Fracture strength of all-ceramic (Y-TZP) three- and four-unit fixed dental prostheses with different connector design and production history

Deyar Jallal Hadi Mahmood, Ewa H Linderoth, Per Vult von Steyern, Ann Wennerberg

Abstract

◎ Aims: To investigate how different default settings for connector design of two different CAD/CAM systems, i.e. to compare how different radii of curvature in the embrasure area of the connector affect the fracture strength and the fracture mode of 3-unit and 4-unit all-ceramic FDPs made from Y-TZP and to investigate how the number of pontics affect the fracture strength of Y-TZP.

Material and methods: 32 all-ceramic three (3Z) and four (4Z) -unit, Y-TZP, FDP cores were made and divided in 4 groups, with 8 FDP in each group. The groups 3Z:1 and 4Z:1 were generated with a mechanical scanner, Procera® Forte and the FDPs in group 3Z:2 and 4Z:2 were generated with an optical scanner, NobelProcera® Scanner. The connector dimensions were set to 3 mm x 3 mm and core was set to 0.7 mm. The design of radius of the gingival and occlusal embrasures in the connector areas was set according to default settings and the manufacturer's recommendations. All the FDP cores were subjected to heat treatment, thermocycled for 5,000 cycles, preloaded for 10,000 cycles to simulate ageing and finally loaded to fracture.

Results: Regarding connector design a significant difference was found between group 3Z:1 and 3Z:2 ($P<0.05$), and group 4Z:1 (50% of the FDPs fractured during preloading 30-300N) and 4Z:2 ($P<0.05$). An extra pontic decreased the fracture strength up to 45%. Conclusions: The default settings of the two different CAD/CAM systems had a great impact on the fracture strength. It is important that a CAD/CAM system is equipped with possibilities to design a connector that fulfils the clinical demands of mechanical function and longevity. The most crucial factor for the load-bearing capacity is the design of the radius of the gingival embrasures. Increasing the number of pontics from three to four decreases the load-bearing capacity nearly twice.

Key words

All-ceramic FDPs, connector-design, CAD/CAM, radii, 3-4 unit FDPs

Department of Prosthetic Dentistry, Faculty of Odontology, Malmö University, Malmö, Sweden

Brotthållfasthet hos helkeramiska (Y-TZP) tre- och fyra-leds broskelett: betydelsen av connectordesign och produktionshistorik

Deyar Jallal Hadi Mahmood, Ewa H Linderoth, Per Vult von Steyern, Ann Wennerberg

Sammanfattning

Yttriumoxidstabiliserad zirkoniumdioxid (Y-TZP) används frekvent inom tandvården. Helkeramiska broar måste dimensioneras och designas med hänsyn till bromaterialets specifika materialegenskaper. Särskilt kritisk är utformningen av connectorområdet. Rekommendationerna avseende connectordimensionerna för anteriora 3- och 4-ledsbroar av Y-TZP varierar mellan 2 till 4 mm i vertikal riktning och mellan 2 till 4 mm i horisontell riktning. Utöver dimensioneringen måste hänsyn tas till connectorns utformning då connectorns gingivala radie har en stor betydelse för konstruktionens totala brotthållfasthet. Vid belastning utsätts broar med liten gingival radie för högre spänningskoncentrationer i connectorområdet jämfört med broar med större gingival radie. Den gingivala utformningen av connectorer kan ha en ännu större betydelse för brotthållfastheten om antalet hängande led utökas från ett till två.

Syftet med föreliggande studie var att undersöka hur förinställningarna avseende design av connectorer i två olika CAD/CAM-system påverkar brotthållfastheten, hur den gingivala radiens storlek påverkar brotthållfasthet och frakturmönster samt undersöka hur brospannets längd, dvs antalet hängande led påverkar brotthållfastheten hos 3- och 4-ledsbroar av Y-TZP.

Två olika CAD/CAM-system användes för att framställa totalt 32 broskelett av Y-TZP, varav 16 st 3-ledsbroar med ett hängande led och 16 st 4-ledsbroar med två hängande led. Broskeletten delades in i 4 grupper baserat på antal led och utformning av connectorernas gingivala radie. För att framställa grupperna 3Z:1, 4Z:1 användes en mekanisk scanner, Procera®Forte och för grupperna 3Z:2, 4Z:2 användes en optisk scanner, NobelProcera® Scanner.

