Effect of Sintering Speed and Polishing or Glazing on the Failure Load of Monolithic Zirconia Fixed Partial Dentures

Ali Hafezeqoran, DDS, MSc
Department of Prosthodontics, Faculty of Dentistry, Tabriz University of Medical Sciences, Tabriz, Iran.

Pouya Sabanik, DDS
Private practice, Tabriz, Iran.

Roodabeh Koodaryan, DDS, MSc
Department of Prosthodontics, Faculty of Dentistry, Tabriz University of Medical Sciences, Tabriz, Iran.

Purpose: To evaluate the effect of sintering speed and polishing or glazing on the failure load (FL) of monolithic zirconia fixed partial dentures (FPDs). Materials and Methods: A total of 40 three-unit FPDs extending from the mandibular first premolar to the first molar were evaluated. The prepared typodont teeth were scanned, and the prostheses were designed. Afterwards, the prostheses were milled from monolithic zirconia blanks. The samples were divided into classic and speed sintering groups (n = 20 each). Half of the samples in each group (n = 10) were polished with an electric handpiece according to the manufacturer’s instructions, and the other half (n = 10) were glazed. All of the samples were thermocycled for 3,500 cycles between 5°C and 55°C in water baths. The FL was calculated in Newtons with the three-point bending test. Results: The mean ± SD FL values were as follows: classic sintering/polished group = 2,026.5 ± 172.8 N; classic sintering/glazed group = 1,917.58 ± 174.45 N; speed sintering/polished group = 1,787.58 ± 145.81 N; and speed sintering/glazed group = 1,719.6 ± 143.9 N. There was a significant difference in the mean FL between the two sintering methods (P < .001), with the classic sintering group exhibiting the highest FL. Conclusion: Classic sintering of the monolithic zirconia FPDs led to the maximum amount of FL and strength.

Yttrium-stabilized polycrystalline zirconia (Y-TZP) has been introduced to dental science over the last few decades. Y-TZP requires veneering with glass-ceramics or feldspathic porcelain for fulfilling esthetic purposes.1–3 Zirconia has been used in different clinical situations because of its esthetic potential, biocompatibility, dimensional and chemical stability, and high fracture resistance compared to other dental ceramics.4–6 However, there are some clinical problems when using Y-TZPs. Because of their opacity, they need to be veneered, and the veneered layer is prone to cohesive fracture.2,7–9

High-translucency monolithic zirconia restorations were developed by adding different impurities and alterations in different zirconia phase ratios to eliminate the veneer layer. However, the impact of these alterations on the mechanical properties and performance of zirconia is not yet completely understood. Monolithic zirconia has an easier manufacturing process compared to veneered zirconia and allows for reduced restoration thickness and tooth preparation (only half a millimeter for posterior restorations) at a lower cost.10,11

According to results in the literature, zirconia is a suitable substructure ceramic for long FPDs in stress-bearing areas.11 According to previous reports, different zirconia sintering parameters significantly affect zirconia restorations. Changes in sintering parameters such as final temperature, dwelling time, and total sintering time can...
affect the particle size, translucency, and biaxial flexural strength of translucent zirconia restorations.\textsuperscript{12} The first word that may come to mind when thinking about the impact of digital technologies is “speed.” Speed sintering protocols have been developed in response to the increasing demand for time- and cost-effective chairside one-visit restorations, and, in some cases, for preventing grain growth for better optical properties.\textsuperscript{13,14}

