Research Article

JRAD Journal of Research and Advancement in Dentistry

J Res Adv Dent 2016;5:1:67-74.

A Comparative Study of Fracture Resistance of "Zirconia" and NiCr Metal Framework for Fixed Partial Denture

K Rohan Kumar[1*] Molugu Mallikarjun[2] A Kalyan Chakravarthy[3] P Gautham[4] Anil Kumar[5] M Mahesh Babu[6]

[1]PG Studentr, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
[2]Professor and Head, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
[3]Reader, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
[4]Reader, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
[5]Senior Lecturer, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
[6]Senior Lecturer, Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.

ABSTRACT

Aims and objectives: The purpose of this study was to evaluate the load bearing capacity of four unit spanned Zirconia copings, and compare with NiCr metal coping for fixed partial denture frameworks and its feasibility to use it in the posterior region .

Materials and methods: Total 20 samples,10 zirconia and 10 NiCr four unit fixed partial denture framework with a uniform 0.8-mm-thick core, connector dimension 4*4mm and chamfer finish line and pontics with anatomic form were constructed, artificial aging was performed for zirconia samples to mimic the the oral environment conditions. All the samples were tested for load bearing capacity using universal testing machine.

Results: The fracture resistance was recorded, tabulated and analyzed statistically. The results showed NiCr metal frameworks have higher load bearing capacity than zirconia frameworks and were statistically significant (NiCr -2117.60 N, Zirconia- 1251.70 N), and the values of zirconia showed twice the load bearing capacity seen the posterior molar region.

Conclusion: Fixed partial dentures fabricated from zirconia following the proper design parameters and aging protocols are capable of withstanding the mastigatory forces in the posterior molar regions of the oral cavity, although further prolonged aging experiments and prospective clinical trials are required to prove their fitness for clinical use.

Keywords: zirconia bridge, zirconia coping, All ceramic FPD, Flexural strength of zirconia.

INTRODUCTION

The first feldspathic porcelain crown was introduced by Land[1] following which the interest and demand for nonmetallic and biocompatible restorative materials increased for clinicians and patients. In 1965, McLean[2] pioneered the concept of adding Al_2O_3 to feldspathic porcelain to improve mechanical and physical properties. All-ceramic dental materials can be unique in their chemical composition as well as in their structure and therefore demonstrate different material properties. In dentistry there are three different groups of ceramics: polycrystalline ceramics, glass infiltrated ceramics and glass ceramics[3]. Veneer ceramics are feldspathic porcelains which consist almost entirely of an amorphous glass phase and therefore deliver ideal optical characteristics for the veneering. Glass ceramics and glass infiltrated

Received: Apr. 23, 2015: **Accepted:** Aug. 18, 2015
***Correspondence Dr. K Rohan Kumar.**
Department of Prosthodontics, Meghna Institute of Dental Sciences, Nizamabad, India.
Email: kandukurirohan69@gmail.com

pISSN 2278-0076
eISSN 2321-9270

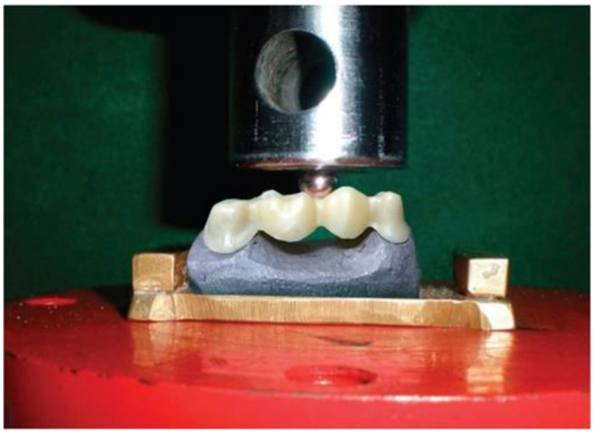


Fig 1: Load bearing capacity of Group I Samples- Zirconia frameworks.

