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The influence of support properties and complexity on fracture strength and fracture mode of all-ceramic fixed dental prostheses

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Abstract

Objective. When a new material is released, clinical studies are indicated. For the clinical studies to be defensible, *in-vitro* studies, which are as clinically relevant as possible, must be performed. The aim of this study was to investigate how the choice of material used for supporting tooth analogues and support complexity influences test results concerning the fracture strength of fixed dental prostheses (FDPs) made from a brittle material: Y-TZP. ***Material and methods.*** Twenty-four FDPs were produced in Y-TZP. The FDP cores were subjected to heat treatment to simulate veneering and then thermocycled for 5000 cycles to simulate ageing. The FDPs were divided into three groups and were cemented on tooth-supporting analogues made from aluminium, polymer and DuraLay®. The FDPs were preloaded for 10,000 cycles and finally loaded to fracture. ***Results.*** There were no significant differences in load to fracture or fracture mode between the groups cemented on polymer and DuraLay tooth analogues. The FDPs cemented on aluminium tooth analogues showed a significantly higher load at fracture and a different fracture mode. ***Conclusions.*** Within the limitations of this *in-vitro* study, the following could be concluded. To achieve mutually comparable results there is a need for a standardized, simple test set-up for *in-vitro* testing of all-ceramic FDPs intended for cementation upon natural teeth. Resilient, non-complex and resilient, complex tooth analogues give comparable test results when the test set-up is unchanged in all other aspects. Non-resilient (with an elastic modulus equivalent to or higher than that of aluminium) tooth analogues give high and unrealistic load-at-fracture values together with adverse fracture modes compared to FDPs failing in clinical situations.

Key Words: *All-ceramic, dental porcelain, fixed dental prostheses, Y-TZP, zirconia*

Introduction

When a new material is released, clinical studies are indicated. For those clinical studies to be defensible, *in-vitro* studies, which are as clinically relevant as possible, must be performed. The purpose of *in-vitro* tests is often to determine the specific properties of a material and the tests can be performed in many different ways [1]. Properties such as tensile strength and reliability are some important considerations for the structural design of a dental restoration. Reliability for mechanical applications is generally assessed by a comparison of the design requirements with the material properties, including the safety margin. The concepts vary depending on which material group is to be subjected to testing. When *in-vitro* testing a metal, the properties might be adequately described by means of the mean and standard deviation of its strength. However, when testing ceramic materials, the test methods must reflect the range and distribution of strength, with consideration being paid to the brittle nature of the material [2,3]. Hence, different test methods are used depending on the material to be tested, taking into account the specific and different properties of each material. The tensile strength of a material can be measured with uniaxial tensile strength tests, three- or four-point bending tests or biaxial bending tests, which include the ring-on-ring, ball-on-ring and piston-on-three-balls tests [3–5]. Fracture toughness can be measured with various modifications of indentation tests [6,7]. Tests of dental materials are performed with standardized stylized beams, discs or norm crowns [5,8–10]. Such methods are all based on highly standardized test specimens but, when the materials are to be used for dental reconstructions, it is often necessary to

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evaluate not only the material but also the design, since the overall strength is highly dependent on the geometry of the reconstruction [11]. In the clinical situation, crowns and bridges are supported by a combination of different structures with differing properties, such as the elastic modulus. In relation to the framework for, and the design of, a fixed dental prosthesis (FDP), the complexity of the supporting system in the oral cavity creates stress patterns that must be considered in the *in-vitro* situation [12,13]. Clinical factors such as supporting bone, periodontal ligament, abutment teeth with varying root anatomy and dentine cores with different shapes and qualities are factors that should be mimicked as far as possible in *in-vitro* studies [14]. Hence it is desirable to test dental restorations with respect to both materials and design approaches. In addition, environmental factors such as moisture, repeated loads, support and temperature changes must also be considered, particularly with regard to a complex construction like an FDP [3,10].

