Flexural strength of a layered zirconia and porcelain dental all-ceramic system

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Statement of problem. New processing techniques have facilitated the use of zirconia core materials in all-ceramic dental prostheses. Zirconia has many potential advantages compared to existing core materials; however, its performance when layered with porcelain has not been evaluated.

Purpose. This study investigated the strength of a wide variety of layered zirconia and porcelain beams to determine whether the inclusion of zirconia cores results in improved strength.

Material and methods. Eight types of layered or simple zirconia and porcelain beams (n = 10), approximately fixed partial denture–size, were made of a tetragonal polycrystalline zirconium dioxide partially stabilized with yttria core (Lava System Frame) and a feldspathic dental porcelain (Lava Ceram veneer ceramic). Elastic moduli of the materials were measured using an acoustic method. Maximum force and modulus of rupture were determined using 3-point flexural testing and a universal testing machine. Descriptive statistical methods were used.

Results. Beams with porcelain tensile surfaces recorded mean tensile strengths or moduli of rupture from 77 to 85 MPa, whereas beams with zirconia tensile surfaces recorded moduli of rupture almost an order of magnitude higher, 636 to 786 MPa. The elastic moduli of the porcelain and zirconia materials were 71 and 224 GPa, respectively. Crack propagation following initial tensile cracking often involved the porcelain-zirconia interface, as well as bulk porcelain and zirconia.

Conclusion. The layered zirconia-porcelain system tested recorded substantially higher moduli of rupture than have been previously reported for other layered all-ceramic systems. (J Prosthet Dent 2005;94:125-31.)

CLINICAL IMPLICATIONS

Because tensile failure is known to be the dominant clinical failure mechanism of all-ceramic restorations, it is expected that zirconia-based systems may reduce clinical failure or widen the range of clinical applications in comparison to weaker systems. The results of this study also highlight the critical importance of core thickness and of using unveneered core in areas of high tensile stress.

Predictable and esthetic nonmetal fixed partial dentures are desired by dentists and patients.[1] Fine-grained solid-sintered engineering ceramics are the strongest and toughest alternatives to metal materials.[2] Unlike many other solid-sintered ceramics, alumina and zirconia are relatively inexpensive and can be processed to achieve a somewhat translucent and tooth-colored appearance. McLean[3] first introduced high-purity solid-sintered alumina preforms for use in crowns and fixed partial dentures (FPDs). However, the processing of fine-grained solid-sintered ceramics to form customized FPDs or customized FPD frameworks is relatively difficult and expensive due to their high sintering temperatures, hardness, toughness, and pronounced sintering shrinkages.[2,4] Saddoun developed a process to create glass-infiltrated alumina cores, the In-Ceram system (European Patent 864,007,810, 1986). Anderson used CAD-CAM technology to enlarge refractory dies to compensate for sintering shrinkage with a proprietary high temperature and pressure firing to produce almost pure solid-sintered alumina copings, the Procera system.[4] Zirconia is more difficult to process

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than alumina and has only recently achieved widespread use in dentistry. Similar technologies have been applied to produce glass-infiltrated as well as solid-sintered zirconia cores and frameworks for FPDs (Lava All-Ceramic System; 3M ESPE, St Paul, Minn).[5,6] Solid-sintered zirconia has several potential advantages compared to alumina, including increased strength, increased toughness, decreased elastic modulus, and the remarkable property of transformation toughening.[7-9] Transformation toughening is the phenomenon whereby the normally tetragonal zirconia crystals undergo a lattice reorganization, when mechanically stressed, to change into a monoclinic form so that they effectively swell, thus tending to heal growing cracks and toughen the overall structure. Transformation toughening primarily affects long crack growth, or cracks of several millimeters in length, so dental prostheses may not benefit from this phenomenon.

