ACP AMERICAN COLLEGE OF PROSTHODONTISTS®

Fracture Strength and Fatigue Resistance of All-Ceramic Molar Crowns Manufactured with CAD/CAM Technology

Mohammed Zahran, BDS, MSc,[1] Omar El-Mowafy, BDS, PhD, FADM,[2] Laura Tam, DDS, MSc,[3] Philip A. Watson, DDS, MScD,[4] & Yoav Finer, DMD, PhD, FRCD(C), MSc[5]

[1]Ph.D Candidate, Department of Prosthodontics, Faculty of Dentistry, University of Toronto, Toronto, ON, Canada
[2]Professor, Department of Restorative Dentistry, Faculty of Dentistry, University of Toronto, Toronto, ON, Canada
[3]Associate Professor, Department of Restorative Dentistry, Faculty of Dentistry, University of Toronto, ON, Canada
[4]Professor and Head, Department of Biomaterials, Faculty of Dentistry, University of Toronto, ON, Canada
[5]Assistant Professor, Department of Biomaterials, Faculty of Dentistry, and Institute of Biomaterials and Biomedical Engineering, University of Toronto, Toronto, ON, Canada

Keywords
Fracture strength; zirconium oxide ceramics; feldspathic porcelain; cyclic loading; fatigue resistance.

Correspondence
Omar El-Mowafy, Restorative Dentistry, Faculty of Dentistry, University of Toronto, 124 Edward St., Toronto, ON, Canada, M5G 1G6. E-mail: oel.mowafy@utoronto.ca

This work was previously presented at the 84th General Session & Exhibition of the IADR, Brisbane, Australia.

Accepted January 26, 2007

doi: 10.1111/j.1532-849X.2008.00305.x

Abstract

Purpose: All-ceramic crowns are subject to fracture during function, especially in the posterior area. The use of yttrium-stabilized zirconium-oxide ceramic as a substructure for all-ceramic crowns to improve fracture resistance is unproven. The aim of this study was to compare fracture strength and fatigue resistance of new zirconium-oxide and feldspathic all-ceramic crowns made with computer-aided design/computer-aided manufacturing (CAD/CAM).
Materials and Methods: An ivorine molar was prepared to receive an all-ceramic crown. Using epoxy resin, 40 replication dies were made of the prepared tooth. Twenty feldspathic all-ceramic crowns (Vita Mark II) (VMII) and 20 zirconium-oxide crown copings (In-Ceram YZ) (YZ) were made using CAD/CAM technique (CEREC-3D). The YZ copings were sintered and veneered manually with a fine-particle ceramic (VM9). All crowns were cemented to their respective dies using resin cement (Panavia F 2.0). Ten crowns in each group were subjected to compressive fatigue loading in a universal testing machine (instron). The other ten crowns from each group were loaded to fracture at a crosshead speed of 1 mm/min. Data were statistically analyzed using independent *t*-test and Fisher's exact test at $\alpha = 0.05$.
Results: There was a significant difference between the survival rates of the two materials during the fatigue test ($p < 0.001$). All VMII crowns survived without any crack formation, while all YZ crowns fractured (40%) or developed cracks (60%). All the YZ crown fractures occurred within the veneering layer during the fatigue test. There was no significant difference in mean fracture load between the two materials ($p = 0.268$). Mean fracture loads (standard deviation) in N were: 1459 (492) for YZ crowns and 1272 (109) for VMII crowns.
Conclusions: The performance of VMII crowns was superior to YZ crowns in the fatigue test. The premature fractures and cracks of the YZ crowns were attributed to weakness in the YZ veneer layer or in the core/veneer bond.

The increasing demand for esthetics, combined with health and environmental concerns about some metallic restorations, has stimulated research in metal-free, tooth-colored restorations. All-ceramic restorations have several advantages, including life-like appearance,[1] biocompatibility,[2] wear resistance, and color stability.[3] Disadvantages include less-than-ideal marginal adaptation, excessive wear of the opposing dentition, aggressive preparation design,[1] technique sensitivity,[4] and susceptibility to fracture. The ability of all-ceramic restorations to withstand occlusal forces is compromised by the presence of two types of inherent flaws:[5] fabrication defects (internal voids, porosities, or microstructural features that arise during processing) and surface cracks (defects on the surface as a result of machining and grinding process).[6] Failure begins with microscopic damage resulting from the interaction of preexisting defects with applied loads.[5] Failure can also occur because of impact forces or subcritical crack growth,[7,8] which is enhanced in an aqueous environment.[9]

Table 1 Physical properties of materials used in the study

Material	Elastic modulus (GPa)	Coefficient of thermal expansion (10^{-6} K^{-1})	Flexure strength (MPa)	Fracture toughness (MPa × $m^{1/2}$)
Vita Mark II (Vita Zahnfabrik)	63*	8.8*	103†	1.26‡
VM9 (Vita Zahnfabrik)	–	8.8-9.2¶	100¶	–
In-Ceram YZ (Vita Zahnfabrik)	210¶	10.5¶	>900¶	5.9¶
Panavia F 2.0 (Kuraray America, Inc.)	9.6¶	–	79¶	–
Epoxy resin die material (Viade Products, Inc.)	12.9§	–	–	–

*Jedynakiewicz,[25] †Bindl et al,[26] ‡Thompson et al,[27] ¶Manufacturer's data, §Neiva et al.[28]

In a systematic review of clinical complications in fixed prosthodontics,[10] all-ceramic crowns showed an 8% incidence of complications, with crown fractures being the most common. Molars showed a higher fracture rate (21%) than premolars and anterior teeth (7% and 3%, respectively). Kelly reported similar results with higher fracture rates in posterior crowns compared to anteriors.[11]