Although CAD/CAM technology has been an effective advancement in dentistry, the adjustment of a restoration in the patient’s mouth is inevitable. This adjustment could be on the occlusal and proximal surfaces due to premature contacts or the buccal and lingual surfaces due to soft tissue trauma.\textsuperscript{15,16} Because of the zirconia surface hardness, special diamond burs are used for this process, which render the surface rough. Moreover, the restoration’s rough surface can cause severe abrasion on the opposite tooth enamel or restoration, and it can even cause dental plaque adhesion or stain formation.\textsuperscript{17,18} Glazing and polishing are known as two ways to smooth the surface of zirconia restorations. Although glazing is one of the most common methods for creating a very smooth surface in restorations, it has been reported that, after 6 months, the glazed layer might be compromised, exposing the underlying zirconia.\textsuperscript{19} Intraoral polishing systems are considered one of the best solutions for re-smoothing the restoration surface to prevent abrasion of the opposite tooth. Polishing can also improve the esthetics and longevity of restorations.\textsuperscript{20,21}

Few studies have evaluated the effects of sintering speed (interactions between changing the sintering temperature, time, and heating conditions) on the biaxial flexural strength and the failure load (FL) of monolithic translucent zirconia regarding the effect on its long-term performance.\textsuperscript{22} Although many studies have evaluated porcelain polishing, there is little information about the effects of this process on zirconia restorations.\textsuperscript{16,17,23,24} Therefore, this study aimed to investigate the effect of sintering speed (classic vs speed sintering cycles) and polishing vs glazing on the FL of monolithic zirconia FPDs. This study hypothesized that classic sintering vs speed sintering and polishing vs glazing would have no significant effect on the FL of FPDs.

**MATERIALS AND METHODS**

Abutments (mandibular first premolar and first molar) were prepared on a standard typodont model (Prosthetic Restoration Jaw Model, PRO2001-UL-HD-M-32, Nissin) with 6-mm occlusocervical height, 1-mm–depth chamfer finish line, and a 6-degree convergence angle. Afterwards, all the line angles and point angles were rounded off, and occlusal grooves were simulated.

To perform the FL test, an abutment model that is the restoration is needed. Accordingly, an impression was made from the master model with additional silicone (Panasil Putty Fast, Kettenbach) impression material and then poured with an acrylic resin pattern (Pattern Resin, GC). The acrylic model was invested and cast using a base metal alloy to fabricate the metal testing model (Rexillium III, Pentron). The metal model was finished with carbide finishing burs and polished with the special nickel-chromium polishing stones (Dura-Green, Shofu). Marginal integrity and the possibility of distortion in the metal testing model were checked under a stereomicroscope (Stereoscopic Zoom Microscope, Nikon) at ×20 magnification.

The prepared abutments were scanned (Rainbow Scanning, Dentium), and FPDs were designed with a thickness of 1 mm in the central groove and axial walls using CAD software (DentalCAD, Exocad). The pontics measured 8 mm in buccolingual width. The connectors measured 3 × 3 mm², and the cement gap was set to 45 μm.

The designed FPDs were milled from four monolithic pre-shaded (A3 color) zirconia blanks with a height of 22 mm (Lot 2015407195-1047, Lot 2015407195-1046, Lot 2015407196-1005, and Lot 2015407196-1001; inCo-ris TZI C milling blocks, Dentsply Sirona) using a dental lab milling unit (CORITEC 350i, Imes-icore). After the milling process, all of the FPDs were separated from the blanks and air-cleaned to prevent any milling residues according to the manufacturer’s recommendations. The specimens were then randomly divided into two groups: (1) a classic sintering group and (2) a speed sintering group (20 FPDs for each group). According to the manufacturer’s instructions, restorations were sintered in a furnace (ZYrcomate T, Vita Zahnfabrik). The procedures in the first group consisted of heating at 25°C/minute to 800°C, then at 15°C/minute to the final temperature (1,510°C), dwelling for 2 hours, then cooling to 200°C with the firing chamber closed. In the second group, the speed sintering protocol was carried out: heating at 99°C/minute to 1,100°C, then at 50°C/minute to the final temperature (1,510°C), dwelling for 30 minutes, then cooling at 99°C/minute down to 800°C, dwelling for 5 minutes, and then to 25% of the highest temperature (400°C). The total sintering time was 61 minutes. Finally, the marginal integrity was evaluated using a condensation light-body silicone (Speedex Light Body, Coltene).

