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Three-Dimensional Finite Element Analysis of the Effect of Incomplete Seating of Cemented Fixed Dental Prostheses

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Abstract - *This study investigated the stress distribution patterns of two finite element models of a stylized fixed dental prosthesis-cement-abutment tooth system, one with the prosthesis completely seated and the other manipulated to be incompletely seated. Maximum equivalent von Mises stress varied according to direction and location of load, with vertical loading of the pontic of the completely-seated FDP (2.9 MPa) and oblique loading of the premolar of the incompletely-seated FDP (80.8 MPa) producing the least and the highest values, respectively. Total deformation of the restored system showed variations, although different cements had minimal effect on stress and on deformation. Under the conditions studied, a fixed dental prosthesis that had not been verified as fully seated on its abutments prior to cementation, could, with repeated loading cycles, be predicted to suffer a greater risk of fatigue, and thus clinical failure.*

KEY WORDS: Prosthesis fit, finite element analysis, fixed dental prosthesis

INTRODUCTION

The clinical and technical processes involved in the construction of a fixed dental prosthesis (FDP) should theoretically ensure its exact seating on the abutment teeth. In practice this is seldom the case, and 'clinical acceptability' is the conventional term that guides standards of care[1]. Accepted clinical protocols require that an imperfectly seated or 'rocking' FDP be corrected before cementation. Failure to do so would detract from achieving good marginal fit and casting-to-abutment adaptation. Good marginal fit of FDP retainers is generally regarded as an important requirement for successful treatment[2-4]. Marginal deficiencies have been linked to fracture and dissolution of luting cement, and secondary caries[2,5-7].

Although geometric form is widely regarded as a key factor to be designed into any non-adhesively-cemented restoration for it to resist dislodgement by functional stresses[8], some studies have questioned the absoluteness of such a relationship[9,10]. Firstly, the stabilizing property of FDP abutments with increasing taper is a *progressive,* rather than an *all-or-nothing* phenomenon, as is generally held[11]. Secondly, close adaptation of casting to abutment, and thus minimum and even cement film thickness, are key to achieving resistance; it follows that uneven and thick cement interfaces, as would occur with incompletely-seated retainers, could potentially affect resistance by altering the distribution and transmission of functional stresses within the restoration-abutment complex[12,13]. However, whereas the effects of marginal discrepancies have been investigated, little is known about the interfacial stresses that may be induced, and, more importantly, how they might contribute to clinical failure. For reasons similarly linked to possible biomechanical complications, implant restorations which exhibit 'misfit' have attracted much research interest[14-17].

The complexities of the clinical scenario described make the problem difficult and expensive to investigate by clinical trials. Two-dimensional finite element analysis (FEA) has been used to characterize the effect of occlusal forces on stress magnitudes and distribution in the luting cement beneath (fully-seated) full coverage restorations[18,19], but more advanced computer modelling techniques that allow three-dimensional (3-D) manipulation of clinical variables are needed when the structures under investigation are not easily simplified, or are non-axisymmetric[20].

This 3-D FEA study investigated the magnitude and distribution of stresses of a mathematically-modelled 3-unit posterior cast metal FDP, a variant of which was systematically manipulated to produce a controlled level of incomplete seating (or 'misfit') in relation to its abutments, after cementation with different luting cements and loading under different conditions.

MATERIALS AND METHODS

A 3-D computer-aided design (CAD) model of a stylized 3-unit gold alloy FDP, cemented onto a mandibular second premolar and second molar as abutments, was created using SolidWorks 2005. The overall dimensions of the model were 32 mm mesiodistally, 10 mm buccolingually, and 24 mm superior-inferiorly. Tooth dimensions were each 22 mm in length, while prepared crowns were 5 mm in diameter for the premolar and 9.6 mm mesiodistally and 6.8 mm buccolingually for the molar (measured at the tooth preparation margin). Abutment preparations had a 6° taper with all line angles except the cavosurface margins

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rounded to a radius of 0.5 mm. A circumferential preparation margin of 0.5 mm in width was modelled, so that minimum retainer thickness was 0.5 mm, reaching up to 2 mm in occluso-axial areas. Mean connector dimensions were 3 mm (horizontally) by 5 mm (vertically), and 4 mm (horizontally) by 5 mm (vertically) for the mesial and distal ends of the pontic, respectively. Cement film thickness was set at 0.1 mm in the fully-seated model.

The CAD model was imported into ANSYS 11.0 for conversion to a finite element (FE) model (Fig. 1). The following were assumed: 1. tooth enamel on both abutments was completely replaced by the retainers; 2. all material surfaces were perfectly smooth and without flaw; 3. interfaces were bonded perfectly, without delamination; 4. retainers, cement and teeth were homogeneous, linear elastic and isotropic; and 5. there were no fatigue effects.

A variant of the first CAD model, in which the molar abutment was positioned 0.1 mm occlusally along its long axis, represented a case of an incompletely-seated FDP, assuming the FDP to have been seated in a vertical path (as it would also be in the fully-seated scenario). A further assumption of the model was that, whereas the fully-seated model would have had a cement film thickness that was uniform for both abutments, the incompletely-seated version did not: instead, it was determined by geometrically-induced factors. Other than the systematic 'extrusion' of the molar abutment in the incompletely-seated variant, no other displacements of roots were permitted.

