ORIGINAL ARTICLE

Fracture resistance of porcelain veneered zirconia crowns with exposed lingual zirconia for anterior teeth after thermal cycling: An in vitro study

CrossMark

Fatemeh A. Amir Rad [a], **Faysal G. Succaria** [b], **Steven M. Morgano** [c,*]

[a] Department of Prosthodontics, Dubai School of Dental Medicine, Dubai, United Arab Emirates
[b] Department of Restorative Sciences and Biomaterials, Boston University Henry M. Goldman School of Dental Medicine, Boston, MA, USA
[c] Department of Restorative Dentistry, Rutgers School of Dental Medicine, Newark, NJ, USA

Abstract *Statement of problem:* In some clinical conditions minimally invasive complete crown tooth preparations are indicated. This is especially true when gross removal of tooth structure would weaken the remaining tooth or violate the vitality of the dental pulp.

Objective: The purpose of this study was to investigate the influence of (1) exposed lingual zirconia with veneered zirconia crowns, and (2) reduced lingual thickness of monolithic lithium disilicate crowns on the fracture resistance of the crowns after cyclic loading. Metal-ceramic crowns with exposed lingual metal served as controls.

Materials and methods: Twenty-four maxillary central incisor crowns were fabricated in identical shape on metal testing dies in 3 groups: metal-ceramic crowns (MC, $n = 8$), veneered zirconia crowns (VZ, $n = 8$), and monolithic lithium disilicate crowns (MO, $n = 8$). A conservative preparation design with 0.75 mm lingual clearance was used for each crown system. All crowns were cemented to their corresponding crown preparations with self-adhesive resin cement (Multilink Automix). The crowns were subjected to 1000 cycles of thermal cycling, then cyclic loading of 111 N by means of a stainless steel ball, and 50,000 cycles of loading were applied for the fatigue test. Fatigue loading was followed by a continuously increasing compressive load, at a crosshead

* Corresponding author at: Rutgers School of Dental Medicine, Department of Restorative Dentistry, RSDM Rm D830110 Bergen Street, Newark, NJ 07101, USA.
  E-mail addresses: smorgano@bu.edu, smm519@sdm.rutgers.edu (S. M. Morgano).

speed of 1 mm/min until failure. The compressive load (N) required to cause failure was recorded. Means were calculated and analyzed with one-way ANOVA and the Tukey HSD test ($\alpha = .05$).

Results: There was a significant difference between MO vs. MC ($P = .0001$), MO vs. VZ ($P = .0001$), and VZ vs. MC ($P = .012$).

Conclusions: There was a significant difference in the mean fracture resistance of MC, VZ, and MO crowns in this in vitro study. The MC group recorded the highest mean fracture strength.

1. Introduction

Three fundamental criteria traditionally considered in the selection of materials for complete-coverage restorations include: strength, esthetics, and fit. Clinical longevity is a critical outcome measure related to these selection criteria (Abbate et al., 1989; Vahidi et al., 1991).

Metal-ceramic restorations have been available for over 50 years. During this period, substantial improvement in alloy substrates and veneering porcelains has resulted in widespread acceptance of metal-ceramic restorations, and continued research efforts have led to a more detailed, practical understanding of metal-ceramic systems (Kelly et al., 1996). Dental ceramic technology is a rapidly advancing area of dental materials research and development (Anusavice and Phillips, 2003). Dental ceramics have the potential to reproduce the depth of translucency, depth of color and texture of natural teeth (O'Brien et al., 1985).

Dental crowns fabricated as multilayered structures may have different stress distributions and load-bearing ability when compared with monolithic restorations. Therefore, differences in mechanical behavior and incidence of fracture could be expected (Santana et al., 2009). The ultimate aim of using all-ceramic systems is to provide crowns with sufficient mechanical strength to resist occlusal forces while maintaining excellent esthetics and biocompatible properties.

