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Comparative analysis on intaglio surface trueness, wear volume loss of antagonist, and fracture resistance of full-contour monolithic zirconia crown for single-visit dentistry under simulated mastication

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PURPOSE. This analysis aimed to evaluate the intaglio surface trueness, antagonist's wear volume loss, and fracture resistance of full-contour crowns of (Y, Nb)-stabilized fully-sintered zirconia (FSZ), 4 mol% or 5 mol% yttria-stabilized partially sintered zirconia (4YZ or 5YZ) with high-speed sintering. **MATERIALS AND METHODS.** A total of 42 zirconia crowns were separated into three groups: FSZ, 4YZ, and 5YZ (n = 14). The intaglio surface trueness of the crowns was evaluated at the inner surface, occlusal, margin, and axial areas and reported as root-mean-square, positive and negative average deviation. Half of the specimens were aged for 120,000 cycles in the chewing simulator, and the wear volume loss of antagonist was measured. Before and after chewing, the fracture load was measured for each group. The trueness values were analyzed with Welch's ANOVA, and the wear volume loss with the Kruskal-Wallis tests. Effect of the zirconia type and aging on fracture resistance of crowns was tested using two-way ANOVA. **RESULTS.** The intaglio surface trueness measured at four different areas of the crown was less than 50 µm, regardless of the type of zirconia. No significant P in wear volume loss of antagonists were detected among the groups ($P > .05$). Both the type of zirconia and aging showed statistically significant effects on fracture resistance ($P < .05$). **CONCLUSION.** The full-contour crowns of FSZ as well as 4YZ or 5YZ with high-speed sintering were clinically acceptable, in terms of intaglio surface trueness, antagonist's wear volume loss, and fracture resistance after simulated mastication. [J Adv Prosthodont 2022;14:173-81]

KEYWORDS
Intaglio surface trueness; Wear volume loss; Fracture resistance; Fully-sintered zirconia; High-speed sintering

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INTRODUCTION

Recently, zirconia ceramics are currently widely used as the materials of choice for prosthodontic treatment, expanding their applications along with the advances of digital technology in clinical dentistry.[1] It has excellent mechanical properties, tooth-like aesthetics in case of increased yttria content, and has been optimized for computer-aided design and manufacturing for many years.[1,2] Generally, in order to reduce the wear of milling tools and chances of phase transformation, dental zirconia is used for milling as in the form of a pre-sintered or partially-stabilized block and produced as white- stage products that require further sintering to final density.[3] According to the conventional protocol, the required sintering time is in the range from 6 to 12 hours to complete. Since it is a time-consuming process, the reduction in sintering time within 90 minutes, so-called fast sintering or high-speed sintering, has recently been focused as a possible alternative pathway.[4-6] High-speed sintering may not significantly change the dimensional accuracy of the milled prostheses, while it could decrease the translucency of dental zirconia.[4,5] It reduced the amount of surface wear of sintered objects, revealed from the two-body wear simulation.[6] The fracture resistance of high-speed sintered product remained consistent or slightly increased compared to those of the products from conventional protocol.[6] The sintering temperature, holding time, and total sintering time may change the optical and mechanical properties of dental zirconia.[7]

As an alternative to changing sintering protocol, the fully-sintered zirconia blocks could be answers for saving production time and the pathway to single-visit dentistry.[8] In comparison to partially-sintered zirconia prostheses, fully-sintered zirconia prosthesis showed decreased volume proportion of pores, increased strength, and precise fit.[9] However, due to high surface hardness of conventional fully-sintered zirconia blocks, the increased wear of machining tool was reported to be the problem.[8,10] To resolve these concerns, a fully-sintered (Y, Nb)-TZP (tetragonal zirconia polycrystal) block with low surface hardness has recently been developed for clinical use. It has excellent machining capability and dimensional accuracy. It also minimizes the chances of phase deformation during fabrication by the presence of stabilizing dopants such as Nb.[11,12]

