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RESEARCH AND EDUCATION

Fracture resistance of monolithic zirconia crowns with different occlusal thicknesses in implant prostheses

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ABSTRACT

Statement of problem. The use of monolithic zirconia crowns in implant prostheses is increasing, especially when the interdental space is insufficient. However, fractures have been reported in clinical practice.

Purpose. The purpose of this study was to determine the minimal thickness of a complete zirconia crown used for an implant prosthesis in the posterior dental region.

Material and methods. Fifty complete zirconia crowns were produced using a computer-aided design/computer-aided manufacturing technique. In each group, 5 crowns of varying thicknesses (0.4, 0.5, 0.6, 0.7, and 0.8 mm) were subjected to cycles of vertical and 10-degree oblique compressive loading at 5 Hz and 300 N in a servohydraulic testing machine. Five finite element models comprising 5 different occlusal thicknesses (0.4, 0.5, 0.6, 0.7, and 0.8 mm) were simulated at 2 loading angles (0 and 10 degrees) and 3 loading forces (300, 500, and 800 N). Data were statistically analyzed, and fracture patterns were observed with a scanning electron microscope.

Results. Cyclic loading tests revealed that the fracture resistance of the specimens was positively associated with prosthesis thickness (P<.01). Low von Mises stress values were obtained for prostheses with a minimal thickness of 0.7 mm under varying loading directions and forces.

Conclusions. Zirconia prostheses with a minimal thickness of 0.7 mm had a high fracture resistance and the lowest stress values. Therefore, dentists and laboratory technicians should carefully choose the optimum thickness of zirconia prostheses. (J Prosthet Dent 2016;115:76-83)

Zirconia was first used in the medical field to produce artificial femoral heads.[1] Since the early 1990s, however, applications have expanded, particularly into the implant field of dentistry. Zirconia has high mechanical toughness, superior esthetic properties, and desirable biocompatibility.[2] Zirconia implants have exhibited a bone-implant-contact comparable with that of titanium implants[3-5] and enhanced bone-implant-contact in animal studies.[6] A popular clinical application that affects the light reflection of the supporting soft tissue is the zirconia abutment composition,[7] such as the computer-aided design/computer-aided manufacturing (CAD/CAM) complete zirconia abutment, zirconia friction-fit to a titanium abutment, and zirconia bonded to titanium.[8] Recently, complete zirconia monolithic crowns and fixed dental prostheses have gained attention[9-11] because of their good mechanical properties, low wear of the enamel antagonist,[12] and lack of metal color. They also reduce the major clinical complications that result from the fracturing of veneering porcelain and zirconia frameworks,[13] particularly in an implant prosthesis.

Although zirconia has a relatively high elastic modulus (215 GPa) and high flexure strength (1000 MPa) that exceed those of many metal alloys, clinicians and technicians suggest that the minimal thickness for natural dentition should be 0.5 mm.[14] Comparisons of different ceramic specimens by Lawn et al[15] and Deng

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Clinical Implications

The thickness of a zirconia crown affects the survival rate of an implant prosthesis. A thickness exceeding 0.7 mm is recommended to account for clinical deviation during operations and to preserve space for occlusal adjustment.

et al[16] have shown that fully dense yttria tetragonal zirconia polycrystal (Y-TZP) has the highest critical load strength at low thicknesses (0.1 to 1.0 mm). Lawn et al[17] used different ceramic flat-layer specimens loaded with spherical indenters on their top surfaces and reported that radial cracking, which is governed by strength, is highly dependent on the thickness of the ceramic layer. The fracture strength also revealed a quadratic relationship with the thickness of the ceramic layer. Kelly[18] reported that tensile stresses are highly sensitive to the ratio of elastic moduli between the ceramic and the cement and dentin and, to a much lesser extent, the thickness of the ceramic and the cement. For a more relevant approach, Kelly[18] also suggested using a stiffer substrate and a finite element method (FEM) to solve for stresses as a function of load. Hsueh et al[19] used FEM to determine that the location of maximum tensile stress changes with the thickness ratio between the veneering and framework materials. Hamburger et al[20] showed that the fracture risk of ceramic materials is positively associated with the layer thickness, and Silva et al[21] pointed out that increasing the thickness in the proximal and lingual cervical margin of zirconia-based prostheses should prevent further veneer cracking. The benefit of monolithic zirconia crowns is decreased clinical implant prosthesis failure, especially in patients with insufficient interarch distance.

