The Effect of Implant-supported All Ceramic Cantilever Bridge on Bone Remodelling

C. Rungsiyakull[1, a], P. Rungsiyakull[2], Q. Li[1], W. Li[1], M.V. Swain[2]

[1]School of Aerospace, Mechanical and Mechatronic Engineering
[2]Faculty of Dentistry
The University of Sydney, Sydney, NSW 2006, Australia

Email: c.rungsiyakull@usyd.edu.au

Available Online at: www.austceram.com/ACS-Journal-2009vol1.asp

Abstract
This paper aims at providing a preliminary understanding in a biomechanics effect of dental implant supported cantilever bridge on dental bone remodelling. The dental crown is made of all-ceramics. The 2D finite element model contains ceramic bridge, implant fixture, abutment and a section of bone which were constructed bases on computerised tomography (CT) images. To facilitate the study in bone remodelling in this paper, the strain energy density (SED) obtained from 2D plane stress finite element analysis (FEA) model is used as the mechanical stimulus to determine the bone turn-over in the cancellous and cortical bones. The results examined the changes in bone density and stiffness as a result of bone remodelling over a period of 48 months. The displacement of the bridge was also taken into account to explore the relationship between the stiffness of the structure and bone remodelling.

Keywords: Bone remodelling, Cantilever Bridge, Dental Implant, Osseointegration, Ceramic crown, Finite element.

1. INTRODUCTION

In 1969, Brånemark and colleagues reported a successful attempt of endosseous dental implant treatment [1]. Since then, osseointegrated dental implants have had a profound influence in contemporary dentistry and been clinically accepted to achieve the desirable outcomes for the management of partially and fully edentulous patients. Despite significant success in clinic, implant failures and various complications have been reported continuously [2]. Occlusal overloading and poor bone quality at the implant site are significant factors related to the implant failures. As more and more extensive applications of implant supported dental prostheses, an indepth understanding in this respect appears urgent [3]. The incorporation of cantilever extensions in implant-supported fixed partial dentures (FPDs) is sometimes one of the essential alternatives in case that unfavourable local condition, such as bone defects, must be involved. Cantilever configurations appear an effective alternative in both the partially edentulous and the full-arch situations. In general, the full-arch implant-supported fixed prostheses have demonstrated relatively high long-term successful rate in the patient with edentulous mandible [4]. However, these treatment modalities completely change the mechanical loading environment. It is unclear yet how such a mechanical change could determine the biological remodelling of bone in a way of either apposition or resorption near the implant site [5]. In-vitro studies, including finite element analysis, revealed that the stress occurred mainly at the bone crest adjacent to the distal surface of the implant that was facing the cantilever extension [6, 7]. The cantilever length of the extension is an important factor that affects on the stress concentration [5, 8]. Several animal studies have investigated on the reaction of the peri-implant bone loss to the mechanical load [9-12]. It was demonstrated that marginal bone resorption was not enhanced by static load [12], continuous load [11], or long-term excessive load [10]. Although several studies failed to demonstrate the relationship between marginal bone resorption and mechanical loading, the study by Isidor et al. [9] revealed that overload induced by non-axial forces caused the loss of osseointegration in a monkey sample over an 18-month period.

Clinical studies have recommended a relationship between the excessive loading and peri-implant bone loss [5, 13]. Several studies have been performed to analyse whether or not a cantilever extension on an implant-supported fixed dental prosthesis increases the amount of peri-implant bone loss [14-16]. For the implant-supported full-arch fixed prostheses, a 6-year follow-up study by Lindquist et al. [17] found that the length of cantilever significantly enhanced the amount of peri-implant bone loss at the anterior region. Their subsequent study finally reported that the cantilever extension might not jeopardize the stability of the peri-implant bone level in 15 years [18]. For the implant-supported partially fixed prostheses, some additional studies were conducted [14, 15, 19]. Romeo et al. [19] reported the overall cumulative implant survival rate was 97%, and the prostheses success rate was 98%. Mesial cantilever prostheses registered a lower success rate (97.1%) than distal cantilever prostheses (100%). They also concluded that there was a comparable medium-term prognosis between conventional FPDs and cantilever FPDs. Another study by Wennstrőm et al. [15] was also in an agreement with an observation that the presence of cantilever extensions did not have an effect on peri-implant bone loss over a 5-year follow-up examination. Although recent medium-term study[14] reported the mean peri-implant bone loss for the FDPs with cantilevers was 0.23 mm and 0.09mm for FDPs without cantilever, there were no statistically significant differences found between these two groups. Cantilever on FDPs may not lead to a higher implant failure rate and more bone loss around supporting implants compared with implants supported conventional FDPs. However, technical complications were observed in the cantilever group. Cantilevers, bucco-lingual offset of the restoration, and excessive height of the abutment-crown with complex lateral forces will each increase the stress on the implant surface. The in-vitro study reported that the weakest region appeared across the pontic–abutment interface in the composite bridges [20], and this result was confirmed with the numerical analysis that a high stress concentration occurs around the incisal portion of the adhesive interfaces between the pontic and abutment.[21]