Samtliga broar utsattes för värmebehandling (simulerad porslinspåbränning), termocykling (5°C – 55°C / 5000 cykler), cyklisk förbelastning (30 – 300N / 10 000 cykler) och belastades slutligen till brott. Samtliga broar fakturerade i connectorområdet och efter okulär besiktning kunde man konstatera att samtliga frakturer startat i connectorns gingivala del. Medelvärdena för last vid brott var i grupp 3Z:1 734 N, grupp 4Z:1 405 N, grupp 3Z:2 910 N samt i grupp 4Z:2 520 N.

Med reservation för de begränsningar som en *in-vitro* studie innebär dras följande slutsatser: Förinställningarna i de olika CAD/CAM systemen hade avgörande betydelse för brotthållfastheten. Eftersom connectorområdet utgör den svagaste punkten i en helkeramisk brokonstruktion är det avgörande att CAD/CAM systemens standardinställningar tillåter en konstruktionsdesign som uppfyller de mekaniska och kliniska kraven.

Storleken på den gingivala connectorradien har avgörande betydelse både för brotthållfastheten och för frakturmönster. Genom att öka antalet hängande led från ett till två hängande led minskar brokonstruktionens brotthållfasthet med ca 50 %.

Introduction

Yttria-stabilised tetragonal zirconia polycrystals (Y-TZP) was introduced in dentistry in the nineteen-nineties. Pure zirconium-dioxide (zirconia) exists in three crystalline phases at different temperatures. The material is unsuitable for structural or mechanical applications in those stages. By adding one of several oxides, yttria, ceria, calcia or magnesia (so called stabilizers), which dissolve within the crystal structure of zirconia, it is possible to densify the material in the tetragonal phase range to yield a fine-grained microstructure consisting almost completely of tetragonal grains, metastable at room temperature (Y-TZP) (8,16,29).

If a crack is formed in the material, each tetragonal grain has the potential to transform into a monoclinic one with a volume expansion of 4-5% near the crack tip and in this way inhibit further propagation of the crack. This mechanism is described as transformation toughening or phase transformation (16), giving the material great potential for stress-bearing and wear-resistant applications even if the transformation is a one way progression. However, when the transformation takes place, the crack hindering effect cannot prevent further crack growth (29).

To ensure optimal clinical performance, the dimensions of a fixed dental prostheses (FDPs) in general and of the connectors in particular, must be adequate since stress concentrations of the framework are developed at the intersections of the pontics. Important factors are the connector diameter, the diameter and the number of the pontics and the radius of curvature at their intersection (17,27).

Recommendations for connector dimensions for Y-TZP vary from 2 to 4 mm in occluso-gingival height and 2 to 4 mm in bucco-lingual width (6,14,17,28,30,31). Furthermore, it has been shown that, in order to reduce the fracture probability when designing all-ceramic FDPs, the shape of the connector is an important factor to consider (9,20,21). In particular, the radius of curvature at the gingival embrasure plays a significant role in the load-bearing capacity. FDPs with small gingival embrasure radii are subjected to high stress concentrations in the connector area during loading, compared to FDPs with large embrasure radii. Previous studies show that a larger radius of the gingival embrasure results in significantly higher fracture resistance compared to FDPs with smaller embrasure radii. In conclusion, it has been shown that not only the material properties, but also the design of an all-ceramic FDP, play an important role for the strength of the reconstruction. Hence, it is important for both dental technicians and dentists to have control of the design parameters as well as the material used (1,20).

Computer-aided design and computer-aided manufacturing (CAD/CAM) systems are used for producing Y-TZP FDPs. There are several CAD/CAM-systems on the market but in each system the design is basically dependent on the design limitations of the CAD-software and the milling properties of the CAM-system (1,9,20,21). Many of the design parameters can be adjusted to fit each case to be milled individually, while other are not and instead fixed as default settings determined by the manufacturers. The limitations in the CAD/CAM systems have sometimes led to FDPs with a particular type of connector design. It differs from the other designs, having small radii approximally, connected with a straight beam, this have been shown to reduce the strength of all-ceramic FDPs (1,30,31). Thus, it could be possible that the design limitations of different CAD/CAM system could have an influence on the fracture strength, and consequently the clinical outcome of all ceramic FDPs. The aims of this study were:

- to investigate how different default settings of two different CAD/CAM systems, i.e compare how different radii of curvature in the embrasure area of the connector affect the fracture strength and the fracture mode of 3-unit and 4-unit all-ceramic FDPs made from Y-TZP.
- to investigate how the length of the FDPs (i.e. the number of pontics) affect the fracture strength of Y-TZP.

Material and methods

A total of 32, all-ceramic, Y-TZP, 16 three- and 16 four-unit FDP cores with differing design were made. There were one or two intermediate pontics, supported by end abutments. The FDPs were then divided according to design in four groups (3Z:1, 3Z:2, 4Z:1 and 4Z:2), including 8 FDPs in each group.

