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In Vitro Study of Fracture Load and Fracture Pattern of Ceramic Crowns: A Finite Element and Fractography Analysis

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Abstract

Purpose: This in vitro study investigated the null hypothesis that metal-free crowns induce fracture loads and mechanical behavior similar to metal ceramic systems and to study the fracture pattern of ceramic crowns under compressive loads using finite element and fractography analyses.

Materials and Methods: Six groups (n = 8) with crowns from different systems were compared: conventional metal ceramic (Noritake) (CMC); modified metal ceramic (Noritake) (MMC); lithium disilicate-reinforced ceramic (IPS Empress II) (EMP); leucite-reinforced ceramic (Cergogold) (CERG); leucite fluoride-apatite reinforced ceramic (IPS d.Sign) (SIGN); and polymer crowns (Targis) (TARG). Standardized crown preparations were performed on bovine roots containing NiCr metal dowels and resin cores. Crowns were fabricated using the ceramics listed, cemented with dual-cure resin cement, and submitted to compressive loads in a mechanical testing machine at a 0.5-mm/min crosshead speed. Data were submitted to one-way ANOVA and Tukey tests, and fractured specimens were visually inspected under a stereomicroscope (20×) to determine the type of fracture. Maximum principal stress (MPS) distributions were calculated using finite element analysis, and fracture origin and the correlation with the fracture type were determined using fractography.

Results: Mean values of fracture resistance (N) for all groups were: CMC: 1383 ± 298 (a); MMC: 1691 ± 236 (a); EMP: 657 ± 153 (b); CERG: 546 ± 149 (bc); SIGN: 443 ± 126 (c); TARG: 749 ± 113 (b). Statistical results showed significant differences among groups ($p < 0.05$) represented by different lowercase letters. Metal ceramic crowns presented fracture loads significantly higher than the others. Ceramic specimens presented high incidence of fractures involving either the core or the tooth, and all fractures of polymer crown specimens involved the tooth in a catastrophic way. Based on stress and fractographic analyses it was determined that fracture occurred from the occlusal to the cervical direction.

Conclusions: Within the limitations of this study, the results indicated that the use of ceramic and polymer crowns without a core reinforcement should be carefully evaluated before clinical use due to the high incidence of failure with tooth involvement. This mainly occurred for the polymer crown group, although the fracture load was higher than normal occlusal forces. High tensile stress concentrations were found around and between the occlusal loading points. Fractographic analysis indicated fracture originating from the load point and propagating from the occlusal surface toward the cervical area, which is the opposite direction of that observed in clinical situations.

When single crowns are indicated, mechanical properties (such as flexure strength, modulus of elasticity, and fracture toughness), marginal adaptation, and esthetic appearance are essential factors for determining which system to use,[1] but the functional aspect should be considered first. Although incorporation of a metal substructure has been shown to improve the fracture resistance of ceramic crowns,[2-4] metal-free reinforced restorative systems have become popular because of the less favorable esthetic appearance of metal ceramic crowns.[3] The routine use of metal-free crowns has resulted in an increasing number of fractured restorations, despite leucite and lithium disilicate reinforcements.[2]