Bone remodelling is defined as a process where bone changes its internal microstructure and external morphology to adapt to the new loading conditions. It was Wolff who first related the bone microstructure to the mechanical loading from a mechanics perspective. A certain change in the mechanical loading and its transfer will alter the bone's response and turnover. Mechanical stress can have both positive and negative consequences for bony tissues [22]. The longevity and stability of dental implants can be improved if a positive healing process takes place. A literature review conducted by Isidor [23] reported that bone is believed to function within the strain range of approximately 50 – 1500 microstrains [22]. If the peak load on bone results in strains of 1500 – 3000 microstrains, a mild overload occurs and is compensated by gaining more bone. If the strains exceeds a threshold, i.e. more than 25,000 microstrains, severe bone micro-fracture can suddenly occur, leading to a bone loss. On the other hand, if the bone strain does not exceed 50–100 microstrains, bone is not properly stimulated and will result in a net loss of bone density. Thus, an optimal functional strain is essential to maintain the bone mass and density, thereby contributing to stabilization of dental implantation.

Mechanically, when occlusal loads are applied to a dental crown, they will be transferred to the bone through the implant. When there are two materials with different mechanical properties (i.e. stiffness), between dental implants and bone, the stress will be highest where the materials have first contact. The highest stress will therefore be expected in the most coronal portion of the supporting bone. Ideal restorations can promote the positive bone remodelling and minimize the healing time. However, it is a challenge to predict whether the loaded implant will induce bone remodelling properly. Superstructure designs of dental implants, especially cantilever design, may have an impact on bone strains around dental implants when occlusal loads are applied. However, the relation between the existing occlusal design and bone remodelling has not been well published. This study will explore the effect of two-unit cantilever fixed dental prostheses on bone remodelling by using the finite element method (FEM) as a tool for remodelling simulation. The role of cantilever bridge on bone's response is justified and therefore to provide a new and effective approach to the improvement of restorative longevity.

2. MATERIALS AND METHODS

2.1 Finite element model

The 2D model of a two-unit implant supported an all-ceramic cantilever bridge which consists of dental implant fixture and implant abutment; all-ceramic bridge and a section of bone were constructed as shown in Fig. 1. The dental implant fixture was based on Brånemark system® (Noble Biocare, Sweden) Mark III model Regular Platform (RP) with a 3.75 mm diameter and 10 mm long.

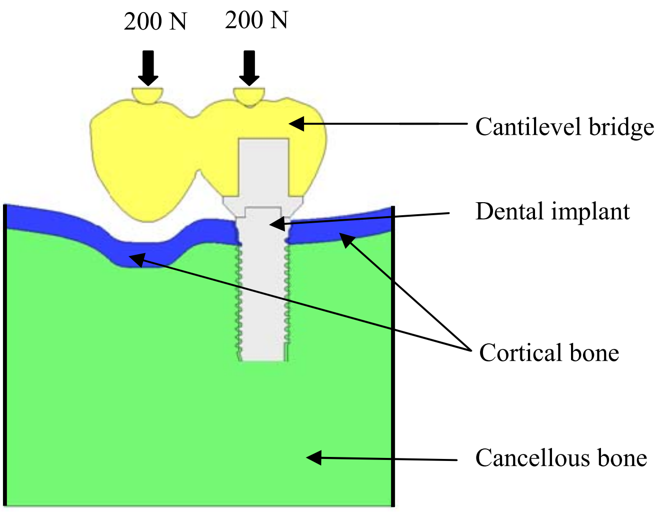


Fig. 1 Two-dimensional plane stress finite element model

Table 1: Material Properties [28, 29]

Materials	**Young's modulus (GPa)**	**Poisson's Ratio**
Titanium alloy (Ti6A14V)	110	0.35
Ceramic	63	0.33
Cortical Bone (Initial stage)	13.7	0.30
Cancellous Bone (Initial stage)	1.37	0.30

The abutment selected for this study was a titanium temporary abutment, Brånemark system® regular platform which locked onto the implant fixture. The implant fixture and abutment model were reconstructed by using SolidWorks 2007 (SolidWorks Corp. USA). A mandibular section of bone was constructed by using the images from computerised tomography (CT) scan technology which then was processed in Rhinoceros 3D (Robert McNeel & Associates, Seattle, USA) where a mesial-distal cross sectional view was then produced. The bone section was modelled with 2 segments; the upper shell with average thickness around 1.5 mm representing the cortical bone and the rest representing the cancellous bone which was assumed to be perfectly bonded with a cortical layer.