Clinically, dental restorations are subject to cyclic forces ranging from 60 to 250 N during function and up to 500 to 800 N for short periods;[5] however, the range varies according to location: maximum occlusal forces are 400 to 890 N in the molar region, 222 to 445 N in the premolar area, 133 to 334 N in the cuspid area, and 89 to 111 N in the incisor region.[12] In an average individual, the number of cycles of mastication per day ranges from 800 to 1400;[5] however, it may reach up to 2700.[13] This translates to a range from 290,000 to 10^6 cycles/year. This number should be reduced by a factor ranging from 5 to 20, because not all chewing cycles are active (i.e., loaded during the chewing cycle).[13] In addition, all-ceramic restorations are affected by thermal changes and oral fluids. A decrease in strength after thermal and/or mechanical fatiguing has been reported for different ceramic materials.[14–21]

Many attempts have been made to increase the fracture strength of all-ceramic restorations. The use of yttrium-oxide partially-stabilized zirconium oxide crystals in dental ceramics has been shown to increase their flexural strength.[22] The tetragonal crystals in these zirconium oxide ceramics are metastable and can be transformed into larger monoclinic crystals with the application of stress from cracks or flaws.[23] This phenomenon is beneficial in hindering crack growth and increasing fracture toughness, hence, it is referred to as "transformation toughening."

In-Ceram YZ (YZ) (Vita Zahnfabrik, Bad Sackingen, Germany) is one of the available yttrium-oxide partially-stabilized zirconium oxide ceramics machined using a computer-aided design/computer-aided manufacturing (CAD/CAM) device. According to the manufacturer's product information, the material has a flexural strength greater than 900 MPa and a fracture toughness of 5.9 MPa/$m^{-1/2}$. Therefore, dental restorations made using these zirconium oxide systems should resist fracture better than traditional all-ceramic restorations, especially in areas where high occlusal loads are anticipated. However, zirconium oxide ceramics are not transparent and cannot be stained to create good esthetic results. They must therefore be veneered with suitable veneering porcelain to enhance the esthetic results.[24] Unfortunately, this multilayer arrangement increases the complexity of stress distribution within the restoration, making it difficult to predict its performance in clinical situations.

The aim of this study was to compare the fracture and fatigue resistance of new zirconium all-ceramic crowns to feldspathic crowns fabricated using CAD/CAM (CEREC 3, Sirona Dental Systems GmbH, Bensheim, Germany).

Materials and methods

Table 1 lists the materials used and some of their physical properties.

Tooth preparation

An ivorine mandibular molar replica was prepared to receive an all-ceramic crown following dimensions recommended for the CEREC 3 system (Fig 1). The tooth had an occlusal reduction

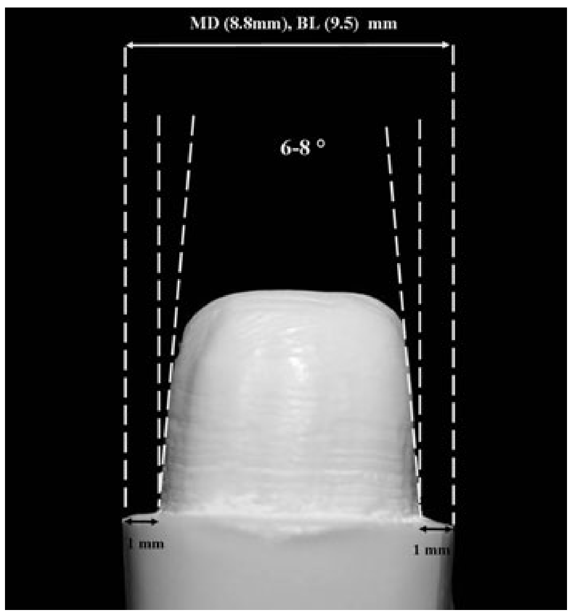

Figure 1 Preparation dimensions of the master die (buccal view).

of 1.5 mm and minimum axial reduction of 1 mm with a 6 to 8° total angle of convergence (3 to 4° on each side). The gingival margin was prepared with a circumferential shoulder at least 1 mm wide. Lingual and facial surfaces were prepared in two planes, and all line angles were smoothed to reduce stress concentration. The prepared ivorine molar was used as a master die to fabricate 40 prepared tooth replicas using a highly filled epoxy resin (Viade Products, Inc. Camarillo, CA) with a modulus of elasticity similar to that of human dentin (12.9 GPa). This material responded to 34% phosphoric acid etching by forming a surface microroughness suitable for bonding.[28]

Radiographic images of all prepared tooth replicas were made to verify that the replicas contained no significant voids. The replicas were also measured faciolingually and mesiodistally using a digital caliper at predetermined locations to verify dimensional accuracy. These measurements revealed a low variability of about 10 to 20 μm.

Impressions and restoration design

The prepared molar replicas were inserted in a dentiform with mesial and distal adjacent ivorine teeth in place. CEREC imaging liquid (polysorbate liquid, Vita Zahnfabrik) was used to create a sticky surface on the molars and the adjacent teeth. The surfaces were then covered with a thin layer of optical reflective medium (titanium dioxide, Vita Zahnfabrik). An optical impression of the prepared tooth with its adjacent teeth was then captured using the CEREC 3 intraoral camera.

An artificial ivorine unprepared mandibular molar was adjusted occlusally in the central fossa region to create the morphology to receive a loading ball of 3 mm diameter. This molar was inserted into a dentiform with adjacent teeth, and an optical occlusal impression was captured using the intraoral camera. CEREC 3 and CEREC software (Version 2.40 R1800) were used to design 20 crowns according to the correlation mode (to correlate the anatomy of the crowns to the adjusted mandibular molar) and 20 crown copings.

