Clinical Performance of a Lithia Disilicate–Based Core Ceramic for Three-Unit Posterior FPDs

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Purpose: The purpose of this research project was to determine the clinical success rate of a lithia disilicate–based core ceramic for use in posterior fixed partial dentures (FPD) as a function of bite force, cement type, connector height, and connector width. ***Materials and Methods:*** Thirty ceramic FPD core frameworks were prepared using a heat-pressing technique and a lithia disilicate–based core ceramic. The maximum clenching force was measured for each patient prior to tooth preparation. Connector height and width were measured for each FPD. Patients were recalled yearly after cementation for 2 years and evaluated using 11 clinical criteria. All FPDs were examined by two independent clinicians, and rankings from 1 to 4 were made for each criterion (4 = excellent; 1 = unacceptable). ***Results:*** Two of the 30 ceramic FPDs fractured within the 2-year evaluation period, representing a 93% success rate. One fracture was associated with a low occlusal force and short connector height (2.9 mm). The other fracture was associated with the greatest occlusal force (1,031 N) and adequate connector height. All criteria were ranked good to excellent during the 2-year recall for all remaining FPDs. ***Conclusion:*** The performance of the experimental core ceramic in posterior FPDs was promising, with only a 7% fracture rate after 2 years. Because of the limited sample size, it is not possible to identify the maximum clenching force that is allowable to prevent fracture caused by interocclusal forces. *Int J Prosthodont 2004;17:469–475.*

The advent of an esthetics-conscious society and the associated demand for fracture-resistant, tooth-colored prosthetic materials have led to the development of tougher, more esthetic restorative materials intended to withstand greater occlusal forces. Ceramic prostheses have found a niche because of the increasing demand for natural-appearing restorative materials.

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Through the years, several technologies, including crystalline reinforcement, thermal tempering, and chemical strengthening, have been adopted to improve the fracture resistance of ceramics.[1] These procedures have paved the way for modern ceramic systems, including those based on leucite-reinforced glass, glass infiltration of partially sintered alumina, high-density alumina and zirconia, and a lithium disilicate glass-ceramic. These ceramics were initially used in limited restorations such as Class I inlays but rapidly progressed to complete-coverage onlays and crowns. With continued improvements in strength, these ceramics were later used for anterior fixed partial dentures (FPD), producing unmatched esthetics with their depth of color and increased translucency.

As the demand continued for improved esthetics and fracture resistance, new core ceramics were developed. Using the technology mentioned earlier, these stronger ceramics were used as cores to serve as frameworks for the more esthetic, but weaker, veneering ceramics. These ceramics can sometimes be indicated for use in posterior FPDs, depending on the esthetic demands of the patient. Clinical studies have shown that survival of all-ceramic crowns is generally higher in the anterior area (93% to 98%) and gradually decreases in the posterior region of the oral cavity (85% to 94%).[2,3] The increased occlusal loading associated with posterior areas has deterred placement of ceramic prostheses. Clinical studies involving posterior ceramic FPDs made from lithium disilicate glass conclude that an acceptable success rate can be achieved if design requisites are followed and adequate connector dimensions are maintained.[4] Other studies show a 90% success rate after 5 years with In-Ceram (Vident) FPDs.[5]

Several cements were developed to enhance the use of ceramic prostheses. The linear expansion of resin-modified glass-ionomer luting cements ranges from 0.4% to 3.1% after storage in 0.9% saline solution at 37°C for 6 months.[6] Resin-modified glass-ionomer cements that exhibit high hygroscopic expansion may cause fracture of ceramic crowns.[7,8] A high degree of cross-linking in the low-expansion resin-modified glass-ionomer cements is claimed to greatly reduce hygroscopic expansion and the risk for fracture of ceramic crowns. Resin cements, on the other hand, exhibit minimal hygroscopic expansion ($< 0.2\%$) and are believed to be better suited for luting of ceramic crowns, although they do not provide the potential for sustained release of fluorine ions. Resin-modified glass-ionomer cements have also been associated with higher solubility, causing marginal ditching[9,10] and increased tooth sensitivity after luting.[11]

The aims of this research were to test the following hypotheses:

1. Three-unit FPDs made of a high-strength core ceramic would exhibit good to excellent clinical performance (based on 11 evaluative criteria) and would adequately resist fracture in posterior situations (excluding third molars) if fabricated with the minimum connector size (4 mm $\times$ 4 mm).
2. A reinforced glass-ionomer cement (ProTec CEM, Ivoclar Vivadent), when used to cement core ceramic crowns in posterior FPDs, would be associated with significantly lower marginal quality but similar fracture resistance compared with a dual-cure resin cement (Variolink II, Ivoclar Vivadent).
3. There would be no significant difference in tooth sensitivity associated with FPDs cemented with glass-ionomer cement and dual-cure resin cement.

