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Simulation of clinical fractures for three different all-ceramic crowns

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Comparison of fracture strength and fracture modes of different all-ceramic crown systems is not straightforward. Established methods for reliable testing of all-ceramic crowns are not currently available. Published in-vitro tests rarely simulate clinical failure modes and are therefore unsuited to distinguish between the materials. The in-vivo trials usually lack assessment of failure modes. Fractographic analyses show that clinical crowns usually fail from cracks initiating in the cervical margins, whereas in-vitro specimens fail from contact damage at the occlusal loading point. The aim of this study was to compare three all-ceramic systems using a clinically relevant test method that is able to simulate clinical failure modes. Ten incisor crowns of three types of all-ceramic systems were exposed to soft loading until fracture. The initiation and propagation of cracks in these crowns were compared with those of a reference group of crowns that failed during clinical use. All crowns fractured in a manner similar to fracture of the clinical reference crowns. The zirconia crowns fractured at statistically significantly higher loads than alumina and glass-ceramic crowns. Fracture initiation was in the core material, cervically in the approximal areas.

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There has been a steady development of stronger and more versatile ceramic materials for dental use in the last decades. Many variables are decisive for the suitability of a crown material, for instance color, translucency, and fracture strength. Strength estimates by laboratory tests indicate that alumina, zirconia, and lithium disilicate glass-ceramics should be able to withstand relevant forces present during oral function for single crowns (1–3). Nevertheless, fractures are one of the main problems reported in clinical follow up for ceramic restorations (4). Several different types of fracture may occur during clinical use. Fractures may be cohesive in the veneering ceramic only (chipping) or adhesive between core and veneer (delamination), or bulk fracture of the core itself may occur. The proportion of the different types of clinical fracture modes is poorly documented. Comparison of different materials is not straightforward as they are used in different ways and are rarely compared in larger randomized clinical trials. Chipping and delamination may have several different causes, such as defects in the veneering material, incorrect cooling rates during veneering, a weak chemical bond between the core and the veneer, and traumatic occlusion. Why core fractures occur during clinical function is less obvious. There are currently no established in-vitro methods for testing the fracture strength of all-ceramic crowns. Laboratory tests commonly used to assess the fracture strength of crown-shaped specimens tend to produce fractures starting from the contact damage caused by the loading device (3, 5–7). These types of fractures are not observed in crowns that have fractured during clinical use (8–12). Therefore, it is not certain whether the in-vitro results are applicable for use in clinical decision making (5). In a previous study it was found that soft occlusal loading on alumina-based ceramic crowns attached to abutment models with a low modulus of elasticity and a Poisson's ratio similar to that of dentin, can simulate clinical fractures for molar crowns (12, 13). The epoxy abutment is compressed during axial loading, which results in a slight bulging of the abutment. The bulging creates tension in the cervical region of the crown fixed to the abutment and fracture occurs at the cervical margin. The load at fracture registered with this method was clinically relevant. It is important to assess this method for different types of preparations and different types of crown materials in order to evaluate further the method's general applicability and clinical relevancy (13).

The aim of this study was to evaluate the fracture patterns and load at fracture of alumina, zirconia, and glass-ceramic incisor crowns using a newly developed test that induces clinically relevant fractures.

Material and methods

Thirty crowns were produced to fit a model of a prepared central incisor (tooth no. 21; Fig. 1). A deep chamfer

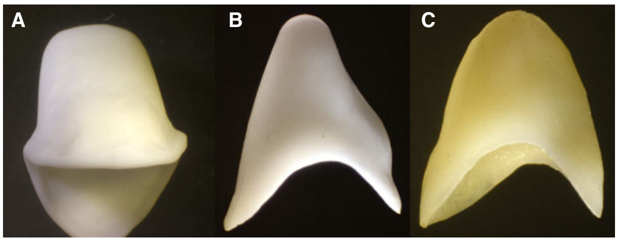

Fig. 1. Specimens as received from the dental technician before veneering. (A) Alumina core, frontal view. (B) Zirconia core, mesial view. (C) Glass-ceramic core, distal view. Note that the glass-ceramic core has thicker walls than both the alumina and zirconia cores.

