applied upon an implant, and the load is partially transferred to the bone. This phenomenon is due to one of the principles of engineering, that is, when two materials are in contact with each other, and one of them is loaded, the stresses will be higher at the materials' initial point of contact.

This explains why the cervical region of the implant is the site where the greatest micro‑deformations occur independently of the type of bone and the design of the implant, the configuration of the prosthesis and the load. The results of the current FEA for the osseointegrated model are in accordance with the findings of Hansson.[7]

Using FEA, Hansson showed that a conical implant abutment interface at the level of the bone crest decreases the peak bone‑implant interfacial stress as compared with the flat top interface. For the friction fit implant abutment interface, this peak interfacial shear stress was located at some depth in the marginal bone. In this study, significantly larger stress values were seen in the neck area versus the apex area among all models in all conditions, which is consistent with the results of other studies. Stresses induced by occlusal load are initially transferred from implant to the cortical bone, and a small amount of remaining stress spreads to cancellous bone.[8]

Higher stress values are observed in cortical bone because of higher modulus of elasticity and bone density compared to the cancellous bone.[8]

Richter has reported that the highest stress in the crestal bone is a result of a transverse load and clenching at centric contacts. The width of almost every natural tooth is greater than the width of the implant used to replace the tooth. The greater the width of a transosteal structure, the lesser the magnitude of stress transmitted to the surrounding bone. The cross‑sectional shape of the natural tooth at the crest is biomechanically optimized to resist lateral loads, implants, however, are almost round in cross section, which is less effective in resisting lateral bending loads thereby concentrating loads in the crestal region. The mean values of axial displacement of teeth in the socket are 25-100 m, whereas the range of motion of osseointegrated dental implants has been reported approximately 3-5 m. The elastic modulus of the tooth is closest to bone compared to the available implant biomaterials. Hence, under similar loading conditions implant generates greater stresses and strain at the crest of bone than a natural tooth. [9]

Himmlova et al proposed maximum implant diameter seems to affect stress peaks at the cortical bone not the trabecular region, while stress values and distribution at the cancellous bone are primarily influenced by implant length.[10]
M.M. El-Zawahry conducted a three dimensional finite element study on dental implant design and showed Von Mises stresses changed with

increasing implant diameter under bending loading and for small implant diameter increasing implant length dramatically decrease maximum tensile stress generated in cortical bone under tension loading, proving that implant diameter is the dominant factor not the implant length.[5]

From the results of the study, it is shown that the narrow occlusal table, irrespective of their connection type has reduced the stress generated. This shows that the width of the implant diameter has got a significant influence on the stress generated on the implant, as well as on the bone. Typically, a 30% to 40% reduction in the implant-diameter in a molar region has been suggested because any dimension larger than this implant diameter can cause cantilever effects and eventual bending movements in single‑ implant prostheses. A narrow occlusal table reduces the chance of offset loading and increases axial loading, which eventually can decrease the bending movement.

Misch has described how a narrow occlusal table can improve oral hygiene and reduce the risk of porcelain fracture. The proposed key factors to control bend overload in posterior restorations were reduced the inclination of cusps, centrally oriented contacts with a 1–1.5 mm flat area, a narrowed occlusal table, and elimination of cantilevers. As the wider occlusal table will increase stress on the abutment screws, the diameter should be reduced in width compared with natural teeth in non esthetic regions of the mouth.[11]

Analysis of finite elements was shown to be a versatile and promising methodology for analyzing stress concentrations in implant dentistry, but it is worth emphasizing that the FEA is an approximate virtual simulation of clinical situations, presenting certain limitations. Therefore, the results derived from this FEA may be considered to be reasonably accurate and acceptable.

Limitations of the present study is Finite element Analysis may not exactly simulate the clinical situation completely since it is a mathematical in-vitro study, Natural bone is viscoelastic, isotrophic and hetrogenous material but in present study all materials are assumed to be linearly elastic and homogenous.

Conclusion:

The friction fit connection is superior to the passive fit connection, as the friction fit creates wedging effects to improve the implant abutment joint stability against the lateral force and helps to transfer the loading force along the conical surface to distribute the stress on the implant, ultimately reducing biological and biomechanical complications. A narrow diameter may increase axial loading and decrease non-axial loading for the implants thereby reducing the stress on the implant, implant abutment interface, and bone. Thus, it is recommended that the size of the occlusal table to be 30% to 40% smaller for molars. The implant dentistry would greatly benefit if it were provided the means to predict how bone and implant components would behave considering each patient’s unique jaw anatomy, quality of bone, amount of bone, amount of occlusal force exerted on the prosthesis angulation of abutment etc. Finite element analysis with all its inherent limitations is a valuable instrument in pursuing that goal.

Finite element analysis is an upcoming and significant research tool for biomechanical analyses in biological research.It is an ultimate method for modelling complex structures and analysing their mechanical properties. In Implantology, finite element analysis has been used to study the stress pattern in various implant components and also in the peri-implant bone.

References:

1. Devaraju K. Rao SJ, Joseph JK, Kurupati SKR. Comparison of biomechanical properties of different implant-abutment connections. Indian J Dent Sci 2018;10:180-183.
2. Shelat S, Kularashmi BS, Annapoorani H, Chakravarthy R. Effect of two different abutment types on stress distribution in the bone and around an implant under two loading conditions. J Dent Implant 2011;1:80-85.
3. Quaresma SE, Cury PR, Sendyk WR, Sendyk C. A finite element analysis of two different dental implants:Stress distribution in the prosthesis, abutment, implant and supporting bone. J Oral Implantol. 2008;34(1):1-6.
4. Sarfaraz H, Paulose A, Shenoy KK, Hussain A. A three dimensional finite element analysis of a passive and friction fit implant abutment interface and the influence of occlusal table dimension on the stress distribution pattern on the implant and surrounding bone. J Indian Prosthodont Soc 2015;15:229-36.

5. MI EI-Anwar, MM EI-Zawahry. A three dimensional finite element study on dental implant design. Journal of Genetic Engineering and Biotechnology 2011;9:77-82.
6. Shetty M, Prasad DK, Shetty NHG, Jaiman R. Implant abutment connection: Biomechanical perspectives. NUJHS 2014 Jun4;(2):47-53.
7. Sarfaraz H, Paulose A, Shenoy KK, Hussain A. A three dimensional finite element analysis of a passive and friction fit implant abutment interface and the influence of occlusal table dimension on the stress distribution pattern on the implant and surrounding bone. J Indian Prosthodont Soc 2015;15:229-36.
8. Hansson S. Clin Implant Dent Res.2000;2(1):33-41. Implant-abutment interface: biomechanical study of flat top versus conical. Clin Implant Dent Relat Res.2000;2(1);33-41.
9. Richter EJ. In vivo horizontal bending moments on implants. Int J Oral Maxillofac implants 1998;13:232-44.
10. Himmlova L, Dostalova T, Konvickova S. Influence of implant lengthand diameter on stress distribution: a finite element analysis. J Prosthet Dent. 2004 Jan;91(1):20-25.
11. MI EI-Anwar, MM EI-Zawahry. A three dimensional finite element study on dental implant design. Journal of Genetic Engineering and Biotechnology 2011;9:77-82.