To perform an *in-vitro* test, tooth analogues may be used as abutments. The analogues can be made from various prefabricated or cast metals or different kinds of resins [13–20]. Extracted human or bovine teeth may also be used [21–24].

Abutments made of one of many different metal alloys, with or without artificial periodontal membranes, have frequently been used. One advantage claimed for this use is that the residual deformation during loading is limited compared to when using other, less stiff, materials. Another advantage is that the abutments can be reused, thus reducing costs. The artificial periodontal membrane is claimed to produce a load situation close to that found in the oral cavity, simulating the physiological mobility of the natural tooth [13,15,17–19].

Another group frequently used is resin-based materials, such as epoxy resins. Resins, apart from being inexpensive, have mechanical properties very similar to those of natural tooth, giving support close to that provided by nature [12,25].

Extracted natural teeth can be used to simulate the oral conditions as closely as possible with regard to modulus of elasticity, bonding characteristics, thermal conductivity and strength [20–23,26–28]. The primary disadvantages of using natural teeth are that they are non-uniform in condition and shape [29]. Accessibility can be a problem and sometimes there is a need for ethical approval. Examples of variations in test set-up are given in Table I. Hence, combinations of the different materials mentioned above may be used and the design may vary, which might influence the result. If a rigid support is used that hinders the movement of the abutment teeth, the load-bearing capacity of the FDPs tested will probably increase and consequently affect the result, giving unrealistic fracture data [20,30]. By comparison, a study design that takes the supporting tissues, dentine cores, periodontal ligament and cortical bone into consideration might result in decreased load-at-fracture values [13,22].

Aim

The aim of this study was to investigate how the choice of material used for supporting tooth analogues and support complexity influences test results relating to the fracture strength of FDPs made from the brittle material Y-TZP.

Material and methods

A total of 24 three-unit, anterior, all-ceramic, Y-TZP FDPs, with one intermediate pontic, supported by end abutments were made.

Preparation

A plastic model of an upper jaw (KaVo YZ, OK VZ 623 0401 180; KaVo Dental, GmbH, Biberach, Germany) was used. Two abutment preparations were made on the left central incisor and left canine. The aim was to design a preparation with a 120° chamfer and an angle of convergence of 15°. The left lateral incisor was removed. Subsequently, a full arch A-silicone (Flexitime Mono Phase; Heraeus Kulzer, GmbH, Hanau, Germany) impression was taken and poured with die stone (Everest® Rock, Type 4 die stone; KaVo Dental) to produce a master cast (Figure 1).

The master cast was scanned once with a mechanical scanner (Procera® Forte; Nobel Biocare, Zürich, Switzerland) and the data were transferred to a computer equipped with computer-aided design (CAD) software (Procera CAD Design C3D, version 2.00) where the intended design of the FDP was established. The connector dimensions were set to 3 mm × 3 mm and the minimum thickness of the core was set to 0.7 mm. The radius of the gingival and occlusal embrasures in the connector areas was 0.6 mm according to the default settings in the CAD program and according to the manufacturer's recommendations. The CAD data for the FDPs were subsequently sent to a production centre (Procera production centre, Stockholm, Sweden), where they were used for production of the 24 FDPs in Procera Zirconia bridge material using 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

Table I. Mean force fracture strength of the three-unit all-ceramic FDPs.

