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Fracture Strength of Yttria-Stabilized Zirconium-Dioxide (Y-TZP) Fixed Dental Prostheses (FDPs) with Different Abutment Core Thicknesses and Connector Dimensions

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Keywords
Yttria-stabilized zirconium-dioxide; fixed dental prosthesis; fracture strength; connector dimension; abutment core thickness.

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Abstract

Purpose: The aim of this study was to investigate the fracture strength and fracture mode of yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) posterior three-unit FDPs with varying connector dimension and abutment core thickness.
Materials and Methods: Seventy 3-unit posterior FDP cores made of Y-TZP were divided into 7 groups with varying connector dimensions and abutment core thicknesses. All the FDPs underwent a simulated aging process including veneering, firing applications, thermocycling, and cyclic preloading. Finally the FDPs were subjected to load until fracture.
Results: Significant difference was seen between the different subgroups ($p < 0.05$). Groups with the same connector dimension showed no significant difference in fracture strength. All fractures of the specimens involved the connector.
Conclusions: Within the limitations of this in vitro study, it can be concluded that the strength of an all-ceramic Y-TZP FDP beam depends more on the connector dimension than on the thickness of the abutment core. Results indicate that the minimum abutment core thickness of an all-ceramic Y-TZP FDP might be reduced, compared to the recommended thickness, without reducing the strength of the reconstruction. This indication, however, needs to be verified by further studies before being considered generally applicable.

When the decision is made to use a fixed dental prosthesis (FDP) to replace one or more missing teeth, many parameters may influence the prognosis and the clinical performance. The location and size of the tooth gap, for example, will not only impose requirements on the dimensions of the FDP but will also influence the choice of material from which it is made.[1,2] Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) has been used as an alternative material to porcelain fused to metal (PFM) due to its biocompatibility and more favorable esthetics.[3] In addition, Y-TZP exhibits significantly higher flexural strength and toughness than earlier known dental ceramics.[4]

Y-TZP is derived from zirconium-dioxide (ZrO_2) but is considerably more stable, tougher, and stronger than the latter. Pure ZrO_2 exhibits crystal phase transformations in the temperature range between sintering and room temperature. Crystal phase transformations build up detrimental internal stresses within the material, thus making it unsuitable for construction purposes. By contrast, the phase transformations in Y-TZP are governed by yttria dopants in a way that makes the material tougher and stronger. If a crack occurs in the material, the surrounding tetragonal crystals turn into a monoclinic structure, resulting in a local volume increase in the crack tip area, preventing further crack propagation.[5-9] This property of preventing crack propagation has been claimed to make Y-TZP suitable for use in larger reconstructions.[8]

It has been shown that posterior Y-TZP FDPs have a survival rate of 74% to 100% over a 5-year period.[9-15] In addition to material properties, the dimensional design of an FDP is important for the clinical outcome. Dimensional requirements vary depending on the location of the missing teeth to be replaced and on the length of the span.[8] The connector area is a weak point of an all-ceramic FDP due to being the thinnest part of an irregularly shaped beam.[16] Therefore, studies have been conducted to determine recommendations for adequate connector dimensions and design.[4,17-22] Although there is no clear consensus as to what can be considered an adequate connector dimension, values of 2 to 5 mm have been suggested. It has been assumed that the higher the loads the construction will be exposed to, the greater the height of connector required. This is illustrated in the theory of deflection of a beam, where the height cubed is inversely proportional to the deflection.[23] The functional load on an FDP, however, is not always vertically oriented in the mouth. On the contrary, studies have shown that the principal direction of the loads during function often deviates from the long axis of the teeth, resulting in a stress pattern that often is more horizontally rather than vertically oriented.[24] When designing all-ceramic FDPs, it is also important to take into consideration that the posterior region, where the largest forces occur, does not always allow enough space for the desired height. This was demonstrated in a clinical study investigating connector dimensions of 115 FDPs. The mean available space (height) was 3.6 mm in the posterior and 4.4 mm in the anterior region, where inverse values would have been desirable.[25] Bahat et al have additionally shown that the tensile strength of an FDP can be further improved by designing the connector areas with a large gingival radius. It was concluded in their study that the tensile strength can be increased by 20%[17] by increasing the connector's gingival radius from 0.6 to 0.9 mm on a Y-TZP FDP.