Metal-ceramic restorations are reputed to be the gold standard in dentistry, offering acceptable esthetics and long-term structural performance (Donovan, 2009; Napankangas and Raustia, 2008). Over the past 60 years, different designs and techniques for the fabrication of metal-ceramic restorations have been developed and improved (Warpeha and Goodkind, 1976; Shelby, 1962; Straussberg et al., 1966; Shoher and Whiteman, 1983; Brecker, 1956; Goodacre et al., 1977). Materials proposed as an alternative must be as reliable as metal-ceramics, particularly with regard to fracture rate and marginal adaptation (Pilathadka and Vahalova, 2007; Heintze and Rousson, 2010). A survey of several dental laboratories indicated that metal-ceramic restorations fabricated with high gold or high noble alloys are more expensive than zirconia-substructure crowns (Donovan, 2009).

Patients' demands for tooth-colored crowns without a metal substructure have driven substantial efforts toward increasing the strength and reliability of dental ceramic systems (Raigrodski, 2006). Fracture-strength studies of crown systems, within their limits, provide data relative to the load-bearing capacity of crowns in simulated clinical situations (Ku et al., 2002). Different in vitro and in vivo studies have been conducted, attempting to evaluate the reliability and fracture resistance of alternative dental ceramic systems and to define the factors that affect the longevity of these restorations. Al-Dohan et al. (2004) tested the shear bond strength of 4 veneering porcelains to the corresponding all-ceramic substructure materials, with metal-ceramic crowns serving as the control, and reported no statistically significant difference for 3 of the porcelain systems when compared to the control group. Coelho et al. (2009a,b) and Guess et al. (2009) fatigue tested zirconia-substructure, porcelain-veneered crowns using different testing methods, different veneering techniques, and different crown systems. Single load-to-failure tests conducted by Coelho et al. (2009a) resulted in fractures through the zirconia core. Fatigue testing that resembled occlusal loading (Coelho et al., 2009a,b) (Guess et al., 2009) resulted in surface damage to the zirconia veneering ceramics with chipping. To develop a clinically relevant testing method, a variety of clinically important variables, such as type of luting agent, bonding technique, the presence of water, substructure material, and preparation design must be considered (Friedlander et al., 1990; Kelly et al., 2010). Clinical follow-up of zirconia porcelain-veneered crowns has suggested a promising alternative to metal-ceramic crowns (Ortorp et al., 2009; Sailer et al., 2006, 2007).

Preservation of tooth structure is essential, especially for situations where gross removal of tooth structure would weaken the remaining tooth or violate the vitality of the dental pulp, for example teeth that are thin facio-lingually. When palatal clearance is limited, the use of a veneered zirconia-substructure crown with the palatal surface only in zirconia might be an option for all-ceramic crowns. Although the commonly recommended minimal thickness for a lithium disilicate monolithic crown is 1 mm, the absolute minimal allowable thickness of lithium disilicate monolithic crowns has not been studied scientifically as an option for certain clinical circumstances where palatal clearance is limited.

The aim of this study was to investigate the influence of (1) exposed lingual zirconia with zirconia porcelain-veneered crowns and (2) reduced lingual thickness of monolithic lithium disilicate crowns on fracture resistance of the crowns after thermal cycling and cyclic loading. The null hypothesis was that there would be no difference in the mean fracture resistance of zirconia veneered crowns with exposed lingual zirconia, monolithic lithium disilicate crowns, and metal-ceramic crowns with a metal lingual surface.

2. Materials and methods

2.1. Tooth preparation

A maxillary central incisor resin tooth (Ivorine tooth; Columbia Dentoform, Long Island City, NY, USA) was fixed in a plaster block, with the plaster 1 mm below the cemento-enamel

junction (CEJ). A crown preparation with a 6-degree convergence angle, 1.2-mm chamfer finish line on the facial and proximal surfaces, 0.75 mm chamfer finish line on the lingual surface, and an incisal reduction of 2 mm was manually prepared. A milling machine (SCHICK Dentalgeräte, Type S3, Schemmerhofen, Germany) ensured a standardized 6-degree convergence angle for the preparation. All sharp angles were rounded and cervical margins were located 1.0 mm above the CEJ (Fig. 1). Thirty metal dies were manufactured in chromium cobalt (Cr–Co) alloy (Cynoprod International, Beirut, Lebanon) by copy-milling the tooth preparation.