Preparation

For the fabrication of the master model for the three-unit FDPs, a plastic model of an upper jaw (KaVo YZ, OK VZ 623 0401 180, KaVo Dental, GmbH, Germany) was used. Two abutment preparations were made on the left central incisor and the left canine, the left lateral incisor was removed. Two abutment preparations were made for the four-unit FDPs on the right central incisor and the left canine, the left central incisor and the left lateral incisor were removed. The aim was to design a preparation with a 120° chamfer and an angle of convergence of 15°. Subsequently a full arch A-silicone (Flexitime Mono Phase, Heraeus Kulzer, GmbH, Germany) impression was made and poured with die stone (Everest® Rock, Type 4 die stone, KaVo Dental, GmbH, Germany) to produce two master cast.

Scanning

Two different scanners were used to perform the FDPs. Data for groups 3Z:1 and 4Z:1 was generated with a mechanical scanner, Procera® Forte (Nobel Biocare, Zurich, Switzerland). The master casts were scanned and the data was transferred to a computer equipped with CAD software (Procera CAD Design C3D, version 2.60) where the intended design of the FDPs was established. The connector dimensions were set to 3 mm x 3 mm and the minimum thickness of the core was set to 0.7 mm. The design of radius of the gingival and occlusal embrasures in the connector areas was set according to the default settings in the CAD program, and according to the manufacturer's recommendations. Data for the FDPs in group 3Z:2 and 4Z:2 were generated with an optical scanner, NobelProcera® Scanner (Nobel Biocare, Zurich, Switzerland). The master casts were scanned and the data of the NobelProcera® optical scanner was transferred to a computer equipped with CAD software (NobelProcera 3D prosthetic design, version 4.1.10.4.) where the intended design of the FDPs was established according the protocol for the first two groups. The CAD data for the FDPs was subsequently sent to a production center (Procera® Production center, Stockholm, Sweden) where it was used for the production of the 32 FDPs in Procera® Zirconia bridge material, following the regular production line. Finally the FDPs were sent back to the Faculty of Odontology, Malmö University, where they were to be tested.

Heat treatment

All FDP cores were subjected to heat treatment in a calibrated porcelain furnace (Ivoclar P 500, Ivoclar Vivadent AG Schaan/Liechtenstein) to simulate the firing cycles of the veneering porcelain (Nobel Rondo™ Zirconia, Nobel Biocare, Gothenburg, Sweden) recommended for the core material used. Four firing programs were used: Liner, Dentin 1, Dentin 2 and finally Glaze as shown in Table 1 (Table 1.), all according to the manufacturer's instructions.

Ageing procedures

Two stages of ageing were performed for all FDPs. In the first stage, all FDP cores underwent thermocycling (LTC Multifunctional Thermocycler, LAM Technologies electronic equipment, Italy) using a small basket controlled by a device driver. All FDP cores underwent 5,000 thermocycles in two water baths at temperatures of 5 °C and 55 °C. The cores were placed in a basket for transfer between the two baths. Each cycle lasted 60 seconds, 20 seconds in each bath and 10 seconds to complete the transfer between the baths. After thermocycling, the cores were dried.

Supporting tooth analogues

Tooth analogues for the testing procedure were made simultaneously with the production of the FDPs, as inspection blocks from the same CAD-file as the one used to produce the FDP cores at the production center (Procera® Production center, Stockholm, Sweden). According to previous studies 32 inspection blocks of a polymer material were made to enable precision check and to support the FDPs during testing (20).

Cementation

Prior to cementation, the tooth analogues were steam cleaned and subsequently treated with ED primer II A and B (Kuraray Medical INC, Tokyo, Japan) which was applied to the cementation surfaces according to the manufacturer's instructions.

The FDPs of all four groups were luted onto the tooth analogues with Panavia F 2.0 luting cement (Kuraray Medical INC, Tokyo, Japan) using both light and Oxyguard II (Kuraray Medical INC, Tokyo, Japan), according to the manufacturer's recommen-

Table 1. General Firing Program. NobelRondo™ Zirconia Ceramics

Firing Program	Pre-heating temp.	Pre-heating drying time (min)	Heating Rate Y-TZP	Firing temp.	Holding time (min)	Vacuum (hPa)
Liner firing	575 °C	8	45/55 °C	930 °C	1	50
Dentin firing 1	575 °C	9	45/55 °C	910 °C	1	50
Dentin firing 2	575 °C	8	45/55 °C	900 °C	1	50
Glaze firing	575 °C	5	45/55 °C	880 °C	1	-

dations. During setting of the cement, all FDPs were loaded in the direction of insertion with a force of 15 N for a period of 60 seconds. After cementation the FDP cores were placed in a plastic container, with water covering the bottom surface and a sealing lid, to create a moist environment to prevent desiccation of the luting cement.