Occlusion also affects the survival rate of implant prosthesis. In the maximal intercuspal position, obtaining light contact when an occlusion record is heavy and no contact when an occlusion record is light is considered a reasonable clinical approach.[22] For implant occlusion, researchers suggest avoiding excessive premature contacts, large occlusal tables, and steep cusp inclinations because of the lack of periodontal ligament.[23] However, occlusion is always complex and difficult to simulate because it may be affected by masticatory cycles and force. A cyclic contact fatigue test could be an effective method of simulating the impact of occlusion. Additionally, FEM may provide an auxiliary method of investigating the changing nature of stress fields as thickness increases and correlating locations of stress with observed cracking modes. Anatomically, axial loading in the posterior region can vary from 42 to 412 N with the implant prosthesis, and the occlusal contact can vary from 9 to 17 minutes per day.[24] A previous study indicated that the maximal voluntary occlusal force was 545.7 N in men and 383.6 N in women.[25] Specimens subjected to 1 to 1.2 million chewing cycles[26,27] in the computer-controlled chewing simulator under 300 N[28] could simulate 5 years of clinical functional loading over posterior teeth restoration.

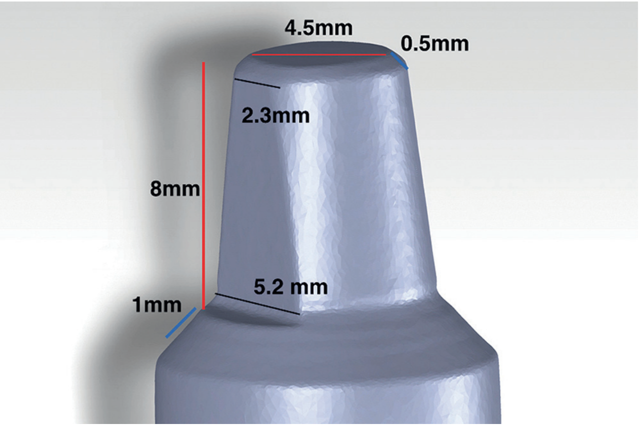

Figure 1. Design and size of dental implant abutment featuring antirotation surface.

The purpose of this study was to determine the minimal thickness of a complete zirconia crown for an implant prosthesis in the posterior region. The hypothesis was that a complete zirconia crown in an implant prosthesis should have a minimum thickness of 0.5 mm.

MATERIAL AND METHODS

An implant prosthesis in the molar area was prepared to receive a complete zirconia crown. The G power analysis was used to estimate the required sample size; assuming 10 test groups, an effect size of .76, the probability of Type I error of .05, and the power of .90. Sample size was thus determined to be 5 per group. Fifty complete zirconia specimens comprising different occlusal thicknesses (0.4, 0.5, 0.6, 0.7, and 0.8 mm) were produced using a Cercon base (DeguDent) and the CAD/CAM technique and then densely sintered at 1350°C (Cercon heat; DeguDent). The specimens were attached to the implant die without a cement space but with adequate friction retention. Figure 1 shows that the stability of the prosthesis was enhanced by the antirotation surface design of the dental implant abutment. Five specimens in each group were tested vertically, and another 5 were tested obliquely (10 degrees) at 5 Hz and at 300 N in a servohydraulic testing machine (MTS 810; MTS System Corp). A vertical load was applied to the occlusal point, and an oblique 10-degree load was applied to the marginal ridge of the specimens. If the specimen exhibited no obvious fracture patterns, the test machine automatically stopped after 1 million impacts under vertical loading and after 300 000

