Effect of core design on fracture resistance of zirconia-lithium disilicate anterior bilayered crowns

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PURPOSE. The effect of core design on the fracture resistance of zirconia-lithium disilicate (LS2) bilayered crowns for anterior teeth is evaluated by comparing with that of metal-ceramic crowns. **MATERIALS AND METHODS.** Forty customized titanium abutments for maxillary central incisor were prepared. Each group of 10 units was constructed using the same veneer form of designs A and B, which covered labial surface to approximately one third of the incisal and cervical palatal surface, respectively. LS2 pressed-on-zirconia (POZ) and porcelain-fused-to-metal (PFM) crowns were divided into "POZ_A," "POZ_B," "PFM_A," and "PFM_B" groups, and 6000 thermal cycles (5/55 °C) were performed after 24 h storage in distilled water at 37 °C. All specimens were prepared using a single type of self-adhesive resin cement. The fracture resistance was measured using a universal testing machine. Failure mode and elemental analyses of the bonding interface were performed. The data were analyzed using Welch's t-test and the Games-Howell exact test. **RESULTS.** The PFM_B (1376. 8 ± 93.3 N) group demonstrated significantly higher fracture strength than the PFM_A (915.8 ± 206.3 N) and POZ_B (963.8 ± 316.2 N) groups (P<.05). There was no statistically significant difference in fracture resistance between the POZ_A (1184.4 ± 319.6 N) and POZ_B groups (P>.05). Regardless of the design differences of the zirconia cores, fractures involving cores occurred in all specimens of the POZ groups. **CONCLUSION.** The bilayered anterior POZ crowns showed different fracture resistance and fracture pattern according to the core design compared to PFM. ***[J Adv Prosthodont 2020;12:181-8]***

KEYWORDS: Lithium disilicate; Zirconia; Metal-ceramic crown; Fracture resistance

INTRODUCTION

Zirconia has made significant contributions to meeting the challenge of low-fracture-resistant dental porcelains.[1,2] However, the low translucency of zirconia caused by its non-transparent optical property presents a limitation in its clinical application.[3,4] Although a method of using feldspathic porcelain with conventional metal ceramic has been introduced to resolve this problem, various complications associated with this method have been reported, such as ceramic chipping.[5-9] In particular, chipping in porcelain-veneered zirconia crowns has been found to occur at a significantly higher frequency than that causing chipping when using conventional (porcelain-fused-to-metal, PFM) crowns.[10] To overcome this problem, zirconia-lithium disilicate (LS2) bilayered crowns, using LS2 heated pressed-on-zirconia (POZ), have been introduced.[11,12] Owing to the relatively improved translucency compared to monolithic zirconia and excellent fracture resistance in the zirconia core than all ceramic crowns,[13] these crowns have been applied clinically for anterior and posterior prosthetic treatment cases requiring more esthetic results. However, there are very few studies on their clinical applicability. Moreover, studies on the bonding force between zirconia and LS2 have

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reported experimental comparisons of shear- and tensile-bond strengths according to liner treatments.[14-21] However, there is a paucity of studies on fracture resistance when crowns are used to fabricate a maxillary anterior prosthesis and when an intraoral load is applied by the antagonistic mandibular anterior teeth such as that observed during the anterior guidance process.[12] Additionally, with these bilayered ceramic crowns, the shape of the zirconia core must be changed according to the location of the zirconia-LS2 junction. Thus, the thickness and shape of the LS2 veneer ceramic can be altered.[22,23] Therefore, comparative analysis of fracture resistance according to such alterations is clinically significant. In this study, conventional crowns were designated as the control group based on fracture resistance, being compared according to zirconia core design in maxillary anterior LS2 bilayered crowns.[24,25] The aim of this study is to investigate the effect of the coping designs of PFM and POZ on the fracture resistance of the prosthesis. The hypothesis of this study is that both PFM and POZ have no effect on the fracture resistance according to the difference in the substructure.