finish line was used because this is recommended by most manufacturers of dental ceramics. The model was made as a natural tooth preparation with rounded edges and a finish line that curved apically on the buccal and palatal surfaces. Ten crowns were made with alumina cores, 10 with zirconia cores, and 10 with glass-ceramic cores (Table 1). The cores of alumina and zirconia were made to a uniform thickness of 0.6 mm. The glass-ceramic cores were built up to an anatomic form that requires only a thin layer of veneering ceramic (Fig. 1). The cores were veneered with veneering ceramics that were suited for the core materials according to the manufacturers' recommendations. The crowns were luted with zinc phosphate cement (De Trey Zink; Dentsply DeTrey, Konstanz, Germany) to 30 identical epoxy models of the preparation. Excess cement was removed and after a 5-min setting time the crowns were placed in distilled water at 37°C for at least 24 h. The crowns were subsequently loaded at the incisal edge with a flat, steel cylinder of 30 mm in diameter cushioned with a 2-mm-thick rubber disc of hardness 90 Shore A (EPDM 90) to avoid contact damage (Fig. 2). The load was applied in a universal materials testing system at 0.5 mm min^{-1} until fracture occurred (Lloyd Instruments, Leicester, UK). The procedure was performed with the crowns immersed in water at 37°C. Load at fracture was recorded. The procedure was recorded by video camera for all crowns.

The fractured crowns were analyzed using fractographic methods (14). The fracture start of the main, primary fracture was localized for each crown. Secondary fractures and general crack propagation was also mapped. The fracture analysis results were compared with the video recordings to verify the crack start and crack propagation. The fracture features were compared with a reference group of 25 crowns (from 21 incisors and four premolars) fractured in clinical use and obtained from Norwegian dentists. Non-parametric statistics were used for the comparison between groups (IBM SPSS statistical software, Armonk, NY, USA). The level of significance was set to 0.05.

Results

All crowns, except one zirconia crown, fractured through both core and veneer. There were statistically significant differences among the groups (Kruskall–Wallis test $P < 0.001$, Fig. 3). The fracture loads of the zirconia-based crowns were statistically significantly higher than were those of the other two materials (Mann–Whitney U-test $P < 0.001$). There was no statistically significant difference between the alumina and glass-ceramic crowns (Mann–Whitney U-test $P = 0.25$). The fracture modes achieved with the test methods resembled the fracture patterns observed in clinical failures of the retrieved reference crowns (Fig. 4). The fracture modes are presented in Fig. 5.

All primary core fractures were initiated in the cervical area in the approximal region, as observed in the clinical reference group. One crown with a zirconia core delaminated completely between the core and the buccal veneering ceramic (adhesive fracture). This fracture started on the incisal edge from contact damage. The core of this crown remained intact. A similar fracture was observed in one of the clinical reference crowns with zirconia although this was a cohesive fracture within the veneer. All the other zirconia crowns exhibited partial delamination of the veneering ceramic in addition to the core fractures. No delamination was observed on the other two crown types. Minor secondary chipping of the veneering ceramic was observed in all crowns (cohesive fractures). Twenty-three of the 30 test crowns exhibited secondary fractures in the core material. These were mainly caused by bifurcations of the main fracture (16 specimens) and new cracks starting at the opposite side of the crown margin compared with the start of the primary fracture (seven specimens). These findings are similar to those of the clinical reference group.

