Effect of Clinical Sandblasting on the Surface Roughness of Zirconia Cores for all Ceramic Crowns and their Fracture Resistance after the Addition of Repair

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Objective: This in vitro study aims to evaluate the effect of clinical sandblasting with 50 µm aluminum oxide and 30 µm silica-coated particles on the surface roughness of zirconia cores and the subsequent effect on their fracture resistance after veneering with composite using a specific repair kit.

Materials and Methods: Zirconia cores (n=21) were digitally designed and milled from ZirCAD LT B1 (IPS e.max® ZirCAD) blocks using arum 5x-300 Pro (ARUM DENTISTRY). These cores were randomly divided into three groups: Group A: n=8, sandblasted with 50 µm aluminum oxide, and veneered with packable Z350 composite. Group B: n=8, sandblasted with 30 µm silica-coated particles and veneered with packable Z350 composite. Group C: control group (n=5), sandblasted in the laboratory with 110 µm aluminum oxide and veneered with porcelain (Vintage Zr PRO - SHOFU Dental GmbH). All the specimens were tested for surface roughness by the TAYLOR-HOBSON profilometer. After adding veneering material, all the specimens were subjected to a fracture resistance test through a universal testing machine.

Results: One-way ANOVA test showed a significantly higher surface roughness in group B compared to group A. Fracture resistance values showed no significant difference between all the groups.

Conclusion: Silica-coated particles produced higher surface roughness than aluminum oxide alone. The fracture resistance values of all the groups were above the acceptable clinical limit.

Keywords: Aluminum-oxide, Intra-oral repair, Sandblasting, Silica, Surface Roughness, Zirconia.

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Introduction

Dental esthetics is the science and art of improving the appearance and function of the teeth, oral cavity, and facial symmetry by applying specific knowledge and techniques. [1] Ceramics were used as restorative materials in dentistry in the late 1700s, relying on their ability to match the shape and color of natural teeth. [2] Metal alloys (including all-metal and metal-ceramic), ceramics, and resin-based composites are the primary materials used for indirect dental restorations. [3]

All-ceramic indirect restorations are gaining in popularity among patients, mostly because of their ability to match the optical and esthetic features of natural enamel and dentine without compromising biocompatibility or chemical endurance. [4] Zirconium oxide ($ZrO2$ – zirconia) as a base material has shown considerable promise among all-ceramic prosthetic materials because of its high strength, high toughness, high corrosion resistance, and outstanding esthetical effects. [5]

There are certain complications following the placement of all-ceramic crowns and fixed partial dentures. One of the most prevalent issues with indirect restorations is the chipping of the veneering porcelain. [6]

Replacement of a failed restoration could be a viable approach, but it is time-consuming and costly and has the danger of injuring the prepared abutment, so it is not always the most practical option. [7] Repairing the shattered porcelain intraorally, on the other hand, especially in challenging medical or prosthetic situations, is quite simple and provides the patient and dentist with a cost - and time-efficient alternative, effectively restoring both function and esthetics. 8 However, there is limited proof of their success. In addition, the growing variety of materials and production technologies necessitates the creation of therapeutic repair protocols that are appropriate. [9]

For proper repair, the surface of the indirect restoration should be subjected to a pre-treatment to create micromechanical retention with the repair material. [10] Air-particle abrasion (sandblasting), one of the most widely used surface treatment processes, forms micro undercut areas and hence improves the micro-mechanical retention of bonding agents. [11,12] Tribochemical silica coating uses silica-coated alumina sand particles to enhance the bond between resin and zirconia restorative materials. [13]

This research aims to investigate the effect of clinical sandblasting with different powders on the surface roughness of zirconia cores and their fracture resistance after the addition of repair material.

Materials and Methods

Abutment preparation: Lower right first molar was digitally designed using exocad software (Exocad 3.0 Galway) according to the dimensions of a natural tooth. [14] Using the same software, a digital preparation was performed with the following guidelines: Occlusal reduction: 1.5 mm at the center and 2 mm at the cusp tip with 45-degree beveling. The other surfaces were reduced with a taper of 6 degrees. Finishing line: 1.3 mm all around the shoulder with internal roundation, as shown in figure 1 a. 21 definitive Poly Methyl Methacrylate (PMMA) (DETAX, Germany) casts were digitally printed using Arum 5x-300 Pro milling machine corresponding to the previously mentioned parameters. Each cast was attached to a 3D-printed, cylindrical acrylic block (MAZIC, VERICOM CO., LTD.) with a height of 20 mm, and a diameter of 16 mm (Fig 1 b). These are considered the abutments to receive the cores and the crowns.

