Viscoelastic Finite Element Stress Analysis of the Thermal Compatibility of Dental Bilayer Ceramic Systems

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Purpose: The aim of this study was to test the hypothesis that viscoelastic finite element analyses can reliably predict the effect of geometry on maximum tensile stresses in bilayer screening tests that are used to determine thermal compatibility. ***Materials and Methods:*** Three-dimensional viscoelastic finite element models of a beam, cylinder, disk, sphere, central incisor crown, molar crown, and posterior three-unit fixed partial denture (FPD) were used to calculate residual stresses after simulated bench cooling. Four compatible and four incompatible systems were evaluated. ***Results:*** The highest residual tensile stresses for all material combinations were associated with the three-unit FPD. Residual tensile stresses ranged from 5.4 MPa in the disk for a compatible combination to 262 MPa in the three-unit FPD for an incompatible system. Residual tensile stresses in the three-unit FPD ranged from 16.8 MPa to 44.0 MPa for the compatible systems and from 175 MPa to 262 MPa for the incompatible systems. ***Conclusion:*** Based on finite element calculations, it is predicted that all-ceramic dental prostheses with an average thermal contraction mismatch (500°C to 25°C) greater than ± 1.0 ppm/K will likely exhibit a relatively high percentage of failures in clinical use compared with systems having smaller thermal contraction mismatch between core and veneering ceramics. *Int J Prosthodont 2009;22:56–61.*

Ceramic dental restorations are currently a popular choice because of their superior esthetics and biocompatibility compared with metal-ceramic systems. Ceramic crowns have success rates that are nearly comparable to those of metal-ceramic systems, but ceramic fixed partial dentures (FPDs) have not demonstrated similar success, especially in posterior regions.[1–4] While the causes of failure of dental restorations are multifaceted, an important contributor to instantaneous and/or delayed formation of cracks is the thermal contraction mismatch between core and veneering ceramics. Currently, there are no guidelines for manufacturers to follow concerning the maximum allowable thermal contraction mismatch between ceramics. A rule of thumb, however, has been that the coefficients of thermal expansion should match as closely as possible or the coefficient of the core material should be slightly higher than that of the veneering ceramic to ensure that compressive residual stresses are induced in the weaker veneering layer.[5] However, it is not necessarily clear what expansion or contraction coefficients should be used to determine thermal mismatch. Should average coefficients be based on dilatometry data measured between two temperatures on the heating curve or the cooling curve?

ISO Standard 6872[6] for dental ceramics recommends that expansion be measured at 5°C/min to 10°C/min for two specimens fired twice and two specimens fired four times between 25°C and 500°C (or T_g, whichever is lower). Manufacturers should report the average value and standard deviation for expansion

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coefficients measured between 25°C and 500°C (or T_g). However, there is some evidence that manufacturers do not necessarily follow these guidelines. For example, the technical data sheets for IPS Empress 2 (Ivoclar Vivadent) core and veneering ceramics report average coefficients of expansion calculated between 100°C and 400°C while values for IPS e.max press (Ivoclar Vivadent) are given between 100°C and 400°C as well as between 100°C and 500°C. The expansion coefficient range for Vita VM 13 (Vident) ceramic reported by the manufacturer is 13.1 to 13.6 ppm/K without a specific temperature range. Researchers have also not been consistent in providing complete information on thermal expansion/contraction data when reporting on thermal compatibility studies or clinical studies involving the fracture of dental restorations caused by a large thermal contraction mismatch.

Simple geometries that have been used to determine the effect of thermal contraction mismatch on residual stresses[7,8] and deflections[9] in ceramic-ceramic systems include beams and disks. In addition, viscoelastic finite element analyses have been used to calculate transient and residual stresses in simple screening geometries[10,11] and in three-unit FPDs.[12] Although a direct comparison is not possible, residual stress levels in ceramic-ceramic systems that are reported to be thermally incompatible were generally much larger than those systems that are thought to be compatible.

To date, the authors are not aware of any studies that evaluate the validity of the various screening tests used to determine thermal compatibility of ceramic-ceramic systems. Therefore, the objective of this study was to use viscoelastic finite element analyses to determine transient and residual stresses in seven geometries composed of two core ceramics in combination with four veneering ceramics. Four of these combinations are considered to be thermally compatible and four are considered to be thermally incompatible.