Numerous factors should be considered in order to produce zirconia-based dental prostheses with long-term clinical success. Proper seating of dental prosthesis is important and it depends on the dimensional accuracy of the intaglio surfaces.[13,14] Failure to proper seating can result in thick cementation layers, creating residual stresses on the tensile surface due to viscoelastic deformation during cyclic loading.[15] The marginal area of fixed prosthesis is one of the most frequent source of prosthetic complication, closely related to plaque retention, secondary caries and periodontal disease in case of misfit.[16-19] To minimize marginal discrepancy, evaluation of intaglio trueness of dental prosthesis is important in terms of manufacturing accuracy.[20]

The fracture resistance of dental prosthesis is an important factor in clinical situation where occlusal forces should be tolerated.[21] The fracture resistance of all-ceramic crown is reported to be affected by the crown thickness, crown shape, and axial wall height affect.[22,23] Moreover, clinical failure of all-ceramic restoration is a multifactorial event that includes patient characteristics, masticatory loading, material properties, and fatigue phenomena.[24] 3 mol% yttria-stabilized tetragonal zirconia polycrystal (3Y-TZP) has high fracture toughness and flexural strength due to transformation toughening.[25,26] However, 4 mol% (or 5 mol%) yttria-stabilized partially sintered zirconia (4Y- or 5Y-PSZ) has higher proportion of cubic phase than 3Y-TZP, which has reduced martensitic transformation ability and decreased mechanical properties.[25]

Considering the clinical situations, the performance of the prosthesis should be examined under cyclic loading and hydrothermal aging.[27] The wear of occlusal surfaces of the prosthesis also needs to be simulated in the physiologic environment.[6] Fatigue failure is described as the formation of subcritical cracks in pre-existing flaws in ceramic materials as a result of intermittent stress below the material's typical fracture strength in an aqueous environment.[28] Zirconia ceramics are known as susceptible to hydrothermal aging, a process in which water molecules cross through the grain boundaries and cause phase transformation, resulting in aggravation of mechanical properties.[29,30] The consequences of hydrothermal aging are surface roughening, enhanced wear rates, reduced mechanical properties and even catastrophic failure.[31,32]

The purpose of this study is to evaluate the intaglio trueness, wear volume loss of antagonistic teeth, and fracture resistance of full-contour monolithic crowns, fabricated using fully-sintered (Y, Nb)-TZP and partially-sintered 4Y-PSZ (or 5Y-PSZ) with high-speed sintering (90-minute protocol). The first null hypothesis was that no difference would be found in terms of intaglio trueness of the crowns, regardless of the type of zirconia. The second null hypothesis was that the wear volume loss of antagonistic teeth would be the same, irrespective of the type of zirconia. The third null hypothesis was that no difference would be found in the fracture resistance among the different zirconia groups, regardless of artificial aging.

MATERIALS AND METHODS

Considering the recommended tooth preparation for full-contour monolithic crown (occlusal area: 1.5 mm, axial surface: 1 - 1.5 mm, chamfer margin width: 1 mm, rounded line angle), the mandibular first molar was prepared as an abutment using standardized acrylic resin teeth (Simple Root Tooth Model; Nissin Dental Products, Kyoto, Japan). The prepared resin tooth abutment was digitized with an 7 µm-accuracy laboratory scanner (T500; Medit, Seoul, Korea). A total of 42 metal abutments were fabricated using Co-Cr alloy powders (SP2 CoCr; Eos GmbH, Krailling, Germany) and the direct metal laser sintering method (EOSINT M270; Eos GmbH, Krailling, Germany). A full-contour crown was virtually designed on the prepared abutment using a dental CAD software (Dental Designer; 3Shape, Copenhagen, Denmark), with a cementation layer thickness of 25 µm, beginning at a distance of 1 mm from the chamfer margin.