There are, however, limited recommendations for the abutment core thickness of Y-TZP FDPs. Several manufacturers recommend preparations that provide space of at least 0.7/1.5 mm (core only/core + veneer) to allow sufficient thickness for the prosthetic material. The space obtained from the preparation is then allocated to both core and veneer materials. Hence, the recommended abutment core thickness of Y-TZP has been suggested based on the recommendations for PFM (i.e., > 0.5 mm)[12,26,27] and general guidelines for the all-ceramic crown core thickness of 0.4 to 0.7 mm. These values, however, have been developed empirically and have not been verified by clinical trials.[28] Reducing the core of a Y-TZP single crown from 0.5 to 0.3 mm decreased the fracture resistance by 35%.[29] In a prospective clinical study by Roediger et al, 99 FDPs with a core thickness of 0.4 mm were evaluated after 50 months. The results showed a core fracture at an atypical location for an FDP; namely, on the anterobuccal side of the abutment, and not through the usually fracture-prone connectors.[10,16] The explanation for this fracture, as the authors suggested, was probably due to a locally reduced core thickness.

There has not, however, been any investigation into the combined influence of both connector dimensions and abutment core dimensions on the fracture strength of Y-TZP FDPs. Hence, the aims of the present study were to investigate how the strength of 3-unit Y-TZP FDPs is influenced by varying abutment core thicknesses and connector dimensions, and which of the two parameters investigated have the highest impact on fracture load and mode; and to contribute to the development of recommendations regarding the minimum recommended core dimension for Y-TZP FDPs under the null hypothesis that core and connector dimension affects the strength of the FDP equivalent.

Materials and methods

Based on a master model in die stone (Fig 1), consisting of two end abutments representing the maxillary first premolar and first molar, 70 3-unit FDPs with one intermediate pontic were manufactured of Y-TZP. The abutments had a 120° chamfer preparation and a convergence angle of 15°. The master model was scanned once with a laboratory scanner (3Shape D700 Scan; 3Shape A/S, Copenhagen, Denmark) and, based on the data collected, 3-unit FDPs with varying dimensions were designed (3Shape CAD Design Software; 3Shape A/S). Standard default settings were used for all the FDPs. Seven .stl files were then sent to a production center for manufacture of 70 3-unit FDPs in Abradere Zirconia FDP material (Abradere Zirconia; Biomain AB, Helsingborg, Sweden). The specimens were divided into three main groups according to their abutment core thickness. The first group (the control group) had 10 specimens with an even abutment core thickness of 0.7 mm and a connector cross-sectional diameter of 3 mm. The remaining two groups consisted of FDPs with 0.5 and 0.3 mm abutment core thicknesses. Those two main groups were further divided into three subgroups depending on the connector's cross-sectional diameter: 3 mm × 3 mm, 3.5 mm × 3.5 mm, 4 mm × 4 mm. All subgroups contained 10 specimens each (Table 1).

All FDP cores were heat-treated in a porcelain furnace (Ivoclar P 500; Ivoclar Vivadent AG, Schaan, Lichtenstein) to simulate the firing cycles of the veneering porcelain (IPS e.max Ceram; Ivoclar Vivadent AG). The firing program implemented was ZirLiner (960°C), Wash (750°C), Dentin 1 (750°C), Dentin 2 (960°C), and Glaze (725°C), according to the manufacturer's recommendations. The FDPs then underwent thermocycling (LTC Multifunctional Thermocycler; LAM Technologies Electronic Equipment, Firenze, Italy) with the following program: 5000 cycles in two water baths tempered to 5°C and 55°C. Each cycle lasted 60 seconds: 20 seconds in each bath and two times 10 seconds to complete the transfer between the baths.