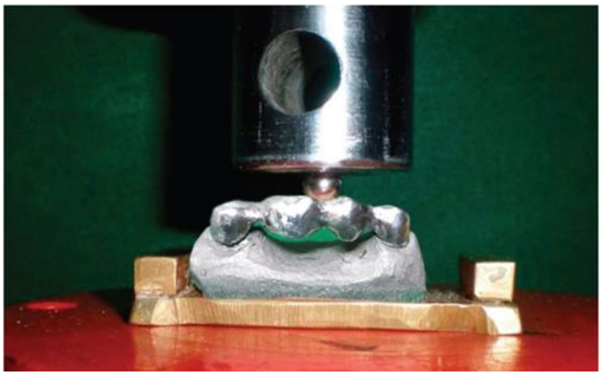


Fig 2: Load Testing Of Group II Samples- NiCr metal framework.

Fig 3: Group I Samples After Load Testing.

Fig 4: Group II Samples After Load Testing.

ceramics are multi-phase materials and contain crystalline constituents (e.g. leucite crystallites in the glass ceramic, Al2O3-crystals in infiltrated ceramics etc.) in addition to an amorphous glass phase. The clinical shortcomings of these materials such as brittleness, crack propagation, low tensile strength, wear resistance, and marginal accuracy, continued to limit their use only to the anterior region. The clinical shortcomings of these materials such as brittleness, crack propagation, low tensile strength, wear resistance, and marginal accuracy, continued to limit their use only to the anterior region.

Aluminia and zirconia are the only two polycrystalline ceramics suitable for use in dentistry as framework materials able to withstand large stresses. These materials are shown to provide both necessary esthetics (tooth color) and strength[3].Yttriumoxide (Y_2O33%mol) is added to pure zirconia to control the volume expansion and to stabilize it in the tetragonal phase at room temperature. This partially stabilized zirconia has high initial flexural strength and fracture toughness[4]. Tensile stresses at a crack tip will cause the tetragonal phase to transform into the monoclinic phase with an associated 3-5% localized expansion. The volume increase creates compressive stresses at the crack tip that counteract the external tensile stresses. This phenomenon is known as transformation toughening and retards crack propagation. 600N is set as bench mark, which is considered to be the lower of static load bearing capacity for clinically acceptable bridges in the posterior regions, materials undergo cyclic fatigue loading and stress corrosion fatigue in the oral environment, the material used in these regions should have almost twice the load bearing capacity, as various studies concluded that as time progresses the load bearing capacity will almost reduced to half its original value due to above mentioned factors when used long term in oral cavity[5].

Zirconium oxide (zirconia), with its excellent strength and biocompatibility, bridges for the posterior region are considered as an indication[6], as many studies[7,8] have compared its survival and ability to withstand heavy occlusal forces with porcelain fused metal ceramic bridges, these studies were limited to 3 unit bridges. The

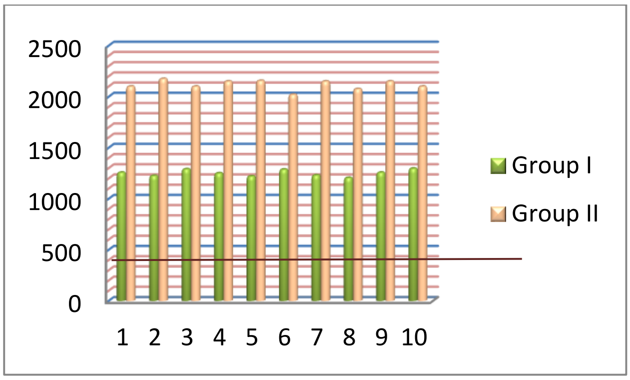
Table 1: Comparison of peak load of Group I and Group II.