This 2D FE model represented a plane stress problem with a thickness of 13 mm from the averaged CT data in the aspect. The deformation in the implant along cervical-apical was expected. Therefore, both side edges of the model were constrained as the boundary condition. A mechanical load of 200 N was applied on top of the bridge at two locations as shown in Fig. 1. A ball loading was used in order to avoid the stress singularity at loading point [24]. The frictional contact was used for the interface implant fixture and bone region and between the bridge and dental abutment. The friction coefficient and shear modulus were set as 1.0 and 240 MPa [25]respectively. The interface between implant fixture and abutment and between cortical and cancellous bone were assumed to be perfectly bonded.

The model was meshed using 6 node quadratic triangular elements (CPS3). A global element edge length of 0.2 mm was implemented after taking a convergence study similarly to [26], whereby model accuracy and computational cost were optimal. All the biomaterials associated were presumed to be linear elastic, homogenous, and isotropic for the sake of simplification in the remodelling analysis [27]. The corresponding properties such as Young’s modulus and Poisson ratio were determined from the literature [28, 29] and are summarized in Table 1. All the FE analyses were performed by using ABAQUS software (Dassault Systemes 2008) version 6.7.1 on a personal computer (Intel Quad Core 3GHz).

2.2 Bone remodelling algorithm

Adaptive bone remodelling processes have been studied by many researchers in long bone community since 1970s, which all assumed that this process was regulated by mechanical stimuli detected by some internal sensors within the bone. Once the mechanical loading condition deviates from the specific homeostatic level, bone will respond by changing its density. The changing rate of the apparent density (ρ) is based on the differences between the remodelling stimulus (S) and the thresholds of bone remodelling (i.e. $K(1+s)$ and $K(1–s)$) [30].

Frost [31] suggested that if there is a small deviation in the mechanical stimulus, the bone remodelling process would not occur, thus an inhibitory value or threshold was proposed. As such, either bone apposition or resorption will not take place as long as the stimulus signal is in between these inhibitory threshold values. This hypothesis was incorporated with bone remodelling theories later and was named as a "lazy zone" or "dead zone" [30, 32, 33], as shown in Fig. 2. Mathematically, the remodelling algorithm can be expressed as,

$$\frac{d\rho}{dt} = \begin{cases} C_f\left[S - K(1+s)\right]; S > K(1+s) \\ 0 \qquad ; K(1-s) \le S \le K(1+s) \\ C_r\left[S - K(1-s)\right]; S > K(1-s) \end{cases} \tag{1}$$

where C_f and C_r are the rate constants for apposition and resorption, respectively, t is time, s is the width of lazy zone and K is assumed as the reference force for adaptive activity.

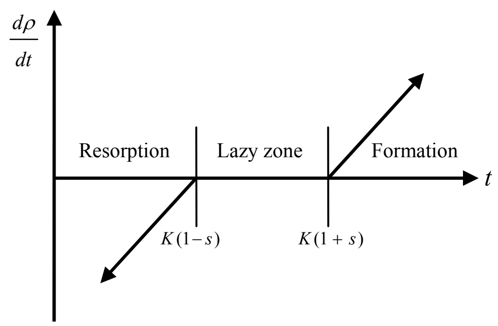


Fig. 2 Bone remodelling algorithm bases upon the Eq. (3)

Different mechanical stimuli have been proposed for long bone, like femur or tibia, remodelling activities. The strain energy density (SED) has been proven fairly effective in predicting bone's biological responses and will be adopted in this paper [25]. Therefore, the remodelling stimulus S can be obtained by the following calculation of SED per unit density, as

$$S = \frac{U}{\rho} \tag{2}$$

where U is the strain energy density (strain energy per unit of volume) and ρ is the local density (bone mass per unit of volume). Hence, Eq. (1) then becomes:

$$\frac{dp}{dt} = \begin{cases} C_f\left[\frac{U}{\rho} - K(1+s)\right] \text{ if } \frac{U}{\rho} > K(1+s) \\ 0 \qquad \text{ if } K(1-s) \le \frac{U}{\rho} \le K(1+s) \\ C_r\left[\frac{U}{\rho} - K(1-s)\right] \text{ if } S < K(1-s) \end{cases} \tag{3}$$

In the FE framework, SED can be calculated from averaging the strain energy density values at all the integration points of an element,

$$U = \frac{1}{2}\{\sigma_{ij}\}\{\varepsilon_{ij}\} \tag{4}$$

where U is the strain energy, σ_{ij} and ε_{ij} are the stress and strain tensors, respectively.

