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Reliability of Reduced-thickness and Thinly Veneered Lithium Disilicate Crowns

ABSTRACT

The present investigation hypothesized that the reliability of reduced-thickness monolithic lithium disilicate crowns is high relative to that of veneered zirconia (Y-TZP) and comparable with that of metal ceramic (MCR) systems. CAD/CAM first mandibular molar full-crown preparations were produced with uniform thicknesses of either 1.0-mm or 2.0-mm occlusal and axial reduction, then replicated in composite for standard crown dies. Monolithic 1.0-mm (MON) and 2.0-mm CAD/CAM lithium disilicate crowns, the latter with a buccal thin veneer (BTV) of 0.5 mm, were fabricated and then sliding-contact-fatigued (step-stress method) until failure or suspension (n = 18/group). Crack evolution was followed, and fractography of *post mortem* specimens was performed and compared with that of clinical specimens. Use level probability Weibull calculation (use load = 1,200 N) showed interval overlaps between MON and BTV. There was no significant difference between the Weibull characteristic failure loads of MON and BTV (1,535 N [90% CI 1,354–1,740] and 1,609 N [90% CI 1,512–1,712], respectively), which were significantly higher than that of Y-TZP (370 N [90% CI 322–427]) and comparable with that of MCR (1,304 N [90% CI 1,203–1,414]), validating the study hypothesis.

KEY WORDS: biomaterial(s), ceramics, glass-ceramic, prosthetic dentistry, prosthodontics, restorative materials.

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INTRODUCTION

Compared with other all-ceramic systems, yttrium oxide partially stabilized tetragonal zirconia polycrystal (Y-TZP) exhibits superior mechanical properties, due partly to a transformation toughening mechanism (Christel *et al*., 1989; Tinschert *et al*., 2000). Although zirconia fixed partial dentures have been most commonly investigated (Tinschert *et al*., 2005; Vult von Steyern *et al*., 2005; Sailer *et al*., 2006), the sparse information regarding single crowns has shown similar results concerning the principal failure mode, which involved fractures within the veneering ceramic (Cehreli *et al*., 2009; Ortorp *et al*., 2009; Groten and Huttig, 2010). A method to reproduce these failure modes in the laboratory has been developed using mouth-motion fatigue-testing in water (Coelho *et al*., 2009a,b; Silva *et al*., 2010), where mechanical testing has confirmed the susceptibility of zirconia crowns to porcelain veneer failure. Considering that porcelain veneer fracture in Y-TZP restorations is a complex phenomenon—one likely explained by differences in coefficient of thermal expansion (CTE) between framework and porcelain, residual thermal stresses from cooling, and core/framework design, among other factors—there is still much to be clarified (Zarone *et al*., 2011). Contributing to the continued use of these crowns have been specific changes in the core design aimed at optimizing porcelain support, as well as improvements observed in reliability and reduced fracture sizes of the porcelain veneer relative to a uniform thickness core for Y-TZP crowns (Silva *et al*., 2010).

The lithium disilicate glass-ceramic system, whether CAD/CAM-processed or heat-pressed, is indicated either as a full-contour (monolithic) restoration or as a core for subsequent porcelain veneering. Recent 2-year clinical recalls on full-contour lithium disilicate crowns have shown promising results in terms of structural integrity, with no mechanical failures such as fracture or chipping (Fasbinder *et al*., 2010; Reich *et al*., 2010). In agreement with these clinical findings, 2-mm-thick full-contour molar crowns of IPS e.max CAD subjected to sliding contact fatigue-testing have demonstrated significantly higher reliability than porcelain-layered Y-TZP crowns (Guess *et al*., 2010). Although the clinical results are promising, occlusal clearances of less than 2 mm may reduce the reliability of lithium disilicate full-contour crowns. This concern warrants investigation. In addition, the use of veneering porcelains to improve esthetics would require a reduction in core thickness that could also limit crown mechanical performance in the posterior region.

Since metal ceramic (MCR) crowns are generally considered the "gold standard", and Y-TZP restorations are of significant interest for clinical use, both materials should be compared with lithium disilicate crowns tested with the same methodology.

