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Zirconia-reinforced lithium silicate crowns: Effect of thickness on survival and failure mode

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ABSTRACT

Objective. To evaluate the reliability and failure mode of zirconia-reinforced lithium silicate (ZLS) molar crowns of different thicknesses.
Methods. Monolithic ZLS molar crowns (0.5 mm, 1.0 mm, and 1.5 mm thickness) were modeled and milled using a CAD/CAM system (n = 21/group). Crowns were cemented on dentin-like epoxy resin replicas with a resin cement. The specimens were subjected to single load-to-failure test for step-stress profiles designing. Mouth-motion step-stress accelerated-life test was performed under water by sliding an indenter 0.7 mm lingually down on the distobuccal cusp until specimen fracture or suspension. Use level probability Weibull curves and reliability were calculated and plotted. Polarized-light optical microscope and scanning electron microscope (SEM) were used to characterize fracture patterns.
Results. Irrespective of crown thickness, beta (β) values were higher than 1 and fatigue accelerated failures. While 0.5 mm ZLS crowns exhibited a significant reduction in the probability of survival at 200 N, 300 N and 400 N mission loads (69%, 41% and 19%, respectively), no significant difference was observed between 1.0 mm and 1.5 mm crowns. Both thicknesses have maintained the survivability at approximately 90%. Failure primarily comprised bulk fracture where radial cracks originated from the cementation surface beneath the indenter loading trail and propagated towards the cervical margin.
Significance. 1.5 mm- and 1.0 mm-thickness monolithic ZLS crowns presented higher probability of survival compared to 0.5 mm crowns. Bulk fracture was the chief failure mode, regardless of thickness.

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1. Introduction

Over the last two decades, a shift towards the indication of all-ceramic restorations has been sought in dentistry to meet higher esthetic demands [1]. Compared to other all-ceramic systems, high-strength polycrystalline ceramics, such as zirconia-based systems, have gained attention for use as a framework material, especially for high-load requirement reconstructions [2]. Nonetheless, long-term clinical studies have reported an expressive range in the survival rates (98% to 67%, after 5 and 10 years, respectively) with veneering ceramic chipping and delamination from the inner core encompassing the main complications associated with high-strength infrastructures (11%, after, on average, 2.2 years in function) [3,4]. Hence, literature findings have suggested that despite pragmatic investigation of the reasons and alternatives to overcome chipping, it still represents the mostly reported technical complication [1–4].

Accordingly, the fabrication of monolithic translucent restorations, where the lower fracture toughness porcelain veneer is eliminated, seems to be a watershed event in the restorative field to enhance survivability [5]. Full-contour restorations are increasingly favored over bilayered ones since they minimize problems associated with complex fabrication process, as well as, greater structural integrity can be accomplished by eliminating the veneering ceramic and its required bonded interface [1–10]. All the aforementioned facts associated with the effectiveness of adhesive cementation in improving the critical resistance of all-ceramic rehabilitations have also supported the indication of thinner restorations, which meet the criteria of minimally invasive dentistry and preservation of healthy tooth structure [10–17].

Reinforced glass-ceramic systems have been increasingly indicated for monolithic rehabilitations due to continuous enhancements of their mechanical properties associated with their inherent outstanding optical properties. Undoubtedly, the use of monolithic glass-ceramics became remarkable after the development of lithium disilicate system (LD) [5], which has demonstrated almost 4-fold higher flexural strength relative to traditional glass-ceramics [18,19]. Clinical data of monolithic LD crowns performance indicate promising survival rates after short and long-term evaluations (~96% and 83%, after 5 and 10 years) [7,20–22]. Although favorable results, recent investigations have shown that bulk fracture is still the most common reason for rehabilitation failure (2.3–19%, after ~5 years), mainly in areas subjected to greater masticatory load [7,21]. Despite the broad use of LD systems, materials scientists have attempted to develop a more reliable glass-ceramic with further increased strength by microstructure reinforcement.

Zirconia-reinforced lithium silicate (ZLS) system features a complex microstructure, which is composed of lithium metasilicate and lithium orthophosphate crystals embedded in a glassy matrix containing zirconia [23–25]. The very fine crystal size exhibited in the ZLS system (0.5–1.0 μm), which is approximately 4–6 times smaller than LD crystals (2.0–3.0 μm), provides a higher percentage of glass content (roughly 50%) than LD (30%) [23,26]. Such microstructure supports outstanding optical properties to ZLS blocks that have previously demonstrated similar and higher mean translucency values compared to feldspathic and LD blocks, respectively [19]. Despite the higher glass content, the homogeneous dispersion of zirconia into the glassy matrix is claimed to improve the mechanical properties of the ceramic structure by toughening mechanisms [26]. Indeed, ZLS ceramic discs have shown a significantly higher flexural strength relative to LD systems (approximately 100 MPa higher) [19,27,28]. Additionally, ZLS fracture toughness, which describes the material resistance to crack propagation and may be related to longevity in service, has exhibited higher values (2.31–2.65 $MPam^{1/2}$) when compared to LD (1.88–2.01 $MPam^{1/2}$) [26,27]. Preliminary data of fracture resistance behavior obtained from single load-to-fracture (SLF) test have also indicated that monolithic ZLS crowns outperformed LD and conventional zirconia crowns (1742 N, 1565 N and 1267 N, respectively) [29].

Nonetheless, it has been well-defined that a complex environmentally-assisted mechanical scenario other than overload catastrophic failure plays a synchronized role in ceramic damage initiation, accumulation, and failure [30–35]. The cumulative damage triggered by cyclic forces associated with water molecules presence at the crack tip results in the chemically-assisted slow crack growth leading to ceramic failure at relatively lower stress levels compared to its ultimate characteristic stress, mainly in reduced-thickness rehabilitations [16,17,30–35]. Hence, literature data regarding ZLS system lacks laboratory studies that closely reproduce in-service conditions, which can be obtained using mouth-motion step-stress accelerated-life (SSALT) test providing a reliable prediction of its clinical behavior [9,10,36–44]. Accordingly, the current study sought to evaluate the reliability and failure mode of ZLS monolithic molar crowns of different thickness using SSALT test. The postulated null hypotheses were that: (i) survival predictions would not be affected by thickness in monolithic ZLS molar crowns; and (ii) failure mode would not vary in different thicknesses of monolithic ZLS molar crowns.

