A Load-to-Fracture and Strain Analysis of Monolithic Zirconia Cantilevered Frameworks

THESIS

Presented in Partial Fulfillment of the Requirements for the Degree Master of Science in
the Graduate School of The Ohio State University

By

Faris Alshahrani, BDS, RCD(C)

Graduate Program in Dentistry

The Ohio State University

2016

Thesis Committee:

Advisor: Burak Yilmaz, DDS, PhD

Member: Jeremy Seidt, MS, PhD

Member: Edwin McGlumphy, DDS, MS

Member: William Brantley, BS, MS, PhD

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Abstract

Fixed implant-retained cantilevered restorations for edentulous patients have been successfully used in implant dentistry. In recent years, with the improvements in CAD/CAM technology, zirconia has been in use as a more affordable alternative to metal frameworks for fixed restorations due to the rising cost of noble metals. The dimensions of implant-retained fixed cantilevered dentures are important to prevent mechanical and biological complications. Information on the optimum thickness and cantilever length for improved strength of zirconia frameworks is limited in the literature.

The aim of this study was to investigate the effect of cantilever length and occluso-cervical thickness on the load-to-fracture and strain distribution of zirconia frameworks.

Twenty-seven rectangular prism-shaped specimens (6 mm thick buccolingually) were fabricated using a CAD/CAM milling technique. Specimens were prepared in 9 groups (n=3) according to their vertical dimensions (6×6 mm, 8×6 mm and 10×6 mm) and cantilever loading distance (7 mm, 10 mm and 17 mm). All specimens were heat-treated in a porcelain furnace and thermocycled for 20000 cycles before the tests. Each framework was secured using a clamp attached to the first 20 mm of the framework. Three-dimensional image correlation (3D-DIC) technique was used for a full-field measurement of strain during testing. A load-to-fracture test was used until the specimens

fractured. Maximum force and the principal strain data were analyzed by ANOVA using the maximum likelihood estimation method.

The frameworks tested in this study fractured at load values ranging from 1065 - 8870 N. The effect of occluso-cervical thickness and cantilever length was significant for the load-to-fracture ($P < .001$). No statistically significant interaction was observed between the two factors. The strain values for all specimens were similar. No statistically significant effects were found for occluso-cervical thickness and cantilever length or between them on the strain distribution.

Increased occluso-cervical thickness allowed the cantilever to withstand higher loads. Decreased cantilever length allowed the frameworks to withstand higher loads. The occluso-cervical thicknesses and cantilever lengths of zirconia frameworks tested withstood the maximum bite force that has been reported in the literature. However, the properties of components in the dental implant-abutment-framework assembly should be taken into consideration when interpreting these results.

Acknowledgments

I would like to thank my advisor Dr. Burak Yilmaz for his guidance and support and believing in this project. I would also like to thank my committee members, Dr. Edwin McGlumphy, Dr. Jeremy Seidt and Dr. William Brantley, for being great mentors, for their help with the study and their valuable input throughout the process. A special thank you to Dr. Johnston for his timely help with the statistics.

Vita

- 2008..... Bachelor in Dental medicine & Surgery
BDS, Faculty of Dentistry, King Abdulaziz
University, Jeddah, Saudi Arabia.
- 2014..... Certificate in advanced prosthodontics.
Ostrow School of Dentistry, University of
Southern California, Los Angeles, USA.
- 2016..... Fellow of The Royal College of Dentists of
Canada, Prosthodontics

Fields of Study

Major Field: Dentistry

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Chapter 1: Introduction

The use of zirconia in dentistry has increased significantly in the last decade. One of the uses of zirconia in dentistry is in implant-supported prosthesis.¹ In recent years zirconia has been used as an alternative material to metal frameworks due to the rising cost of noble metals, which make the zirconia framework fabrication more affordable.² The increased interest in zirconia as a framework material can be attributed to the improvements in computer-aided design/ computer-aided manufacturing (CAD/CAM) technology, its excellent intrinsic biocompatibility,^{3,4} good mechanical properties,^{3,5} and improved esthetics due to the absence of underlying gray color of metal framework.⁶

