ISSN NO. 2320-5407

Journal homepage: http://www.journalijar.com

INTERNATIONAL JOURNAL OF ADVANCED RESEARCH

RESEARCH ARTICLE

3D Finite Element Analysis of Two Modified Implant-Abutment Connections

Dr.Mohammed Suleman Alruthea, Dr.Mostafa Omran Hussein

1. Lecturer of Prosthodontics, Prosthodontic Department, Faculty of Dentistry, Qassim University Kingdom of Saudi Arabia

2. Assistant Professor of Prosthodontics, Prosthodontic Department, Faculty of Dentistry, Qassim University Kingdom of Saudi Arabia

Manuscript Info

Manuscript History:

Received: 26 October 2014
Final Accepted: 23 November 2014
Published Online: December 2014

Key words: Implant-abutment connection, crestal bone, 3D Finite element, bone stress.

Corresponding Author

Dr.Mostafa Omran Hussein

Abstract

The present study focused on modifying the implant-abutment connection in order to minimize crestal bone stress. Two modifications, based on two strategies, were studied. Three virtual 3D models were created including implant-abutment complex placed on bony tissues. The first model represented the conventional design; the second was modified by creating relief at implant hex area while a shock absorber was incorporated at the third design. Using finite element software, two simulations per design were applied, one at vertical 100 N load and the second by oblique (45º) 30 N load. Maximum and minimum principle stresses of the cortical and cancellous bones were assessed. The results of the vertical load showed highest stress value at the second design followed by conventional design then the third design was the least value. The oblique load generated highest value at the second design followed by third design then conventional design. Within the limitations of the study, it could be concluded that, although the proposed designs were not fully successful to reduce crestal bone stress but they changed the stress distribution of the conventional design.

Introduction

Risk factors in implant dentistry could be classified as general, biomechanical, and esthetic considerations (Werking, 2009). (Porter & Von Fraunhofer, 2004) believed that the main predisposing factor of implant failure are poor bone quality, chronic periodontitis, systemic diseases, smoking, implant location, short implants, eccentric loading, inadequate number of implants, para-functional habits and absence/loss of implant integration with hard and soft tissues. They also added that inappropriate prosthesis design might also contribute to implant failure. Between all these factors of implant failure, overloading is the most annoying predisposing factor to implantologists. (Kitamura et al., 2004) clarified that marginal bone loss was the initial clinical finding detected. However, they claimed that marginal bone loss could happen, to a certain extent, due to biomechanical adaptation of bone to stress. Further, as bone resorption progresses, the increasing stresses in the cancellous bone and implant under lateral load may result in implant failure.

Various researches focused on revealing the optimal implant design that enhance prosthesis longevity (Okumura et al., 2010; Porter & Von Fraunhofer, 2004; Valderrama et al., 2011; Werking, 2009). (Valderrama et al., 2011) evaluated implant design to determine their impact on the behavior of the crestal bone. They recommended the use of surface treated non-machined collar as criteria for interface between implant and marginal bone. Moreover, (Heitz-Mayfield et al., 2013) conducted a comparative study among designs of three premium commercial implant systems and their influence on crestal bone preservation. According to their findings, the Astra Tech Osseospeed implant profile reduced marginal bone significantly and showed greater bone preservation.

(Freitas Jr et al., 2012) reported that during implant placement, the insertion torque was reduced in implant macro designs that incorporated cutting edges, and lesser insertion torque was generally associated with decreased micromovement. However, they showed that the insertion torque and implant profile design were irrelevant. (Peñarrocha-Diago et al., 2013) compared clinically two different implant different neck features and prostheses platform connection (machined with external hex and rough-surfaced with switching platform) regarding their influence on peri-implant marginal bone loss before and after loading. After 12 months, they confirmed that the implant-abutment connection type had a significant impact on peri-implant crestal bone levels. On other hand, a retrospective study on the influence of the implant-abutment connection type on the crestal bone resorption revealed no significant results on short-term period (Lin et al., 2013).

Using nonlinear dynamic finite element analysis (FEA), (Covani et al., 2013) studied mechanical efficiency of four different types of connections (modified internal hex connection using collar, an internal hex connection, a standard connection without hex, and an external hex). They showed that the collar-modified internal hex connection was more resistant to loosening and distortion than the traditional hex. Furthermore, (Dittmer et al., 2011) evaluated in-vitro the load bearing capacities and failure modes of six commercial implant systems with different implant-abutment connection designs. They confirmed that the implant–abutment connection design had a major role in bearing load and deformation during function. Similarly, (Freitas-Júnior et al., 2012) proved by laboratory and FEA, that the use of regular internal hex connection was preferred than platform-switching external hex connection.