2. Materials and methods

2.1. *Specimen fabrication*

A CAD-based three-dimensional (3D) model of a mandibular first molar was generated (Ceramill Mind CAD software, Amann Girrbach, Koblach, Austria). To simulate a preparation for monolithic all-ceramic restorations, the proximal and occlusal walls were uniformly reduced to: (i) 1.5 mm, (ii) 1.0 mm and (iii) 0.5 mm thickness. The CAD-modeled preparation replicas (n = 21/group) were milled using dentin-like fiber-reinforced epoxy resin cylinders (Accurate Plastics Inc., New York, NY, USA) and stored in deionized water at 37 °C for 21 days to provide hygroscopic expansion and minimize dimensional alteration prior to cementation, as detailed elsewhere [11,45,46]. Such a material was selected as the substrate due to its similarity in terms of adhesion (9 MPa) and elastic modulus (18.6 GPa) to hydrated dentin (6.5 MPa and 18 GPa, respectively) [45].

Based on the 3D model of the mandibular first molar previously designed, monolithic zirconia-reinforced lithium

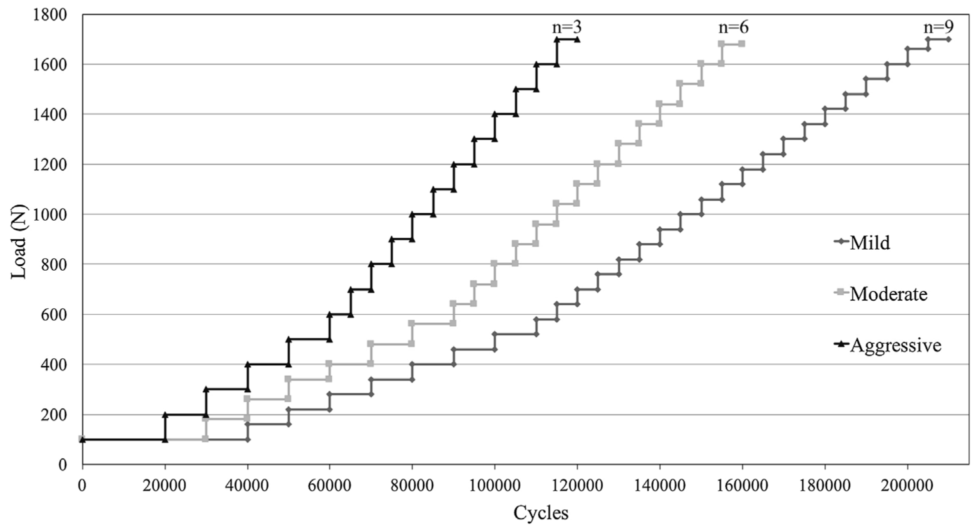

Fig. 1 – Step-stress accelerated-life testing profiles designed for the fatigue sequence, in which a mild, a moderate, and an aggressive loading approach was determined as a function of number of cycles. Out of eighteen specimens in each group, 9 were assigned to mild, 6 to moderate, and 3 to the aggressive fatigue profile.

silicate (ZLS) (Vita Suprinity, Vita Zahnfabrik, Bad Säckingen, Germany) crowns with different thicknesses (0.5 mm, 1.0 mm, and 1.5 mm) were modeled and milled using computer-aided design/computer-aided manufacturing (CAD/CAM) blocks through Ceramill Motion 2 system (Amann Girrbach, Koblach, Austria) (n = 21/group). After milling procedure, crowns were subjected to crystallization firing (840 °C, 8 min), polishing (Vita Suprinity Polishing Set, Vita Zahnfabrik, Bad Säckingen, Germany), and glazing firing (800 °C, 1 min) (Vita Akzent Plus, Vita Zahnfabrik, Bad Säckingen, Germany) following the manufacturer guidelines.

ZLS crowns were cemented on their epoxy resin counterparts using a dual-cure resin cement (RelyX^{TM} Ultimate, 3 M Oral Care, St. Paul, MN, USA). Prior to cementation, the replicas were acid etched with 10% hydrofluoridric acid (Vita Ceramics Etch, Vita Zahnfabrik, Bad Säckingen, Germany) for 120 s [46]. The intaglio surface of the ceramic crowns was also acid etched with 10% hydrofluoridric acid (Vita Ceramics Etch, Vita Zahnfabrik, Bad Säckingen, Germany) for 20 s [47]. After etching, all samples were washed and rinsed, gently air dried and cleaned with distilled water in an ultrasonic bath (2 min) to remove any residual acid. For adhesive bonding, a thin layer of adhesive (Scotchbond^{TM} Universal, 3 M Oral Care, St. Paul, MN, USA) was applied with a microbrush on both ceramic and epoxy resin substrate for 20 s following 5 s of gentle air-drying. The resin cement was directly applied on the internal surface of the crown using the cement automix dispenser. After crown-to-replica setting, the assembly was maintained under a load of 10 N to promote uniform cement spreading and the excess was removed. The crown margin was light-cured for 20 s on each surface. Thereafter, pairs of crowns and replicas were vertically positioned into a polyvinyl siloxane matrix (Express, 3M Oral Care, St. Paul, MN, USA) to standardize the embedding and pouring of acrylic resin (Dencôr, Clássico, São Paulo, Brazil) into a 25 mm-diameter polyvinyl chloride tube (PVC). Prior to mechanical testing, the samples were stored in deionized water at 37 °C for 5 days to provide complete polymerization and hydration of the cement layer [46].

2.2. *Mechanical testing*

In accelerated life testing using time-varying stress application, such as step-stress accelerated-life testing (SSALT), the load is successively increased, step-by-step, to levels that exceed normal operating conditions in order to accelerate failures without affecting predictions accuracy [36,48,49]. Therefore, to determine the step-stress profiles, three crowns per group were subjected to the single load-to-failure (SLF) testing. The specimens were mounted in a universal testing machine (Model 5566, Instron, Norwood, MA, USA) and a load was applied axially through a tungsten carbide indenter (r = 3.18 mm) on the central fossa of the occlusal surface using a 10 kN load cell at a loading- rate of 1 mm/min [9,10,36–44].

Based on the SLF data, three different step-stress profiles, mild, moderate and aggressive, were designed (Fig. 1). The profiles were named based on the load increase rapidness in which a tested sample would reach a certain load level, where ZLS crowns allocated into the mild profile took longer number of cycles to reach the same load level than crowns allocated into the aggressive profile. Besides, the use of 3 SSALT profiles was based on the need of failure distribution along different step loads, which allows for a better statistic prediction narrowing confidence intervals [36]. Usually, a designed profile started at a load that was approximately 30% of the mean value obtained from SLF, and end at a load of approximately 60% of the same value [9,10,36–44]. Out of eighteen crowns per group, 9 were assigned to mild, 6 to moderate, and 3 to the aggressive fatigue profile following the distribution ratio of 3:2:1 to mild, moderate and aggressive, respectively [9,10,36–40,43,44]. The rationale behind this distribution ratio is that the accuracy of the statistic prediction is inversely proportional to its cycling length; therefore, 9 samples are favored in the mild relative to 3 in the aggressive [36].

