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ORIGINAL ARTICLE

Fracture strength of monolithic all-ceramic crowns made of high translucent yttrium oxide-stabilized zirconium dioxide compared to porcelain-veneered crowns and lithium disilicate crowns

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Abstract

Objective. The aim of the study was to provide data on the fracture strength of monolithic high translucent Y-TZP crowns and porcelain-veneered high translucent Y-TZP crown cores and to compare that data with the fracture strength of porcelain-veneered Y-TZP crown cores and monolithic lithium disilicate glass-ceramic crowns. ***Materials and methods.*** Sixty standardized crowns divided into six groups ($n = 10$) were fabricated: monolithic high translucent Y-TZP crowns, brand A, monolithic high translucent Y-TZP crowns, brand B, veneered high translucent Y-TZP crown cores, brand A, veneered high translucent Y-TZP crown cores, brand B, heat-pressed monolithic lithium disilicate crowns and veneered Y-TZP crown cores. All crowns were thermocycled, cemented onto dies, cyclically pre-loaded and finally loaded to fracture. ***Results.*** The monolithic Y-TZP groups showed significantly higher fracture strength (2795 N and 3038 N) compared to all other groups. The fracture strength in the veneered Y-TZP group (2229 N) was significantly higher than the monolithic lithium disilicate group (1856 N) and the veneered high translucent Y-TZP groups (1480 N and 1808 N). ***Conclusions.*** The fracture strength of monolithic high translucent Y-TZP crowns is considerably higher than that of porcelain-veneered Y-TZP crown cores, porcelain-veneered high translucent Y-TZP crown cores and monolithic lithium disilicate crowns. The fracture strength of a crown made of monolithic high translucent Y-TZP is, with a large safety margin, sufficient for clinical use for the majority of patients. Porcelain-veneered Y-TZP crown cores show higher fracture resistance than monolithic lithium disilicate crowns.

Key Words: *Dental ceramics, glass ceramics, oxide ceramics, Y-TZP, zirconia*

Introduction

Fixed dental restorations based on yttrium oxide-stabilized tetragonal zirconia polycrystals (Y-TZP) have become popular thanks to its favorable mechanical properties, excellent biocompatibility and good esthetics when veneered with porcelain. An increasing number of studies on zirconia-based restorations show survival rates similar to conventional high noble metal-ceramic restorations, often considered to be the gold standard for fixed dental prostheses [1]. The risk of complete fracture is minimal and only a few complete fractures have been registered in studies on tooth-supported zirconia-based restorations and none in studies on implant-supported ones [2,3]. Almost all studies on fixed dental prostheses, however, report chip-off fractures, or more correctly classified as veneering material fractures, of the porcelain-veneer. Although few fractures have led to the removal of restorations, it is still a problem that needs to be considered [1–3].

A veneering material fracture creates a rough surface and sometimes sharp edges that need adjustment in the form of polishing or repair [2,4]. If the fracture affects the esthetic appearance or leads to functional impairments, such as loss of occlusal or approximal contacts, replacement might be necessary [2,4].

Different factors influencing the risk of veneering material fractures have been discussed, such as the mechanical properties of the core and veneer materials, design, thermal conductivity and differences in the thickness ratio and the coefficient of thermal expansion between the core and veneer materials [1,2,5,6]. Although the mechanical properties of

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the veneering porcelain have been improved and are now comparable with those used for metal-ceramic restorations [7], the problem with veneering material fractures still remains. Changing the veneering material from porcelain to glass-ceramic has not resulted in improved performance [8,9]. Laboratory and clinical studies have also failed to show any significant differences between application techniques: hand-layered and press-on veneers [10–12]. Much attention has focused instead on the importance of an adapted anatomical core design for veneer support as well as slow-cooling firing protocols to avoid stress formation in the veneer material during porcelain firing [6,8,13–16]. However, no clinical data confirming these hypotheses are available yet.

A recently introduced alternative technique utilizing milled veneer structures made of a lithium disilicate glass-ceramic material, fused onto Y-TZP cores by using 'CAD-on' technology, shows promising results compared with conventional veneering materials [12,17]. This technology, however, has design constraints that limit clinical usability.