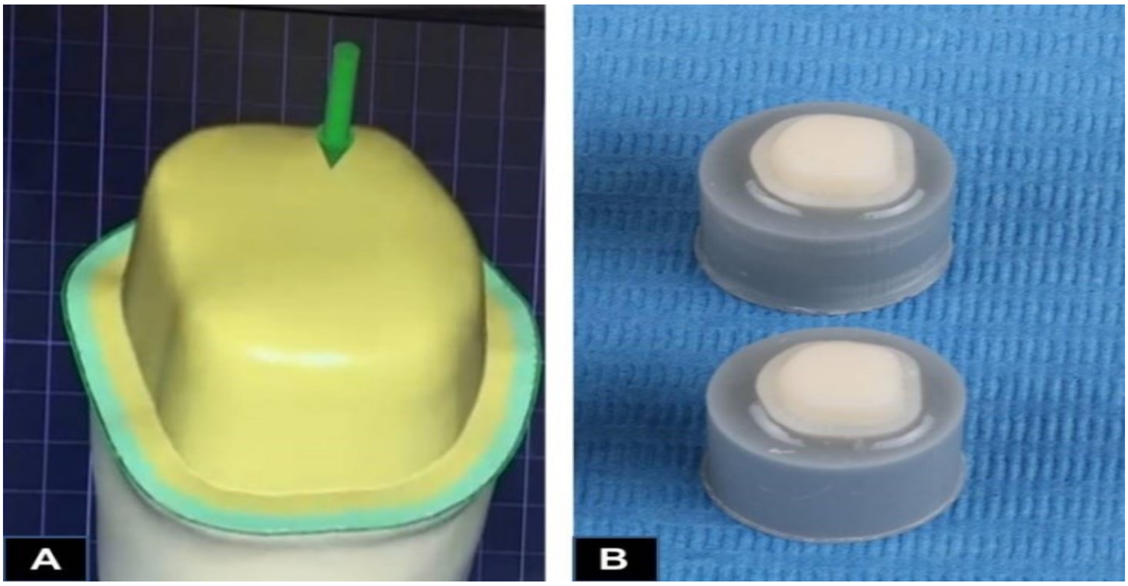


Figure 1. (a) Digital design of the prepared abutment; (b) 3D-printed PMMA definitive dies.

Fabrication of Zirconia cores and crowns: Using Exocad 3.0 Galway software, zirconia cores (n=21) were designed with a thickness of 0.7 mm (Fig 2 a) and milled from ZirCAD LT B1 (IPS e.max® ZirCAD) blocks using arum 5x-300 Pro (ARUM DENTISTRY) (Fig 2 b) with a total time of 20-30 minutes. After milling, the sprues of zirconia were cut by using high-speed diamond bur (Komet 5862, USA). Sintering was done according to Ivoclar manufacturer instructions using AUSTROMAT 674i sintering furnace (DEKEMA Dental-Keramiköfen GmbH) with a total time of 9 hours, 31 minutes, and 38 seconds, reaching a maximum temperature of 1500 °C. These cores were randomly divided into three groups:

Group A n=8 (sandblasted clinically with 50 µm aluminum oxide).

Group B n=8 (sandblasted clinically with 30 µm silica-coated particles).

Group C n=5 (control group: sandblasted in the laboratory with 110 µm aluminum oxide). Specimens of the control group were veneered with porcelain (Vintage Zr PRO - SHOFU Dental GmbH) according to manufacturer instructions.