The enhanced esthetic appearance of ceramic restorations is a result of improved light transmission through the restoration as compared to metal ceramic crowns.[5,6] Increased fracture resistance of ceramic systems when metal reinforcement was eliminated has been obtained by the addition of chemical components such as aluminum oxide, leucite, and lithium disilicate.[7,8] For polymer crowns, the improvement was due to higher filler content and the inclusion of multifunctional monomers that increase the cross-linking between the polymeric chains.[9] Their fracture resistance exceeded normal occlusal forces,[10] and finite element analyses (FEA) showed similar stress distributions for both ceramic and polymer crowns.[11,12] Feldspathic ceramic has been shown to be the weakest material among ceramic systems. Incorporation of leucite and lithium disilicate contents has improved the fracture load,[13] although the improvement in the lithium disilicate ceramic was attributed to the particle size and distribution and not necessarily the change in composition.[7] Considering that any restoration has a risk of fracture, it is important to ensure that ceramic crowns have sufficient resistance to support occlusal forces and, in case of fracture, do not compromise the tooth. Clinical studies have reported values for normal occlusal forces.[14] However, different material compositions and unpredictable behavior in clinical situations make it difficult to establish the actual forces on a crown. The strength of restorative materials, represented by fracture loads, is usually determined in laboratory tests. Clinical validity of such failure testing has been questioned.[15] Failure might be a result of design deficiency, material deficiency (fabrication process), or in situ stress-induced conditions.[16] The main critique is that under laboratory conditions, failure begins at the load point, which is different from clinical situations, where crack propagation occurs from the bonded interface toward the crown surface[15-18] due to critical flaws and high tensile stress in the bonded area. Subcritical crack growth, which is crack propagation at load levels lower than the fracture resistance, can also significantly decrease survival time of dental ceramics.[19] Repetitive low-level loading can cause preexisting subcritical flaws to slowly grow until failure occurs at a level of loading insufficient to cause failure of the new prosthesis.[20] In combination with laboratory load-to-failure tests, FEA and fractography are used to evaluate mechanical behavior of restorations and compare the results with clinical conditions. FEA allows investigation of stress distributions,[21-26] and fractography can reveal the origin of a crack and propagation direction and history.[16,17,19,20,23,27,28] Fractography includes the examination of fracture surfaces and the description and interpretation of fracture markings used to understand failure events of brittle materials.[16] For in vitro laboratory tests, bovine roots are commonly used due to their similarity to human dentin[29-31] and the ease in standardizing the dentin substrate.

The aim of the present study was to compare the fracture load of single ceramic and polymer crowns with metal ceramic under compressive loads. The null hypothesis was that the ceramic systems have fracture loads and mechanical behavior similar to the metal ceramic system. The fracture behavior was investigated through two dimensional (2D) FEA and fractographic analysis.

Materials and methods

Comparisons were made among six groups ($n = 8$). Conventional metal ceramic (CMC) crowns with a metal coping (NiCr) and feldspathic ceramic (Noritake) as the control; modified metal ceramic (MMC) crowns, as in the control group, but with aluminum-reinforced ceramic in the buccal cervical area (collarless); ceramic crowns reinforced by lithium disilicate (Empress II) (EMP); ceramic crowns reinforced by leucite (Cergogold) (CERG); ceramic crowns reinforced by leucite fluoride-apatite (d.Sign) (SIGN); polymer crowns (Targis) (TARG). All materials used were manipulated in accordance with manufacturers' recommendations. Composition and manufacturer information are listed in Table 1.

Out of 200 bovine teeth stored in 0.2% Tymol solution for up to a month, 115 with regular and similar roots were selected. They were sectioned with a double-sided diamond disc (KG Sorensen, Barueri, Brazil) to obtain roots 15-mm long. Then, 48 roots with similar volume (cervical mesiodistal diameter of 5 to 6 mm, and buccolingual diameter of 6 to 7 mm), shape, and canal diameter were chosen. Root canals were filled with gutta percha and Fill Canal cement (Technew Com. Ind. Ltda, Rio de Janeiro, Brazil) and then prepared with a 1.59-mm diameter cylindrical rotary cutting instrument (Fast steel bur 1/16, Twill, São Paulo, Brazil) to a depth of 10 mm, cleaned with hydrogen

Table 1 Restorative material characteristics

Material	Composition	Manufacturer (location)	Batch nr.
Durabond MS	Ni-Cr	Marquat S/A (São Paulo, Brazil)	BUE-109
Noritake	Feldspathic ceramic	Noritake Kizai Co. Ltda (Aichi, Japan)	OC823
Vitadur	Aluminum-reinforced ceramic	Vita Zahnfabrik (Bad Sackingen, Germany)	61007
IPS Empress II	Lithium disilicate- reinforced ceramic	Ivoclar Vivadent (São Paulo, Brazil)	912247
Cergogold	Leucite-reinforced ceramic	Degussa (São Paulo, Brazil)	0019/1
IPS d.Sign	Leucite fluoride- apatite* reinforced ceramic	Ivoclar Vivadent (São Paulo, Brazil)	C12729
Targis	Laboratory-processed composite resin	Ivoclar Vivadent (São Paulo, Brazil)	C42167

*According to the manufacturer.