Unfortunately, current processing technologies cannot make zirconia frameworks as translucent as natural teeth, nor can they provide internal shade characterization or facilitate customized shading. Therefore, zirconia cores or frameworks must be veneered with porcelain to achieve acceptable esthetics. A series of studies on layered all-ceramic structures showed that a veneer of relatively weak porcelain may result in failure at low loads should the porcelain veneer be placed in tension.[10-14] Although clinical failure of all-ceramic restorations is a very complex process involving patient variables, dynamic loads, restoration geometry, material properties, fatigue phenomena, and multiple failure modes, in vitro models may help to elucidate mechanical parameters known to influence fracture by tensile failure.[15-30] Tensile failure is believed to be the dominant clinical failure mechanism of all-ceramic restorations.[31-33] Previously reported studies on layered ceramic beams are replicated with zirconia core materials in this experiment.[10-11] The strengths of a wide variety of solid and layered zirconia and feldspathic porcelain beams were investigated with respect to their moduli of rupture (or tensile strength), loads to initial failure, failure modes, and the elastic moduli of their constituents.

MATERIAL AND METHODS

Zirconia-porcelain beams with 8 different configurations were fabricated as illustrated in Figure 1, using previously described techniques.[10,11] The zirconia core (tetragonal polycrystalline zirconium dioxide partially stabilized with yttria) (Lava System Frame; 3M ESPE) and the feldspathic dental porcelain (Lava Ceram Veneer Ceramic; 3M ESPE) had matching coefficients of thermal expansion. The beams were polished sequentially, with 800-grit silicon carbide (Saint-Gobain Advanced Ceramics, Boron Nitride Products, Amherst, NY) as the final abrasive.[9-11] Each test group contained 10 specimens with dimensions of approximately 44 mm in length, 4 mm in thickness, and 4 mm in width. This sample size had previously been found to be sufficient to discern trends in prior studies of 3 different types of layered dental ceramic and porcelain beams.[10,11] To ensure a high degree of uniformity, the thickness and width dimensions of individual beams varied by less than 0.05 mm, as measured with digital traveling micrometers with an accuracy of 0.0004 mm (Model 1337; Boeckler Instruments, Tuscon, Ariz) and a toolmaker's microscope (TM; Unitron, Bohemia, NY). Actual measurements, not group mean values, were used for all calculations. The test span was standardized to 40 mm, thus achieving a uniform span-to-depth ratio of 10:1, as well as simulating a long-span posterior FPD.[34] The 4×4-mm cross-section was chosen so as to bracket the sizes in the manufacturer's guidelines for FPD connector cores.

The elastic moduli of the solid veneering porcelain and solid zirconia core materials were measured because they are independently interesting and because they are needed to determine moduli of rupture. This was performed using a previously described nondestructive acoustic method.[35,36] The specimens were subjected to a 3-point flexural testing using a screw-driven universal testing machine (Model 1122; Instron Corp, Canton, Mass) at a crosshead speed of 0.25 mm per minute, and a chart speed of 50.8 mm per minute.[37-46] A 3-point load fixture (WTF-SB; Wyoming Test Fixtures, Laramie, Wy), containing linear bearings and rotatable 3.2-mm diameter cylindrical supports, was used to test the specimens. Factors including dimension, span, load rate, and surface finish, which are known to affect the mean and the distribution of strength values of dental porcelain, were standardized in this study.[37-46]

Failure forces were determined from the universal testing machine chart recorder. Formulae previously derived and published by the authors were used to calculate the tensile stresses at failure, or modulus of rupture (MOR).[10,11,47] Group mean values and standard errors were calculated for tensile strength, or MOR, and for maximum load.