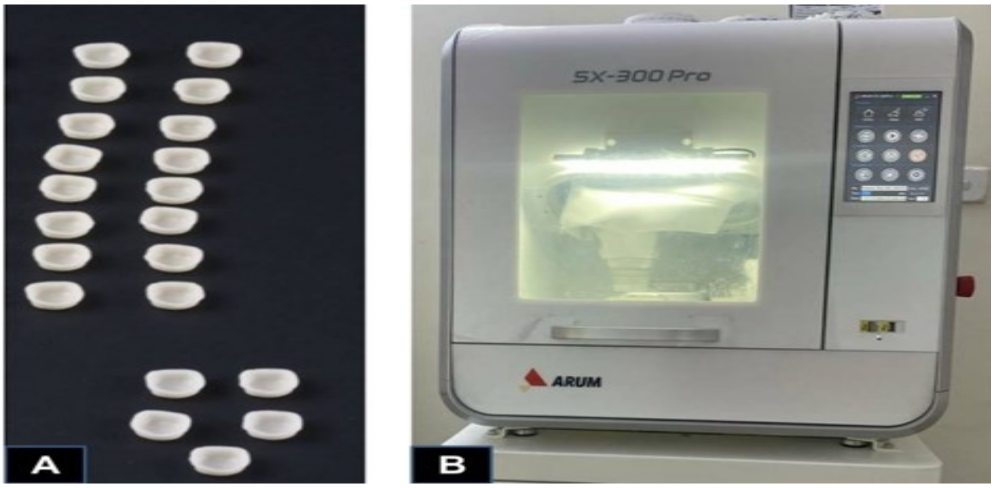


Figure 2: (a) Zirconia cores; (b) Milling machine.

Sandblasting and surface roughness test: Specimens of the control group (n=5) were sandblasted in the laboratory using 110 µm aluminum oxide (Korox; BEGO Medical) at 2 bar pressure from 50 mm distance for 10 seconds at 90° angle as a setting for laboratory work. The specimens from group A were clinically sandblasted with 50 µm aluminum oxide (Dentify GmbH, Germany), while the specimens from group B were clinically sandblasted with 30 µm silica-coated particles (3M™ Cojet™ Sand). Clinical sandblasting parameters were set as follows: 2.5 bar pressure from 10 mm distance for 10 seconds at 90° angle by using a clinical sandblasting device (AquaCare - Velopex International). The sandblaster handle was attached to a customized dental surveyor (Ney Surveyor, Ney Dental, Bloomfield, CT, USA) to allow for standard movement during the process of sandblasting (Fig 3 a, b). All the samples were tested for surface roughness using the TAYLOR-HOBSON profilometer (Fig 3 c). Specimens of the control group were tested twice, before and after sandblasting. Three readings for each specimen were recorded at a 1 mm distance between each line, one in the center and the other two at a 1 mm distance above and below, and the mean value was calculated.