Fourteen monolithic crowns were fabricated for each of the following groups, and the number of specimens per each group was referred to the previous studies.[33,34] First, the fully sintered (Y,Nb)-TZP crowns (FSZ group, Perfit FS; Vatech mcis, Gyeonggi-do, Korea) were manufactured with a CAM software (Millbox; CIMsystem, Milano, Italy) and a 4-axis wet milling machine (Cori TEC one; imes-icore, Eiterfeld, Germany). Second, the 4Y-PSZ crowns (4YZ group, KATANA STML; Kuraray Noritake Dental Inc., Tokyo, Japan) were manufactured with a CAM software (HyperDent; Follow-me! Technology, Munich, Germany) and a 5-axis milling machine (Arum 5X-300; Doowon, Daejeon, Korea). After milling, the crowns were subjected to the 90-minute high-speed sintering protocol suggested by the manufacturer. The details of the firing schedule were as follows: heating rate of 35°C/min for 45 minutes, holding at 1560 - 1565°C for 30 minutes, and cooling rate of 45°C/min for 15 minutes. Third, the 5Y-PSZ crowns (5YZ group, KATANA UTML; Kuraray Noritake Dental Inc., Tokyo, Japan) were manufactured with the same milling protocol and sintering schedule as the 4YZ group.

To calculate the intaglio surface trueness of all zirconia crowns, the surfaces were scanned using an intraoral scanner (i500; Medit, Seoul, Korea) with an *in vitro* precision of 5 µm. Because the digitization using a laboratory scanner was partly limited by the scanning angle and distance to the intaglio surface of the crown, the intraoral scanner was utilized for the experiment after calibration according to the manufacturer's instructions. The scanning process was performed in the absence of ambient light and the calibration was performed at each stage of the specimen scanning procedure. Each scan data of the intaglio surface of the zirconia crown was superimposed on the internal area of the virtual crown from the CAD file (reference data) using an inspection software (Geomagic Control X; Geomagic Inc., Morrisville, NC, USA). After defining the internal area of the virtual crown as the region of interest, the best-fit alignment was performed using an iterative closest point algorithm, based on point-to-point measurements. The root-mean-square (RMS), positive average deviation (+AVG), and negative average deviation (-AVG) values were determined between the intaglio surface scan data and reference data for each of the four inspection regions: inner surface, occlusal, margin, and axial area.

The half of the zirconia crowns of each group (n = 7) were cemented on the metal abutments with a self-curing resin cement (RelyX-U200; 3M-ESPE, St.

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Paul, MN, USA), for the thermal cycling and mechanical loading in a chewing simulator (Chewing Simulator CS-4.8; SD Mechatronik, Feldkirchen-Westerham, Germany). The disto-palatal cusp of the human maxillary molar was adjusted to a spherical shape using a diamond bar (Dia-Burs EX-21F; MANI, Tochigi, Japan) and embedded in self-curing acrylic resin (Vertex Self-Curing; Vertex Dental, Soesterberg, Netherlands), serving as an enamel antagonist. The crown-abutment specimen was loaded with 50 N at 1.5 Hz for 120,000 cycles as 6-month clinical service.[35,36] Simultaneous thermal cycling was performed in water by changing the temperature from 5°C to 55°C. Additionally, a 0.7 mm sliding movement from the central fossa to the buccal cusp was performed to simulate physiological mastication. The thermal cycling and mechanical loading parameters employed in this study are listed in Table 1. To assess surface wear, the occlusal surface of each antagonist tooth was digitized before and after the chewing cycle using a laboratory scanner (T500; Medit, Seoul, Korea). Using an inspection software (Geomagic Control X; Geomagic Inc., Morrisville, NC, USA), the scan data before and after cyclic loading were superimposed to compute the wear volume loss (mm^3).

For the fracture resistance evaluation, the artificially aged crown-abutment specimens were mounted on the universal testing machine (Instron 8871; Instron, Norwood, MA, USA) to perform the load-to-failure test using a 10 kN load cell. A 5 mm diameter stainless-steel spherical indenter was used to make contact with the central fossa of each crown. A 2 mm urethane sheet was placed on its occlusal surface to evenly distribute the load. Compressive loading was carried out at 0.5 mm/min crosshead speed until catastrophic fracture occurred, at which point the load (N) at failure was recorded. The crowns not artificially aged for each group were also assessed for the fracture resistance in the same manner described above.