A crude estimate for the lower limit of the mechanical porcelain-zirconia interfacial shear strength was calculated, using previously derived and published formulae, for situations in which the load necessary to cause interfacial delamination could be measured and differentiated from that which caused tensile failure (Group ZZZP, Fig. 2).[10,47] This lower limit for shear bond strength was determined based on the assumption that the tensile porcelain remained intact until the delamination load was attained. However, because this was not

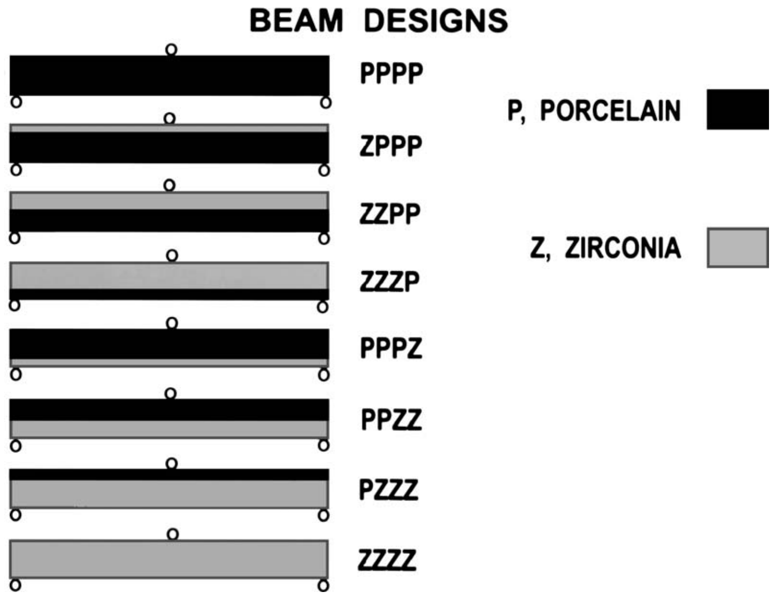

Fig. 1. Beam designs and test configuration. In this 3-point test, lower surface of each beam was placed in tension, whereas upper surface was placed in compression.

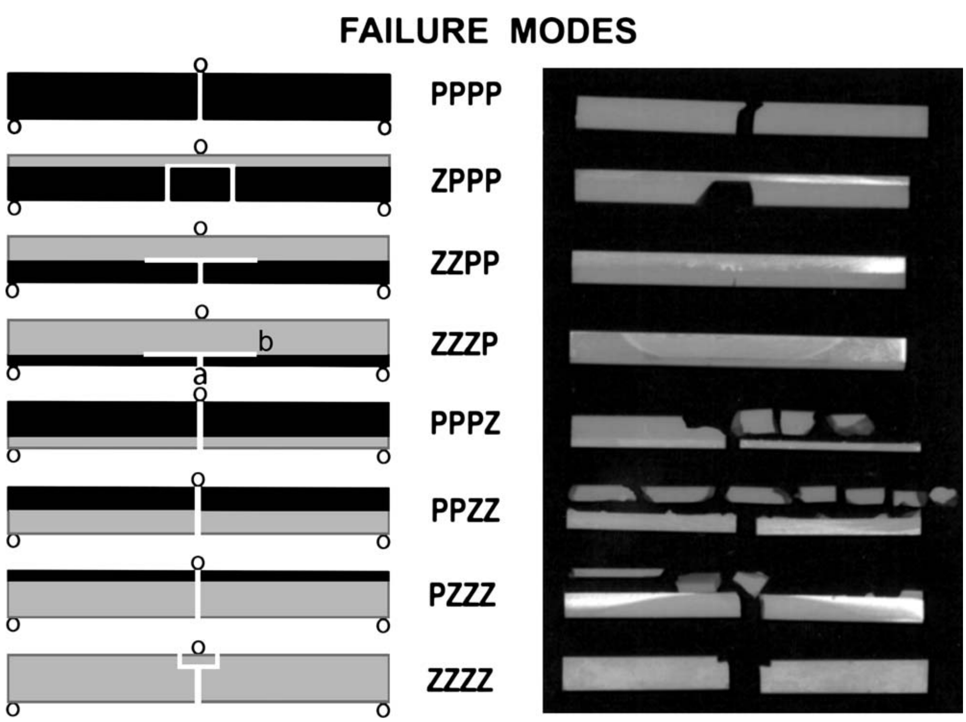

Fig. 2. Failure modes of solid and layered porcelain-zirconia beams. Schematic diagrams of initial failure modes *(left)* and representative corresponding specimens after testing *(right)*. In group ZZZP, 2 separate failure loads could be differentiated and measured. Initial failure *(a)* represented tensile porcelain failure (1.7 MN, Fig 4); second failure *(b)* represented subsequent porcelain-zirconia delamination (1.2 MN). *P,* Porcelain; *Z,* zirconia.