Sample No.	Peak load (N)	
	Group I	Group II
1	1257	2100
2	1225	2175
3	1290	2100
4	1250	2150
5	1220	2156
6	1287	2020
7	1230	2150
8	1205	2075
9	1257	2150
10	1296	2100
Mean ± SD	1251.70 ± 31.81	2117.60 ± 47.36
t-test	124.452	141.389
Df	9	9
Significance (p-value)	0.000*	0.000*
95% Confidence Interval of the Difference		
Lower	1228.95	2083.72
Upper	1274.45	2151.48

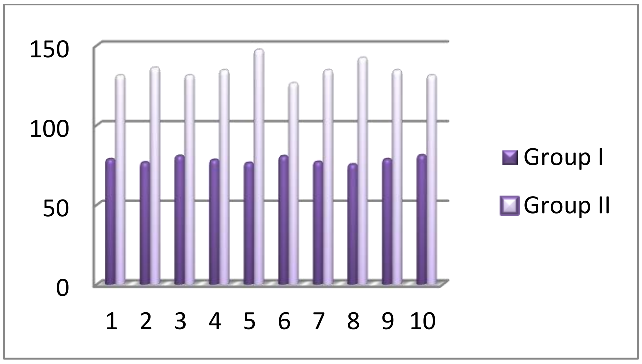
Table 2: Comparison of fracture resistance of Group I and II.

Sample No.	Fracture Resistance (Mpa)	
	Group I	Group II
1	78.56	131.25
2	76.56	135.94
3	80.62	131.25
4	78.13	134.37
5	76.25	147.25
6	80.44	126.25
7	76.88	134.37
8	75.31	142.19
9	78.56	134.37
10	81	131.25
Mean ± SD	78.23 ± 1.98	134.85 ± 5.98
t-test	124.466	71.25
Df	9	9
Significance (p-value)	0.000*	0.000*
95% Confidence Interval of the Difference		
Lower	76.81	130.57
Upper	79.65	139.13

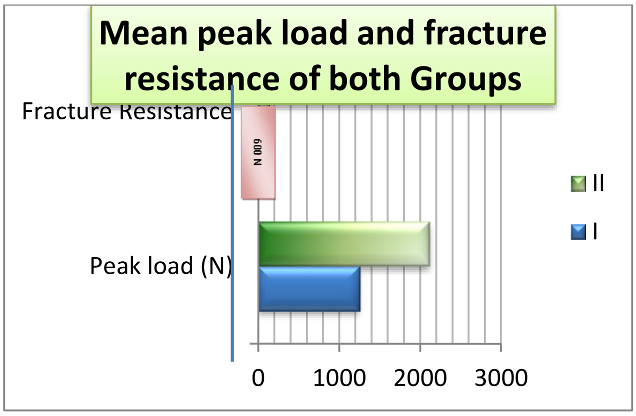
Graph 1 : Comparision Of Peak Load Of Group I And Group II Samples



Graph II: Comparision Of Fracture Resistance Of Group I And Group II Samples



Mean peak load and fracture resistance bars of the tested frameworks.



Graph 3: A line at the **600 N** denotes the maximum force in the posterior molar region.

aim of this study is to compare the in vitro load bearing capacity of long span (i.e 4 units zirconia coping) with Ni-Cr metal coping for porcelain fused metal bridges and its ability to withstand the occlusal load in posterior regions .

MATERIALS AND METHOD

In a maxillary jaw plastic model (Nissin - PRO2001-UL-SP-FEM 32) first premolar(24) and second molar(27) were prepared to accommodate a four-unit all-ceramic fixed partial denture. The chamfer type preparation allowed a crown wall thickness of 0.8 mm. The occlusal surfaces were reduced by 1.5 mm and the convergence angle was adjusted to 10^0. The prepared teeth were placed in their correct positions in the maxillary arch, and a full-arch impression was made of the prepared teeth with a silicone impression material **(**3 M ESPE**)**. The impression was poured in a resin material (Alpha-Die –Top;Schutz dental,Rosbach ,Germany) to create a preparation model that was invested, burned out, and cast in a nickel chromium alloy (Bellabond, Bego). In this way, a nickel chromium model of a three-unit FPD with fixed dies was fabricated. The surfaces of the dies were smoothed with rubber polishers, first with coarser, then with finer rubber cusps and points (Shofu Dental). The metal master model was used as a working cast to fabricate the FPDs and to evaluate their fracture strength in an axial load test.