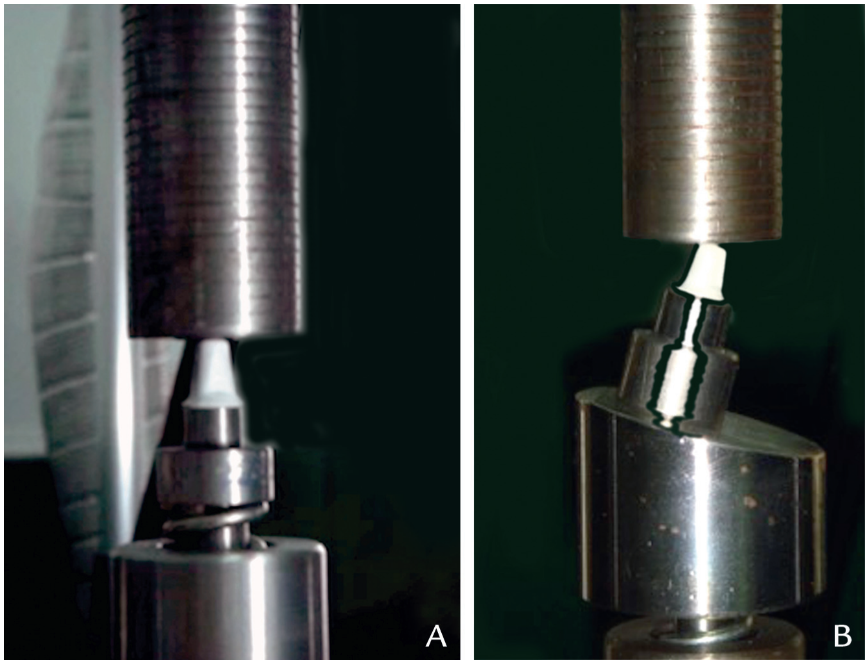

Figure 2. Specimen holder with spring to simulate buffering of human tissue. A, Vertical load test specimen. B, Oblique 10-degree load test specimen.

impacts under oblique loading. The specimen holder and the operating program of the test machine were adjusted to simulate the human masticatory cycle. The specimen holder had 3 parts (Fig. 2): a heat-treated steel rod with a tapered tip to accommodate the specimen, a spring for simulating cyclic occlusal contact and position control, and a precise adapter for the test machine.

The computer program, which controls the test machine, had a 2-stage control to ensure that the force exerted on the specimen was consistent. The first stage, which simulated the natural occlusal displacement, was position control. This stage ensured that the stroke distance between a load cell and a specimen remained at a specific value. The second stage was load control. This stage restricted the force exerted by the load cell to ensure that the specimen sustained a consistent force during the test cycle. The test was stopped when the specimen broke or when the number of impacts reached 1 million.

The dental implant abutment models were constructed using a CAD program (Pro/ENGINEER Wildfire 2.0; Parametric Technology Corp). Five finite element (FE) models with 5 occlusal thicknesses (0.4, 0.5, 0.6, 0.7, and 0.8 mm) were constructed in the posterior mandibular area. All models were combined through Boolean operations by CAD software (Pro/ENGINEER Wildfire 2.0; Parametric Technology Corp). The Young modulus of zirconia was taken as 220 GPa and the Poisson ratio as 0.3.[29] The Young modulus of Ti-6Al-4V was taken as 110 GPa and the Poisson ratio as 0.33.[30]

The materials used in the models were assumed to be isotropic, homogeneous, and linearly elastic. A

Table 1. Results of different occlusal thicknesses of zirconia specimens after cycling test under different loading direction

	Loading Direction		
Thickness (mm)	Vertical	Oblique 10 Degrees	*P**
0.4	2564 ±975[a]	2933 ±2129[I]	NS[†]
0.5	8480 ±2009[b]	31 151 ±11 936[II]	<.01
0.6	17 012 ±5910[c]	300 000[III]	<.01
0.7	1 000 000[d]	300 000[III]	<.01
0.8	1 000 000[d]	300 000[III]	<.01
P[‡]	<.01	<.01	

Different superscript letters in a column indicate statistical significance among groups ($P<.05$; post hoc Tukey test); 1 000 000 cycles stands for total number of cycles that specimens were not broken under vertical loading set; 300 000 cycles stands for total number of cycles that specimens were not broken under oblique loading set.
*Mann-Whitney *U* test (2 independent samples).
†No statistically significant difference ($P>.05$).
‡Kruskal-Wallis test (K independent samples).