Another possible way to avoid veneer fractures is to exclude the veneer and instead use monolithic crowns made exclusively from the high-strength core material [17–22]. This is possible by using CAD/CAM-technology where monolithic crowns can be milled in full anatomical contour from blanks. The production of monolithic crowns has other advantages such as reduced production time and improved cost-effectiveness. Instead of building up the porcelain in several layers and firing in multiple firing cycles, the monolithic crown can be individualized by using staining techniques commonly employed with glass-ceramics for more than three decades [18,23,24].

High-strength glass-ceramics based on lithium disilicate have been used in different combinations since their launch in the late 20th century [23,24]. This material combines good esthetics with high strength and has been suggested for monolithic crowns to be fitted in high load situations [23,24]. The material possesses approximately half the strength of Y-TZP, but 3-times the strength of the veneering porcelain that is used in combination with Y-TZP [7,24]. For veneered Y-TZP-based crown cores, the veneering material is the weak link in the construction being prone to fractures [23]. It has, therefore, been suggested that the overall strength of monolithic lithium disilicate crowns may exceed that of veneered Y-TZP-based crown cores [25]. Furthermore, it has been claimed that monolithic lithium disilicates display fracture resistance comparable to the metal-ceramic gold standard [25].

There are many situations, however, where a stronger material is needed to withstand loads and wear, e.g. in posterior areas and in cases of bruxism [23,24]. In such situations, monolithic Y-TZP could be adequate and a possible treatment alternative if it was not for the poor optical properties of the material [23]. Many patients might not accept restorations with the poor esthetic performance of traditional Y-TZP.

However, the introduction of high translucent Y-TZP, individualized using staining techniques, enables the esthetics to be sufficiently improved and makes restorations of this material an attractive treatment alternative [18,21]. By using monolithic Y-TZP, the survival and success rate of all-ceramic restorations might be improved and consequently the range of indications could be expanded.

Despite the fact that the optical properties are better than for traditional Y-TZP, the high translucent Y-TZP still leaves a lot to be desired from an esthetic viewpoint. It is known to the authors that some dental laboratories add porcelain to the high translucent Y-TZP cores to improve appearance, sometimes in the absence of instructions from the manufacturer.

Little information exists as to how these restorations, both monolithic and porcelain-veneered high translucent Y-TZP, perform in comparison with other treatment alternatives.

The aim of the study was to provide data on the fracture strength of monolithic high translucent Y-TZP crowns and porcelain-veneered high translucent Y-TZP crown cores and to compare that data with the fracture strength of porcelain-veneered Y-TZP crown cores and monolithic lithium disilicate glass-ceramic crowns under the null hypothesis that the result will be equal in all groups.

Materials and methods

A total of 60 standardized crowns divided into six groups of 10 were to be fabricated for the study. Each group was to undergo thermocycling, cyclic preload and load to fracture according to a protocol. The groups were as follows: (1) monolithic high translucent Y-TZP crowns, brand A (MTZ), (2) monolithic high translucent Y-TZP crowns, brand B (MTN), (3) veneered high translucent Y-TZP crown cores, brand A (VTZ), (4) veneered high translucent Y-TZP crown cores, brand B (VTN), (5) monolithic lithium disilicate crowns (MEM) and (6) veneered Y-TZP crown cores (VZN), as shown in Table I.

Master die

An impression of a metal die resembling a molar crown preparation, with a 120° cervical chamfer and a 20° angle of convergence, was made in an A-silicon impression material (President, Coltene® AG, Altstätten, Switzerland). The impression was poured with die stone (esthetic-base® gold, Dentona

Table I. Overview of the groups and materials used.

Abbreviation of the group	Crown design	Material	Veneering material	Manufacturer
MTZ	Monolithic	High translucent Y-TZP Z-CAD® HTL	—	Metoxit AG, Thayngen, Switzerland
MTN	Monolithic	High translucent Y-TZP NexxZr® HT	—	Sagemax® Bioceramics, Inc., Federal Way, WA
VTZ	Porcelain-veneered core	High translucent Y-TZP Z-CAD® HTL	IPS e.max® Ceram[a]	Metoxit AG, Thayngen, Switzerland
VTN	Porcelain-veneered core	High translucent Y-TZP NexxZr® HT	IPS e.max® Ceram[a]	Sagemax® Bioceramics, Inc., Federal Way, WA
MEM	Monolithic	Lithium disilicate glass-ceramic IPS e.max® Press	—	Ivoclar Vivadent® AG, Schaan, Liechtenstein
VZN	Porcelain-veneered core	Y-TZP NexxZr® HS	IPS e.max® Ceram[a]	Sagemax® Bioceramics, Inc., Federal Way, WA

HTL, High translucent; HT, High translucency; HS, High strength.
[a]Manufacturer Ivoclar Vivadent® AG, Schaan, Liechtenstein.