Crown fabrication

Twenty monolithic crowns were fabricated from Vita Mark II (VMII) blocks (shade A 3.5 on Vitapan, 14 mm long, Vita Zahnfabrik), and 20 crown copings were fabricated from YZ cubes using the CEREC 3 milling unit (serial number 02527). Cutting diamonds were changed after milling eight crowns or eight crown copings. Following each instrument change, the milling unit was calibrated using the CEREC calibration kit.

The VMII crowns were glazed using an Akzent glazing kit (Vita Zahnfabrik) according to the manufacturer's instructions as follows: predrying temperature of 600°C, increase temperature at the rate of 58°C/min with closing time of 6 minutes and a final firing temperature of 950°C with a holding time of 1 minute.

The YZ copings had a 0.7 mm occlusal wall thickness and 0.5 mm circumferential wall thickness. The copings were sintered and veneered with porcelain (VM9 system, shade A3.5, Vita Zahnfabrik) following manufacturer's instructions. The sintering process began at <200°C, followed by an increase in temperature at the rate of 600°C/h, with a final sintering temperature of 1530°C and cooling phase at 600°C/h. The duration of the program including the cooling phase was approximately 8 hours. The veneering porcelain was applied in layers (base dentin "dentin wash," dentin, enamel, and Akzent finishing agent). The "effect bonder" layer was eliminated, because it has been reported that the best bonding between YZ and VM9 is achieved when a dentine wash is fired directly to the zirconium oxide core following the firing protocol of the effect bonder.[29]

A special furnace (Vita Vacumat 4000 Premium, Vita Zahnfabrik, serial number 20010184) was used to sinter the YZ core and to fire the veneering porcelain. To minimize variability among crowns, one Vita-certified and experienced technician applied the veneering porcelain to the YZ copings. The technician was provided with a loading ball and a representative VMII crown, and was instructed to reproduce as closely as possible the size and anatomy of the VMII crown.

Using a caliper (Buffalo Dental Manufacturing Co., Syosset, NY), all crown thicknesses at the central fossae were verified to have a minimum thickness of 1.5 mm. The maximum mesiodistal and buccolingual dimensions at the height of contour were measured with a digital caliper (Mitutoyo Corporation, Tokyo, Japan). There was no statistically significant difference between the means of the crowns' dimensions for each material when compared using independent *t*-test ($\alpha = 0.05$).

Cementation

Before cementation, each crown was seated on its respective tooth replica to check its marginal fit. A measuring microscope (Mitutoyo Corporation) was used to determine the marginal gap at three points (at the middle and near the two line angles) on each surface. Any crown with a mean marginal gap more than 150 μm was excluded. None of the YZ crowns was rejected, while five crowns in VMII group were rejected and new crowns were remade.

The internal surfaces of VMII crowns were treated with 9.6% hydrofluoric acid etching gel (Pulpdent Corporation, Watertown, MA) for 2 minutes.[30] The etched internal surface was cleaned using a water spray, followed by ultrasonic cleaning in distilled water for 60 seconds.[31] Because hydrofluoric acid does not etch YZ copings effectively, the internal surfaces of YZ crowns were grit-blasted for 5 seconds with 50 μm Al_2O_3 particles using a microetcher under 2 bar pressure.[32] The prepared surfaces of molar teeth replicas were etched with 40% phosphoric acid (Kuraray America, Inc., New York, NY) for 1 minute. The etched surfaces were then rinsed with water spray and dried with compressed oil-free air.

A bonding/silane coupling agent containing adhesive phosphate monomer (Clearfil SE bond/porcelain bond activator, Kuraray America, Inc.) was applied to the internal surfaces of both material-type crowns according to the manufacturer's recommendations.

All crowns were cemented with dual-polymerized phosphate-modified resin cement (Panavia F 2.0, Kuraray America, Inc.) according to the manufacturer's instructions. The crowns were initially seated on their respective replicas using finger pressure, the excess cement was removed, and an air-sealing gel (Oxyguard, Kuraray America, Inc.) was applied to the crown margins for 3 minutes as recommended by the manufacturer. The crowns were then placed under a static load of 22 N for 5 minutes.[33] After load removal, each crown was light-polymerized (Optilux 501, Kerr Demetron, Danbury, CT) for 20 seconds on each surface. One hour after cementation, the crowns were stored in 37°C distilled water for 1 week.

The dimensions of each specimen from the occlusal surface of the crown to the apical surface of the replica were measured before and after cementation to verify crown seating using a digital caliper (Mitutoyo Corporation). None of the crowns demonstrated an increase in vertical dimension greater than 50 μm; therefore, none were rejected.

Mechanical cyclic fatigue and fracture tests

All cemented crowns with their respective dies were mounted in resin material (SR Ivolen, Ivoclar Vivadent, Schaan, Liechtenstein) with dimensions suitable for attachment to a loading jig in a water bath. A 3-mm diameter stainless steel ball was placed on the previously adjusted area on the central fossa of the crown. All crowns were loaded in distilled water at room temperature to mimic the hydrolytic effect of saliva on the ceramic (static fatigue).

Each material group was randomly divided into two subgroups of 10 specimens each. Specimens of one subgroup were uniaxially-loaded in a universal testing machine (Instron 8501, Instron, Canton, MA) using a crosshead speed of 1 mm/min, and the load at fracture was recorded. Specimens of the other subgroup were subjected to mechanical cyclic loading at loads ranging between 50 and 600 N for 500,000 cycles at a frequency of 20 Hz. The Instron machine was adjusted to stop if the deformation increased more than 0.15 mm. After cyclic loading, the crowns were examined using a light microscope. Crowns were classified as cracked, fractured, or no cracks and/or fractures. The load-displacement data from the Instron machine were imported to a software program (Microsoft Office Excel, Redmond, WA), and load-displacement curves were created. For each curve, the first drop was marked, and the corresponding load (N) was recorded as load at failure.

