Esthetic dental ceramics are commonly bilayered systems, meaning they have an inner layer of high-strength core ceramic and an outer layer of esthetic veneering ceramic. The load-bearing ability of a bilayered ceramic crown is widely assumed to improve with increased core ceramic thickness for a given total thickness; for example, for a ceramic crown with a total thickness of 1.5 mm, having a 1-mm-thick core layer results in a stronger crown than having a 0.5-mm-thick core.\(^1\)\(^-\)\(^5\) Unfortunately, maximizing the thickness of the stronger opaque core frequently reduces esthetics. By this reasoning, dental professionals must always attempt to balance esthetics and function when designing dental ceramic crowns.

Interestingly, recent theoretical work suggests that it is actually the total ceramic thickness and not the core thickness that is the significant determinant of load-bearing ability. Analytical modeling by Hsueh and Miranda\(^6\) suggested that failure loads are relatively core-thickness independent of radial cracking of a supported bilayered ceramic. This implies that if a constant total thickness of ceramic is maintained, the ratio of the esthetic porcelain instead of the stronger core porcelain can be increased without reducing the load-bearing ability of the ceramic crown. Calculated failure loads for veneered lithium disilicate specimens are shown in Figure 1, using the formula of Hsueh and Miranda.\(^6\)

Considering this interesting analytical solution, it was our intention to definitively explore the role of core thickness ratios for a constant total bilayer thickness under cyclic fatigue conditions in the presence of water. The project gathered mean load-to-failure data under cyclic fatigue conditions, using a standard staircase (up-and-down) sensitivity test over a set number of cycles. This allows testing of the hypothesis under appropriate conditions for clinical relevance.\(^7\)
Understanding the mechanisms by which crowns fail clinically is of primary importance so that we may replicate these conditions experimentally. It is well accepted that the clinical bulk fracture of ceramic crowns often initiates from the cementation surface. This conclusion is derived from finite element analysis (FEA) and fractographic analyses of clinically failed specimens. Failure is believed to begin when compressive forces on the occlusal surface result in tensile stresses (bending) at the ceramic-cement interface. The tensile stresses, which arise from differences in the elastic moduli of ceramic and substrate, result in bending at the ceramic-cement interface and eventually lead to crack initiation (Fig. 2).

In an in vitro protocol, the variables that influence the stress conditions present during contact loading of a dental ceramic should be understood so that the failure conditions can be replicated. Some of the most important variables include load, contact area, elastic modulus of the 2 materials in contact, bonding, exposing the ceramic to a high number of low loads, and presence of water. All these factors must be accurately replicated in vitro in order for test data to be relevant to clinicians.

Unfortunately, most in vitro studies involve monotonic load-to-failure testing. Monotonic failure loads do not predict cyclic behavior at low loads/high cycle values (as occur intraorally) because there are significant differences in damage accumulation. This is especially true when in the presence of water. For example, monotonic bending tests of zirconia typically find that airborne-particle abrasion increases bending strength presumably due to development of a protective compressive stress layer. However, under cyclic fatigue conditions, Zhang et al found that airborne-particle abrasion compromised the fatigue strength of both alumina and zirconia by between 20% and 30%, finding that “In fatigue, small flaws introduced by particle abrasion can outweigh any...strengthening effect from compression associated with surface damage or...phase transformation.” Furthermore, in vitro failure loads and contact pressures often far exceed the loads found in vivo. Contact pressures at wear facets during maximal clenching are approximately 40 MPa, whereas monotonic loads range from 1000 MPa to 5000 MPa due to loading with small-diameter spheres. Failure under cyclic loading may involve the same mechanism as monotonic loading but using mean loads that approximate clinically relevant numbers. Kelly et al reported that mean failure loads of dental ceramics dropped well within the range for clinical failure when specimens were cyclically loaded under aqueous conditions.

Both the contact area and the load determine the surface contact stresses and the stress distribution throughout the ceramic. In vivo, wear facets can be as large as 1 to 3 mm in diameter. Spherical indenters used in vitro must have diameters that replicate clinical contact surface areas to reproduce the proper stress state. Until recently, studies used either a small-diameter sphere to propagate a crack or examined surface damage under Hertzian sliding contact. Blunt loading with a small diameter sphere creates a “Hertzian” stress state. This stress state is characterized by the development of a ring-shaped crack located outside the contact area, which will eventually develop into a subsurface cone crack. Interestingly, fractographic analysis of clinically failed crowns has provided evidence that failures do not involve

**Clinical Implications**

Esthetics can be maximized in bilayer prostheses by keeping the structural layer as thin as possible without reducing durability.
surface damage at or below wear facets. In fact, Quinn et al. reported the only Hertzian crack ever observed from a clinical specimen but determined that it was the result of edge chipping following fracture. Kelly et al. determined that the appropriate stress state and crack system can be developed using pistons having an elastic modulus equal to or less than enamel with diameters in the size range of clinical wear facets.