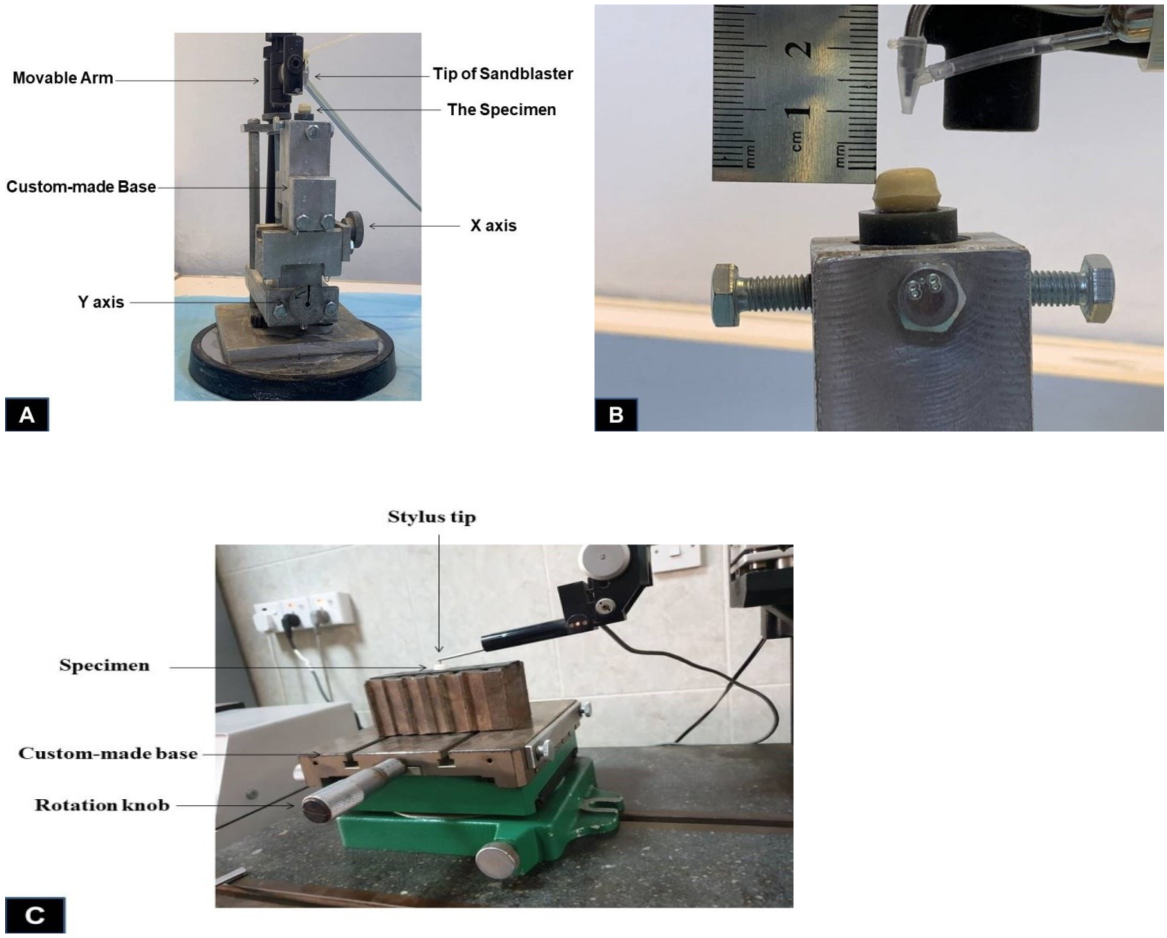


Figure 3: (a) Sandblasting technique; (b) Distance between the tip of the sandblaster and the specimen; (c) Surface roughness test.

Cementation: All the specimens were cemented using dual-cure self-adhesive resin cement TheraCem (Bisco, Schaumburg, USA) following manufacturer instructions. A 5-kg weight was used to keep the samples in place during the primary cement setting to ensure uniform seating pressure.

Application of Repair Material and Fracture Load Test: Composite build-up was performed on all the specimens from groups A and B using light-cure resin composite (3M Filtek Z350 XT). A transparent mold with a thickness of approximately 1 mm (Fig 4 a) was fabricated by using clear polyvinyl siloxane (EXACLEAR; GC Corp) on a randomly chosen specimen from the control group to control the thickness of the composite material. A layer of veneer wax (Renfert GmbH) was added beneath the finishing line of the definitive die to block the undercut and control the fit of the mold (Fig 4 b). Before the addition of the veneering composite, the porcelain repair kit (Intraoral Repair Kit, Bisco Inc., Schaumburg, IL) was used according to the manufacturer’s instructions; one coat of Z-Prime Plus was applied and dried with an air syringe for 3-5 seconds. A thin layer of porcelain bonding resin was applied and spread evenly on the surface, then air-thinned for 3-5 seconds. For each specimen, the mold was loaded with two capsules of (3M Filtek Z350 XT) and secured over the specimen; excess composite was removed with a micro brush and then light cured for 20 seconds for each occlusal, buccal, lingual, mesial, and distal surface using light curing pen (Eighteeth, Changzhou, China) at an intensity of 1000 mW/cm² from a distance of 1-2 mm. After the removal of the transparent mold (Fig 4 c), each surface was light-cured for another 20 seconds. The specimens were then kept in

37 °C distilled water for one week. All the specimens were then subjected to 1000 rounds of thermal cycling between 5±2 ºC and 55±2 ºC for 30 seconds in each bath and 5-second intervals between the baths. All the specimens were then loaded in a universal testing machine (TERCO MT 3037 Terco I&S AB, Sweden). Each specimen was secured in a custom-made metallic base, and the pressure was applied through a vertically movable rod with a semi-spherical head of 6 mm in diameter (Fig 4 d), with a cross speed of 1 mm/min. The loading piston was positioned at the center of the occlusal surface (Fig 4 e). In order to make sure that the position was correct, it was checked by three examiners.