Second stage of ageing of FDP cores

In the second stage of ageing, all FDP cores underwent cyclic preloading at loads between 30 and 300 N, comprising 10,000 cycles and a load profile in the form of a sine wave at 1 Hz. In all groups the force was applied with a stainless steel intender, 2.5mm in diameter, placed centrally on the incisal edge of the pontic at the three-unit FDPs, and centrally on the lateral incisor pontic at the four-unit FDPs, to avoid sliding during loading. During preloading all FDPs were stored in distilled water and mounted at a 10° inclination relative to the vertical plane.

Load to fracture

In the final stage of testing, the FDPs were mounted in a testing jig at a 10° inclination and subjected to a load applied by a universal testing machine (Instron 4465, Instron Co. Ltd, Norwood, MA, USA). The crosshead speed was 0.255mm/min and the load was applied with a stainless steel intender placed centrally on the incisal edge of the pontic at the three-unit FDPs, and centrally on the lateral incisor pontic at the four-unit FDPs. In the final step of the procedure, the FDPs were loaded until a fracture occurred, whereupon the loads at fracture were registered. Fracture was defined as a visible fracture through the entire construction.

Table 2. Fracture load (N), mean fracture load and standard deviation

Group	3Z:1 3-unit	3Z:2 3-unit	4Z:1 4-unit	4Z:2 4-unit
Significance	P<0.05		P<0.05	
FDP No				
1	753	805	300*	575
2	749	953	300*	512
3	849	697	499	455
4	735	1070	300*	564
5	767	843	508	430
6	830	997	530	574
7	610	921	300*	503
8	575	999	499	547
Mean(N)	734	910	405	520
SD	96	122	112	55

*FDPs fractured at 30-300 N preload

Throughout the test period, whenever the FDP cores were not being actively tested, they were stored in distilled water.

Statistical analyses

Any differences between groups were tested by means of the student's t-test and were performed to detect any significant differences ($p < 0.05$) between the groups.

Results

The fracture data are listed in Table 2. During the cyclic mechanical preload (30-300N) four of the FDPs in group 4Z:1 fractured.

All the 32 FDPs in the present study were examined to establish the fracture modes which were distributed as follows:

- In group 3Z:1 six of the FDPs fractured at the distal connector area on the left central incisor and through the pontic. Two FDPs fractured at the mesial connector area of the left canine and through the pontic. All FDPs fractured at the sharp mesial corner of the connector.
- In group 4Z:1 seven of the FDPs fractured at the mesial area of the connector on the left lateral incisor and through the pontic. One FDP fractured at the mesial connector area of the left canine and through the pontic. All FDPs fractured at the sharp mesial corner of the connector.
- In group 3Z:2 four of the FDPs fractured at the distal connector area on the left central incisor and through the pontic. Four FDPs fractured at the mesial connector area on the left canine and through the pontic. All fractures were located in the center of the connector area, where the connector is thinnest.
- All the FDPs in group 4Z:2 fractured at the mesial connector area on the left lateral incisor and through the pontic. All fractures were located in the center of the connector area, where the connector is thinnest.

Discussion

To simulate clinical conditions the FDPs underwent different ageing treatments. According to some studies, the temperatures that the core is subjected to during porcelain firing may decrease the mechanical properties of the ceramic material. In the case of Y-TZP a possible explanation is that machine grinding initiates the tetragonal to monoclinic transformation, creating a compressive layer, and that these residual stresses are relaxed during porcelain firing reversing the transformation. Transformation from the monoclinic to the tetragonal meta-stable phase may occur as soon as a given temperature is reached, regardless of the holding time (10,13,30,31).

Heat-treating Y-TZP to simulate porcelain firing, a preheating process with or without actual veneering, can be carried out, both to exclude the risk that the porcelain itself affects the results but also because the veneer material is more complicated to build up in a standardised way, as generally veneering requires a high level of skilled workmanship. It was therefore decided to exclude this production step in the present study.

The FDPs can be thermo-cycled in order to simulate ageing and to expose the materials to fatigue. The change in temperature creates stresses corresponding to mechanical stresses in the mouth. The wet environment may also affect the materials by enhancing micro-crack growth due to stress corrosion. The strength degradation rate is a slow process affecting the all-ceramic material differently depending on several micro-structural parameters, such as – in the case with Y-TZP - yttrium distribution and concentration, the distribution of flaw populations and grain size (2,5,38).