Materials and Methods

All patient recruitment and treatment was performed at the University of Florida College of Dentistry, Graduate Prosthodontic Clinic, by prosthodontic faculty. Patients were initially screened to exclude individuals with medical contraindications to dental treatment, parafunctional habits, and inability to ensure residence in the area for the next 5 years. Inclusion criteria were a missing posterior tooth in a quadrant (first premolars through second molars) that could be restored with a three-unit FPD, periodontal pockets of less than 4 mm for each abutment, no periodontal disease, vital abutment teeth, and a crown:root ratio of at least 1:1. A patient could have multiple FPDs placed, as long as the above-mentioned criteria were met. The following baseline data were obtained for each selected subject:

- General medical history and physical examination
- Primary casts made with irreversible hydrocolloid impression material
- Bite force measurement made with a gnathodynamometer
- Pocket depths of abutment teeth
- Periapical radiographs of abutment teeth

The maximum occlusal force exerted by each subject was measured prior to commencing treatment using a bite force gauge that has been reported previously.[12] The purpose of these measurements was to analyze the influence of occlusal force on the survival of the FPDs. A total of 30 FPDs were fabricated with the core ceramic for 21 patients, all of whom were recalled each year for 2 years. Three clinicians performed treatment, and one technician using an in-house laboratory accomplished all lab work. Of the 21 patients, 18 were women and 3 were men, with ages ranging from 30 to 62 years. The three-unit FPDs were located in the posterior area, with canines serving as the most anterior abutment and second molars as the most posterior abutment. All FPDs were opposed by natural dentition. The dimensions for tooth reduction included at least 1 mm of axial reduction, 2 mm of occlusal reduction, and incorporation of a shoulder or a deep chamfer margin design with rounded line angles. Final impressions were made using a dual impression technique with high- and low-viscosity polyvinyl siloxane in a stock tray. Provisional acrylic resin FPDs were made and cemented with provisional cement. FPDs were processed by heat pressing the core ceramic (Ivoclar Vivadent) and applying stain and glaze as necessary. The heat-pressed ceramic system uses the lost-wax technique, whereby the FPD is waxed to its proper shape, contoured, and invested in a special flask with a special type of investment material. The desired shade of a precerammed ceramic cylinder is plasticized at 1,100°C and pressed under vacuum and pressure into the mold of the investment.[13]

The ceramic FPDs were inspected to ensure that the incisogingival height and curvature of the gingival embrasure of the connectors were adequate to resist

Table 1 Descriptive Outline of FPDs

FPD position*	Height × width of mesial connector (mm)	Height × width of distal connector (mm)	Bite force (N)
23–25	4.1 × 5.0	4.0 × 6.1	284
24–26	4.5 × 5.4	4.0 × 5.7	155
43–45	4.5 × 3.8	4.2 × 4.4	781
25–27	4.7 × 7.2	4.3 × 8.2	781
35–37	4.0 × 4.9	3.9 × 6.2	781
24–26	4.5 × 7.0	4.5 × 6.8	382
24–26	3.4 × 6.4	4.3 × 6.4	266
13–15	4.3 × 5.4	4.3 × 6.5	266
34–36	2.9 × 5.2	3.4 × 6.0	373
24–26	3.4 × 7.3	3.8 × 8.3	373
34–36	3.6 × 5.5	3.0 × 6.2	373
13–15	5.4 × 4.5	4.8 × 5.3	364
23–25	5.4 × 5.9	5.3 × 6.6	NA
23–25	5.1 × 5.4	4.8 × 6.5	222
13–15	3.8 × 7.2	4.2 × 7.7	515
23–25	4.2 × 5.4	4.8 × 6.0	NA
13–15	6.0 × 5.1	5.1 × 5.6	NA
13–15	5.6 × 5.8	4.9 × 6.6	564
34–36	4.0 × 5.1	3.5 × 6.3	1,031
34–36	3.9 × 4.6	5.2 × 5.6	218
13–15	5.7 × 6.8	4.7 × 7.7	204
43–45	4.4 × 5.2	4.9 × 6.0	204
33–35	4.4 × 5.2	5.1 × 5.9	204
13–16	4.7 × 5.5	4.9 × 7.1	719
25–27	5.2 × 5.8	5.2 × 6.7	719
34–36	3.0 × 5.7	4.5 × 6.7	NA
34–36	5.5 × 4.2	5.4 × 4.8	435
14–16	4.3 × 5.0	4.0 × 5.7	795
35–37	4.3 × 5.7	4.3 × 6.2	364
14–16	4.9 × 5.7	4.6 × 5.9	631