AG, Dortmund, Germany) to make a master die for the production of the 60 crowns (Figure 1).

Manufacture of crown cores and monolithic crowns

Two different wax-ups were scanned using a double scan technique: (1) full anatomy for the monolithic crowns in the groups MTZ and MTN and (2) reduced anatomy for the crown copings that were to be veneered in the groups VTZ, VTN and VZN (Figure 1). Subsequently to each scan of the wax-ups the master die was scanned. The scans were made in a dental laboratory scanner (3Shape D700, 3Shape Dental System, CAD Design Software, 3Shape A/S, Copenhagen, Denmark). Two CAD-files were produced in this way, one for the 20 monolithic crowns and one for the 30 veneered crown cores, all to be made in the pre-sintered Y-TZP materials in each case according to the manufacturers' instructions.

The internal cement space was set at 0.055 mm and the distance to the margin line at 0.35 mm, also according to the manufacturers' recommendations. The dimensions of the cores for the veneered crowns were to allow for the veneering porcelain in accordance with the manufacturers' recommendations (Figure 1).

Depending on the brand of Y-TZP, the CAD-files were then sent to either a production center (OpenMill® AB, Malmö, Sweden) for the groups MTN, VTN and VZN or to a dental laboratory (Teknodont AB, Malmö, Sweden) for the groups MTZ and VTZ, where the crowns were milled and sintered either to full anatomy or shaped as crown copings to be veneered in a final production step, depending on group.

The group MEM was double scanned as described above, with the same dimensions as the full anatomy crowns, but using another dental laboratory scanner (Everest® Scan pro 4101, Everest® Energy CAD Software Version 1.x, Kavo Dental GmbH, Biberach, Germany) (Figure 1). From the CAD-file produced in this scanning procedure, burn-out pre-shapes (Everest® C-Cast Ronde, Kavo Dental GmbH) for

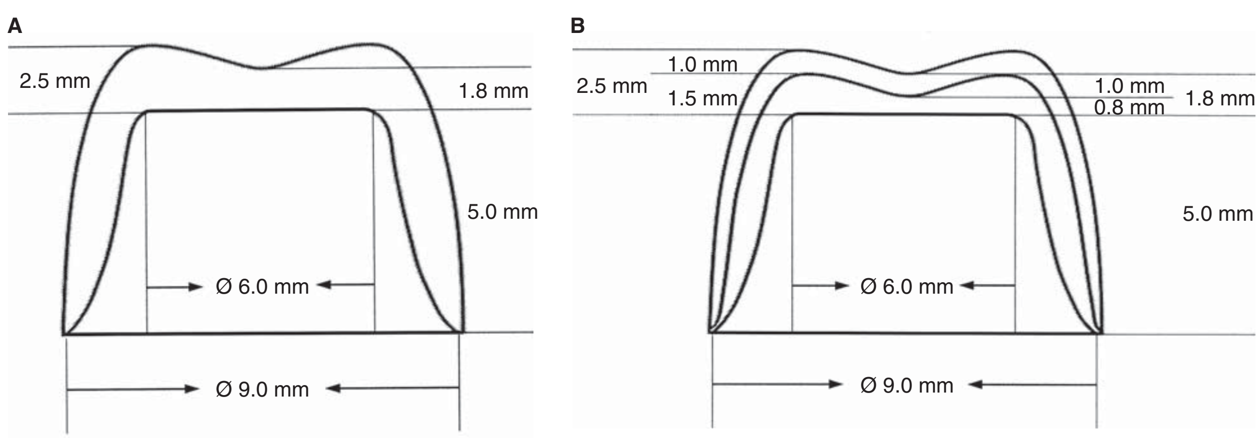

Figure 1. Schematic illustration of the design and dimensions of the crowns. (A) Monolithic crown, group MTZ, MTN and MEM. (B) Porcelain-veneered crown core, group VTZ, VTN and VZN.

Table II. Firing parameters[a] for IPS e.max® Ceram and the number of crowns in the furnace per firing (n).