3-dimensional FE mesh of a 0.4 mm thickness model comprising 167 772 elements and 236 264 nodes was constructed using 10-node tetrahedral elements. The applied vertical forces were 300, 500, and 800 N, and the applied oblique force was 300 N at 10 degrees from the marginal ridge of the prosthesis. The lower border of the implant-prosthesis complex was constrained as the boundary condition. On the basis of the literature,[31,32] von Mises stress values (EQVs) were defined as the ductile material such as metallic implants and principal stress to distinguish between tensile and compressive stress. The EQVs were detected in diffeent loading directions.

Data were compared by the Kruskal-Wallis test, and the Spearman correlation was revealed using a statistical program (SPSS Statistics for Windows, v19; IBM Corp). The microscopic conditions of the specimens

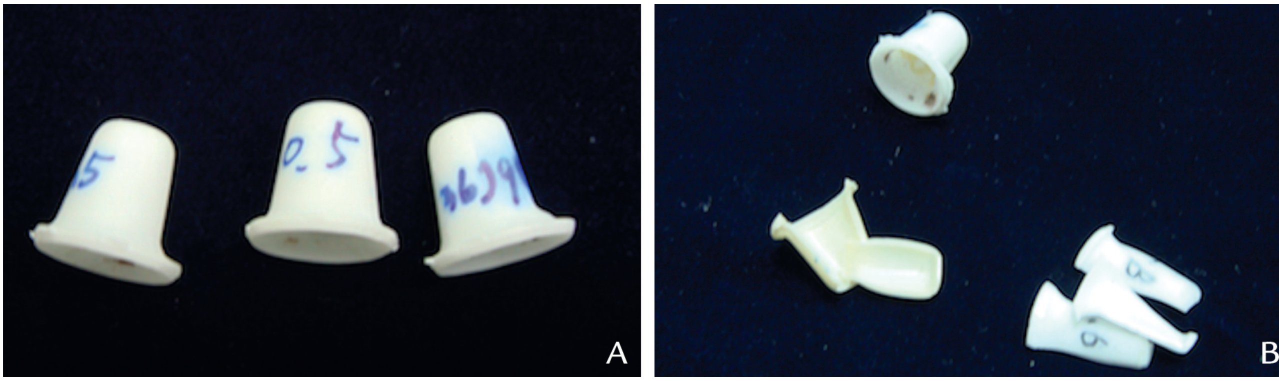

Figure 3. Specimens with even thicknesses. A, 0.5 mm. B, 0.6 mm and intact morphology (top) and fractured segments (bottom).