2.2. Crown fabrication

Three groups of specimens were constructed comprising 10 specimens within each group. Two of the specimens in each group were used for the pilot study, resulting in 8 specimens for each final experimental group (Table 1 and Fig. 2). For group VZ, complete-contour wax patterns (Scan wax for CEREC inLab modeling wax, Sirona Dental Systems, LLC, Charlotte, NC) were fabricated by using a silicone index. Dimensions of the wax patterns were 10.5 mm in length, 9 mm in width at the incisal portion and 8 mm in width at the cervical portion. Dimensions were measured with an electronic digital caliper (TRESNA, Guilin Guanglu Measuring Instrument Co., Ltd., Guangxi province, China). Wax patterns were cut back to provide space for porcelain on the facial, incisal and lingual surfaces, with the full contour of cingulum area and lingual fossa retained in wax. A 90-degree marginal finish line was designed in the wax pattern for the future junction of the zirconia and veneering porcelain at the junction of incisal and middle third on the lingual surface. The 0.5-mm thickness of the facial and incisal surfaces and 0.75-mm thickness of the lingual surfaces were measured with the use of a dental crown wax/metal dial caliper (TM Dental Lab, Inc., Staten Island, NY, USA). A dual-scan procedure was used to create and merge the data sets from the scanned die and the scanned wax pattern by using the Cerec InLab optical scanning unit (InEos, Sirona Dental Systems, LLC, Charlotte, NC, USA).

The zirconia copings were fabricated from partially-sintered zirconia blocks (MO 0/C15 IPS e.max ZirCAD, Ivoclar Vivadent AG, Schaan, Liechtenstein) with a uniform wall thickness of 0.5 mm except at the area apical to the finish line produced for the veneering ceramics. A virtual spacer layer of 30 μm was used. After the milling procedure, the copings were

Table 1 Groups of specimens used for property analysis.

Group	N
Veneered zirconia (VZ)	8
Metal-ceramic (MC)	8
Monolithic (MO)	8

sintered (Sirona inFire HTC sintering furnace, Sirona Dental Systems, LLC) and an overpressing technique was used to veneer the copings. First, ZirLiner (IPS e.max® ZirLiner, Ivoclar Vivadent AG) was applied and fired. Then the complete-contour wax pattern (Renfert GEO classic beige opaque, Renfert GmbH, Hilzingen, Germany) was developed on the coping. Following investing, the preheating and heating procedures were accomplished and porcelain ingots (LT A2 IPS e.max ZirPress, Ivoclar Vivadent) were used for overpressing. After divestment, the reaction layer was removed. The area to be ground was allowed to remain wet. Pressing sprues were removed with a fine diamond disk (911H 6911H, Komet Medical, Brasseler GmbH & Co., Lemgo, Germany) and the crowns were glazed (IPS e.max Ceram Glaze paste and liquid, Ivoclar Vivadent AG).

For group MC, the scanned and merged data sets from the group VZ dual-scan procedure were used to fabricate the patterns for the poly methyl methacrylate (PMMA) copings by using Cynoprod PMMA disks (Cynoprod International) with a uniform wall thickness of 0.5 mm on all surfaces except the lingual surface apical to the finish line for the veneering porcelain. The patterns were invested with special speed investment (Presto Vest II; SILADENT, Goslar, Germany). Lodestar alloy (Ivoclar Vivadent) was used for casting in an induction casting machine (RDO Induction. LLC, Model SUPER-CAST3, Washington, NJ, USA). The metal copings were finished by using tungsten carbide rotary instruments (EF fine s-cutters H295EF, Komet Medical, Brasseler GmbH & Co.) and evaluated on their dies under low magnification (4×) (Mantis compact microscope, Vision Engineering Ltd., Surrey, UK) for marginal accuracy. Copings were adjusted by using a silicone disclosing medium (Fit Checker II; GC Corporation, Tokyo, Japan) until the best possible fit was achieved. After airborne particle abrasion with 50-μm aluminum oxide and oxidation, 2 layers of opaque paste (IPS InLine/IPS InLine PoM Opaquer, Ivoclar Vivadent) were applied and fired. A fully anatomical wax pattern was made over each cast coping
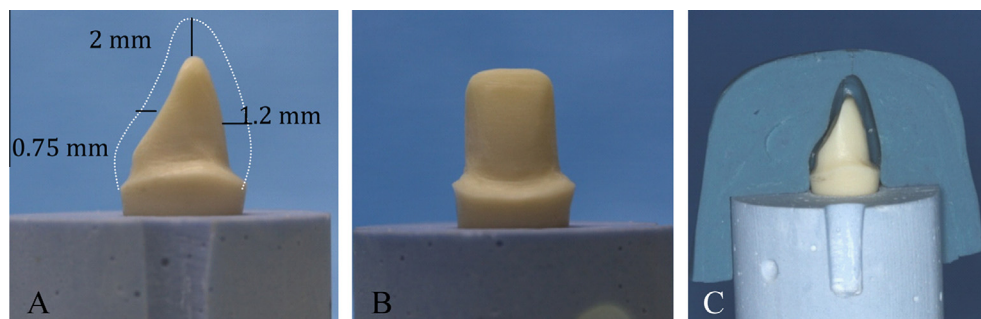
Figure 1 (A) Proximal view of tooth preparation with 2 mm of incisal reduction, 0.75 mm of lingual reduction and 1.2 mm of facial reduction. Chamfer finish line is deeper on the facial surface compared with the lingual chamfer; (B) Facial view of preparation; (C) Silicone index was used as a reference for preparation.