MATERIALS AND METHODS

Maxillary central incisor (#11) models of the bilayered crowns were divided into experimental and control groups. Models having a veneer covering that spanned from the entire labial surface to the incisal 1/3 and cervical 1/3 points of the palatal surface were designated as POZ_A and POZ_B, respectively, in the experimental group. PFM_A and PFM_B were similarly added to the control group (Fig. 1).

The dentiform (Pro2002-UL-UP-FEM-28, Nissin Dental, Kyoto, Japan) of the artificial maxillary central incisor was scanned (D700, 3Shape A/S, Copenhagen, Denmark), followed by computer-aided design (CAD) and computer-aided manufacturing (CAM) fabrication of a customized traditional Ti abutment to have the shape of the final prosthesis.

The cores of PFM_A and POZ_A were designed to form most of the palatal surface, except at the incisal margin, formed with zirconia and cobalt-chrome, respectively (Fig. 1A). PFM_B and POZ_B were designed to present

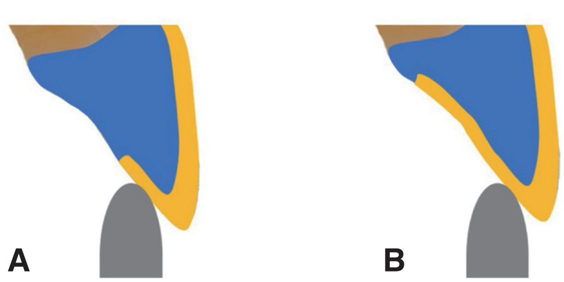

Fig. 1. Schematic of the static loading process: (A) PFM_A and POZ_A, (B) PFM_B and POZ_B.

most of the palatal surface, except for the cervical region, formed using LS2 and feldspathic porcelain veneer (Fig. 1B). The shapes of designs A and B, formed during the planning stage, were ultimately verified using three-dimensional analysis.

A zirconia block (Zirtooth, HASS, Gangneung, Korea) was used to fabricate the fully sintered zirconia core for POZ_A and POZ_B using a CAD program (Zirkonzhan Nesting, Zirkonzahn GmbH, Gais, Italy) and a CAM milling machine (Zirkonzahn M5, Zirkonzahn GmbH). This was done while referencing the scanned data of the final prosthesis and the Ti abutment shape. The samples were sintered for 2 h in a 1,500°C furnace (Austromat μSiC, Dental-Keramiköfen GmbH, Freilassing, Germany) to prepare 20 core samples for the experimental group.

Lost-wax and heat-pressing techniques were used to press the LS2 veneer. After forming a sprue on the prepared zirconia core using the same method as traditional prosthesis fabrication, it was covered with investment material (HSTM Investment, Microstar Dental, Lawrenceville, GA, USA). Then, the wax pattern was removed from the furnace, and the mold was preheated for 30 min at 860°C. An LS2 ingot (Amber LiSi-POZ, HASS) was softened at 915°C, and a pressing machine (Rosetta press, HASS) was used for LS2 pressing. Afterwards, the casting was cooled at room temperature, and the investment material was removed. Subsequently, a high-speed handpiece (Ti-Max X600L, NSK, Tokyo, Japan) and a diamond bur (SR-11, MANI Inc., Utsunomiya, Tokyo, Japan) were used to cut off the sprue. The remaining investment material was completely removed using the air-particle abrasion (APA) method (Basic master, Renfert, Hilzingen, Germany).

To minimize errors occurring from differences between the traditional fabrication method of the metal-ceramic crown and the CAD/CAM for the zirconia core, a sintering metal block made of cobalt-chrome alloy (Soft Metal, LHK, Daegu, Korea) was used to fabricate the core of the PFM (control group) with the same shapes and thicknesses of the zirconia core (experimental group). Moreover, to minimize shape errors during the metal-ceramic crown veneering process, compared with the LS2 veneer in the experimental group, the same wax pattern was scanned and veneered for all samples by an experienced dental technician. Furthermore, sintering and machining were performed in the same manner as that performed in the clinical prosthesis fabrication.