For the statistical analysis, the means and standard deviations of the intaglio surface trueness measurements (µm, RMS, +AVG, and -AVG), fracture load values (N), and the volume loss (mm^3) of the enamel antagonists were calculated for all groups. To compare the RMS, +AVG, and -AVG values in each of the four regions, the normality was assessed with Shapiro-Wilk tests ($P > .05$). As the equal variance criterion was not satisfied for the Levene's test, Welch's analysis of variance (ANOVA) was performed, and post-validation was carried out with the Games-Howell test. The Kruskal-Wallis test was used to determine the statistical significance of the wear volume loss of the enamel antagonists. A two-way ANOVA was used to investigate the impact of two factors on fracture resistance: the type of zirconia and the artificial aging (thermo-cycling and mechanical loading), as well as their interactions. One-way ANOVA was conducted to determine if the fracture load values before and after aging were different for each group. A post-hoc pairwise comparison was performed and Bonferroni's correction was applied. A paired t-test was employed to determine whether there was a significant difference between the groups before and after aging. Statistical software (IBM SPSS Statistics v25.0; IBM Corp., Armonk, NY, USA) was used for all analyses with a statistical significance (P) of .05.

RESULTS

In terms of intaglio surface trueness of zirconia crowns, the measured values of three groups (FSZ, 4YZ, and 5YZ) in four different areas (inner surface, occlusal, margin, and axial area) are shown in Table 2, Figure 1 and Figure 2. The FSZ group showed significantly larger mean RMS values than the 4YZ and 5YZ groups in the margin ($P < .001$) and occlusal area ($P <$

Table 1. Parameters for chewing simulation used in this study

Transverse path	Speed (mm/s)	Weight	Thermal cycling	Cycle
Downward: 2 mm Lateral: 0.7 mm	Up: 55 Down: 30 Forward: 30 Back: 55	5 kg	Dwell time: 60 s Temperature: 5°C - 55°C	Number: 120,000 Frequency: 1.5 Hz

.001). The FSZ group also showed larger positive average deviation (+AVG) value in the occlusal area than the other groups (P < .001). The negative average deviation (-AVG) value was significantly larger in the FSZ group than the other groups in the margin (P = .003), occlusal (P = .001), and inner surface areas (P = .001).

After chewing simulation for 120,000 cycles, the measured wear volume loss (mean ± SD) of the enamel antagonists were 1.49 ± 1.58 mm^3, 2.49 ± 1.33 mm^3, and 2.05 ± 1.24 mm^3 for FSZ, 4YZ, and 5YZ

Table 2. Intaglio surface trueness measurements (µm, root-mean-square, positive average deviation, and negative average deviation) of three zirconia groups, measured in four different areas of inspection: inner surface, occlusal, margin, and axial surfaces

Areas	RMS (Mean ± SD) Groups FSZ	4YZ	5YZ	*P*	+AVG (Mean ± SD) Groups FSZ	4YZ	5YZ	*P*	-AVG (Mean ± SD) Groups FSZ	4YZ	5YZ	*P*
Inner surface	33.72 ± 2.35	26.94 ± 5.91	29.56 ± 4.51	.001	27.26 ± 3.62	22.21 ± 6.10	24.81 ± 5.74	.051	-24.88 ± 3.15	-20.12 ± 2.96	-21.79 ± 2.96	.001
Occlusal	29.81 ± 4.46	21.11 ± 2.97	22.39 ± 3.68	< .001	25.29 ± 4.85	14.58 ± 1.59	15.10 ± 5.02	< .001	-24.62 ± 5.03	-18.89 ± 3.28	-18.89 ± 4.11	.001
Margin	37.96 ± 3.83	28.71 ± 6.30	29.16 ± 4.66	< .001	14.39 ± 5.76	17.11 ± 5.16	17.31 ± 3.51	.225	-34.02 ± 8.74	-25.51 ± 5.86	-26.45 ± 4.79	.003
Axial	36.69 ± 7.29	29.79 ± 9.85	34.72 ± 7.21	.085	32.14 ± 6.31	26.22 ± 9.23	30.09 ± 8.94	.171	-15.69 ± 5.99	-11.00 ± 4.88	-13.67 ± 6.35	.111

FSZ, Fully-sintered (Y,Nb)-TZP group; 4YZ, Partially sintered 4Y-PSZ group; 5YZ, Partially sintered 5Y-PSZ group; RMS, root-mean-square; SD, standard deviation; +AVG, positive average deviation; -AVG, negative average deviation of four zirconia crown groups measured in four different areas of inspection. Trueness (RMS value), positive average, negative average of the zirconia crowns of the 3 different groups.