peroxide, washed, and dried with paper points. Cylindrical NiCr (Durabond MS, Marquat S/A, São Paulo, Brazil) cast dowels, 12-mm long, received a circumferential groove to increase mechanical retention for a composite resin core and were cemented in the canal, after being air-abraded with aluminum oxide (N. Martins e Teixeira Ltda, São Paulo, Brazil), with zinc phosphate cement (Vigodent S/A, Rio de Janeiro, Brazil). Cores were prepared with composite resin (Z250, 3M ESPE, São Paulo, Brazil) placed in three increments. Each increment was polymerized for 20 seconds, under at least 500 mW/cm^2 (Optilight II, Gnatus, São Paulo, Brazil) at a distance of 3 mm. Cores were prepared at high speed and finished at low speed using a 2143 diamond rotary cutting instrument (KG Sorensen, São Paulo, Brazil). The characteristics of the preparations were as follows: 6° axial convergence, 1-mm rounded shoulder, and rounded internal angles, based on the study by Burke.[6] The finished core was 3-mm high in the central area and about 3.5-mm high in the cuspal areas. Shoulder width was controlled using the rotary instrument (1.2-mm diameter) as reference, and axial convergence was checked using a device previously calibrated. Impressions of the preparations using polyether (Impregum) were made after 10 days of storage in saline solution to allow hygroscopic expansion of the composite resin core.

Manufacturers' instructions were followed for all restorative systems used, and all restorations had the final form of a premolar with axial thickness of approximately 1 mm in the cervical third and 2 mm in the occlusal surface checked with a caliper. Before cementation, crowns were inspected in a stereomicroscope (20x) (LEICA MS 5, Leica Microscopy Systems Ltd, Heerbrugg, Switzerland). Two SIGN crowns initially cracked were replaced. Restoration bonding surfaces were air-abraded with 50 μm aluminum oxide (N.Martins e Teixeira Ltda) at 2 bar for 10 seconds, followed by ultrasonic cleaning (Thornton Inpec Eletronica Ltda, São Paulo, Brazil) in distilled water for 60 seconds.[1,3,32] Next, nonmetal bonding surface areas were covered with a ceramic primer (Dentsply, São Paulo, Brazil) followed by an adhesive system (Single Bond, 3M ESPE USA, St. Paul, MN), which was polymerized for 20 seconds. Preparation surfaces were conditioned with phosphoric acid (30 seconds), washed, and cleaned, and had the same adhesive system applied and polymerized for 20 seconds.

A dual-cure resin cement (RelyX ARC, 3M ESPE) was applied on the internal aspect of the restoration after manipulation in accordance with the manufacturer's instruction. The restoration was then placed on the preparation and maintained under a static load (500 g) for 5 minutes. After excess cement was eliminated, each surface (buccal, mesial, lingual, distal) was polymerized for 40 seconds. Low-speed silicon abrasive instruments (KG Sorensen, Barueri, SP, Brazil) were used along the margin to eliminate adhesive and/or cement residues. After cementation, all restorations were stored in saline solution at room temperature for 5 days and embedded in autopolymerizing polystyrene resin cylinders (Sales e Antunes Com. e Representações Ltda, Uberlândia, Brazil). An artificial periodontal ligament was reproduced using a polyether impression material (Impregum).[33] Seven days after cementation, the specimens were submitted to compressive loads in a mechanical testing machine (EMIC 2000 DL, São José dos Pinhais, Brazil) at a 0.5-mm/min crosshead speed, using a 6-mm diameter stainless steel sphere attached to a rod. After log transformation to normalize the data recorded at the moment of fracture, they were submitted to one-way ANOVA ($\alpha = 0.05$) and Tukey multiple comparisons tests using the SPSS 12.0 Program (Chicago, IL). The type of fracture was evaluated by one examiner at a magnification of 20x with a stereomicroscope, based on the studies of Burke [6,34] and according to the following classification (Fig 1): type I—cervical fracture/crack; type II—cohesive fracture not involving metal or tooth; type III—cohesive fracture involving any interface; type IV—fracture involving the core (root preserved); type V—Fracture involving root.

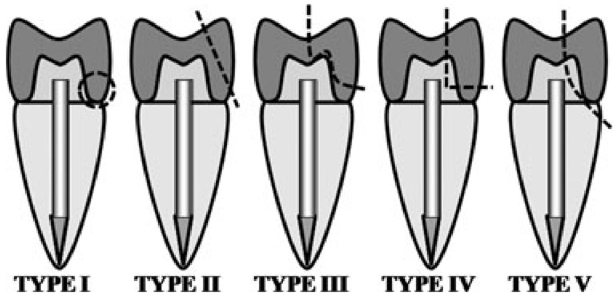

Figure 1 Fracture patterns.