Zinc phosphate and resin cements were investigated, representing examples of high and low elastic moduli, respectively. Table 1 lists the mechanical properties of the materials used to construct the FE models. A vertical load of 70 N was applied alternately to the molar, the pontic and the premolar to study the effects on the structures of each of the fully- and incompletely-seated FDPs. A 45° load of the same magnitude was similarly applied to study the effect of oblique, non-axial loading. The structure was constrained at the outer surfaces of the cortical bone, while the teeth, FDP, cement and the inner surface of the cortical bone were free too deform. No periodontal ligament was included, which, being less constraining to the tooth, would have reduced the reaction to loading at the tooth and FDP levels, so leading to lower stresses. By omitting the periodontal ligament, the scenario which we studied is, therefore, the worst case in terms of induced stresses and strains. For the same purpose of investigating exclusively the passive response of the materials in question, cancellous bone was excluded from the model.

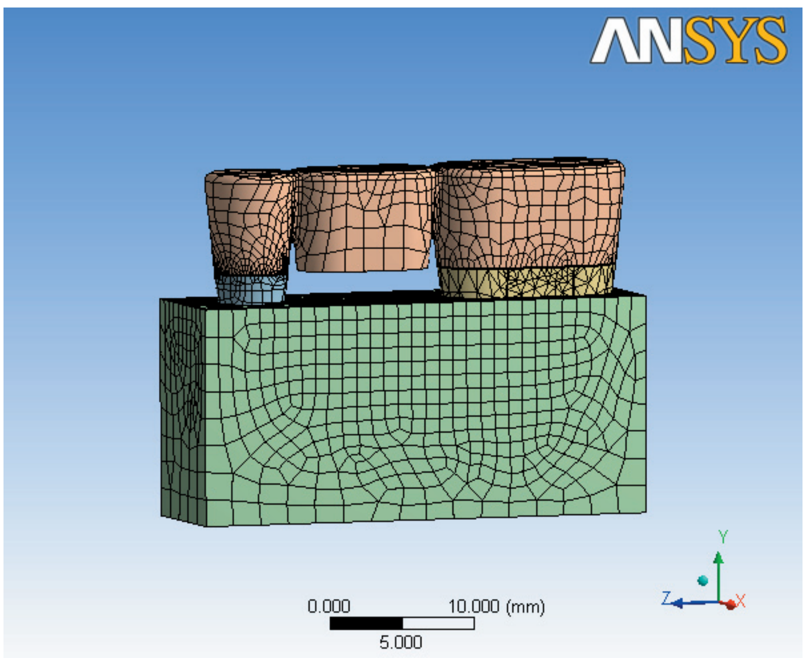


Figure 1. *Three-dimensional finite element model of fixed dental prosthesis on two abutments includes a simulated monolithic prosthesis, a cement layer, and cortical bone. No periodontal ligament, cancellous bone or pulp space were modelled.*

Table 1. *Material properties*[21]

Material	*Elastic modulus E (GPa)*	*Poisson's ratio υ*	*Ultimate tensile stress UTS (MPa)*
Dentine	16.0	0.31	
Type III gold alloy	100.0	0.33	
Zinc phosphate cement	13.7	0.33	9
Resin cement	8.0	0.33	40
Cortical bone	14.7	0.30	

Results were displayed as stress contour plots to indicate regions of maximum stress concentration. Stress distributions for each of the conditions were compared qualitatively to permit detection of factors that could be predictive of cement microfracture, and thus of clinical failure. Because of the large number of models, only summaries of the results are presented, focusing on the variables that could lead to cement microfracture. Statistical analyses in computer-simulated FEA studies are rarely applied given their inherent invariance, which also obviates the need for repeated calculations[22].

RESULTS

Maximum equivalent von Mises stress for the completely-seated scenarios subjected to a vertical load were generally low, regardless of whether load was applied to the molar, the pontic or the premolar (Table 2). Maximum stress concentrations were at the inferior aspect of connectors, and, depending on the location of loading, also at the cervical margins, albeit with a lower level of stress (Fig. 2). Total deformation in any part of the structure was minimal, but it should be recalled that the model excluded a periodontal ligament (Fig. 3). Presence of zinc phosphate or resin cements had little effect on levels of stress and deformation.

Maximum equivalent von Mises stress increased substantially when an oblique load of 45° was applied to any of the units of the completely-seated FDP. The cement would seem to accommodate the load by allowing the FDP to rotate relative to the tooth, appearing to be compressed on the loaded side, with opening of the margin on that side; on the opposite side there is closure of the margin (Fig. 4). Deformation was also greater than in the vertically-loaded system.

In the incompletely-seated scenario, with the molar positioned occlusally by 0.1 mm, loading the molar vertically caused it to behave much as in the fully-seated situation. Little stress was produced at the premolar because the deformation was taken up at the connector between the pontic and premolar. However, as load was applied more anteriorly on the pontic and the premolar, stress distributions changed substantially in the presence of a 'cantilever-like' situation (and which the absence of a periodontal ligament also accentuates). In this scenario, maximum equivalent von Mises stress ranged between 7.2-46.9 MPa (Table 2).