The linear relationship between elastic modulus (E) and density (ρ) for the cortical bone is formulated [25]

$$E = -23.93 + 24\rho \tag{5}$$

where the elastic modulus (E) is expressed in GPa and density (ρ) in g/cm^3 [34]. O'Mahony et al [35] proposed the relationships of the cancellous bone density and elastic modulus as

$$E = 2.349\rho^{2.15} \tag{6}$$

where the elastic modulus (E) is measured in GPa and density (ρ) in g/cm^3.

These formulations have been implemented in an ABAQUS environment by using Python programming language for facilitating the bone remodelling calculation.

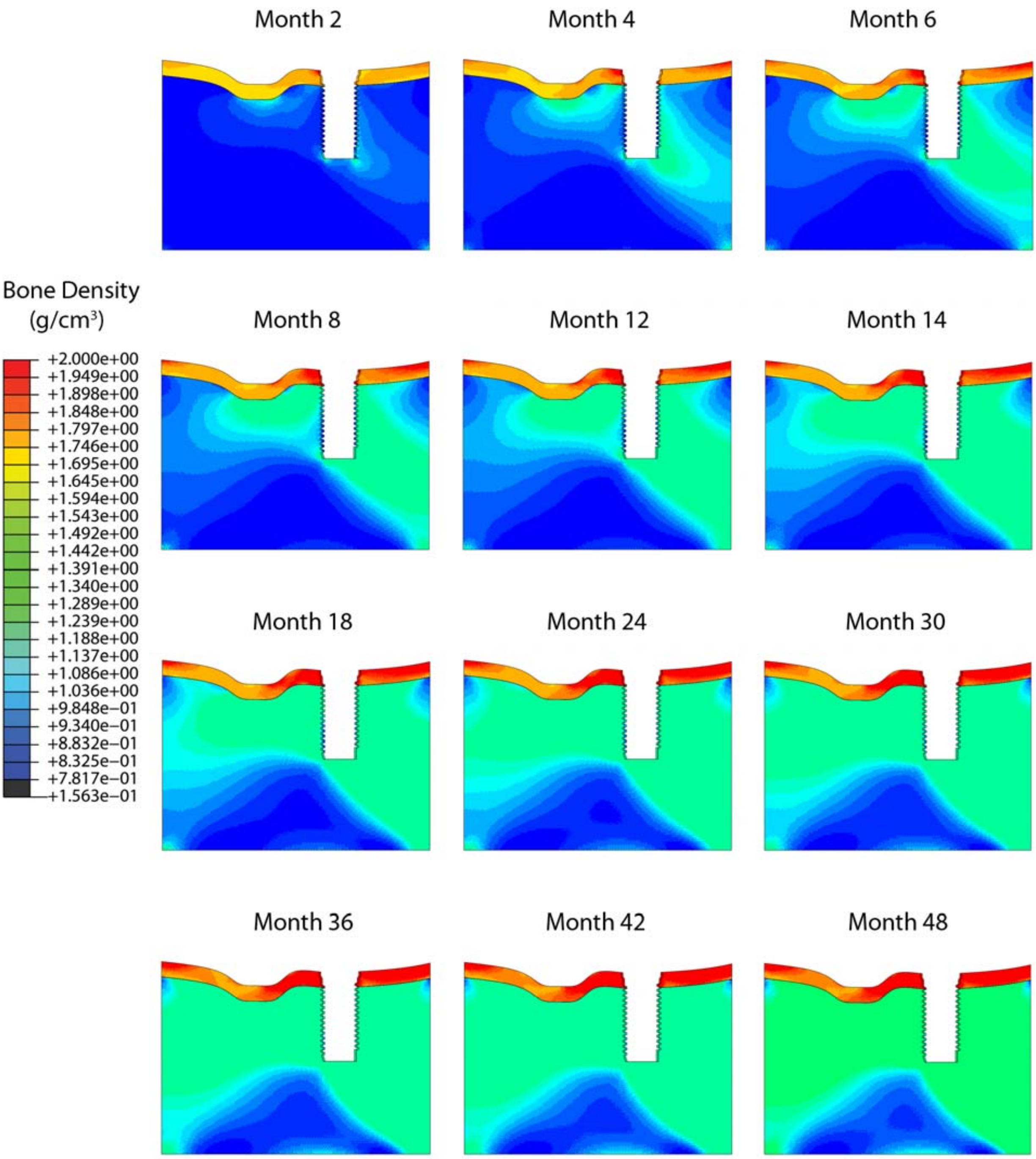
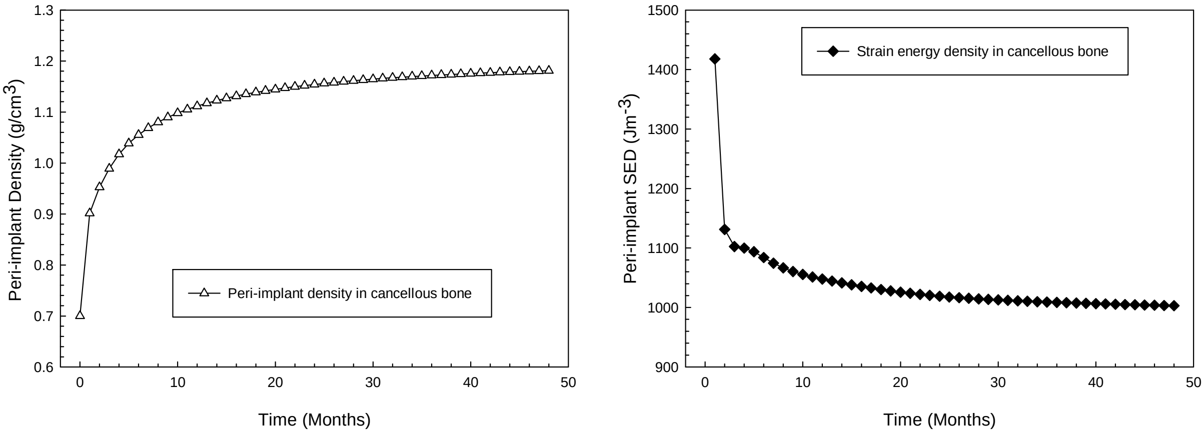
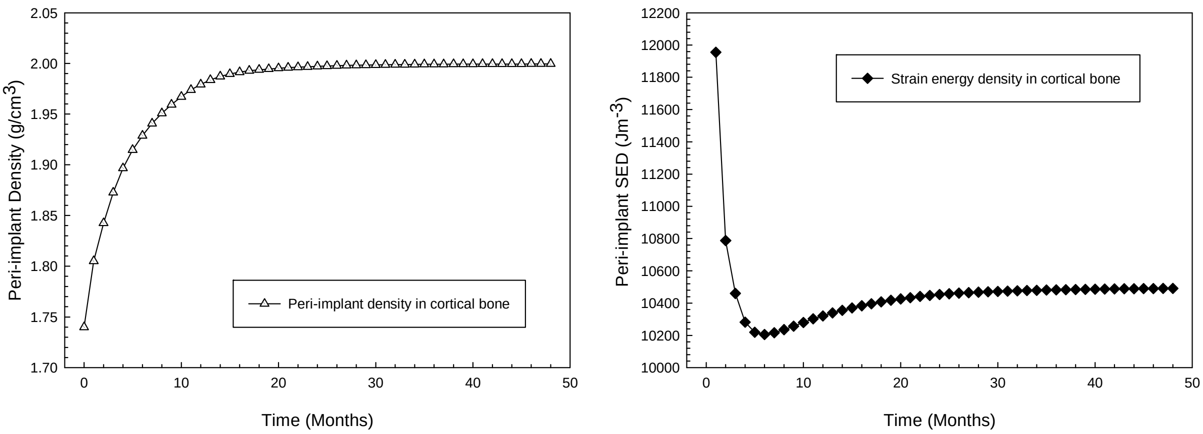


Fig 3 The contour of bone density during the bone remodelling period



a) Density progression *b) SED progression*

Fig. 4 Average peri-implant density and strain energy density in the cancellous region



a) Density progression *b) SED progression*

Fig. 5 Average peri-implant density and strain energy density in the cortical region

3. RESULTS AND DISCUSSION

In this study, 48 months are chosen as the remodelling duration, in which the changes in the bone density, strain energy density and stiffness are compared to investigate the bone remodelling activities. Fig. 3 plots the overall density distributions in the cancellous and cortical bone during remodelling process. Overall, it can be clearly seen that bone density increases over the period of time which provides the evidence that bone has been engaged and the remodelling activities take place. The magnitude of bone density is gradually increasing, especially within the first 12 months, and then the remodelling rate gradually decreases until it becomes stabilised after 30 months, which implies that the remodelling progress approaches to an equilibrium status of bone turnover. Furthermore, the progression of density seems to initiate from the peri-implant region, confirming that the most concentrate region of mechanical stimulus is around the implant fixture. After month 48 of remodelling, the majority of the cancellous region is occupied by the denser bone which would imply the signal of implantation success.