(Shafie & White, 2014) reported that the internal connections are preferred that other connections due to their superior esthetic and excellent microbial seal between the abutment and implant. Similarly, (Gracis et al., 2012) reviewed both internal and external connection to clarify their role in incidence of screw fracture or loosening. They clarified that there is no correlation between incidence of fracture and connection type whatever abutment material used. In contrast, loosening of abutment screws was the most frequently occurring technical complication accompanied external connection type. According to the FEA simulated by (Pessoa et al., 2010), the use of immediate loading protocol seems to override the biomechanical environment of the dental implant. They showed that the implant-abutment connection design had less influence on the biomechanical environment than implant overloading and primary stability in immediate loading protocol.

(Morton, Stanford, & Aquilino, 1998) evaluated the role of using resilient abutment components on measured strain under dynamic loading. They concluded that polyoxymethylene abutment components did not affect measurable bone strain compared to titanium. The use of zirconia abutments were also tested mechanically with different implant-abutment connections and no benefits were shown (Truninger et al., 2012).

(Chapman & Kirsch, 1989) were pioneers in testing shock-absorber concept to change occlusal load dissipation on implants. They used T-scan occlusal analysis system to assess the efficacy of using this new concept on occlusal load distribution. Their findings were encouraging and supported the research hypothesis. On the same track, (Richter, Orschall, & Jovanovic, 1990) conducted a study to disclose the role of using built-in compliance resembling the tooth mobility to optimize force distribution in implant-tooth connection cases. The results showed a remarkable reduction of stress values with a uniform distribution. Moreover, (Carvalho, Vaz, & Simões, 2004) compared conventional and modified novel dental implants implanted in a cadaveric mandible and loaded dynamically. The novel implant was modified by elastomeric material barrier (stress shock absorber) interposed between the implant and an artificial crown. The modified implant was able to dissipate forces surrounding bone and enhance load distribution. However, (Hoffmann & Zafiropoulos, 2012) published a review about implant-tooth connection and they suggested, according to the clinical studies collected, the use of rigid connection between tooth and implant.

The aim of the current study was to reduce the marginal bone stress created under loading of the dental implant using two suggested modifications at the implant-abutment connection. Two hypotheses were drawn, one seeks changing the pattern of stress distribution, and the other relies on using resilient implant-abutment connection.

Materials & Methods

I. Implant Design

A 3D model of a dental implant (Nobel replace select, Nobel Biocare, Switzerland) and their abutments were created using the CAD designing software (MOI v2, Triple squid software design, USA). Both implants dimension (4.3 mm diameter x 13 mm length) and shape were simulated to represent the selected implant. Abutment was also designed to fit fixture hex and to be tightened by screw instead of using tri-channel of the original implant type. Three 3D models were designed by the CAD modeling software, (figure 1). The implant models were categorized as follows:

Model I: was the control group where fixture and abutments were connected through the whole surface representing conventional rigid connection.

Model II: where fixture-abutment connection was modified by creating a space to minimize contact at the coronal part of the fixture.

Model III: where a continuous layer of shock absorber was added to occupy the interface between fixture and abutment.

Implant fixture of the second group was modified by cutback area surrounding counterpart of the fixture hex to limit contacts between fixture and abutment to collar top and tightening screw areas, Fig 1, B. Fixture of the third group was modified by exchanging areas of contact between fixture and abutment-screw complex by polymeric shock absorber material, seen as a pink color, Fig 1 C. All implant assemblies were exported as a (SAT) file extension to be imported later to the design modeler module of the finite element software (ANSYS Workbench v 14 package; ANSYS, Inc., Canonsburg, PA, USA). Within the design modeler, both compact and cancellous bones were designed as piece of mandibular body with dimension (30x35 mm). The cross-section of the designed bone was created by simulating mandibular body cross-section at premolar area. The Compact bone layer was designed by creating variable thickness Shell surrounding cancellous bone. After bone modeling the implant was inserted by Boolean Subtraction operation within bone layers, (figure 2).