Mouth motion step-stress fatigue testing was performed by sliding a tungsten carbide indenter (r = 3.18 mm) 0.7 mm lingually down the distobuccal cusp towards the central fossa, beginning 0.5 mm lingual to the cusp tip, simulating aspects of natural occlusion using an electrodynamic fatigue testing machine (EL-3300 Enduratec, Bose, Minnetonka, MN, USA) [9,10,36–44]. Crowns were immersed in distilled water during fatigue testing. Because ceramic materials are sensitive to load/stress rate, a clinically relevant load rate (1000 N/s) was set [30,42]. As a result, the loading frequency, including load application, slide and lift-off phases, varied from 0.3 Hz at 1700 N to 3 Hz at 100 N [42]. At the end of each step-stress cycle, all specimens were inspected under polarized-light optical microscope (Axio Zoom v.16, Zeiss, Oberkochen, Germany) for cracks and ceramic surface damage evaluation. The test was conducted until crown failure (considered as chip-off or bulk fracture) or survival until a maximum load of 1700 N [41,42].

Data analysis consisted of an underlying life distribution that describes the life data collected at different stress levels and a life-stress model that quantifies the manner in which the life distribution changes across different stress levels [36,48]. Thus, the Weibull Distribution was chosen to fit the life data collected in SSALT and its probability density functions (pdfs) is given by: $\int(T) = \frac{\beta}{\eta}\left(\frac{T}{\eta}\right)^{\beta-1} e^{\left(\frac{T}{\eta}\right)^{\beta}}$, where η = scale parameter, and β = shape parameter. Considering the time-varying stress model (x(t)) of SSALT, a life-stress relationship was selected to quantify the path from the overstress pdfs and extrapolate a use level pdf considering the cumulative effect of the applied stresses (x(t)), and it is commonly referred as the cumulative damage model. In such a model, the inverse power law (IPL) relationship would be given by $L\,(x(t)) = (\alpha/x(t))^{\eta}$, where L = life data, and x(t) = stress. Then, the IPL-Weibull pdf (where η is replaced by the IPL) was given by: $\int(t, x(t)) = \beta\left(\frac{x(t)}{\alpha}\right)^{n}\left(\int_0^t\left(\frac{x(t)}{\alpha}\right)^{n} du\right)^{\beta-1} e^{-\left(\left(\frac{x(t)}{\alpha}\right)^{n} du\right)^{\beta}}$. From the extrapolated use level pdf, a variety of functions could be derived, including reliability $R(t,\,x\,(t)) = e^{-\left(\left(\frac{x(t)}{\alpha}\right)^{n} du\right)^{\beta}}$. Moreover, parameters estimation was accomplished via maximum likelihood estimation (MLE) methods, and confidence intervals were approximated using the Fisher matrix approach [36,48,50]. Hence, the use level probability Weibull curves (probability (%) of failure versus number of cycles) with a set load of 300 N at 90% two-sided confidence interval were calculated and plotted (Synthesis 9, Alta Pro, Reliasoft, Tucson, AZ, USA). Also, the reliability was calculated for completion of a mission of 100,000 cycles at 100, 200, 300, 400, 600 and 800 N and the differences between groups were identified based on the non-overlap of confidence interval (90%)(Synthesis 9, Alta Pro, Reliasoft).

2.3. *Microscope imaging analysis*

Failed specimens were first inspected in a polarized-light optical microscope (Axio Zoom v.16, Zeiss, Oberkochen, Germany) and then representative samples were sputter coated with gold alloy (Emitech K650, Emitech Products Inc., Houston, TX, USA) for scanning electron microscopy (SEM) analysis (Hitachi S-3500, Hitachi Instruments, Tokyo, Japan) and fractographic evaluation.

3. Results

The mean load-to-fracture values used to design step-stress profiles (±standard deviation) for 1.5 mm, 1.0 mm, and 0.5 mm ZLS crowns were 2109 ± 489 N, 1276 ± 414 N and 718 ± 202 N, respectively. All crowns failed by bulk fractures.

Use level Weibull probability plot at a use load of 300 N for 1.5 mm, 1.0 mm, and 0.5 mm ZLS crowns (90% confidence bounds) are shown in Fig. 2. The beta (β) values derived from use level probability Weibull calculation were 1.58 (0.66–3.75), 1.17 (0.57–2.40) and 3.69 (2.27–5.99) for 1.5 mm, 1.0 mm, and 0.5 mm crowns, respectively (Table 1). Such values indicate that, irrespective of thickness, failure rate increased over time and fatigue damage accumulation was an acceleration factor for failure.

Predicted reliability for completion of a mission of 100,000 cycles at a range of load magnitudes is presented in Table 1. All ZLS crowns thicknesses demonstrated high probability of survival at 100 N (up to 92%). While 1.5 mm and 1.0 mm ZLS crowns kept their survivability higher than 90% at a set load of 200 N, a significantly reduced reliability was demonstrated by 0.5 mm crowns (6̃9%). Similarly, the cumulative damage reaching high-load magnitude missions, either for 300 N or 400 N, would result in significantly lower probability of survival for 0.5 mm crowns (41% and 19%, respectively) relative to 1.0 mm (95% and 86%) and 1.5 mm crowns (99% and 94%). Indeed, no significant difference was observed between 1.0 mm and 1.5 mm ZLS crowns for all calculated missions corresponding to physiological masticatory molar forces (200–400 N), and both thicknesses maintained the probability of survival at approximately 90%. Only at a set mission of 800 N 1.0 mm crowns (1̃5%) showed significantly reduced reliability relative to 1.5 mm crowns (8̃0%).

The chief failure mode for ZLS monolithic crowns comprised bulk fracture (Figs. 3 and 4). Careful examination of the ZLS crowns from the occlusal aspect revealed the quasiplastic deformation at the loading area represented by sliding-contact induced cracks on the distobuccal cusp (Fig. 4a). During fatigue testing of the ZLS crowns, cracks usually initiated at the area directly below the indentation trail and/or at the cementation interface and propagated as the load and number of cycles increased (Figs. 3 and 4b). Fractographic analysis of failed ZLS crowns has shown that radial cracks dominated and the fracture originated predominantly at the cementation interface (Figs. 3 and 4b). In addition, telltale fractographic marks including hackles and arrest lines that suggest the direction of crack propagation (DCP) were identified (Figs. 3 and 4b–f). The fracture originated beneath the indenter sliding path on the cementation surface with crack front propagating upward and sideways through the proximal and cervical margins of the sample. Such assumption was confirmed by the presence of multiple arrest lines with their concave part pointing towards the occlusal indentation area (Fig. 4). Also, hackles lines propagating towards the cervical margins corroborates with the suggested DCP (Fig. 4). In addition, two 1.5 mm crowns survived the set profile and one showed chip-off fracture.