Firing program	Pre-heating temperature (°C)	Pre-heating time (min)	Heating rate (°C/min)	Firing temperature (°C)	Holding time (min)	Vacuum start (°C)	Vacuum stop (°C)	Cooling (°C)	n^b
Liner[c]	403	4.00	40	960	1.00	450	959	—	5
Wash[d]	403	4.00	40	750	1.00	450	749	—	5
Dentin 1st/2nd[e]	403	4.00	40	750	1.00	450	749	—	1
Glaze[f]	403	6.00	60	725	1.00	450	724	450	5

[a]Current 29 March 2012.
[b]Number of crowns simultaneously in the furnace per firing.
[c]IPS e.max® Ceram ZirLiner 1/ZirLiner Build-Up Liquid allround, Ivoclar Vivadent® AG, Schaan, Liechtenstein.
[d]IPS e.max® Ceram Dentin A1/Build-Up Liquid allround, Ivoclar Vivadent®AG.
[e]IPS e.max® Ceram Dentin A1/Build-Up Liquid allround, Ivoclar Vivadent®AG.
[f]IPS e.max® Glaze Paste, Ivoclar Vivadent® AG.

pressing the IPS e.max Press crowns (IPS e.max® Press, Ivoclar Vivadent®AG, Schaan, Liechtenstein) were milled (Everest® Engine 4140, Milling Pin ZS 1 Long 029800, Milling Pin ZS 3 long 028461, Kavo Dental GmbH) and invested, pressed, divested, stained and finally glazed, all according to the manufacturer's recommendations.

Veneering of the crown cores in groups VTZ, VTN and VZN

The 30 crown copings with reduced anatomic shape were veneered with a glass-ceramic porcelain (IPS e.max® Ceram, Ivoclar Vivadent® AG). A specially-made knife was used to standardize the shape and dimensions of the porcelain-veneer during veneer build-up, according to a technique described in detail in previous studies [26,27].

The thickness of the porcelain-veneer was a uniform 1 mm and the amounts and proportions of the porcelain-powder and liquid were measured and standardized. The same dental technician performed the veneering. The firing cycles were carried out in a calibrated furnace (Programat® EP 5000, Ivoclar Vivadent® AG) and each crown underwent five firings: liner, wash, dentin 1 and 2 and finally glaze firing, all steps according to the manufacturer's recommendations (Table II).

Surface treatment of the monolithic Y-TZP crowns in groups MTZ and MTN

The 20 monolithic Y-TZP crowns were manually polished by one dental technician for 1 min using a goat hair brush (Robinson's Bristle Brushes Soft no. 11, Buffalo Dental MFG, Syosset, NY) and a polishing paste (Zircon-Brite, Dental ventures of America, Inc., Corona, CA, USA) with a rotation speed of 5000 rpm.

Thermocycling

All 60 crowns underwent 5000 thermocycles in two baths, 5 and 55°C, in a specially constructed thermocycling device. Each cycle lasted 60 s, 20 s in each bath and 10 s for transfer between the baths.

Cementation

Sixty dies were used to support the crowns during mechanical testing. The dies were made in a polyoxymethylene resin material (DuPont™ Delrin®acetal resin, E.I DuPont de Nemours & Comp, Neu Isenburg, Germany/Wallins Mekaniska i Eslöv AB, Eslöv, Sweden) having the same shape and dimensions as the master die and with mechanical properties resembling those of the natural tooth [28,29]. Before cementation the dies were sandblasted with 110 µm aluminum oxide powder at a pressure of 2 bars and finally steam cleaned. All cementation surfaces of the crowns were treated according to the manufacturers' recommendations, comprising thorough cleaning of all groups and additional etching of the glass-ceramic MEM group with hydrofluoric acid (IPS® Ceramic Etching Gel, Ivoclar Vivadent®AG). Subsequently, all crowns were luted using a universal primer (Monobond® Plus, Ivoclar Vivadent®AG) and a resin cement (Multilink Implant®, Ivoclar Vivadent®AG).

A standardized seating load of 15 N was applied to the crowns in the direction of insertion during

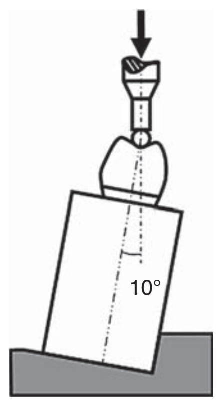

Figure 2. Schematic illustration of the test set-up at cyclic pre-load and load to fracture.