The biomechanics became more complex when the incompletely-seated FDP was loaded obliquely, with maximum equivalent von Mises stress increasing substantially to 80.8 MPa (concentrated at the connectors) when the premolar of the resin-cemented FDP was loaded (Fig. 5). Total

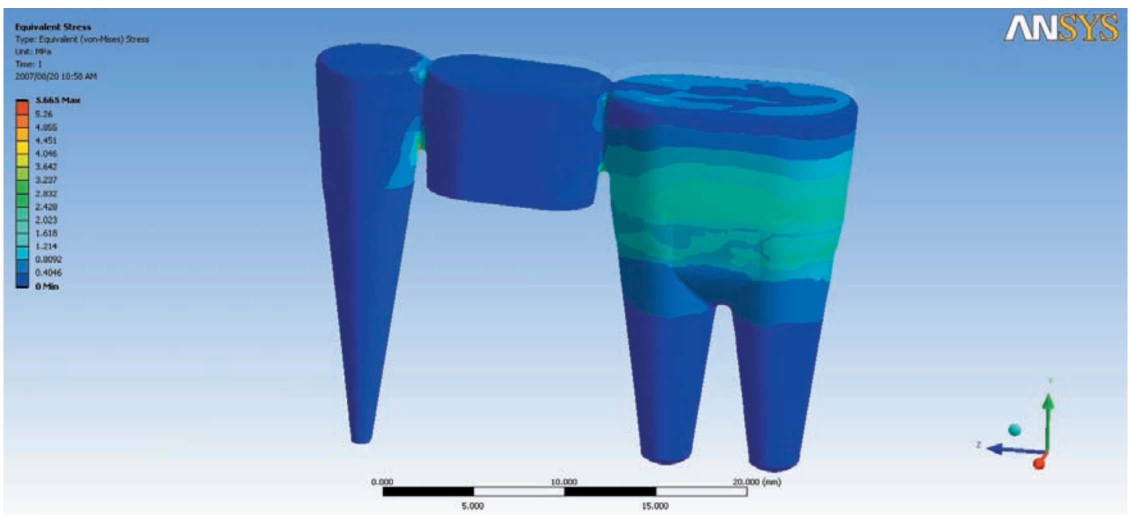


Figure 2. *Equivalent von Mises stress distribution within the completely-seated 3-unit FDP cemented with zinc phosphate under 70N vertical load applied to the molar only.*

Note: Since the actual deformation would not be visible to the observer, it is customary with FEA to display results in magnified form (to the order of x200 in this case). The shadow indicates the baseline position.

Table 2. *Maximum equivalent von Mises stress (MPa) for completely- and incompletely-seated fixed dental prostheses under different conditions of load directions and points of application and different cementing materials*

	Completely-seated				*Incompletely-seated*			
Load	*90° Vertical*		*45° Oblique*		*90° Vertical*		*45° Oblique*	
Cement	*ZnP*	*Res*	*ZnP*	*Res*	*ZnP*	*Res*	*ZnP*	*Res*
Premolar	8.2	8.1	52.0	54.0	46.1	46.9	78.4	80.8
Pontic	3.0	2.9	28.3	29.6	30.1	33.0	45.0	46.9
Molar	5.7	5.6	41.7	43.0	7.4	7.2	42.2	43.8

ZnP=Zinc phosphate cement; Res=Resin cement

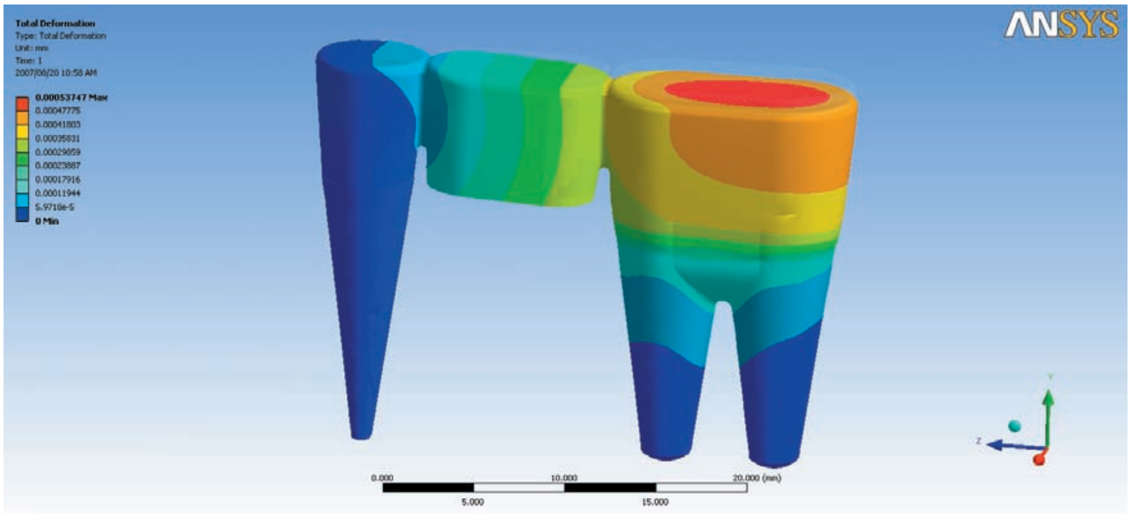


Figure 3. *Total deformation of the completely-seated 3-unit FDP cemented with zinc phosphate under 70N vertical load applied to the molar only.*

Note: Since the actual deformation would not be visible to the observer, it is customary with FEA to display results in magnified form (to the order of x200 in this case). The shadow indicates the baseline position.