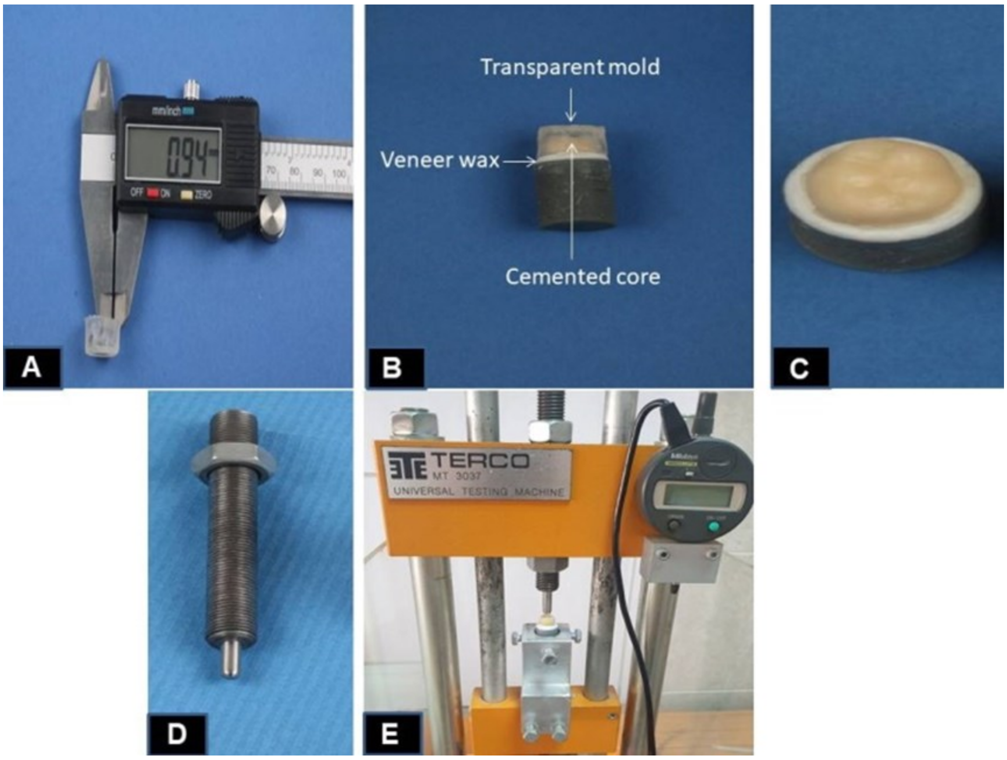


Figure 4: (a) Thickness of the mold; (b) Adaptation of the mold; (c) Addition of the repair material; (d) Custom-made indenter; (e) position of the specimen in the universal testing machine.