Individual resin abutments were produced in DuraLay® (Reliance Dental MFG Co., Worth, IL) for each FDP. This was done by reproducing the master model in die stone (Vel-Mix, Kerr, Romulus, MI) using an A-silicone impression (Flexi Time Correct Flow, Flexi Time Heavy Tray; Heraeus Kulzer GmbH, Hanau, Germany). A metal dowel was centered in each of the two abutments to stabilize the following wax-up of two root replicas. Subsequently, A-silicone impressions were made and used as a pattern for the production of 140 resin abutments, according to previous studies.[17,19,30,31] The FDPs were luted with Panavia F2.0 (Kuraray Medical Inc., Okayama, Japan) according to the manufacturer's recommendation. Cementation was performed with a 15 N load in the direction of insertion in 60 seconds. Individual support for the entire structure was made in the form of acrylic blocks. The FDP cores were fixated in these blocks with die stone (Vel-Mix).

The FDPs were thereafter preloaded with forces between 30 and 300 N at a frequency of 1 Hz in 10,000 cycles (Developed at the Faculty of Odontology, Department of Material Science and Technology, Malmö in collaboration with PAMAKO AB, V.Ingelsta, Malmö, Sweden). The force was applied with a stainless steel ball, 4 mm in diameter, placed distally on the occlusal surface of the pontic. Finally the FDPs were mounted in a testing jig at a 10° inclination and underwent load to fracture in a universal test machine (Instron 4465, Instron Co. Ltd., Norwood, MA). The crosshead speed was 0.255 mm/min, and the load was again applied with a stainless-steel ball, 4 mm in diameter, placed in the same position as during preloading. Preload and load until fracture were performed in distilled water. Fracture was defined as a visible fracture through the entire construction.

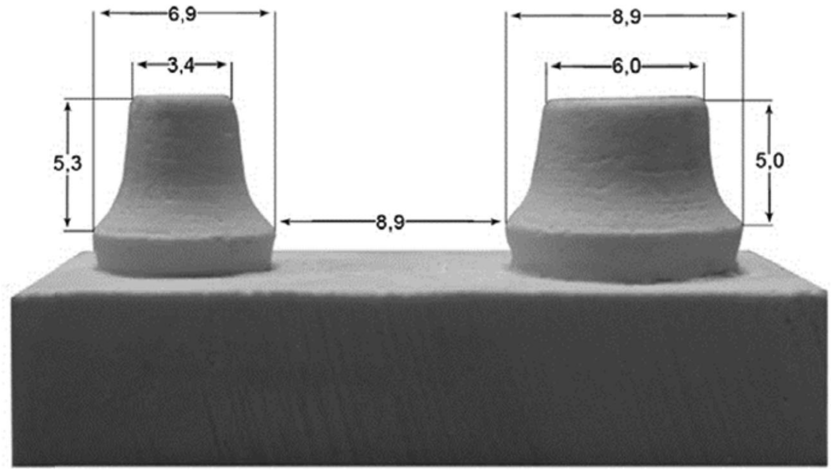

Figure 1 Master model and its measurements; 120° chamfer preparation and a convergence angle of 15°.

Table 1 Group classification

FDP group no	0.3/3	0.3/3.5	0.3/4	0.5/3	0.5/3.5	0.5/4	0.7/3
Core thickness (mm)	0.3	0.3	0.3	0.5	0.5	0.5	0.7
Height (mm)	3	3.5	4	3	3.5	4	3
Width (mm)	3	3.5	4	3	3.5	4	3

To minimize and allocate potential sources of error between the test groups, each step was carried out with one FDP from each group (i.e., seven FDPs per step). All FDPs were stored in a humid environment between trials. One-way ANOVA and Tukey's test were performed to detect any significant differences ($p < 0.05$) between the groups.