As previously discussed, crack initiation during occlusal loading results from tensile stresses, scaling with the differences in the elastic moduli of the ceramic and substrate. Therefore, finding a substructure material that has an elastic modulus similar to that of hydrated dentin is important. Finite element analysis predicts that the elastic modulus of the substructure material will in is important. Finite element analysis predicts that the elastic modulus of the substructure material will influence failure loads. The substructure material found to have the closest elastic modulus to hydrated dentin in recent ceramic bulk failure experiments was a woven glass fiber-filled epoxy (NEMA grade G10; Accurate Plastics). In these experiments, blunt contact stress-strain slopes for dentin and the G10 material were indistinguishable, meaning that the elastic moduli were also identical. The resin cement bond strength to dentin was slightly lower than that of the G10 material. Given these findings, G10 was an appropriate choice for the substructure material in an attempt to replicate the differences in the elastic moduli between the ceramic and dentin. The effect of water is rarely evaluated but is very important in the propagation of cracks. Numerous studies have provided evidence for the strength-degrading effects of water during ceramic failure. This holds true for both monotonic and cyclic loading of dental ceramics. The phenomenon by which water participates in reducing fatigue strength is termed “chemically assisted crack growth” or “static fatigue.” It has been determined that water acts at crack tips to reduce the strength of glasses and ceramics. Water specifically acts to break the metal oxide bonds with subsequent production of hydroxides. This results in the slow propagation of a crack (slow crack growth) until it reaches the critical size for fracture. Not only does water have access to the external surface of dental crowns, it also has access to the internal surface. Dentine tubules and transport from dental cements provide an aqueous environment for the internal surface of a crown. The presence of water should be taken into account in a protocol design, with both the external and internal surfaces of the ceramic being exposed to water.

Our intention was to design and execute a protocol that appropriately replicated the variables important in the clinical bulk fracture of ceramic crowns. The purpose of this study was to address the hypothesis that the thickness of the core ceramic has minimal influence on the strength of a bilayered ceramic at a constant total thickness. If the results of the experiments support the hypothesis that the thickness of the high-strength core ceramic has minimal influence on the fatigue lifetime of a bilayered ceramic, it will have ramifications for all dental professionals. If we could determine that core ceramic thickness does not influence fracture behavior, dental clinicians could fabricate ceramic crowns using the minimal core thickness possible. This would enable dentists to optimize esthetics without compromising function.

**MATERIAL AND METHODS**

Slightly oversized specimens of lithium disilicate were sectioned from presintered computer-aided design/computer-aided manufacture (CAD/CAM) blocks (Ivoclar Vivadent AG) to create cores with thicknesses of 0.5, 0.75, 1.0, and 1.5 mm. These cores were then sent to the manufacturer (Ivoclar Vivadent Inc) to be cerammed and then to a ceramic machinist (BOMAS Machine Specialties) to ensure accurate core thicknesses. The lithium disilicate cores were veneered with IPS e.max Ceram (Ivoclar Vivadent Inc) to the core/veneer thicknesses of 0.5/1.0 mm (n=29), 0.75/0.75 mm (n=42), 1.0/0.5 mm (n=40), and 1.5/0.0 mm (n=48). The veneered specimens were sent to the same ceramic machinist to ensure total thickness was 1.5 mm for all specimens (Fig. 3).

Bases were fabricated from an epoxy-glass fiber material (NEMA grade G10; Accurate Plastics), which presents elastic behavior similar to that of hydrated dentin. Channels were cut into the NEMA G10 bases to allow deionized water to hydrate the cement during storage and testing, following a recently published protocol. All specimens were cemented (Multilink Automix; Ivoclar Vivadent Inc) to the bases. Ceramic specimens were etched with 5% hydrofluoric acid for 30 seconds, rinsed, and then silanated (Monobond Plus; Ivoclar Vivadent Inc). A primer (Multilink; Ivoclar Vivadent Inc) was applied to the bases for 30 seconds and then air-dried. Before cementation, each disk and base was placed on a micrometer (Mitutoyo digital micrometer; Mitutoyo Corp), and the unit was calibrated to zero. The ceramic specimen and base were then cemented to each other using the micrometer to ensure a cement space of 50 μm. Specimens were tested after 2 weeks of storage in deionized water at 37°C.

