Evaluation of Fracture Resistance of Varying Thicknesses of Zirconia Around Implant Abutment Cylinders

Cheryl J. Park, DDS
Jin-Ho Phark, DDS, Dr Med Dent
Winston W. Chee, DDS

Zirconia is becoming increasingly used as a restorative material for implant-supported restorations; however, information is lacking with respect to the minimum thickness of zirconia surrounding the implant components. The purpose of this study is to evaluate the resistance to fracture of different thicknesses of zirconia luted to implant components. Thirty cylinders of zirconia (Prettau, Zirkonzahn) with 13-mm height, designed with indented occlusal surface for loading, and varying wall thicknesses (0.5 mm, 1 mm, 1.5 mm; n = 10/group) were milled using a computer-aided design/computer-aided manufacturing system (Modellier, Zirkonzahn), after which they were sintered. Titanium temporary cylinders (ITCS41, Biomet3i) were attached to 30 implant analogs (ILA20, Biomet3i) that were embedded into polymethylmethacrylate blocks (Palapress Vario, Heraeus Kulzer) with dimensions of $4.5 \times 1.8 \times 2$ cm. Zirconia specimens were cemented to the titanium cylinders using a self-adhesive, dual-cure resin cement (Panavia SA, Kuraray). Load to failure test was performed under compression until fracture using a universal testing machine (Instron5965, Instron) at a crosshead speed of 0.5 mm/min and measured in N (Newton). Statistical analysis was performed using 1-way analysis of variance and Tukey B test at $\alpha = .05$ (SPSS19, IBM). Mean load to failure was 1059.94 N, 2019.46 N, and 4074.79 N for groups 0.5 mm, 1 mm, and 1.5 mm, respectively. Values were significantly different between the groups ($P < .05$). Study limitations are that it is in vitro, specimens do not replicate tooth dimensions, and forces are static and directed toward the occlusal portion of each specimen. Within these limitations and considering the average human bite force, a thickness of 0.5 mm to 1 mm of this particular type of zirconia around this type of implant component can avoid fracture with these dimensions.

Key Words: implant-supported zirconia, fracture, zirconia thickness, peri-implant component thickness

INTRODUCTION

In the rapidly changing field of dentistry, the drive to achieve an esthetic outcome while maintaining the strength and durability of a restoration has resulted in zirconia being selected as a material of choice in a growing number of indications.[1,2]

Zirconia is a polycrystalline ceramic that exists in 3 allotropic forms. Upon cooling, zirconia goes through phase changes from cubic (>2370°C) to tetragonal (1170°C–2370°C) and finally to a monoclinic phase.[3]

For dental restorations, the tetragonal phase is the most desirable because of its excellent physical and superior optical properties when compared with metallic restorative materials. The tetragonal form of zirconia is unstable at room temperature; however, this can be mitigated by adding a small amount of yttria to stabilize this allotropic form at room temperature.[4]

The major advantage of zirconia is high fracture resistance, which is represented by the high flexural strength (900–1200 MPa) and fracture toughness of 9–10 MPa.m$^{1/2}$ compared with other all-ceramic core materials, whose flexural strength (alumina-based glass infiltrated ceramic, 400 Mpa; densely sintered alumina, 650 MPa) and fracture toughness (alumina-based glass infiltrated ceramic, 4.78 MPa.m$^{1/2}$; densely sintered alumina, 4.48 MPa.m$^{1/2}$) are inferior.[5–10]

These physical properties can be attributed to "transformation toughening," which can be described as an allotropic change from tetragonal to monoclinic when stress is applied.[1,4,11] This phenomenon results in 4.4% volumetric increase of the grains surrounding the crack tip, which retards crack propagation.[11]

The use of zirconia has been growing in single- and multiple-unit implant-supported restorations. The most commonly reported clinical complications with zirconia-based restorations has been reported to be chipping of layered porcelain.[12–14] The failure of the zirconia framework itself has been sparsely reported in the literature.[15] Recently, publications are emerging that describe use of implant-supported, multi-unit, complete-arch restorations with zirconia frameworks.[5,16–20] Many authors describe complete-arch implant-supported zirconia restorations in the literature with various framework designs.[15–17,21–26]