Results

Significant difference was seen between the three subgroups ($p < 0.05$). No significant difference was seen between the groups with the same connector dimensions (Table 2). All fractures of the specimens involved the connector. The main fracture pattern, F1, started from the mesial corner of the distal connector and propagated up through the pontic where the load was applied (Fig 2). Two other types of fractures were seen in groups 0.3/4 and 0.5/4. One type, F2, had vertical fractures in direct contact between the connector and distal abutment (Fig 2). The other type, representing only one specimen, showed a fracture lengthwise from one abutment to the other.

Table 2 Load at fracture (N) of FDP cores with varying core thickness and connector dimension

FDP core no	0.3/3[I,IV]	0.3/3.5[II,IV]	0.3/4[III,IV]	0.5/3[I,V]	0.5/3.5[II,V]	0.5/4[III,V]	0.7/3[I]
1	1045	1311	1391	1213	1268	1267*	831
2	950	1103	1654	881	1111	1653	830
3	1001	1300	1562	1095	1198	1899	1021
4	789	1323	1510	824	1342	1825	1036
5	1111	1076	1694	861	1328	1577	985
6	929	967	1803	901	1175	1674	1185
7	967	1179	1303	914	949	1864	701
8	1103	1313	1621	669	1047	1828	852
9	840	985	1570	1019	1344	1472	725
10	1034	1304	1735	1109	1138	1719	753
Mean	977	1186	1584	949	1190	1678	892
SD	105	143	153	160	133	198	158

*Lengthwise fracture from one abutment to the other.
[I,II,III]No significant difference between groups with the same connector dimension.
[IV,V]Significant difference between the three subgroups.

Discussion

The complexity of the oral cavity makes high demands on a material's mechanical properties. Esthetics plays an important part but must not affect the reconstructions' mechanical behavior during clinical function.[1,24,32] The material Y-TZP, however, exhibits esthetic improvements and also has acceptable mechanical properties for the manufacture of FDPs.[3,4]

An important part of the survival aspect is the need to create an optimal design for the restoration. With regard to the FDP, particular emphasis has been placed on the design of the connector; however, there is a lack of scientific basis for recommendations on abutment core thickness of Y-TZP FDPs, with only a few studies highlighting this.[12,26,27] To the authors' knowledge, studies illustrating the relationship between abutment core thickness and the connector design, and its influence on fracture strength, have yet to be published. A systematic review by Heintze and Rousson demonstrated less than 1% of Y-TZP FDP core fractures over a 3-year period.[33] Thus, it seems the material's ability to resist forces in the mouth can be considered to be acceptable.

In the present study, specimens were divided into three main groups depending on their abutment core thickness (Table 1). The control group was provided with an even abutment core