A total of 40 prostheses [PFM_A (n = 10), PFM_B, (n = 10), POZ_A (n = 10), and POZ_B (n = 10)] were cemented to the titanium-alloy abutments using self-curing resin cement (G-Cem One, GC Corporation, Tokyo, Japan) (Fig. 2).

The samples were stored in 37°C distilled water for 24 h, after which a 6000-cycle thermal cycling treatment (soak in 5°C cold water and 55°C warm water for 30 s each) was applied to the samples.

All samples were then mounted on a customized metal jig to apply the static loading at the same point. The samples were fixed with a slope of approximately 30° relative to

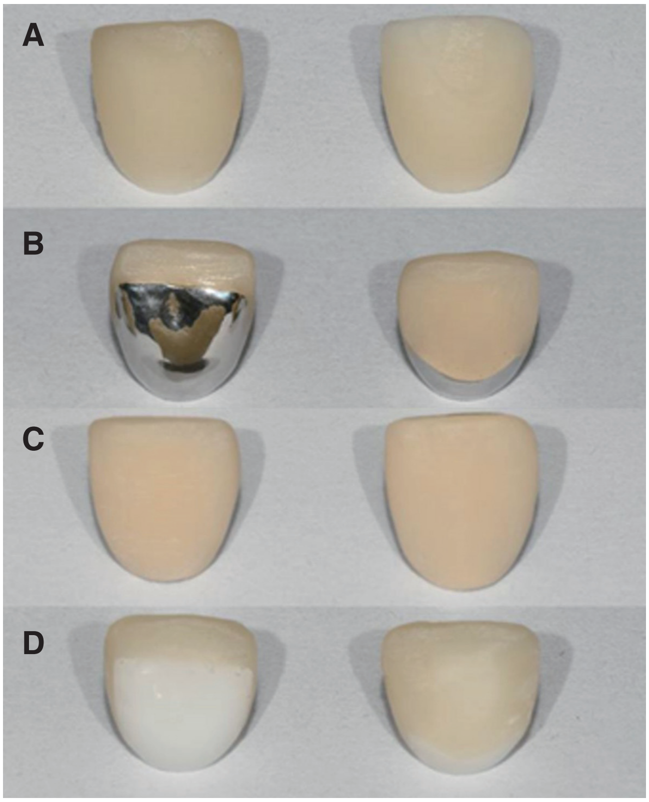

Fig. 2. Specimens. (A) Labial view of PFM_A (left) and PFM_B (right), (B) Palatal view of PFM_A (left) and PFM_B (right), (C) Labial view of POZ_A (left) and POZ_B (right), (D) Palatal view of POZ_A (left) and POZ_B (right).

their long axis to create measurement conditions similar to those when contacting the mandibular anterior teeth in clinical settings. The static loading was set to a rate of 1 mm/s, and a universal testing machine (Model 5982, Instron, Norwood, MA, USA) was used to record the maximum load value when the initial fracture occurred. The fracture patterns of all specimens were observed to determine whether the veneer alone or with the core was included.

Three randomly selected fragments from each group were used for interface analysis using a scanning electron microscope (SEM; Quanta 250 FEG, FEI, Hillsboro, OR, USA).

Energy dispersive X-ray spectroscopy (EDS; Ametek EDAX Apollo XP, Mahwah, NJ, USA) in conjunction with SEM was used to analyze the ion distribution on the interface between the veneer and core on both experimental and control groups.

Statistical analysis was then performed using the SPSS statistics program (IBM SPSS 23.0, IBM Corp., Armonk, NY, USA). To identify the differences in the fracture resistance according to the differences in the core design for all experimental and control specimens, the normal distribution of data and homogeneity of variance were checked. Because the normal distribution was satisfactory and the homogeneity of variance was not equally distributed, Welch's t-test was performed at a 95% confidence interval instead of one-way analysis of variance. The post-hoc Games-Howell test was then performed.