Since few reports of framework fracture exist, monolithic zirconia restorations are being used both with tooth-borne or implant-supported restorations to maximize the durability of

Herman Ostrow School of Dentistry, University of Southern California, Los Angeles, Calif.
* *Corresponding author,* email: jongheep@usc.edu
DOI: 10.1563/aaid-joi-D-16-00204

the restoration and to avoid complication of porcelain chipping.[17,26] Although it is tempting to extrapolate from the evidence of the strength of monolithic zirconia, there is limited information available to support widespread clinical use for implant-supported zirconia prostheses.

When machining zirconia restorations, it is more common to use a soft machining process in which the non- or partially-sintered zirconia is machined. This reduces wear on the machining bits used to form the restoration and reduces the induction of cracks upon machining. After formation of the restoration, it is sintered to affect a phase change to the stronger tetragonal allotrope. The sintering process causes an approximate volumetric shrinkage of 25%. Although compensated for with a computer algorithm, this step can cause a misfit from the zirconia structure to the implant. This can be compensated by cementing metal abutments into the implant zirconia framework to serve as a connection to the implant platform. This also avoids complications of forces from the screw to the zirconia frame and of differential wear that arises when zirconia has a direct connection to the implant platform.[27–30]

However, information is still lacking regarding the optimal wall thickness of zirconia surrounding the abutment components (Figures 1 and 2). Therefore, this study was conducted to isolate the factor of material thickness around the abutment components by using a simplified specimen design.

The purpose of this study is to evaluate the resistance to fracture of zirconia with different wall thicknesses luted to implant components. This may provide information toward the design of zirconia prostheses for implant-supported restorations. The null hypothesis is that there will not be any significant differences of fracture resistance between the varying thicknesses of zirconia luted to implant components.

Materials and Methods

Thirty cylinders of monolithic Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) zirconia of 13-mm height and varying wall thicknesses (0.5 mm, 1 mm, 1.5 mm) were designed using computer-aided design (CAD) software (Modellier Software, Zirkonzahn, Gais, Switzerland). All specimens were designed with an indented occlusal surface for experimental loading. Samples were milled out of a monolithic 18-mm height zirconia block (Prettau Zirconia, Zirkonzahn) using a 5 + 1 axis milling unit (M5, Zirkonzahn) and sintered according to the manufacturer's recommendations. The final thickness was confirmed using a digital caliper (Absolute Digimatic Caliper, Mitutoyou, Sakada, Japan).

Thirty external hex implant analogs (ILA20, Biomet 3i, Palm Beach Gardens, Fla) were embedded into self-curing acrylic resin (Palapress Vario, Heraeus Kulzer, Wehrheim, Germany) with ratio of 10 g powder:7 mL liquid using a silicone mold (dimensions: 4.5 cm × 1.8 cm × 2 cm). Titanium hexed abutment screws were torqued to 20 Ncm to connect titanium temporary cylinders (ITCS41, Biomet 3i LLC) to the implant analogs according to the manufacturer's recommendations. The zirconia cylinders were cemented to the titanium cylinders using a self-adhesive, dual-cure resin cement (Panavia SA, Kuraray, Tokyo, Japan). Excess cement was removed with microbrushes, and the specimen was light cured using a light-emitting diode–curing unit (10 seconds, 1000 mW/cm^2; Valo, Ultradent, South Jordan, Utah). Prepared specimens were kept in room temperature in a dry environment for 24 hours before fracture testing was performed.

All specimens were subjected to load to failure test under compression until fracture using a universal testing machine (Instron 5965, Instron, Norwood, Mass). Fracture strength data were recorded in Newton (N) by the computer controlling the universal testing machine and was exported to an Excel spreadsheet (Excel 2016, Microsoft, Redmond, Wash). Data for the dependent variable (wall thickness) and independent variable (fracture strength) were entered in separate columns and subsequently transferred to a statistical software (SPSS 19, IBM, Armonk, NY) for further analysis. All tested specimens were included, and data were not transformed for analysis.[31]

The primary objective of this investigation was to evaluate the fracture strength of milled, monolithic Y-TZP zirconia cylinders with varying wall thickness (0.5, 1.0, and 1.5 mm), which were adhesively cemented to titanium temporary cylinders. One-way independent analysis of variance (ANOVA) with post hoc Tukey B test was used to evaluate differences between the 3 groups differing in wall thickness.