Figure 2 Three groups of specimens, from left: MO, VZ, and MC crowns cemented on the corresponding metal dies.

by using the silicone index made for the fabrication of the previous group of crowns.

Wax patterns were invested, and copings were overpressed with a special porcelain ingot (Small 2 IPS InLine PoM, Ivoclar Vivadent), which possessed the appropriate coefficient of thermal expansion with respect to the metal alloy substructure. The pressed crowns were adapted to the metal dies. Finally, the crowns were glazed. The lingual metal surface was polished after porcelain application by using gold polishing points (Silicone polisher #1101, 1102, 1103, Brasseler USA, Savannah, GA, USA) and buffing wheels (Felt Cloth, Flannel Cloth and Leather Buff Brasseler) and universal polishing paste (Ivoclar Vivadent AG).

For the fabrication of monolithic lithium disilicate crowns, complete-contour wax patterns (Renfert GEO classic beige opaque, Renfert GmbH) were developed on the metal dies according to the silicone index. Following investing, the preheating and heating procedures were accomplished. They were invested and pressed with a small ingot of special porcelain (LT A1 IPS e.max Press, Ivoclar Vivadent) and glazed. After glaze firing, all crowns were cemented onto their corresponding metal dies with self-adhesive universal resin cement (Multilink Automix, Ivoclar Vivadent, Amherst, NY).

2.3. Fracture testing

The crown-die assemblies were stored in distilled water at 37 °C for at least 48 h and then subjected to 1000 cycles of thermal cycling in a custom-made thermal cycling machine. Each 60-s-long cycle consisted of 15 s of time in 2 baths of 5 °C and 55 °C, with 2 transport times of 30 s between the 2 baths. After thermal cycling, each specimen was positioned in a testing jig with a chamber filled with distilled water at 37 °C. The crowns were subjected to wet cyclic loading in a custom-made cyclic loading machine (50,000 cycles) with 111 N by means of an 8-mm diameter stainless steel ball, applied at 135° to the long axis of the crown at a rate of 1 Hz. The load was applied to the lingual aspect of the specimens, 2.5 mm below the incisal edge.

The specimens that did not display bulk fracture were further tested with a load-to-failure test in a universal testing machine (Instron Model # 4202; Instron Corp, Canton, Mass) at a crosshead speed of 1 mm/min until failure. The compressive load (N) required to cause fracture was recorded for each specimen. To distribute the applied force over a larger area

and avoid loading stress peaks on the veneering material, a 1-mm thick layer of polyethylene vacuum-forming shell (Henry Schein, Port Washington, NY, USA) was placed between the piston and the crown. After loading in the universal testing machine, each test specimen was examined to determine the mode of failure.