Reference	Core material	Mean force at fracture (N)	Tooth-supporting analogues	Connector dimensions (mm)
Tinchert et al., 2001	IPS Empress	490	Wiron 99, Bego	4 (height)
Kheradmandan et al., 2001	IPS Empress II	293	Human[a]	4 × 4
Oh et al., 2002	Lithium disilicate-based glass-ceramic	673[g]; 944[f]	Epoxy resin	4 × 5
Chitmongkolsuk et al., 2002	IPS Empress II	950	Human[a]	4 × 4
Tinchert et al., 2001	IPS Empress II	1350	Wiron 99, Bego	4 (height)
Kheradmandan et al., 2001	In-Ceram Alumina	240	Human[a]	4 × 4
Vult von Steyern et al., 2000	Vita In-Ceram	510[b]; 606[c]	DuraLay[a]	3 (diameter)
Beuer et al., 2008	In-Ceram Alumina	659	Co–Cr[a]	12 mm^2
Tinchert et al., 2001	In-Ceram Alumina	950	Wiron 99, Bego	4 (height)
Vult von Steyern et al., 2005	Procera Alumina	378	Duralay[a]	3 (diameter)
Vult von Steyern et al., 2005	Procera Alumina	604	Titanium[a] abutment	3 (diameter)
Beuer et al., 2008	In-Ceram Zirconia	770	Co–Cr[a]	12 mm^2
Beuer et al., 2008	In-Ceram YZ	1042	Co–Cr[a]	12 mm^2
Tinchert et al., 2001	In-Ceram Zirconia	1650	Wiron 99, Bego	4 (height)
Sundh et al., 2006	Vita YZ	1100	Steel	3 × 3
Att et al., 2007	Procera Zirconia	1396	Human[a]	3 × 3
Att et al., 2007	Vita YZ-Cerec	1630	Human[a]	3 × 3
Sundh et al., 2006	Denzir-M	1300	Steel	3 × 3
Sundh et al., 2005	Denzir	1611[e]	Steel	3 × 3
Att et al., 2007	DC-Zircon	1823	Human[a]	3 × 3
Tinchert et al., 2001	DC-Zircon	2250	Wiron 99, Bego	4 (height)
Sundh et al., 2005	Denzir	3480[d]	Steel	3 × 3
Bahat et al., 2009	Procera Zirconia	651[b,h]; 831[b,i]	Duralay[a]	3 × 3; 3 × 3

[a]Design set-up mimicking periodontal ligament.
[b]Chamfer preparation.
[c]Shoulder preparation.
[d]Framework as delivered after machining was not subjected to dynamic loading in water.
[e]Frameworks heat-treated in a way similar to veneering with Vita D and subjected to dynamic loading in water.
[f]Occlusal radius 0.25 mm; gingival radius 0.90 mm.
[g]Occlusal and gingival radii 0.25 mm.
[h]Occlusal and gingival radii 0.6 mm.
[i]Occlusal and gingival radii 0.9 mm.

firing cycles of the veneering porcelain (Nobel Rondo™ Zirconia; Nobel Biocare) recommended for the core material used. Four firing programmes were used: Liner (930°C); Dentin 1 (910°C); Dentin 2 (900°C); and Glaze (890°C), as shown in Table II, all according to the manufacturer's instructions.

Ageing procedures

Two stages of ageing were performed. In the first stage, all FDP cores underwent thermocycling (LTC Multifunctional Thermocycler; LAM Technologies Electronic Equipment, Ferenze, Italy) using a small basket controlled by a device driver. All FDP cores underwent 5000 thermocycles in two water baths at temperatures of 5°C and 55°C. The cores were placed in a basket used to cycle them between the two baths. Each cycle lasted 60 s: 20 s in each bath and 10 s to complete the transfer between the baths. After thermocycling, the cores were dried and then randomly divided into three groups.

Supporting tooth analogues

Three types of support were made: two inspection blocks to be milled using the same CAD file as that used to produce the FDP cores; and one complex support to be made using a technique described in previous studies [10,11]. During the previous production step at the production centre (Procera production centre), inspection blocks were made to enable precision checking. Such inspection blocks can be made from aluminium or a polymer material (both at Procera production centre). Eight inspection models were ordered from each material to support the FDPs in Groups A and P (Figure 2).