RESULTS

In the control group, PFM_B exhibited a statistically significantly lower fracture strength than PFM_A ($P < .001$). In the experimental group, POZ samples showed no statistically significant differences between the different designs, unlike the PFM samples ($P > .05$). POZ_B showed a statistically significantly lower fracture strength than PFM_B ($P < .011$). However, POZ_A did not show statistically significant fracture resistance with either PFM_A or PFM_B ($P > .05$) (Table 1, Table 2).

PFM specimens that did not contain metal coping showed both chipping and failure of the entire veneer. On the other hand, the POZ specimens containing both LS2 veneer and zirconia coping showed fractures (Fig. 3).

Fragments of the metal-ceramic crown (control group) clearly showed oxidized layers in the metal, veneering porcelain, and interface at low and high magnifications (× 100) (Fig. 4A and 4B). However, fragments of the LS2-POZ crown (experimental group) did not show such distinct interfaces at low (× 100) or high (× 1000) magnification (Fig. 4C and 4D). Results at × 10000 magnification showed LS2-specific spindle crystals (Fig. 5A) and zirconia-specific polycrystalline ceramics (Fig. 5B).

EDS at intervals differentiated by metal, oxidation layer, and feldspathic porcelain of the metal-ceramic crown fragment confirmed the presence of elements at each interval representing the chrome-cobalt alloy, oxide film, and feldspathic porcelain (Fig. 6). The high content of chrome and cobalt, representing the metal core, was also confirmed. Relatively increased oxygen content was confirmed at the oxidation layer. However, oxygen and silicon, the main constituents of glass, were detected as the main constituents in the feldspathic porcelain layer.

Unlike the results of typical metal-ceramic crown fragment analyses, LS2 POZ crown fragments did not show clearly differentiated reaction (oxidation) layers. Although

Table 1. Mean (± standard deviation) fracture resistance (N)

Group	Fracture resistance (Mean ± SD)
PFM_A	915.8 ± 206.3^{ac}
PFM_B	1376.8 ± 93.3^{b}
POZ_A	1184.4 ± 319.6^{abc}
POZ_B	963.8 ± 316.2^{ac}

Same superscript letters identify statistically similar groups ($P < .05$).

Table 2. Post hoc test of differences in fracture resistance (N)

(I) Group	(J) Group	(I-J) Mean difference	± SD	*P*	95% CI	
					Lower bound	Upper bound
PFM_A	PFM_B	-460.958*	71.617	< .001	-672.23	-249.68
	POZ_A	-268.546	120.304	.158	-614.24	77.15
	POZ_B	-48.035	119.419	.977	-390.95	294.88
PFM_B	PFM_A	460.958*	71.617	< .001	249.68	672.23
	POZ_A	192.412	105.293	.314	-126.83	511.65
	POZ_B	412.923*	104.279	.011	96.92	728.93
POZ_A	PFM_A	268.546	120.304	.158	-77.15	614.24
	PFM_B	-192.412	105.293	.314	-511.65	126.83
	POZ_B	220.511	142.191	.430	-181.37	622.39
POZ_B	PFM_A	48.035	119.419	.977	-294.88	390.95
	PFM_B	-412.923*	104.279	.011	-728.93	-96.92
	POZ_A	-220.511	142.191	.430	-622.39	181.37

* Significant difference ($P < .05$)

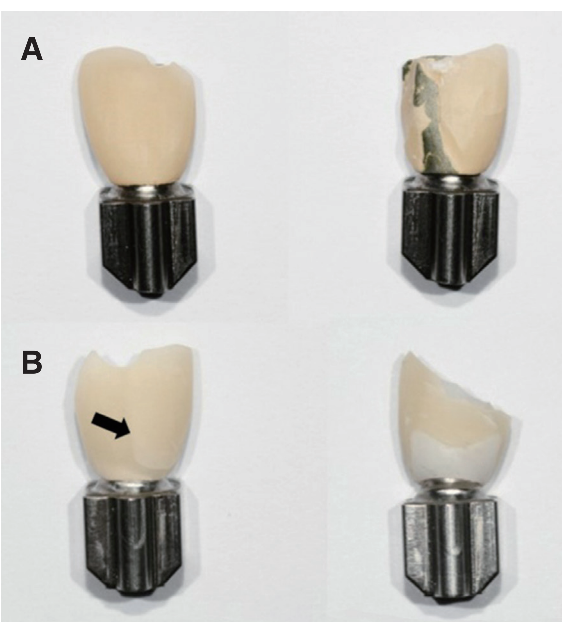

Fig. 3. Fracture pattern. (A) Chipping (left) and veneer fracture (right) of PFM, (B) Veneer and core fracture of POZ, Arrow indicates fracture line through labial surface.