Results

Mean load to failure was 1059.94 N, 2019.46 N, and 4074.79 N for groups 0.5 mm, 1 mm, and 1.5 mm, respectively (Table 1, Figure 3). Load to failure values ranged from a minimum of 503.22 N in the 0.5-mm group to 4647.75 N in the 1.5-mm group.

In all samples tested, screw loosening was noted after load to fracture. None of the lab analogs or the temporary abutments were damaged when visually inspected. A fractured specimen is shown in Figures 4 and 5.

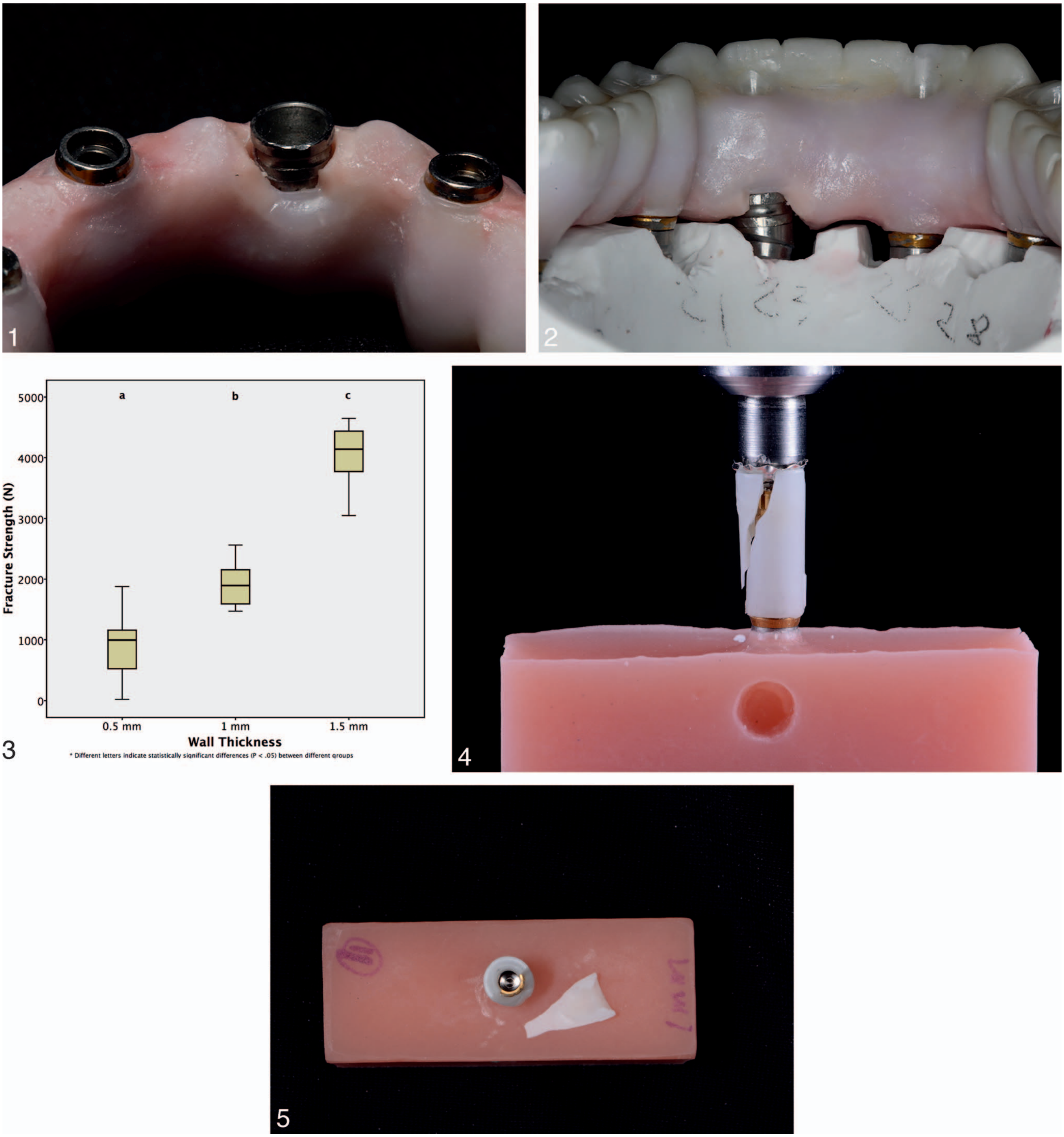
Prerequisites for ANOVA (normality and homoscedasticity) according to Kolmogorov-Smirnov and Levene tests were met ($P > .05$). ANOVA showed an overall significant difference between the groups ($F = 135.43$, $df = 2$, $P < .001$). Groupwise comparison using Tukey B test showed a significant difference between the groups ($P < .05$; Table 2).

