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Effect of collar diameter and cement space thickness on fracture resistance of two types CAD/CAM zirconia retained implant restorations

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Effect of Collar Diameter and Cement Space Thickness on Fracture Resistance of Two Types Computer Aided Designs/Computer Aided Manufacturing Zirconia Retained Implant Restorations

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Abstract

Purpose: The purpose of the study was to evaluate the effect of collar diameter and cement space thickness on fracture resistance on two types of implants retained computer-aided design/computer-aided manufacturing CAD/CAM zirconia crowns. **Patients and methods:** Zirconia crowns were designed and constructed by computer-aided design/computer-aided manufacturing and then divided into two groups; group (1) crowns were made from bilayer zirconia while group (2) crowns were made from monolithic zirconia. The two groups were divided into two subgroups according to the collar diameter of the implant; subgroup (A) with a collar diameter of 2.5 mm, and subgroup (B) with a collar diameter 4.5 mm. Furthermore, all subgroups were divided into three divisions according to cement space thickness; division 1 with a cement space thickness 50 μm, division 2 with a cement space thickness 100 μm, and division 3 with a cement space thickness 200 μm. All crowns were adhesively bonded to their corresponding implant abutments by using dual cure self-adhesive resin cement and subjected to thermo-mechanical fatigue simulating 3 months of clinical situations fracture resistance was measured until the fracture of crowns and then the fractured parts were evaluated under electron microscope. **Results:** A statistically significant difference ($P \leq 0.05$) in the fracture resistance values among tested groups was found. The tests showed significant impact of cement space thickness on the fracture strength of zirconia crowns ($P \leq 0.05$). **Conclusions:** Increasing cement space parameter setting significantly improved the fracture strength of implant-retained zirconia restorations. There was no statistically significant difference between the narrow collar diameter of the implant and the normal diameter. Fracture resistance of monolithic zirconia crowns was significantly higher than bilayered zirconia crowns.

Keywords: Cement space, Fracture resistance, Implant diameter, Monolithic zirconia

1. Introduction

Zirconia ceramics have been widely tested and have several applications in dentistry. Zirconia restoration can be machined from a mono-block without veneering or veneered with appropriate ceramic on a zirconia framework. However, ceramic veneer chipping is said to be among the most prevalent reasons of zirconia repair failure [1].

Improvements in computer-aided design/computer-aided manufacturing (CAD/CAM) design and manufacturing have facilitated fabrication of the monolithic mono-block restorations, resulting in the development of highly translucent zirconia blocks that allow the fabrication of Y-TZP monolithic anterior and posterior fixed prostheses without the issue of veneering/porcelain chipping [2].

The distance between the top edge of peri-implant soft tissue at the margins and the implant platform must be used to calculate the collar dimension of the prosthesis required to create a natural biological width and implant emergence profile for crowns

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based on dental restorations [3]. Abutment choice is based on the Gingival Height Index, which is established by the gingival sulcus depth, which it is guided by the occlusal height and vertical space enabled for the restoration. Deep sulcus in some patients typically requires longer abutment collars than shallow gingiva in other patients [4].

The collar diameter design includes horizontally repositioning the implant-abutment interface by employing an implant with a smaller neck or collar diameter, which was advantageous since it required vertical interspace less in dimension to construct the biological width of the existing peri-implant tissues. It indicates that, the distance which exists from the prosthetic abutment and the implant's collar is reversely related to the loss of crystal bone height, and that the implant collar diameter maintains peri-implant bone to an obviously identifiable level [5].

Other than that, the biological benefits will reduce stress induced in the bone that surrounds the implant. Stress presented on the crystal bone decreased when the collar diameter of the implant was decreased by 10% or 20% (from 4.5 mm to 4.0 mm) using the standard diameter of the abutment and irrespective to the micro-threads presented, smooth surface and the applied force direction (90 or 15°). However, some studies have examined the implant collar diameter design concept in the posterior region of the mandible biomechanically [5,6].