*Fédération Dentaire Internationale tooth-numbering system.

fracture when subjected to normal bite forces. The minimum recommended dimensions—for premolar FPDs, 4 mm × 4 mm, and for molar FPDs, 4 mm × 5 mm—were ensured for each FPD when esthetics and gingival contour permitted. Connector heights and widths were measured for each FPD using a Boley gauge. FPDs were cemented using either a resin-reinforced glass-ionomer cement (Protec CEM; n = 14) or a dual-cure resin cement (Variolink II; n = 13) using a random-number table.

Patients were recalled after cementation annually for 2 years and evaluated for the following clinical criteria: *(1)* tissue health; *(2)* secondary caries; *(3)* occlusion; *(4)* proximal contact; *(5)* marginal integrity; *(6)* absence of sensitivity to percussion, heat, cold, and air; *(7)* color match; *(8)* surface texture; *(9)* absence of wear of opposing teeth; *(10)* anatomic contour; and *(11)* cracks/chips or fracture. This system was derived from the California Dental Association quality assessment evaluation system.[14] FPDs were examined by two independent clinicians who did not prepare the teeth or cement the prostheses, and rankings of each criterion were made from 1 to 4 (4 = excellent; 3 = good; 2 = unacceptable, needs repair or replacement in the near future; and 1 = unacceptable, needs immediate replacement). All clinicians/evaluators were subjected to several calibration exercises that consisted of tabletop analysis of marginal openings as well as slide evaluations of different clinical situations.

Data were analyzed by logistic regression analysis of the variables, with $\alpha = .05$. The consistency of the examiners' scores was evaluated using a test for the standard deviation (SD) of interexamination.

Results

The consistency of the examiners' scores was evaluated and defined as the SD of interexamination. The SDs for each criterion were: *(1)* 0.35, *(2)* 0.33, *(3)* 0.13, *(4)* 0.27, *(5)* 0.42, *(6)* 0.00, *(7)* 0.40, *(8)* 0.13, *(9)* 0.00, *(10)* 0.38, and *(11)* 0.13. The lower the SD, the higher the agreement between the two examiners, with an SD of 0 indicating 100% agreement in scoring.

Two of the 30 ceramic FPDs fractured within the 2-year evaluation period (one survived for 529 days, the other for 750 days), representing a 93% success rate. One FPD fracture occurred in the subject exhibiting the greatest clenching force (1,031 N), whereas the other fracture was associated with a connector height of 2.9 mm (Table 1).

Table 2 Clinical Performance of Surviving FPDs at 1- and 2-Year Recall Examinations

Criterion*	Good rating	Excellent rating
Year 1 (n = 29)		
1[†]	5	23
2	0	29
3	1	28
4	4	25
5	10	19
6	2	27
7	8	21
8	1	28
9	0	29
10	11	18
11	1	28
Year 2 (n = 28)		
1	4	24
2	0	28
3	0	28
4	4	24
5	13	15
6[‡]	3	24
7	13	15
8	1	27
9	1	27
10	2	26
11	0	28

*See Materials and Methods section for criteria specifications.
[†]Not including one patient with poor tissue health that resolved after the first recall.
[‡]Not including one patient with marked sensitivity to cold and percussion.