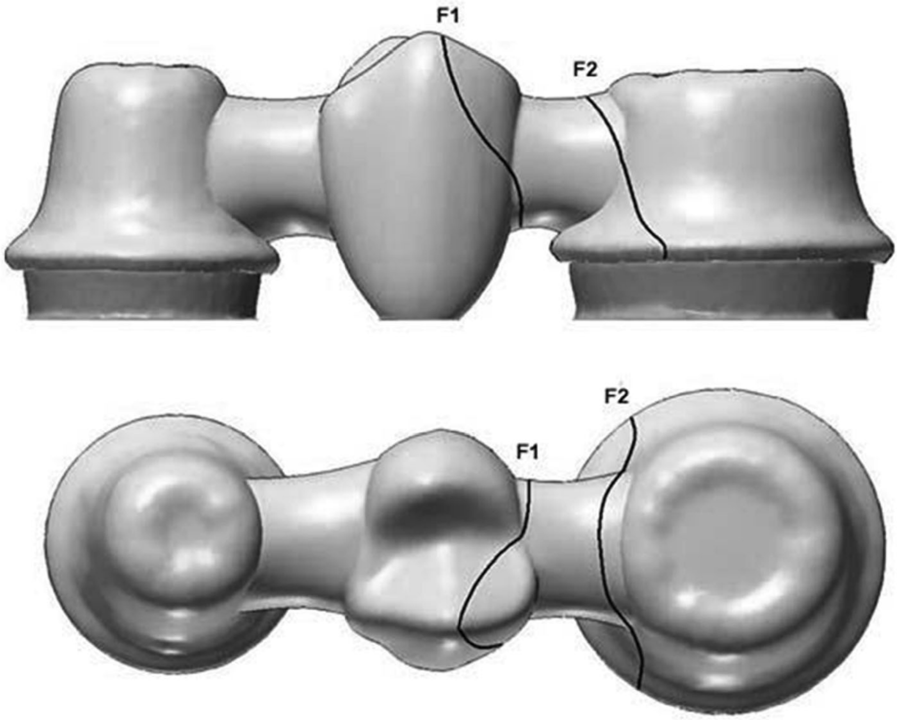

Figure 2 Fracture patterns.

thickness of 0.7 mm and a 3 mm connector diameter. This design is based on recommendations from previous studies and has been used in clinical studies of all-ceramic FDPs.[19,34] The 0.5 mm abutment core thickness was included because it is a minimum recommendation made by several manufacturers. The specimens belonging to the 0.3 mm group were chosen to analyze the fracture pattern when the abutment core thickness was below the recommendations mentioned above. Furthermore, according to manufacturers, 0.3 mm is considered to be the minimum producible thickness of pre-sintered Y-TZP.

The 0.3 and 0.5 mm groups were further divided into three subgroups (Table 1). In longer FDPs it has been shown that a diameter measuring <3 mm fractures easily during preloading.[19] Therefore, this study used a larger proven connector design. The selected connector dimensions (3 mm × 3 mm; 3.5 mm × 3.5 mm; 4 mm × 4 mm) were based on usage and recommendations from previous studies.[17,19,34,35] The purpose of this grouping was to analyze whether fractures continued to occur in the connector even if the abutment core thickness was reduced and to compare the results with studies mentioned above. A connector design of 3 mm × 3 mm has been suggested as suitable for a 3-unit anterior FDP; however, in this study, posterior constructions were investigated.[34] It is recommended that Y-TZP-based FDPs with long spans and molar replacements be designed with at least 4 × 4 mm connectors.[19,34] All specimens in this study were designed with a large gingival radius, as it has been shown to increase the fracture strength.[17,36]

To compare the various groups, they were designed to be as identical as possible and only varied in relation to the factors studied. It was possible to do this by using CAD/CAM technology, changing the specific parameters of a single template from default settings, thereby retaining the basic design of all the FDPs. The intention of this in vitro study was to mimic actual treatment procedures, the aging process, and the oral environment in an adequate manner. The choice of materials for the abutments has been shown to be important for the load-at-fracture value and the fracture mode. DuraLay® abutments were chosen, as it has been reported that they are appropriate for this type of study.[31] The firing program was performed according to the manufacturer's recommendation for veneering the porcelain (IPS e.max Ceram) to Y-TZP. Studies have shown that firing cycles may affect the mechanical properties of Y-TZP.[35,37–39] To include its possible mechanical changes, the materials were exposed to the standardized firing programs. Zirconia may be affected by low temperature degradation.[27] Therefore, a simulated oral environment was produced by thermocycling. It was possible, when transferring the specimens between the two baths that the difference in temperatures might lead to stresses within the material. Exposure to moisture in itself can also affect the material's properties by increasing slow-crack growth during loading.[5,6] Therefore, both preloading and load to fracture were performed in an aqueous environment. The number of cycles and time intervals for thermocycling and preloading is consistent with similar studies on the Y-TZP to allow for comparisons with those studies.[17,19,30,31] The FDPs were cemented with Panavia F2.0. This resin cement has been recommended for use with Zirconia.[40]

According to the results (Table 2), there was no statistically significant difference in fracture strength between an abutment core thickness of 0.3, 0.5, or 0.7 mm if the connector dimension was unchanged. A significant difference was seen, however, between the subgroups for each main group (i.e., the connector dimension is still the most crucial factor for the fracture strength).