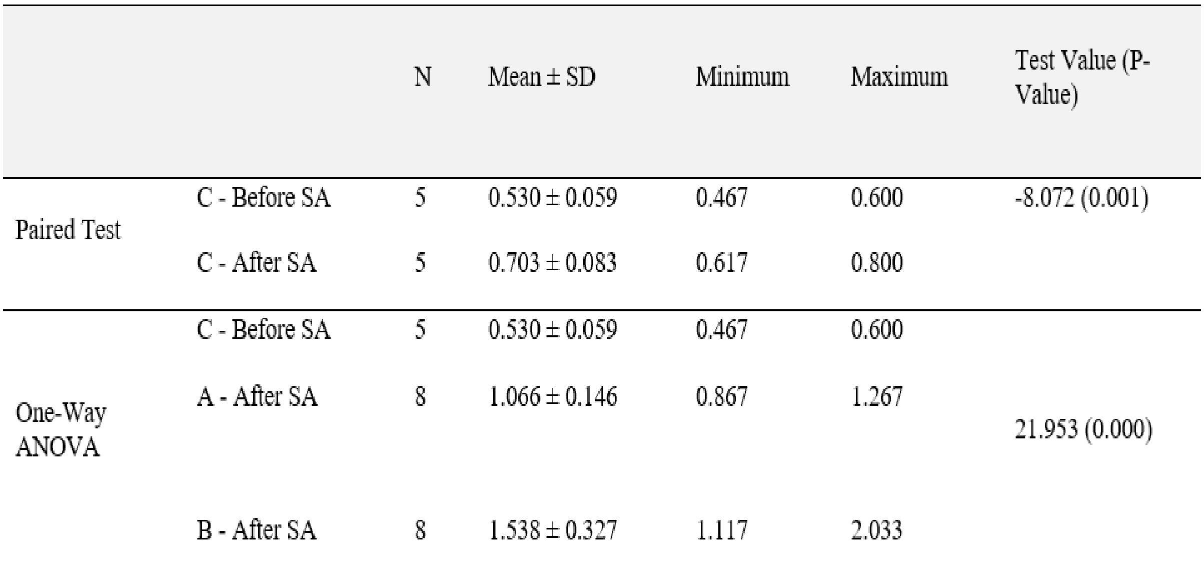
Statistical Analysis: Following a one-way analysis of variance (ANOVA), the normality and homogeneity of variance were tested using Shapiro-Wilk and Levene's tests, respectively. Paired t-tests were computed for groups measured twice, and independent two-sample t-tests were used to compare two groups. The Bonferroni test was used to detect multiple comparisons among the experimental groups. SPSS version 25 was used to run all the statistical tests. A significant difference was set at P<0.05.

Results

Surface roughness: Table 1 shows descriptive statistics of surface roughness measurements where the data is present as mean ± SD. The dataset met both the normality and homogeneity assumptions, which required performing a one-way ANOVA test to determine whether there were significant differences between the groups. The Shapiro-Wilk and Levene's tests were used to accomplish this, and the ANOVA test shows that a statistically significant difference occurred between the groups based on extracted p-value (0.000), As a result, an additional test was required as per Table 2.

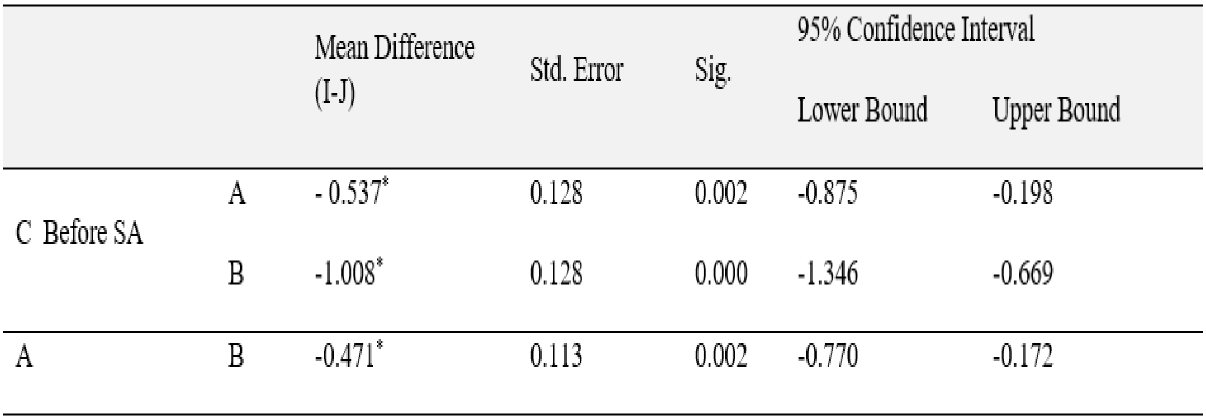
Fracture Resistance: table 3 shows descriptive statistics of fracture resistance measurements where the data is present as mean ± SD. Based on the ANOVA results, it can be reported that the differences were not statistically significant as the p-value (0.066).

Table 1: Statistical Paired T-Test and One-Way ANOVA test for surface roughness values. (µm).