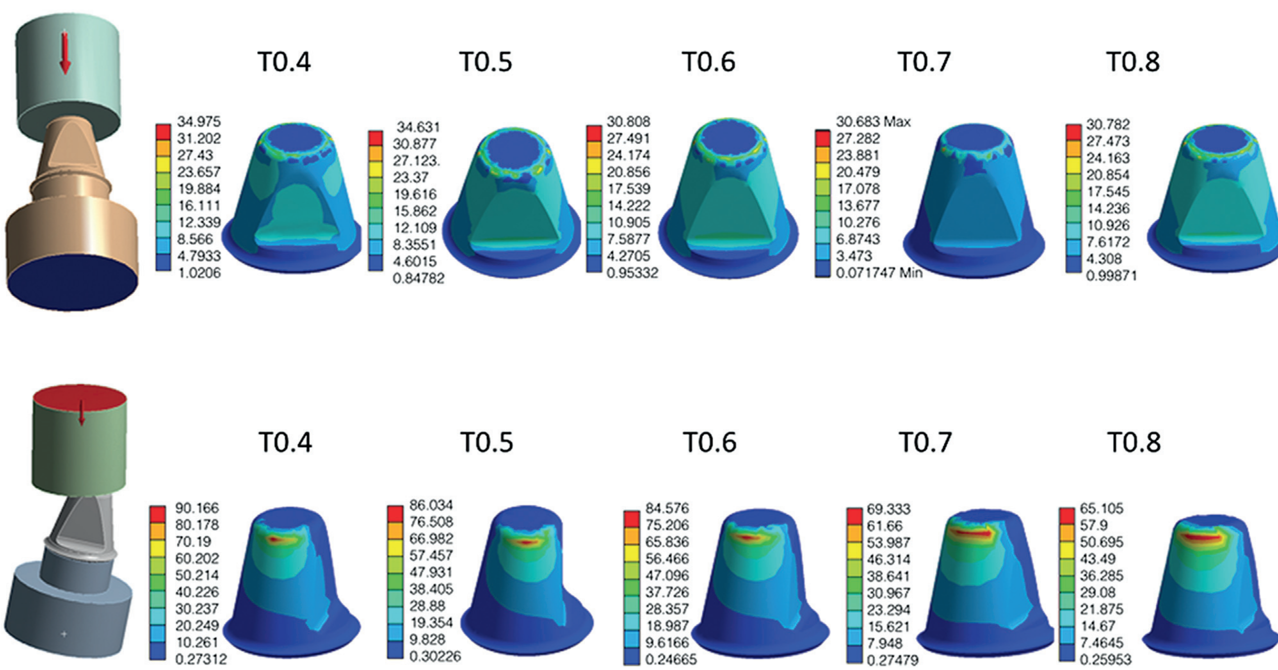

Figure 4. Distributions of stresses in zirconia crowns. (Top) Vertical load of 300 N. (Bottom) Oblique 10-degree load of 300 N on marginal ridge. From left to right are loading models: T 0.4, T 0.5, T 0.6, T 0.7, and T 0.8. T, thickness (mm).

were observed by scanning electron microscope imaging (JSM-6360; JEOL).

RESULTS

Under vertical loading, the mean number of cycles until breakage was 2564, 8480, and 17 012 for 0.4, 0.5, and 0.6 mm specimens, respectively (Table 1). The 0.4 and 0.5 mm specimens showed obvious breakages that separated specimens into 2 or 3 fragments. Specimens thicker than 0.7 mm showed no visible fracture lines or fragments; the test machine was stopped after 1 million impacts. Under oblique loading, the mean number of cycles upon 0.4 and 0.5 mm thick specimens was 2933 and 31 151, respectively. Specimens thicker than 0.6 mm showed no visible fracture lines or fragments; the test machine was stopped at 300 000 impacts (Fig. 3).

The specimen thickness and the number of cycles had a significant positive association (r=.89). Specimen breakage was significantly associated with prosthesis thickness (P<.01) (Table 1). Loading directions significantly differed (P<.01) for all tested thicknesses from 0.5 to 0.8 mm, except for 0.4 mm.

Stress concentrated on the corner area of the occlusal-axial wall of the prosthesis when forces were applied in different directions. The peak EQV was higher for oblique loading than for vertical loading (Fig. 4). Comparisons of different thicknesses under different vertical loading showed that a thickness exceeding 0.6 mm had a low EQV, whereas a thickness of 0.7 mm had the lowest EQV