To resemble the distance covered by a four-unit dental bridge from a second upper molar to a first upper pre-molar, the stainless steel holder is prepared with a distance of 23 mm between the centres of the master models. Using an addition silicone impressions are made of the stainless steel master models holder 20 die-stone models are made.

GROUP I: 10Die models for preparing zirconia coping using CAD/CAM.

GROUP II: 10Die models for casting procedure to prepare metal coping.

Group I

A die is scanned by a contact free optical processor (coritec 3d scanner for 12-15 minutes for a 4 unit fixed partial dental prosthesis .The CAD software designs an enlarged framework that is milled by using the Cad/Cam, esthetics System(imes icore, germany) with dimensiond of 0.8mm thick core ceramic and connector areas of the substructure between the abutment retainer and the pontics were modeled with an occlusogingival height of 4mm , ten identical stylized ceramic substructures intended for dental all-ceramic four unit fixed partial denture are made of isostaic pressed, Pre-sintered Y-TZP zirconia(Tanzadent , germnay) it takes 75-90 minutes to mill a 4 unit fixed partial denture prosthesis, the framework can be colored by sintering in a special automated oven for 8 hours .

Group II

Wax patterns were prepared on the die models for four unit metal copings, the wax patterns were invested and casting procedure was done using Ni-Cr metal pellets(Bellabond metal pellets temp 1350c) in induction casting machine (Bego), trimming and polishing procedures were performed and care was taken to maintain the metal thickness to 0.8mm.

AGING

The FPDs were fixed onto the abutments by means of a glass-ionomer cement (Ketac-Cem; 3M Espe, Seefeld, Germany). During their 200 d of storage in distilled water at 36 ^{0}C, the FPDs were subjected to thermal and mechanical cycling (TMC). A total of 10^4 thermal cycles, between 5 and 55 ^{0}C (30 s dwell time at each temperature), and 10^6 cycles of mechanical loading with 100 N as the upper limit (load frequency 2.5 Hz), were applied successively.

STATISTICS

The results obtained were tabulated and subjected to statistical analysis using SPSS (Statistical Package for Social Sciences) Version 20.0 statistical Analysis Software.Mean and Standard deviations were calculated for all the measurements of peak load and fracture resistance for both Zirconia (Group I) and metal framework (Group II). **One-sample student 't' test** was used to compare means of various parameters and to obtain the probability levels. The values were.. represented to know their significance. P-value < 0.05 is considered as level of significance. The **95% Confidence Interval of the Difference** was also calculated for all respective groups.

RESULTS

The specimens were loaded until fracture, this was carried out in an Universal testing machine for the three – point flexural test . The rate of load application was adjusted to 0.5 N/s at a crosshead speed of 1 mm/ min with the force transferred to the occlusal connector area between teeth 25 and 26 via a tungsten carbide ball (diameter 6.0 mm) on an interposed tin foil (thickness 0.2 mm).A sudden decrease in force of more than 15 N was regarded as an indication of failure, and the maximum force up to this point was recorded as force at fracture. The load causing the initial fracture was recorded [Fig1and 2]. Fracture resistance was then automatically calculated by the equipment software and displayed. All the data were recorded, tabulated and subjected to statistical analysis. The mean and standard deviation (SD) for each group were determined. The data were analyzed for differences using student t test to determine statistically significant differences between the means. The load bearing capacity and fracture resistance was recorded and is tabulated in [Tables 1-2], respectively graphs were drawn using the same values[Graph 3 and 4].

The values obtained for respective groups are:

Load bearing capacity group I **- 2117.60N.** group II **- 1251.70 N,**

Fracture resistance group I **- 134.85Mpa.** group II– **78.23 Mpa.**

DISCUSSION

In the present study, the fracture resistance of four-unit all-ceramic FPDs was investigated in vitro. Compared to in vivo studies, in vitro experiments are less expensive, easier to reproduce, and less exposed to influences that are at best known qualitatively.