Table 1

Details of the materials used in the trial

Material group	Material composition	Brand name	Manufacturer	Veneering material
Alumina	Al_2O_3 100%	Vita In-Ceram AL for inLab	Vita Zahnfabrik, Bad Säckingen, Germany	Ducera AllCeram, Degudent, Hanau, Germany
Zirconia	ZrO_2 99% (HfO_2/Y_2O_3) Al_2O_3 0.25%	Starceram Z-Al-Med HD	HD, H.C. Stark Ceramics, Selb, Germany	IPS e-max Ceram, Ivoclar Vivadent, Schaan, Liechtenstein
Glass-ceramic	$LiSi_2O_5$ 70% SiO_2 30% (K_2O, P_2O_5, ZrO_2, ZnO)	IPS e.max Press	Ivoclar Vivadent	IPS e-max Ceram, Ivoclar Vivadent

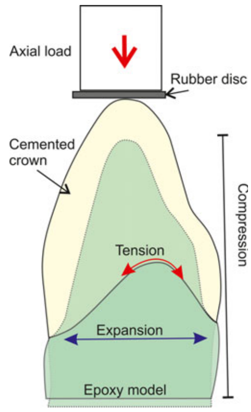

Fig. 2. The test set up. The crowns are cemented to an epoxy model of the tooth preparations. Axial load is applied on the incisal edge at 1 mm min^{-1} (red arrow), but cushioned to avoid contact damage. The load causes the epoxy model to compress vertically and expand horizontally (blue arrow), which causes tension at the cervical margin of the crown. The area of expected concentration of tension is marked with a red arrow at the top of the curvature.

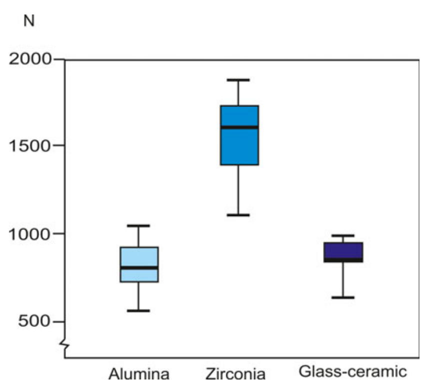

Fig. 3. Box-plot of the load at fracture in Newton (N) for the three different groups. The median values are indicated as horizontal lines in the box, and the bars represent the maximum and minimum values.

Discussion

The provoked in-vitro fracture modes closely matched the in-vivo fracture modes observed in retrieved reference crowns, indicating that the test method is well suited for simulating clinical stress in all-ceramic crowns. Vertical tension in the cervical crown margin seems to have initiated fractures from the cervical margin in the approximal area. The tension is difficult to measure but seems to be created by clinically relevant axial loads. The differences among materials are therefore relevant for clinical application. The finding that zirconia was stronger than the other two materials is in accordance with previous findings and assumptions (15–18). However, the mean fracture load was lower than in previous studies of crown-shaped specimens (3, 19–24). This is probably caused by the differences in crack initiation. In the present study the fractures were initiated in the cervical margin as a result of tension created by the expanding abutment, as observed in a previous study that used the same method (13). The cervical margin is the thinnest area of the crown and the area most susceptible to machining damage (25). The occlusal surface is usually the thickest area of the crown; furthermore, the occlusal surface is supported by the occlusal area of the tooth. In previous in-vitro trials the fractures have been initiated from occlusal contact damage (5) or from cone cracks under the occlusal surface (10, 26). With a rigid abutment material, the tensile stress at the cervical margin will be lower. Therefore, higher loads are required to obtain fractures. However, few clinical core–veneer crowns have had the same fracture mode as presented in these in-vitro tests.

Clinically fractured crowns may have many different fracture modes depending on the material, design, and load. Chipping is probably the most common fracture occurring in clinical use. However, it is more complicated to retrieve such fractures without damaging the remaining crown. Bulk fractures or chipping have been shown to start from contact damage on the occlusal surface in monolayer crowns without high-strength cores (26). Furthermore, fractures starting from the inner surface of the core in the occlusal area as a result of radial cracks have been presented both in clinical failures (10, 26, 27) and in laboratory failures (28). The focus of this study was to simulate the fracture modes observed in the retrieved clinical fractures we received and thus the aim was to avoid the other types of fracture modes. To fully comprehend the behavior of all-ceramic crowns, all clinical failure modes must be investigated further. Clinical trials rarely include thorough analyses of failure modes. So far, it seems that chipping is more predominant than core failures, especially for zirconia crowns (4, 29). Nevertheless, zirconia core failures do occur in spite of the high strength, and it would be of interest to find out why.