The stress distribution in the tooth-dowel-core-restoration complex under a static load application was evaluated using FEA.[35,36] A 2D numerical model was created from a longitudinal slice of the CMC group, consisting of a premolar with endodontic treatment, metal dowel, composite resin core, and restoration. The same general geometry was adapted for each group to reflect its specific characteristics. The geometry and characteristics of the specimens were defined using computer-aided design (CAD) software (Mechanical Desktop, AutoCAD V6, Autodesk, Barcelona, Spain) (Fig 2A) and exported to FEA software (Ansys 9.0, Ansys Inc., Houston, TX) using IGES format. Two-dimensional finite element models were created by identifying each separate structure and meshing them with quadrilateral plane stress elements (Ansys element type PLANE 183). The material properties applied to the various materials are listed in Table 2.[37–43] All properties were assumed isotropic, linear-elastic, and homogeneously distributed. Perfect adhesion was assumed between the structures (crown/cement, cement/core, core/dowel, dowel/cement, and cement/dentin). Two static loads of 45 N were applied perpendicular to the occlusal surface, simulating the contact loads with the tooth-sphere in the experiments (Fig 2B). Model movements were restricted in all directions at the external lateral outline and base of the cylinder. Maximum principal stresses were used to evaluate the stress state (Fig 2C). Additionally, stress values were recorded at 11 points along the cement/core interface (Fig 3). For fractographic analysis three steps were followed:[44] (1) visual inspection to select specimens containing visible markings of fracture, (2) examination under stereomicroscope magnification (10x to 40x) to select areas of interest for further investigation under the SEM, (3) SEM examination under high-resolution close-ups of the regions of interest. The stereomicroscope examination started at the mesial cervical area moving toward the occlusal and finishing at the distal

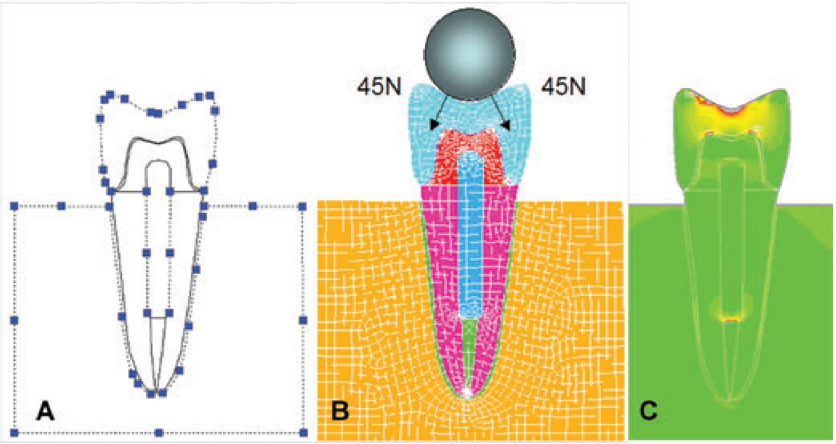

Figure 2 General representative definitions of 2D numerical models obtained for all groups: (A) External and internal contours; (B) Finite element mesh of numerical model showing mechanical properties of each structure and static load application; (C) Stress distribution (maximum principal stress).

cervical area. After stereomicroscope examination, the fractured specimens were goldsputter coated with gold/palladium in high vacuum (SCD 050, Bal-tec AG, Balzers, Liechtenstein) for SEM observation (JSM – 5600, JEOL Ltd. Tokyo, Japan).[20,28] Failure markings from representative fracture surfaces such as arrest lines, hackles, and wake hackles were searched.[27,45] An arrest line is a well-defined line produced when a crack comes to a halt before resuming its propagation, often in a slightly different direction.[17] Arrest lines are also indicators of the direction of propagation, as the beginning of a crack event is always located on the concave side of the first arrest line. Hackles are lines on the fracture surface that run in the local direction of cracking and are commonly formed when a crack moves rapidly.[17] They separate parallel portions of the propagating crack on slightly different planes. Wake hackle is a trail (wake) emanating from a pore (or other irregularity) and is created by the crack front advancing along the sides of the pore before continuing on slightly different planes.[17] Thus, wake hackles are indicators of the direction of crack propagation.