In the next stage, each group was randomly divided into glazing or polishing subgroups (10 FPDs for each). In the polishing subgroups, 10 FPDs were polished with a zirconia polishing system (Disasynt Plus and Diacera, Eve Ernst Vetter) with the following steps, based on the manufacturer’s instructions: prepolishing (coarse grit) and polishing (medium grit). The specimens were mounted on a polishing jig. Polishing was performed for each polisher in one direction for 20 seconds (using...
a timer) with water and air cooling with a low-speed micromotor (MX2 LED, Bien-Air Dental) with a 1:1 handpiece (Paris U 1:1 LUX, Eur-Med) at 10,000 rpm. In the glazing subgroups, the FPDs were glazed, and the glaze powder was mixed with the glaze liquid (Akzent Plus Glaze Powder, Vita Zahnfabrik) and then applied in a thin layer using a ceramic brush, followed by firing in the ceramic furnace (Programat P310, Ivolcar Vivadent) according to the manufacturer’s recommended glaze firing cycle. Accordingly, the time for increasing temperature was 11.37 minutes, and the final temperature was 950°C; eventually, the holding time at the final temperature was 1 minute.

All of the procedures were performed by a single operator in this study to reduce variability and simulate the clinical procedure (P.S.).

Thermocycling was performed for all FPDs to stimulate the clinical aging process. All 40 samples were subjected to 3,500 thermal cycles (THE-1100, SD Mechatronik) between 5°C and 55°C in water baths for 30 seconds each, with a dwell time of 10 seconds.

Finally, all the FPDs were cleaned with a mixture of distilled water and acetone in an ultrasonic cleaner (Sonic 6MX, James Products) for 5 minutes and then rinsed with distilled water and air-dried.

All FPDs were subjected to a three-point bending test in a universal testing machine (H25KS, Hounsfield). The specimens were mounted on the metal testing model and loaded at 0.5 mm/minute at the pontic center with a 6-mm–diameter steel rod until fracture occurred. The maximum load was immediately recorded after a sudden decrease in force as the FL (in Newtons). All of the measurements were conducted at room temperature (23°C), and after subjecting each sample to a three-point bending test, the testing model was examined for any distortion under a stereomicroscope (Stereoscopic Zoom Microscope) at ×20 magnification.

Statistical analyses were conducted using SPSS version 17 (IBM). Descriptive statistics (mean ± SD) and Kolmogorov-Smirnov test were calculated for the measured parameters. FL values were analyzed using two-way analysis of variance (ANOVA) at a significance level of P < .05.

RESULTS

The mean and SD FL values are presented in Table 1. The statistical analysis confirmed that the highest FL values were recorded in the classic sintering group, and this difference was statistically significant (P < .001). There was no significant difference in FL values between the polished and glazed subgroups (P > .05). The data indicated no interaction between sintering speed and glazing or polishing; the effect of sintering speed on the mean FL in both the glazing and polishing subgroups was not statistically significant (P > .05).

DISCUSSION

The null hypothesis can be partially rejected based on the results because the sintering speed affected the FL of monolithic zirconia FPDs. However, polishing vs glazing had no significant effect on the FL. The FL was found to be significantly higher in the classic sintering cycle compared to the speed sintering cycle, which is consistent with previous reports.12,22 Some rational arguments for this finding have been presented. Speed sintering cycles provoke more nonhomogenous grain patterns than classic sintering cycles, and a homogenous grain pattern generates a dense zirconia restoration with fewer pores.25,26 Also, classic sintering provokes larger grain size composition, and this polycrystalline structure allows for coalescence and growth between the grains.27–30

Guazzato et al31 reported that new upgrades in the zirconia ceramics’ flexural strength were related to the growth in the amount of the monoclinic phase. In addition, surface treatments, sintering conditions, and tensile stresses could trigger tetragonal to monoclinic phase transformation.32,33 In this regard, the sintering temperature is the crucial one, altering the phase composition and microstructure of zirconia. Sintering protocols that intensify heating and cooling temperatures can cause tensile stresses on the polycrystalline structure of zirconia.8,34 Accordingly, alteration in sintering parameters can cause phase transformation and tensile stress simultaneously. Sintering the zirconia at a higher temperature impacts the grain size and yttria, and the uneven distribution of yttria could increase the cubic phase, which has negative influences on flexural strength. Stawarczyk et al12 concluded that zirconia had the highest flexural strength at the final sintering temperatures below 1,550°C.