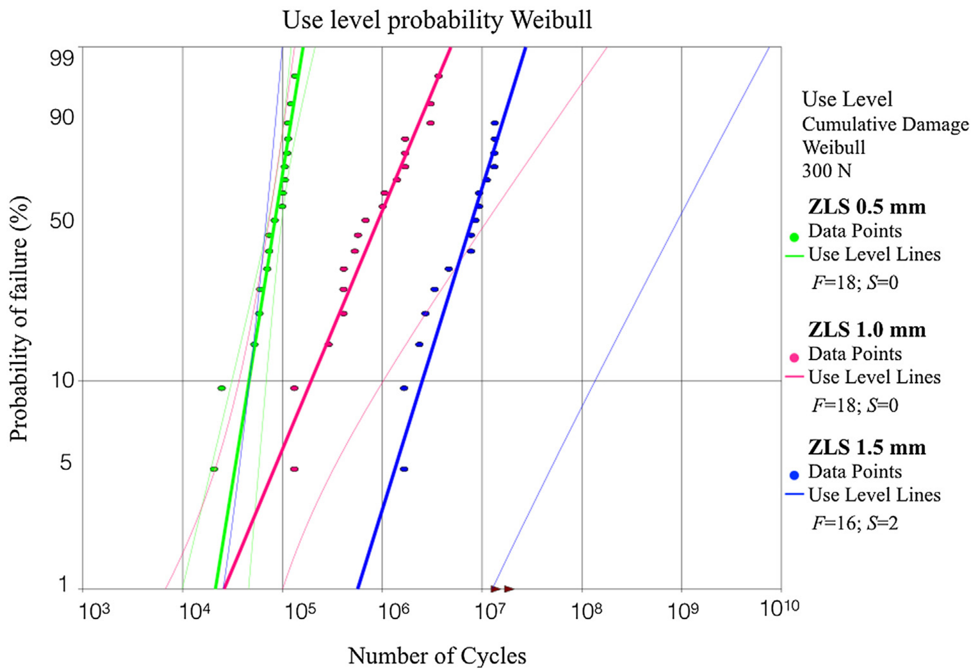

Fig. 2 – Use level probability Weibull curves (90% two-sided confidence interval) evidencing the probability of failure as a function of elapsed cycles for different-thickness ZLS crowns at 300 N of set load (the two crowns of 1.5 mm thickness that did not fail were considered suspended in the statistical analysis).

Table 1 – Calculated reliability (%) for a given mission of 100,000 cycles at set loads of 100, 200, 300, 400, 600 and 800 N.

	1.5 mm			1.0 mm			0.5 mm		
	Lower bound	Reliability	Upper bound	Lower bound	Reliability	Upper bound	Lower bound	Reliability	Upper bound
100 N	99.64	**100 aA**	100	98.32	**99.92 aA**	100	66.87	**91.94 bA**	98.26
200 N	97.87	**99.97 aAB**	100	91.14	**98.94 aAB**	99.88	45.65	**68.99 bAB**	83.88
300 N	93.99	**99.80 aABC**	99.99	77.52	**95.25 aAB**	99.07	25.23	**41.27 bBC**	56.62
400 N	87.65	**99.21 aABC**	99.95	58.91	**86.72 aBC**	96.23	0.06	**19.41 bCD**	37.73
600 N	67.71	**94.44 aBC**	99.17	20.92	**52.29 abCD**	76.43	0	**0.02 bD**	21.04
800 N	41.75	**79.31 aC**	94.03	0.02	**14.96 bD**	37.96	0	**0.07 bD**	12.81
Beta (ß)	0.66	**1.58**	3.75	0.57	**1.17**	2.40	2.27	**3.69**	5.99

Different lowercase letters mean statistical difference between crown thicknesses. Different uppercase letters mean statistical difference between missions. Differences between groups were identified based on the non-overlap of 90% two-sided confidence interval.

This fracture had a typical shell-shaped morphology with a maximum length extending all the way to the cervical margin. Different from samples that presented bulk fracture, the inspection from the buccal aspect of the crown indicated that the cohesive fracture originated immediately bellow the indentation trail. The presence of several arrest lines and hackles lines indicated the DCP from the occlusal contact area to the proximal and cervical margins.

4. Discussion

Recently, zirconia-reinforced lithium silicate (ZLS) glass-ceramic blocks for CAD/CAM use have been launched aiming to improve optical and mechanical properties relative to previous glass-ceramic systems [19,23–27]. Preliminary data of in vitro mechanical tests and the absence of long-term clinical trials have encouraged the current study to mimic clinical failures through mouth-motion fatigue and to estimate the lifetime of ZLS monolithic molar restorations of 1.5 mm, 1.0 mm and 0.5 mm thicknesses [11,16,19,26,27,29,39].

The interpretation of the use level probability plots indicates that, regardless of ZLS crown thickness, the beta (β) values were higher than 1 and fatigue accelerated failures [36,48,50]. Hence, the cumulative damage triggered by cyclic forces associated with stress corrosion resulting in the chemically-assisted slow crack growth was an acceleration factor for ZLS crowns failure [30–35].

The projected reliability for completion of a mission of 100,000 cycles at clinically relevant molar masticatory loads (200–400 N) demonstrated a very limited survival prediction for 0.5 mm crowns relative to 1.0 mm and 1.5 mm crowns [51,52]. While 0.5 mm ZLS crowns exhibited a significant reduction in the reliability, 69%, 41% and 19% at 200 N, 300 N and 400 N, respectively, no significant difference was observed between 1.0 mm and 1.5 mm crowns, with both thicknesses