Materials and Methods

The seven screening geometries include a beam, cylinder, disk, sphere, central incisor crown, molar crown, and a three-unit FPD. Dimensions of the core ceramic beam were 1.1 mm (height) × 4.0 mm (width) × 25 mm (length) with a 0.6-mm-thick layer of veneering ceramic. This geometry is the same as that used by Taskonak et al[7] to determine residual stresses in bilayer beams. The 16-mm-diameter core ceramic disks were 0.8 mm thick with a 1.0-mm veneer layer. The hollow-core cylinders (9 mm high and 11 mm inner diameter) had a wall thickness of 0.8 mm and were coated with a 1.0-mm veneer layer. The solid-core spheres (6.3 mm diameter) were coated with a 1.8-mm-thick veneering ceramic layer. The simulated central incisor crown had a 0.4-mm thick core framework with an incisal edge thickness of approximately 1.5 mm. The molar crown was modeled in a symmetric geometry with an average core thickness of 0.8 mm and an occlusal veneer thickness of 1.0 mm. The three-unit FPD consisting of a premolar abutment, a first molar pontic, and a second molar abutment, was based on a model developed by Goetzen et al.[13] The core framework had a thickness of approximately 0.4 mm and a variable veneer thickness of 0.5 mm to 1.0 mm in the occlusal area.

The material combinations selected for this study were the same as those reported by DeHoff et al[11] to represent thermally compatible and thermally incompatible ceramic systems. The core materials selected included IPS Empress 2 (E2C) and an experimental core material (EXC) (Ivoclar Vivadent), both of which were lithia-disilicate-based glass-ceramics. Four different commercial dental veneering ceramics were used: IPS Empress 2 (E2V) and IPS Eris (ERV) (Ivoclar Vivadent), Vita VMK68 body ceramic (VV) (Vident), and Finesse dentin (FV) (DENTSPLY Ceramco). Four compatible core-veneer groups (CG) and four incompatible groups (IG) were selected for the study. These combinations are given in Table 1. The thermal, elastic, viscoelastic, and contraction properties for these materials are given in a previous study.[11]

Calculation of transient and residual stresses for each geometry required an initial heat transfer analysis, followed by a viscoelastic structural analysis. All analyses were conducted using the ANSYS Finite Element program (ANSYS). The heat transfer analyses were completed using the thermal elements Solid77 for the cylinder, disk, and sphere, and Solid90 for the beam, crowns, and three-unit FPD. The viscoelastic structural analyses employed the axisymmetric element, Visco88, for the cylinder, disk, and sphere and the three-dimensional element, Visco89, for the beam, crowns, and three-unit FPD. Finite element models used for the cylinder and sphere were the same as presented by DeHoff et al.[11] The finite element model for the bilayer beam comprised 800 elements and 4,565 nodes for the core while the veneer layer was composed of 600 elements and 3,601 nodes. The finite element model for the disk is a two-dimensional mesh that is rotated 360 degrees to form the disk. The finite element models of the beam and disk are fairly simple and are not shown. However, the models for the simulated central incisor crown, the symmetric section of a molar half-crown, and the three-unit FPD are shown in Figs 1a to 1c. The core framework of the anterior crown was modeled with 2,637 elements and 12,488 nodes while the veneering layer was composed of 12,130 elements and 31,927 nodes. The core framework of the molar half-crown was modeled with

Table 1 Core Ceramic-Veneering Ceramic Combinations

Core ceramic matierals	Commercial veneering ceramics			
	Empress 2 (E2V)	Eris (ERV)	Vita VMK68 (VV)	Finesse (FV)
Empress 2 (E2C)	CG	CG	IG	IG
Experimental material (EXC)	CG	CG	IG	IG

CG = Compatible core-veneer goup; IG = Incompatible core-veneer group.

1,392 three-dimensional elements and 4,873 nodes while the veneer layer was modeled by 1,106 elements and 3,957 nodes. The core framework of the three-unit FPD was modeled with 25,017 tetrahedral elements and 41,524 nodes while the veneer layer was composed of 27,443 elements and 49,679 nodes. In each case, the number of elements was selected to provide stable stress results. The number of nodes depends on the elements selected for each geometry. Transient temperatures were determined for each screening geometry by assigning a constant convective coefficient on all exposed surfaces that resulted in an initial cooling rate of approximately 680°C/min. The initial temperature in each case was 700°C. This was also the reference temperature for the viscoelastic analyses.