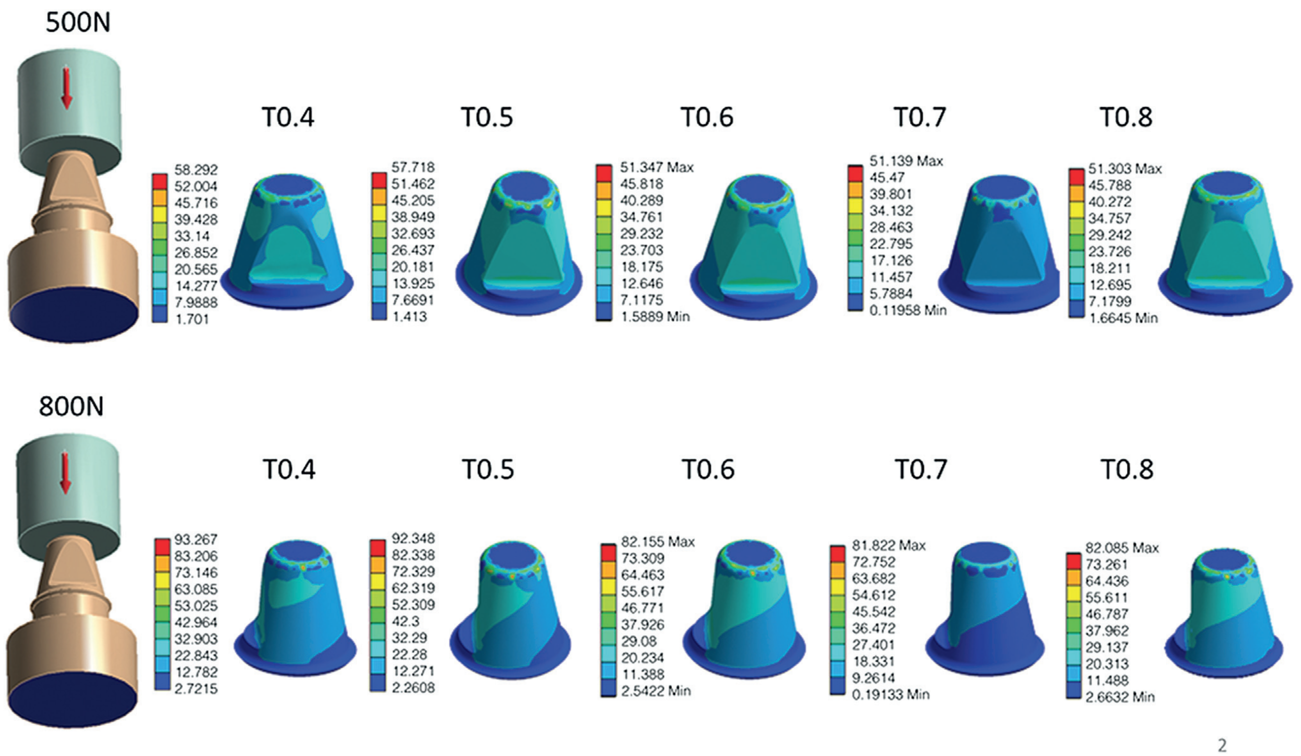

Figure 5. Distributions of stresses in zirconia crowns. (Top) Vertical load of 500 N. (Bottom) Vertical load of 800 N on marginal ridge. From left to right are loading models: T 0.4, T 0.5, T 0.6, T 0.7, and T 0.8. T, thickness (mm).

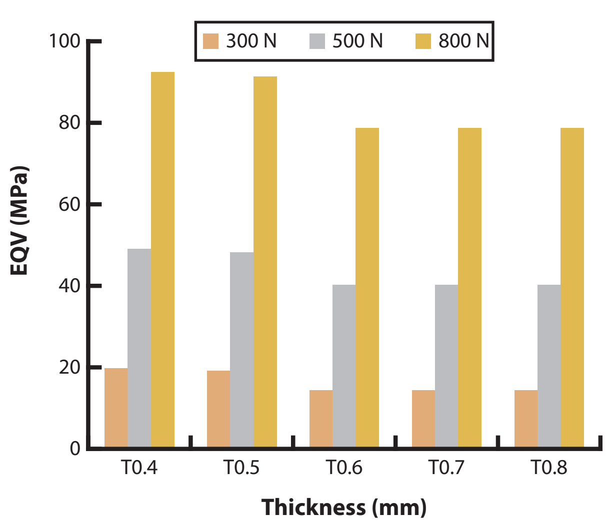

Figure 6. Comparison of EQVs in zirconia prosthesis with 5 thicknesses under vertical loading of 300, 500, and 800 N. Lowest value occurred when thickness was 0.7 mm.

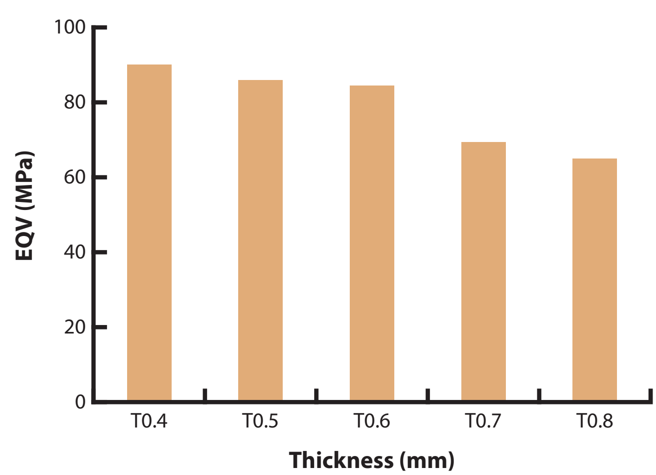

Figure 7. Comparison of EQVs in zirconia prosthesis with 5 thicknesses under oblique loading of 300 N. Lowest value occurred when thickness was 0.8 mm.