This investigation tested the hypothesis that the reliability of reduced-thickness monolithic lithium disilicate crowns is higher than that of zirconia (Y-TZP) and comparable with that of metal ceramic systems.

MATERIALS & METHODS

A CAD-based 3-D model of a mandibular first molar full-crown preparation was generated (Pro/Engineer Wildfire 5.0, PTC, Needham, MA, USA) with uniform preparation occlusal reductions of 1.0 mm or 2.0 mm, and proximal axial wall reduction of 1.5 mm. Crowns investigated were full-contour lithium disilicate (IPS e.max CAD, Ivoclar Vivadent, Schaan, Liechtenstein [hereafter Ivoclar]) of either 1.0-mm occlusal (MON) or 2.0-mm occlusal dimension, the latter with a buccal thin veneer (BTV) of fluorapatite ceramic (IPS e.max Ceram, Ivoclar). Before final maturation, a sample at "blue block" stage was sent for marginal fit evaluation. Final crowns were cemented to resin-based composite dies (Tetric EvoCeram, Ivoclar) and aged 30 days (Coelho *et al*., 2009a,b; Silva *et al*., 2010, 2011) with a self-curing resin-based dental luting material (Multilink Automix, Ivoclar) according to the manufacturer's instructions. These were then stored in water for a minimum of 7 days prior to mechanical testing.

Following incubation time, specimens (n = 3, each group) were mounted in a universal testing machine (Model 5566 [dual-column tabletop], Instron, Norwood, MA, USA), and single-load-to-fracture (SLF) tested at 1 mm/min with a WC indenter (r = 3.18 mm) on the distobuccal cusp inner incline. Based on these data, samples (n = 18, each group) were subsequently exposed to mouth-motion step-stress fatigue in water (Coelho *et al*., 2009a,b; Silva *et al*., 2010, 2011). Sliding contact fatigue-testing (SCFT) was performed by sliding the WC indenter (r = 3.18 mm) 0.7 mm lingually down the distofacial cusp incline toward the central fossa, beginning at 0.5 mm lingual to the cusp tip, at 2 Hz with a fatigue chewing simulator (ELF-3300, EnduraTEC, Bose, Eden Prairie, MN, USA). At the end of each load cycle step, all specimens were inspected under polarized light stereomicroscopy (MZ APO stereomicroscope, Leica Microsystems Ltd., Heerbrugg, Switzerland) for cracks and ceramic surface damage evaluation. The load at which an initial crack was observed was noted, and testing continued until either specimen fracture or completion of testing (suspension). Use level probability Weibull plots (unreliability *vs.* cycles) were calculated at 90% confidence intervals for a use load that generated the best data comparison for MON and BTV. Load at failure Weibull probability calculations x (characteristic failure load) of MON and BTV were compared with those of MCR (PdAg, Superior Plus, Jensen Industries, North Haven, CT, USA; (Silva *et al*., 2011), as well as with hand-layer-veneered Y-TZP crowns with conventional (uniform 0.5-mm thickness) and modified frameworks (0.5-mm thickness; 1 mm thick lingually; built up 2.5 mm occlusally to lower the veneering porcelain bulk of the cusps; Silva *et al*., 2010) with 2-mm crown preparation.

Failed and suspended samples were embedded in epoxy resin (Epofix Resin, Struers, Ballerup, Denmark), sectioned, and inspected for evaluation of crack pattern (MZ APO stereomicroscope, Leica Microsystems Ltd., Heerbrugg, Switzerland). Images of clinical and laboratory fracture specimens were then compared.

RESULTS

The SLF mean value (n = 3) and standard deviation were 2,474 ± 630 N and 2,052 ± 151 N for MON and BTV, respectively.

Use level probability Weibull plots with use load of 1,200 N (best data fit) for MON and BTV are shown in Figs. 1A and 1B. The wide confidence bounds are the result of a large number of test suspensions in both groups (*i.e*., samples that cracked but did not fracture throughout the fatigue profiles). The Weibull modulus (beta, β) values of 0.94 (MON) and 0.90 (BTV) showed that fatigue was not a strong acceleration factor for failure.