Cyclic preloading in an aqueous environment can be performed to simulate ageing of the material in the oral cavity during function. It has been reported that ceramic materials show an abrupt strength degradation and transition in damage mode after multi-cyclic loads compared to static loading tests. Hence, it is essential to consider fatigue and environmental influence, as water in the saliva enhances crack-growth in a ceramic reconstruction when subjected to small alternating forces during mastication in the clinical situation (14,35).

According to several previous studies a test method and tooth analogues, with a low elastic modulus, was used, since the test method must reflect the range and distribution of strength, with consideration being paid to the brittle nature of the material (18,29,37).

In the present study the connector design and core thickness of the FDPs in all groups was made according to the default settings of the CAD software in order to find out how the settings for the connector design, i.e. their consequences upon the design, would affect the over-all load-bearing capacity of the FDPs. Default settings are standard in many CAD/CAM systems, as an help for the operating dental technician but they might also be an adaption to the limitations of the size and form of the milling pins or the numbers of axis in the CAM system. The default setting is often designed by software programmers without specialist knowledge of designing FDPs, the tools of the software is locked and can´t be adjusted by the operating dental technician. Those limitations of the default settings might result in FDPs with inadequate design. The intention with using two different CAD/CAM systems and compare the results (Table 2.) was to establish two different connector designs (Figure 1. and 2.), presuming that the default settings would be different from each other since the first system is older than the second one. In the group 3Z:1 and 4Z:1 the default settings of the CAD system resulted in FDPs with a straight beam shaped connector with a small radii at the intersection area, creating a sharp corner or notch, in the transition area. (Figure 1) (14,30,31). A different system was used to scan and design the FDPs in group 3Z:2 and 4Z:2 which resulted in a connector design more according to the recommendations, i.e. with an arch shaped connector with larger gingival embrasure radii and smooth transmissions in the intersection areas (1,3,7,9,10,15,19,20,21,23,25).

All the FDPs in the present study fractured in the connector area which corresponds with previous

Figure 1. Connector design group 3Z:1, 4Z:1

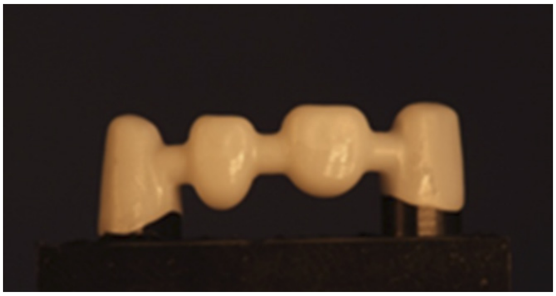
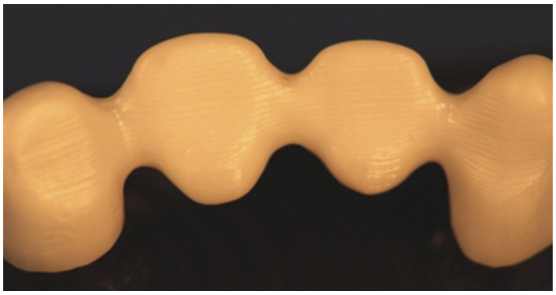


Figure 2. Connector design group 3Z:2 and 4Z:2



studies of all-ceramic FDPs (1,7,9,12,17,18,20,21,22,26,30,31,32,34).

Comparing the different connector design according to fracture strength a significant difference was found between group 3Z:1 and 3Z:2 ($P<0.05$), and group 4Z:1 and 4Z:2 ($P<0.05$). This shows that a design of the FDPs, with an arch shaped connector, larger gingival embrasure radii and smooth transmissions in the intersection areas gives a minimum 20% higher load-bearing capacity, which supports the recommendations in a previous study (1).

When analysing the fracture mode it could be stated that there is a relation between the design of the FDPs and the fracture mode. A 3-point bend test was used to test the FDPs. Such test is known to create compressing loads on the loaded side and tensile stress on the opposite side, causing a crack initiation there (22,23). It is also known that ceramic materials are extremely sensitive to notches or indentations and that such on the tensile stress side probably will be where a crack originates. To prevent a crack in the ceramic material caused by tensile load, increasing the radii in the insertion area and design the connector with a large radii of curvature is important factors to consider in order to increase the all over fracture strength of the construction (1). In group 3Z:1 and 4Z:1 with the straight beam design and a small radius at the insertion area, the total fractures seems to have started at the sharp corner (Figure 3.). In group 3Z:2 and 4Z:2 the total fractures seems to origin in the zenith of the arch shaped connectors, which is logic as the bulk of material is thinnest there (Figure 4.)