Zirconia's exceptional mechanical characteristics make it ideal for biomedical applications, particularly implantology. The crystalline phase transition is an important component in the biomechanical characteristics of zirconia [7].

In reviewing the strength and failure mechanism of the ceramic materials, the cyclic loading and the water presence must be involved in the laboratory research to correlate *in vivo* and *in vitro* biomechanical studies with each other. Regardless of these recommendations, static load fracture examination has been used in several implant-supported restorations in in-vitro studies [8]. These laboratory findings questioned the validity of the static load test, raising questions about the clinical usefulness of static load in the procedures of testing fracture, however, this test was used in many studies previously [9–11].

When selecting all-ceramic zirconia crowns, the most important criteria presented are mechanical strength, the critical space and clinical results, up to long periods. Minor spaces have been investigated in some *in vitro* and *in vivo* research. Improper crown or bridge cementation can result in cement breakdown and plaque formation. Buildup, marginal leakage, secondary caries, and eventual crown collapse are all possibilities. The clinically acceptable critical gap was 120 mm [12].

Several studies on large cement spaces for CAD/CAM systems have been conducted. When the cement space is big, the critical gap is small but it can cause a very huge internal gap, which negatively affects the cement's mechanical qualities. A worse interior fit has been observed to increase the chance of veneer failure [12–14].

The appropriate cement space dimensions are recommended to be 20–40 μm per wall. Thus the diameter of the tooth preparation must be less than the internal diameter of the full coverage crown by 40–80 μm. If so, a crown will not sit correctly during bonding if the cement gap is too small. If parts are set too far apart, the crowns might become loose on the teeth and readily detach during usage [14].

Consequently, the current study was to assess collar implant diameter effect and cement space thickness on two types of implants retained CAD/CAM zirconia restorations investigating fracture resistance of both crowns.

2. Patients and methods

The study design was approved by the Ethical Committee of the Faculty of Dental Medicine for Girls, Al-Azhar University and give Ethical code of REC-CR-23-08.

72 implant analogs with collar diameters of 2.5 mm and 4.5 mm (implant direct USA) were used in this study. Titanium abutments (implant direct USA) were screwed to these implant analogs representing mandibular first molar according to the instructions provided by the manufacturer. After that, all samples were centralized in a plastic cube (2 cm height, and 2 cm diameter) filled with epoxy resin (CMB. International, Egypt). The elastic modulus of Epoxy resin is approximately 12 GPa, which in turn mimics the human bone (18 GPa) [15]. Implant analogs were inserted in epoxy resin blocks parallel to their long axis facilitated by a dental surveyor (Dentaurum, Germany).

2.1. Samples grouping

The samples in this study were divided into two groups according to the crown fabrication technique in group (1) the crowns made of multilayered CAD/CAM zirconia, while group (2) crowns made from CAD/CAM monolithic zirconia. Then each group was divided into another two subgroups according to the collar diameter of the implant used, where in subgroup (A) the implant collar diameter was 2.5 mm, while in subgroup (B) the implant collar diameter was 4.5 mm. Furthermore, each sub group was divided into three divisions according to the cement space thickness used; division 1 with cement space thickness 50 μm, division 2 with a cement space thickness 100 μm, and division 3 with cement space thickness 200 μm.

2.2. Crown design

To produce 3D digital pictures of all teeth, every sample was scanned with an extraoral digital scanner (3-Shape, Copenhagen, Denmark) which has an ExoCad software, as shown in Fig. 1.

The thickness of occlusal surface of crowns were 1.5 mm for the monolithic zirconia and 0.8 mm for the frameworks made of zirconia with porcelain veneering layer, in which each of them assembled an anatomic form. While the buccal/lingual thickness was set at 1.4 mm, the cement space thickness was chosen depending on the sample division (50, 100, and 200 m).