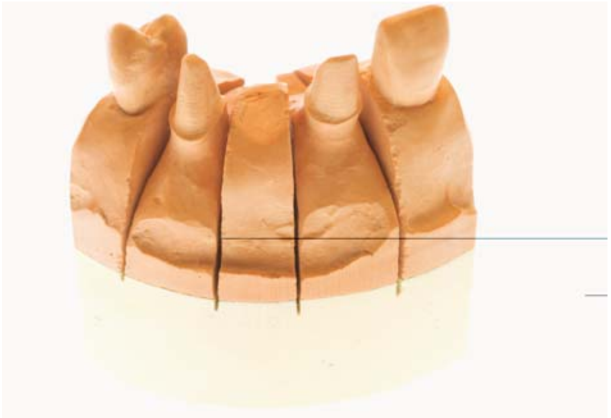

Figure 1. Master cast.

For the first group (Group A), eight inspection blocks (Procera production centre) made from aluminium were used as supporting tooth analogues.

For the second group (Group P), eight inspection blocks (Procera production centre) made from a polymer material were used as supporting tooth analogues.

The fabrication of supporting tooth analogues for the third group (Group D) was done by reproducing the abutments of the master cast, first and third incisor. They were reproduced using A-silicon impression material (President; Coltene AG, Altstätten, Switzerland), to make two tooth-like end abutments on which the FDPs could be cemented and fixed in acrylic blocks according to a previous study [16]. Impressions were taken of the central incisor and canine abutments using A-silicon impression material (President), and then poured in die stone (Everest Rock, Type 4 die stone; KaVo Dental) with a metal dowel pin centred in each abutment to stabilize the following build-up of a wax-up. Grooves were made on the dowel pins to facilitate retention of the wax.

The above-mentioned reproduction abutments were used only to create the shape for the final test model and were not used in the test. They were copied in a second step using an A-silicone impression (President) and subsequently poured with inlay pattern resin (DuraLay®; Reliance Dental MFG Co., Worth, IL), thus creating the final model for the experiment.

A total of 16 tooth analogues, eight incisors and eight canines, were made.

Cementation

Prior to cementation, the inspection blocks (Procera production centre) made from aluminium in the first group (Group A) were steam-cleaned and then treated with metal adhesive primer (Alloy primer; Kuraray Medical Inc., Okayama, Japan), which was applied to the cementation surfaces according to the manufacturer's instructions (Figure 3).

The inspection blocks (Procera production centre) made from a polymer in the second group (Group P) were steam-cleaned and subsequently treated with ED primer II A and B (Kuraray Medical Inc.), which was applied to the cementation surfaces according to the manufacturer's instructions (Figure 4).

The DuraLay tooth-supporting analogues in the third group (Group D) were cleaned and treated with ED primer II A and B (Kuraray Medical Inc.), which was applied to the cementation surfaces according to the manufacturer's instructions (Figure 5).

The FDPs of all three groups were luted onto the reproduction abutments with Panavia F 2.0 luting cement (Kuraray Medical Inc.) using both light and Oxyguard II (Kuraray Medical Inc.), according to the manufacturer's recommendations. 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 s. The cemented FDPs in the third group (Group D) were fixed in holes in acrylic blocks with die stone (Vell-mix; Kerr, Romulus, MI) subsequent to cementation. Before the start of mechanical testing, the FDP cores were placed in a plastic container, with water covering the bottom surface and a sealing lid, which created 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. The force was

Table II. The general firing programme (NobelRondo Zirconia Ceramics). Extended cooling was not employed.

	Pre-heating temperature (°C)	Pre-heating drying time (min)	Heating rate	Firing temperature (°C)	Holding time (min)	Vacuum (hPa)
Liner firing	575	8	55	930	1	50
Dentin firing 1	575	9	55	910	1	50
Dentin firing 2	575	8	55	900	1	50
Glaze firing	575	5	55	880	1	–