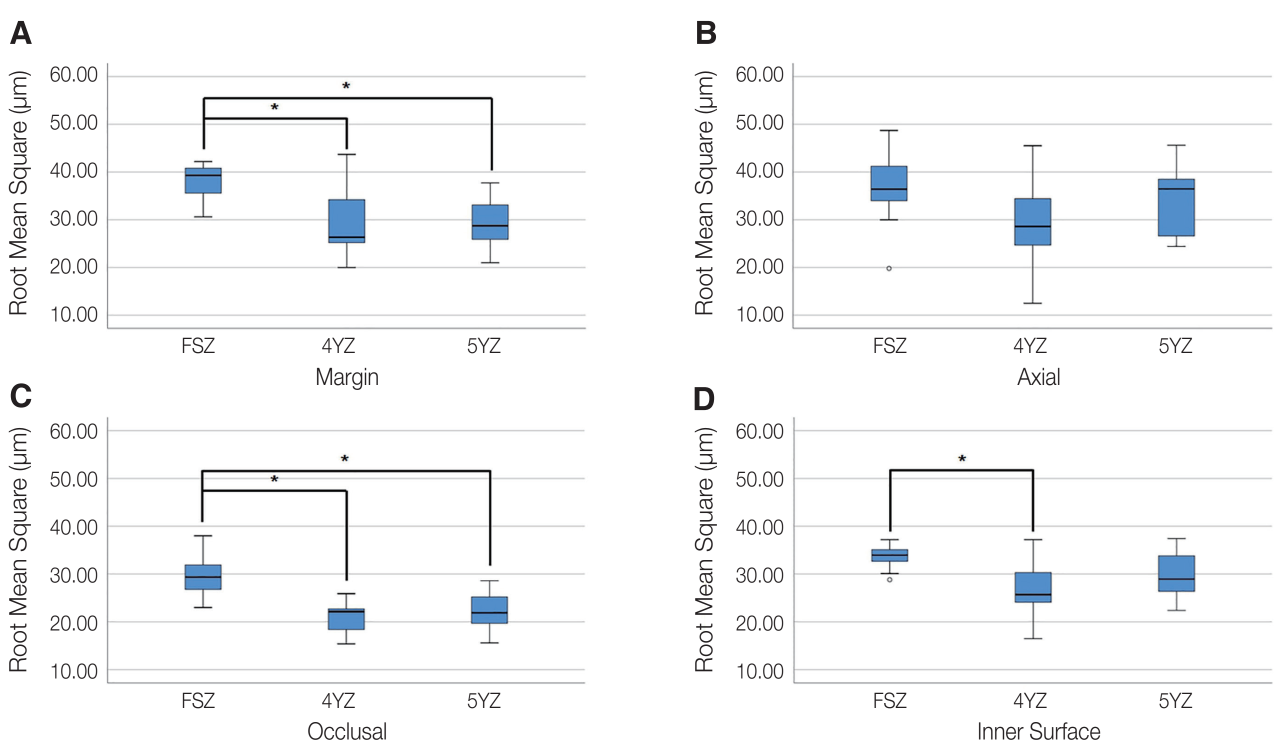

Fig. 1. Intaglio surface trueness measurements (µm, root-mean-square) of three zirconia groups measured in four different areas of inspection. (A) Margin, (B) Axial area, (C) Occlusal area, (D) Inner area. Statistically significant difference (P < .05) was marked with asterisk (*). FSZ, Fully-sintered (Y,Nb)-TZP group; 4YZ, Partially sintered 4Y-PSZ group; 5YZ, Partially sintered 5Y-PSZ group.