For the patients included in the study, the mean maximum clenching force was 461 ± 246 N. All surviving FPDs fell into the good to excellent category for the 11 criteria during the first and second recall examinations, except for two instances (Table 2). Poor tissue health was recorded for one patient during the 1-year recall examination. This condition was resolved without professional intervention before the 2-year recall, possibly because of more detailed oral hygiene instructions. During the 2-year recall, sensitivity to cold and percussion was noted in one patient, although further observation is necessary to determine the possible cause and need for replacement if symptoms persist.

Data were analyzed as a function of time via logistic regression with a term included for maximum clenching force. The results indicated no statistically significant effect of maximum occlusal force on the incidence of chipping or fracture at either 1- or 2-year recall exams ($P = .445$ and .230, respectively). No significant difference in marginal integrity was observed between the two cements after 1 and 2 years ($P > .050$). Variolink II resin cement exhibited a greater proportion of excellent scores than Protec CEM ($P = .001$). Some marginal washout was observed for Protec CEM. There was no statistically significant difference in tooth sensitivity ($P = .961$) associated with the two cements. Scanning electron microscopic (SEM) analysis was performed on the fractured surfaces to reveal that the critical flaw originated from gingival areas of the distal connector.

Discussion

The success rate for the core ceramic FPDs was 93% within a period of 2 years. Fractures occurred in 2 of the 30 FPDs. Other clinical studies of all-ceramic FPDs show comparable longevity of 90% to 93% within a 5-year observation period.[5,15] This is lower than the survival rates of 95% to 97.7% reported for metal-ceramic FPDs after 5 to 7.5 years.[16,17] Despite the slightly lower survival rate, ceramic FPDs are still indicated specifically for esthetic reasons. Considering the greater esthetic demands of the general population, it is easier to achieve esthetic results with ceramic prostheses with less tooth reduction compared with metal-ceramic restorations (1.0 to 1.5 mm versus 1.2 to 1.7 mm for anterior restorations).[18]

Failure of ceramic FPDs often results from complicated stress patterns introduced during the process of mastication. Because of the brittle nature of ceramics, tensile stresses are tolerated poorly and often result in fracture. Ceramic prosthesis failure often occurs at the connector site along the gingival area.[19,20] This type of fracture occurs because of tensile stresses within the connector caused by flexure (Fig 1).[21] In contrast, a cantilevered FPD results in tension developing within the occlusal connector area. One of the fractured FPDs (mandibular right first premolar to first molar) was seen in a patient who showed the highest maximum clenching force in the study (1,031 N). The patient initially reported chewing on something hard and then hearing a crack. Several weeks later, the FPD was loose. Figure 2 shows the application of a dye to reveal fracture on both the mesial and distal connector areas. SEM analysis on the distal connector (Fig 3) revealed the critical flaw originating from the gingival area. SEM analysis of the mesial connector (Fig 4) revealed the critical flaw located on the occlusal surface. It can be surmised that the fracture occurred initially on the distal connector, originating from the critical flaw along the gingival area. This produced a cantilever effect on the mesial connector, leading to its eventual fracture from the occlusal surface. Since ceramics lack the ability to deform plastically, concentrated tensile stresses can lead to failure. This was confirmed by fractographic analyses that support the recommendation of an increased radius of curvature at the gingival embrasure to improve fracture resistance.[22]

The other fractured FPD (mandibular left first molar to first premolar) also fractured along the distal connector, although segments of this FPD were not recovered. The connector heights were 2.9 and 5.2 mm.