Detailed data entry was performed with SPSS statistical software (Version 17.0.1, SPSS Inc., Chicago, Ill, USA) and Microsoft Excel Workbook 2007 (Microsoft Office, Redmond, Washington, USA) for analysis. The loads at fracture were registered. To evaluate statistical significance among the 3 groups (95% confidence interval), a one-way analysis of variance (ANOVA) was conducted to compare the means. The factors were the 3 different material combinations. Means and standard deviations were calculated for each group. To evaluate specific implications between individual groups, the Tukey HSD test was performed. All statistical testing was performed with $P < 0.05$ as the level of significance.

3. Results

None of the 24 specimens had any visible cracks or fractures after cyclic loading, and they all were loaded until failure occurred. The curve of failure load on the $x–y$ plot displayed an incremental increase until fracture occurred.

The means, SDs, minimal, and maximal fracture loads for the 3 groups are listed in Table 2. Mean fracture resistance for the control group, MC, was 9369 ± 734 N. Mean fracture resistance values for the experimental groups, MO and VZ, were 1650 ± 945 N and 2782 ± 283 N, respectively. The comparative bar graphs of the means and SDs for each group are presented in Fig. 3.

The one-way ANOVA indicated that there was a significant difference (= .0001) in fracture resistance among the 3 groups. The Tukey HSD post hoc test revealed a significant difference (= .0001) between the MO and MC, and the VZ and MC groups. The fracture resistance of MO and VZ groups also differed significantly (= .012) from each otherTable 3).

Table 2 Descriptive statistics of MO, VZ, and MC groups.

Group	Mean	SD	Minimum	Maximum
MO	1650.203	944.7801	597.492	3248.115
VZ	2782.148	283.0352	2450.425	3300.022
MC	9369.009	733.9286	8300.005	10,500.121
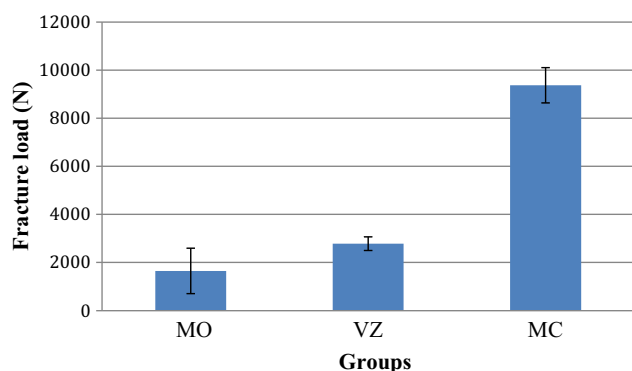
Figure 3 Mean fracture loads and SD of MO, VZ, and MC groups.

Table 3 Statistical analysis.

ANOVA	
$F = 275.9$	$P = .0001$
Post hoc test	
MO vs. MC	$P = .0001$
VZ vs. MC	$P = .0001$
MO vs. VZ	$P = .012$

There were substantial differences in the manner in which MO, VZ and MC crowns fractured. MO specimens fractured catastrophically, cracks displaying through the entire thickness of these units. These MO specimens shattered macroscopically into 2 or 3 pieces. VZ specimens demonstrated cracking of the zirconia substructure and chipping of the veneering layer. In some specimens, multiple cracks were visible through the veneering porcelain. The MC specimens demonstrated superficial cracking that affected only the ceramic layer, occasionally with a few chips breaking away.

4. Discussion

This study was conducted to compare the fracture resistance values of MC, VZ and MO crowns under simulated clinical conditions. The null hypothesis was rejected, because significant differences in fracture resistance values were found among the 3 groups. The mean fracture resistance of MC crowns was higher than that of MO and VZ crowns, while the fracture resistance of VZ crowns was higher than that of MO crowns.

Testing the fracture resistance of crowns is not a standard procedure, such as a bending test for a geometrically well-defined bar (Ku et al., 2002). In the current study, every effort was made to create clinically relevant, standardized and uniform specimens. Fracture-resistance studies after cyclic loading and thermal cycling in wet conditions, within their limits, can provide some indication of the load-bearing capacity of crowns under simulated clinical situations.