the case, the estimate must only be considered to be a lower limit and not an accurate estimate.[10,47] A mean and its associated standard error were calculated for this estimate of the lower limit of porcelain-zirconia interfacial shear strength.

RESULTS

All beams appeared to undergo tensile failure, with crack initiation from the central portion of the tensile undersurfaces of the beams, allowing appropriate application of formulae (Fig. 2). However, crack

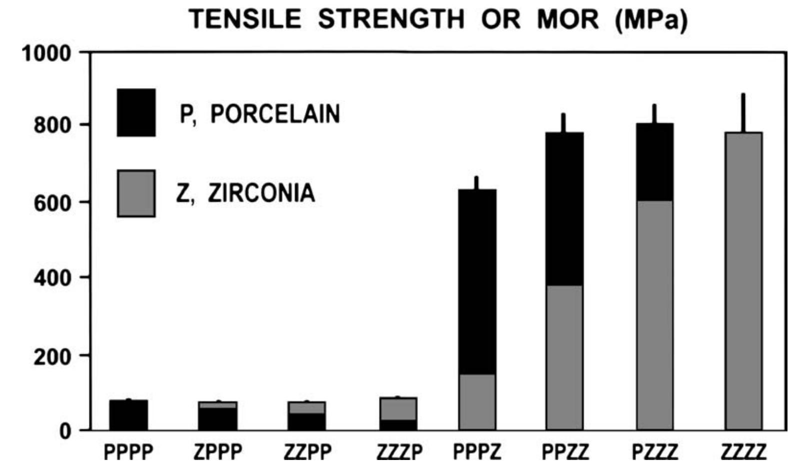

Fig. 3. Tensile strength, or MOR, mean values and standard errors of solid and layered porcelain-zirconia beams (MPa).

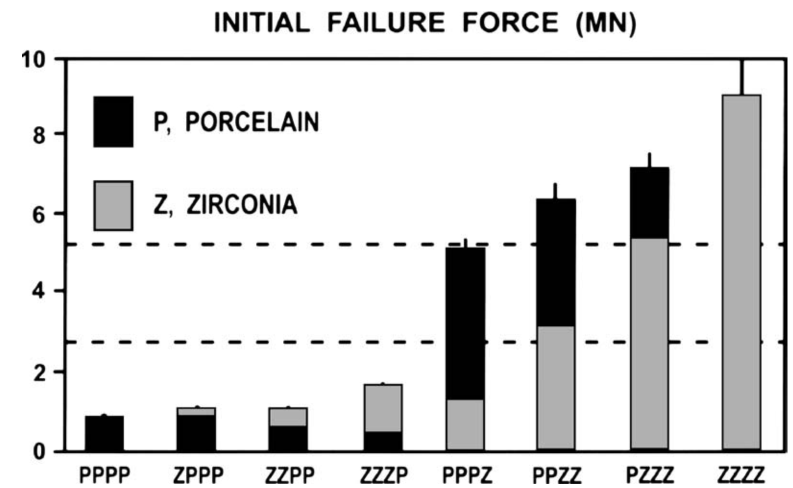

Fig. 4. Maximum force mean values and standard errors of solid and layered porcelain-zirconia beams (MN). *Dashed lines* indicate upper and lower ranges of reported values for maximum occlusal forces.[15-18]

progression sometimes differed among beam designs. Notably, in the layered beams that had porcelain tensile or undersurfaces (ZPPP, ZZPP, and ZZZP), cracks were deflected laterally when the stronger zirconia layers were reached. In the ZZZP beams, 2 distinct failure loads were noted by the chart recorder. The first failure load (Fig. 2, *a*) occurred upon initial tensile failure; the second (Fig. 2, *b*) occurred on lateral crack deflection and the initiation of delamination. Catastrophic failure following initial tensile crack progression also often involved the porcelain-zirconia interface as well as the bulk porcelain and zirconia (Fig. 2). The initial failure load was used in all calculations.