Table 2 Mechanical properties of tooth structures and materials used in the FEA: Elastic modulus (E) and Poisson's ratio (ν)

Material	E (GPa)	ν
Zinc phosphate cement[33]	13.7	0.33
Dentin[34]	18.6	0.30
Gutta percha[34]	0.00069	0.45
Polyether[35]	0.0689	0.45
Polystyrene resin[35]	13.7	0.30
Laboratory-processed resin[35]	18.8	0.24
Aluminum-reinforced ceramic[36]	418	0.22
Luting resin cement[37]	5.1	0.27
Feldspathic ceramic[37]	69	0.30
Lithium disilicate ceramic[38]	120.0	0.25
Leucite-reinforced ceramic[38]	65	0.23
Composite resin[39]	16.6	0.24
Ni/Cr alloy*	200.0	0.30

*According to the manufacturer.

Results

Based on the data and normal distribution after log transformation, one-way ANOVA showed significant differences among the groups (Table 3). Table 4 summarizes the means of fracture load, results of Tukey ($p < 0.05$) multiple comparisons, and type of fracture for all groups. The fracture loads of metal ceramic groups (CMC and MMC) were significantly greater than those of metal-free groups (EMP, CERG, SIGN, and TARG). Figure 4 shows a comparison of minimum, mean, and maximum fracture resistance values (N) for all groups tested.

In the CMC and MMC groups only 3 out of 16 fractures involved the root, and no displacement of the metal coping

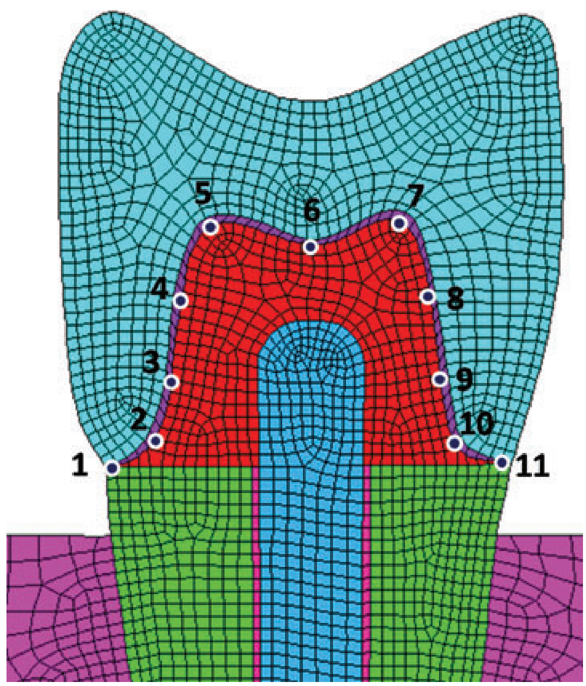

Figure 3 Schematic representation of 11 stress measuring points along the cement/core interface.

Table 3 Results of one-way ANOVA for fracture load (log-transformed data)

	Sum of square	DF	Mean square	F	Sig.
Between groups	11.660	5	2.332	37.876	0.000
Within groups	2.586	42	0.062		
Total	14.246	47			

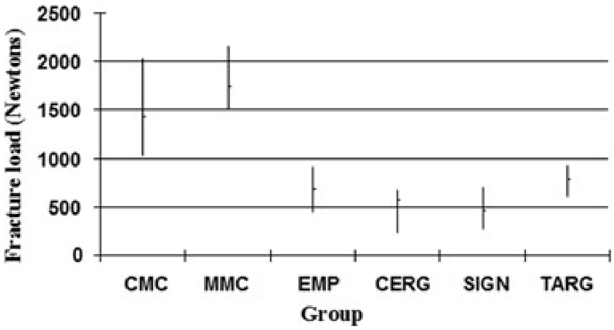

Figure 4 Minimum, mean, and maximum fracture resistance values of all groups. Horizontal line indicates mean values.

was observed; however, fractures in all specimens from groups EMP, CERG, and SIGN were extensive, with displacement of a cusp, and involved either the core (type IV) or the root (type V). All fractures of group TARG involved the root in a catastrophic manner (type V). Type I and type II fractures were not observed.