Table III. The results for fracture strength (N): load at fracture for each specimen, arithmetic mean, standard deviation (SD), statistical differences and fracture mode: Cohesive/Total (*n*).

	Group					
Specimen	MTZ	MTN	VTZ	VTN	MEM	VZN
1	2932	2991*	1971	2036	1587	2215
2	2509	3321*	1274	1936	1875	2290
3	2706	2510*	1544	1768	2144	2344
4	2801	2922*	1614	1731	1889	2215
5	2581	3444*	1052	1504	1778	2161
6	2745	3285*	1209	1852	1675	2095
7	2993	3003*	1304	1969	1998	2146
8	3003	3012*	1521	1854	1981	2259
9	2993	3020*	1736	1770	1821	2133
10	2686	2875*	1574	1664	1807	2431
Mean (N)	2795[a]	3038*[a]	1480	1808[b]	1856[b]	2229
SD	± 179	± 264	± 272	± 156	± 161	± 104
Fracture mode Cohesive/Total (*n*)	0/10	0/0*	2/8	5/5	0/10	3/7

SD, Standard deviation.

*No fractures occurred and the loading was terminated due to acoustic events and load drop, thus the highest measurable load of each specimen was recorded and the mean was based on these measurements.

[a,b]Groups denoted with the same superscripted letter indicate no statistical significant difference in fracture strength ($p > 0.05$) between the groups.

cementation and light curing. Finally the cement excess was removed using a surgical blade (AESCULAP® no. 12, Aesculap AG & Co, Tuttlingen, Germany) and the crowns were stored in a moist environment for 24 h before the mechanical test was done.

Cyclic pre-load and load to fracture

All crowns were cyclically pre-loaded in a wet environment for 10,000 cycles using a specially constructed pre-load device (MTI Engineering AB, Lund, Sweden/Pamaco AB, Malmö, Sweden). The loads were between 30–300 N with a load profile in the form of a sine wave at 1 Hz and a 10° load angle relative to the vertical plane, as shown in Figure 2. The load was applied on the occlusal surface of the crowns with a centrally placed stainless steel ball (2.5 mm diameter) and a 1 mm thick plastic foil (Erkoflex, Erkodent® Erich Kopp GmbH, Pfalzgrafenweiler, Germany) placed between the ball and the crown.

Finally the crowns were mounted as before in a testing jig at a 10° angle relative to the vertical plane and loaded to fracture in a wet environment using a universal testing machine (Instron model 4465, Instron®, Canton, MA) (Figure 2). The load was again applied centrally on the occlusal surface of the crowns with a steel ball (2.5 mm diameter) and a 1 mm thick plastic foil placed between the ball and the crown. The crosshead speed was 0.255 mm/min.

Load at fracture (N) was registered and a fracture was defined as visible fracture or as occurrence of acoustic event and load drop.

The fracture surfaces were examined and analyzed under a light microscope (Leica DFC 420, Leica Application Suite v. 3.3.1, Leica Microsystems CMS GmbH, Wetzlar, Germany) to determine the type of fracture: total, adhesive or cohesive.

Statistics

One-way ANOVA, Tukey's test was used to determine differences in fracture strength between the groups (IBM SPSS Statistics 20, SPSS Inc., Chicago, IL). Differences in fracture mode were determined using Fisher's exact probability test. The level of significance was set to $\alpha = 0.05$.

Results

Fracture strength

The monolithic groups MTZ and MTN showed significantly higher ($p \leq 0.001$) fracture strength compared to all other groups (Table III). The fracture strength was significantly higher ($p \leq 0.001$) in the veneered group VZN in comparison with the monolithic lithium disilicate group MEM and the veneered high translucent groups VTZ and VTN.

The monolithic lithium disilicate group MEM showed significantly higher ($p \leq 0.001$) fracture strength than the veneered high translucent group VTZ, but the result equaled ($p > 0.05$) that of the VTN group.

Fracture mode

Two fracture types were observed: total fracture through the entire crown and cohesive fracture through the porcelain-veneer only. The monolithic groups MTZ and MEM showed only total fractures, whereas the veneered groups VTZ, VTN and VZN exhibited both total and cohesive fractures but no adhesive fractures (Table III). There were significantly more ($p < 0.05$) total fractures in group MTZ and MEM compared to group VTN.