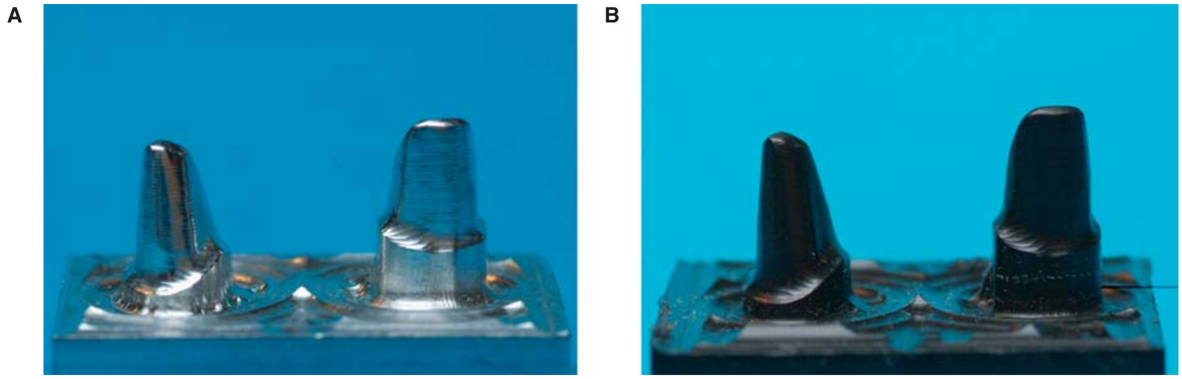

Figure 2. (A) Inspection block milled from aluminium. (B) Inspection block milled from polymer material.

applied with a stainless-steel intender, 2.5 mm in diameter, placed centrally on the incisal edge of the lateral incisor pontic 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). The cross-head speed was 0.255 mm/min and the load was applied with a stainless-steel intender, again placed as described above. Throughout the test period, whenever the FDP cores were not being actively tested they were stored in water. 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. Any difference between groups was tested by means of Student's *t*-test. Differences in fracture modes were calculated using Fisher's exact probability test.

Results

The fracture data are listed in Table III. All the FDPs fractured in the connector area, which corresponds to previous studies of all-ceramic FDPs [11,12,14,18–23,30–40].

The fracture modes differed significantly ($P < 0.01$) between Group A and the two other groups and were distributed as follows:

- In Group A, 75% of the FDPs fractured through both connectors and 25% fractured through one connector only.
- All the FDPs in Group P fractured through one connector only (six of the FDPs through the mesial part of the connector and through the pontic and two in the same manner but through the distal part of the connector and the pontic).
- All the FDPs in Group D fractured through the distal part of the connector and through the pontic.

All the crack propagation that led to fracture started at the gingival side of the connector.

None of the abutments failed during testing.

The fracture loads were significantly higher in Group A compared to Groups P ($P < 0.001$) and

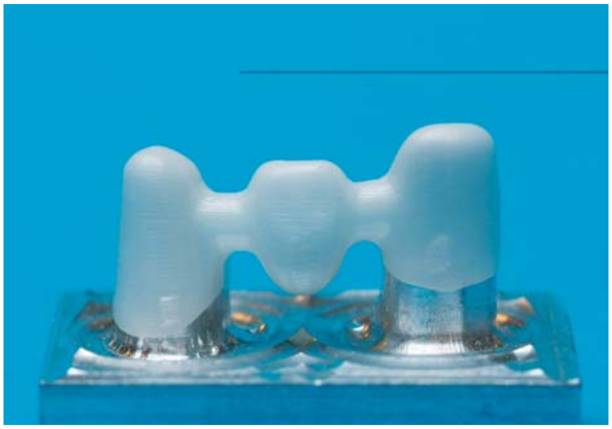

Figure 3. Y-TZP FDP cemented on aluminium tooth-supporting analogue.