Using load at failure during SCFT, we calculated Weibull 2-parameter probability multiplots. Weibull modulus (β) and characteristic failure load (eta, η, the load at which 63.2% of specimens would fail) were calculated and plotted. These data were compared with our previous results for Y-TZP and MCR crowns using contour multiplots (see Figs. 1C, 1D, and Table). As can be seen, the conventional Y-TZP group presented the lowest characteristic failure load value. Modified Y-TZP showed characteristic failure load value higher than that of conventional Y-TZP but lower than those of MCR, MON, and BTV.

The chief failure mode for MON and BTV was bulk fracture. Sectioned suspended specimens (Fig. 2) showed the competing failure modes of water-induced surface partial cone cracks (Zhang *et al*., 2005; Kim *et al*., 2008) located under the beginning of the indenter sliding trail—in tandem with flexural radial cracks coming from the crown–cement interface toward the occlusal surface. Whereas a series of incomplete partial cones formed along the wake of the moving indenter, only one usually dominated and competed with a radial crack.

Images of a tested suspended specimen loaded under the fatigue protocol used in this study are shown in Figs. 3a-3c. Its crack pattern significantly resembles the clinical fracture of an e.max restoration that cracked intra-orally (Figs. 3d-3f).

DISCUSSION

Interpretation of use level probability plots (β < ~1) for MON and BTV indicates that, regardless of the step-stress level in which samples were tested, the failures were associated with stress level (load) rather than number of cycles (damage accumulation). Such failure behavior is similar to that observed for metal ceramic crowns (Silva *et al*., 2011) but differs from that of Y-TZP with either conventional or modified core/framework designs, where fatigue accelerates the failure (Silva *et al*., 2010). Based upon the above β interpretation, the specimens' load at failure (sample fracture) observed during fatigue testing was