Results

Transient stresses were determined at each time step used in the thermal analysis, and residual stresses were calculated at room temperature. Because ceramics are brittle materials and are most likely to fail when the maximum tensile stress reaches the tensile strength, the maximum principal stress (S1) that occurs anywhere in the model is presented. Moreover, although a dental restoration can fail because of transient stresses during cooling from the firing temperature, the level of residual stresses are of greater concern. A restoration might survive the cooling process only to have failure occur later because of high residual stresses, either as delayed fracture before placement or because of superimposed occlusal loading in service.

Shown in Figs 2a to 2e are residual principal stress maps for the beam, disk, central incisor crown, molar crown, and three-unit FPD, respectively. Stress plots for the cylinder and sphere can be found in a previous study.[11] Only the plots for the compatible system (EXC/ERV) are included in this article. The maximum residual principal stresses for all geometries and ceramic combinations are represented as a bar graph in Fig 3. In general, the maximum residual tensile stresses for the thermally compatible combinations (E2C/E2V, E2C/ERV, EXC/E2V, and EXC/ERV) are smaller than those of the thermally incompatible combinations (E2C/VV, E2C/FV, EXC/VV, and EXC/FV) for all geometries. Also, it can be noted that residual stresses for the crowns and three-unit FPD are generally higher than those for the simple geometries for each ceramic combination. An alternative presentation of the stress results is shown in Fig 4. Only the results for the cylinder, disk, molar half-crown, and three-unit FPD are presented. Of special importance is the sharp increase in stresses at thermal contraction mismatch values less than –1.18 ppm/K. An interesting comparison is shown in Fig 5, which presents the probability of failure of crowns[14] along with the failure pattern of spheres,[11] both plotted against average thermal contraction coefficient difference. Although the mismatch values were measured for different ceramic combinations and by different cooling rates in a dilatometer, the failure trends are remarkably similar.

Discussion

The authors are aware of only two studies that attempt to establish a maximum mismatch value between core and veneering ceramics that would predict probable failure of ceramic dental restorations as a result of thermally induced residual stresses. Steiner et al[14] fired nine commercially available body porcelains on an IPS Empress central incisor core and determined that no crowns failed for absolute values of thermal contraction mismatch less than 0.58 ppm/K. The average thermal contraction values were determined on the cooling curve between 500°C and 25°C with a cooling rate controlled at 5°C/min. DeHoff et al[11] used a commercial core ceramic and an experimental core ceramic to fabricate open-ended cylinders and core ceramic spheres. The core cylinders and spheres were veneered with one of four commercial dental ceramics representing four thermally compatible groups and four thermally incompatible groups. The thermally compatible groups had average alpha mismatch values between –0.39 ppm/K and +1.02 ppm/K (positive values indicate alpha of the core greater than that of the veneering ceramic). The thermally incompatible groups had average alpha mismatch values between -1.26 ppm/K and -3.01 ppm/K. Average alpha values were calculated between 500°C and 25°C based on cooling curve data measured in a dilatometer after shutting off the furnace power. Of the thermally incompatible cylinders, 100% failed. None of the thermally compatible combinations failed. Among the spheres, 100% of the thermally incompatible systems failed, 16% of one of the thermally compatible systems