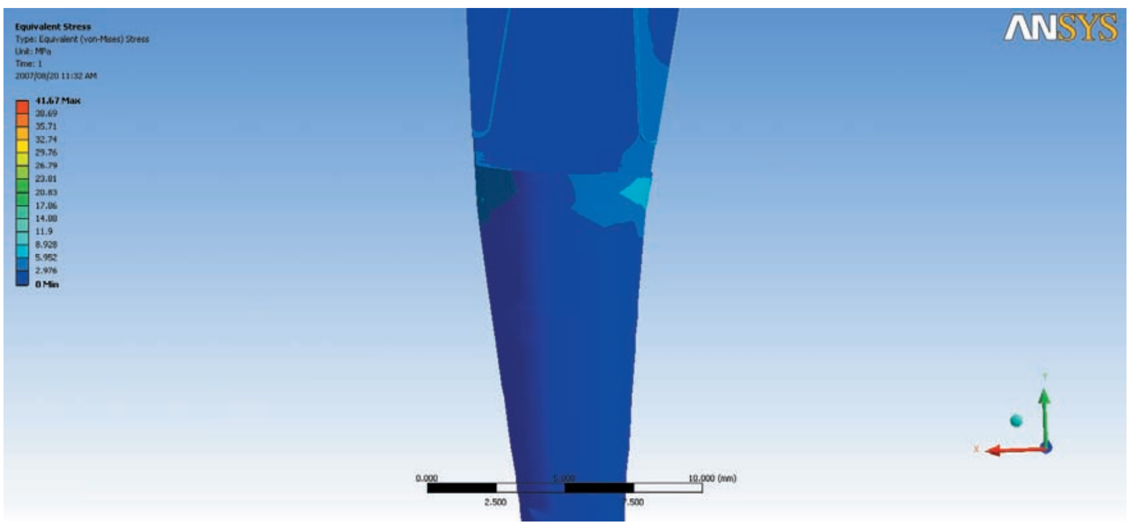


Figure 4. *Bucco-lingual section showing equivalent von Mises stress distribution within the completely-seated 3-unit FDP cemented with zinc phosphate under 70N oblique load applied to the molar only. There is opening of the margin on the loaded side and closing on the opposite side.*

Note: Since the actual deformation would not be visible to the observer, it is customary with FEA to display results in magnified form (to the order of x200 in this case). The shadow indicates the baseline position.

deformation was correspondingly large, with the cement allowing rotation of the FDP to produce displacements in the horizontal plane (the molar acting as a vertical axis for clockwise movement, Fig. 6a) and vertical plane (the opposite margins most likely acting as fulcra, Fig. 6b). As before, cement had only a small effect due to the difference in stiffness. When the pontic and molar of the incompletely-seated FDP (cemented with resin cement) were loaded obliquely, von Mises stress was 46.9 MPa, less than when the premolar was loaded, and again concentrating stress at the connectors as in the completely-seated scenario.

DISCUSSION

The general consensus from follow-up studies of FDPs placed in a variety of clinical settings is that secondary caries is the commonest cause of failure, followed by mechanical failure, in particular loss of retention[2-4]. Furthermore, secondary caries and loss of retention are frequently associated with the *same* clinical failure[2].

In this regard, the possible ways in which stresses may be induced in the cemented FDP system are pertinent. Of these, errors introduced during the fabrication process,