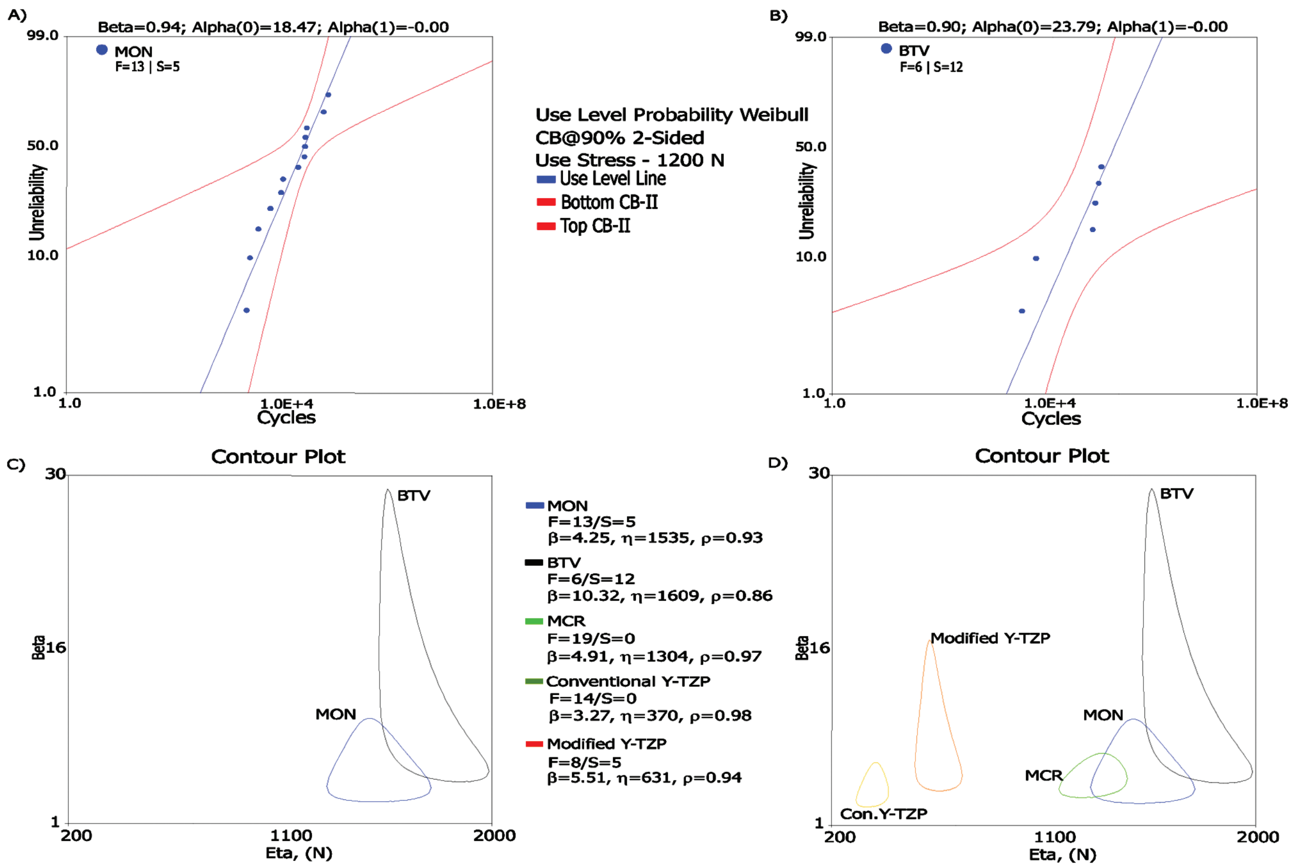

Figure 1. Use level probability Weibull plots of MON **(A)** and BTV **(B)** for a use load of 1200 N. Failed specimens are presented as blue points. Two-sided 90% confidence intervals are displayed as red lines. (A) Plot of MON (n = 18 specimens: 13 fractures, 5 suspensions), β = 0.94. (B) Plot of BTV (n = 18 specimens: 6 fractures, 12 suspensions), β = 0.90. Overlaps between confidence levels show no statistical significance. **(C,D)** Contour plots using 90% confidence bounds for the relationship between shape parameters (β) and characteristic failure load (η).

used to calculate the Weibull probability distribution and respective characteristic failure load values.

The high characteristic failure load values observed for MON and BTV seem related to the material's microstructure: Its processing results in a glass-ceramic with fine, elongated grains of approximately 1.5 µm in length and 0.4 µm in diameter of 70% crystal volume incorporated into a glass matrix (Fasbinder *et al.*, 2010). These elongated lithium disilicate crystals inhibit crack propagation, and cracking has been shown to occur only through the residual glass phase (30–40% of its volume; Apel *et al.*, 2008). This microstructure may explain the fractures occurring only at very high loads, and the resulting high reliability (Guess *et al.*, 2010; Heintze *et al.*, 2011), as well as the positive short-term clinical trials results for full-contour lithium disilicate crowns (Fasbinder *et al.*, 2010; Reich *et al.*, 2010).

The stiff contact generated by the WC ball may produce a more concentrated stress distribution potentially leading to different failure modes compared with an enamel and particularly with three-body contact. However, concentrated load testing as conducted here has been shown to produce clinically relevant fractures (Coelho *et al.*, 2009a,b; Silva *et al.*, 2010, 2011). At the point of loading, contact pressure increases monotonically with an expanding contact circle (Lawn, 1998; Lawn *et al.*, 2002),

Table. Weibull Moduli Beta (β) and Characteristic Strength Eta (η) Values for the Groups Tested, with 90% Confidence Intervals (Upper and Lower Limits)

Groups	β (upper–lower limits)	η (upper–lower limits)
Conventional Y-TZP	3.27 (2.27–4.72)[a]	370 N (322–427)[a]
Modified Y-TZP	5.51 (3.02–10.06)[a]	631 N (548–725) [a]
MCR	4.91 (3.69–6.53)[b]	1304 N (1203–1414)[b]
MON	4.25 (2.74–6.57)	1535 N (1354–1740)
BTV	10.32 (5.90–18.05)	1609 N (1512–1712)