Four YZ crowns in which the zirconium oxide core remained intact after either fatigue or fracture testing were sectioned axially in a buccolingual direction through the loading area for further light microscopy and scanning electron microscopy (SEM) examination. Two other YZ crowns from the fatigue subgroup were selected for surface examination with SEM. For SEM, the sections were sputter-coated with 7 nm of platinum in a Polaron E5100 coating unit (Polaron Equipment, Ltd., Bedford, UK) and examined using SEM (Hitachi S-2500, Hitachi, Mito City, Japan).

Statistical analysis

The data were analyzed using a software program (Version 13.0, SPSS, Inc., Chicago, IL). All statistical tests were two-tailed and performed at an alpha level of 0.05. Results of the crown examination after the fatigue test were analyzed using chi-square test followed by Fisher's exact test. An independent sample *t*-test was used to compare fracture strength of the crowns made from the two materials.

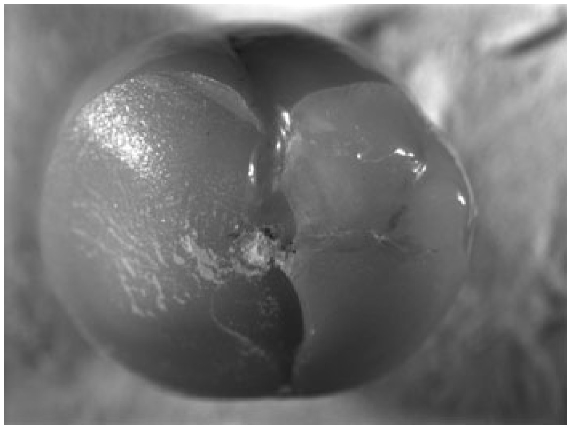

Figure 2 In-Ceram YZ crown fractured during fatigue testing.

Results

Fatigue test

All ten VMII crowns survived the fatigue test without any evidence of fractures and/or cracks, while the veneering porcelain of four YZ crowns fractured during the fatigue test (Fig 2) and the remainder had cracks (Fig 3). Chi-square analysis of the fatigued crowns revealed a significant difference between the two groups ($p < 0.001$). Because the expected proportion in about 67% of the cells was less than 5, the chi-square results should be considered cautiously, and Fisher's exact test is preferred. Due to the limitation of the statistics software, this test can be used only for 2×2 tables. Therefore, the data were modified to create a 2×2 table format, and the fractured and cracked crowns were combined into a single group. The Fisher's exact test on the new 2×2 table revealed a statistically significant difference between the two materials ($p < 0.001$).

Fracture test

Nine out of ten VMII crowns showed total fracture through the whole crown thickness. In contrast, all YZ crown fractures occurred within the veneering porcelain.

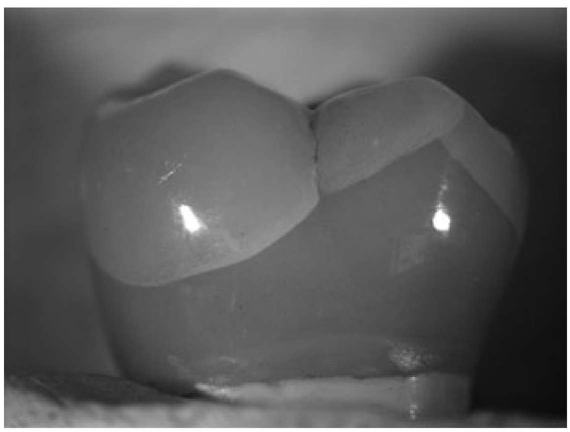

Figure 3 In-Ceram YZ crown cracked during fatigue testing.

Table 2 Fracture strength (N) of Vita Mark II and In-Ceram YZ

	N	Median	Mean	SD	Maximum	Minimum
Vita Mark II	10	1279	1272	109	1484	1038
In-Ceram YZ	10	1511	1459	492	2374	665

The YZ subgroup revealed a higher mean fracture load (1459 N) than VMII (1272 N); however, the independent sample *t*-test revealed no statistically significant difference ($p = 0.268$). The median, mean, standard deviation, and minimum and maximum fracture loads are shown in Table 2.

Microscopic examination

Cracks were observed in the veneering layer (Figs 4 and 5). The core material showed a homogenous structure compared to the veneering porcelain where obvious flaws and voids were detected (Fig 6).

Surface examination of the fractured fatigued crowns revealed a combination of veneering porcelain fracture and delamination. The fracture within the veneering porcelain involved the different layers used to build up the veneering layer (Fig 7). Some of the SEM pictures revealed exposed zirconium oxide crystals (Fig 8), which might be due to delamination of the veneering porcelain (adhesive failure) or due to fracture of the core ceramic (cohesive failure).

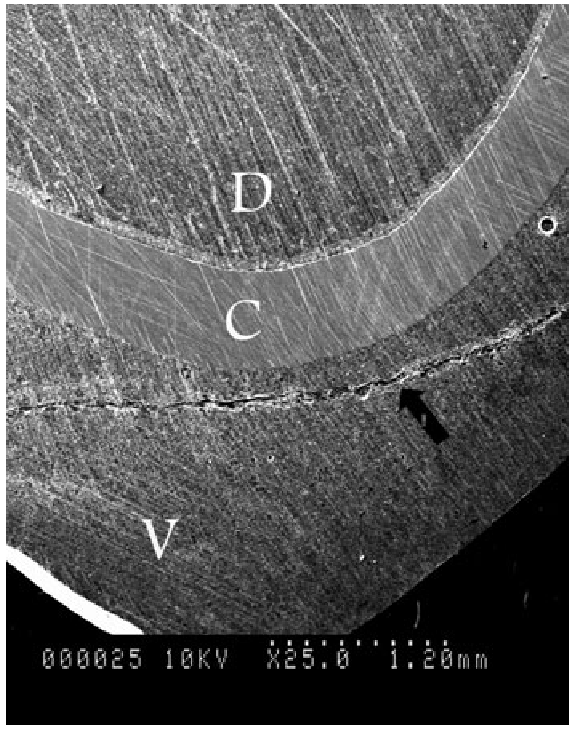

Figure 5 SEM image of In-Ceram YZ fatigued crowns revealing crack in the veneering layer (black arrow) (C = core; D = die; V = veneer).