The results in this study show that the design affects both fracture strength and the fracture mode. The results emphasises that, to be able to take advantage of the properties of the ceramic materials all FDPs must be properly designed, i.e. without any sharp angels, especially in the transition areas between coping or pontic and connector. The design in the 3Z:1 and 4Z:1 group is a design contraindicated to all recommendations and to previous studies since the radius of curvature at the gingival embrasure plays a significant role in the load-bearing capacity. FDPs with small gingival embrasure radii are subjected to higher stress concentrations in the connector area during loading, compared to FDPs with large embrasure radii. Even if the studied material in the present study is relatively limited the result is according to previous studies that show that a larger radius of the gingival embrasure between each unit results in significantly higher fracture resistance compared to FDPs with smaller embrasure radii (1,3,7,9,10,15,19,20,21,23,25). With an oval shaped connector designed in the CAD-software the radius of the gingival embrasure increases in size. As the more increased radius of connector curvature at the gingival embrasure and between each unit, the higher the total fracture strength of the FDP (1,3,9,10,15,19,20,21,23,24).

According to the manufacture Nobel Biocare, "Procera® zirconia has a flexural strength 1120 MPa and is recommended for single crown up to long span FDPs". The load-to-fracture values are in all groups significantly lower than the values unveiled by the manufacturer, which probably can be explained by the design of the FDPs.

Comparing the mean values in group 3Z:1 and 4Z:1 the shorter FDPs show that the extra pontic in group 4Z:1 decreased the fracture strength with at least 45 %. There would probably be a greater decrease than shown here due to the estimated values of the FDPs that fractured during preloading. In groups 3Z:2 and 4Z:2 the extra pontic in group 4Z:2 decreased the fracture strength with 43%. Due to *Luthy et al* (19) increasing the number of pontics, i.e. from three to four, will decrease the over-all load-bearing capacity approximately two times. The results in the present

Figure 3. Fracture mode group 4Z:1. Crack showed with arrow

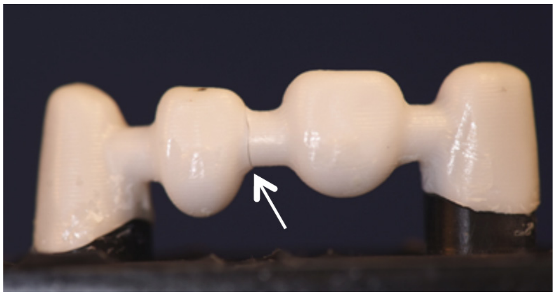
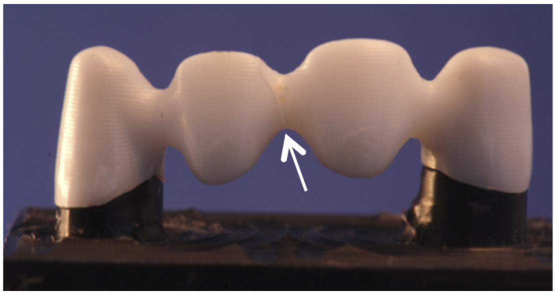


Figure 4. Fracture mode group 4Z:2. Crack showed with arrow



study support their findings, but demand some comments. The only groups that can be compared giving a fair result are 3Z:2 and 4Z:2 since half the FDPs in group 4Z:1 fractured during the preloading process and therefore can´t be compared to group 3Z:1. In group 4Z:1 the connector design is the most probable cause to the premature fractures.

Despite that uncertainty the statistics of this study supports the theory that regardless connector design increasing the length of the FDP span from three units to four units decreases the load bearing capacity approximately twice.

The design of the all-ceramic FDPs can be most crucial for the clinical survival. Therefore it is most important that both dentists and dental technicians are aware of the importance of using the correct settings of the CAD/CAM-system and optimizing the design in order to create a FDP that will fulfil the clinical demands.

Within the limitations of this *in-vitro* study it can be concluded that:

- The default settings of the two different CAD/CAM systems had a great impact on the fracture strength. It is important that a CAD/CAM system is equipped with possibilities to design a connector that fulfils the clinical demands of mechanical function and longevity.
- The design and the size of the radius of the gingival embrasures is crucial for the load-bearing capacity but also affects the fracture mode of the FDPs.
- Increasing the number of pontics from three to four decreases the load-bearing capacity nearly twice. Increasing the number of pontics even more would probably weaken the FDPs even more, but further research must be conducted to be able to predict the results.