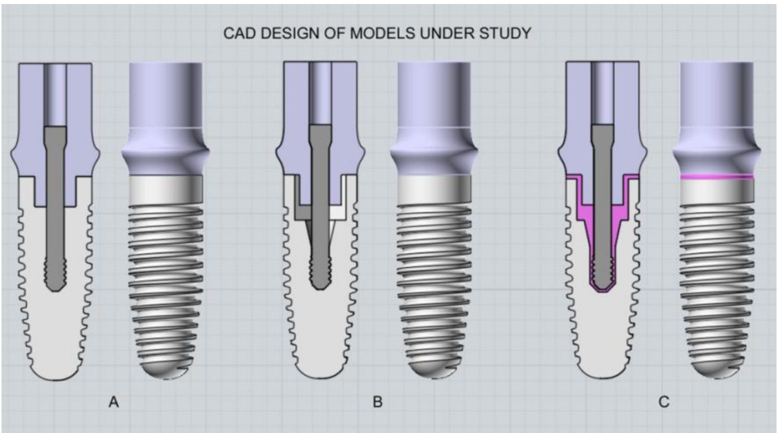

Figure 1: Representative of different design modifications used, A; control; B & C model 2 and 3 respectively.

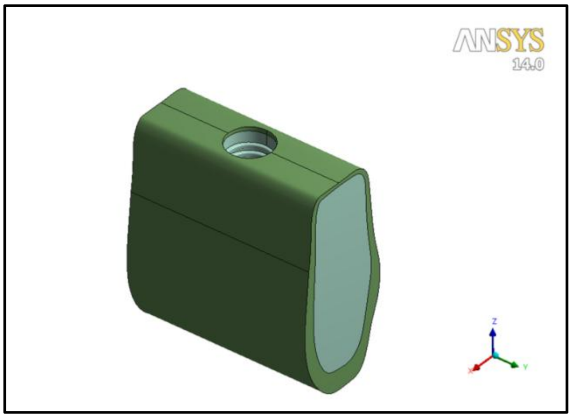

Figure 2: Isometric view of the 3D model of both compact and cancellous bone in design modeler module of the ANSYS software.

Table 1: Material properties used to be assigned to model components.

TABLE 1 **Material properties**		
Material	**Elastic Modulus (GPa)**	**Poisson's Coefficient (μ)**
Implant	113.8	0.342
Compact bone	20	0.3
Cancellous bone	2	0.4
Shock absorber	3	0.28

II. Material Properties

Linear isotropic material properties were considered for this Simulation. Thus, each part of implant assembly as defined by two engineering values, Young's modules and Poison's ratio. Accordingly, two values were assigned for each item of the model, table (1).

III. Meshing process

Meshing was done by converting the CAD 3D model into elements and nodes. Tet-10 element type was selected and used to mesh the model with minimizing element Size in areas of Contact that had particular Concerns, (figure 3).

IV. Boundary condition

As a linear analysis was predefined, bonded contact were assigned to all Contact areas between neighboring components. Each model was loaded according to conditions. Firstly, a 100 N vertical load was applied on the top of abutment surface to distribute the force widely. Second Condition considered oblique load by applying 30 N at 45° angle in a lingo buccal direction simulating lateral force executed at working Side of mastication. All models were constrained in both loading conditions with Zero degree of freedom at the bone cross-section ends mesially and distally.

V. Identifying Results

The current study focused on the influence of the design modification on the supporting tissues. Accordingly, maximum and minimum principle Stress was evaluated because of the brittle behaviors of both compact and cancellous bones. The maximum values of both types of Stress representing studied tissues were Collected and tabulated. Furthermore, a qualitative analysis of each condition was interpreted as a scale of Color bands from red to blue represent the stress values.

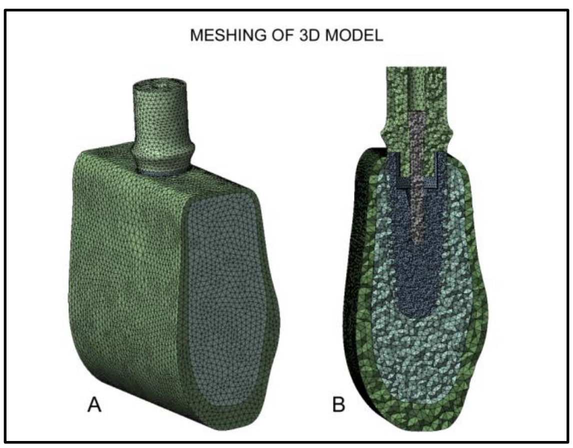

Figure 3: Meshing of the whole assembly with mesh refinement at critical areas. A; isometric view while B; is a cross-sectional view.

Results

All data were collected and tabulated including both studied parameters (Maximum principle stress and minimum principle stress), table 2 & 3.