Sources: [a]Silva *et al.*, 2010; [b]Silva *et al.*, 2011.

which is actually related to indenter radius. In Hertzian contact, the elastic modulus of the indenter enters the calculations only as it relates the contact area to the load applied (Lawn, 1998; Lawn *et al.*, 2002). For a sphere of radius *r* and normal load *P*,

$$a^3 = 4kPr/3E, \text{ where } k = (9/16)[(1-\nu^2) + (1-\nu'^2)E/E']$$

and where E is the elastic modulus, and the prime notation is for the indenter material. The mean contact pressure p_0 is:

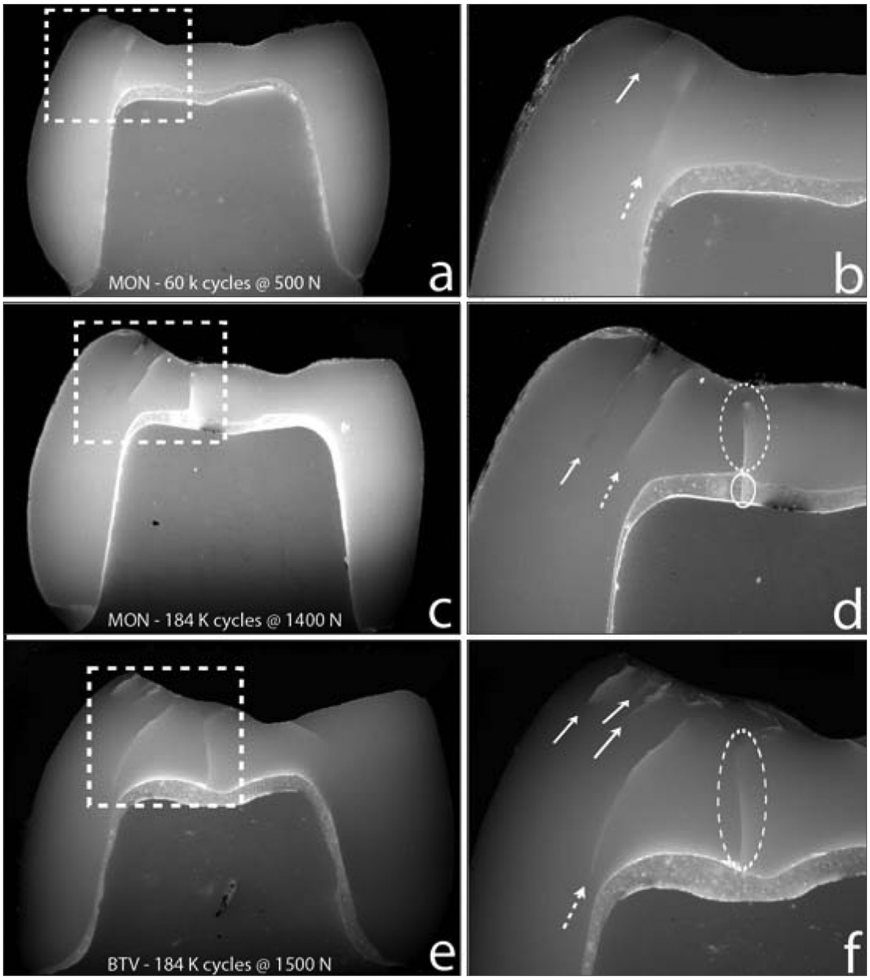

Figure 2. Cross-sections of polarized optical stereomicroscopy of MON (**a,b** and **c,d**) and BTV (**e,f**) of test suspended samples show competing failure modes after SCFT. Dashed boxes in a, c, and e define detail areas shown in b, d, and f, respectively. (a,b) Test suspended sample at 60 K cycles at 500-N load. Note cone (white solid arrow) and inner cone (white dashed arrow) cracks. No radial crack could be seen at this sectioned site. (c,d) Test suspended sample after 184 K cycles at 1,500-N load. The white dashed oval evidences radial cracks extending from the cement–ceramic interface upward to the indentation site. Note in (d) cement fracture (solid oval) continuous to the radial crack. (e,f) Deeper cone and inner cone cracks after 184 K cycles at 1,500-N load. Note radial crack starting from the cement–ceramic interface (white dashed oval).