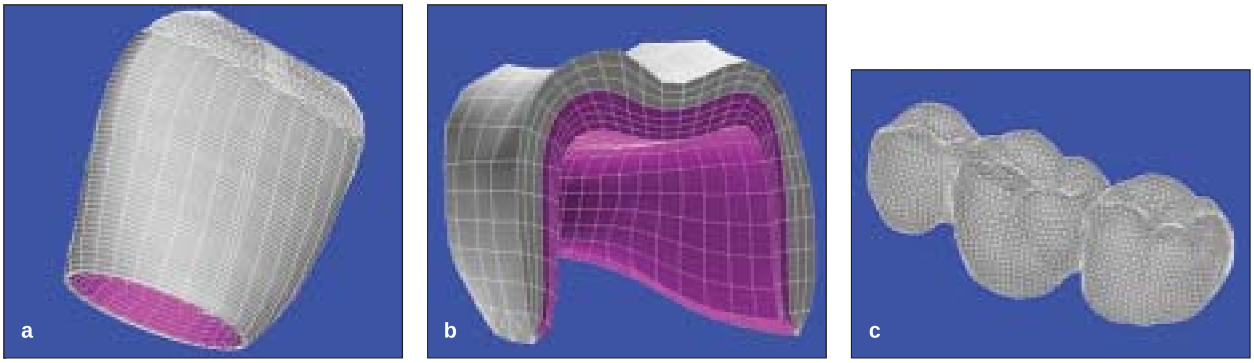

Fig 1 Finite element model of **(a)** a simulated central incisor crown, **(b)** the symmetric section of a molar crown with a core thickness of 0.8 mm, and **(c)** the three-unit FPD.

failed, and none of the remaining compatible combinations failed. While no thermal contraction mismatch limit was specified, it is worth noting that none of the specimens failed for thermal contraction mismatch values of -0.39 ppm/K or -0.61 ppm/K, which is in agreement with the results of Steiner et al.[14]

The results of this study suggest that simple screening tests are not likely to identify ceramic combinations that are ensured of success in clinical situations because of the many variables that affect long-term survival. However, there is sufficient evidence to suggest that placing a limit on thermal contraction coefficient differences is appropriate and manufacturers and dental lab technicians should be aware of these limits. Note that the thermal contraction mismatch limit on the negative side ($\alpha_c < \alpha_v$) seems to be critically important. For example, 100% of the cylinders and spheres[11] with a mismatch value of –1.26 ppm/K failed, but none failed for the combination with a mismatch value of –0.61 ppm/K. The trends for positive mismatch are less clear in that no cylinders failed for positive mismatch values of +0.8 ppm/K and +1.02 ppm/K, and only 16% of spheres failed at +0.8 ppm/K and 0% at +1.02 ppm/K. For simulated crowns,[14] 0% failed at +0.3 ppm/K, 12.5% failed at +1.05 ppm/K, and 37.5% failed at +1.2 ppm/K. Finite element calculations show a sharp increase in residual tensile stresses for mismatch values below -1.0 ppm/K and relatively constant values for the positive mismatches (Fig 4). Thus the finite element results tend to agree with the experimental evidence in that high residual tensile stresses are associated with the combinations that failed, and lower values are associated with those that did not fail or had low failure rates.

It is important to emphasize that many factors are involved in the eventual performance of any ceramic-ceramic combination in clinical situations. The finite element analyses reported here were limited to specific material combinations and particular geometries. Also, the contraction coefficients were based on cooling data measured in a dilatometer, whereas manufacturers generally report such data based on the heating curve. The average thermal contraction coefficients are based on the temperature range from 500°C to 25°C on the cooling curve, while some manufacturers report values from 100°C to 400°C on the heating curve. The ISO standard 6872 suggests that average thermal expansion coefficients should be reported between 25°C to 500°C (or T_g, whichever is lower). The ISO standard also specifies that standard deviations should be reported, but no limits are specified.

As an indication of the difficulties involved in specifying mismatches, the data for IPS Eris veneering ceramic yield average contraction coefficients of 10.79 ppm/K (500°C to 25°C) and 9.41 ppm/K (100°C to 400°C) while the manufacturer reports 9.75 ppm/K (100°C to 400°C). For the Empress 2 core material the data yield 10.18 ppm/K (500°C to 25°C) and 10.46 ppm/K (100°C to 400°C), while the manufacturer reports 10.60 ppm/K (100°C to 400°C). Thus the thermal contraction mismatch that is reported for E2C/ERV is –0.61 ppm/K, whereas if the mismatch were based on the manufacturer's published values it would be +0.85 ppm/K. In order for studies involving thermal contraction mismatch to be useful, there must be some agreement in the dental research community to standardize the measurement and reporting of thermal expansion/contraction coefficient data. It is recommended that data from the cooling curve measured in the dilatometer with the power off be used and that the average coefficients be reported from 500°C to 25°C.