I. Maximum and Minimum principle stress at vertical load

1.1 Compact bone

Under 100 N vertical load the maximum principle stress was variably expressed at the studied models. The maximum principle stress of the of the second model were the highest value between the studied models (5.7436 MPa) followed by value of control model (5.4066 MPa) then the value of the third model which was the lowest recorded value (4.6007 MPa), table 2. The minimum principle stress value recorded at the third model (1.6517 MPa) was higher than the other models whose showing approximately similar values, (1.6403 MPa), table 2.

According to the color coded scale, the stress distribution of the studied models were close to the control group. A band of higher maximum principle stress values were seen in cross-sectional view at crestal area surrounding implant collar. From the occlusal view, a rim of stress concentration was also detected at areas of fixed constrains, figure 4.

Similarly, the minimum principle stress was concentrated at the interface between the crestal bone and the implant fixture as seen in cross-sectional view of figure 5. Some bands of stress concentration was also seen at the constrain areas, figure 5.

1.2 Cancellous bone

The maximum principle stress expressed in the second model was higher than the other models (0.84111 MPa) followed by value of the control model (0.80182 MPa) while the third model showed the lowest value among studied models (0.78546 MPa), table 2.

Similarly, the minimum principle stress of the second model was highest value among all models (0.20856 MPa) followed by the control group (0.19244 MPa) and then the lowest value recorded at the third model (0.16462 MPa), table 2. The crestal cancellous bone zone surrounding implant collar showed higher stress color bands of all models. Generally, all bone surrounding threaded areas showed stress concentration than zones far from implant-bone interface, figure 4.

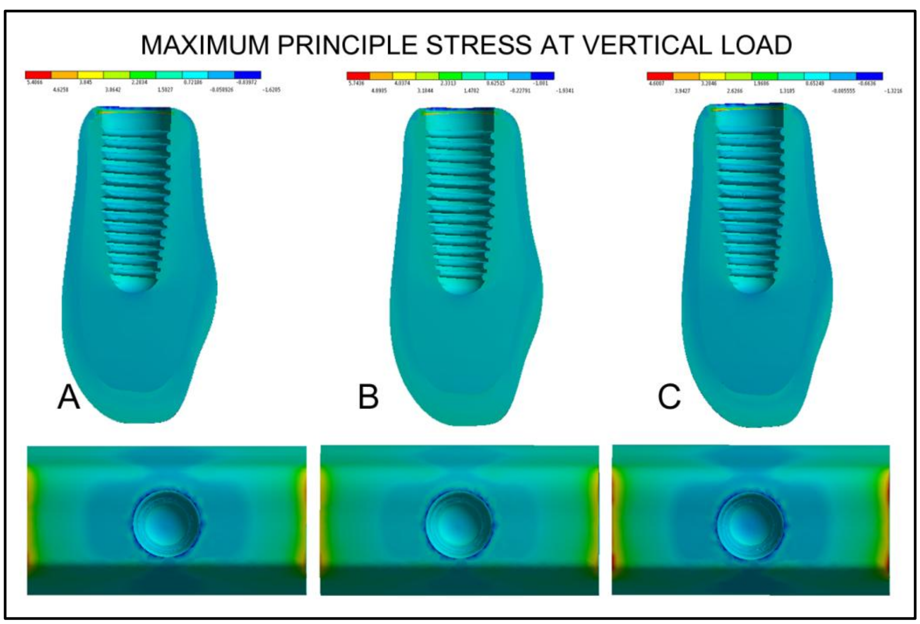

Figure 4: Maximum principle stress at vertical load seen in compact and cancellous bones. A, cross-sectional and top view of control model, B, cross-sectional and top view of the second model and C, cross-sectional and top view of the third model.

Table 2: Maximum and Minimum principle stress (in MPa) of compact and cancellous bones under vertical load.

TABLE II Vertical Load 100 N			
Group	**Tissue**	**Maximum Principle Stress**	**Minimum Principle Stress**
CONTROL	compact	5.4066	1.6403
	cancellous	0.80182	0.19244
MODEL II	compact	5.7436	1.6403
	cancellous	0.84111	0.20856
MODEL III	compact	4.6007	1.6517
	cancellous	0.78546	0.16462