Discussion

The recommendations for a clinically relevant in vitro load-to-failure test for all-ceramic restorations described by Kelly were followed in this study,[34] including using a die material with elastic modulus similar to dentin, preparing the teeth or dies according to clinical guidelines, using all-ceramic crowns with clinically relevant dimensions, and using a reliable, commonly used luting cement. An aqueous environment was maintained during cyclic-loading. A preliminary study was conducted to ensure that the epoxy die material responded favorably to phosphoric acid-etching and bonded well to resin cement.

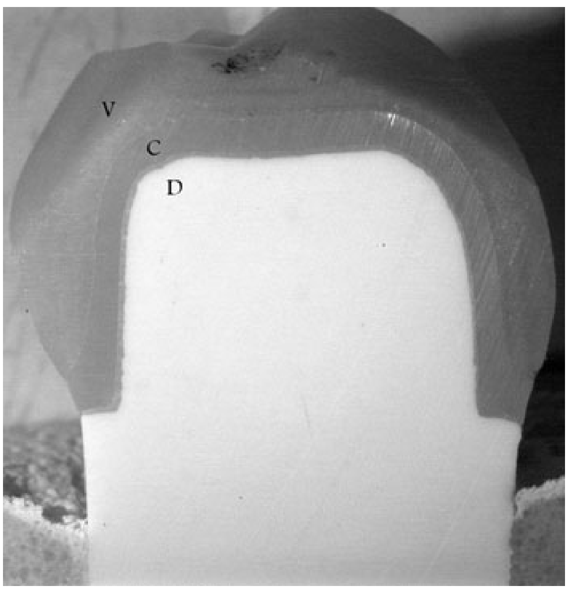

Figure 4 Light microscope image of In-Ceram YZ fatigued crowns revealing veneering layer fracture (C = core; D = die; V = veneer).

Clinically, all-ceramic restorations commonly fail through slow crack growth resulting from fatigue caused by masticatory stresses. The crowns in this study were fatigued using a range of forces similar to which crowns in the molar region might be subjected. The crowns were loaded for 500,000 cycles. This represents 10 years of normal function in the oral cavity,[13] or as suggested by Kelly, about half a year of continuous bruxism.[34]

As all the forces in the cyclic loading were vertical and other forces were ignored due to the loading machine limitations, the clinical implication of the results in this study should be limited to the vertical loading situation. The use of a higher frequency (20 Hz) in the cyclic loading test rather than a lower frequency (1 to 2 Hz) as observed in chewing cycles was unavoidable because of the budget limitation accompanied by the increase in the number of cycles and the sample size. Such high frequency may lead to more heat generation compared to 1 to 2 Hz, and may not give a time for stress relaxation. Kelly et al[35] used 20 Hz frequency for cyclic loading of leucite-reinforced all-ceramic crowns using a staircase approach between 100 to 600 N, with a 100 N step size for 10^6 cycles in water, and were able to measure fracture loads reasonable for clinical relevance.

The use of a small ball increased the contact pressure in the crown system compared to the clinical contact pressure[34].

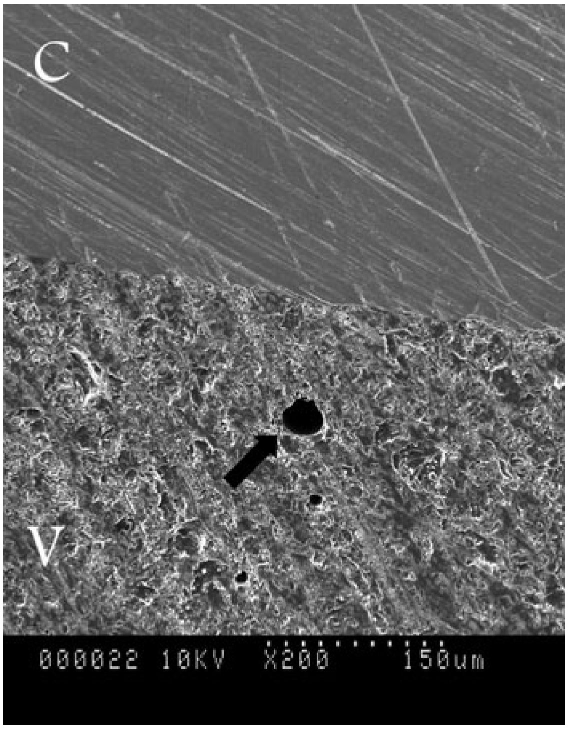

Figure 6 SEM image of In-Ceram YZ fatigued crowns revealing void within the veneering layer (black arrow) (C = core; V = veneer).