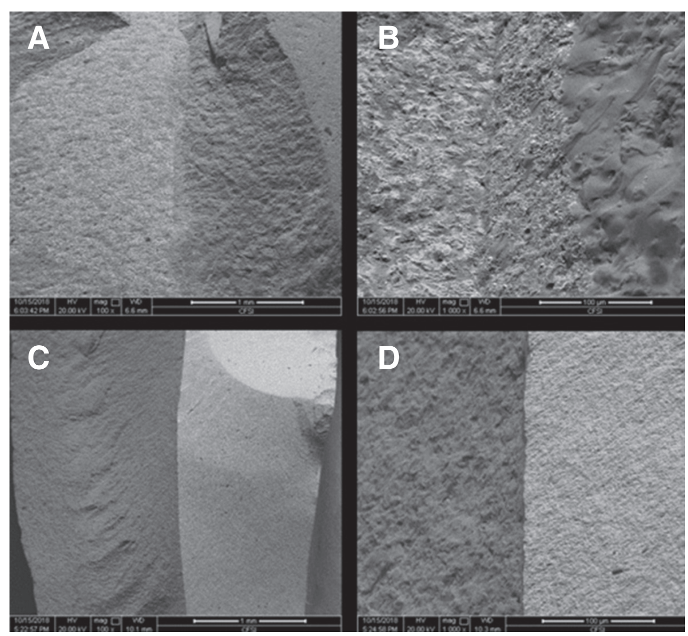

Fig. 4. Scanning electron microscope (SEM) images. (A) Control group (PFM) at low magnification (× 100), (B) Control group (PFM) at high magnification (× 1000), (C) Experimental group (POZ) at low magnification (× 100), (D) Experimental group (POZ) at high magnification (× 1000).

the heat-pressed region was comprised mostly of oxygen and silicon, which are also the constituents of LS2, the zirconia region contained oxygen and zirconium as the main constituents (i.e., oxidized zirconium). Moreover, the reaction layer was difficult to observe using SEM, but a pattern of partial ion diffusion was observed via EDS mapping (Fig. 7).

DISCUSSION

The aim of this study is to investigate the effect of the coping designs of PFM and POZ on the fracture resistance of the prosthesis. According to the results, the null hypothesis was rejected because the control group (PFM) caused a significant difference in the fracture resistance with regard to the coping design. On the other hand, in the experimental group (POZ), the null hypothesis was not rejected because there was no significant difference in the fracture resistance

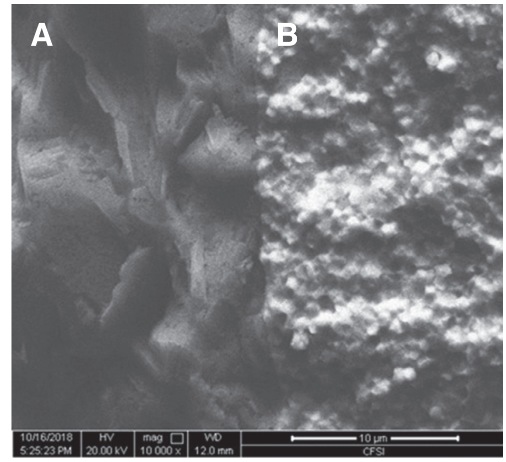

Fig. 5. SEM images of experimental group (POZ) at × 10000 magnification. (A) LS2-specific spindle crystals, (B) Zirconia-specific polycrystalline structure.