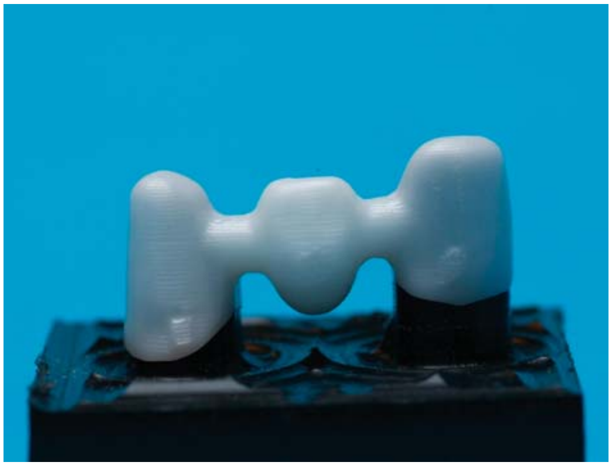

Figure 4. Y-TZP FDP cemented on acrylic tooth-supporting analogue.

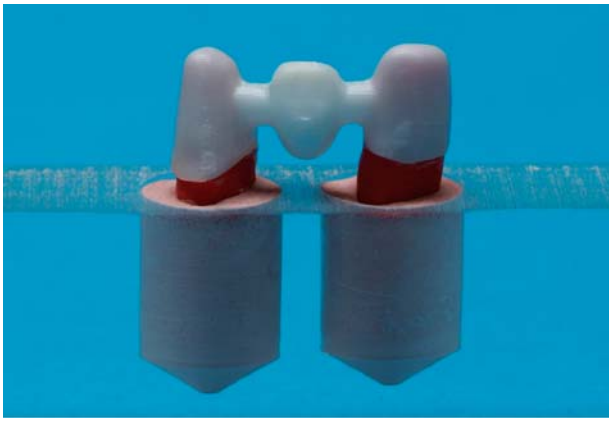

Figure 5. Y-TZP FDP cemented on DuraLay tooth-supporting analogue.

D ($P < 0.001$). There was no significant difference in fracture loads between Groups P and D.

Discussion

To the authors' knowledge, there have been no previous comparisons of different materials for tooth analogues made in the same study. However, the results from other studies vary, clearly showing the difficulties in comparing results from studies where the test set-up varies.

The present study shows that the choice of material for the tooth analogues is crucial for the fracture strength of all-ceramic reconstructions during testing. There was no significant difference in fracture strength between the groups with tooth analogues made from DuraLay and polymer, whereas the stiff tooth analogues made from aluminium showed a doubled fracture load value compared to Groups P and D.

The tooth analogues were made in two completely different ways, affecting both precision and complexity. The analogues made from aluminium and polymer were machine-milled at the same production centre and with support from the same data file as was used for the design and milling of the FDPs, which should indicate ultimate fit and precision. The complex processing of the tooth analogues made from DuraLay probably means, however, that every step in the processing chain leads to deviation from ultimate fit and precision. Despite the presumed differences in precision between the machine-milled and hand-made tooth analogues, there was no significant difference in the results.

The machine-milled tooth analogues were of simple construction, being made in one piece of the same material. The tooth analogues made from DuraLay were hand-made from three different components to mimic clinical factors assumed to be relevant, resulting in a more complex construction [1,11]. Despite the variation in complexity between Groups P and D, the result was not affected.

Regardless of the precision and complexity of the analogues, the results show significant differences between stiff and resilient tooth analogues. Previous studies have shown that the choice of material for the tooth analogues in a study like the present one could have a significant and consequential influence on the result. Hence, rigid tooth analogues and rigid supports which hinder the movement of the abutment teeth will increase the load-bearing capacity of the FDPs tested and consequently affect the result, giving unrealistic fracture data [20,30]. By comparison, a study design that took the supporting tissues, dentine cores, periodontal ligament and cortical bone into consideration resulted in decreased load-at-fracture values [13,22].

To determine whether other studies have come to the same conclusions as the present study, i.e. that there is an obvious correlation between the load at fracture and the choice of material for the tooth analogues, comparisons were made with other studies testing three-unit, all-ceramic FDPs [12–14,18–22,41,42] (Table I). These comparisons showed that there were no obvious trends or connections between the choice of material for the tooth analogues and the resulting load at fracture.