The material forming the tensile surface of the beams was of critical importance (Fig. 3). Beam designs with porcelain tensile surfaces recorded mean tensile strengths, or moduli of rupture, from 77 to 85 MPa, whereas beam designs with zirconia tensile surfaces recorded moduli of rupture almost an order of magnitude higher, 636 to 786 MPa. An increasing proportion of zirconia tended to increase the force-bearing capacity of beams of similar configuration (Fig. 4). The mean elastic moduli of the porcelain and zirconia materials (with their associated standard errors) were 70.7 (0.6) and 224.4 (0.8) GPa, respectively. The mean estimate for the lower limit of shear bond strength of the porcelain-zirconia interface was 4.6 MPa, with a standard error of 0.1 MPa (n = 10).

DISCUSSION

Because the beams with the strong core material on their tensile surfaces recorded up to elevenfold larger moduli of rupture and up to tenfold larger forces to failure than when the weak porcelain was placed on the tensile surfaces (Figs. 3 and 4), it is strongly recommended that the undersurfaces of FPD connectors and other areas of high tensile stress not be veneered with porcelain at all. Because of the clear trend that an increase in the relative thickness of the strong core material increased the load to failure (Fig. 4), whether the core was placed in tension or in compression, it is recommended that prostheses be designed with as thick a core and as thin a porcelain veneer as possible. Although FPDs may be primarily loaded in a vertical occlusal-to-gingival direction, their complex shapes and human masticatory habits may cause prostheses to be loaded in many different ways.[15] Therefore, the maximum amount of core material in all potential areas of high stress is recommended.

This layered zirconia-porcelain system recorded substantially higher moduli of rupture (786-794 MPa) than comparable solid-sintered alumina-porcelain (504-510 MPa) and glass-infiltrated alumina-porcelain (340-520 MPa) layered beams that had been tested in a similar configuration with their half- or full-thickness core layers placed in tension.[10,11] It is important to note that because MOR is a fundamental mechanical property, meaningful comparisons can be made of MOR values among different studies as long as comparable specimen preparation techniques and test parameters are used. Because tensile failure is believed to be the dominant clinical failure mechanism of all-ceramic restorations,[31-33] it is reasonable to expect that zirconia-based systems could reduce clinical failure or widen the range of clinical applications compared to alumina-based systems.

The MOR, or flexural tensile strength, and the elastic modulus of the simple solid-sintered zirconia beams in this study were comparable to prior data for other zirconia-based materials of similar formulation.[7-9] Although much work has been focused on zirconia ceramics during the last decade, most of this work has been performed in industrial, not academic, settings. Consequently, abundant data are available in nonrefereed commercial product technical data sheets, but much less has been reported in refereed scientific journals. Review of many commercial product data sheets suggests that the data in this study are consistent with the mid-range performance of similar partially yttria-stabilized zirconias. However, the manufacturer's technical product profile lists a value from a 3-point flexural strength test that is substantially higher than that achieved in this experiment—1272 versus 786 MPa. This and some other strength values listed by the manufacturer fall within the upper ranges of those reported in the refereed and nonrefereed literature. It is possible that such a difference could be the result of differing specimen and test configuration, surface preparation, and experimental conditions. However, the experimental parameters used in this study fall within widely established guidelines for the flexural testing of brittle materials.[37-46] The strength values recorded in this current study are not indicative of the transformation toughening phenomenon in these prosthesis-sized specimens, and it is unlikely that most dental prostheses would benefit from this phenomenon.