The contact pressure is influenced by the ratio of the elastic modulus of the dental porcelain to the elastic modulus of the loading ball, and by the radius of the loading ball.[36] Alternative methods could have included the use of a loading ball with a modulus of elasticity lower than that of the stainless steel ball, a tin sheet between the load applicator and crown as stress breaker, or a stainless steel loading piston with its end machined to a curvature equivalent to 40- to 50 mm diameter to reproduce clinical contact pressure.[34] In this study, both materials were

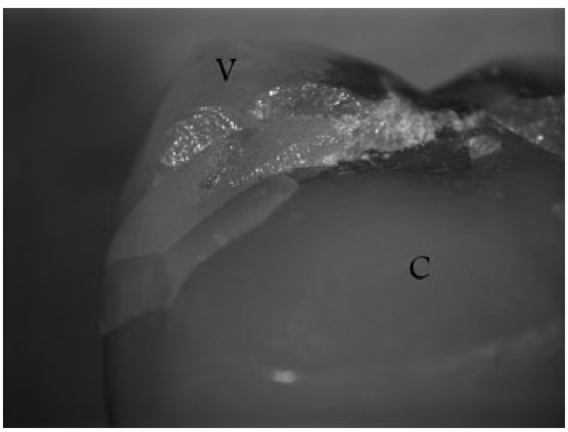

Figure 7 Light microscope image of In-Ceram YZ fatigued crowns revealing fracture involving multiple layers of the veneering porcelain (C = core; V = veneer).

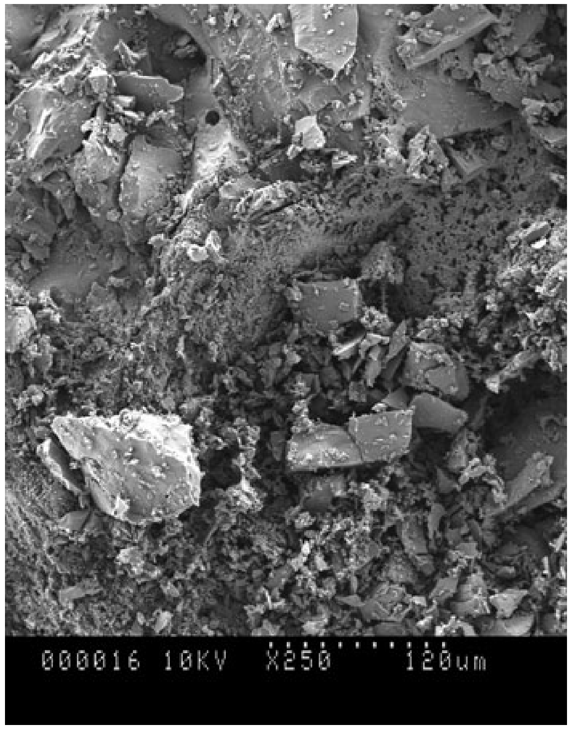

Figure 8 Zirconium oxide crystals exposed as a result of veneer delamination or cohesive failure within the core material.

subjected to the same load frequency and contact pressure. The fracture strength data in this study should be considered relative, not as "absolute" values, and extrapolation of these in vitro strength data to the clinical performance must be considered cautiously and within the limitation of the study.[11]

Several studies reported the mean fracture strength of VMII molar crowns to range between 600 and 3000 N.[15,31,37,38] Their findings varied as a result of using different methodologies. In this study, the mean fracture strength of VMII crowns was 1272 N and was within the previously reported range. All VMII crowns except one showed catastrophic fracture involving the whole thickness of the ceramic crown. This is the expected mode of fracture for monolithic all-ceramic crowns.[39] None of the VMII crowns subjected to cyclic loading demonstrated any evidence of cracking. Previous studies did not report any failure of VMII crowns during fatigue testing.[15,37]

In the current study, 40% of fatigued YZ crowns fractured within the veneering porcelain during the fatigue test, and the remainder had cracks. Zirconium oxide ceramic systems were recently introduced to the marketplace, and the literature is lacking information about the fracture strength of YZ crowns.[40-44] Only one study subjected molar zirconium oxide (Procera AllZirkon, Nobel Biocare, Goteborg, Sweden) crowns to 10,000 mechanical fatigue (30 to 300 N) cycles at 1 Hz.[43] None of the fatigued crowns were reported to have any fractures or cracks during cyclic fatiguing. The dichotomy between the results of this current study with that study can be explained by the lower range of load and the fewer cycles used in that study, and perhaps by the different properties of the two proprietary zirconium oxide systems.

Guazzato et al[45] studied the fracture mode of bilayer porcelain/zirconia samples and found that fracture initiation was through the development of a Hertzian cone crack at low force. When the force increased, this crack propagated and approached the core with an acute angle. As the crack reached the core–veneer interface, it was hindered, and extended laterally parallel to the interface. This lateral extension resulted in delamination of the veneering porcelain, and in 80% of the cases, was followed by crushing of the contact area. Similarly, all YZ crowns in this study started their fracture with chipping or delamination of the veneering porcelain. Examination with light microscopy and SEM revealed that the fracture was a combination of chipping within the veneering porcelain and delamination at the core–veneer interface. Interfacial failure was the basic cause of failure for Y-TZP crowns in other in vitro studies.[42-44] Data collected from failed clinical crowns showed a similar failure mode for multilayer crowns.[46]

In general, the fracturing of multilayer crowns starts at their weakest part. In the case when a stronger and stiffer core substructure is veneered with weaker porcelain, the failure typically occurs in the weak veneering porcelain or at the bond between the core and veneer.[47] The results of this study suggest that the weakness in the YZ crowns also lies in the veneering porcelain or at the core–veneer interface.

The product profile for VM9 reported a flexural strength of 100 MPa. This flexural strength could be lower because of the residual tensile stresses or structural flaws that could develop during processing.[45,46,48] SEM examination revealed multiple voids within the veneering porcelain, which are expected to develop as a result of the human factor in building up these crowns.