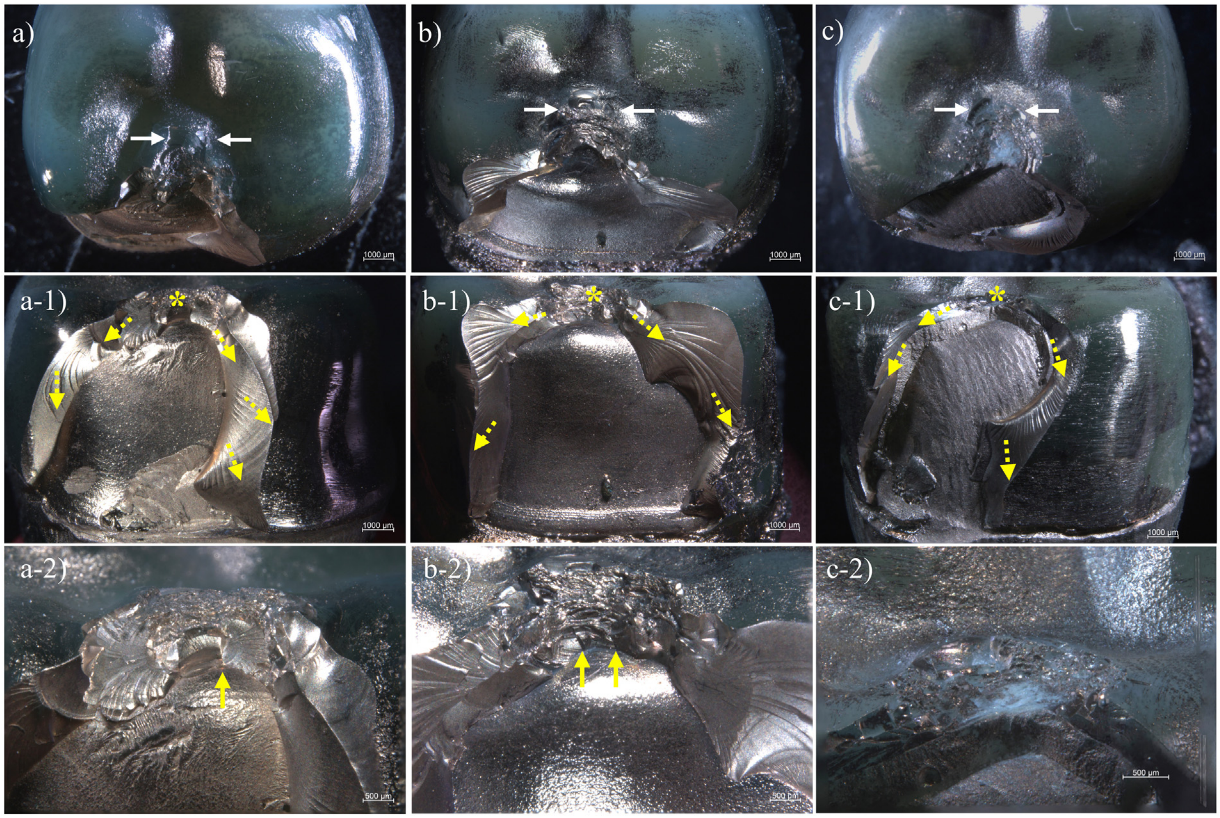

Fig. 3 – Irrespective of thickness, the chief failure mode in monolithic ZLS crowns subject to sliding-contact mouth-motion step-stress accelerated-life test comprised bulk fracture. Micrographs of representative fractured crowns of 1.5 mm- (a), 1.0mm- (b), and 0.5 mm-thickness (c) show the quasiplastic deformation on the loading area (white arrow). Also, the presence of fractographic marks moving from the oclusal surface of the crown (*) towards the cervical margin indicate the direction of crack propagation (yellow arrows). Higher magnification images of the buccal surface show the formation of radial cracks (yellow arrows), which indicates the suggested origin of fracture (cementation interface). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

maintaining the probability of survival at approximately 90%. It has been determined that the load necessary to lead to bulk fracture in ceramic restorations is proportional to the square of the ceramic thickness [53,54]. The present data endorsed this statement as decreasing the thickness of a monolithic ZLS crown to 0.5 mm had a significant impact on increasing its failure risk and affecting its predicted survival [11–14,16]. Such decrease in mechanical performance has been validated using finite element analysis that demonstrated greater stress concentration at both occlusal and cementation surfaces for ultra-thin ZLS molar restorations (≤0.5 mm) [11,16]. Thus, the first postulated null hypothesis that the probability of survival would not be affected in reduced-thickness monolithic ZLS molar crowns was rejected since 1.5 mm and 1.0 mm ZLS crowns outperformed 0.5 mm crowns.

Substantial data are available regarding the estimated lifetime of all-ceramic molar crowns. In comparison with high-strength polycrystalline systems, monolithic ZLS restorations (99%, after 100,000 cycles at 200 N), even as thin as 1.0 mm (98%, after 100,000 cycles at 200 N), have exhibited a significantly higher probability of survival relative to veneered zirconia restorations, regardless of framework design and veneering material (48%–99%, after 50,000 cycles, and ~23% after 100,000 cycles, at 200 N) [10,37–40,43,44]. Recently, much effort has been taken to develop highly-translucent zirconia systems for use as monolithic restorations and studies are warranted to estimate their mouth-motion SSALT fatigue lifetime for different-thickness restorations [46].

One potential cause of the decreased mechanical performance of ultra-thin (≤0.5 mm) glass-ceramic restorations has also been considered to lie on the effect of acid etching [55]. A recent investigation has shown that acid etching may produce ceramic microstructural alterations not only superficially but also internally and the depth effect is directly related to acid type, concentration, application time and ceramic glass content [55]. Conventional hydrofluoridric acid (HF) at a concentration of 10% applied for 20 s, as used in the current study [47], has demonstrated the potential of dissolving approximately 0.3 mm of the glassy phase in leucite and LD systems. Thus, treating glass-ceramic materials, mainly in reduced-thickness restorations, using strong etching protocols may increase defect population and, consequently, the risk of crack propagation under tensile stress [55]. Future investigations must confirm the extent of these findings for ZLS system and, also, evaluate the potential benefits of using one-step ceramic self-etching primers. Such simplified products present softer