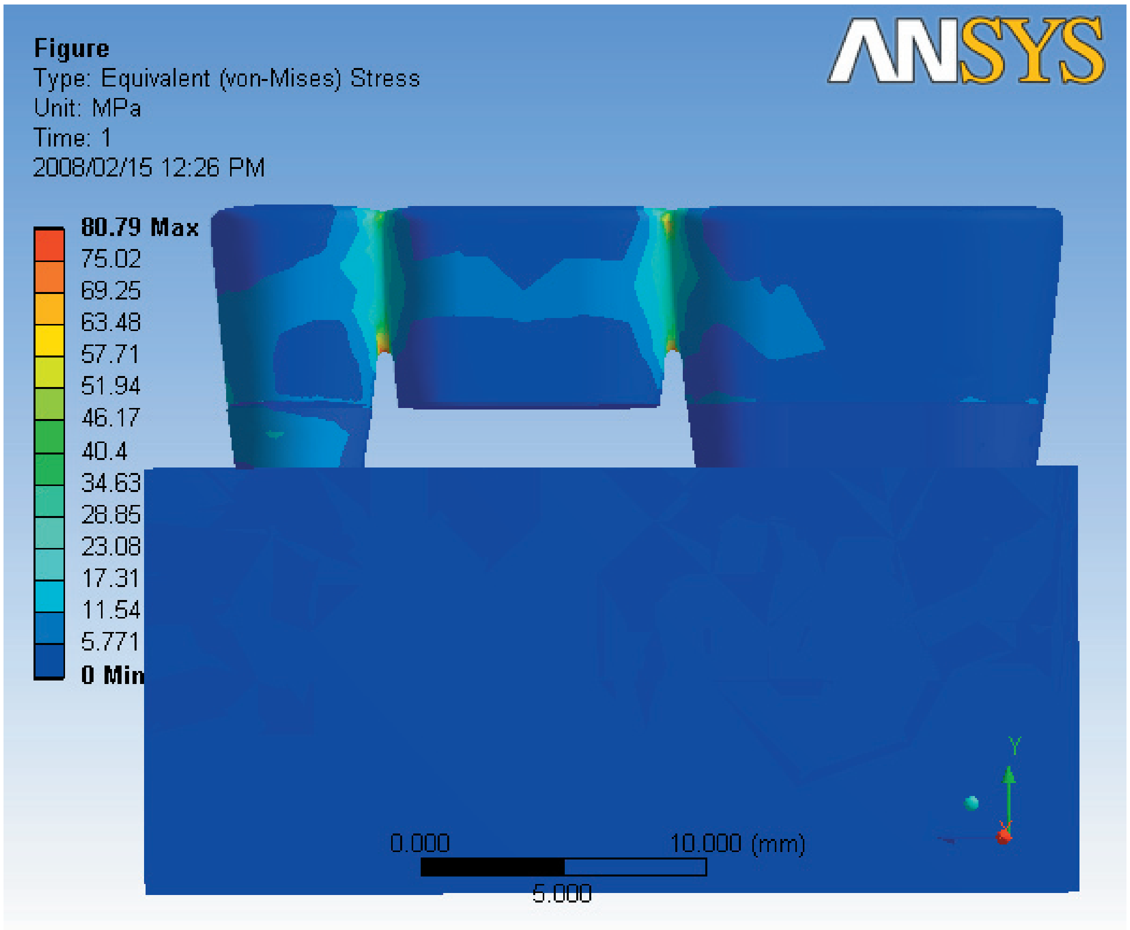


Figure 5. *Equivalent von Mises stress distribution within the incompletely-seated 3-unit FDP cemented with resin cement under 70N oblique loading applied to the premolar only.*

Note: Since the actual deformation would not be visible to the observer, it is customary with FEA to display results in magnified form (to the order of x200 in this case). The shadow indicates the baseline position.

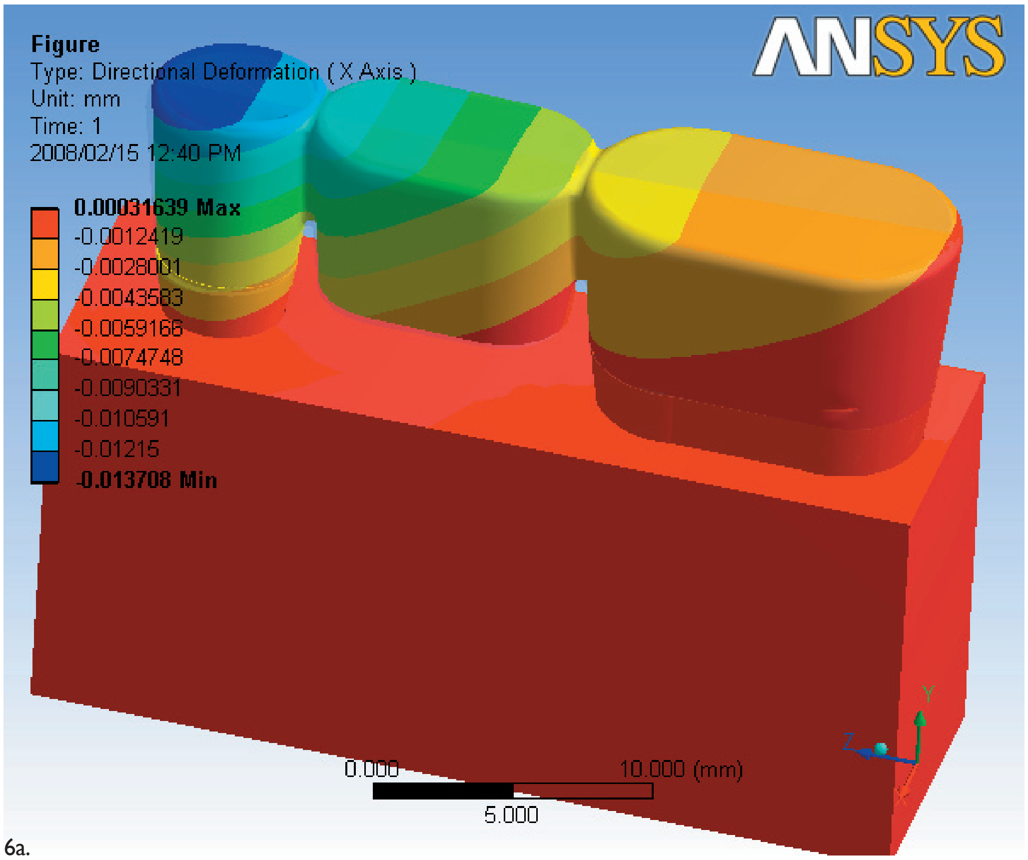
be they at the clinical or laboratory stages, are important. For example, structural design faults of FDP frameworks such as under-dimensioning of connectors can arise from space shortages[23]. The cyclic flexure of the structure that might then occur, even in normal function, can lead to fatigue failure in the luting cement layer[24]. Our finding that connectors showed maximum stress concentration even when their modelled dimensions easily exceeded minimum recommendations, confirms the importance of connectors in structural design.

Another way in which stresses might be induced is by cementing a FDP that was not verified as being completely seated prior to cementation. It was on the possible effects of this type of 'misfit' that our study focussed. Using the FEA method, the predictive potential for failure of poorly-fitting compared to well-fitting FDPs was assessed. FEA does so by indicating the physical response of the system to a given load[25]. The results reported here are for von Mises stress, which is a combination of stress components in all directions over all planes at a particular point, thus identifying the most likely place for material fracture, and thus failure, to occur.