Acknowledgment

The Y-TZP FDPs and the supporting test analogues were kindly provided by Nobel Biocare, Zurich, Switzerland.

The Panavia F 2.0 cement were kindly provided by RH Dental ApS, Gentofte, Denmark.

References

1. Bahat Z, Mahmood DJ, Vult von Steyern P. Fracture strength of three-unit fixed partial denture cores (Y-TZP) with different connector dimension and design. Swed Dent J 2009;33(3):149-59.
2. Chevalier J, Loh J, Gremillard L, Meille S, Adolfson E. Low-temperature degradation in zirconia with a porous surface. Acta Biomater 2011 Jul;7(7):2986-93.
3. Dittmer MP, Kohorst P, Borchers L, Stiesch M. Influence of the supporting structure on stress distribution in all-ceramic FPDs. Int J Prosthodont 2010 Jan-Feb;23(1):63-8.
4. Fahmy NZ. Influence of veneering materials on the marginal fit and fracture resistance of an alumina core system. J Prosthodont 2011 Jan;20(1):45-51.
5. F.F. Lange, GL. Dunlop, B.I. Davis. Degradation During Aging of Transformation-Toughened ZrO2-Y2O3 Materials at 250°C. J Am Ceram Soc 1986;69:237–40.
6. Filser F, Kocher P, Weibel F, Luthy H, Scharer P, Gauckler LJ. Reliability and strength of all-ceramic dental restorations fabricated by direct ceramic machining (DCM). Int J Comput Dent. 2001 Apr;4(2):89-106.
7. Fischer H, Weber M, Marx R. Lifetime prediction of all-ceramic bridges by computational methods. J Dent Res 2003 03;82(3):238-42.
8. Guazzato M, Quach L, Albakry M, Swain MV. Influence of surface and heat treatments on the flexural strength of Y-TZP dental ceramic. J Dent 2005 01;33(1):9-18.
9. Kamposiora P, Papavasiliou G, Bayne SC, Felton DA. Stress concentration in all-ceramic posterior fixed partial dentures. Quintessence Int 1996 10;27(10):701-6.
10. Kelly JR. Perspectives on strength. Dent Mater 1995 Mar;11(2):103-10.
11. Kelly JR, Tesk JA, Sorensen JA. Failure of all-ceramic fixed partial dentures in vitro and in vivo: analysis and modeling. J Dent Res 1995 06;74(6):1253-8.
12. Kern M, Fechtig T, Strub JR. Influence of water storage and thermal cycling on the fracture strength of all-porcelain, resin-bonded fixed partial dentures. J Prosthet Dent 1994 03;71(3):251-6.
13. Kheradmandan S, Koutayas SO, Bernhard M, Strub JR. Fracture strength of four different types of anterior 3-unit bridges after thermo-mechanical fatigue in the dual-axis chewing simulator. J Oral Rehabil 2001 Apr;28(4):361-9.
14. Koutayas SO, Kern M, Ferraresso F, Strub JR. Influence of framework design on fracture strength of mandibular anterior all-ceramic resin-bonded fixed partial dentures. Int J Prosthodont 2002 May-Jun;15(3):223-9.
15. Larsson C, Holm L, Lövgren N, Kokubo Y, Vult von Steyern P. Fracture strength of four-unit Y-TZP FPD cores designed with varying connector diameter. An in-vitro study. J Oral Rehabil 2007 09;34(9):702-9.
16. Lin J, Shinya A, Gomi H, Shinya A. Finite element analysis to compare stress distribution of connector of lithia disilicate-reinforced glass-ceramic and zirconia-based fixed partial denture. Odontology. 2012 Jan;100(1):96-9.
17. Luthardt RG, Holzhuter MS, Rudolph H, Herold V, Walter MH. CAD/CAM-machining effects on Y-TZP zirconia. Dent Mater 2004;20:655-662.
18. Luthardt RG, Holzhuter M, Sandkuhl O, Herold V, Schnapp JD, Kuhlisch E. Reliability and properties of ground Y-TZP-zirconia ceramics. J Dent Res 2002 Jul;81(7):487-91.
19. Luthy H, Filser F, Loeffel O, Schumacher M, Gauckler LJ, Hammerle CH. Strength and reliability of four-unit all-ceramic posterior bridges. Dent Mater 2005 Oct;21(10):930-7.
20. Mahmood DJ, Linderoth EH, Vult Von Steyern P. The influence of support properties and complexity on fracture strength and fracture mode of all-ceramic