The mechanism of adhesion between zirconium oxide ceramics and their veneering porcelain has not as yet been established. The chemical structure at the core-veneer interface was analyzed using energy-dispersive X-ray. This revealed that some of the veneer elements diffused into the zirconium oxide layer to a depth of 8 to 10 μm.[47] Two studies reported the core-to-veneer bond strengths for different zirconium oxide systems.[49,50] The core–veneer bond strengths of the studied systems ranged between 14 and 40 MPa, and were either comparable to or stronger than that of the control group (metal-ceramic bond strength).[50] All studied systems showed cohesive fractures of the veneers, interfacial core–veneer fractures, or a combination of both.

In general, the core–veneer bond can be severely affected by the flaws and/or stresses at the interface. Interfacial flaws might develop during veneering porcelain application due to poor wettability of the core by the veneering porcelain.[49,51] Interfacial stresses can arise as a result of firing shrinkage of the veneering porcelain,[47] the mismatch in the coefficient of thermal expansion for the core and the veneer,[49] or the transformation of the tetragonal crystals to monoclinic crystals,[52,53] which can be initiated during the multiple firing of veneering porcelain on the zirconium oxide core.[54]

In our study, VMII crowns showed more fatigue resistance than YZ. This finding could be explained by the complex nature of stresses and flaws that might develop during fabrication of the multilayer YZ crowns compared to the ones found in homogenous monolithic VMII crowns.

Conclusions

Within the limitations of this in vitro study, the following conclusions can be drawn:

1. There was no statistically significant difference between the mean fracture loads of the VMII and YZ crowns; however, this finding should be interpreted with caution as the two materials behaved differently under conditions of cyclic loading.
2. VMII crowns showed better fatigue resistance compared to YZ and should be expected to better resist crack propagation when used clinically.
3. VMII and YZ showed different fracture modes. Most VMII crowns showed total crown fracture involving the whole crown thickness, while all YZ crowns had their fractures in the veneering layer. The addition of the veneer layer to the core is the weak link in the YZ crowns, and attention is needed to improve core–veneer bond.

Acknowledgments

This work was supported by a grant from the Faculty of Dentistry Research Committee, University of Toronto, and through donated materials from Patterson Dental, Kuraray, and Vident, to whom the authors are grateful.

References

1. Rosenstiel SF, Land MF, Fujimoto J: Contemporary Fixed Prosthodontics. St. Louis, MO, Mosby, 2001, pp. 262-271
2. Sjogren G, Sletten G, Dahl JE: Cytotoxicity of dental alloys, metals, and ceramics assessed by millipore filter, agar overlay, and MTT tests. J Prosthet Dent 2000;84:229-236
3. Siervo S, Pampalone A, Siervo P, et al: Where is the gap? Machinable ceramic systems and conventional laboratory restorations at a glance. Quintessence Int 1994;25:773-779
4. Blatz MB: Long-term clinical success of all-ceramic posterior restorations. Quintessence Int 2002;33:415-426
5. Drummond JL, Eliades G: Ceramic Behavior Under Different Environmental and Loading Conditions. Dental Materials In Vivo: Aging and Related Phenomena. Chicago, IL, Quintessence, 2003, pp. 35-45
6. Rosenstiel SF, Land MF, Fujimoto J: Contemporary Fixed Prosthodontics. St. Louis, MO, Mosby, 2001, pp. 643-672
7. Evans AG: Slow crack growth in brittle materials under dynamic loading conditions. Int J Fract 1974;10:251-259
8. Evans AG, Johnson H: Fracture stress and its dependence on slow crack growth. J Mater Sci 1975;10:214-222
9. Wiederho SM: Moisture assisted crack growth in ceramics. Int J Fract Mech 1968;4:171-177
10. Goodacre CJ, Bernal G, Rungcharassaeng K, et al: Clinical complications in fixed prosthodontics. J Prosthet Dent 2003;90:31-41
11. Kelly JR: Dental ceramics: current thinking and trends. Dent Clin North Am 2004;48:viii, 513-530
12. Anusavice KJ, Phillips RW: Phillips' Science of Dental Materials. Philadelphia, PA, Saunders, 1996, p. 66
13. Wiskott HW, Nicholls JI, Belser UC: Stress fatigue: basic principles and prosthodontic implications. Int J Prosthodont 1995;8:105-116