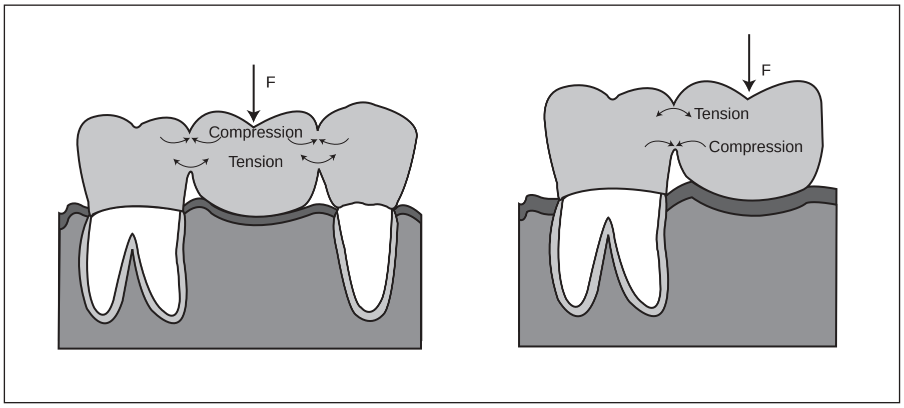
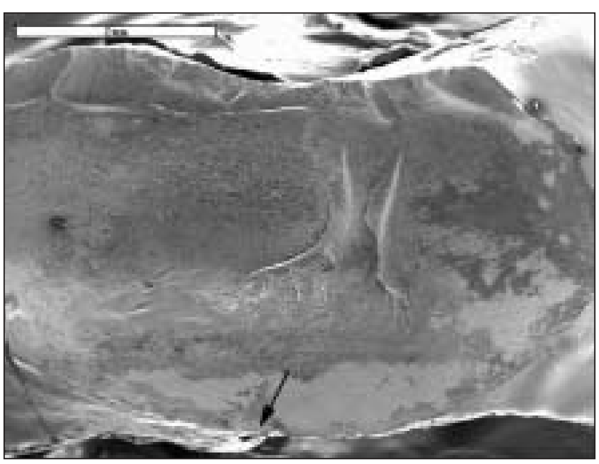


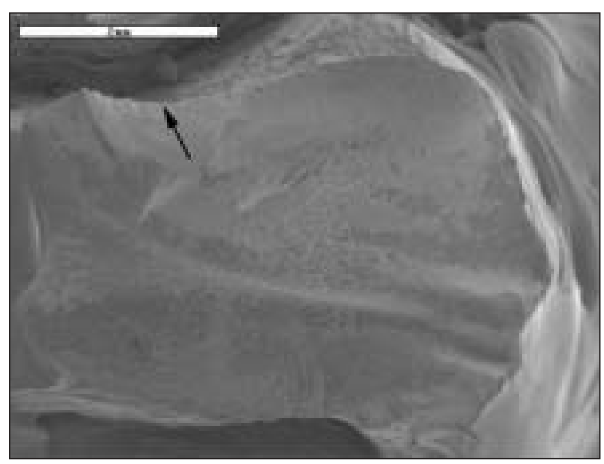
Fig 1 Compression and tension forces in a three-unit FPD *(left)* and cantilevered FPD *(right)* when flexural force *(F)* is applied. (Adapted from Anusavice.[21])

Fig 2 *(right)* Fractured FPD in patient exhibiting high occlusal force; distal fracture occurred first, followed by mesial fracture caused by cantilever forces. Reprinted from Anusavice KJ. Informatics systems to assess and apply clinical research on dental restorative materials. Adv Dent Res 2003;17:43–48.

Fig 3 *(below)* SEM analysis of FPD in Fig 2 shows critical flaw in distal connector site. Fracture originates gingivally *(arrow)* and propagates occlusally, consistent with tension/compression diagram in Fig 1.

Fig 4 *(below right)* Mesial connector shows critical flaw *(arrow)* originating from occlusal surface and moving gingivally to complete fracture, consistent with cantilever forces.





The manufacturer recommends a minimum connector height of 4.0 mm, as some studies[20,23] have shown a 40% to 50% stress reduction at this level compared with shorter connectors, although this becomes difficult to achieve as crown height becomes shorter posteriorly. The connector design is often dictated by embrasure contour, strength, cleansability, and esthetics; these criteria will have to be considered, and consequences weighed, prior to recommending a ceramic FPD. The fracture resistance of ceramic crowns is controlled by several factors, including ceramic thickness.[23,24] Therefore, the thicker the ceramic, the less likely it is to fracture.