fixed dental prostheses. Acta Odontol Scand. 2011 Jul;69(4):229-37.
21. Mollers K, Patzold W, Parkot D, Kirsten A, Guth JF, Edelhoff D. Influence of connector design and material composition and veneering on the stress distribution of all-ceramic fixed dental prostheses: a finite element study. Dent Mater 2011 Aug;27(8):e171-5.
22. Oh WS, Anusavice KJ. Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures. J Prosthet Dent 2002 May;87(5):536-42.
23. Oh W, Gotzen N, Anusavice KJ. Influence of connector design on fracture probability of ceramic fixed-partial dentures. J Dent Res 2002 09;81(9):623-7.
24. Onodera K, Sato T, Nomoto S, Miho O, Yotsuya M. Effect of connector design on fracture resistance of zirconia all-ceramic fixed partial dentures. Bull Tokyo Dent Coll 2011;52(2):61-7.
25. Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. Biomaterials 1999 Jan;20(1):1-25.
26. Plengsombut K, Brewer JD, Monaco EA,Jr, Davis EL. Effect of two connector designs on the fracture resistance of all-ceramic core materials for fixed dental prostheses. J Prosthet Dent 2009 Mar;101(3):166-73.
27. Pospiech P, Rammelsberg P, Goldhofer G, Gernet W. All-ceramic resin-bonded bridges. A 3-dimensional finite-element analysis study. Eur J Oral Sci 1996;104(4):390-5.
28. Pospiech P, Rammelsberg P, Unsold F. A new design for all-ceramic resin-bonded fixed partial dentures. Quintessence Int 1996 11;27(11):753-8.
29. Quinn GD, Studart AR, Hebert C, VerHoef JR, Arola D. Fatigue of zirconia and dental bridge geometry: Design implications. Dent Mater 2010 Dec;26(12):1133-6.
30. Raigrodski AJ, Chiche GJ. The safety and efficacy of anterior ceramic fixed partial dentures: A review of the literature. J Prosthet Dent 2001 Nov;86(5):520-5.
31. Richard H. J. Hannink, Patrick M. Kelly, Barry C. Muddle. Transformation Toughening in Zirconia-Containing Ceramics. Journal of the American Ceramic Society March 2000;83:461-87.
32. Richerson DW. Design approaches in modern ceramic engineering. Boco Raton, FL: Taylor & Francis Group, LLC; 2006.
33. Sundh A, Molin M, Sjogren G. Fracture resistance of yttrium oxide partially-stabilized zirconia all-ceramic bridges after veneering and mechanical fatigue testing. Dent Mater 2005: 21(5):476-82.
34. Vult von Steyern P. All-ceramic fixed partial dentures. Studies on aluminum oxide- and zirconium dioxide-based ceramic systems. Swed Dent J Suppl 2005;(173):1-69.
35. Vult von Steyern P, al-Ansari A, White K, Nilner K, Derand T. Fracture strength of In-Ceram all-ceramic bridges in relation to cervical shape and try-in procedure. An in-vitro study. Eur J Prosthodont Restor Dent 2000 Dec;8(4):153-8.
36. Vult von Steyern P, Carlson P, Nilner K. All-ceramic fixed partial dentures designed according to the DC-Zirkon technique. A 2-year clinical study. J Oral Rehabil 2005 03;32(3):180-7.
37. Vult von Steyern P, Jönsson O, Nilner K. Five-year evaluation of posterior all-ceramic three-unit (In-Ceram) FPDs. Int J Prosthodont 2001;14(4):379-84.
38. Vult von Steyern P, Kokubo Y, Nilner K. Use of abutment-teeth vs. dental implants to support all-ceramic fixed partial dentures: an in-vitro study on fracture strength. Swed Dent J 2005;29(2):53-60.
39. Zarone F, Russo S, Sorrentino R. From porcelain-fused-to-metal to zirconia: clinical and experimental considerations. Dent Mater 2011 Jan;27(1):83-96.
40. Zhang Y, Kim JW. Graded structures for damage resistant and aesthetic all-ceramic restorations. Dent Mater 2009 Jun;25(6):781-90.
41. Yoshimura M, Noma T, Kawabata K, Somiya S. Role of H2O on the degradation process of Y-TZP. Journal of Materials Science Letters 1987;6:465-7.

Corresponding author:
PhD Student Deyar Mahmood
Department of Prosthodontics
Faculty of Odontology
Malmö University
SE-205 06 Malmö, Sweden
E-mail: deyar.mahmood@mah.se