The mean failure loads for all the examined specimens were well above the incising forces normally exerted within the anterior region of the mouth, based on the assumption that mean masticatory force in the anterior region ranges from 89 to 111 N, with an added safety margin of 200 N (Santana et al., 2009). The VZ design modification in this study was based on what has been previously proposed for metal-ceramic

crowns (Warpeha and Goodkind, 1976; Shelby, 1962; Straussberg et al., 1966; Shoher and Whiteman, 1983; Brecker, 1956; Goodacre et al., 1977). However, such a thin metal substructure design had never been clinically evaluated because fracture of the metal substructure is unlikely as a result of its ductility, and survival rates are known to be high with such systems (Donovan, 2009).

Several factors, such as preparation design, crown thickness, method of luting, method of cyclic loading and thermal cycling can influence the results (Friedlander et al., 1990). Therefore, the results of different studies cannot be compared directly. In the present study, there was a significant difference between MC and VZ crowns, which can be explained by not veneering the lingual area of the VZ crowns. The lingual portion of anterior crowns is more influenced by the masticatory and habitual loads. When this portion of the crown was kept unveneered (i.e., the crack and fracture susceptible veneering porcelain was omitted), a significant difference in the failure mode of MC and VZ crowns was observed.

In the study by Quinn et al. (2010) the chips in both types of specimens detached once the veneer/substructure intersection was reached, and fracture did not continue into the substructure. Guess et al. (2009) demonstrated in their study that the cracks stopped at the veneer-core interface regardless of the veneering technique applied. Crack propagation into the zirconia substructure could not be observed in the study by Guess et al. (2009). However, in the present study the substructure itself was exposed to load and, in some of the VZ specimens, the failure did start from the zirconia with extension to the veneer/substructure intersection. In the MC specimens in the present study, the substructure was not susceptible to cracks and fractures; a very high measure of loading was transmitted to the facial, proximal and incisal veneering porcelain and cracks and chips occurred.

A significant difference was revealed between the mean fracture strengths of VZ and MO groups, and, in part, this can be related to the high crystalline content of the zirconia-based material that resulted in better mechanical properties. Small porosities and flaws in the microstructure of the pressed lithium disilicate crowns may be related to the fabrication process of the material or polishing methods, and may act as stress raisers leading to a catastrophic effect on the fracture resistance of these crowns. Also, it has been reported that etched and bonded ceramic crowns experience lower failure rates when compared with crowns luted with other types of cements and without an etching technique (Kelly et al., 2010). Lithium disilicate can be etched with 5% hydrofluoric acid applied for 20 s. Commonly the dental laboratory technician performs the etching protocol; however, this procedure can be accomplished by the dentist also. Because of the etchability of lithium disilicate, it achieves part of its strength from the micro-mechanical bonding with resin cement and the underlying tooth structure. Zirconia cannot be etched to achieve this strength from bonding (Pilathadka and Vahalova, 2007). However, a self-adhesive resin cement was used in the present study, and this cement can develop a bond to zirconia, although this bond is not as strong as a bond achieved from an etching technique. The lower load to failure values measured for the MO specimens compared with the VZ group could be attributed to the lack of bonding to the metal dies used in the current study. The airborne particle abrasion process used to retrieve the pressed-crowns in the VZ and MO groups could have damaged the zirconia

and lithium disilicate surfaces, which could have negatively influenced the mechanical properties of these 2 materials.

In the current study, lingual wall thickness for the MO crowns was less than the 1 mm recommended by the manufacturer. The fracture load for this group varied in the range of 597–3248 N with a mean of 1650 N. Although the mean of 1650 N may seem promising with the use of this material in reduced thickness for crowns, 2 specimens fractured at a load of 597 N and 874 N. Therefore, use of this material in reduced palatal thickness for patients with parafunctional habits may be a concern. Furthermore, because the specimens recorded a large SD and range of failure loads, predictable outcome for these crowns mainly depends on technique-sensitivity and handling of this material. One study reported that the load to cause failure from cracks increased as the square of the crown's thickness increased (Wolf et al., 2008). Therefore, the load to cause bulk fracture of a lithium disilicate crown can be expected to diminish rapidly as the thickness is lowered, so caution is advised if adequate tooth reduction is not provided. The MC crown was by far the strongest crown and can be recommended when clearance is limited.