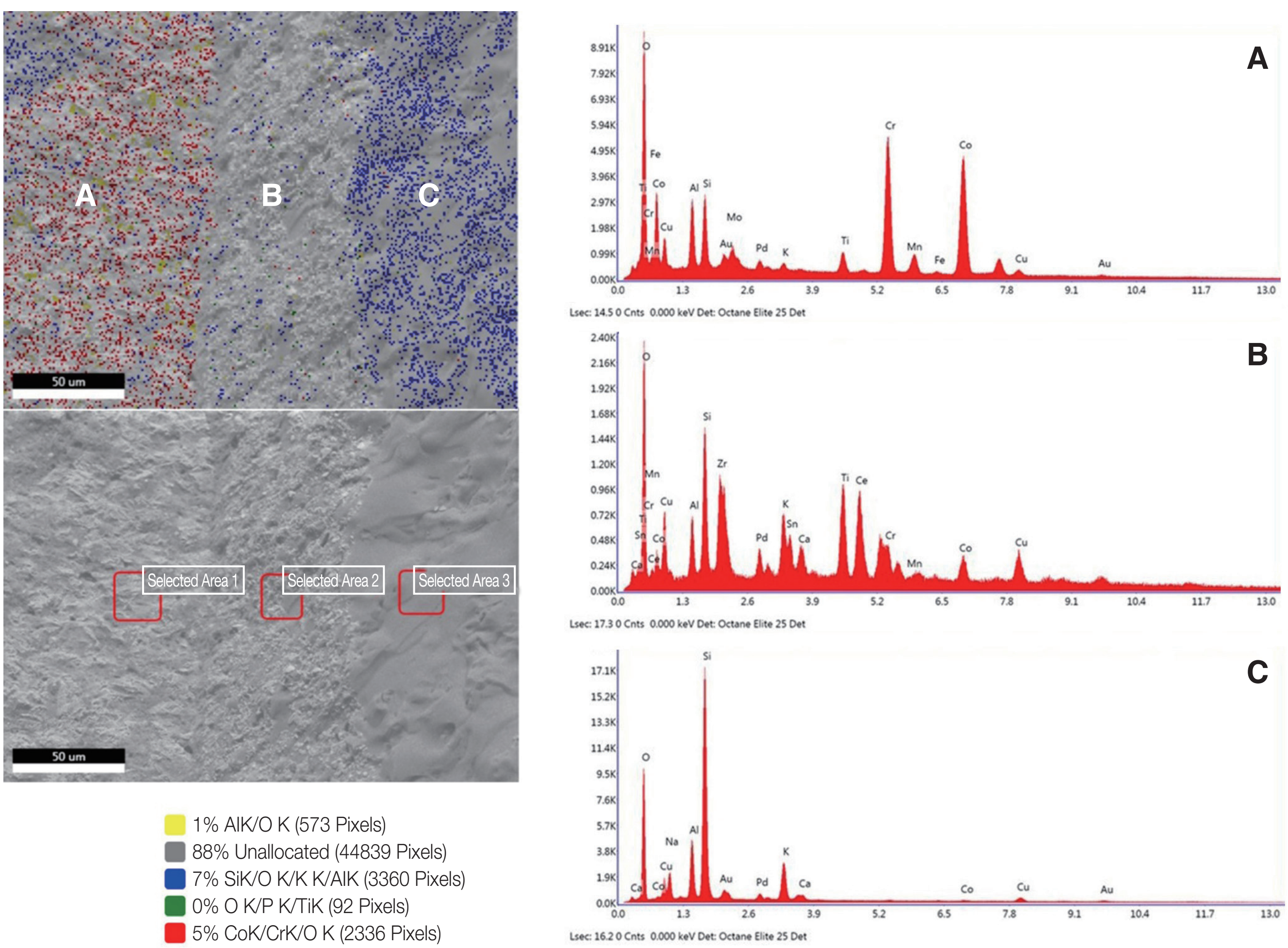

Fig. 6. Energy dispersive X-ray spectroscopy (EDS) mapping analysis of the control group (PFM). (A) Presenting chrome-cobalt alloy, (B) Oxide layer, (C) Feldspathic porcelain.