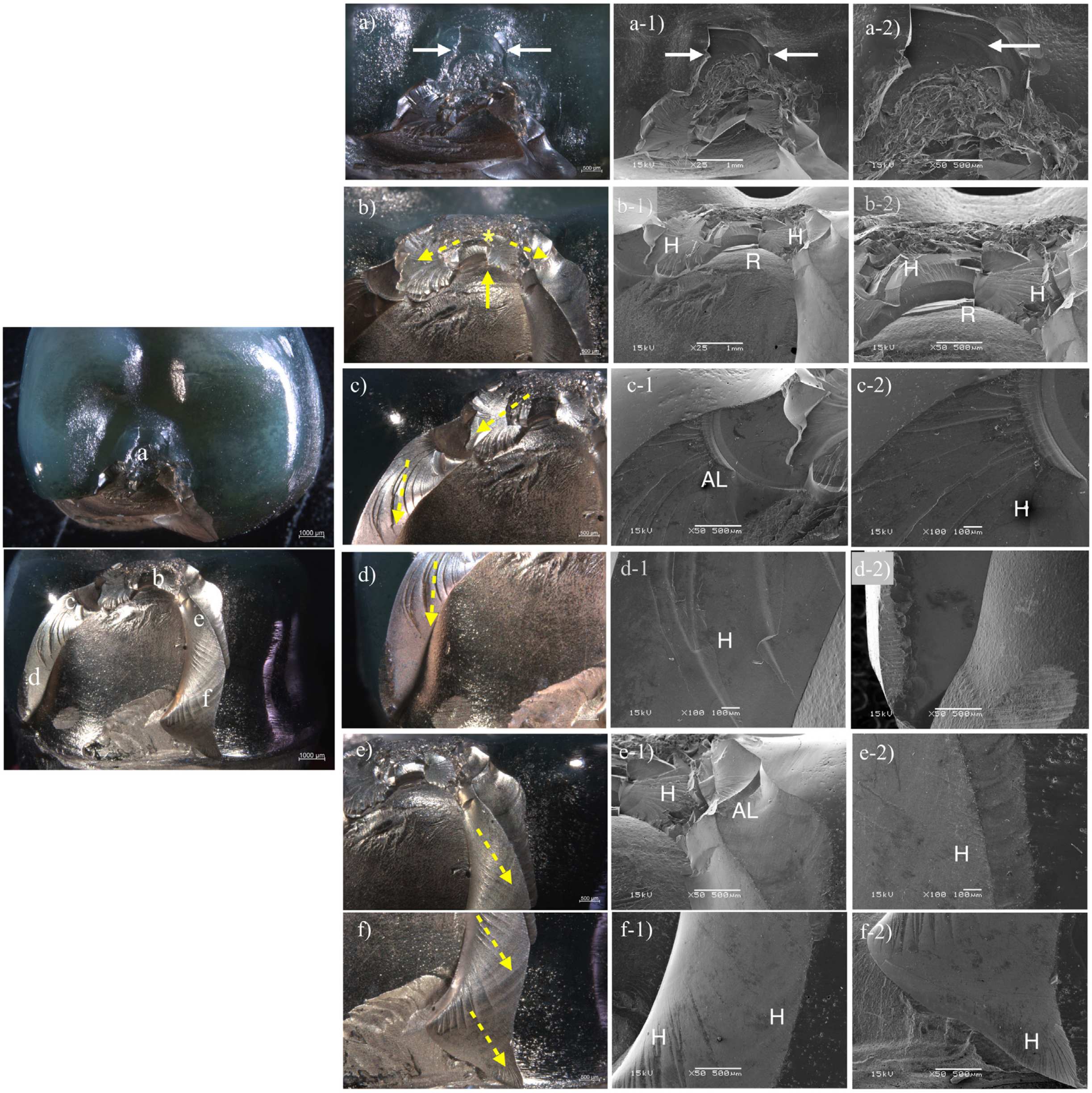

Fig. 4 – Occlusal view images of a representative fractured crown depict the quasiplastic deformation on the loading area (white arrows) (a). Further magnifications focus on the sliding-contact induced cracks on the distobuccal cusp (white arrows). Buccal view micrographs show the fracture originated (*) on the cementation surface with a predominant radial crack (R) front propagating upward and sideways all the way through the proximal and cervical margins of the crown (b–f). This statement is supported by the presence of multiple hackle lines (H) and arrest lines (AL), especially demonstrated in the SEM images, which indicates the direction of crack propagation (yellow arrows) (c–f). Arrest lines are related to direction of propagation as the beginning of a crack event is always located on the concave side of the first arrest line. Hackles run in the same direction of cracking and they separate parallel portions of the propagating crack that are on slightly different planes. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

compounds that have produced smoother etching patterns and similar bond strength when compared to conventional surface treatment (HF and silane, separately) [56,57].

With respect to failure, it is very important to understand the interplay between the competing fracture modes in order to reveal the weakest link of a determined system and define a practical procedure to improve it and/or develop a more fatigue-resistant restorative system [34]. Occlusal-like loading in a monolithic ceramic causes contact damage on the ceramic surface forming incomplete ring cracks that can extend deep into the material by the environmentally-assisted slow crack growth. Thus, cone and median cracks can be generated mechanically under the loading area and extend by hydraulic pumping of fluid into the microcracks [30–34]. The addition of the tangential component in sliding-contact loading skews the tensile stress field with concomitant development of cone cracks at the trailing edge contact [31,34,35]. Apart from occlusal cracks, radial cracks can form at the cementation interface. The substrate beneath the loading area can deform triggering tensile stresses at the intaglio surface [8,10,11,30,31,33–35]. Irrespective of crown thickness, fractographic analysis of failed ZLS crowns has proven that radial cracks dominated and the fracture originated at the cementation interface and propagated towards the proximal and cervical margins of the sample. Multiple hackles lines and arrest lines were also identified and supported the designated direction of crack propagation (DCP). Such failure mode corroborates with previous literature findings of different-thickness monolithic glass ceramic crowns subjected to SSALT test [9–11,42]. Thus, the second postulated hypothesis that failure mode would not vary in different thicknesses of monolithic ZLS molar crowns was accepted since bulk fracture comprised the primary failure mode. Previous finite element analysis evaluating the stress distribution in a monolithic ZLS crowns has demonstrated a tensile zone on the cementation interface that validates the suggested crack origin in the current study [11,16]. Also, increased failure risks in critical thickness restorations (<1 mm) relies on the enhanced flexural stress on the intaglio cementation surface and radial cracks prevalence [10,11]. Therefore, designing of monolithic ZLS ceramic crowns involves thicker ceramic layer (≥1 mm) to prevent early bulk fracture from cementation surface radial cracks [11,30].

Moreover, SSALT test was performed using anatomically-shaped monolithic molar crowns and followed typical chewing parameters to reproduce in-service conditions, in which the indenter applied the load on the cusp incline, slided, and lifted-off to its initial position to start a new cycle [30,42]. Despite the laboratory-obtained failure mode (chiefly bulk fracture) being similar to previously reported clinical failures of bilayered all-ceramic reconstructions, there is currently limited evidence concerning the clinical failure modes of ZLS crowns, considering that the material is somewhat new on the market [7]. Hence, if the predominant failure mode observed in the current study involving catastrophic failure is confirmed by clinical studies, the main disadvantage of this material would be the loss of repair potential and the mandatory need of crown replacement, which adds time and costs to both patients and professionals. In contrast, if chipping had been the main failure mode, repair would have been possible either by repolishing or by recontouring with composite resin. In fact, extensive fractographic work has been performed on all-ceramic restorations, primarily on bilayered reconstructions, and the results highlight that whereas a failure mode may predominate in some materials, failure origin does vary and originate not only from the occlusal surface and cementation interface, but also from the cervical margin [7,20,21,58–60]. Such findings emphasize the importance of comparing clinical- versus laboratory-obtained failure modes validating the results of life data analysis [58,59].