The fracture toughness of zirconia ranges from 6 to 10 MPa m^{1/2}. In addition to high fracture toughness values zirconia has shown good flexural strength, ranging from 900 to 1200 MPa.^{7,8,9,10} Three crystallographic shapes of zirconia exist at different temperatures: cubic, tetragonal, and monoclinic. The monoclinic phase is stable at room temperature and up to 1170° C, tetragonal phase between temperatures 1170 - 2370° C and cubic phase over 2370° C.¹¹ The most desirable phase in dentistry is the tetragonal phase. Zirconia can be stabilized in tetragonal phase also at room temperature by adding stabilizing oxides such as CaO, MgO, CeO₂ or Y₂O₃.⁶ The three most commonly used types of zirconia are:

1) Y-TZP – Yttria Tetragonal Zirconia Polycrystals

2) GTA – Glass Infiltrated Zirconia-toughened Alumina

3) PSZ– Partially Stabilized Zirconia

Y_2O_3 is the most commonly used stabilizer of dental zirconia material, and yttria partially stabilized tetragonal zirconia is the most commonly used form of zirconia in dentistry.

The superior mechanical properties of zirconia are a result of transformation toughening. Upon cooling, zirconia undergoes a transformation from the tetragonal to the monoclinic phase. The phase transformation is accompanied by a 4% to 5%-volume increase of the crystals. This increase in volume creates high compressive stresses within the material that prevent crack propagation.^{12,13} The local stresses created by a crack can lead to tetragonal to monoclinic transformation at the crack tip, and this will prevent crack propagation (Figure 1)¹⁴. Moreover surface grinding, sandblasting, stress or temperature changes can initiate the tetragonal to the monoclinic transformation.^{15,16}

Figure 1. Illustration of the stress-induced transformation toughening process

Another phenomenon related to the transformation ability of the tetragonal phase is low temperature degradation. In the presence of water molecules, or body fluids such as saliva, slow surface transformation to the stable monoclinic phase occurs^{17,18}. In a moist environment, water molecules can penetrate the zirconia and lead to transformation of the outer tetragonal grains of zirconia to monoclinic grains¹⁹. Transformation of the outer tetragonal grains to monoclinic grains is accompanied with increase in volume and creates stress on surrounding grains. Increasing the volume of surface and the grains allow more water molecules to penetrate and more transformation of the grains in deep layers of the material.^{20,21} (Figure 2)²⁹

The low temperature degradation is affected by many factors. Increasing the grain size affects the low temperature degradation resistance. A decrease of degradation resistance for grain sizes above 0.6 μm has been reported.²² In another study, grain size bigger than 1 μm shows large amount of *t-m* transformation as well as decrease in strength.²³

The sintering temperature controls the amount of the cubic phase in zirconia. When the sintering temperature reaches 1500° C during a long sintering time (5h), the amount of the cubic phase in zirconia increases. In the presence of large amounts of cubic phase, the resistance to low temperature degradation is reduced.²⁴

The density of zirconia plays a role in the low temperature degradation process. The lesser the density of zirconia, the greater the amount of degradation. The high density minimizes water molecule access to the bulk of the material and as well results in the increase of mechanical properties.²⁰ Surface stress also has a role in low temperature

degradation. Because the stresses decrease the energy needed for phase transformation, the tensile surface stresses caused by machining can initiate the *t-m* transformation.^{25,26} On the other hand, compressive stresses do not have a major effect on phase transformation.^{27,28}

Figure 2. Illustration of low temperature degradation in zirconia

Using zirconia in moist environments seems to be more of a concern for hip prostheses than dental restorations. Failures of many hip prostheses were reported and it seems that

autoclave sterilization had caused the degradation.²⁹ For dental prostheses, the studies evidenced that low temperature degradation does not have major influence on the clinical behavior or mechanical properties of zirconia.³⁰

The use of cantilevered prosthesis frameworks can be beneficial for sites where anatomic structures such as the maxillary sinuses and inferior alveolar nerves prevent the ideal placement of dental implants in the posterior regions. The additional occluding units that may be provided by the cantilever portion can provide function and esthetics that would otherwise not be achievable.³¹ However, cantilevers in implant dentistry may adversely influence the biomechanics of implant restorations^{32,33} and may result in mechanical and/or biological complications.