		N	Mean ± SD	Minimum	Maximum	Test Value (P-Value)
Paired Test	C - Before SA	5	0.530 ± 0.059	0.467	0.600	-8.072 (0.001)
	C - After SA	5	0.703 ± 0.083	0.617	0.800	
One-Way ANOVA	C - Before SA	5	0.530 ± 0.059	0.467	0.600	21.953 (0.000)
	A - After SA	8	1.066 ± 0.146	0.867	1.267	
	B - After SA	8	1.538 ± 0.327	1.117	2.033	

Table 2: Bonferroni pairwise comparison test result.



		Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
C Before SA	A	- 0.537*	0.128	0.002	-0.875	-0.198
	B	-1.008*	0.128	0.000	-1.346	-0.669
A	B	-0.471*	0.113	0.002	-0.770	-0.172

Table 3: Descriptive statistical result for fracture resistance measure per group (N).

Groups	N	Mean ± SD	Minimum	Maximum	One-Way ANOVA F-value (P-value)
C	5	1848.000 ± 342.155	1240.000	2040.000	3.164 (0.066)
A	8	1475.000 ± 333.766	1130.000	2140.000	
B	8	1400.000 ± 302.844	1020.000	2020.000	

Effect of Surface Roughness on Fracture Resistance: linear regression modeling was built to highlight the variability between surface roughness and fracture resistance. Figure 5 shows that there was a strong significant positive correlation between the surface roughness and fracture resistance in A and B groups.

Discussion

Zirconia is being used extensively in prosthetic dentistry because of its excellent biocompatibility, low cytotoxicity, chemical stability, high mechanical strength, superior fatigue resistance, high fracture resistance, and Young's modulus comparable to that of stainless steel alloy. [15] Zirconia is an oxide ceramic with three different crystalline structures (monoclinic, tetragonal, and cubic). [16] Below 1170°C, zirconia transforms into the weakest of its three allotropes, monoclinic zirconia. Cubic zirconia forms at temperatures over 2370°C, whereas tetragonal zirconia forms between 1170 and 2370°C. [17]

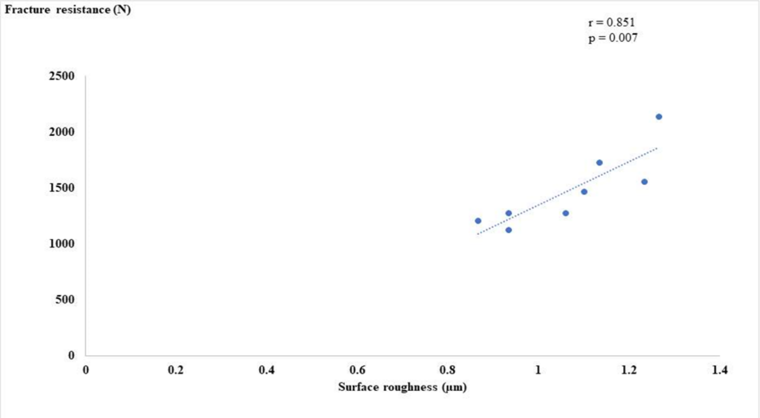


Figure 5. (a) Correlation between surface roughness and fracture resistance in A group.

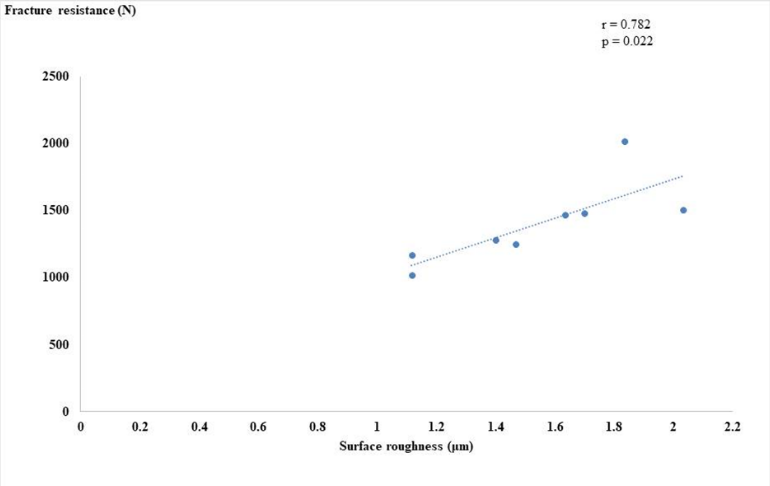


Figure 5. (b) Correlation between surface roughn ess and fracture resistance in B group.