Elastic modulus is a measure of the rigidity or stiffness of a material, or the ratio of stress to strain. The mean value for the core zirconia's elastic modulus measured in this study (224 GPa) was in close agreement with the manufacturer's value (210 GPa). In contrast, solid-sintered aluminas are reported to have elastic moduli values in the range of 350 to 400 GPA, almost double that of the zirconias.[2,8] The relatively low elastic modulus of zirconia may have implications for the extremely complex stress distributions within occlusally loaded prostheses in vivo.[12,21]

The MOR, or flexural tensile strength, and the elastic modulus of the feldspathic porcelain recorded in this study were comparable with prior data for other dental porcelains.[10,11,46,47] However, the manufacturer's technical product profile lists a value from a 4-point flexural strength test that is slightly higher than that achieved in this 3-point flexural experiment—85 and 77 MPa, respectively. The mean value for the elastic modulus of the veneering porcelain in this test (71 GPa) was slightly lower than the manufacturer's value (80 GPa) but is consistent with previously reported data for feldspathic porcelains.[10,11,47]

In the layered beams that had porcelain tensile or undersurfaces (ZPPP, ZZPP, and ZZZP), cracks were deflected laterally when they reached the stronger zirconia layers. This can be interpreted in 2 ways. First, the crack deflection could be a consequence of the superior ability of zirconia to resist crack propagation. Secondly, the interlaminar crack deflection could also indicate a relatively poor zirconia-to-porcelain bond. The clinical implication of this finding is that this system could have a tendency to produce porcelain ‘‘pop-off’’ rather than catastrophic failure. Of course, any type of damage is unwelcome, but ‘‘pop-off’’ might be considered a lesser evil.

Had the porcelain and zirconia layers and their interfaces behaved in a theoretically ‘‘ideal’’ manner, it would have been expected that all beams with zirconia tensile or undersurfaces (PPPZ, PPZZ, PZZZ, and ZZZZ) would have displayed like failure modes. Clearly, this was not the case (Fig. 2). Therefore, it is probable that either residual stresses remained from zirconia sintering, porcelain firing, or finishing and polishing procedures, that the interfacial bond was relatively poor, or that both effects were present. A less than perfect bond is consistent with observed patterns of crack propagation (Fig. 2). When a solid-sintered alumina-porcelain system was previously tested in a similar manner, it was found to behave in a theoretically ‘‘ideal’’ manner.[11,13] However, when a glass-infiltrated alumina-porcelain system was previously tested in a similar manner, it was found not to behave in a theoretically ‘‘ideal’’ manner, and substantial porcelain debonding occurred during testing.[10] Initially, that system was thought to have a less than optimal core-to-porcelain bond due to accumulation of infiltration glass on the core surface, but subsequent investigations demonstrated that the apparent delamination often occurred just inside the porcelain layer close to the interface, not within the infiltration glass.[19,22] The exact mechanism of apparent interfacial bond failure in the current investigation of the solid-sintered zirconia-porcelain system is unknown.

Estimates for porcelain-ceramic interfacial bond strengths have rarely been attempted, so meaningful comparison of this data to prior work cannot be made.[10,23] The authors recognize the limitations of the crude estimate for the lower limit of the porcelain-zirconia bond. Standard tests for porcelain-ceramic bond strength measurements have yet to be established. As interfacial porcelain debonding was discerned in this study on zirconia and in prior studies on other all-ceramic systems, a method to quantify porcelain-ceramic bond strengths would be useful.[10,19,22]

The maximal forces resisted by the beams in this study should only be compared to similar tests of equivalent beam dimension and equivalent testing configuration. In contrast, the MOR data can be usefully compared to any other tests that followed the normal guidelines for 3-point flexural testing of brittle ceramics because the effects of specimen shape and test configuration have been computed. The beams used in this study had an extremely long span (40 mm) and bracketed the cross-sectional dimensions recommended by the manufacturer for large posterior prostheses.