The evidence available regarding load-to-fracture of cantilevered zirconia frameworks is mainly limited to opinion and expert clinical reports, and there is no clear consensus as to what can be considered as an adequate occluso-cervical thicknesses and cantilever length.³⁴

The zirconia fixed partial dentures without cantilevers have been extensively studied and the load-to-fracture value has varied between 1192 N and 2131 N.^{35,36,37} The strain distribution on the loaded zirconia fixed partial dentures without cantilevers was studied using finite element analysis, and it has been shown that the connector region is the weak point for these restorations.^{38,39}

Well-designed, clinically relevant laboratory studies are needed to investigate the effect of occluso-cervical thickness and cantilever length on the load-to-fracture value of zirconia frameworks. The aim of this study was to investigate how the load-to-fracture

and strain distribution of zirconia frameworks were influenced by varying cantilever lengths and occluso-cervical thicknesses.

The null hypotheses tested were that the specimens with different cantilever lengths and occluso-cervical thicknesses would have similar load-to-fracture values.

Chapter 2: Materials and Methods

Twenty-seven frameworks were fabricated using a CAD/CAM milling technique. All specimens were 6 mm in bucco-lingual width and prepared in 9 different groups (n=3) according to their vertical thicknesses (6×6 mm, 8×6 mm and 10×6 mm) and cantilever loading distances (7 mm, 10 mm and 17 mm) (Figures 3 and 4). The frameworks were milled from zirconia blocks using a CAM machine (Zirkonzahn, Italy). All specimens were heat-treated in a porcelain furnace to simulate the firing cycles of the veneering porcelain. The firing program was implemented according to the manufacturer recommendations. The specimens underwent thermocycling with the following program: 20,000 cycles in two water baths (5° C and 55° C). Each cycle was 60 seconds: 20 seconds in each bath and two times (10 seconds each) to complete the transfer between the baths. Each framework was secured by using a clamp attached to the first 20 mm of each framework (Figure 5). Before loading in the testing machine, the load frame was placed to contact the framework 3 mm away from the free end (Figure 6). The length of the frameworks represented the number of teeth used as the cantilever (7 mm = premolar, 10 mm = molar and 17 mm = premolar and molar). The crosshead speed was set at 1 mm/min.

Static load was applied in a vertical direction to the zirconia framework using a biaxial servohydraulic load frame and universal testing machine (Instron Model 1321;

Instron, Norwood, MA, USA) (Figure 7). The maximum load-to-fracture values were recorded as indicated by a sudden load drop. All specimens were loaded from 0 Newton (N) until fracture occurred. The broken fragments were collected. The ultimate failure load level at maximum load capacity was recorded in Newtons (N).

The maximum bending stress was also (σ) determined by the general relationship:

$$\sigma = \frac{My}{I}$$

Since $M=FL$,

$$\sigma = \frac{FLy}{I}$$

F is the load-to-fracture (N)

L is the cantilever length (m)

y is the distance from neutral axis (m)

I is the area moment of inertia (m^4)

The maximum force data were analyzed by an analysis of variance (MIXED procedure, SAS ® Proprietary Software 9.3, SAS Institute, Inc., Cary, NC, USA) using the maximum likelihood estimation method⁴⁰ to eliminate the needs for normality within the groups and for equality of variances between the groups. For this analysis, two factors were the occluso-cervical thickness and cantilever length, and the statistical model included the interaction between these two factors. Resolution of the significance factors from this analysis was obtained using the Tukey test, with an overall $\alpha=0.05$.