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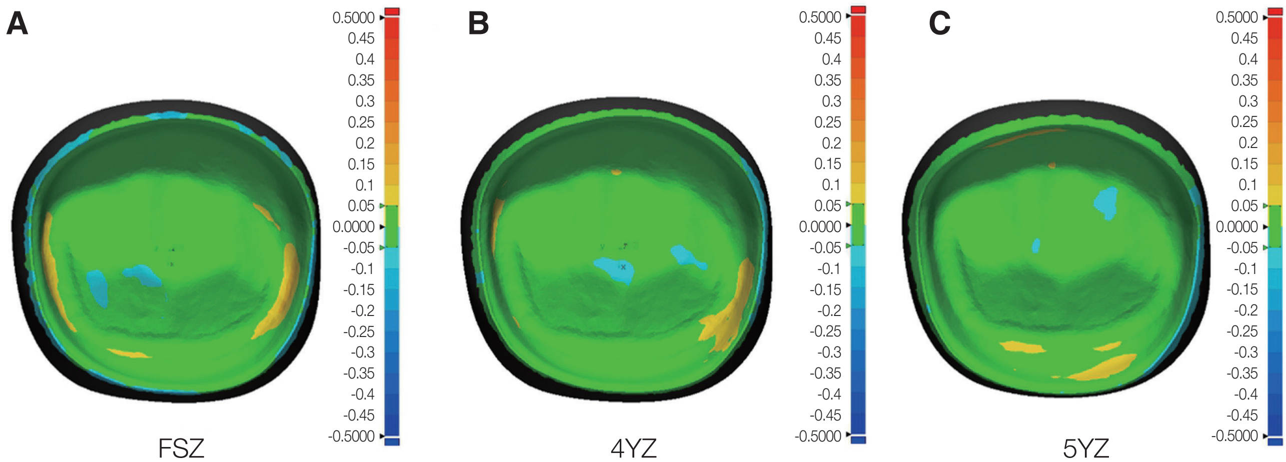


Fig. 2. Representative color deviation maps of intaglio trueness of 3 zirconia crown groups (positive deviation, yellow to red, negative deviation, cyan to blue, deviation below 50 µm, green). (A) FSZ, Fully-sintered (Y,Nb)-TZP group; (B) 4YZ, Partially sintered 4Y-PSZ group; (C) 5YZ, partially sintered 5Y-PSZ group.

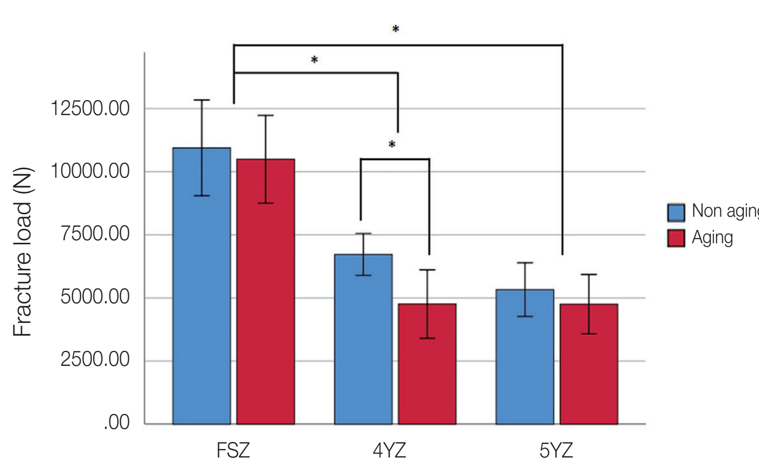
groups, respectively. No statistically significant differences were detected among the groups (*P* = .95). In terms of fracture resistance, the fracture load values of the FSZ group were larger than those of the other groups both before and after artificial aging (Fig. 3). Both the type of zirconia (*P* = .001) and artificial aging (*P* = .026) affected significantly on the fracture resistance of the monolithic crowns; however, there was no interaction between two factors (*P* = .295). Regardless of artificial aging, the FSZ group showed the highest fracture resistance among the zirconia crown groups (Fig. 3, Table 3).