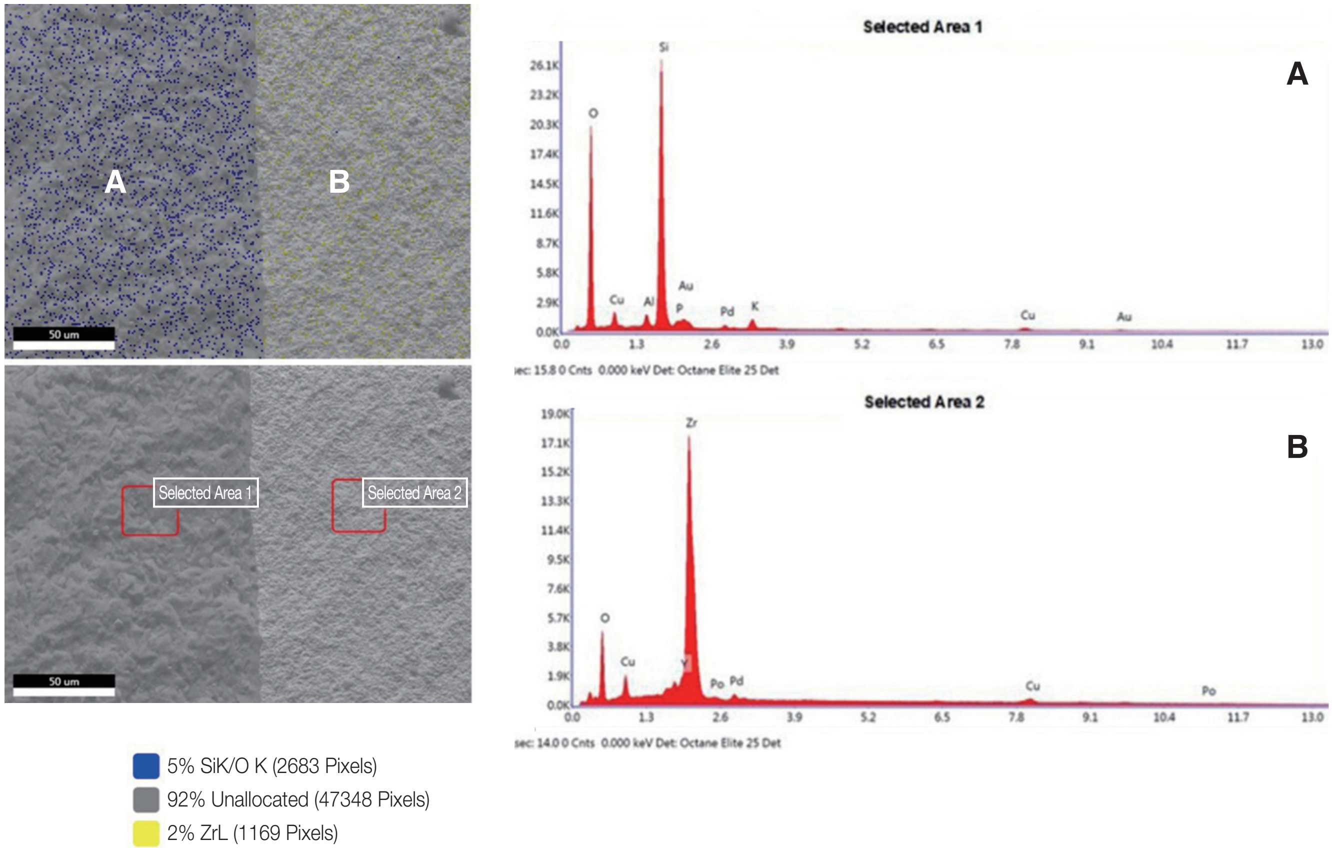

Fig. 7. Energy dispersive X-ray spectroscopy (EDS) mapping analysis of the experimental group (POZ). (A) Heat-pressed LS2 region was comprised mostly of oxygen and silicon, (B) Zirconia region contained oxygen and zirconium as the main constituents.

with regard to the coping design.