The finding, in the present study, that all core fractures started in the approximal area indicates that this is the weakest point of the crowns or the area that is most exposed to tension. This may be explained by the differences in mechanical properties between dentin and dental ceramics. Ceramics are stiff and brittle and deform very little before fracture during loading (30, 31). However, dentin is elastic and will deform during loading (32, 33). A high occlusal load will compress the dentin, resulting in an increase in the diameter of the tooth cervically (bulging), which in turn will cause tension at the crown margins (hoop forces) (Fig. 2). A previous study revealed that an expansion of 0.1% was sufficient to cause a fracture of alumina crowns for molars (13). It was not possible to measure reliably the bucco–palatal expansion of the cervical crown margin for the present specimens, but it is reasonable to assume that an expansion of approximately 7 μm has occurred. The crown margins are usually not even or leveled, creating stress foci in the areas that curve toward the occlusal surface (e.g. approximal). This curve is often located where the crown height is lowest, which gives a shorter axial wall. The combination of the curvature and the shorter axial wall in this area is

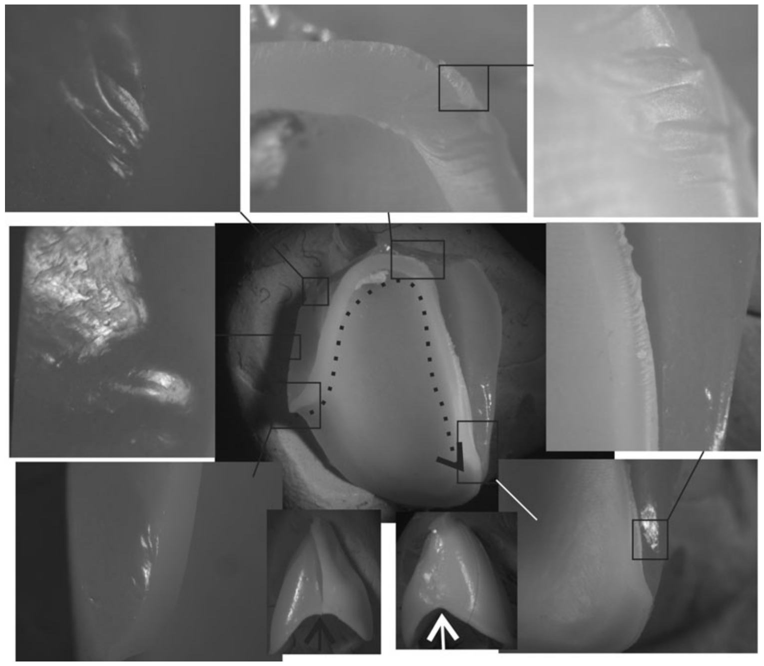

Fig. 4. Fractographic map of a zirconia crown fractured in vitro by simulation of clinical fracture. The boxes indicate size and location of the magnified images surrounding the central image. Dotted arrow indicate crack propagation, Black arrow indicate fracture origin and white arrow indicate fracture end.

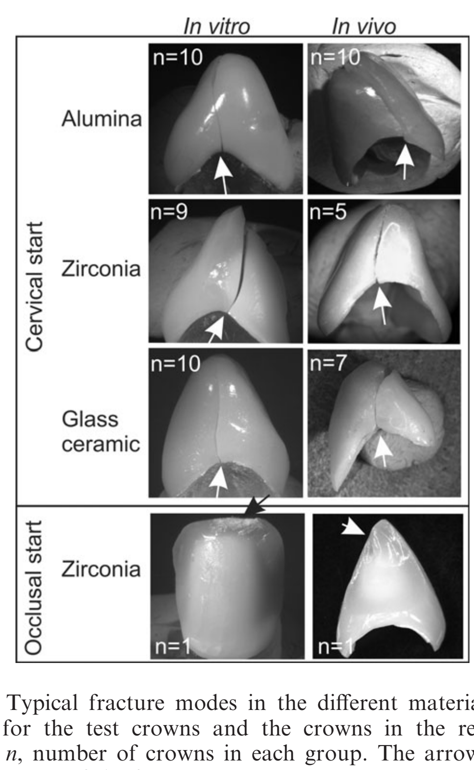

Fig. 5. Typical fracture modes in the different material categories for the test crowns and the crowns in the reference group. *n*, number of crowns in each group. The arrows indicate the start of the fracture.