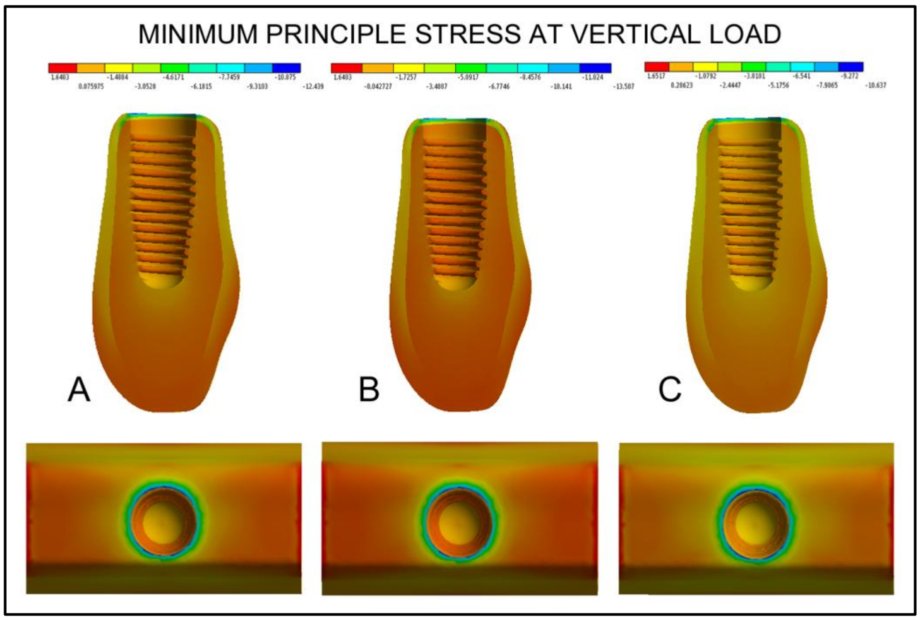

Figure 5: Minimum principle stress at vertical load seen in compact and cancellous bones. A, cross-sectional and top view of control model, B, cross-sectional and top view of the second model and C, cross-sectional and top view of the third model.

II. Maximum and Minimum principle stress at oblique load

2.1 Compact bone

By applying the load in an oblique direction, the value of maximum principle stress of the second model was the highest value among studied models (15.113 MPa) followed by the third model (14. 07 MPa) then the control model (13.077 MPa), table 3.

The minimum principle stress of the control model showed highest value among studied models (1.8649 MPa) followed by the third model (1. 86 MPa) then the lowest value was recorded at the second model (1.8552 MPa).

Upon applying the oblique load, the stress distribution of the maximum principle stress was changed to show higher stress concentration on one side of the crestal bone while minimal stress on the opposite side, figure 6. The same finding was detected during evaluating stress distribution of the minimum principle stress. A band of stress concentration was recognized at one side with lesser values at the opposite side, figure 7.

2.2 Cancellous bone

The maximum principle stress of the cancellous bone seen in the third model was the highest value (1.1508 MPa) followed by the second model (1.1159 MPa) and finally the lowest value recorded in the control group (1.1001 MPa), table 3.

The minimum principle stress recorded at the cancellous bone of the second model was the highest (0.10303 MPa), while the third model was (9.67E-02 MPa) and the control model was the lowest (7.30E-02 MPa).

The minimum principle stress concentration was limited to the area surrounding collar part of the implant away from the loaded side, figure 6, 7.

Table 3: Maximum and Minimum principle stress (in MPa) of compact and cancellous bones under oblique load

TABLE III Oblique load 30 N			
Group	**Tissue**	**Maximum Principle Stress**	**Minimum Principle Stress**
CONTROL	compact	13.077	1.8649
	cancellous	1.1001	7.30E-02
MODEL II	compact	15.113	1.8552
	cancellous	1.1159	0.10303
MODEL III	compact	14. 07	1. 86
	cancellous	1.1508	9.67E-02