14. 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;92:551-556
15. Chen HY, Hickel R, Setcos JC, et al: Effects of surface finish and fatigue testing on the fracture strength of CAD-CAM and pressed-ceramic crowns. J Prosthet Dent 1999;82:468-475
16. Kappert HF, Knode H, Schultheiss R: Strength of the In-Ceram system under mechanical loads and thermocycling in artificial saliva. Dtsch Zahnarztl Z 1991;46:129-131
17. Kern M, Douglas WH, Fechtig T, et al: Fracture strength of all-porcelain, resin-bonded bridges after testing in an artificial oral environment. J Dent 1993;21:117-121
18. Ohyama T, Yoshinari M, Oda Y: Effects of cyclic loading on the strength of all-ceramic materials. Int J Prosthodont 1999;12:28-37
19. Sobrinho LC, Cattell MJ, Glover RH, et al: Investigation of the dry and wet fatigue properties of three all-ceramic crown systems. Int J Prosthodont 1998;11:255-262
20. Sobrinho LC, Glover RH, Knowles JC, et al: Comparison of the wet and dry fatigue properties of all ceramic crowns. J Mater Sci Mater Med 1998;9:517-521
21. Strub JR, Beschnidt SM: Fracture strength of 5 different all-ceramic crown systems. Int J Prosthodont 1998;11:602-609
22. Guazzato M, Quach L, Albakry M, et al: Influence of surface and heat treatments on the flexural strength of Y-TZP dental ceramic. J Dent 2005;33:9-18
23. Guazzato M, Albakry M, Ringer SP, et al: Strength, fracture toughness and microstructure of a selection of all-ceramic materials. Part II. Zirconia-based dental ceramics. Dent Mater 2004;20:449-456
24. White SN, Miklus VG, McLaren EA, et al: Flexural strength of a layered zirconia and porcelain dental all-ceramic system. J Prosthet Dent 2005;94:125-131
25. Jedynakiewicz NM: Vitablocks Mark II CEREC Porcelain. CAD-CAM in Restorative Dentistry: CEREC Method. Liverpool, UK, Liverpool University Press, 1993, pp. 147-148
26. Bindl A, Luthy H, Mormann WH: Fracture load of CAD/CAM-generated slot-inlay FPDs. Int J Prosthodont 2003;16:653-660
27. Thompson JY, Bayne SC, Heymann HO: Mechanical properties of a new mica-based machinable glass ceramic for CAD/CAM restorations. J Prosthet Dent 1996;76:619-623
28. Neiva G, Yaman P, Dennison JB, et al: Resistance to fracture of three all-ceramic systems. J Esthet Dent 1998;10:60-66
29. Giordano R, Abrosa C, Ober R, et al: Bond strength of porcelain to zirconia. J Dent Res 2005;84 (Special Issue A): (Abstract 544)
30. Blatz MB, Sadan A, Kern M: Resin-ceramic bonding: a review of the literature. J Prosthet Dent 2003;89:268-274
31. Attia A, Kern M: Fracture strength of all-ceramic crowns luted using two bonding methods. J Prosthet Dent 2004;91:247-252
32. Blatz MB: Cementation of zirconium-oxide ceramic restorations. Pract Proced Aesthet Dent 2004;16:14
33. Pallis K, Griggs JA, Woody RD, et al: Fracture resistance of three all-ceramic restorative systems for posterior applications. J Prosthet Dent 2004;91:561-569
34. Kelly JR: Clinically relevant approach to failure testing of all-ceramic restorations. J Prosthet Dent 1999;81:652-661
35. Kelly JR, Hunter BD, Brenyo MR, et al: Simulating clinical failure during in vitro testing of all-ceramic crowns. J Dent Res 1998;77 (Special Issue B):778 (Abstract 1175)
36. Lawn B: Fracture of Brittle Solids (ed 2). Cambridge, UK, Cambridge University Press, 1993, pp. 249-306
37. Attia A, Abdelaziz KM, Freitag S, et al: Fracture load of composite resin and feldspathic all-ceramic CAD/CAM crowns. J Prosthet Dent 2006;95:117-123
38. Lampe K, Luthy H, Mormann WH, et al: Fracture Load of All-ceramic Computer Crowns. CAD/CIM in Aesthetic Dentistry: CEREC 10 Year Anniversary Symposium. Chicago, IL, Quintessence, 1996, pp. 463-482
39. Kelly JR, Giordano R, Pober R, et al: Fracture surface analysis of dental ceramics: clinically failed restorations. Int J Prosthodont 1990;3:430-440
40. Luthardt RG, Sandkuhl O, Reitz B: Zirconia-TZP and alumina–advanced technologies for the manufacturing of single crowns. Eur J Prosthodont Restor Dent 1999;7:113-119
41. Snyder MD, Hogg KD: Load-to-fracture value of different all-ceramic crown systems. J Contemp Dent Pract 2005;6:54-63
42. Sundh A, Sjogren G: A comparison of fracture strength of yttrium-oxide- partially-stabilized zirconia ceramic crowns with varying core thickness, shapes and veneer ceramics. J Oral Rehabil 2004;31:682-688
43. Vult von SP: Fracture strength of two oxide ceramic crown materials after cyclic preloading and thermocycling. Swed Dent J Suppl 2005;173:1-21
44. Wood KC, Berzins DW, Luo Q, et al: Resistance to fracture of two all-ceramic crown materials following endodontic access. J Prosthet Dent 2006;95:33-41
45. Guazzato M, Proos K, Quach L, et al: Strength, reliability and mode of fracture of bilayered porcelain/zirconia (Y-TZP) dental ceramics. Biomaterials 2004;25:5045-5052
46. Thompson JY, Anusavice KJ, Naman A, et al: Fracture surface characterization of clinically failed all-ceramic crowns. J Dent Res 1994;73:1824-1832
47. Aboushelib MN, Kleverlaan CJ, Feilzer AJ: Microtensile bond strength of different components of core veneered all-ceramic restorations. Part II: zirconia veneering ceramics. Dent Mater 2006;22:857-863
48. Isgro G, Pallav P, Van Der Zel JM, et al: The influence of the veneering porcelain and different surface treatments on the biaxial flexural strength of a heat-pressed ceramic. J Prosthet Dent 2003;90:465-473
49. Aboushelib MN, de JN, Kleverlaan CJ, Feilzer AJ: Microtensile bond strength of different components of core veneered all-ceramic restorations. Dent Mater 2005;21:984-991
50. Al-Dohan HM, Yaman P, Dennison JB, et al: Shear strength of core-veneer interface in bi-layered ceramics. J Prosthet Dent 2004;91:349-355
51. Kelly JR, Tesk JA, Sorensen JA: Failure of all-ceramic fixed partial dentures in vitro and in vivo: analysis and modeling. J Dent Res 1995;74:1253-1258
52. Ardlin BI: Transformation-toughened zirconia for dental inlays, crowns and bridges: chemical stability and effect of low-temperature aging on flexural strength and surface structure. Dent Mater 2002;18:590-595
53. Chevalier J, Deville S, Munch E, et al: Critical effect of cubic phase on aging in 3mol% yttria-stabilized zirconia ceramics for hip replacement prosthesis. Biomaterials 2004;25:5539-5545
54. Sundh A, Molin M, Sjogren G: Fracture resistance of yttrium oxide partially-stabilized zirconia all-ceramic bridges after veneering and mechanical fatigue testing. Dent Mater 2005;21:476-482