probably the reason that most fractures start here. Finite element analyses have found increased concentration of stress in the cervical margin with increased difference in axial wall height in the bucco–palatal walls compared with the mesial–distal walls (34). However, the observed fractures in the test specimen and the reference crowns do not follow the shortest path across to the other side. This indicates that the fracture is not caused only by the shorter wall height.

The zirconia crowns displayed delamination of the veneer during fracture, whereas the other two crown types only had minor chipping within the veneering ceramics. These findings are in accordance with clinical findings where chipping is reported more frequently for zirconia-based restorations than for other all-ceramic restorations (31, 35–37). Moreover, event though adhesive fractures were found in all zirconia crowns, these occurred at a higher load than the fracture load in the other groups. The crown with total delamination differed significantly from all other test crowns, but was similar to an in-vivo crown that had been removed as a result of fracture of veneering ceramic only. This type of delamination fracture is usually called 'chipping' in clinical trials. Such a failure would nevertheless have been classified as a 'clinical failure' and would require replacement owing to the large impact on function and esthetics. Chipping damage is more frequently observed in clinical trials and has been successfully simulated in vitro previously. All these findings indicate that the results are clinically relevant and can be used for clinical decision making. Furthermore, they indicate that the test method is well suited for comparing load at fracture and fracture modes of different all-ceramic crowns.

The glass-ceramic crowns achieved similar fracture-strength values as the alumina-based crowns, even though glass-ceramic is a weaker material, as measured in flexural strength tests of bar-shaped specimens (15).

This may be explained by the differences in the geometry of the cores at the crown margin (Fig. 1). The alumina cores had a uniform thickness of 0.6 mm with a thin edge at the crown margin to allow room for veneering ceramic all the way down to the margin, allowing the opaque, uniform color of the alumina to be covered. The glass-ceramic cores are more translucent than alumina and can be made in many different tooth colors. It is not necessary to have veneering material all the way to the margin on these crowns in order to achieve optimal esthetics. These cores were built up to an anatomic form and with a thicker crown edge than on the alumina and zirconia copings. The glass-ceramic cores were covered with thinner layers of veneer and glaze than the alumina and zirconia coping to achieve an anatomically correct crown. When evaluating the fracture strength in this trial it must be taken into consideration that the glass-ceramic crowns can probably achieve better strength values clinically when adhesive luting agents are used (38). The information from the manufacturers indicates that both conventional and adhesive luting agents can be used. However, adhesive cementation is recommended for optimal strength.

Several in-vitro studies have been performed to assess the importance of the geometry of the finish line, but the results of the different trials are ambiguous. Some trials find that the deeper the chamfer, the stronger the crown; others find no significant difference (7, 39–41). As most of these trials have loaded the crowns in such a way that the fractures start occlusally or on the incisal edge, it is not surprising that the effect of the finish-line design is limited. One study, using a method similar to that of the present study, found that the geometry of the finish line is of great importance to both fracture initiation and fracture strength for dome-shaped glass structures (11). Further studies, using clinically relevant test methods, must be performed to verify the effect of finish-line geometry for dental ceramic crowns in order to recommend changes in the design of the restoration or the preparation. The newly developed test method is well suited for assessing the clinically relevant fracture load of different all-ceramic crown materials. All the tested crowns fractured at loads that indicate suitability for clinical use in anterior crowns.

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Conflicts of interest – The authors declare that no conflicts of interest exist for any of the products or equipment mentioned in the paper.

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