2.3. Crown fabrication

Milling of zirconia crowns was done by using an MC X5 milling machine (Sirona Dental System, Germany). Then all samples were sintered following the manufacturer's guidelines at 1450 °C for 4 h and 50 min.

In Group 1, samples were milled from Ceramill Zolid FX multilayer zirconium oxide blocks (Amann Girrbach AG, Koblach, Austria). After that these samples were veneered with IPS E.max Ceram (Ivoclar Vivadent, Schaan, Liechtenstein) as recommended by the manufacturer. To ensure standardization of applying the porcelain veneer layer, the specially constructed knife was used to set up veneer layer dimensions as per some previous studies in which the porcelain was placed in multiple layers (liner, wash, dentin 1, dentin 2, and gloss) using specified brushes as recommended with the guidelines set by the manufacturer [16,17].

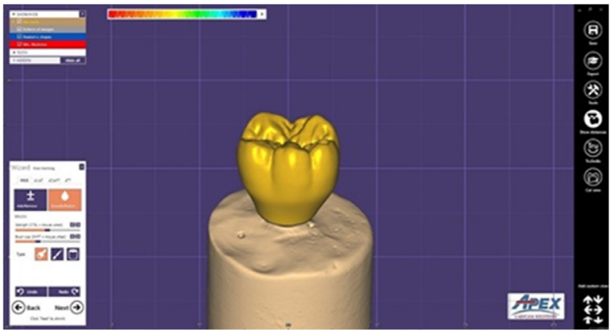

Fig. 1. Designing the crowns using ExoCad software.

The dimensions of this porcelain veneer layer were set according to manufacturer; in which axial wall was 0.5 mm and occlusal surface was 0.7 mm. All samples were designed by the same technician to ensure standardization. A digital micrometer was used (Langcheng company, China) to check the thickness of porcelain veneer layer.

In group (2) the samples were milled from Monolithic zirconia blocks (KATANA Zirconia HT). Following that, all samples were stained with Ivocolor Stains and glazed with Ivocolor Glaze Paste FLUO (Ivoclar Vivadent) at 710 °C, with a heating rate of 45 °C/min and long-term cooling to 450 °C according to the manufacturer guidelines.

2.4. Crown cementation

The samples were thoroughly cleaned in ultrasonic bath filled with ethanol before being dried with air. Then the samples were cemented to their respective abutments after being sandblasted for 5 s with aluminium oxide (AL2O3), 110 m grit, 10 mm distance, 3.5 bar compressed air pressure, and then cleaned with a steam cleaner and compressed air, as recommended by the manufacturer [18].

Cementation was done by Theracem, a self-adhesive resin cement (BISCO, USA) in a specially designed cementing device with a static load of 2 kg. Excess cement was removed after initial light cure for 2 s using a sharp scaler at the gel stage of the resin. Then resin cement was chemically cured for approximately 2 min, followed by final 20 s light curing for each surface.

2.5. Fracture load test

All samples were then stabilized in a universal loading machine with an average load of 50 N for 120 000 cycles [19]. After that an axial compression loading was applied on all samples by a crosshead with speed (0.5 mm/min) with the aid of a universal testing machine. Failure load and data were collected in a stress- strain curve with the associated digital software. Analysis of the fractured samples was done by using a stereomicroscope to identify the mode of failure. Fracture mode analysis was performed using a magnifying lens to classify the failure modes according to Burke classification, as shown in Table 1.

2.6. Failure modes of zirconia crowns

Fractured samples were further examined using an electron microscope to determine the mode of failure. Although there were five modes of fracture

Table 1. *Fracture modes of zirconia restorations according to classification of Burke's [20].*

Fracture Mode	Definition
Mode 1	Minimal fracture or crack in crown
Mode 2	Less than half of crown lost
Mode 3	Crown fracture through midline (half of crown is displaced or lost)
Mode 4	More than half of crown is lost
Mode 5	Severe fracture of implant abutment and/or crown

according to the Burke classification, results showed that the fracture patterns recorded can also be classified into one of the following three types:

(a) Type 1 represented failure in the zirconia/resin cement interface (Adhesive failure).
(b) Type 2 represented failure within abutment or resin cement (Cohesive failure).
(c) Type 3 represented a mixed pattern of failure seen at zirconia/resin cement interface and within abutment (Adhesive and Cohesive failures).