It can be observed that the bone density contour at the bottom of the cancellous region has a triangular shape with the lower magnitude of bone density. This is because the mechanical load is transferred from the implant fixture to the boundary on both sides and the flexure of the mandible is negligible.

The change of the density in peri-implant cancellous region was plotted in Fig. 4a. It can be seen that the density of the peri-implant in the cancellous region increases throughout the remodelling period, indicating that the bone in this region responds to the occlusal loading positively by adapting its morphology to accommodating the new biomechanical environment. The progression of peri-implant density growth may also represent the degree of osseointegration as well. In general, the greater the bone density, the better the implant stability. It is observed that the increase rate of density is high in the first few months, particularly from month 0 to month 4, and then the rate gradually decreases until the end of the remodelling period concerned, where the average bone density of 1.18 g/cm^3 can be archived. It should be pointed out that the bone loss due to over load stress which occurs clinically is not considered in this present work for simplicity. However, it would not be difficult to incorporate this feature in the future investigations.

Fig. 4b exhibits the progression of strain energy density (SED) over 48-month period. The strain energy density magnitude decreases sharply in the first three months and then the rate gradually decreases until the end of the remodelling period, suggesting that the mechanical strain concentration in the peri-implant cancellous region can be greatly affected by the bone apposition in the first few months as a result of initial osseointegration and initial stability of the implantation, which is in a good agreement with the progression of bone density in the cancellous region.

The change of the density in peri-implant cortical region in Fig. 5a also exhibits a similar magnification pattern to that in the cancellous region as shown in Fig. 4a. Overall the average density increases dramatically within the first 12 months then becomes stable onward. The average peri-implant density after month 48 reaches 1.99 g/cm^3, which is slightly less than the maximum cortical bone density of 2.0g/cm^3 suggested by the published data [36]. It is interesting to note that the remodelling rate of the cortical bone seems to reach an equilibrium as early as month 12 while those of cancellous bone seems to become stabilised at a later stage. This implies that the bone in cortical region is more sensitive to the mechanical load than that in the cancellous area. Therefore, bone turnover requires a shorter period of time to approach equilibrium for healing process.

Fig. 5b presents the SED in the cortical region which also represents the mechanical stiffness of the structure. The magnitude of SED decreased sharply in the first 6 months and then recovered a little before reaching equilibrium at month 24. This may be because such a very high strain concentration in the peri-implant may not be fully relaxed by the bone apposition process due to the maximum density of 2.0 g/cm^3 which was applied in the oral bone remodelling algorithm. After month 48 remodelling progression, the averages of SED in the cortical region dropped by 12.2% compare with 29.3% in cancellous region. It is interesting to note that, unlike the change in the density where the bone in the cortical region shows a quicker morphological response to the mechanical load, the SED in cortical region has little change compared with that in the cancellous region. This may be because the Young's modulus in the cancellous bone has a nonlinear relationship to the bone density as stated in Eq. 6, indicating that a small change in the bone density may result in a great change in the Young's modulus value which finally affects the magnitude of resultant SED or the mechanical stiffness of the structure.

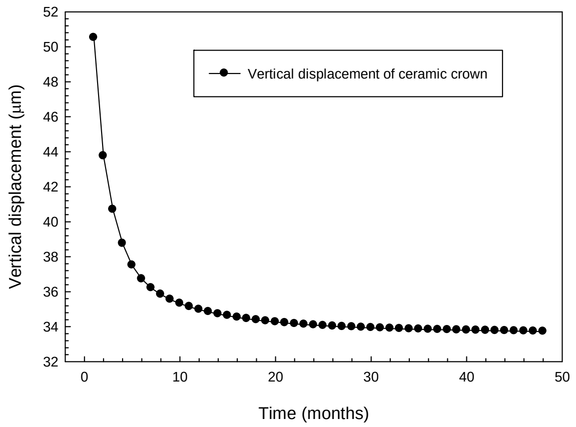


Fig. 6 The vertical displacement on the crown of the ceramic bridge

Fig. 6 represents the displacement of the ceramic bridge. The displacement shows the same deduction trend as SED, representing a measure of structural stabilities and stiffness. From mechanical perspective, the higher the SED, the lower the stiffness or the less stable the structure.