The lowest load at fracture for a single specimen was measured at 669 N, a value below maximum bite force in the posterior region; however, similarity to maximum bite force is seen in the group with the lowest mean value of 892 N.[41] This indicates that all the dimensions investigated would be acceptable for posterior 3-unit FDPs.

The fracture mode showed that a majority of the fractures started from the mesial corner of the distal connector and propagated up through the pontic towards the loading site (Fig 2: F1). In addition to the fracture described above, groups 0.3/4 and 0.5/4 also showed two other types of fracture patterns with high fracture strength values. The type that showed vertical fractures in direct contact between the connector and distal abutment might possibly be due to the well-extended connector design for these specific groups. It is conceivable that this fracture pattern was due to the increased connector diameter, which gives the construction an increased fracture resistance and moves the weakest point away from the typical break point through the connector (Fig 2: F2). The third type showed a fracture lengthwise from one abutment to the other. Only one specimen showed this type of fracture, at a slightly lower fracture strength value (Table 2: 05/4, FDP core no 1). The probable cause in this case is that the steel ball was placed more mesially than the intended placement, which was distal to the occlusal surface to the pontic. The force then caused a two-point load between the buccal and palatal cusps of the pontic. This might have given rise to a more pronounced "wedge-effect" that better explains the fracture pattern.

According to this study, it appears that, despite the reduced thickness of the material, fracture will still occur in the connector with acceptable values for a 3-unit FDP. Thus, it might be possible to decrease the abutment core thickness. The advantages of a thinner core are the possibility of preserving tooth structure, which may lead in turn to a reduced risk of pulpal complications. Several studies have shown the relationship between the preparation depth and the risk of adverse pulpal reaction.[42–45] Another aspect relates to esthetics, since a thinner core is believed to provide increased translucency and allows more space for the veneer porcelain, which leads to improved options in terms of esthetics.[46] These two aspects are of minor importance since a decrease in preparation depth of 0.2 mm is not only hard to achieve, but also difficult to control. The oral anatomy may limit the design and does not always meet the design recommendations for the material.[25] According to the present in vitro study, it seems that a core thickness of 0.3 mm is acceptable for a 3-unit FDP in Y-TZP, as long as the connector is designed according to given recommendations. It is possible that a thinner abutment core thickness might give equivalent results.

The greatest abutment core and connector thickness showed the highest fracture strength, but with varying fracture patterns including both the connector and the abutment. The results also showed that the abutment core thickness was less important for the fracture strength and that strength is primarily determined by connector design, i.e., the connector dimensions have greater significance than the abutment core thickness. At a minimum conceivable dimension of the core, it is probable that fractures will occur solely at the abutment, provided there is a proper connector design; however, this was not demonstrated in this study, as this type of fracture only occurred with the largest connector. Therefore, further studies with reduced dimensions need to be conducted to be able to recommend this minimum abutment core thickness. It appears that it is possible to reduce the core thickness further than given recommendations; however, one should bear in mind whether this brings any clinical benefit, since it is questionable whether it is clinically achievable.

With regard to the connector dimensions, this study shows that a diameter of 3 mm might be acceptable for a posterior 3-unit FDP, and a diameter exceeding 3.5 mm is not clinically necessary. With regard to abutment core thickness, it appears that 0.3 mm is acceptable for a 3-unit FDP in Y-TZP, as long as the connector is designed according to given recommendations.

This study does not illustrate how a direct load applied to the abutment affects the core. It is possible that the stress pattern caused by such loads may lead to more fractures of the abutment core starting from its cementation surface (inside the crown). More studies are needed before any recommendations for reducing the abutment core thickness can be made for general clinical use.