$p_0 = P/\pi a^2$ or $p_0 = (3E/4\pi k)a/r$ combining the equations.

The maximum tensile stress in the specimen occurs at the contact circles and is:

$$\sigma_m = ½\,(1\text{-}2v)p_o.$$

The maximum tensile stress is outside the area of contact. Hence, indenter elastic modulus affects the contact area and thus reduces the mean contact pressure. The value of k would be 1.125 for a glass indenter (E = 90 GPa) and 0.169 for the WC indenter with an E = 600 GPa, given that the Poisson's ratio for both glass and WC is about 0.25. Bhowmick *et al*. (2007) investigated the role of indenter modulus on cone fracture evolution and compared, among others, a glass indenter with that of a WC indenter on glass substrate with uni-axial loading. They noted a 40% decrease in load to cone cracking with use of the carbide as opposed to the glass indenter. With mouth-motion testing in water, far fewer cycles are required to initiate partial cone cracking as compared with uni-axial loading (Kim *et al*., 2007). Glass on porcelain fatigue-testing leads to very high friction forces, promoting partial cone cracking. WC on porcelain, while having higher contact pressure, has a lower coefficient of friction, particularly after a few cycles. Currently, we are conducting studies on mouth-motion contact damage on porcelain comparing WC, glass, and Steatite ball indenters. We have routinely utilized WC as a standard, and the results reported here can be related to our previous findings (Coelho *et al*., 2009a,b; Silva *et al*., 2010, 2011) for loads required to cause fatigue damage.

It must be acknowledged that, by eliminating the low fracture toughness of veneering porcelain and the interface encountered in bilayered systems, different failure modes may occur (Rekow and Thompson, 2007; Rekow *et al*., 2011). As to the core materials evaluated in the present study, only veneer fracture is expected in both the metal ceramic and Y-TZP crown systems. Core failure is a rare event in Y-TZP crowns, where porcelain cohesive failures predominate, thus compelling researchers to elaborate on the topic pragmatically (Guess *et al*., 2011). One approach to limit Y-TZP crown veneer chipping is core/framework design modification to provide porcelain support (Bonfante *et al*., 2009). The resulting system characteristic failure load has proved higher than that of a conventionally designed core but significantly lower than that of the metal ceramic. This suggests that more effort is needed toward the comprehension and solution of Y-TZP porcelain chipping.

Concerning the failure modes of the full-contour lithium disilicate crowns, it appeared that radial cracks predominated over partial cone cracks, regardless of core thickness. Since high loads were required for final bulk fracture in our testing, the reliability results are reflected in the clinical observation of no failures (in crowns of unknown thicknesses) in the short term (Fasbinder *et al*., 2010; Reich *et al*., 2010). Our investigation of reduced thickness (1 mm), as compared with the previously reported 2 mm (Guess *et al*., 2011), and the latter's reduction to accommodate a 0.5-mm veneering porcelain (group BTV) were thus an attempt to understand further the system's mechanical performance under challenging structural and loading scenarios.

The mouth-motion step-stress fatigue used in the present study found failure modes similar to those observed clinically.