It is obvious that the occlusal displacements decrease rapidly within the first year then converge to the stable values thereafter, which exhibits the same pattern as the SED.

The stiffness of the bridge after 12 months of the remodelling period increases by 30.7% while the total increasing of 33.2% is archived after the 48-month period. This implies that the remodelling process in the first year is crucial for the initial healing of the bone after surgery and initial stability of the implantation which is in a good agreement with the experimental reports and clinical studies.

4. CONCLUSIONS

Within the limitation of this study, the numerical simulations showed the remodelling consequence of an implant supported cantilever bridge (fixed partial denture). The bone in the peri-implant cortical region seems to be more sensitive to the biomechanical responses than that in the peri-implant cancellous region. Overall, the remodelling rate in both the regions is very high in the first few months, which confirms that it is an important healing period for the bone to establish a proper osseointegration and initial stability after implantation.

However, the present remodelling algorithm allows the high strain concentration in the bone to be relaxed by increasing the bone density. Therefore, the greater mechanical loading that transferred onto the supporting bone will results in greater magnitude of bone density, which clinically, may not true because an overload may damage the bony tissues. The bone resorption due to overload stress may need to be incorporated with the bone remodelling algorithm in the future studies.

5. ACKNOWLEDGEMENTS

This study has been supported by Australian Research Council (ARC) and the first author is grateful the Faculty Scholarship at The University of Sydney.

6. REFERENCES

1. Branemark, P.I., et al., *Intra-osseous anchorage of dental prostheses. I. Experimental studies.* Scandinavian Journal Plastic Reconstruction Surgery 1969. **3**: p. 81-100.
2. Berglundh, T., L. Persson, and B. Klinge. *A systematic review of the incidence of biological and technical complications in implant dentistry reported in prospective longitudinal studies of at least 5 years*. 2002.
3. Lindh, T., et al., *A meta-analysis of implants in partial edentulism.* Clinical Oral Implants Research, 1998. **9**(2): p. 80-90.
4. Naert, I., et al., *A study of 589 consecutive implants supporting complete fixed prostheses .2. prosthetic aspects.* Journal of Prosthetic Dentistry, 1992. **68**(6): p. 949-956.
5. Rangert, B., et al., *Bending overload and implant fracture: a retrospective clinical analysis.* Int J Oral Maxillofac Implants, 1995. **10**(3): p. 326-34.
6. Rodriguez, A.M., S.A. Aquilino, and P.S. Lund, *Cantilever and implant biomechanics: a review of the literature, Part 2.* J Prosthodont, 1994. **3**(2): p. 114-8.
7. Akca, K. and H. Iplikcioglu, *Finite element stress analysis of the effect of short implant usage in place of cantilever extensions in mandibular posterior edentulism.* Journal of Oral Rehabilitation, 2002. **29**(4): p. 350-356.
8. Rodriguez, A.M., et al., *Evaluation of strain at the terminal abutment site of a fixed mandibular implant prosthesis during cantilever loading.* J Prosthodont, 1993. **2**(2): p. 93-102.
9. Isidor, F., *Loss of osseointegration caused by occlusal load of oral implants - A clinical and radiographic study in monkeys.* Clinical Oral Implants Research, 1996. **7**(2): p. 143-152.
10. Heitz-Mayfield, L.J., et al., *Does excessive occlusal load affect osseointegration? An experimental study in the dog.* Clinical Oral Implant Research, 2004. **15**: p. 259-268.
11. Barbier, L. and E. Schepers, *Adaptive bone remodeling around oral implants under axial and nonaxial loading conditions in the dog mandible.* International Journal of Oral &

Maxillofacial Implants, 1997. **12**(2): p. 215-223.