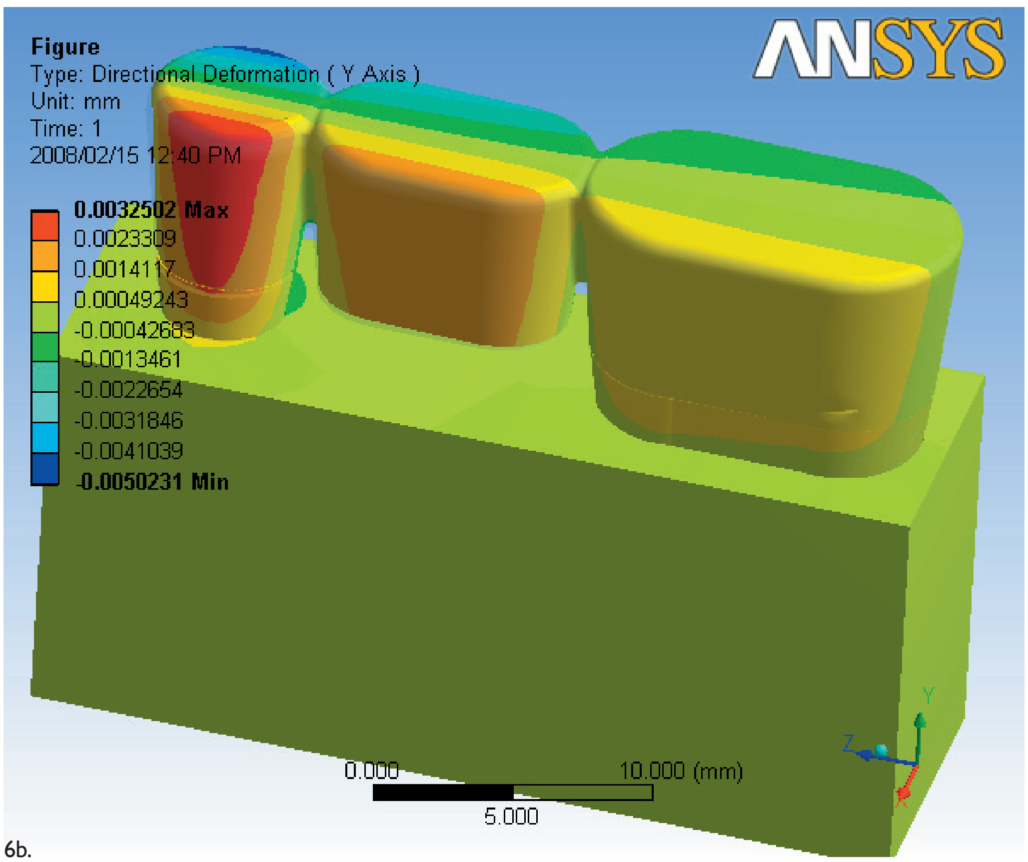
While relatively little is known about the biomechanical effects of incompletely-seated FDPs, implant 'misfit' has been quite extensively investigated. Results for implants have, however, been equivocal[15,17,26]. Thus, the clinical relevance of the presumed implications of implant 'misfit'[14] is unclear since clinicians do not know the effects of non-passive fit, nor the threshold of 'misfit' beyond which damage might occur[27]. With tooth-supported fixed prostheses still being far more widely prescribed than implant-supported ones[28], these questions would seem to be at least as important for conventional prostheses as they are for implant-supported ones.

Our overall findings confirmed the importance of good fit as shown by the generally low levels of maximum stress and deformation seen when the FDP was vertically loaded. Loading directly onto an abutment not surprisingly concentrated stress at that tooth, with any load transmission from this point to the rest of the system taking place through the connectors. When the pontic was loaded, flexure occurred, causing opening of the most distal and mesial margins of the FDP, and closure of those approximating the pontic area. In this regard, the importance of proper connector dimensioning for limiting such flexure has been discussed[23] .

The role of smaller cross-sectioned components, such as connectors, acting as mechanical isolators was clearly a strong influence in our model due to the omission of the periodontal ligament. This was considered both necessary and reasonable for the purpose of modelling a system that would permit less movement of the FDP system in order



6a.



6b.

Figure 6. *Directional deformation in the (a) horizontal (X) and (b) vertical (Y) planes of the incompletely-seated 3-unit FDP cemented with zinc phosphate under 70N oblique loading applied to the premolar only.*

Note: Since the actual deformation would not be visible to the observer, it is customary with FEA to display results in magnified form (to the order of x200 in this case). The shadow indicates the baseline position.

to observe the passive material response rather than one modified by the influence on the structure of a less constraining periodontal ligament. On the same basis, cancellous bone was omitted from the model, with the role of cortical bone being solely to constrain the movement of the roots and the outer limits of the structure. In addition, omission of the pulp as a void has been shown to have no effect on the magnitude of the coronal stress field[29]. In clinical terms, this set of boundary conditions is likely to represent a prosthesis that has been loaded relatively heavily causing the tooth to 'bottom out' in the alveolar housing[30]. Notwithstanding the fact that this represents a worst case scenario in terms of stresses and strains, the biological conditions that give rise to physiological tooth mobility[31], and their potential to permit *apparent* seating of an incompletely-seated FDP, are clearly relevant and needs further study.

Application of an oblique load in the completely-seated scenario increased stress and deformation. In this regard, certain conclusions from experimental data relating to implants may be applicable to conventional fixed prostheses, viz. the force vectors applied to teeth during function are multi-vectorial, ranging from fully vertical through horizontal, and with the transverse forces considered most detrimental because of the relative weakness of the components in tension and shear[32]. In a study investigating the effect of implant offset on induced stress in supporting bone, a 15° change in angle of force application produced a more than 3-fold increase in von Mises stress with a prosthesis of 6 mm in height[33]. Such a pattern was also observed in the present study with stress increasing up to 10-fold when the direction of load changed from 90° to 45°.