No significant differences were found between the other groups.

Discussion

The stress patterns created in the oral cavity are complex and it is difficult to assess the loads a dental restoration must resist in order to withstand the environmental impact during function over time. Therefore, when testing a material, it is often best to evaluate clinically shaped restorations under environmental conditions close to those present in the oral cavity and to compare the results with clinical data on the maximum loads that might occur in the oral cavity [5,30,31]. It is also prudent to allow for a reasonable margin of safety as *in-vitro* studies never completely mimic the clinical situation and the wide strength scatter that lies within the properties of dental ceramics [30–33].

In the present study, the specimens were shaped as crowns and when bi-layered materials were used, they were layered as recommended by the manufacturers for restorations intended for clinical use. During testing, environmental aspects were considered in the laboratory set-up. To simulate aging of the materials, resembling the fatigue process in the oral environment, thermocycling and cyclic pre-loading in a wet environment was used, a combination of methods accepted and commonly employed in *in-vitro* studies [8,15,27,30,31,34,35]. These testing procedures have been discussed in previous studies as having a detrimental effect on dental ceramics which resembles the clinical situation [8,30,31,34,36]. Dental ceramics are susceptible to slow crack growth. During cyclic loading, especially in a wet environment with water molecules causing stress corrosion at the crack tip, cracks tend to propagate even under small loads, thus degrading the material [5,30–32]. This fatigue significantly reduces the fracture strength of the ceramic material, making it prone to fracture under loads below the initial fracture strength of the material. Since dental restorations must be able to resist the stresses in the oral environment, artificial aging is an essential part of an *in-vitro* study, preventing obtaining unrealistically high fracture loads. There is, however, no consensus regarding the appropriate procedure for aging [37]. Therefore, the number of cycles, for instance, varies between 100–50,000 cycles in studies and is often arbitrarily set. According to an ISO standard [38] 500 cycles in a temperature range of 5–55°C is indicated as an appropriate aging procedure. The number of cycles in the present study was set to 5000 cycles, a number frequently used, based on previous studies and with a margin of safety with regard to the ISO standard.

The sample size was based on previous studies with a similar study design, where significant differences between ceramic materials with sample sizes of 10 have been shown and taking the level of significance, the dispersion (standard deviation) expected of ceramic materials and what could be considered to be a clinically relevant difference, into account [8,27]. Therefore, no power analysis was considered necessary. Polishing of the monolithic Y-TZP crowns was performed as studies have shown lower wear of enamel antagonists as well as high wear resistance of polished monolithic Y-TZP materials compared to glazed [18–22]. Furthermore, there are no glaze materials or firing parameters specifically intended for monolithic Y-TZP available or provided by the manufacturers.

The die material used had a modulus of elasticity close to that of the natural tooth which has been pointed out as important when testing dental restorations in order to gain relevant strength data comparable with the clinical situation [29,30,35].

The preparation design used, with a deep cervical chamfer, was the same for all groups in accordance with the manufacturer's recommendations for all material combinations included in the study. The cement used was also the same for all groups and was again recommended by the manufacturer for both the Y-TZP groups and the lithium disilicate group. The reason for choosing a cement system developed for implant components rather than a cement system developed for bonding to tooth substance was that the latter would not have cured sufficiently due to the absence of tooth substance. According to the manufacturer, however, the properties of the cement are no different from the properties of one used for bonding to tooth substance. Furthermore, the monolithic crowns and the veneered crown cores, respectively, had a similar design and dimensions and the same porcelain was used for all the veneered crown cores, facilitating comparisons between the groups.

Attempts have been made to estimate the human maximum bite force, but estimates vary greatly and the literature presents a wide range of suggestions on the requirements for fracture strength of a dental restoration [5,8,30,39–42]. The average maximum bite force varies from one patient to another and intra-individually over time [39–42]. Moreover, the range varies markedly from one area in the mouth to another, increasing from the incisors to the third molar, being ~ 90–340 N in the anterior region, 220–450 N in the premolar region and 400–900 N in the molar region [40–42]. All materials tested in the present study presented results that exceeded the expected average maximum loads with a large safety margin, indicating sufficient fracture strength, with the lowest mean value 1480 N (VTZ). Nevertheless, clinical studies have shown that veneering material fractures, and not complete fractures, are a common complication in the clinical situation [1,2]. This indicates that even though load at fracture values may be sufficient, the overall performance of the restoration is not, which calls for stronger materials than the veneered Y-TZP-restorations susceptible to veneering material fractures.