Chipping or fracture of the veneering ceramic is the leading cause of clinical failure in veneered zirconia-based restorations. [18] Depending on the composition and microstructure of the material, a variety of intraoral healing techniques for ceramic restorations have been studied and documented .[19–21]

In the present study, following the manufacturer's recommendations, the preparation parameters were chosen to give adequate thickness for both zirconia cores and veneering material. Since the modulus of elasticity of PMMA is comparable to that of human dentin, [22,23], this material was selected for the manufacturing of definitive dies.

Before applying the repair material, airborne-particle abrasion was selected as the surface treatment approach in this investigation. Airborne-particle abrasion (usually sandblasting with alumina particles) is commonly used to remove contamination, roughen the substrate surface, and modify the wettability and energy of the substrate. [24] In the present study, the sandblasting parameters, including the size of particles, distance, pressure, and duration, have been selected based on the previous study performed by Okada et al, [25] as this combination yielded the optimum flexural strength results. In terms of sandblasting effect, the results of the present study showed that the surface roughness of group B was significantly higher than group A. This could be explained by the fact that silica particles become embedded onto the substrate surface and promote chemical bonding at the ceramic-resin interface , [6–28]. However, Nagaoka et al [29] observed that certain silica particles were not embedded into the zirconia surface and did not combine with other particles, which revealed that the tribochemical reaction was incomplete. Literature shows controversy regarding the values of surface roughness following sandblasting, results of this study go in line with the findings of De Queiroz et al, [30] which stated that air abrasion using silica-coated alumina particles at 2.5 bar pressure produced more favorable 3D surface roughness characteristics for micromechanical retention when compared with alumina particles alone. However, in a study performed by Turp et al, [31] sandblasting with 50 µm aluminum oxide showed higher surface roughness values than silica-coated particles, in contrast to the results of the current study. This might be attributed to the different types of zirconia used in that study. Laboratory sandblasting with 110 µm aluminum oxide showed lower surface roughness values when compared with other groups, this could be attributed to the larger distance between the tip of the sandblaster and the specimen. [32]

The zirconia cores for both groups A and B were treated with a specific repairing kit which provides the basis for chemical bonding as it contains 10-methacryloyloxydecyl dihydrogen phosphate (MDP) which forms phosphate-oxygen-zirconium bonds with zirconia. [33] Nagaoka et al, [34] postulated that 10-MDP monomer is either adsorbed onto the zirconia surface by hydrogen bonding or interacts with zirconia through ionic bonding. Yue et al, [35] stated that the combination of air abrasion and treatment with MDP-based products resulted in high bond strength values and chemical affinity as a consequence of improved surface wettability from air abrasion and higher bond strength from the treatment with MDP-based primers. However, Sanohkan et al , [36] observed that the shear bond strength values between zirconia ceramic and resin composite were not significantly influenced by the use of different primers.

In terms of fracture resistance, the results of the present study go in line with the findings of Alsadon et al, [37] which stated that the outcomes of crowns made from a zirconia coping and veneered with light-cured composite were not statistically different from those veneered with feldspathic porcelain. The sandblasting treatment performed to zirconia surfaces appears to create protective compressive residual stresses from the tetragonal-to-monoclinic transition, hence enhancing flexural strength. [24] However, Okada et al ,[25] reported that excessive sandblasting pressure led to the production of microcracks that lowered the flexural strength. All the specimens in the present study exhibited fracture resistance values higher than the maximum bite force in the molar region as documented by Varga et al [38]. It is important to note that this investigation was conducted in vitro; accordingly, fracture resistance ratings may differ from clinical settings as only vertical axial forces were applied; however, in clinical situations, lateral forces and fatigue loading also play a significant effect.

Conclusion

The following conclusions can be drawn from this study:

1. Clinical sandblasting of zirconia cores with silica-coated particles produced higher surface roughness values
2. In terms of fracture resistance, all the specimens showed values above the acceptable clinical limit.

Conflicts of interst

The authors reported no conflicts of intersts

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