5. Conclusion

Based on the current results, the indication of ZLS system for tooth-supported monolithic molar crowns seems very promising, however, caution is advised when intending to use ultra-thin restorations (≤0.5 mm thickness). While 1.0 mm and 1.5 mm crowns exhibited high reliability at clinically relevant molar loads, a significantly reduced survivability was demonstrated for 0.5 mm crowns. Failure primarily comprised bulk fracture, in which the critical radial crack originated from the cementation surface beneath the loading trail and propagated towards the cervical margin.

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REFERENCES

[1] Guess PC, Schultheis S, Bonfante EA, Coelho PG, Ferencz JL, Silva NR. All-ceramic systems: laboratory and clinical performance. Dent Clin North Am 2011;55:333–52, ix.
[2] Pjetursson BE, Sailer I, Makarov NA, Zwahlen M, Thoma DS. All-ceramic or metal-ceramic tooth-supported fixed dental prostheses (FDPs)? A systematic review of the survival and complication rates. Part II: multiple-unit FDPs. Dent Mater 2015;31:624–39.
[3] Miura S, Kasahara S, Yamauchi S, Okuyama Y, Izumida A, Aida J, et al. Clinical evaluation of zirconia-based all-ceramic single crowns: an up to 12-year retrospective cohort study. Clin Oral Investig 2018;22:697–706.
[4] Schwarz S, Schroder C, Hassel A, Bomicke W, Rammelsberg P. Survival and chipping of zirconia-based and metal-ceramic implant-supported single crowns. Clin Implant Dent Relat Res 2012;14(Suppl. 1):e119–25.
[5] Silva LHD, Lima E, Miranda RBP, Favero SS, Lohbauer U, Cesar PF. Dental ceramics: a review of new materials and processing methods. Braz Oral Res 2017;31:e58.
[6] Baldassarri M, Stappert CF, Wolff MS, Thompson VP, Zhang Y. Residual stresses in porcelain-veneered zirconia prostheses. Dent Mater 2012;28:873–9.
[7] Sailer I, Makarov NA, Thoma DS, Zwahlen M, Pjetursson BE. All-ceramic or metal-ceramic tooth-supported fixed dental prostheses (FDPs)? A systematic review of the survival and complication rates. Part I: single crowns (SCs). Dent Mater 2015;31:603–23.
[8] Zhang Y, Mai Z, Barani A, Bush M, Lawn B. Fracture-resistant monolithic dental crowns. Dent Mater 2016;32:442–9.
[9] Guess PC, Zavanelli RA, Silva NR, Bonfante EA, Coelho PG, Thompson VP. Monolithic CAD/CAM lithium disilicate versus veneered Y-TZP crowns: comparison of failure modes and reliability after fatigue. Int J Prosthodont 2010;23:434–42.
[10] Silva NR, Bonfante EA, Martins LM, Valverde GB, Thompson VP, Ferencz JL, et al. Reliability of reduced-thickness and thinly veneered lithium disilicate crowns. J Dent Res 2012;91:305–10.
[11] Abu-Izze FO, Ramos GF, Borges ALS, Anami LC, Bottino MA. Fatigue behavior of ultrafine tabletop ceramic restorations. Dent Mater 2018;34:1401–9.