Maximal occlusal forces measured in humans have been reported with ranges of 2.7 to 5.2 MN, depending on gender, age, measurement technique, location measured, head position, and state of the dentition, among other factors.[15-18] According to the results of this study, such maximal loads would be sufficient to cause porcelain failure, but not core failure, on the tensile surfaces of FPDs of similar dimensions to the extremely long beam designs tested (Fig. 4, groups ZPPP, ZZPP, and ZZZP). For this reason, it is again recommended that porcelain not be placed on the tensile undersurfaces of connectors or other areas of high tensile stress. Similarly, the results of this study suggest that an FPD with a very thin core might not withstand maximal occlusal forces, even when the core is placed in tension and the porcelain is protected in compression (Fig. 4, group PPPZ). For this reason, it is recommended that thin cores not be used, even when the tensile undersurface is composed of core material. However, it is important to note that these test beams had exceptionally long spans, 40 mm, and that routine masticatory loads are much smaller than maximal occlusal forces. Had shorter spans been used, the maximal forces the beams could resist would have been expected to rise in inverse proportion to the length of the span.[10-11,37-46] Hence, a beam with a 20-mm span of the same cross-sectional dimensions as used in this study, and with a zirconia tensile surface, would be expected to resist approximately 2 to 3 times the upper limit of occlusal forces measured in humans (Fig. 4).[15-18] A 20-mm span may represent a 3- to 4-unit FPD.[34]

The bilayer beam mechanical model used in this study has been validated by finite element analysis and correlated with failure behavior.[13,34] Although it can identify important trends and has relevance to more complex clinical situations, it does have some disadvantages.[12,20,21] It has a much simpler geometry than an FPD. It lacks thinner stress-concentrating connectors. It does not have outer layers of porcelain on both the compressive and tensile surfaces, as FPDs often do. It is not supported by flexible dentin, a flexible periodontal ligament, or flexible bone. However, the same mechanical principles do apply to crowns and FPDs. The model is also relevant because both all-ceramic crowns and FPDs are thought to fail most often by crack initiation during tensile loading.[31-33]

Predictive models, such as the finite element analysis bilayer beam models, are widely used to study well-defined systems with known parameters. However, this current study suggests that factors such as less than optimal interfaces and residual stresses should be included in theoretical models, increasing their complexity and necessitating the initial interrogation of such factors. Investigation of the effects of residual stresses, various interfacial bond strengths, and processing defects by relatively efficient theoretical methods could be most enlightening.

It is important to note that quasi-static mechanical strength tests, used in this study, are only a first step toward predicting clinical performance. Dental ceramics are susceptible to the effects of chemical fatigue, or stress-corrosion, as well as to the effects of cyclic mechanical fatigue.[24-26] However, comparative quasi-static mechanical testing does provide a basis for initial comparison, and stronger dental ceramic systems are known to have superior clinical performance compared to weaker systems.[27-30]

CONCLUSIONS

1. The mean elastic moduli (and associated standard errors) of the porcelain veneer and zirconia core materials were 70.7 (0.6) and 224.4 (0.8) GPa, respectively.
2. All types of solid or layered beams tested initially underwent tensile failure, with crack initiation from the central part of the tensile undersurfaces of the beams.
3. Crack progression differed among beam designs. Notably, in the layered beams that had porcelain tensile or undersurfaces, cracks were deflected laterally when the stronger zirconia layers were reached.

4. The material forming the tensile surface of the beams was of critical importance. Beam designs with porcelain tensile surfaces recorded mean tensile strengths or moduli of rupture from 77 to 85 MPa, whereas beam designs with zirconia tensile surfaces recorded moduli of rupture almost an order of magnitude higher, from 636 to 786 MPa.

5. An increasing thickness of zirconia increased the load bearing capacity of the beams.

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