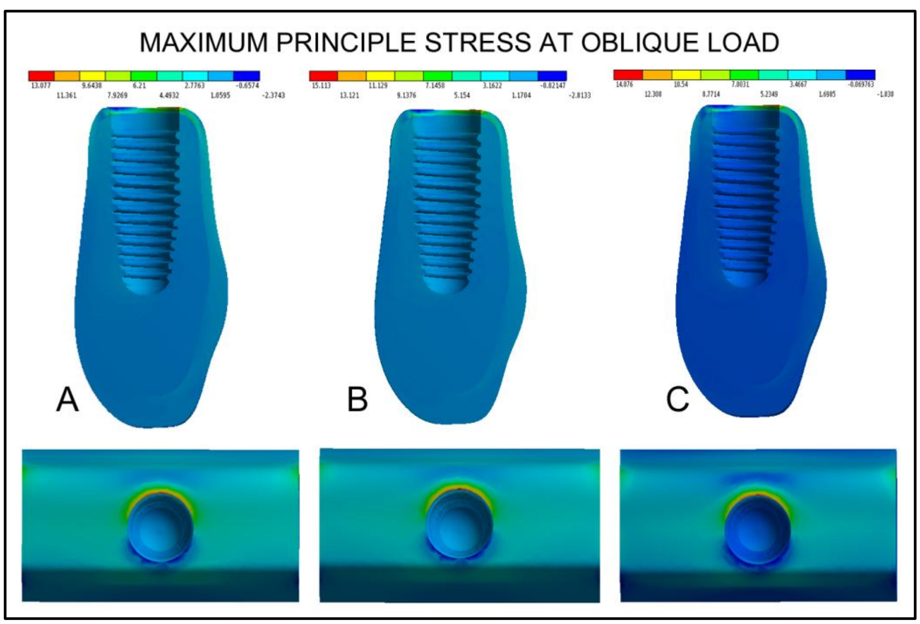

Figure 6: Maximum principle stress at oblique load seen in compact and cancellous bones. A, cross-sectional and top view of control model, B, cross-sectional and top view of the second model and C, cross-sectional and top view of the third model.

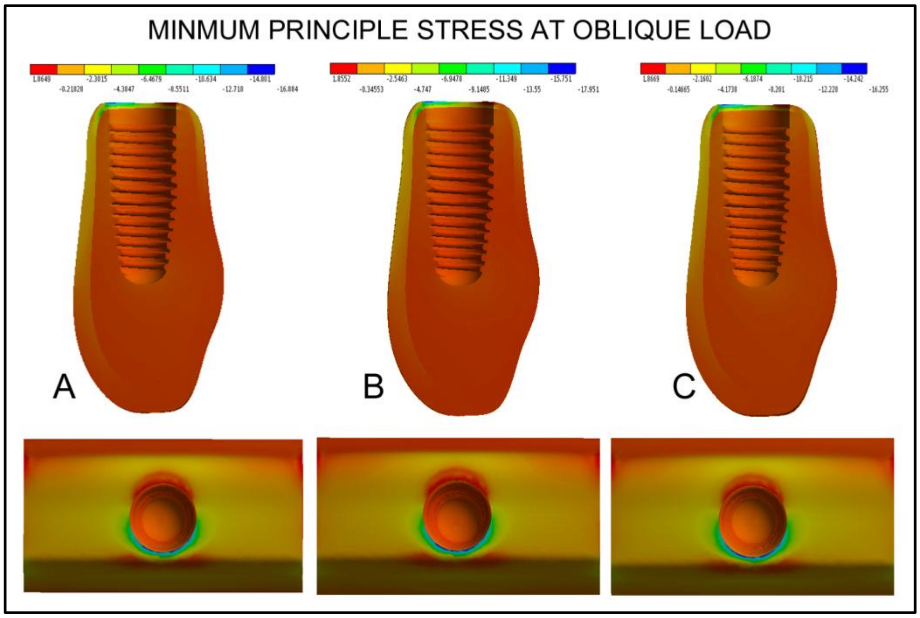

Figure 7: Minimum principle stress at oblique load seen in compact and cancellous bones. A, cross-sectional and top view of control model, B, cross-sectional and top view of the second model and C, cross-sectional and top view of the third model.

Discussion

Implant longevity is a major demand for all implantologists. Great efforts were exerted along several years of research to formulate clear and conclusive criteria of the optimal design, material, surface topology, and procedures for implantation. The proper design and characteristics of implant-abutment connection or engagement are one of the priorities if durable implant would be considered. Consequently, special concern of that connection might change the overall loading and distribution of generated stress in the bone surrounding the fixture (Porter & Von Fraunhofer, 2004; Werking, 2009).

The current study focused on using two different strategies to modify stress values and distribution in the proposed designs. The first strategy based on reducing stress transfer from abutment to fixture at the crestal zone. This area is considered a key for most stress concentration the surrounding bone expected to bear (Heitz-Mayfield et al., 2013; Kitamura et al., 2004; Valderrama et al., 2011). Thus, any design that is capable to minimize stress at crestal bone will be desirable. Consequently, the strategy was relied on minimizing contact between fixture and abutment at this critical part and transfer load to a safer position near the center of the fixture. This strategy was also expected to get benefit of changing stress distribution by changing the position of actual fulcrum of fixture movement. However, unfortunately the proposed design was not sufficiently able to produce the desired effect. The main cripples were two challenging points. Firstly, reliving the fixture-abutment contact at this critical area requires incomplete seal at crestal bone, which is unacceptable in any sound design (Schwarz, Hegewald, & Becker, 2014; Seetoh et al., 2011; Steinebrunner et al., 2004). Secondly, transferring the load to the center of the fixture resulted in transferring more load on the fixture-screw interface, which is another stress critical area (Pessoa et al., 2010; Seetoh et al., 2011), especially when applying the oblique load. Even during the vertical load, the stress was transferred directly to the fixture through its minimal contact with abutment.