The linear expansion of approximately 0.36% caused by water absorption of Protec CEM hybrid ionomer appears to pose a minimal risk for fracture of glass-ceramic crowns. This is in agreement with other research showing no significant difference in the fracture incidence of ceramic crowns when cemented with resin-modified glass-ionomer.[25,26] In addition to the use of explorers, impressions of subgingival margins are necessary to confirm the marginal integrity scores and extent of washout. Resin-modified glass-ionomer has been shown to exhibit an increased incidence of marginal cement degradation as a result of marginal breakdown and attrition.[27,28] Compared to other luting agents such as zinc polycarboxylate and zinc phosphate, glass-ionomer cements show significantly less material loss after continuous erosion cycling.[29] Pulpal sensitivity does not seem to be affected by the use of resin-modified glass-ionomer.[30]

Comparative in vivo and in vitro studies between resin cements and resin-modified glass-ionomer cements have also shown better performance of the resin cements. Resin-modified glass-ionomers exhibit an adhesive bond failure at the cement-ceramic interface, leading to fracture or loss of the prosthesis. Resin cements exhibit a 2% failure rate, whereas resin-modified glass-ionomers show a 15% failure rate.[31] Another study claims that the corrosive acidic environment created by acid-base cements such as zinc phosphate, zinc polycarboxylate, and resin-modified glass-ionomers propagates preexisting flaws in the porcelain, leading to higher failure rates of ceramic prostheses. It also claims that resin cements increase the strength of ceramic prostheses by "healing" surface imperfections.[32]

Conclusions

The performance of a lithia disilicate–based core ceramic is promising, with a success rate of 93% after 2 years. Long-term analysis with a larger sample size is needed to evaluate the effect of occlusal force and connector thickness on fracture resistance of ceramic prostheses to develop standard guidelines for fabrication. Several recommendations can be made from this clinical study:

1. Although there was no correlation between occlusal forces and predisposition to fracture, clinicians should make a conscious effort to minimize high forces on all-ceramic FPDs because of their brittle nature. Patients with parafunctional habits, those with excessive occlusal forces, or those who are prone to occlusal trauma (eg, athletes) should not receive ceramic prostheses.
2. The connector height should be maximized within the limitations of esthetics and gingival health. Based on fractographic analyses, the radius of curvature of the gingival embrasure should be increased as much as is feasible to improve fracture resistance.
3. Because of some degradation of the resin-modified glass-ionomer cement, the use of dual-cure resin cement is recommended for the cementation of all-ceramic restorations for better marginal adaptation.

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Literature Abstract

Gradient surface porosity in titanium dental implants: Relation between processing parameters and microstructure

Titanium dental implants with porous surfaces that have pore sizes between 150 and 400 µm were found to provide space for vascular tissue that lead to mineralized bone growth during healing. Sintering metal powder or titanium beads, electro-discharge compaction and plasma spraying are several methods of creating porous surfaces on dental implants. A method of using microwave sintering to sinter titanium powder that would result in a porous surface was described in this study. By subjecting titanium powder to microwave radiation, the titanium powder could be sintered, resulting in a dense core with a porous surface. The reason for this phenomenon is that heat is generated in the interior of the condensed titanium powder and is dissipated toward the surface, producing a well-sintered core with a porous surface. Titanium disks were sintered at 1.0, 1.25, and 1.5 kW for various durations (10, 20, and 30 minutes) in a semi-industrial microwave furnace. Temperature on the sample surfaces was measured with an optical pyrometer and the maximum temperatures recorded for 1.0, 1.25, and 1.5 kW were 1,220°C, 1,300°C, and 1,340°C, respectively, which was lower than the melting temperature of pure titanium at 1,943°C. However, the specimen core would have a high enough temperature to sinter the titanium powder densely, as claimed by the authors. SEM micrograph of the sintered samples demonstrated a gradient porosity at the surface and the pores appeared interconnected. The thickness of the porous region was about 100–200 µm and the interconnected pore sizes were approximately 30–100 µm, which could provide an ideal surface for osteoblast cell growth. These samples were subjected to tensile testing and the sample that was sintered at 1.25 kW for 20 minutes was found to have the highest strength value, exceeding 400 MPa. This was comparable to reported values for CP titanium with values ranging from 345 to 550 MPa. When sintered to 30 minutes, strength of the titanium samples decreased due to cracks formed at the interface of the porous region and the dense region. The authors concluded that this method of processing titanium provided an ideal porous surface for dental implants and could also provide better stress transfer than a coated surface since the porous surface and the dense core are composed of the same material.

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