The stress distributions in Figure 5 show tensile stress concentration areas under and between the load points, indicated by orange, red, and light-gray (indicating values higher than 10 MPa). The restorations in the metal ceramic and ceramic groups had similar (tensile) stress distributions. For the polymer crowns, tensile stress concentration areas were observed throughout the crown-core-dowel-tooth complex. Small areas of tensile stress concentrations were observed at the top areas of the resin core in the CMC, MMC, and EMP groups and in the area around the lower portion of the dowel for all groups. Figure 6 shows the comparison among the stress levels measured at the points along the cement/core interface, indicating higher stresses under the polymer crown specimens. Specimens with a higher elasticity modulus presented lower stress levels at the cement/core interface.

Fractographic analysis indicated fracture origin at the load point and propagation from the occlusal surface toward the cervical area in most restorations. Figure 7 shows a representative SEM of crack origin and propagation for a polymer and a ceramic crown. The probable failure origin noted in the fractographic analysis coincided with the stress concentration area at the occlusal load point observed in the FEA. Many specimens did not show signs that allowed a reliable fractographic determination of the failure origin.

Discussion

The data showed significant differences of fracture load among the groups and resulted in the rejection of the null hypothesis that the metal-free systems would present fracture loads similar to the metal ceramic system. When selecting a restorative material, the first criterion is sufficient fracture resistance to support masticatory forces and protect the remaining dental structure.[11] The fracture load of the final restoration is the result of the combined effects of bonding between the underlying tooth, the ceramic restoration, and the resin composite cement.[34] In vitro tests are the primary methods used to investigate the fracture strength of restorations, but different methodologies used in different studies, such as the mode and direction of load application, crosshead speed, fracture mode, and root embedding can result in different outcomes,[33] making any comparison difficult. The present study was carefully designed to minimize methodological effects on the final results while simulating clinical conditions. The fracture resistance of ceramic crowns can be increased if a castable or reinforced material is used,[8] as shown by the results of the present study, in which leucite and lithium disilicate reinforcement increased the fracture load when compared to feldspathic ceramic. Leucite-reinforced ceramic crowns were reported to have high fracture resistance, and their fracture tended to involve the underlying tooth.[8] This relationship was also observed in the present study for all-ceramic crown groups in which all fractures involved either the core (type IV) or the tooth (type V) in a proportion of 41% and 58%, respectively. Polymer crowns yielded high fracture load values but with 100% catastrophic fractures (type V). In the metal ceramic groups the low incidence of dental involvement without restoration displacement when the fracture occurred indicates the protective effect of the metal coping for the tooth. When submitted to vertical loads, the stress was concentrated around the loading point for both metal ceramic and ceramic crowns, and the low stress found in the cervical region (Fig 5) may explain the nonoccurrence of fracture type I. But considering that under vertical loads the stress concentrates around the load point, the type of fracture from the ceramic groups in the present study, involving either the core or the root with displacement of the cusp, suggests that the cracks began at the load point, propagated through the tooth/restoration interface and then involved the dental structure because of the strong adhesive bonding at the interfaces. Also, static loads provide no clues about the long-term fracture resistance of crowns under fatigue loading. The use of cyclic loading resulted in the

Table 4 Mean values (± SD), Tukey comparisons ($p < 0.05$) for fracture load and type of fracture from the groups. Different lowercase letters indicate significant differences

Group	Mean values (N) and SD	Type of fracture I	II	III	IV	V
MMC	1691 ±236[a]	0	0	6	0	2
CMC	1383 ±298[a]	0	0	7	0	1
TARG	749 ±113[b]	0	0	0	0	8
EMP	657 ±153[b]	0	0	0	4	4
CERG	546 ±149[bc]	0	0	0	3	5
SIGN	443 ±126[c]	0	0	0	3	5