Figure 3. Different dimensions of the frameworks were used (10×6, 8×6 and 6×6) mm

Figure 4. First 20 mm of the each framework for clamp attachment, and 7, 10 and 17 mm
as cantilever

Figure 5. Framework with fixture attached to testing machine

Figure 6. Schematic of frameworks showing the indenter location

To measure the strain, the three-dimensional image correlation (3D-DIC) technique was applied to all specimens, and calculations were performed using an image correlation software (Vic-3D, version 7.2.4).⁴¹ A pair of high-resolution digital cameras (Point Grey Research, Richmond, British Columbia, Canada) was used in conjunction with digital image correlation (Vic-3D, version 7.2.4, build 194, Columbia, SC, USA) to provide a synchronized stereo view of the models during the experiment (Figure 8). A random dot

pattern was applied to external surfaces of the specimens. As the specimens were tested, the cameras recorded changes in the random dot pattern. Before tests, each camera was independently used to record several images of the same target grid in different views to provide the basis for relating image positions in both cameras to a common three-dimensional coordinate system.

The strains were calculated throughout the entire surface of the specimens until fracture, by using the dot pattern to define correlation areas for each recorded photograph.

The principal strain data were analyzed by a repeated-measures analysis of variance (MIXED procedure, SAS ® Proprietary Software 9.3, SAS Institute, Inc., Cary, NC, USA) using the same maximum likelihood estimation method to eliminate the needs for normality within the groups and for equality of variances between the groups. For this analysis, the two between-subject factors were the occluso-cervical thickness and cantilever length, and the two within-subject factors were the location and the direction of the principal strains. All interactions were included in this statistical model.

Figure 7. Instron model 1321; Instron, Norwood, MA, USA

Figure 8. High-resolution digital cameras

Chapter 3: Results

The frameworks tested in this study fractured at loads ranging from 1065N to 8870 N.

(Figure 9)

The mean load-to-fracture value for the 7 mm cantilever length was 5433 N.

The mean load-to-fracture value for the 10 mm cantilever length was 4406 N.

The mean load-to-fracture value for the 17 mm cantilever length was 2860 N.

The mean load-to-fracture value for the 6 mm occluso-cervical thickness was 2427 N.

The mean load-to-fracture value for the 8 mm occluso-cervical thickness was 4196 N.

The mean load-to-fracture value for the 10 mm occluso-cervical thickness was 6075 N.

Effect	L	H	Estimate	Standard Error	DF	t Value	Pr > t
L	7		5433.00	303.44	18	17.90	<.0001
L	10		4406.38	303.44	18	14.52	<.0001
L	17		2860.18	303.44	18	9.43	<.0001
H		6	2427.79	303.44	18	8.00	<.0001
H		8	4196.06	303.44	18	13.83	<.0001
H		10	6075.70	303.44	18	20.02	<.0001

Table 1. Least Square Means — Load to Fracture

The effect of cantilever length was statistically significant for the load-to-fracture ($P < .001$).

The effect of occluso-cervical thickness was statistically significant for the mean load-to-fracture values ($P < .001$).

No statistically significant interaction was observed between occluso-cervical thickness and cantilever length for the load-to-fracture values.

Effect	Num DF	Den DF	F Value	Pr > F
L	2	18	18.22	<.0001
H	2	18	36.14	<.0001
L*H	4	18	0.82	0.5312

Table 2. ANOVA Summary

Figure 9. Mean load-to-fracture values for all groups

The maximum bending stress value ranged between 342 and 926 MPa. The mean maximum bending stress value for all specimens tested in this study was 727 MPa. The mean maximum bending stress value for each group is summarized in Table 3.

The mean tensile strain value ranged between 0.001 and 0.003, and the mean compressive strain value ranged between 0.001 and 0.002. For all specimens, the mean tensile strain value was 0.003, and the mean compressive strain value was 0.002. The mean tensile and compressive strain values for each group are summarized in Table 4.