Conclusions

Within the limitations of this in vitro study, it could be concluded that:

1. The strength of an all-ceramic (Y-TZP) FDP frame is more dependent on the connector dimension than on the thickness of the abutment core.
2. The minimum abutment core thickness of an all-ceramic (Y-TZP) FDP can be reduced, compared to the recommended thickness, without reducing any of the reconstruction's strength. This indication, however, needs to be verified by further studies before being considered generally applicable.

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References

1. Lundgren D, Laurell L: Biomechanical aspects of fixed bridgework supported by natural teeth and endosseous implants. Periodontol 2000 1994;4:23-40
2. Anusavice KJ: Standardizing failure, success, and survival decisions in clinical studies of ceramic and metal-ceramic fixed dental prostheses. Dent Mater 2012;28:102-111
3. Bachhav VC, Aras MA: Zirconia-based fixed partial dentures: a clinical review. Quintessence Int 2011;42:173-182
4. Filser F, Kocher P, Weibel F, et al: Reliability and strength of all-ceramic dental restorations fabricated by direct ceramic machining (DCM). Int J Comput Dent 2001;4:89-106

5. Manicone PF, Rossi Iommetti P, Raffaelli L: An overview of zirconia ceramics: basic properties and clinical applications. J Dent 2007;35:819-826
6. Piconi C, Maccauro G: Zirconia as a ceramic biomaterial. Biomaterials 1999;20:1-25
7. Casellas D: Fracture toughness and mechanical strength of Y-TZP/PSZ ceramics. Scr Mater 2001;45:213-220
8. Plengsombut K, Brewer JD, Monaco EA, et al: Effect of two connector designs on the fracture resistance of all-ceramic core materials for fixed dental prostheses. J Prosthet Dent 2009;101:166-173
9. Al-Amleh B, Lyons K, Swain M: Clinical trials in zirconia: a systematic review. J Oral Rehabil 2010;37:641-652
10. Roediger M, Gersdorff N, Huels A, et al: Prospective evaluation of zirconia posterior fixed partial dentures: four-year clinical results. Int J Prosthodont 2010;23:141-148
11. Eschbach S, Wolfart S, Bohlsen F, et al: Clinical evaluation of all-ceramic posterior three-unit FDPs made of In-Ceram Zirconia. Int J Prosthodont 2009;22:490-492
12. Sailer I, Feher A, Filser F, et al: Prospective clinical study of zirconia posterior fixed partial dentures: 3-year follow-up. Quintessence Int 2006;37:685-693
13. Schmitt J, Holst S, Wichmann M, et al: Zirconia posterior fixed partial dentures: a prospective clinical 3-year follow-up. Int J Prosthodont 2009;22:597-603
14. 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;32:180-187
15. Sailer I, Gottnerb J, Kanelb S, et al: Randomized controlled clinical trial of zirconia-ceramic and metal-ceramic posterior fixed dental prostheses: a 3-year follow-up. Int J Prosthodont 2009;22:553-560
16. Motta AB, Pereira LC, da Cunha AR, et al: The influence of the loading mode on the stress distribution on the connector region of metal-ceramic and all-ceramic fixed partial denture. Artif Organs 2008;32:283-291
17. 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;33:149-159
18. Conrad H, Seong W, Pesun I: Current ceramic materials and systems with clinical recommendations: a systematic review. J Prosthet Dent 2007;98:389-404
19. Larsson C, Holm L, Lovgren N, et al: Fracture strength of four-unit Y-TZP FPD cores designed with varying connector diameter. An in-vitro study. J Oral Rehabil 2007;34:702-709
20. Luthy H, Filser F, Loeffel O, et al: Strength and reliability of four-unit all-ceramic posterior bridges. Dent Mater 2005;21:930-937
21. Pospiech P, Rammelsberg P, Goldhofer G, et al: All-ceramic resin-bonded bridges. A 3-dimensional finite-element analysis study. Eur J Oral Sci 1996;104:390-395