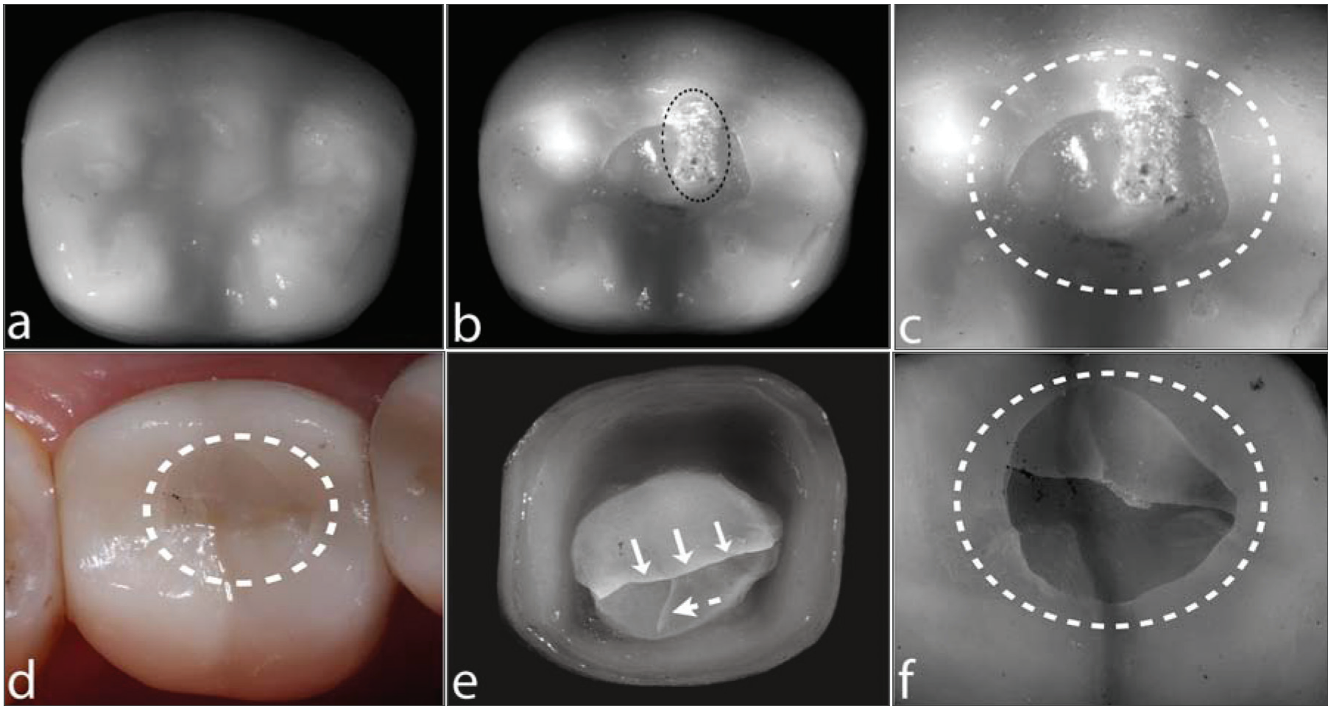

Figure 3. Laboratory (**a–c**) and clinical (**d–f**) fracture modes of e.max crowns. Note that cracks produced in the laboratory tend to form a ring around the indented area. From the intact crown (a) to SCFT crown (b), a trail of the sliding contact is formed (dashed black oval) followed by crack growth mesially and distally. As load increases or step-progresses, the crack propagates in a ring around the indentation area (c). The clinical picture (d) formed the same fracture pattern seen in (c). Polarized optical stereomicroscopy of the intaglio surface (e) of the e.max crown seen in (d) shows mesiodistal (solid white arrows) and buccolingual (dashed white arrow) radial cracks. The polarized image (f) of the occlusal aspect of the crown seen in (d) shows a crack pattern similar to that of the test suspended specimen (c) after SCFT (69 K cycles at 1,400-N load).

This suggests the effectiveness of the method used for the direct comparison of crown systems. Clinical results for lithium disilicate crowns are promising, but longer observation periods are required. Comparison between monolithic and multilayer systems must be approached with caution.

The present investigation hypothesized that the reliability of reduced-thickness monolithic lithium disilicate crowns is higher when compared with that of zirconia (Y-TZP) and at least comparable with those of metal ceramic systems. Similar use level probability Weibull and characteristic failure load calculations were observed between MON and BTV. These monolithic lithium disilicate configurations presented higher values than Y-TZP crowns. Relative to MCR crowns, BTV presented significantly higher failure loads, whereas the thinner MON presented intermediate values, thus validating the study hypothesis.

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