Table 3. Fracture load (N) of the zirconia crowns of the 3 different groups

Aging	Group			*P*
	FSZ	4YZ	5YZ	
Aging	10490.75 ± 1739.02	4756.07 ± 1355.76	4749.78 ± 1174.88	< .001
Non-aging	10945.85 ± 1897.82	6720.43 ± 829.34	5329.84 ± 1061.87	< .001

FSZ, Fully-sintered (Y,Nb)-TZP group; 4YZ, Partially sintered 4Y-PSZ group; 5YZ, Partially sintered 5Y-PSZ group.

Fig. 3. Fracture load (N) of three zirconia groups before and after artificial aging.
FSZ, Fully-sintered (Y,Nb)-TZP group;
4YZ, Partially sintered 4Y-PSZ group;
5YZ, Partially sintered 5Y-PSZ group.
* Statistically significant difference (*P* < .05).



DISCUSSION

Based on the results of this study, the first and third null hypotheses were rejected. The FSZ group showed significantly lower intaglio surface trueness than the other zirconia groups, especially in the regions such as margin and occlusal areas. In this study, the FSZ crown was fabricated using a 4-axis milling machine according to the manufacturer's instruction, which can be less accurate than the 5-axis milling machine, and this would explain the difference in trueness from the crowns of the 4YZ or 5YZ groups.[8,37] Furthermore, if the bur diameter exceeds the curvature of the prepared abutment, particularly the transition from the axial to the occlusal surface, or the path of milling tools does not fully reproduce the morphologies of the intaglio surface of the crown, it could decrease the accuracy of milled prosthesis.[13] The higher negative average deviation in the margin area of the FSZ group than the others could be the result of the aforementioned problem. Nevertheless, the accuracy of the monolithic crowns evaluated in this study is within the clinically acceptable level.[8,38] Based on the RMS values at the margin and inner surface areas showed the interquartile range of the FSZ group was relatively smaller than those of the other groups. Aside from the wear of milling tools, 'hard milling' of dental ceramics without further sintering may be more advantageous than 'soft milling' for production of complex dental prostheses.[2] For the 4YZ and 5YZ groups, the uneven shrinkage after sintering may have contributed relatively wide interquartile range of RMS measurements for intaglio surface trueness.[39] Additionally, the high-speed sintering was reported to affect the internal fit of monolithic crown, resulting in worse fit than the conventional sintering, due to different pattern of temperature change.[4]

According to the manufacturers' information, the (Y, Nb)-TZP block and 4Y-PSZ block showed similar flexural strength (MPa) values, while the 5Y-PSZ block showed relatively lower than those. However, the fracture load (N) was significantly higher for FSZ group than 4YZ group. It is speculated that the material of the support die, the loading angle, the fatigue loading, the specimen thickness, and the cementation procedure have influenced the load-to-failure testing results.[21,23] In addition, relatively high fracture load value may be attributed to the use of Co-Cr alloy for testing abutment with high elastic modulus.[21] Generally, the zirconia crowns undergone high-speed sintering could change the grain size and fracture resistance compared to the conventional sintering.[5,6,30] Despite of high-speed sintering, the fracture load of the 4YZ group was lower than that of the FSZ group, which needs further evaluation.

After artificial aging, the fracture load values changed in all three groups, but only the 4YZ group showed a significant decrease. Previous studies showed that 4Y-PSZ and 5Y-PSZ were less sensitive to hydrothermal aging than 3Y-TZP.[30-32] Higher stabilizer content is responsible for the high resistance to aging and for eliminating the transformation toughening mechanism, also being responsible for the appearance of a great amount of cubic crystals in its microstructure.[40] Since there was less hydrothermal degradation in the FSZ group utilizing (Y,Nb)-TZP under low-temperature aging conditions, the change in strength was relatively minimal. This is mostly attributed to the phase stability of zirconia, which is caused by the Y-Nb alignment in the tetragonal zirconia lattice.[12]

CONCLUSION

Within the limitations of this study, the full-contour crowns of fully-sintered (Y, Nb)-TZP as well as partially sintered 4Y-PSZ or 5Y-PSZ with high-speed sintering were concluded as clinically acceptable for single-visit dentistry, in terms of intaglio surface trueness, antagonist's wear volume loss, and fracture resistance after simulated mastication. The fully-sintered zirconia crown showed higher fracture resistance than the partially sintered zirconia crown with high-speed sintering.

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