Another observation of the results from the finite element analyses is that the maximum residual tensile stresses for the three-unit FPD are generally higher than those for the simple screening geometries for all combinations. This is not surprising since the three-unit FPD is a complex geometry with many areas that could be classified as stress raisers. In clinical situations it is most likely that FPD failures will occur in the connector

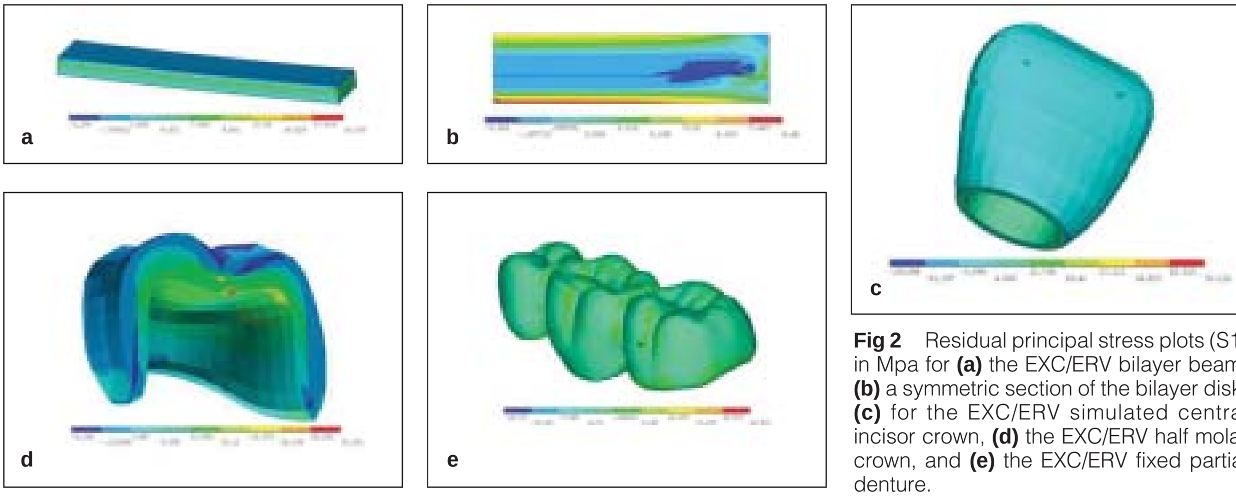

Fig 2 Residual principal stress plots (S1) in Mpa for **(a)** the EXC/ERV bilayer beam, **(b)** a symmetric section of the bilayer disk, **(c)** for the EXC/ERV simulated central incisor crown, **(d)** the EXC/ERV half molar crown, and **(e)** the EXC/ERV fixed partial denture.

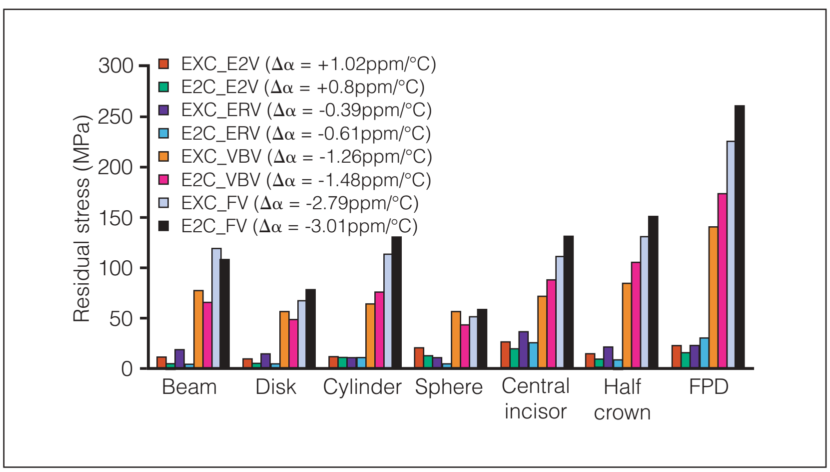

Fig 3 Maximum principal residual stresses (S1) in each geometry for each ceramic combination.

areas, primarily in the weaker veneer. For the particular FPD geometry used for the finite element analyses, the residual tensile stresses were generally higher in the veneer than in the core framework. High stresses did occur in the connector areas of the veneer, but they also existed in other areas as well. Generally, the residual tensile stresses in the veneer were higher at the core/veneer interface compared to those at the surface. This would support the study by Thompson et al[15] in which failures of all-ceramic FPDs originated in the veneer at the core/veneer interface in the connector area. However, a later study[16] did not support this finding.