Guess et al. [25] concluded in an *in-vitro* study that monolithic lithium disilicate crowns possess higher fracture strength than porcelain veneered Y-TZP crown cores. That conclusion could not be confirmed in the present study since the veneered Y-TZP crown cores (VZN) showed significantly higher fracture loads compared to the monolithic lithium disilicate crowns (MEM). Because the study designs were not identical, e.g. regarding the fatigue and loading conditions, the results are not directly comparable. Nevertheless, as the results of the present study were contradictory, it might provide additional information about appropriate treatment alternatives and improvements in crown design. The same porcelain was used in both that study and the present one, so the difference in results is probably not explained by the brand of porcelain. However, the lithium disilicate and the Y-TZP core materials used in the two studies did differ (IPS e.max® Press and IPS e.max® CAD, NexxZr® and IPS e.max® ZirCAD, Ivoclar Vivadent® AG). As the composition of the materials varies and the manufacturers specify different mechanical properties for the materials, this might be one factor which explains the differences in results.

Another possible reason is that the study by Guess et al. [25] used a 0.5 mm evenly thick core to support an unevenly thick (1.0–1.5 mm) porcelain layer, while in the present study anatomically shaped cores were used to support an evenly thick (1.0 mm) porcelain layer recommended for high load situations and that has been shown to significantly increase the strength of a veneered crown core [8,13]. Moreover, previous studies [6,16] have concluded that a thicker porcelain layer on a core with a low thermal diffusivity, such as Y-TZP, increases the risk of residual tensile stresses within the porcelain, hence making the restoration more prone to cohesive veneering material fractures. Differences in the porcelain layering technique and the dexterity of the dental technician who created the veneer by hand could be another reason since the monolithic lithium disilicate is manufactured in a more standardized manner where the technician's personal skills play a minor role.

When comparing the MEM group with the two groups of veneered high translucent Y-TZP crown cores, the load at fracture of one of the latter (VTN) equaled that of the MEM group while the other (VTZ) showed significantly lower fracture loads. The high translucent Y-TZP materials, however, are not primarily intended to be veneered but are developed for monolithic crowns, individualized by staining only.

The monolithic high translucent Y-TZP crowns in group MTZ and MTN presented values for mean loads at fracture that far exceed loads expected during clinical use. Furthermore, all values registered for the MTN group are lower than the potential fracture loads, due to acoustic events and load drops that occurred during the test prior to visible fractures and thereby defined failure. Those events were associated with fracture of the dies rather than of the crowns, implying that the crowns might have resisted higher loads than those registered. Even so, in those cases the loads reached were far above what is clinically relevant for a dental restoration [5,30]. In consistency with the results of the present study, Beuer et al. [17] concluded that monolithic Y-TZP crowns have a higher load-bearing capacity than veneered Y-TZP crown cores.

All crowns in the monolithic groups MEM and MTZ fractured with total fractures. None of the veneered crown cores showed any adhesive fractures comprising the core–veneer interface. This is in accordance with clinical studies on Y-TZP where most fractures are superficial cohesive veneering material fractures and adhesive fractures are rare [43–45]. In fact, in a clinical situation with the restoration remaining *in situ*, fractures may be difficult to assess, thus fractures defined as adhesive might actually be cohesive fractures [2,46].

In summary, monolithic high translucent Y-TZP crowns seem to be a promising treatment alternative, especially for patients with a history of fractured restorations. Clinical studies are needed, however, to confirm the findings of this *in-vitro* study. Furthermore, the material properties of high translucent Y-TZP need to be thoroughly investigated, e.g. regarding abrasiveness, color-resistance, optical properties and degradation of the material in the oral environment over time.

Conclusion

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

- The fracture strength of monolithic high translucent Y-TZP crowns is considerably higher than that of porcelain-veneered Y-TZP crown cores, porcelain-veneered high translucent Y-TZP crown cores and monolithic lithium disilicate crowns.

- The fracture strength of a crown made of monolithic high translucent Y-TZP is, with a large safety margin, sufficient for clinical use for a majority of patients.
- Porcelain-veneered Y-TZP crown cores show higher fracture resistance than monolithic lithium disilicate crowns.

The null-hypothesis is thereby rejected.

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