2.7. Evaluation of fractured segments [21]

A stereoscopic microscope was used to study the broken segments, and the causes of fracture presented were further investigated. The broken samples were extensively disinfected in 95% ethanol using an ultrasonic cleaning mode for 15 min before being coated with a gold-palladium sputter coating.

The test samples were inspected beneath a polarized light microscope (AxioZoom V.16, Carl Zeiss Microscopy, Oberkochen, Germany) shortly after fatigue in order to identify the point of fracture origin, the direction of crack propagation (DCP), hackles, and compression curves, which were referred to by red arrows and white lines that fled in conjunction with the arrest line.

Two samples of each group were then analyzed for the type of the qualitative fractographic evaluation whether cohesive or adhesive fractures under electron microscope scanning (Quanta 250-FEG, FEI, Netherlands).

3. Results

The data was examined with the SPSS 20.0 software (SPSS, Chicago, IL, USA) at a 0.05 significance level. A one-way analysis of variance (ANOVA) was used to determine the covariates' significant impacts. The data were compared using Tukey's multiple comparison test.

The test values of the zirconia materials employed in this investigation regarding their resistance to fracture differed significantly. The differences were verified by one-way ANOVA ($P \leq 0.000$). Comparisons of the test results are shown in Tables 2 and 3.

The test results demonstrated significant variations in fracture resistance ($P \leq 0.05$) between the examined materials when fixed to their abutments.

T test revealed a substantial variation in fracture strength for crowns made with various cement space thicknesses and the same luting agent, as shown in Table 4.

Also, the T test was performed to validate the difference associated between subgroups with various collar diameters, regardless of crown type, as shown in Table 5.

3.1. Fracture types

Observation of the examined samples visually revealed three types of failures mixed, cohesive and/or adhesive fractures. For group 1 (multilayer zirconia) the fractured crowns showed cohesive failure pattern within the veneering layer. While for group 2 (monolithic zirconia) the fractured crowns parts showed mostly bulk fracture (through and through) as shown in Figs. 2 and 3.

Meanwhile, following Burk's classification, the fracture types were mainly mode-II and mode-III

Table 2. *ANOVA results of the fracture resistance evaluation for group 1 with descriptive statistics (mean and standard deviation) (N = 36).*

Group/sub group (bilayer zirconia crowns)	Cement space thickness	N	[a]Mean (MPa)	SD	ANOVA *P*-Value	95% Confidence interval for mean	
						Lower band	Upper band
Sub group A: bilayer crown with collar diameter 2.5 mm	50 μm	$N = 6$	1.135	0.305	0.000	0.917	1.353
	100 μm	$N = 6$	1.955	0.187	0.000	1.821	2.088
	200 μm	$N = 6$	2.417	0.341	0.000	2.253	2.697
Sub group B: bilayer crown with collar diameter 4.5 mm	50 μm	$N = 6$	2.47	0.31	0.000	2.173	2.661
	100 μm	$N = 6$	2.59	0.39	0.000	2.310	2.880
	200 μm	$N = 6$	3.029	0.262	0.000	2.842	3.216

[a] Average fracture resistance was measured in mega Pascals (MPa).

Table 3. ANOVA results of the fracture resistance for group 2 with descriptive statistics (mean and standard deviation) ($N = 36$).