It has been shown that the mechanical properties of the luting cement play a vital role in the concept of resistance[12]. Even though the cement interface is never subjected to purely compressive stresses, the greater the compressive component, the greater is the abutment's capacity to stabilize the cemented restoration[15]. The portion of an abutment's surface that is subjected to a range of compression and shear stresses will vary with the prevailing functional (and parafunctional) force vectors, and when the physical properties of the cement become insufficient to withstand repeated loading, the cement lute fails. More specifically, when the integral of the compressive force vectors decreases below a given threshold level for the system in question, the cement lute becomes overstressed in terms of its resistance to repeated compressive forces. According to this concept, the threshold level would largely depend on the cement's resistance to fatigue in compression[12]. In this regard, a thin, uniform film of zinc phosphate luting cement has been shown to withstand fatigue fracture after repetitive loading[34]. These reports notwithstanding, however, the basic difficulty of predicting long-term clinical performance from short-term laboratory tests remains[35].

Incompletely-seated FDPs showed generally higher stress values than fully-seated FDPs, and much more so when obliquely loaded. The worst case scenario, as simulated by our model using single cycle loading, predicts the most likely areas of failure of the system, viz. the cement, in terms of ultimate strength. However, clinical situations involve low stress and significant cyclic loading, which means that cement failure would best be predicted in terms of fatigue. The ability of a material to withstand an infinite (or very large) number of cycles without failing is defined by its endurance limit. Although many factors affect the endurance limit, for many materials it is at about 40% of its ultimate strength[36]. Ultimate tensile strengths (UTS) of the cements used in this study were 9 MPa and 40 MPa (Table 1). This means that cyclic loads as low as 3.6 MPa could lead to fatigue fracture. In our study, only the vertically-loaded scenarios produced maximum equivalent von Mises stress values which were lower than, or of the order of, the estimated endurance limit. The endurance limit of resin cement (16 MPa) was more favourable in terms of resistance to fatigue failure. Under oblique load, and especially for the incompletely-seated case, stress values were much higher than the endurance limit, easily exceeding even the UTS of both cements. Although 70 N is not at the upper limit of occlusal loading, the worst case scenario simulated by our model would have accentuated the responses seen. In this regard, FEA, like any other basic research method, is used as an initial step and as an aid for planning further laboratory tests and clinical projects that will reduce the assumptions, and thus the inaccuracies, inherent with the FEA method.

Of the cements included in this study, zinc phosphate had the worst combination of values, viz. a high elastic modulus and relatively low UTS, which could lead to cement microfracture. Resin cement, with its medium elastic modulus and high UTS, could be a better choice for resisting microfracture in the clinical situation, and especially so when casting-to-abutment fit is less than optimal, as this study sought to simulate. However, clinicians will be aware of the numerous other factors that need to be considered when making clinical choices. One of these, in the case of resin cements, is its relative lack of long-term clinical data compared to zinc phosphate cement[34].

CONCLUSIONS

Within the scope of this finite element analysis, the following conclusions were drawn:

1. Vertical loading of the completely-seated FDP produced by far the lowest stress to any part of the restored system.
2. Stress increased markedly with an oblique load of 45°, and also varied according to the point of load application.
3. The incompletely-seated FDP exhibited highest maximum stress, and was greater the closer the point of load application was to the location of the 'misfit'; a 'cantilever-like effect' amplified the effect of the 'misfit'.
4. The combination of oblique load and 'misfit' suggested a potential for fatigue failure of the luting cement.

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MANUFACTURERS' DETAILS

- SolidWorks 2005, SolidWorks Corporation, Concord, MA, USA
- ANSYS 11.0, ANSYS Inc., Canonsburg, PA, USA