The mathematical assumptions for the Tukey test were (1) normal distribution of data, (2) homogeneity of variance (homoscedasticity), and (3) independence of observations within and among the groups. All assumptions were met. While normality of distribution was confirmed by Kolmogorov-Smirnov test ($P = .200$), homogeneity of variance was confirmed by Levene test ($P = .648$). Since each specimen was fabricated separately by CAD/computer-aided manufacturing (CAM) procedures, independence of observation within and among groups is ensured (Tables 3 and 4).

Discussion

The null hypothesis was rejected. There is a significant difference of fracture resistance between the varying thicknesses of zirconia around implant components.

Although long-term clinical data on the survival and



Figures 1–5. Figure 1. Example fracture of zirconia framework around implant abutment. **Figure 2.** Example fracture of zirconia framework around implant abutment. **Figure 3.** Load to fracture data (N) of different zirconia wall thicknesses. **Figure 4.** Fractured specimen under load. **Figure 5.** Fractured 1-mm specimen.

complications of implant-supported zirconia prostheses are limited, some in vitro studies have reported fracture resistance of zirconia abutments to be close to 593 N in single implant-supported restorations.[5,32] However, in comparing our experiment with previous studies, we have to note some differences in specimen design. Att et al[32] had zirconia abutments with metal inserts for connection to the implants onto which anatomic zirconia crowns were cemented. Unlike this study, our experimental design consists of a direct connection of the zirconia restoration to titanium abutment. With these differences in mind, we can extrapolate from the fracture pattern of the cervical portion of zirconia in the study by Att et al that this is the area of highest stress concentration, which is vulnerable to fracture because of the minimal volume of zirconia.[32]

To investigate the area vulnerable to fracture around implant-supported zirconia restorations, zirconia cylinders with

TABLE 1

Fracture strength data for each zirconia wall thickness group: sample size, mean, minimum, and maximum fracture strength values (N)*

Wall Thickness, mm	Sample Size, n	Fracture Strength, N Mean ± SD	Minimum	Maximum
0.5	10	1509.94 ± 424.48[a]	503.22	1878.57
1.0	10	2019.46 ± 330.11[b]	1591.13	2561.62
1.5	10	4074.79 ± 466.16[c]	3049.31	4647.75

*Different superscript letters indicate statistically significant differences ($P < .05$) between different groups.

TABLE 3

Output for test of normality (Kolmogorov-Smirnov)*

Wall Thickness		Kolmogorov-Smirnov Statistic	*df*	Significance
Fracture strength, N	0.5 mm	.147	10	.200†
	1 mm	.137	10	.200†
	1.5 mm	.194	10	.200†

*Lilliefors significance correction.
†This is a lower bound of the true significance.

varying thicknesses around titanium abutments were tested in our study. The results suggest keeping a minimum circumferential thickness of 0.5 mm to 1 mm of monolithic zirconia may be sufficient to withstand the average molar occlusal force in humans (597 N in women, 847 N in men).[33] The average fracture resistance of the 0.5-mm group was 1059.94 N, which is above the average molar occlusal force in humans. However, we must consider the lowest fracture resistance of 503.22 N, which was also present in the 0.5-mm group. This suggests that increasing thicknesses will significantly increase strength, and 0.5 mm might be the thinnest acceptable thickness for clinical use. A possible explanation for the lowest fracture strength of the 0.5-mm group could be related to confounding factors such as milling discrepancies of thin sample of zirconia.