Different studies have concluded that the weakest link in fully veneered zirconia crowns is the veneering porcelain (Al-Dohan et al., 2004; Coelho et al., 2009a,b; Ortorp et al., 2009; Fischer et al., 2009; Ashkanani et al., 2008). Along with attempts to improve the strength of the veneering porcelain, it would seem logical to avoid this material on more stress-prone areas, such as the lingual surfaces of the maxillary incisors.

Increased numbers of specimens could have reduced the influence of data variations on the statistical outcome. Furthermore, as with any in vitro study, it remains unclear as to what extent the results may be different in a clinical setting. Higher numbers of loading cycles may be required to represent longer service time.

Scanning electron microscopic (SEM) investigation of the initiation and propagation direction of the cracks and failures would have been beneficial in studying the association between the defect at the loading point and the crack or fracture lines. Use of finite element analysis (FEA) also would be helpful in investigations of stress distribution to evaluate the mechanical behavior of restorations.

In addition, studies are needed to determine the effect of the different polishing and glazing methods on crack propagation on the exposed zirconia surface. Highly polished zirconia is strongly recommended to prevent the wear of the opposing teeth. When the zirconia is exposed in the mouth, intra-oral adjustments of the occlusion are usually necessary. All 3 examined types of restorations seem to be reliable options in clinical conditions of restored teeth where minimally invasive preparations are indicated. The metal-ceramic crown is still the gold standard, considering its highly favorable mechanical properties.

5. Conclusion

This investigation was designed to evaluate clinically relevant fracture resistance of veneered zirconia crowns with exposed lingual zirconia, and the reduced lingual thickness in both zirconia veneered and monolithic lithium disilicate crowns, after thermal cycling and cyclic loading. Within the limitations of this in vitro study, the following could be concluded:

1. MO crowns exhibited lower mean fracture load compared with VZ and MC crowns ($P = .0001$).
2. VZ crowns showed lower mean fracture load compared with MC crowns ($P = .012$).
3. MC crowns exhibited a mean fracture load considerably higher than the load documented for that region of the mouth in vivo.

Ethical statement

The manuscript was prepared following the policies and principles of the Committee on Publication Ethics (COPE).

Conflict of interest

There are no conflicts of interest to declare.

Acknowledgment

References

Abbate, M.F., Tjan, A.H., Fox, W.M., 1989. Comparison of the marginal fit of various ceramic crown systems. J. Prosthet. Dent. 61, 527–531.

Al-Dohan, H.M., Yaman, P., Dennison, J.B., Razzoog, M.E., Lang, B.R., 2004. Shear strength of core-veneer interface in bi-layered ceramics. J. Prosthet. Dent. 91, 349–355.

Anusavice, K.J., Phillips, R.W., 2003. Phillip's Science of Dental Materials, 11th ed. WB Saunders, St. Louis, pp. 656–660.

Ashkanani, H.M., Raigrodski, A.J., Flinn, B.D., Heindl, H., Mancl, L.A., 2008. Flexural and shear strengths of ZrO_2 and a high-noble alloy bonded to their corresponding porcelains. J. Prosthet. Dent. 100, 274–284.

Brecker, S.C., 1956. Porcelain baked to gold – a new medium in prosthodontics. J. Prosthet. Dent. 6, 801–810.

Coelho, P.G., Silva, N.R., Bonfante, E.A., Guess, P.C., Rekow, E.D., Thompson, V.P., 2009a. Fatigue testing of two porcelain-zirconia all-ceramic crown systems. Dent. Mater. 25, 1122–1127.

Coelho, P.G., Bonfante, E.A., Silva, N.R., Rekow, E.D., Thompson, V.P., 2009b. Laboratory simulation of Y-TZP all-ceramic crown clinical failures. J. Dent. Res. 88, 382–386.

Donovan, T.E., 2009. Porcelain-fused-to-metal (PFM) alternatives. J. Esthet. Restor. Dent. 21, 4–6.

Goodacre, C.J., Van Roekel, N.B., Dykema, R.W., Ullmann, R.B., 1977. The collarless metal-ceramic crown. J. Prosthet. Dent. 38, 615–622.

Guess, P.C., Zhang, Y., Thompson, V.P., 2009. Effect of veneering techniques on damage and reliability of Y-TZP trilayers. Eur. J. Esthet. Dent. 4, 262–276.