The possibility to imitate, control, and reproduce an artificial oral environment under exactly defined conditions, and to simulate long-term effects effectively, in a time-saving manner, is a further advantage. Model materials and testing conditions were chosen carefully to imitate clinical reality as faithfully as possible. The abutment teeth and their bases were made of NiCr metal. Finally, the FPDs were luted to their abutments by means of a clinically proven glassionomer cement[9,10] . Preparation of teeth, and dimensions of the framework, and connectors were performed according to the manufacturers recommendations and previous studies i.e:

- Connector dimensions 0f 4mm x 4mm,
- Thickness of coping 0.8mm,
- Finish lines given was chamfer.

To achieve the best reproducibility possible for specimens, frameworks made of presintered Y-TZP, and metal frameworks casted by wax patterns prepared were all copies of one master model. Connector dimensions were taken as 4x4 mm as a study by R.D. Lakshmi et al[11] where they reported increasing the connector dimensions up to 4 mm x 4 mm in 3 unit inlay retained Fixed Dental Prosthesis has a significant improvement in the stress distribution among monolithic zirconia and lithium disilicate materials, making it suitable as a posterior restorative material in all ceramic inlay retained Fixed Dental Prosthesis and also another study by S. Mir Mohammad Rezaei et al[12] reported increasing the connector width decreases the failure probability when a vertical or angled load is applied. The connectors were designed with an elliptical cross-section, which, according to a recent finite element study, performs best of all possible designs investigated (K.-J. Erdelt, F. Beuer, W. Gernet, P. Pospiech, unpublished data).

Chamfer finish lines were preferred as various studies favoured this type of finish line for zirconia bridges and one such study was by Jalalian et al[13] where they concluded chamfer margin could improve the biomechanical performance of posterior single zirconia crown restorations. This may be because of strong unity and round internal angle in chamfer margin. And *thickness of coping*, a study by Sunting et al[14] reported fracture resistance of zirconia crowns with a thickness of 1.0mm can be equal to metal ceramic crowns. Doubling the monolithic zirconia core from 0.6 mm to 1.5 mm increases the fracture resistance of this restorative system threefold.

Artificial aging was performed, and to simulate artificial aging under clinical conditions, two factors had to be taken into account. First, the presence of water in the oral cavity, aqueous environments generally facilitates subcritical crack growth in ceramics and may cause uncontrolled transitions from the tetragonal to the monoclinic structure of Y-TZP [15]. Both produce a decrease in the strength of the ceramic. Second, material strength is degraded by the repeated thermal stressing of prosthetic restorations, which results from temperature changes caused by the consumption of hot and cold food and drink, and by breathing.

The importance of aging the material was reported by Kohorst P et al[16] in their study they stated after aging FPDs made from presintered and fully sintered, materials were capable of withstanding occlusal forces generated in the posterior region. Therefore Y-TZP may be suitable for posterior four-unit all-ceramic FPDs. Palmer et al[8] reported a temperature range between 0^0 and 67^0 C for food and drink, which results in temperature changes between 5^0 and 55^0 C at teeth and dentures. Accordingly, this interval also chosen in other in vitro investigations[17] was adopted for the 10^4 thermal cycles applied in this study.

In the present study, the maximum force recorded before the occurrence of a first force drop of more than 15 N was regarded as force at fracture irrespective of the cause.

Indeed 8 of 10 zirconia FPDs failed with sudden bulk framework fracture at this event, whereas 2 bridges showed cracking beneath the indenter. And for metal frameworks intially they showed sag of the framework and then fracture in the connector region (Fig 3,4).

The values obtained for respective groups are :

Load bearing capacity group I - **1251.70 N,** Load bearing capacity group II - **2117.60N.**

Fracture resistance group I – **78.23 Mpa ,** Fracture resistance group II - **134.85Mpa**

The choice of material proved to have a significant influence on the load capacity of FPDs. The superior load capacity, exhibited by FPDs were fabricated from NiCr metal, these material has been shown to possess higher bending strength compared with an experimental presintered, shaped and afterwards densely sintered zirconia[18].