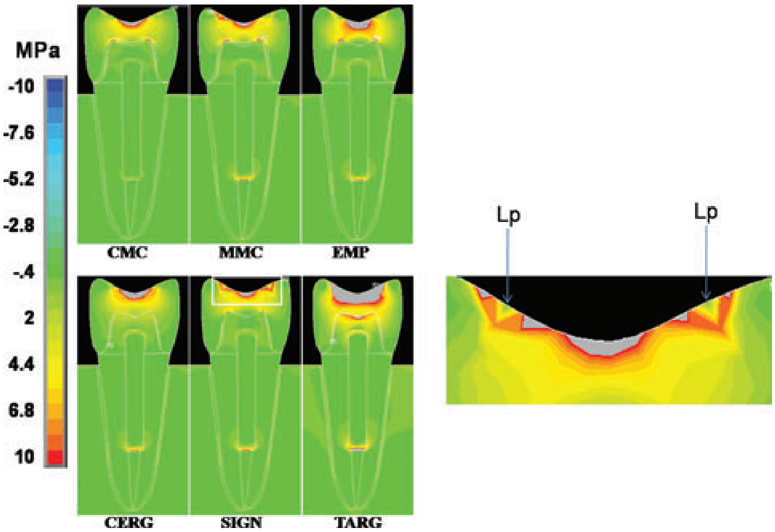

Figure 5 Stress distributions (maximum principal stress) according to a linear color scale. Negative and positive values indicate compressive and tensile stresses, respectively. The light-gray areas are locations where stress concentrations exceed 10 MPa. The framed detail (SIGN group) shows the load points (Lp), indicated by the two arrows, where compressive stresses (small green area) are surrounded by tensile stresses (yellow and red areas), the probable fracture origin.

largest decrease in mechanical properties[7] and could result in data closer to clinical conditions. Nonoccurrence of cohesive fracture without involving any interface (type II) could indicate that the cohesive strength of the materials surpasses the adhesive strength under the applied loading conditions.

Tensile stresses tend to be more critical than compressive stresses for ceramic materials, since their tensile strengths are often an order of magnitude lower than their compressive strengths. Therefore, maximum principle stresses were chosen to assess the stress state in the FEA. If tensile stresses are present, this is the highest tensile component of the three principal stress components. The FEA gave insight into the general stress distribution; however, the strength of a restoration is strongly affected by the presence of flaws or other microscopic defects. Depending on the flaw population within the material, failure may occur at loads lower than the maximum value derived for ideal materials.[25] Although experimentally determined strength values in the present study may not match the quantitative stress levels shown in the FEA because flaws or minor defects within the materials were not modeled, FEA can provide valuable insight for consideration in clinical practice because the general stress distribution pattern is not affected by microscopic defects. Tensile stress concentration at the cementation surface of the ceramic layer was suggested to be the predominant factor controlling ceramic failure.[18] These stresses had a higher potential to cause damage to restorations and dental tissues,[26] which could lead to the fracture origin and propagation from that surface as previously reported.[15-18] But in the present study, FEA showed lower tensile stress levels at the cementation surface than in the area under and between the load points at the buccal and lingual cusps, which could explain the occlusal to cervical direction of fracture seen in the fractographic analysis (Fig 7).

Mechanical properties, geometry, and thickness of the restorative material can directly influence the load distribution in a tooth/restoration complex.[35,36,40] The calculated maximum principle stresses (Fig 5) seem to have an inverse relationship to the elastic modulus of the crown materials. Groups CERG, SIGN, and TARG had higher tensile stress concentrations than the CMC, MMC, and EMP groups, even around the dowel

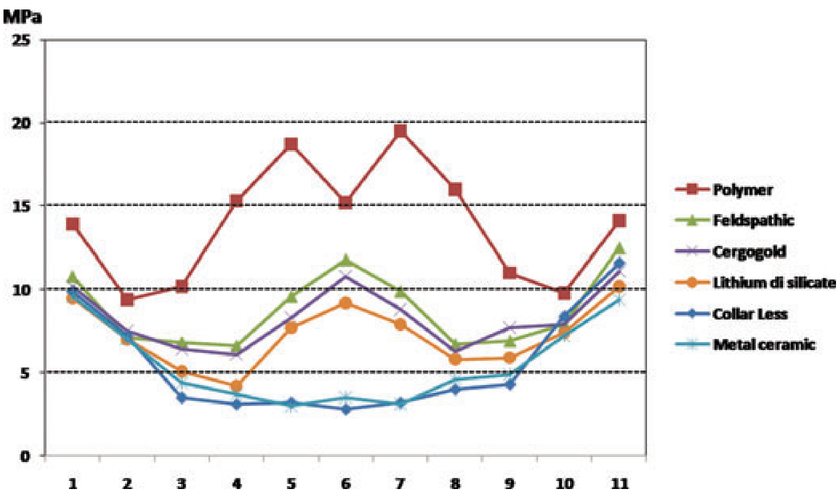

Figure 6 Comparison of the 11 stress measurements along the cement/core interface for all materials.