There may be other factors affecting the fracture load values in a study. The dimensions of the FDPs tested vary depending on whether the FDPs are anteriorly or posteriorly placed. There may also be differences in the length, width and height of the beam [8].

The height and cervical shape of the preparation may affect the result, as can the amount of root of tooth analogue mounted in the test-block, if there is one [14]. A short, wide preparation mounted with a large part of its root supported in the test-block will probably give a higher load-at-fracture value, while a tall, narrow preparation with a smaller part of the root

Table III. Load at fracture (N) for the three-unit all-ceramic FDPs in Groups A, P and D.

	Group		
FDP core no.	A	P	D
1	1562	753	700
2	1704	749	568
3	1728	849	675
4	2103	735	647
5	1646	767	724
6	2189	830	728
7	1964	610	829
8	1641	575	875
Mean	1817	733.5	718
SD	235	96	98

supported in the test-block will presumably give a lower value.

Previous studies show that the gingival and occlusal curvature of the connector have a major impact on the FDP fracture strength. The greater the radius of curvature achieved, and the more perfectly arched the connector is between each unit, the higher the total fracture strength of the FDP. The dimensions of the gingival and occlusal curvature are often determined by either the settings of the CAD program, the limits of milling in the CAM system or both [11,12,33,35]. Furthermore, every arrangement in the test set-up before load to fracture that is intended to simulate a clinical situation may influence the result.

According to some studies, the temperatures that the core is subjected to during porcelain firing may, however, decrease the mechanical properties 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 [18,19,43,44]. To simulate porcelain firing, a preheating process with or without actual veneering can be carried out. Conversely, the veneer material is more complicated to build up in a standardized way, as veneering generally requires a high level of skilled workmanship. It was therefore decided to exclude this production step in the present study. Another reason for this decision was based on the assumption that the veneering itself would not significantly affect the fracture strength in the connector area, since the elastic modulus of the veneering material is much lower than that of the core material (≈100 GPa for porcelain, compared to ≈200 GPa for Y-TZP) [14].

The FDPs can be thermocycled 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. Strength degradation is a slow process affecting Y-TZP differently depending on several microstructural parameters, such as yttrium distribution and concentration, the distribution of flaw populations and grain size [45].

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 multicyclic 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 [13,14,16].

Results from different *in-vitro* studies may be mutually comparable but this does not necessarily mean that they have clinical relevance. However, the result of an *in-vitro* study can affect the clinical choice of a dental material. To be clinically relevant, the fracture strength of a material intended for cementation upon natural preparations should range between 300 and 800 N. This ought to indicate that attempting to achieve exaggeratedly good results, with a test set-up not even trying to mimic natural teeth, should not be the main aim of an *in-vitro* study. Furthermore, every step in the test set-up of a study must be clearly presented or else the significance of that study must be questioned.

In this study, the choice of different materials for the tooth analogues was the most crucial factor affecting the load-at-fracture value and fracture mode. The FDPs tested on stiff aluminium tooth analogues showed both a higher load-at-fracture value and fracture mode with different fractures in both connectors in 75% of that group. The differences in fracture mode indicate that the stiff support supplied by the non-resilient aluminium model results in a fracture mode differing from that normally seen in clinical failures [8,14,39,46]. Unfortunately it is not possible to compare the results in the present study regarding fracture mode with other *in-vitro* studies, since the fracture modes in those studies were not recorded. This indicates that comparing results from different studies in which materials with differing properties are used to support the material tested are of no value.

Conclusions

Within the limitations of this *in-vitro* study, the following could be concluded:

- There is a need for a standardized, simple test set-up to achieve realistic and comparable results when *in-vitro* testing all-ceramic FDPs.
- Resilient, non-complex and resilient, complex tooth analogues give comparable test results when the test set-up is unchanged in all other aspects.
- Non-resilient (with an elastic modulus equivalent to or higher than that of aluminium) tooth analogues give high and unrealistic load-at-fracture values together with adverse fracture modes compared to FDPs involved in clinical failure.

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