(Figs. 4-6). Additionally, comparisons of different thicknesses under oblique loading showed that a thickness exceeding 0.7 mm had a low EQV, whereas a thickness of 0.8 mm had the lowest EQV (Fig. 7).

The scanning electron microscopic images showed beach marks indicating progressive fatigue failure. Alongside these beach marks, the final fracture zone was typically visible where a brittle failure of the material occurred (Fig. 8).

DISCUSSION

Analyses of zirconia specimens show that their thickness affects the fracture resistance.[15] The present study demonstrated that the strength of a zirconia specimen increased with thickness; similar findings have been

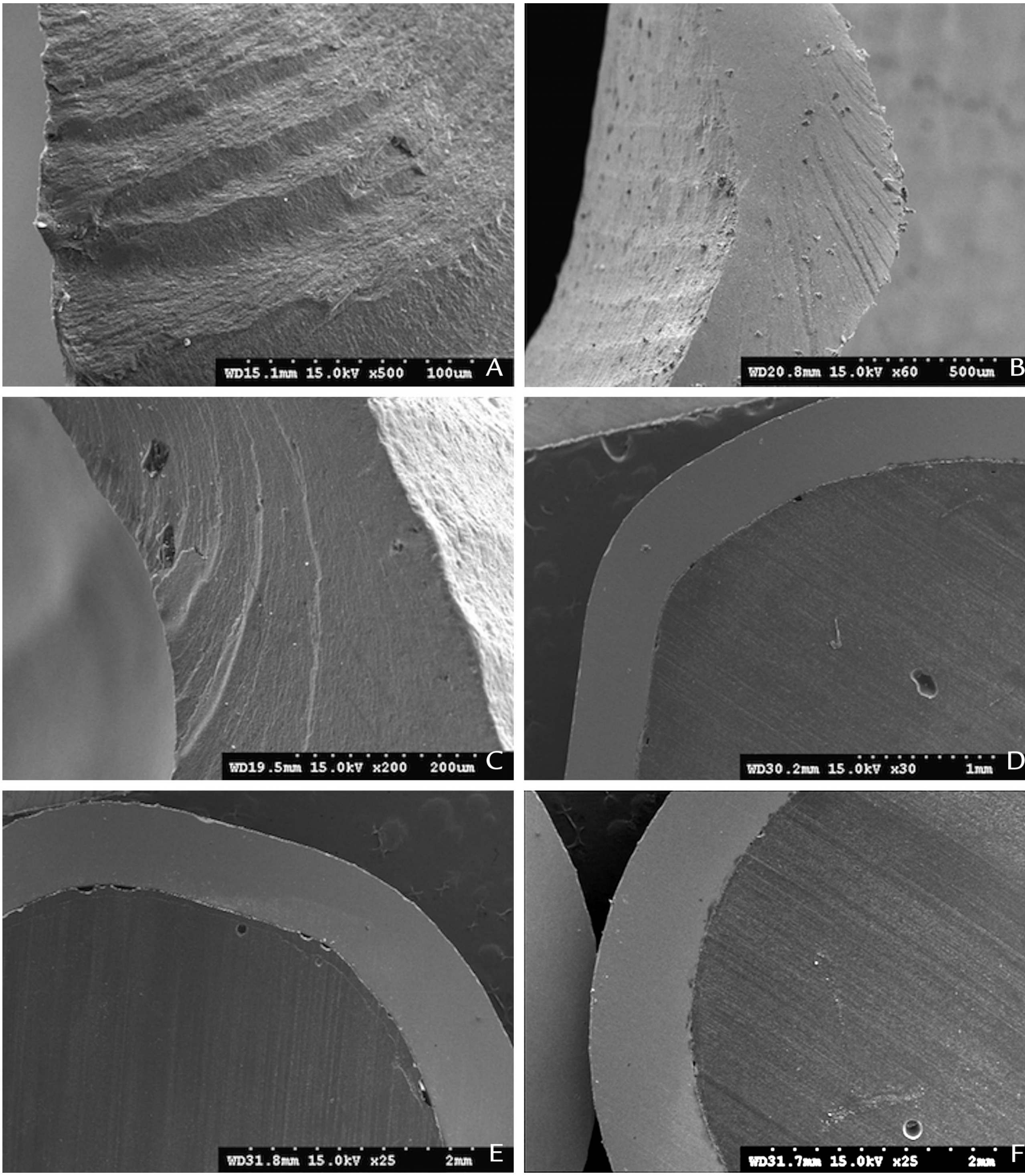

Figure 8. Scanning electron microscopic images showing beach marks from fracture segments. A, 0.4 mm (original magnification, ×500). B, 0.5 mm (original magnification, ×60). C, 0.6 mm (original magnification, ×200). No fracture line or other cracks were found in D, 0.6 mm (original magnification, ×30). E, 0.7 mm (original magnification, ×25). F, 0.8 mm (original magnification, ×25) from specimen cross sections.

reported by Lawn et al[15] and Deng et al.[16] Clinicians and technicians often use the recommended zirconia thickness of 0.5 mm. This study showed that the 0.5 mm thickness of an occlusal surface is insufficient for implant occlusion in the posterior region and that even a thickness of 0.6 mm has a high risk of fracture under vertical loading. A zirconia implant prosthesis should be at least 0.7 mm.

Because the critical load is calculated as the thickness squared, Lawn et al[17] suggested that radial cracks can be avoided by using a minimum thickness of 1.5 mm. However, Kelly[18] recommended that approaches to more relevant testing of ceramic restorations should use a substrate stiffer than epoxy resin in indentation contact tests. Other methods of solving for stresses as a function of load include FEM or iterative mathematical approximation. In a study by Deng et al,[16] comparisons of different ceramic plates revealed that monolithic zirconia specimens had relatively high critical contact loads (800 to 900 N) when the thickness exceeded 0.7 mm. However, the substrate was polycarbonate. In the more relevant method applied in the present study, a heat-treated steel rod was used to simulate the dental implant abutment of the cyclic contact test and FEM when the stress distribution was analyzed.