Table 1 Mean (SD) Failure Load Values for Each Group

<table>
<thead>
<tr>
<th>Sintering protocol</th>
<th>Surface treatment</th>
<th>Failure load, N</th>
<th>Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Classic</td>
<td>Glazed</td>
<td>10</td>
<td>1,917.58 (174.45)</td>
</tr>
<tr>
<td></td>
<td>Polished</td>
<td>10</td>
<td>2,026.52 (172.82)</td>
</tr>
<tr>
<td></td>
<td>Total</td>
<td>20</td>
<td>1,966.00 (177.52)</td>
</tr>
<tr>
<td>Speed</td>
<td>Glazed</td>
<td>10</td>
<td>1,719.64 (143.99)</td>
</tr>
<tr>
<td></td>
<td>Polished</td>
<td>10</td>
<td>1,787.58 (145.81)</td>
</tr>
</tbody>
</table>

*Two-way ANOVA showed no interaction between sintering speed and surface treatment (P = .700).

*Two-way ANOVA showed a significant difference in failure load between the two sintering protocols (P < .001).

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On the other hand, some studies illustrate that diversity in sintering parameters had no impact on the FL or flexural strength of zirconia. These variations in the results might be attributed to different zirconia brands, test conditions, specimen dimensions, and sintering parameters in these studies.

Y-TZP demonstrates tetragonal to monoclinic phase transformation under polishing stress, which increases the FL and flexural strength. It is also important to note that the fewer surface defects on zirconia, the higher the resistance of zirconia. According to Mai et al., the three-step zirconia polishing system's application had no significant effect on the zirconia surface roughness compared to applying the two-step zirconia polishing system.

Kumchai et al. concluded that glazed specimens had significantly lower flexural strength than unglazed ones with the same zirconia brand and that the strength reduction in the glazed zirconia might be due to the glazing material rather than the glaze heating cycle. Similarly, Yener et al reported that the glazing materials could reduce the flexural strength of Ceramill, Cercon, and Zirkonzahn zirconia brands. It was believed that glazing materials decrease the depths of the surface cracks and increase the strength of ceramic restorations. However, this effect of glazing is still ambiguous.

Excessive residual stress in ceramics, along with the addition of other stresses in the oral environment, would provoke crack formation and propagation in the zirconia restoration, reducing the strength. The residual stress can be caused by a thermal expansion mismatch during the firing process and tempering stresses associated with temperature gradients during cooling. The coefficient of thermal expansion (CTE) of tetragonal zirconia is higher than the CTE of monoclinic zirconia; therefore, the CTE of polished zirconia might depend on the degree of phase transformation caused by the polishing procedure. The CTE of porcelain changes during firing. Since the glazing materials have porcelain powder inside, the CTE might change in glazing material during firing, and glaze application also changes the CTE of zirconia restorations. High residual tempering stresses result from high-temperature differences through the specimens; thus, using a speed sintering protocol for ceramic is considered to be a positive movement in this direction. However, in dental arch areas requiring high-strength restorations, it is better to use the classic sintering protocol to achieve maximum strength of zirconia restorations.

CONCLUSIONS

Both sintering protocols had acceptable FL for mastication loads. Fast dental procedures are the most demanding; thus, using a speed sintering protocol for ceramic is considered to be a positive movement in this direction. However, in dental arch areas requiring high-strength restorations, it is better to use the classic sintering protocol to achieve maximum strength of zirconia restorations.

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REFERENCES