22. Raigrodski AJ: Contemporary materials and technologies for all-ceramic fixed partial dentures: a review of the literature. J Prosthet Dent 2004;92:557-562
23. Dupont R: Large ceramo-metallic restorations. Int Dent J 1968;18:288-308
24. Randow K: On the functional deformation of extensive fixed partial dentures. An experimental clinical and epidemiological study. Swed Dent J Suppl 1986;34:1-83
25. Johansson M, Mosharraf S, Karlsson S, et al: A dental laboratory study of the dimensions of metal frameworks for fixed partial dentures. Eur J Prosthodont Restor Dent 2000;8:75-78
26. Beuer F, Stimmelmayr M, Gernet W, et al: Prospective study of zirconia-based restorations: 3-year clinical results. Quintessence Int 2010;41:631-637
27. Rekow ED, Silva NR, Coelho PG, et al: Performance of dental ceramics: challenges for improvements. J Dent Res 2011;90:937-952
28. Marchack BW, Sato S, Marchack CB, et al: Complete and partial contour zirconia designs for crowns and fixed dental prostheses: a clinical report. J Prosthet Dent 2011;106:145-152
29. Reich S, Petschelt A, Lohbauer U: The effect of finish line preparation and layer thickness on the failure load and fractography of ZrO_2 copings. J Prosthet Dent 2008;99:369-376
30. Vult von Steyern P, Ebbesson S, Holmgren J, et al: Fracture strength of two oxide ceramic crown systems after cyclic pre-loading and thermocycling. J Oral Rehabil 2006;33:682-689
31. 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;69:229-237
32. Albashaireh ZS, Alhusein AA, Marashdeh MM: Clinical assessments and patient evaluations of the esthetic quality of maxillary anterior restorations. Int J Prosthodont 2009;22:65-71
33. Heintze SD, Rousson V: Survival of zirconia- and metal-supported fixed dental prostheses: a systematic review. Int J Prosthodont 2010;23:493-502
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. 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;21:476-482
36. Oh WS, Anusavice KJ: Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures. J Prosthet Dent 2002;87:536-542
37. Sundh A, Sjogren G: Fracture resistance of all-ceramic zirconia bridges with differing phase stabilizers and quality of sintering. Dent Mater 2006;22:778-784
38. Guazzato M, Albakry M, Quach L, et al: Influence of surface and heat treatments on the flexural strength of a glass-infiltrated alumina/zirconia-reinforced dental ceramic. Dent Mater 2005;21:454-463
39. Guazzato M, Quach L, Albakry M, et al: Influence of surface and heat treatments on the flexural strength of Y-TZP dental ceramic. J Dent 2005;33:9-18
40. Kim MJ, Kim YK, Kim KH, et al: Shear bond strengths of various luting cements to zirconia ceramic: surface chemical aspects. J Dent 2011;39:795-803
41. Waltimo A, Kononen M: Maximal bite force and its association with signs and symptoms of craniomandibular disorders in young Finnish non-patients. Acta Odontol Scand 1995;53:254-258
42. Cheung GS, Lai SC, Ng RP: Fate of vital pulps beneath a metal-ceramic crown or a bridge retainer. Int Endod J 2005;38:521-530
43. Jackson C, Skidmore A, Rice R: Pulpal evaluation of teeth restored with fixed prostheses. J Prosthet Dent 1992;67:323-325
44. Lockard MW: A retrospective study of pulpal response in vital adult teeth prepared for complete coverage restorations at ultrahigh speed using only air coolant. J Prosthet Dent 2002;88:473-478
45. Stanley HR: Pulpal response to dental techniques and materials. Dent Clin North Am 1971;15:115-126
46. Baldissara P, Llukacej A, Ciocca L, et al: Translucency of zirconia copings made with different CAD/CAM systems. J Prosthet Dent 2010;104:6-12