Group/sub group (Monolithic zirconia crowns)	Cement space thickness	N	[a]Mean (MPa)	SD	ANOVA P-Value	95% Confidence Interval for Mean	
						Lower band	Upper band
Sub group A: monolithic crown with collar diameter 2.5 mm	50 μm	$N = 6$	2.198	0.328	0.000	1.822	1.343
	100 μm	$N = 6$	2.695	0.330	0.000	2.088	2.173
	200 μm	$N = 6$	3.040	0.730	0.000	2.661	2.835
Sub group B: monolithic crown with collar diameter 4.5 mm	50 μm	$N = 6$	2.445	0.341	0.000	3.270	3.475
	100 μm	$N = 6$	2.565	0.4391	0.000	3.377	3.594
	200 μm	$N = 6$	3.480	0.4981	0.000	4.326	4.832

[a] Average fracture resistance was measured in mega Pascals (MPa).

Table 4. T test comparison of fracture strength among subgroups with different cement spacing thicknesses.

Group	Sub group	Cement space thickness (50 μm)	Cement space thickness (100 μm)	Cement space thickness (200 μm)	P value
Group 1: Bilayer zirconia crowns	Subgroup A (2.5 mm collar diameter)	1522.14	1527.77	1654.15	0.127
	Subgroup B (4.5 collar diameter)	1688.75	1804.52	1936.25	0.019875[a]
Group 2: Monolithic zirconia crowns	Subgroup A (2.5 mm)	1857.8	1933.37	2488.42	0.377
	Subgroup B (4.5 mm)	2645.39	3090.91	3110.91	0.011[a]

[a] Differences can be considered statistically significant ($P \leq 0.05$).

Table 5. Comparison between different subgroups according to Fracture Load related to collar diameter.

	Subgroup A1 ($n = 16$)	Subgroup A2 ($n = 16$)	Subgroup B1 ($n = 16$)	Subgroup B2 ($n = 16$)	$KW\chi^2$	P
Fracture load						
Min–max	1493.8–3562.3	1745.1–3527.9	3415.6–5631.3	4881.3–7637.5		
Mean ± SD	2625 ± 915.08	2950.4 ± 705.9	4263.8 ± 867.9	6104.4 ± 992.2	14.554[a]	0.002[a]
Median	2509.38	3078.13	4281.25	5996.88		
$P1$	0.016[a]	0.009[a]	0.016[a]			
$P2$	0.028[a]	0.016[a]				
$P3$	0.917					

[a] Differences can be considered statistically significant ($P \leq 0.05$).

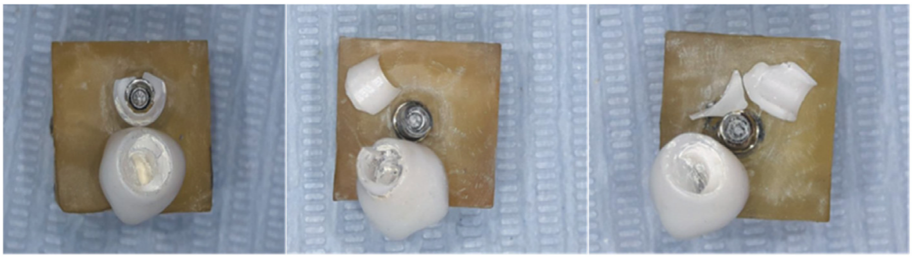

Fig. 2. Fractured patterns of examined samples.

fractures, with no damage to the abutments. In group 1, 16 samples shattered into four parts, while another four samples were entirely split into five pieces.

Samples in group 2 showed that most of them were broken down into more than six pieces (14 samples), while 13 samples shattered into four pieces, and the last three samples were split into five pieces.