Group	Maximum bending stress MPa
6X6 with 7 mm Cantilever	689.13
8X6 with 10 mm Cantilever	690.69
10X6 with 17 mm Cantilever	613.78
6X6 with 7 mm Cantilever	806.15
8X6 with 10 mm Cantilever	736.21
10X6 with 17 mm Cantilever	692.71
6X6 with 7 mm Cantilever	771.98
8X6 with 10 mm Cantilever	781.64
10X6 with 17 mm Cantilever	766.32

Table 3. Mean Maximum Bending Stress for Each Group

No statistically significant effect was observed for occluso-cervical thickness and cantilever length or interaction between them on the strain distribution.

All zirconia frameworks fractured at the area secured by the clamp, with catastrophic crack propagation along occluso-cervical thickness. (Figures 10 and 11)

Group	Tensile strain	Compressive strain
6×6 with 7 mm Cantilever	0.003	0.002
8×6 with 10 mm Cantilever	0.003	0.002
10×6 with 17 mm Cantilever	0.003	0.002
6×6 with 7 mm Cantilever	0.003	0.002
8×6 with 10 mm Cantilever	0.003	0.001
10×6 with 17 mm Cantilever	0.003	0.002
6×6 with 7 mm Cantilever	0.001	0.001
8×6 with 10 mm Cantilever	0.002	0.002
10×6 with 17 mm Cantilever	0.003	0.002

Table 4. Mean Tensile and Compressive Strain for Each Group

Figure 10. Area secured by clamp (Before loading)

Figure 11. Fracture at area secured by clamp (After loading)

Chapter 4: Discussion

The null hypotheses that specimens with different cantilever lengths and occluso-cervical thicknesses have similar load-to-fracture values were rejected.

Results of this study showed that the cantilever length and occluso-cervical thickness are important factors in determining the load-to-fracture values of zirconia cantilever frameworks, and both factors (cantilever length and occluso-cervical thickness) alter the load-to-fracture values independent of each other. Although no statistically significant interaction was found between the cantilever length and occluso-cervical thickness, the clinical outcome may be different. The highest mean load-to-fracture values among the nine groups (7672 N) was found in the group with greatest occluso-cervical thickness (10 mm) and the shortest cantilever length (7mm). In contrast, the lowest mean load-to-fracture values among the nine groups (1543 N) was found in the group with shortest occluso-cervical thickness (6 mm) and the longest cantilever length (17mm).

The lowest mean load-to-fracture of all groups (1543 N) is almost twice the maximum bite force 720 N that has been reported.⁴² However, higher maximum biting forces occur for patients with bruxism. One study examined the bite force in young dentate adults and found that the mean maximum bite force for bruxers was 806 N.⁴³ In another study, the mean maximum bite forces in the molar region were 847 N for men and 597 N for women.⁴⁴ The maximum bite forces for cantilever FPD's were reported between

150 N and 700 N.⁴⁵

In the present study, no statistically significant effect was observed for occluso-cervical thickness and cantilever length or interaction between them on the strain distribution. The measured maximum bending stresses and strains were relatively consistent, independent of the geometry of the cantilever, which can imply: 1) the bending stresses and strains are intrinsic material properties and 2) these properties can be used to evaluate proposed cantilever designs prior to implementation. The three-dimensional image correlation (3D-DIC) technique used in this study showed that tensile and compression stresses were concentrated at the area secured by the clamp, which makes the dimensions of the area where the cantilever starts critical (Figures 12 and 13). A minimum diameter of 4.0 mm is recommended for the connector area of posterior zirconia fixed partial dentures.⁴⁶ The effect of the diameter for the connector area of posterior zirconia fixed partial dentures on the load-to-fracture has been studied, and the general consensus is that increased cross-sectional thickness of connector is desirable.^{47,48} In the observations by other investigators, the axial load on the cantilever produced compressive strain in the lower part of the distal crown and tensile strain at the top part of the crown. The tensile stress was reported to be responsible for the fracture.⁴⁹

Figure 12. Stresses concentrated at the area secured by the clamp (Before loading)

Figure 13. Stresses concentrated at the area secured by the clamp (After loading)