Based on this study, it is recommended that all-ceramic dental systems involving a veneering ceramic should not be used for posterior FPDs. Although the results of this study apply only to the specific material combinations included in the study, unpublished finite element analyses of other combinations by our group suggest that thermal contraction mismatch is a dominating factor in the development of residual stresses of dental restorations during bench cooling. Thus, a monolithic FPD, which would not have thermally induced residual stresses, should perform better in posterior applications than veneered FPDs. However, this would likely lead to greater wear of opposing enamel surfaces. While the thermally induced residual stresses in the veneer may or may not contribute to veneer chipping, the mere presence of a lower strength ceramic in a potentially high stress area is not recommended. There are several published studies that provide credence for this recommendation. Taskonak et al[16] used fractography and fracture mechanics to estimate failure stresses for nonveneered and veneered ceramic FPDs. Calculated failure stresses for the all-core FPDs ranged from 107 to 161 MPa whereas for the veneered FPDs the range was 19 to 68 MPa. Goetzen et al[13] cited unpublished laboratory data to indicate that the fracture strengths of monolithic Cercon Zirconia FPDs were higher than those of veneered specimens. Guazzato et al[17] tested DC-Zirkon monolithic core disks and bilayer disks of DC-Zirkon veneered with Vita D ceramic and determined that there was no

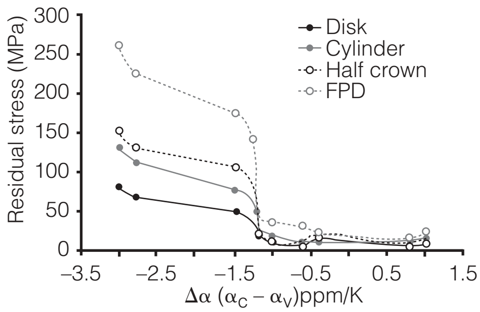

Fig 4 Maximum principal residual stress (S1) as a function of $\Delta\alpha$ for four geometries.

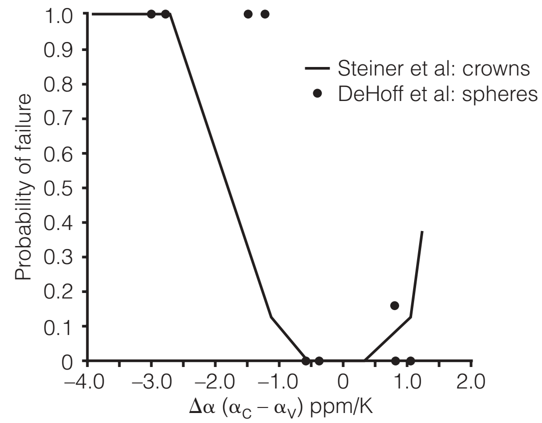

Fig 5 Probability of failure of crowns and spheres as a function of $\Delta\alpha$.

statistically significant difference in the mean flexural strengths of monolithic veneer disks compared with values for core/veneer disks when the veneer was placed on the tensile side. These studies suggest that the presence of a weaker veneering layer increases the probability of failure when an FPD is placed in a high stress area.

As previously indicated, it is also recommended that manufacturers report average thermal expansion/contraction coefficients based on cooling data from 500°C to 25°C. Based on average contraction coefficients determined in this manner, it appears that manufacturers should strive for thermal contraction mismatch values that fall within the range suggested by Steiner et al,[14] ie, $|\Delta\alpha| < 1$ ppm/K. Because of the many variables involved it is unlikely that any screening test can be devised that will identify with certainty whether a system will prove to be clinically successful. Simple geometries can be used to determine the effects of variables such as thickness ratios, cooling rates, loading schemes, and material properties, but in the end, only service history can validate the clinical efficacy of a dental restoration.

Acknowledgments

This study was supported by NIH-NIDCR Grant No. DE06672 and the Mechanical Engineering Department at UNC Charlotte.

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