The second strategy was based on using shock absorber or damping effect by using a polymeric material at fixture-abutment interface. A similar strategy was used effectively in articles to get homogenous support in implant-tooth connection cases (Anders, 1995; Carvalho et al., 2004; Chapman & Kirsch, 1989; Chee & Jivraj, 2006; Richter et al., 1990). This strategy was partially successful to minimize stress on bone at crested area during applying vertical load. This finding was in agreement with (Chee & Jivraj, 2006), (Carvalho et al., 2004) and (Anders, 1995).

Although it could not do the same effect during oblique load. The polymeric material seems to be crashed early and submitted that load to the fixture and subsequently to the crested bone. Accordingly, the mechanical properties of the proposed shock absorber material should be revised to select a material that are capable to perform its function whatever the load direction applied.

As an interpretation of the positive results of the control design, the full contact between fixture and abutment looks to distribute the stress over a surface area at that interface. Although, overloading of the crestal bone could not be avoided by control design.

The current study had certain limitation such as the type of contact used between all interfaces. If frictional contact was selected, it is expected to change stress distribution. The bonded contact was selected an order to facilitate calculation process and the frictional modulus of some contact are lacking in literatures (Murakami & Wakabayashi, 2014). Another limitation was the use of dynamic loading rather than the static loading, which may also affects the results (Duyck et al., 2001; Hoyer et al., 2001; Hussein, 2013).

Some recommendations could be mentioned to improve the studied designs by selecting efficient material as a shock absorbs or using different material properties for abetment itself to do the shock absorber function.

The current Study suggested two proposed designs to minimize crestal bone stress and opened the field for new designs able to perform stress-releasing action. Subsequently, it could be concluded that, although the proposed designs were not fully successful to reduce crestal bone stress but they changed the stress distribution of the conventional design.

References

Anders, I. (1995). Dental implant with shock absorbent cushioned interface: Google Patents.

Carvalho, L., Vaz, M., & Simões, J. (2004). Mandibular strains induced by conventional and novel dental implants. The Journal of Strain Analysis for Engineering Design, 39(3), 291-297.

Chapman, R., & Kirsch, A. (1989). Variations in occlusal forces with a resilient internal implant shock absorber. The International journal of oral & maxillofacial implants, 5(4), 369-374.

Chee, W., & Jivraj, S. (2006). Connecting implants to teeth. British dental journal, 201(10), 629-632.

Covani, U., Ricci, M., Tonelli, P., & Barone, A. (2013). An evaluation of new designs in implant-abutment connections: a finite element method assessment. Implant dentistry, 22(3), 263-267.

Dittmer, S., Dittmer, M. P., Kohorst, P., Jendras, M., Borchers, L., & Stiesch, M. (2011). Effect of implant–abutment connection design on load bearing capacity and failure mode of implants. Journal of Prosthodontics, 20(7), 510-516.

Duyck, J., Naert, I., Rønold, H. J., Ellingsen, J. E., Van Oosterwyck, H., & Vander Sloten, J. (2001). The influence of static and dynamic loading on marginal bone reactions around osseointegrated implants: an animal experimental study. Clinical oral implants research, 12(3), 207-218.

Freitas-Júnior, A. C., Rocha, E. P., Bonfante, E. A., Almeida, E. O., Anchieta, R. B., Martini, A. P., . . . Coelho, P. G. (2012). Biomechanical evaluation of internal and external hexagon platform switched implant-abutment connections: An in vitro laboratory and three-dimensional finite element analysis. Dental Materials, 28(10), e218-e228.

Freitas Jr, A. C., Bonfante, E. A., Giro, G., Janal, M. N., & Coelho, P. G. (2012). The effect of implant design on insertion torque and immediate micromotion. Clinical oral implants research, 23(1), 113-118.

Gracis, S., Michalakis, K., Vigolo, P., Vult von Steyern, P., Zwahlen, M., & Sailer, I. (2012). Internal vs. external connections for abutments/reconstructions: a systematic review. Clinical oral implants research, 23(s6), 202-216.

Heitz-Mayfield, L., Darby, I., Heitz, F., & Chen, S. (2013). Preservation of crestal bone by implant design. A comparative study in minipigs. Clinical oral implants research, 24(3), 243-249.

Hoffmann, O., & Zafiropoulos, G.-G. (2012). Tooth-implant connection: a review. Journal of Oral Implantology, 38(2), 194-200.