A staircase sensitivity (up-and-down) statistical protocol was used. Cyclic loading was done with a servo-hydraulic machine (858 Mini BionixII; MTS) under load control. Cemented specimens were secured in a base...
holder at the bottom of a stainless steel well containing deionized water at room temperature. Disks were centrally loaded using the 2-mm–diameter G10 piston. A sheet of polyethylene (0.1-mm thick) was placed between the piston and disk to further reduce contact stress concentrations. Sinusoidal cyclic loading was applied to ceramic specimens at a frequency of 20 Hz from 10 N to the target load for 500,000 cycles. Initially, 4 to 5 specimens were tested at a step size of 50 N to identify the 50% probability of failure load (that is, the load where either first cracking or survival occurred). After cyclic loading was completed, specimens were examined for subsurface crack formation by transillumination. If cracked, the next specimen was cycled at a lower load. If the specimen did not fail, the subsequent specimen was cycled at a higher load. This step size was decreased to 25 N for formal testing to determine means and standard deviations. For staircase sensitivity analysis, means and standard deviation values were used to compare among groups using ANOVA for summary statistics and a 95% post hoc test.

### RESULTS

In almost all specimens, fatigue cracks were subsurface radial cracks, and only these were used for statistical analysis. The distribution of radial and surface cracks (eg, Hertzian cone cracks) are listed in Table 1.

This distribution was significantly different than expected if surface cracking was randomly distributed. Limited cone cracking was observed only in the 2 thinner core groups, based on total sample size ($\chi^2$ test: $P<.05$) possibly indicating residual tensile stress in veneer and compressive in core. Only subsurface radial cracks were expected based on elastic stress analyses (outlined above in the introductory section). It should also be noted that the 0.5-mm core group was reduced in numbers due to cracking during ceramic machining.

Means and standard deviations (based on staircase sensitivity statistics) for fatigue crack development are shown in Figure 4. If Hertzian cracks appeared within a specimen group, they were excluded from the statistical analysis of fatigue data but were included in the $\chi^2$ test. All veneered groups were significantly stronger than the full-thickness group (ANOVA: $P<.001$; 95% post hoc).

### DISCUSSION

The widely held assumption that the fatigue strength of a bilayered ceramic improves if the thickness of the core is increased appears to be incorrect. Recent data published by Hsueh and Miranda suggested that the failure loads of bilayered systems should be independent of core thickness (dependent only on the square of total thickness). Our results, which resonate with those of Hsueh and Miranda, furthermore suggest that the addition of veneering ceramic may actually improve the fatigue strength of this particular bilayered system.

Residual stresses appear to be present at room temperature in bilayered lithium disilicate. These stresses are generated after sintering upon cooling and originate from differences in the thermal expansion/contraction behaviors of the ceramic layers. Hence, a compressive-tensile residual stress state is formed. IPS e.max Ceram veneering ceramic was developed to be compatible with both lithium disilicate and zirconia. The coefficient of thermal expansion of the veneering ceramic is reported by Ivoclar Vivadent Inc to be lower than that of the core (IPS e.maxCeram = 9.5 × 10^-6/K; IPS e.maxCAD = 10.15 × 10^-6/K) as seen in Figure 5.

If this were correct, as is the case with metal-ceramic systems, the veneer would be in a state of compression and the core in tension. If the veneer were in compression, the likelihood of developing Hertzian cone cracks would decrease rather than increase, as seen in our results. Unpublished work by Kelly and Hill, 2010 (University of Connecticut and Ivoclar Vivadent Inc), testing the same hypothesis with monotonic loading of dry specimens produced even more dramatic results. All 0.5-mm IPS e.max CAD core specimens were tested at a step size of 50 N to identify the 50% probability of failure load (that is, the load where either first cracking or survival occurred). After cyclic loading was completed, specimens were examined for subsurface crack formation by transillumination. If cracked, the next specimen was cycled at a lower load. If the specimen did not fail, the subsequent specimen was cycled at a higher load. This step size was decreased to 25 N for formal testing to determine means and standard deviations. For staircase sensitivity analysis, means and standard deviation values were used to compare among groups using ANOVA for summary statistics and a 95% post hoc test.