12. Gotfredsen, K., T. Berglundh, and J. Lindhe, *Bone reactions adjacent to titanium implants subjected to static load - A study in the dog (I).* Clinical Oral Implants Research, 2001. **12**(1): p. 1-8.
13. Quirynen, M., et al., *A study of 589 consecutive implants supporting complete fixed prostheses .1. periodontal aspects.* Journal of Prosthetic Dentistry, 1992. **68**(4): p. 655-663.
14. Halg, G.A., J. Schmid, and C.H.F. Hammerle, *Bone level changes at implants supporting crowns or fixed partial dentures with or without cantilevers.* Clinical Oral Implants Research, 2008. **19**(10): p. 983-990.
15. Wennstrom, J., et al., *Bone level change at implant-supported fixed partial dentures with and without cantilever extension after 5 years in function.* Journal of Clinical Periodontology, 2004. **31**(12): p. 1077-1083.
16. Romeo, E., et al., *Implant-supported fixed cantilever prostheses in partially edentulous arches. A seven-year prospective study.* Clinical Oral Implants Research, 2003. **14**(3): p. 303-311.
17. Lindquist, L.W., B. Rockler, and G.E. Carlsson, *Bone resorption around fixtures in edentulous patients treated with mandibular fixed tissue-integrated prostheses.* Journal of Prosthetic Dentistry, 1988. **59**: p. 59-63.
18. Lindquist, L.W., G.E. Carlsson, and T. Jemt, *A prospective 15-year follow-up study of mandibular fixed prostheses supported by osseointegrated implants - Clinical results and marginal bone loss.* Clinical Oral Implants Research, 1996. **7**(4): p. 329-336.
19. Romeo, E., et al., *Long-term survival and success of oral implants in the treatment of full and partial arches: A 7-year prospective study with the ITI dental implant system.* International Journal of Oral & Maxillofacial Implants, 2004. **19**(2): p. 247-259.
20. Li, W., et al., *Fibre reinforced composite dental bridge. part I: experimental investigation.* Biomaterials, 2004. **25**: p. 4987-4993.
21. Li, W., et al., *Fibre reinforced composite dental bridge. part II: numerical investigation.* Biomaterials, 2004. **25**: p. 4995-5001.
22. Frost, H.M., *A 2003 update of bone physiology and Wolff's Law for clinicians.* Angle Orthod, 2004. **74**(1): p. 3-15.
23. Isidor, F., *Influence of forces on peri-implant bone.* Clinical Oral Implants Research, 2006. **17**: p. 8-18.
24. Ichim, I., et al., *Modelling of fracture behaviour in biomaterials.* Biomaterials, 2007. **28**(7): p. 1317-1326.
25. Lin, D., et al., *Dental implant induced bone remodeling and associated algorithms.* Journal of the Mechanical Behavior of Biomedical Materials. **In Press, Corrected Proof**.
26. Li, W., et al., *Towards automated 3D finite element modeling of direct fiber reinforced composite dental bridge.* Journal of Biomedical Materials Research Part B-Applied Biomaterials, 2005. **74B**(1): p. 520-528.
27. Rungsiyakull C, L.Q., Li W, Appleyard R, Swain M, *Effect of Particle size of Fully Porous-Coated (FPC) Implants on Osseointegration.* Advanced Materials Research, 2008. **47 - 50**: p. 916-919.
28. Vaillancourt, H., R.M. Pilliar, and D. Mccammond, *Finite-Element Analysis of Crestal Bone Loss around Porous-Coated Dental Implants.* Journal of Applied Biomaterials, 1995. **6**(4): p. 267-282.
29. O'Brien, W.J., *Dental materials and their selection*. 4th ed. 2008, Hanover Park, IL: Quintessence Pub. Co. xiii, 425 p.
30. Huiskes, R., et al., *Adaptive bone-remodeling theory applied to prosthetic-design analysis.* Journal of Biomechanics, 1987. **20**(11-12): p. 1135-1150.
31. Frost, H.M., *The laws of bone structure* 1964: C.C. Thomas (Springfield, Ill).
32. Carter, D.R., *Mechanical Loading Histories and Cortical Bone Remodeling.* Calcified Tissue International, 1984. **36**: p. S19-S24.
33. Weinans, H., R. Huiskes, and H.J. Grootenboer, *Effects of Material Properties of Femoral Hip Components on Bone Remodeling.* Journal of Orthopaedic Research, 1992. **10**(6): p. 845-853.
34. Rho, J.Y., M.C. Hobatho, and R.B. Ashman, *Relations of Mechanical-Properties to Density and Ct Numbers in Human Bone.* Medical Engineering & Physics, 1995. **17**(5): p. 347-355.
35. O'Mahony, A.M., et al., *Anisotropic elastic properties of cancellous bone from a human edentulous mandible.* Clinical Oral Implants Research, 2000. **11**(5): p. 415-421.
36. O'Mahony, A.M., J.L. Williams, and P. Spencer, *Anisotropic elasticity of cortical and cancellous bone in the posterior mandible increases peri-implant stress and strain under oblique loading.* Clinical Oral Implants Research, 2001. **12**(6): p. 648-657.