Hoyer, S. A., Stanford, C. M., Buranadham, S., Fridrich, T., Wagner, J., & Gratton, D. (2001). Dynamic fatigue properties of the dental implant–abutment interface: joint opening in wide-diameter versus standard-diameter hex-type implants. The Journal of prosthetic dentistry, 85(6), 599-607.

Hussein, M. O. (2013). Biomechanical Analysis of Different Implant-Overdenture Loading Protocols under Dynamic loads. Journal of American Science, 9(12).

Kitamura, E., Stegaroiu, R., Nomura, S., & Miyakawa, O. (2004). Biomechanical aspects of marginal bone resorption around osseointegrated implants: considerations based on a three-dimensional finite element analysis. Clinical oral implants research, 15(4), 401-412.

Lin, M.-I., Shen, Y.-W., Huang, H.-L., Hsu, J.-T., & Fuh, L.-J. (2013). A Retrospective Study of Implant–Abutment Connections on Crestal Bone Level. Journal of dental research, 92(12 suppl), 202S-207S.

Morton, D., Stanford, C. M., & Aquilino, S. A. (1998). Evaluation of resilient abutment components on measured strain using dynamic loading conditions. The Journal of prosthetic dentistry, 80(1), 46-51.

Murakami, N., & Wakabayashi, N. (2014). Finite element contact analysis as a critical technique in dental biomechanics: A review. Journal of prosthodontic research, 58(2), 92-101.

Okumura, N., Stegaroiu, R., Kitamura, E., Kurokawa, K., & Nomura, S. (2010). Influence of maxillary cortical bone thickness, implant design and implant diameter on stress around implants: A three-dimensional finite element analysis. Journal of prosthodontic research, 54(3), 133-142.

Peñarrocha-Diago, M., Flichy-Fernández, A., Alonso-González, R., Peñarrocha-Oltra, D., Balaguer-Martínez, J., & Peñarrocha-Diago, M. (2013). Influence of implant neck design and implant–abutment connection type on peri-implant health. Radiological study. Clinical oral implants research, 24(11), 1192-1200.

Pessoa, R. S., Muraru, L., Júnior, E. M., Vaz, L. G., Sloten, J. V., Duyck, J., & Jaecques, S. V. (2010). Influence of Implant Connection Type on the Biomechanical Environment of Immediately Placed Implants–CT-Based Nonlinear, Three-Dimensional Finite Element Analysis. Clinical implant dentistry and related research, 12(3), 219-234.

Porter, J., & Von Fraunhofer, J. (2004). Success or failure of dental implants? A literature review with treatment considerations. General dentistry, 53(6), 423-432; quiz 433, 446.

Richter, E.-J., Orschall, B., & Jovanovic, S. (1990). Dental implant abutment resembling the two-phase tooth mobility. Journal of biomechanics, 23(4), 297-306.

Schwarz, F., Hegewald, A., & Becker, J. (2014). Impact of implant–abutment connection and positioning of the machined collar/microgap on crestal bone level changes: a systematic review. Clinical oral implants research, 25(4), 417-425.

Seetoh, Y., Tan, K. B., Chua, E., Quek, H., & Nicholls, J. I. (2011). Load fatigue performance of conical implant-abutment connections. The International journal of oral & maxillofacial implants, 26(4), 797.

Shafie, H. R., & White, B. A. (2014). Different Implant–Abutment Connections. Clinical and Laboratory Manual of Dental Implant Abutments, 33-46.

Steinebrunner, L., Wolfart, S., Bößmann, K., & Kern, M. (2004). In vitro evaluation of bacterial leakage along the implant-abutment interface of different implant systems. The International journal of oral & maxillofacial implants, 20(6), 875-881.

Truninger, T. C., Stawarczyk, B., Leutert, C. R., Sailer, T. R., Hämmerle, C. H., & Sailer, I. (2012). Bending moments of zirconia and titanium abutments with internal and external implant–abutment connections after aging and chewing simulation. Clinical oral implants research, 23(1), 12-18.

Valderrama, P., Bornstein, M. M., Jones, A. A., Wilson Jr, T. G., Higginbottom, F. L., & Cochran, D. L. (2011). Effects of implant design on marginal bone changes around early loaded, chemically modified, sandblasted acid-etched-surfaced implants: a histologic analysis in dogs. Journal of periodontology, 82(7), 1025-1034.

Werking, C. M. (2009). Risk Factors in Implant Dentistry: Simplified Clinical Analysis for Predictable Treatment. Journal of Prosthodontics, 18(4), 370-371.