3.2. SEM and polarized light analysis

The fragment analysis exhibited that the origin of the fracture was located occlusally, which was in

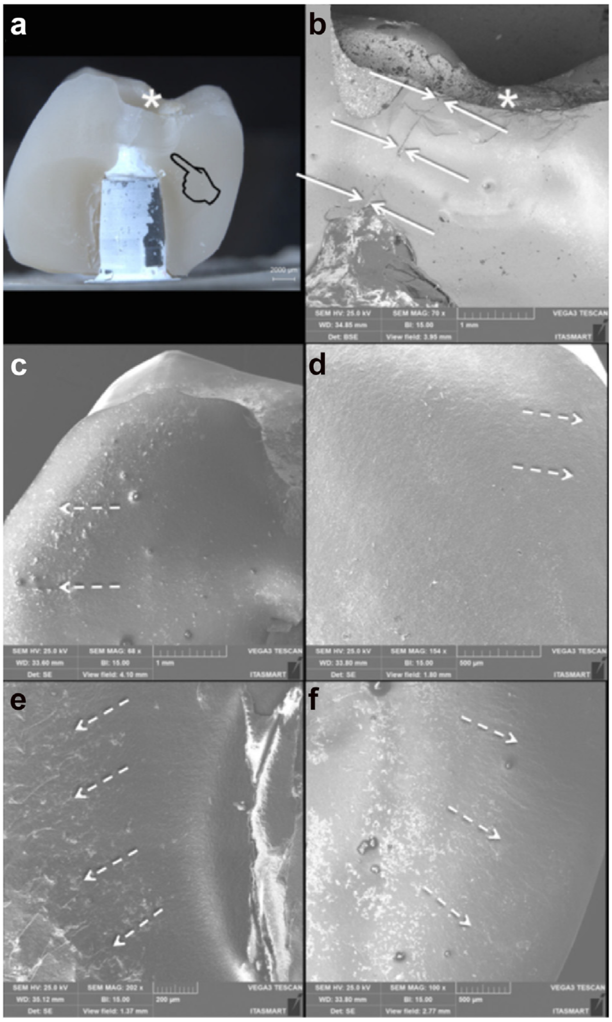

Fig. 3. (a) Illustrated a cracked sample of multilayered zirconia under investigation of Polarized light micrograph in which arrest lines (pointer) with their concave section were pointed towards the origin, i.e., indentation region (asterisk). (b) Further illustration of the indentation region under SEM backscattered magnification, in which this region exhibited a cone fracture (bounded by solid arrows) that extends till the occlusal surface. (c) and (d) showed fracture propagation direction identified by line dots which was illustrated further in magnifications of the left and right cusps appeared in fig A. (e) and (f) showed both sides magnifications of the fractured surface's margins as in fig A, where hackle lines demonstrated path of the crack from the occlusal surface to the fractured edges.

touch with the indenter in the majority of the samples. On the other hand, the zirconia fragments that resulted from the fracture were separated which showed a bulk fracture, as shown in Fig. 3a.

The broken samples revealed various crack propagations, arrest lines, and twisted hackle lines as a consequence of the SEM investigation, as shown in Fig. 3b.

The fracture originated in the occlusal surfaces primarily at the loading surface's contact point. In group 1 multiple halted fracture propagation was found, as shown in Fig. 3c and d.

While in group 2 twist hackle lines were unusually noted, as were halted crack propagation lines, as shown in Fig. 3e and f.

There were no voids discovered at the bonding interface between the implant abutment and the crown, indicating that close contact and a suitable internal fit were achieved. Additionally, the fractured fragment was analyzed using SEM as shown in Figs. 4 and 5.

4. Discussion

Single implant-retained restorations have several documented aesthetic and practical benefits. The preservation of healthy neighboring teeth, decreased bone resorption around the implant, and accessibility for good oral measures between the crown cemented on the implant and neighboring natural teeth are just a few of these advantages [22].

The implementation of (CAD/CAM) technology in dentistry gave permission to dentists for using creative methods in treating and modifying design and fabrication of esthetic restorations as prerequisites for use in the mouth especially in the posterior region [23].