### Table 1. Distribution of subsurface radial and surface cracks (Hertzian cone cracks) in each specimen group

<table>
<thead>
<tr>
<th>Crack Type</th>
<th>0.5-mm Core</th>
<th>0.75-mm Core</th>
<th>1.0-mm Core</th>
<th>1.5-mm Core</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radial</td>
<td>9</td>
<td>20</td>
<td>21</td>
<td>22</td>
</tr>
<tr>
<td>Hertzian</td>
<td>3</td>
<td>2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Sample size</td>
<td>28</td>
<td>41</td>
<td>39</td>
<td>47</td>
</tr>
</tbody>
</table>

### Figure 4. Mean fatigue failure loads (N) and standard deviations for each core thickness group, using staircase sensitivity test.
Furthermore, data presented in Figure 5 date back to 2005, a time when Ivoclar Vivadent Inc noted significant problems with the veneering porcelain for e.max CAD, leading to its reformulation (J.R.K., personal communication, 2005). If data in Figure 5 are correct, this means that absolute values of the differences in expansion coefficients ($|\Delta z|$) would be approximately 0.69 for e.max CAD and 1.27 for e.max ZrCAD. According to the findings of Steiner et al., this would lead to failure probabilities of cracking after the firing of central incisor crowns of approximately 16% for e.max CAD and 38% for e.max ZrCAD. Despite this, both systems have proved highly successful clinically, indicating that the published coefficients may not be correct. Last, coefficients of expansion from slow-heating dilatometry (5°C/min) do not take into account the fact that stresses develop during cooling (not heating), contraction curves can be different from expansion curves, and contraction behavior during rapid oven cooling in the dental laboratory cannot be reproduced during slow heating.

We hypothesize that the results, in addition to preliminary research, strongly suggest that these stresses result in higher protective stresses in the core ceramic. This creates a situation in which tensile stresses are found in the veneer and compressive stresses in the core, strengthening the bilayered ceramic as a whole. Sglavo et al. described the potential benefits of introducing residual stress profiles into the different layers of a bilayered ceramic to improve mechanical properties. Conventionally, metal ceramic crowns have been engineered so that the coefficient of thermal expansion of the metal is higher than that of the porcelain. This mismatch places the porcelain under compression, thereby improving its mechanical properties.

It would be advantageous to verify our results with finite element analysis, paying specific attention to the role residual stresses can play in fatigue strength. Determining and analyzing the role of the coefficient of thermal contraction for the veneer and core, as opposed to the coefficient of thermal expansion, are probably more appropriate for understanding the behavior of bilayered ceramics. As mentioned previously, it is during cooling from the sintering temperature to room temperature that residual stresses are introduced. We believe these residual stresses play a role in the fatigue strength of bilayered ceramics. Direct observation of residual stresses can be accomplished with an optical polarimeter. As mentioned above, Steiner et al. suggested in 1997 that $|\Delta z|$ differences of 1.5 to 1.6 $\times 10^{-6}/K$ would result in complete fracture of bilayered systems. A mismatch between IPS e.max Ceram and zirconia is shown in Figure 5 that would most likely result in some fractures. Therefore, the validity of the data presented in Figure 5 is questionable, and measuring the coefficient of thermal contraction may be more accurate.

If, as it appears, the fatigue strength is independent of core thickness, dentists/technicians should be encouraged to maximize the thickness of the veneering porcelain while maintaining a minimum core thickness. Veneering ceramics closely replicate the optical properties of enamel and dentin because of their amorphous 3-dimensional structure. Core ceramic materials, typically particle-filled glass or polycrystalline ceramic, are less esthetic and more opaque materials. The ability to maximize the veneering layer will, therefore, improve the esthetics and patient satisfaction.

In the future, a protocol that allows crack pop-in to be acoustically identified during cyclic loading of a bilayered dental ceramic should be developed. This will enable the use of lifetime statistical analysis. Extrapolation of accelerated condition data to clinical use loads would be possible with such analysis.

**CONCLUSIONS**

Our results indicate that the addition of veneering ceramic to lithium disilicate cores increases the fatigue strength of the biceramic system. Furthermore, our results suggest that for this system, increasing the thickness of the veneering ceramic, rather than the core ceramic, may have a positive impact on overall fatigue strength.

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