Fischer, J., Stawarzcyk, B., Trottmann, A., Hammerle, C.H., 2009. Impact of thermal misfit on shear strength of veneering ceramic/zirconia composites. Dent. Mater. 25, 419–423.

Friedlander, L.D., Munoz, C.A., Goodacre, C.J., Doyle, M.G., Moore, B.K., 1990. The effect of tooth preparation design on the breaking strength of dicor crowns: Part 1. Int. J. Prosthodont. 3, 159–168.

Heintze, S.D., Rousson, V., 2010. Fracture rates of IPS Empress all-ceramic crowns – a systematic review. Int. J. Prosthodont. 23, 129–133.

Kelly, J.R., Nishimura, I., Campbell, S.D., 1996. Ceramics in dentistry: historical roots and current perspectives. J. Prosthet. Dent. 75, 18–32.

Kelly, J.R., Rungruanganunt, P., Hunter, B., Vailati, F., 2010. Development of a clinically validated bulk failure test for ceramic crowns. J. Prosthet. Dent. 104, 228–238.

Ku, C.W., Park, S.W., Yang, H.S., 2002. Comparison of the fracture strengths of metal-ceramic crowns and three ceromer crowns. J. Prosthet. Dent. 88, 170–175.

Napankangas, R., Raustia, A., 2008. Twenty-year follow-up of metal-ceramic single crowns: a retrospective study. Int. J. Prosthodont. 21, 307–311.

O'Brien, W.J., Johnston, W.M., Fanian, F., 1985. Double-layer color effects in porcelain systems. J. Dent. Res. 64, 940–943.

Ortorp, A., Kihl, M.L., Carlsson, G.E., 2009. A 3-year retrospective and clinical follow-up study of zirconia single crowns performed in a private practice. J. Dent. 37, 731–736.

Pilathadka, S., Vahalova, D., 2007. Contemporary all-ceramic systems, part-2. Acta Med. (Hradec Kralove) 50, 105–107.

Quinn, J.B., Sundar, V., Parry, E.E., Quinn, G.D., 2010. Comparison of edge chipping resistance of PFM and veneered zirconia specimens. Dent. Mater. 26, 13–20.

Raigrodski, A.J., 2006. Materials for all-ceramic restorations. J. Esthet. Restor. Dent. 18, 117–118.

Santana, T., Zhang, Y., Guess, P., Thompson, V.P., Rekow, E.D., Silva, N.R., 2009. Off-axis sliding contact reliability and failure modes of veneered alumina and zirconia. Dent. Mater. 25, 892–898.

Sailer, I., Feher, A., Filser, F., Luthy, H., Gauckler, L.J., Scharer, P., Franz Hammerle, C.H., 2006. Prospective clinical study of zirconia posterior fixed partial dentures: 3-year follow-up. Quintessence Int. 37, 685–693.

Sailer, I., Feher, A., Filser, F., Gauckler, L.J., Luthy, H., Hammerle, C.H., 2007. Five-year clinical results of zirconia frameworks for posterior fixed partial dentures. Int. J. Prosthodont. 20, 383–388.

Shelby, D.S., 1962. Practical considerations and design of porcelain fused to metal. J. Prosthet. Dent. 12, 542–548.

Shoher, I., Whiteman, A.E., 1983. Reinforced porcelain system: a new concept in ceramometal restorations. J. Prosthet. Dent. 50, 489–496.

Straussberg, G., Katz, G., Kuwata, M., 1966. Design of gold supporting structures for fused porcelain restorations. J. Prosthet. Dent. 16, 928–936.

Vahidi, F., Egloff, E.T., Panno, F.V., 1991. Evaluation of marginal adaptation of all-ceramic crowns and metal ceramic crowns. J. Prosthet. Dent. 66, 426–431.

Warpeha Jr., W.S., Goodkind, R.J., 1976. Design and technique variables affecting fracture resistance of metal-ceramic restorations. J. Prosthet. Dent. 35, 291–298.

Wolf, D., Bindl, A., Schmidlin, P.R., Luthy, H., Mormann, W.H., 2008. Strength of CAD/CAM-generated esthetic ceramic molar implant crowns. Int. J. Oral Maxillofac. Implants 23, 609–617.