Because there is little scientific proof that only one type of restoration material fits all of the parameters for effective restoration about implant-supported prostheses, the prosthetic material selection remains problematic [24].

This study aimed to determine fracture resistance of two CAD/CAM zirconia implant retained crowns with varying cement gap thickness and implant collar diameter. There has been no previous research studying the impact of implant collar diameter in the posterior region on the fracture resistance of zirconia crowns used.

Regardless of cement gap thickness utilized with zirconia crowns; group 2 demonstrated significantly higher fracture resistance for both collar diameter measures than the other examined groups. Cohesive failure was found on the tip of the cusps, while a line of fracture was visible along the buccal and lingual portions of the crowns in these samples. This conclusion is consistent with the findings of a recent study that demonstrated some patterns of fracture that resembled the present study in which this previous study illustrated that fracture was the most prevalent reason for ceramic failure [25].

Several studies have shown that using a narrow collar diameter did not affect the survival rate of crowns used in various situations clinically and by using different surgical techniques. In majority of

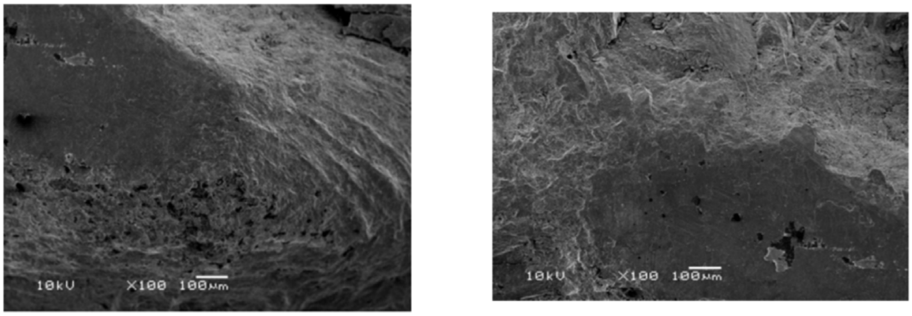

Fig. 4. Fractographic analysis of interfacial fracture in Bilayered zirconia sample under SEM.

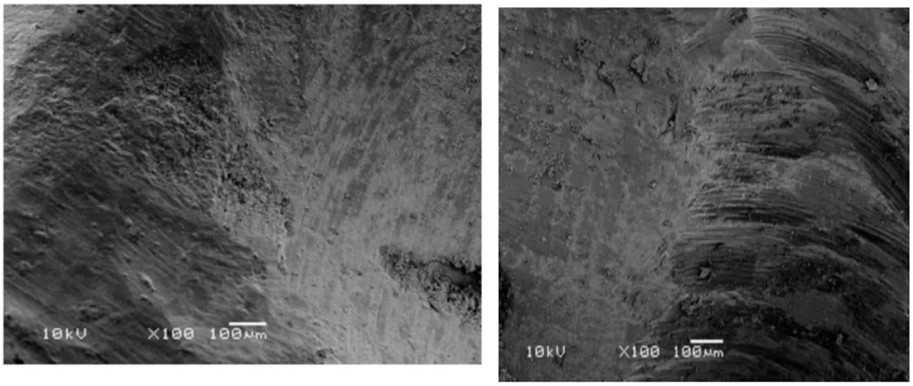

Fig. 5. Fractographic analysis of interfacial fracture monolithic zirconia under SEM.

cases, excellent satisfactory outcomes were observed ensuring cumulative survival rates for long periods of study comparable to those reported in larger diameter implant restorations (between 94 and 100% survival rates) [26–28].

The implant collar configuration seems to be of interest in clinical research. Previous research has shown the place in bone where distribution of stress occurs mostly near the implant's neck upon loading. The fracture resistance of zirconia cement-maintained crowns differed significantly between narrow and wide collar diameters in this investigation [29,30].

Regarding the type of crowns used in this study, group 2 which included monolithic zirconia crown material showed higher fracture resistance than group 1 which included multilayered zirconia crown material regardless of the type of collar diameter or cement space thickness used.