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REFERENCES

1. Shillingburg HT, Hobo S, Whitsett LD, Jacobi R, Brackett SE. *Fundamentals of fixed prosthodontics*, 3rd edn. Chicago: Quintessence, 1997; 392-393.
2. Lindquist E, Karlsson S. Success rate and failures for fixed partial dentures after 20 years of service: Part I. *Int J Prosthodont.,* 1998; **11**:133-138.
3. Holm C, Tidehag P, Tillberg A, Molin M. Longevity and quality of FPDs: a retrospective study of restorations 30, 20, and 10 years after insertion. *Int J Prosthodont.,* 2003; **16**:283-289.
4. De Bakker H, van Maele G, de Moor N, van den Berghe L, de Boever J. A 20-year retrospective survival study of fixed partial dentures. *Int J Prosthodont.,* 2006; **19**:143-153.
5. Randow K, Glantz P-O, Zöger B. Technical failures and some related clinical complications in extensive fixed prosthodontics. *Acta Odontol Scand.,* 1986; **44**:241-255.
6. Palmqvist S, Swartz B. Artificial crowns and fixed partial dentures 18 to 23 years after placement. *Int J Prosthodon.,* 1993; **6**:279-286.
7. Walton T. An up to 15-year longitudinal study of 515 metal-ceramic FPDs: Part 2. Modes of failure and influence of various clinical characteristics. *Int J Prosthodont.*, 2003; **16**:177-182.
8. Caputo AA, Standlee. *Biomechanics in clinical dentistry*. Chicago: Quintessence, 1987; 128.
9. Wiskott A, Nicholls JI, Belser UC. The relationship between abutment taper and resistance of cemented crown to dynamic loading. *Int J Prosthodont.,* 1996; **9**:117-130.
10. Wiskott A, Nicholls JI, Belser UC. The effect of tooth height and diameter on the resistance of complete crowns to fatigue loading. *Int J Prosthodont.*, 1997; **10**:207-215.
11. Trier AC, Parker MH, Cameron SM, Brousseau JS. Evaluation of resistance form of dislodged crowns and retainers. *J Prosthet Dent.*, 1998; **80**:405-409.
12. Wiskott A, Krebs C, Scherrer SS, Botsis J, Belser UC. Compressive and tensile zones in the cement interface of full crowns: A technical note on the concept of resistance. *J Prosthodont.,* 1999; **8**:80-91.
13. Milleding P. Abutment preparation. In Karlsson S, Nilner K, Dahl B (Eds): *A Textbook of Fixed Prosthodontics – The Scandinavian Approach*. Stockholm: Gothia, 2000; 170-172.
14. Weinberg L. The biomechanics of force distribution in implant-supported prostheses. *Int J Oral Maxillofac Implants,* 1993; **8**;19-31.
15. Duyck J, Naert I. Influence of prosthesis fit and the effect of a luting system on the prosthetic connection preload: An in vitro study. *Int J Prosthodont.*, 2002; **15**:389-396.
16. Natali AN, Pavan PG, Ruggero AL. Evaluation of stress induced in peri-implant bone tissue by misfit in multi-implant prosthesis. *Dent Mater.,* 2006; **22**:388-395.
17. Markarian RA, Ueda C, Sendyk CL, Lagana DC, Souza RM. Stress distribution after installation of fixed frameworks with marginal gaps over angled and parallel implants: A photoelastic analysis. *J Prosthodont.*, 2007; **16**:117-122.
18. Anusavice KJ, Hojjatie B. Stress distribution in metal-ceramic crowns with a facial porcelain margin. *J Dent Res.,* 1987; **66**:1493-1498.
19. Proos KA, Swain MV, Ironside J, Steven GP. Influence of cement on a restored crown of a first premolar using finite element analysis. *Int J Prosthodont.,* 2003; **16**:82-90.
20. Kamposiora P, Papavasiliou G, Bayne S, Felton DA. Predictions of cement microfracture under crowns using 3D-FEA. *J Prosthodont.,* 2000; **9**:201-209.
21. O'Brien WJ. *Dental materials and their selection,* 2nd edn. Chicago: Quintessence, 1997; 347-387.
22. Kohal R-J, Papavasiliou G, Kamposiora P, Tripodakis A, Strub J. Three-dimensional computerized stress analysis of commercially pure titanium and yttruium-partially stabilized zirconia implants. *Int J Prosthodont.*, 2002; **15**:189-194.
23. Omar R, Abduljabbar T, Al-Ali K, Smyth M, El-Agouri R. Dimensions of metal framework components of metal-ceramic fixed partial dentures constructed in a dental school setting. *Quintessence Int.,* 2004; **35**:820-825.
24. Junge T, Nicholls J, Phillips K, Libman W. Load fatigue of compromised teeth: a comparison of 3 luting cements. *Int J Prosthodont.*, 1998; **11**:558-564.
25. Proos K, Swain M, Ironside J, Steven G. Finite element analysis studies of an all-ceramic crown on a first premolar. *Int J Prosthodont.,* 2002; **15**:404-412.
26. Miller K, Faulkner G, Wolfaardt J. Misfit and functional loading of craniofacial implants. *Int J Prosthodont.*, 2004; **17**:267-273.
27. Taylor T. Research directions in implant prosthodontics. *Int J Prosthodont.*, 2000; **13**:270-271.
28. Carlsson GE, Omar R. Trends in prosthodontics. *Med Princ Pract.,* 2006; **15**:167-179.
29. Hojjatie B, Anusavice K. Three-dimensional finite element analysis of glass-ceramic dental crowns. *J Biomech.*, 1990; **23**:1157-1166.
30. Jacobsen PH, Wakefield AJ, O'Doherty DM, Rees JS. The effect of preparation height and taper on cement lute stress: A three-dimensional finite element analysis. *Eur J Prosthodont Rest Dent.,* 2006; **14**:151-157.
31. Picton DCA. On the part played by the socket in tooth support. *Arch Oral Biol.,* 1965; **10**:945-55.
32. Wiskott A, Pavone A, Scherrer S, Renevey R, Belser U. Resistance of ITI implant connectors to multivectorial fatigue loading application. *Int J Prosthodont.,* 2004; **17**:672-679.
33. Sutpideler M, Eckert S, Zobitz M, An K-N. Finite element analysis of effect of prosthesis height, angle of force application, and implant offset on supporting bone. *Int J Oral Maxillofac Implants*, 2004; **19**:819-825.
34. Yamashita J, Takakuda K, Shiozawa I, Nagasawa M, Miyairi H. Fatigue behaviour of the zinc phosphate cement layer. *Int J Prosthodont* 2000;**13**:321-326.
35. Mitchell CA, Orr JF. Engineering properties and performance of dental crowns. *Proc Inst Mech Eng.,* 2005; **219**:245-255.
36. Chalmers B. *Physical Metallurgy*. New York: Wiley, 1962; 207.