The most obvious reason for the significant difference in strength can be due material used and discrepancies in both the density and homogeneity of the materials investigated. With regard to the load-bearing capacity of posterior four-unit all-ceramic FPDs, there is only few publications in the literature. One deals with FPDs milled of fully sintered zirconia; the other two discuss FPDs made from presintered zirconia.

In general, they confirm our findings, subject to the usual difficulties encountered when comparing results from different load to failure tests, as a result of the many influential variables being uncontrolled.With FPDs fabricated of fully sintered zirconia, Tinschert et al[19] obtained an average load-bearing capacity of 1,607 N, which is about 27 % higher than the corresponding value presented here. This may be attributed to unaged specimens used in their study and the load being more evenly distributed during testing. Rountree and collaborators found a somewhat lower force at fracture (979 N) than those obtained in this study, but they applied only half of the dynamic loading (50 N).(P. Rountree, F. Nothdurft, P. Pospiech, unpublished).In contrast, Luthy et al[20] measured lower forces at fracture (755N), loaded unveneered frameworks that had comparatively low cross-sectional areas in the connector regions (7.3 mm) and had not been subjected to TMC.

For groups of FPDs tested in this study, the forces at fracture were approximately 200 to 240 % higher than the 600 N benchmark [Graph 3], which is considered to be the lower limit of static load-bearing capacity for clinically acceptable bridges in the posterior region[21], taking into account cyclic fatigue loading and stress corrosion fatigue caused by the oral environment[22].

Because the specimens had undergone a simulated aging that corresponds to approximately 15 months of service life only, prolonged aging experiments are needed to prove sufficient life expectancy. However, zirconium dioxide, either processed by HIP and fully sintered or presintered, is a promising material for fabricating four-unit all-ceramic zirconia FPDs in the posterior region. It combines optimal aesthetic quality and outstanding biocompatibility with high strength, which to date, could be achieved only by porcelain-fused-to-metal restorations. However, clinical trials with different bridge designs are still necessary to provide conclusive evidence of long-term reliability under in vivo conditions.

The present study used Ytria stabilized zirconia material for posterior long span bridge to calculate load bearing capacity and fracture resistance after artificial aging was performed. But the present study did not aim to investigate the effect of various design parameters for constructing a long span bridge and their effect on strength of the bridge, it is not affirm whether or how these factors affected the load bearing capacity and fracture resistance results.

Further studies are needed to evaluate other mechanical properties which play a role in determining the survival of the zirconia material intraorally and compare various brands of zirconia from different manufacturers which claim to have fracture resistance similar to metal.

CONCLUSION

A laboratory investigation of the load bearing capacity and fracture resistance of four unit fixed partial dentures was performed and the following conclusions can be drawn .

1. The mean load bearing capacity and fracture resistance of zirconia compared to NiCr metal coping were significantly different, metal fixed partial denture coping have higher values

2. The load bearing capacity of zirconia fixed partial dentures copings is twice as much as the maximum forces in the posterior molar region (600N) of oral cavity.

3. It is concluded that zirconia fixed partial denture coping with its excellent aesthetic properties with many exceptional mechanical properties (load bearing capacity and fracture resistance) can be used in the posterior regions of oral cavity.

CONFLICT OF INTEREST

No potential conflict of interest relevant to this article was reported.

REFERENCES

1. Land CH. Porcelain dental art: No.II. Dent Cosmos 1903;45:615-20.5.

2. McLean JW, HughesTH. The reinforcement of dental porcelain with ceramic oxides. Br.Dent J 1965;119:251-5.

3. Raigrodski AJ. Contemporary materials and technologies for all ceramic fixed partial dentures: a review of the literature. J Prosthet Dent 2004;92:557-62.