Some in-vitro studies illustrated the maximum loading capacities ranging between 4700 and 6350 N for three types of tertiary yetria stabilized monolithic zirconia crowns on one- or two-pieces zirconia implants (Vita YZ T, Lava Plus and Zerion LT) [28,29].

In contrast, earlier research had shown that identical fatigue measurements that resembled five years of clinical service were utilized to further emphasize the restorative implant complex, according to the findings of the current study different implant collar sizes (4.5 mm), two-piece designs, abutment heights (4.0 mm), adhesive cement options, and crown design changes can all be used to explain differing failure stress ranges. In addition, in comparison with the restorations that depended on locally built or ready-made abutments, the relation of small shape implants to the same attached abutment would result in more thicker restoration walls and, as a result, bigger failure stresses [30].

Fractographic analysis and loading until failure of monolithic and multilayer zirconia restorations revealed that under stress, the fracture of monolithic crowns began in the fissure, whereas chipping of the veneering layer was the most likely fracture for multilayer zirconia. Tensile stresses are imparted to the material in the fissure, which is the least desirable loading condition and eventually culminates in a bulk fracture.

Other studies have found the most common patterns of failure, such as cohesive failure in monolithic zirconia-ceramic restorations, while adhesive failures were most shown in multilayered zirconia ceramic restorations [31,32].

According to the investigators, zirconia's improved fracture resistance could be due to the zirconia core's increased crystalline concentration, fracture toughness, and flexural strength [33].

In relation to cement space thickness, this study found that zirconia restorations with a larger cement space thickness had higher fracture strength than those with a smaller cement space thickness. These findings were consistent with earlier research [31,34]. Previous researches, on the other hand, found that increasing cement spacing thickness was associated with fewer seating discrepancies and thinner cement thickness [35,36].

Decreasing the hydraulic pressure during crown sitting to provide a thinner, well-distributed cement. This was accomplished by increasing the thickness of the created crown's cement gap allowing escape of excess cement when cementing the crown. Despite statistically significant variations in fracture strength between subdivisions, the mean crown fracture strength in all groups surpassed the maximal strength of the bite in the molar area [37].

Crown-manufacturing techniques such as scanning, designing, and milling, for example, may alter the adaptability of the machined restoration, resulting in internal errors and slightly too much contact at the tooth-restoration junction [38].

More importantly, in the final stage of sintering during milling of zirconia restoration, it is susceptible to shrinkage in size by 20–30%; this issue might result in several immature connections within the interface; consequently, elimination or diminution of this prematurity as there is more cement gap might clarify the conclusions of this study, which were in accordance with few prior investigations [39,40].

4.1. Conclusions

With the limitation of this study it could be concluded that,

(a) The larger the implant collar diameter, the more improved the fracture resistance of the crowns compared with the smaller ones.
(b) There were substantial variations in fracture resistance amongst the groups examined, with monolithic zirconia crowns having the greatest fracture resistance values and multilayered zirconia crowns having the lowest.
(c) All obtained fracture resistance values were within the clinically accepted ranges.
(d) Increasing cement-space thickness can improve the fracture resistance of crowns. Applying 100–200 μm. space thickness might be favorable for a better fracture strength of the monolithic crown restoration.

4.2. Recommendations

Further studies are recommended to evaluate the marginal and internal fit of zirconia crowns on implant under mechanical loading conditions have more close estimation of the restoration clinical performance.

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Ethics approval

The study design was approved by the Ethical Committee of the Faculty of Dental Medicine for Girls, Al-Azhar University and give Ethical code of REC-CR-23-08.

Biographical information

This study was conducted at clinic of Fixed Prosthodontics Department, Faculty of Dental Medicine for girls, Al-Azhar University, Cairo, Egypt.

Conflict of interest

There are no conflicts of interest.

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