The dynamic impact used to fracture the zirconia prosthesis was set to 300 N to simulate implant occlusion in the mandibular molar area. However, actual clinical conditions are complex, especially in the case of implant occlusion. The recommended occlusal design includes centered contacts along the implant axis with a flat central fossa.[22] In occlusal schemes for single implant prosthesis, anterior or lateral guidance is suggested for natural dentition, light contact is suggested when an occlusal force is heavy, and no contact is suggested when an occlusal force is light. In most conditions, the implant occlusion is not fully rigid, and the buffer would be provided by the periodontal ligament of the dental antagonist and surrounding alveolar bone. When the dental antagonist is an implant prosthesis in the posterior area, the buffering effect is lessened and the relative rigid impact from occlusion is noticeable. This study used a spring to simulate cyclic occlusal contact and position control to decrease the entirely rigid impact from occlusion. Most implant zirconia prosthesis designs should have a minimum thickness of 0.7 mm. However, when the occlusal impact is more rigid with less buffering, an implant thickness exceeding 0.7 mm is suggested.

The servohydraulic testing and FE results revealed similar trends under vertical loading and oblique loading. Fracture resistance was higher at thicknesses exceeding 0.7 mm than at a thickness of 0.5 mm. However, using a thickness of 0.6 mm is controversial because different loading types have different effects. For example, MTS loading on a specimen with point contact differs from FE loading on a specimen with surface contact. Additionally, because of the limitations of the fixture design, the oblique loading force of MTS is less than 300 N, a phenomenon that is attributable to the component of force (Fig. 2). However, the EQV from 300 N oblique loading was directly applied to the corners of the specimens. The results of the 2 different methods indicate that a thickness of 0.7 mm is recommended for posterior implant zirconia prosthesis designs.

CONCLUSIONS

Considering the limitations of the study, the following were concluded:

1. Cyclic loading tests showed that the fracture resistance of the monolithic zirconia prosthesis was positively associated with thickness.
2. Cyclic loading in a servohydraulic testing machine and persistent loading in FE analysis revealed that a zirconia prosthesis with a thickness of 0.7 mm or greater had the highest fracture resistance and the lowest stress values under vertical and oblique loading.
3. For clinical use, a zirconia prosthesis with a thickness of 0.8 mm is recommended to allow for operative deviation and error in occlusal adjustment.

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