4. Luthardt RG, Sandkuhl O, Reitz B. Zirconia- TZP and alumina--advanced technologies for the manufacturing of single crowns. Eur J Prosthodont Restor Dent 1999; 7:113-9.

5. Behrens A, Nesslauer H, Hauptmann H (2005) Fracture Strength of Sandblasted and Silicatised Coloured and Non-coloured Zirconia.

6. Behrens A, Nesslauer H, Hauptmann H (2005) Fracture Strength of Sandblasted and Silicatised Coloured and Non-coloured Zirconia.

7. Drummond JL. In vitro aging of yttria stabilized zirconia. J Am Ceram Soc 1989; 72: 675–676.

8. Palmer DS, Barco MT, Billy EJ. Temperature extremes produced orally by hot and cold liquids. J Prosthet Dent 1992;67: 325–327.

9. Fleming GJP, Curtis AR, Marquis PM (2005) Alumina abrasion and grinding effects on yttria-stabilized zirconia ceramic .

10. Kern M, Douglas WH, Fechtig T, Strub JR, Delong R.Fracture strength of all-porcelain, resin-bonded bridges aftertesting in an artificial oral environment. J Dent 1993; 2: 117–121.

11. R.D. Lakshmi*, A. Abraham, V. Sekar, A. Hariharan Influence of connector dimensions on the stress distribution of monolithic zirconia and lithium-di-silicate inlay retained fixed dental prostheses A 3D finite element analysis Tanta Dental Journal 2015;15:56-64.

12. S. Mir Mohammad Rezaei , H. Heidarifar, F. Fallahi Arezodar, A. Azary, S. Mokhtar Influence of Connector Width on the Stress Distribution of Posterior Bridges under Loading. Journal of Dentistry TehranTehran, Iran 2011; 2: 67-74

13. E. Jalalian, B. Atashkar, R. Rostami the effect of preparation design on the fracture resistance of zirconia crown copings (computer associated design/computer associated machine, cad/cam system) journal of dentistry, tehran university of medical sciences, tehran, Iran. 2011; 8:212-219.

14. Sun Tinga, Zhou Shanyub, Lai Renfaa, Liu Ruoyub, Ma Shuyuana, Zhou Zhiyinga, Shao Longquanb Load-bearing capacity and the recommended thickness of dental Monolithic zirconia single crowns. J Mech Behav Biomed Mater. 2014; 35:93-101

15. Wiederhorn SM. Moisture assisted crack growth in ceramics. Int J Fracture Mech 1968; 4: 171–177.

16. Kohorst P, Herzog TJ, Borchers L, Stiesch-Scholz M. Load-bearing capacity of all ceramic posterior four-unit fixed partial dentures with different zirconia frameworks. Eur J Oral Sci 2007; 115: 161–166.

17. Gale MS, Darvell BW. Thermal cycling procedures for laboratory testing of dental restorations. J Dent 1999; 27: 89–99.

18. Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of allceramic materials. Part II. Zirconia-based dental ceramics. Dent Mater 2004; 20: 449–456.

19. Tinschert J, Natt G, Mautsch W, Augthun M, Spiekermann H. Fracture resistance of lithium disilicate, alumina, and zirconia-based three-unit fixed partial dentures Int J Prosthodont. 2001;14: 231-8

20. Lu¨thy H, Filser F, Loeffel O, Schumacher M, Gauckler LJ, Hammerle CH. Strength and reliability of four-unit allceramic posterior bridges. Dent Mater 2005; 21: 930–937.

21. Ko¨rber KH, Ludwig K. Maximale Kaukraft als Berechnungsfaktor zahntechnischer, Konstruktionen [Maximal biteforce, a factor in designing dental constructions]. Dent Lab 1983; 31: 55–60.

22. Drummond JL. Ceramic behavior under different environmental and loading conditions. In: Eliades G, Eliades T, Brantley WA, Watts DC, eds. Dental materials in vivo: aging and related phenomena. Chicago: Quintessence Publications, 2003; 35–45. 166.