[12] Attia A, Kern M. Influence of cyclic loading and luting agents on the fracture load of two all-ceramic crown systems. J Prosthet Dent 2004;92:551–6.
[13] Campos F, Valandro LF, Feitosa SA, Kleverlaan CJ, Feilzer AJ, de Jager N, et al. Adhesive cementation promotes higher fatigue resistance to zirconia crowns. Oper Dent 2017;42:215–24.
[14] Preis V, Behr M, Hahnel S, Rosentritt M. Influence of cementation on in vitro performance, marginal adaptation and fracture resistance of CAD/CAM-fabricated ZLS molar crowns. Dent Mater 2015;31:1363–9.
[15] van den Breemer CR, Gresnigt MM, Cune MS. Cementation of glass-ceramic posterior restorations: a systematic review. Biomed Res Int 2015;2015:148954.
[16] Monteiro JB, Riquieri H, Prochnow C, Guilardi LF, Pereira GKR, Borges ALS, et al. Fatigue failure load of two resin-bonded zirconia-reinforced lithium silicate glass-ceramics: effect of ceramic thickness. Dent Mater 2018;34:891–900.
[17] Nasrin S, Katsube N, Seghi RR, Rokhlin SI. Approximate relative fatigue life estimation methods for thin-walled monolithic ceramic crowns. Dent Mater 2018;34:726–36.
[18] Holand W, Schweiger M, Frank M, Rheinberger V. A comparison of the microstructure and properties of the IPS Empress 2 and the IPS Empress glass-ceramics. J Biomed Mater Res 2000;53:297–303.
[19] Sen N, Us YO. Mechanical and optical properties of monolithic CAD-CAM restorative materials. J Prosthet Dent 2018;119:593–9.
[20] Pieger S, Salman A, Bidra AS. Clinical outcomes of lithium disilicate single crowns and partial fixed dental prostheses: a systematic review. J Prosthet Dent 2014;112:22–30.
[21] Rauch A, Reich S, Dalchau L, Schierz O. Clinical survival of chair-side generated monolithic lithium disilicate crowns: 10-year results. Clin Oral Investig 2018;22:1763–9.
[22] Gehrt M, Wolfart S, Rafai N, Reich S, Edelhoff D. Clinical results of lithium-disilicate crowns after up to 9 years of service. Clin Oral Investig 2013;17:275–84.
[23] Belli R, Wendler M, de Ligny D, Cicconi MR, Petschelt A, Peterlik H, et al. Chairside CAD/CAM materials. Part 1: measurement of elastic constants and microstructural characterization. Dent Mater 2017;33:84–98.
[24] Ramos Nde C, Campos TM, Paz IS, Machado JP, Bottino MA, Cesar PF, et al. Microstructure characterization and SCG of newly engineered dental ceramics. Dent Mater 2016;32:870–8.
[25] Furtado de Mendonca A, Shahmoradi M, Gouvea CVD, De Souza GM, Ellakwa A. Microstructural and mechanical characterization of CAD/CAM materials for monolithic dental restorations. J Prosthodont 2019;28:e587–94.
[26] Badawy R, El-Mowafy O, Tam LE. Fracture toughness of chairside CAD/CAM materials - Alternative loading approach for compact tension test. Dent Mater 2016;32:847–52.
[27] Elsaka SE, Elnaghy AM. Mechanical properties of zirconia reinforced lithium silicate glass-ceramic. Dent Mater 2016;32:908–14.
[28] Lawson NC, Bansal R, Burgess JO. Wear, strength, modulus and hardness of CAD/CAM restorative materials. Dent Mater 2016;32:e275–83.
[29] Hamza TA, Sherif RM. Fracture resistance of monolithic glass-ceramics versus bilayered zirconia-based restorations. J Prosthodont 2018;28:e259–64.
[30] Kim JH, Kim JW, Myoung SW, Pines M, Zhang Y. Damage maps for layered ceramics under simulated mastication. J Dent Res 2008;87:671–5.
[31] Rekow ED, Silva NR, Coelho PG, Zhang Y, Guess P, Thompson VP. Performance of dental ceramics: challenges for improvements. J Dent Res 2011;90:937–52.
[32] Wiskott HW, Nicholls JI, Belser UC. Stress fatigue: basic principles and prosthodontic implications. Int J Prosthodont 1995;8:105–16.
[33] Yu Zhang SB, Lawn Brian R. Competing fracture modes in brittle materials subject to concentrated cyclic loading in liquid environments: monoliths. J Mater Res 2005;20:2021–9.
[34] Zhang Y, Sailer I, Lawn BR. Fatigue of dental ceramics. J Dent 2013;41:1135–47.
[35] Kim JW, Kim JH, Thompson VP, Zhang Y. Sliding contact fatigue damage in layered ceramic structures. J Dent Res 2007;86:1046–50.
[36] Bonfante EA, Coelho PG. A critical perspective on mechanical testing of implants and prostheses. Adv Dent Res 2016;28:18–27.
[37] Bonfante EA, Coelho PG, Guess PC, Thompson VP, Silva NR. Fatigue anddamage accumulation of veneer porcelain pressed on Y-TZP. J Dent 2010;38:318–24.
[38] Bonfante EA, Sailer I, Silva NR, Thompson VP, Dianne Rekow E, Coelho PG. Failure modes of Y-TZP crowns at different cusp inclines. J Dent 2010;38:707–12.
[39] Coelho PG, Bonfante EA, Silva NR, Rekow ED, Thompson VP. Laboratory simulation of Y-TZP all-ceramic crown clinical failures. J Dent Res 2009;88:382–6.
[40] Coelho PG, Silva NR, Bonfante EA, Guess PC, Rekow ED, Thompson VP. Fatigue testing of two porcelain-zirconia all-ceramic crown systems. Dent Mater 2009;25:1122–7.
[41] El Zhawi H, Kaizer MR, Chughtai A, Moraes RR, Zhang Y. Polymer infiltrated ceramic network structures for resistance to fatigue fracture and wear. Dent Mater 2016;32:1352–61.
[42] Shembish FA, Tong H, Kaizer M, Janal MN, Thompson VP, Opdam NJ, et al. Fatigue resistance of CAD/CAM resin composite molar crowns. Dent Mater 2016;32:499–509.
[43] Silva NR, Bonfante EA, Rafferty BT, Zavanelli RA, Rekow ED, Thompson VP, et al. Modified Y-TZP core design improves all-ceramic crown reliability. J Dent Res 2011;90:104–8.
[44] Silva NR, Bonfante EA, Zavanelli RA, Thompson VP, Ferencz JL, Coelho PG. Reliability of metalloceramic and zirconia-based ceramic crowns. J Dent Res 2010;89:1051–6.
[45] Kelly JR, Rungruanganunt P, Hunter B, Vailati F. Development of a clinically validated bulk failure test for ceramic crowns. J Prosthet Dent 2010;104:228–38.
[46] Yan J, Kaizer MR, Zhang Y. Load-bearing capacity of lithium disilicate and ultra-translucent zirconias. J Mech Behav Biomed Mater 2018;88:170–5.
[47] Sato TP, Anami LC, Melo RM, Valandro LF, Bottino MA. Effects of surface treatments on the bond strength between resin cement and a new zirconia-reinforced Lithium silicate ceramic. Oper Dent 2016;41:284–92.
[48] Zhao WE, EA. A general accelerated life model for step-stress testing. IIE Trans 2005;37:1059–69.
[49] Kelly JR, Cesar PF, Scherrer SS, Della Bona A, van Noort R, Tholey M, et al. ADM guidance-ceramics: fatigue principles and testing. Dent Mater 2017;33:1192–204.
[50] Nelson W. Accelerated testing: statistical models, test plans and data analysis. John Wiley & Sons; 2004.
[51] Gibbs CH, Mahan PE, Lundeen HC, Brehnan K, Walsh EK, Sinkewiz SL, et al. Occlusal forces during chewing–influences of biting strength and food consistency. J Prosthet Dent 1981;46:561–7.
[52] Hattori Y, Satoh C, Kunieda T, Endoh R, Hisamatsu H, Watanabe M. Bite forces and their resultants during forceful intercuspal clenching in humans. J Biomech 2009;42:1533–8.
[53] Seghi RR, Daher T, Caputo A. Relative flexural strength of dental restorative ceramics. Dent Mater 1990;6:181–4.
[54] Wolf D, Bindl A, Schmidlin PR, Luthy H, Mormann WH. Strength of CAD/CAM-generated esthetic ceramic molar

implant crowns. Int J Oral Maxillofac Implants 2008;23:609–17.
[55] Murillo-Gomez F, Palma-Dibb RG, De Goes MF. Effect of acid etching on tridimensional microstructure of etchable CAD/CAM materials. Dent Mater 2018;34:944–55.
[56] El-Damanhoury HM, Gaintantzopoulou MD. Self-etching ceramic primer versus hydrofluoric acid etching: etching efficacy and bonding performance. J Prosthodont Res 2018;62:75–83.
[57] Siqueira FS, Alessi RS, Cardenas AF, Kose C, Souza Pinto SC, Bandeca MC, et al. New single-bottle ceramic primer: 6-month case report and laboratory performance. J Contemp Dent Pract 2016;17:1033–9.
[58] Oilo M, Gjerdet NR. Fractographic analyses of all-ceramic crowns: a study of 27 clinically fractured crowns. Dent Mater 2013;29:e78–84.
[59] Oilo M, Hardang AD, Ulsund AH, Gjerdet NR. Fractographic features of glass-ceramic and zirconia-based dental restorations fractured during clinical function. Eur J Oral Sci 2014;122:238–44.
[60] Moraguez OD, Wiskott HW, Scherrer SS. Three- to nine-year survival estimates and fracture mechanisms of zirconia- and alumina-based restorations using standardized criteria to distinguish the severity of ceramic fractures. Clin Oral Investig 2015;19:2295–307.