The occluso-cervical thickness of the frameworks used in this study was determined based on different clinical situations for available intra-occlusal space for the restorative materials. The present results showed that increased occluso-cervical thickness allowed the cantilever to withstand higher loads and provided improved fracture resistance. This is in agreement with the results from another study which reported that the load-to-fracture value increase as the connector height increased.⁵⁰ Complete or partial contour zirconia restorations with acceptable aesthetics can be used to achieve appropriate cross sections of the cantilever frameworks, as done in this study.⁵¹

Previous in vitro studies have demonstrated that zirconia has a flexural strength of 900 to 1200 MPa, which is higher than the value found in this study (727 MPa).^{7,8} These previous values were found using a 3-point bending test following ISO standards. However, the design used in this study is not a 3-point bending test design, which could potentially explain the observed different mean value for flexural strength of the tested zirconia specimens. In addition, aging through thermocycling and the applied veneering porcelain firing cycles before the tests to simulate clinical conditions might have influenced the flexural strength value measured in this study. The heat treatment that occurs during firing of the veneering porcelain lowers the strength of zirconia when fabricated in a sintered form.^{52,53} In the present study all frameworks were heat-treated in a porcelain furnace to simulate the firing cycles of the veneering porcelain. The amount of the monoclinic phase on the surface of zirconia specimens can increase by thermocycling.⁵⁴ To simulate the intraoral ageing process, thermocycling for 20,000 cycles was applied on all frameworks.

Complete-arch implant-supported zirconia frameworks provide an alternative treatment option for edentulous patients with potentially less complications compared to hybrid prostheses. Even with the complications of hybrid prostheses that have been reported, including wear and fracture of denture teeth, fracture of the framework, and poor esthetics, they are still considered highly successful prostheses.^{55,56} The most common complication of implant-supported zirconia frameworks is fracture of the veneering porcelain.⁵⁷ A 8-year pilot study observed fractures of the veneering porcelain in eight out of nine implant-supported full-arch zirconia-based fixed dental prostheses, and none

of them showed fractures of the framework.⁵⁸

These kinds of complications can be avoided by fabricating the complete contours of prostheses from zirconia without layering porcelain in the posterior area; this approach may eliminate chipping and provide more bulk for the framework material. A randomized study reported no fractures of the veneering porcelain when implant-supported zirconia prostheses and implant-supported metal–ceramic prostheses were compared.⁵⁹

Recent systematic reviews show that the short-term survival rate for implant-supported zirconia frameworks is excellent.⁵⁷ Zirconia has been used for implant-supported prostheses for a relatively short time, which limits the number of published studies with short follow-up periods.⁶⁰

In the design of this study, the zirconia frameworks were not connected to dental implants to eliminate any factors that may influence the results of the tests such as misfit of the framework and to avoid failure of implant prosthetic components before the frameworks such as the retaining prosthetic or abutment screws. In fact, the retaining screws are designed to fail at 600 N.^{61,62} Not using screw-access channels in the frameworks prevented their affecting the load-to-fracture values of zirconia, thereby enabling the testing of the zirconia and its dimensions only. It has been reported that the fracture resistance of monolithic zirconia was not affected by screw access channels.⁶³

Even though load-to-fracture testing in this study does not completely simulate the masticatory forces, it allows for the interpretation of factors affecting the fracture force and amount of force the zirconia framework can withstand in cantilevered situations.⁶⁴

Further testing should include additional variables such as cyclic loading in a wet environment to show how zirconia will perform. Future study designs should also incorporate the use of implants to observe how zirconia cantilever frameworks perform when other intraoral factors are also involved and affect the interpretation/comparison of results.

Chapter 5: Conclusion

Within the limitations of this study, the following conclusions can be drawn:

- 1) The effect of cantilever length and occluso-cervical thickness were statistically significant for the mean load-to-fracture values ($P < .001$).
- 2) Increased occluso-cervical thickness allowed the cantilever to withstand higher loads.
- 3) Decreased cantilever length allowed the frameworks to withstand higher loads.
- 4) The occluso-cervical thicknesses and cantilever lengths of frameworks tested withstood the maximum bite force that has been reported in the literature.

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