Because the PFM_B was designed to apply the loading to the veneering porcelain relatively far from the metal-porcelain junction, it was expected to be more favorable than PFM_A with respect to porcelain fracturing. In PFM_A, loading was applied near the junction. Moreover, these results were similar to previous studies that reported how such a metal-porcelain junction could be more vulnerable to porcelain fracturing.[24,25] With respect to the POZ samples, there was no statistically significant difference in the fracture resistance between POZ_A and POZ_B. It was determined that this could be attributed to the material characteristics of the zirconia-LS2 prosthesis fabricated with only the bilayered ceramic. In other words, the veneers and cores made of only highly brittle ceramics were affected more by the amount of zirconia core rather than the distance between the zirconia-LS2 junction and the loading point. POZ_A had a zirconia core that was responsible for more of the palatal surface than the POZ_B, whereas the POZ_B had approximately 1/3 of the palatal surface covered by LS2 veneer. As a result, the zirconia in the prosthesis was relatively thinner than that of the POZ_A. Thus, the decrease in the relative percentage of zirconia having relatively excellent mechanical properties was believed to have impacted the decrease in the fracture resistance of the entire prosthesis.

In a previous study that investigated the fracture resistance of the POZ bilayerd ceramic, an increase in flexural strength was reported as the zirconia core thickness increased and the LS2 veneer thickness decreased.[14] Therefore, in the palatal surface of the POZ_A group, the core and the veneer junction positions are in close proximities to the loading point, which may prove to be disadvantageous for fracture resistance. In our study, the fracture resistance of PFM_A was significantly lower than that of PFM_B. However, compared to that of POZ_B, POZ_A had a relatively higher thickness of zirconia coping; thus, unlike the control group (PFM), the experimental group (POZ) did not cause a difference in fracture resistance with regard to the coping design. Moreover, although the LS2 veneer showed relatively inferior mechanical properties to the zirconia core, it was a glass ceramic and displayed the best mechanical properties.[11,12] Thus, the zirconia-LS2 bilayered crown had a relatively weak impact on fracture resistance compared to the metal ceramic. Additionally, the reaction layer generated by ion migration, identified by EDS mapping, was believed to have offset the differences in mechanical properties between the two ceramics.[19-21] In this study, all specimens were fabricated using the same method as that used for clinical prosthesis fabrication to measure the fracture resistances. However, the differences in error were inevitable because of differences in the methods used to fabricate the infrastructure of the experimental group. Shrinkage resulted from full sintering after pre-sintered zirconia fabrication by CAD/CAM, metal-ceramic crown fabrication by veneering, and fabrication of the core by different casting methods.[2-4] Moreover, the samples were fabricated based on maxillary anterior teeth. In clinical situations, adequate anterior guidance rehabilitation is important for prosthodontic treatment. In many cases, however, ceramic-veneer chipping is the most common complication.[5-10] In this study, it was confirmed that the core should be selectively designed according to the prosthetic material. In particular, when considering the design of the prosthesis in the bilayered ceramic, it is necessary to consider the material properties. The load application of the mandibular anterior teeth to the palatal aspect of the maxillary anterior teeth inside the mouth during mandibular protrusive movement was reproduced. However, there was a limitation in that the measured values were based on static loading, rather than dynamic loading, which actually generates stress. Moreover, although thermal cycling treatment was applied as a preload process prior to measurement, masticatory loading was omitted. Additional studies are needed to further investigate the process under dynamic loading. In addition, clinical studies should be conducted to evaluate the success rate to verify the results of in *vitro* studies.

CONCLUSION

The following conclusions were established on obtaining the results of this study. The maxillary central incisor PFM crown with the veneering feldspathic porcelain covering the 1/3 cervical point of the palatal surface showed a higher fracture resistance than that of its counterpart covering only the 1/3 incisal point. On the other hand, the POZ bilayered ceramic crown of the maxillary central incisor with the zirconia core and LS2 veneer did not show a statistically significant difference in fracture resistance with regard to the core design.

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