Despite the lack of long-term clinical evidence, more and more clinicians are choosing to restore their patients with complete-arch implant-supported zirconia restorations.[34] A recent systematic review by Abdulmajeed et al included 9 studies for the survival of complete-arch implant-supported monolithic zirconia fixed dental prostheses (FDPs) and concluded that implant-supported zirconia restorations have high survival in the short term but acknowledged the absence of long-term clinical evidence.[35,36] Even though catastrophic failure of complete-arch zirconia restorations has not been commonly reported, our experiment was undertaken to simplify the implant-supported zirconia restoration design into a single unit for the purpose of isolating and testing the effect of varying thicknesses on the fracture strength of zirconia.

Currently, most long-term clinical data come from Larsson et al,[37] who performed an 8-year follow-up on 9 patients who were restored with complete-arch zirconia cemented onto titanium abutments. Although not a prospective study, after 8 years, none of the restorations showed bulk fracture of framework, with fracture of veneering porcelain in 8 patients.

Others have also reported in case reports regarding their experiences with implant-supported zirconia FDPs, with modified framework designs to minimize porcelain fracture.[19,20] Goldberg et al,[23] Mehra et al,[25] and Venezia et al[26] have maintained the occluding or functioning surfaces in full zirconia layering only the facial surfaces with feldspathic porcelain, reducing the chances of veneering material fracture. These results show a promising future for implant-supported complete-arch zirconia-based FDPs as an acceptable treatment alternative, and therefore, a more systemic approach to isolating the design factor for implant-supported zirconia restorations needs to be investigated with in vitro studies. Although different from in vivo studies, in vitro studies that test framework design factors may build the pathway for consistent framework design in future prospective clinical trials.

Limitations of this study include an in vitro experimental design with specimen morphology deviating from a crown morphology, and the force applied may be different from a clinical scenario in which a smaller area or asymmetric configuration of loading may cause increased stress concentration compared with a vertical static load without aging to a single-unit cylinder of zirconia. There are other confounding factors to consider that are possible effects of a different screw hole morphology on the loading area and thus different crack propagation points leading to a fracture. This study is limited to monolithic zirconia screw-retained implant restorations for the specific brand (Prettau Zirconia, Zirkonzahn); therefore, one should be cautious in relating the results to other zirconia implant crowns with porcelain additions. Future experiments testing varying thicknesses under cyclic loading as well as fracture strength after induced low-temperature degradation are under way to provide further information.

TABLE 2

Summary output for one-way independent analysis of variance (ANOVA)

	One-Way Independent ANOVA – Fracture Strength, N Sum of Squares	*df*	Mean Square	*F*	Significance
Between groups	51875185.864	2	25937592.932	123.387	.000
Within groups	5675738.179	27	210212.525		
Total	57550924.043	29			

TABLE 4

Output for test of homogeneity of variances (Levene)

Test of Homogeneity of Variances – Fracture Strength, N Levene Statistic	df1	df2	Significance
.440	2	27	.648

Conclusion

Considering the average occluding forces in humans, this study suggests that a minimum thickness of 0.5 mm to 1 mm of this particular type of zirconia surrounding the implant abutment may be sufficient to resist fracture of the zirconia prosthesis around this area of stress and minimal volume of material.

Abbreviations

ANOVA: analysis of variance
CAD/CAM: computer-aided design/computer-aided manufacturing
FDP: Fixed dental prosthesis
N: Newton
Y-TZP: Yttria-stabilized tetragonal zirconia polycrystal

Acknowledgments

The authors thank Emil Sahakian and Kazunari Takanashi, RDT, at the Burbank Dental Lab, Los Angeles, Calif, for milling the samples. The study was supported in part by an American Academy of Implant Dentistry Foundation student research grant.

Note

The authors declare that they have no conflicts of interest.

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