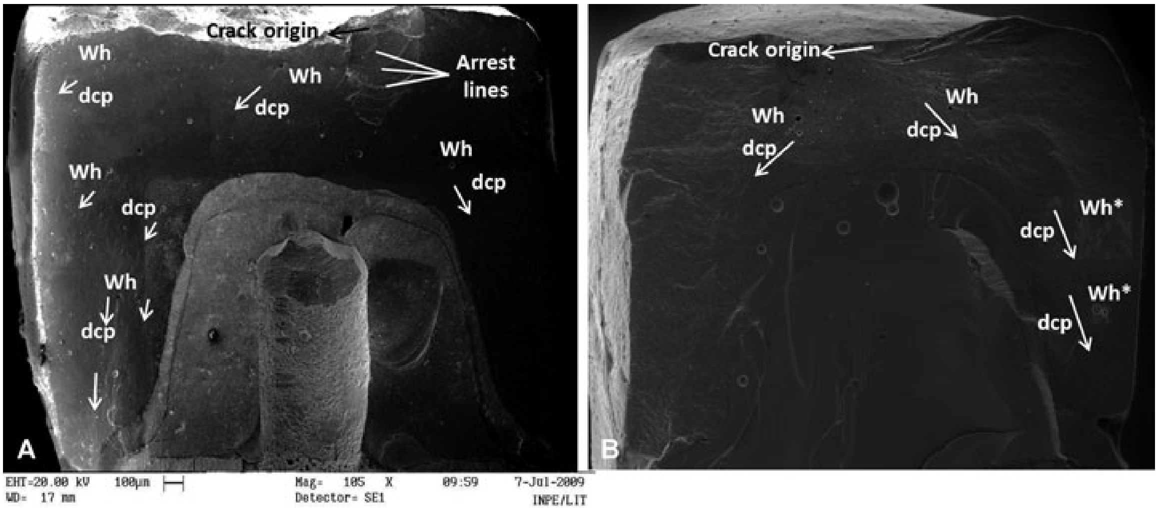

Figure 7 Representative sketch of the crack origin and propagation: (A) Polymer type V fractured specimen; (B) Ceramic type IV fractured specimen. An occlusal edge chip delimited by arrest lines (not seen in B) indicates the probable fracture origin and the initial propagation follows the concavity of the arrest lines. Several wake hackles (Wh) are visible starting from pores within the ceramic restoration indicating that the direction of crack propagation was from occlusal to cervical (black arrows). Wh* are micro wake hackles observed in a higher magnification than the ceramic specimen, as they did not show as clear markings as polymer crowns did.

apical region. Materials with a high elastic modulus, such as ceramics in general, tend to carry more of the load in a dental structure[21] as shown by the stress-level quantification (Fig 6). Thus, the risk of dental fracture may be minimized if the stress concentration in the ceramic structure results in restorative material fracture before the tooth is compromised. Conversely, although the polymer crown had a higher fracture load than ceramics, due to its lower elasticity modulus, a larger amount of the occlusal load was transferred through the tooth structures, resulting in catastrophic fracture for all specimens. This behavior has been previously reported.[4] The catastrophic type of fracture from group TARG would certainly result in tooth loss.

Although polymer crowns demonstrated sufficient fracture load to carry the occlusal forces, their fracture behavior can limit the use of polymer crowns when compared to ceramic systems.[10] Fracture load studies of crown systems, within their limitations, provide some idea of the load-bearing capacity of crowns in simulated clinical situations.[10] Under clinical conditions, the failure of brittle materials is governed by some well-known variables.[12] Data collected from laboratory tests have been considered clinically invalid as a result of an incorrect stress state, failure occurring from contact damage flaws instead of "natural" cementation surface flaws, the production of a great number of fragments, and failure loads too high for clinical significance.[15] In the current study, the fracture origin from occlusal to cervical direction and the number of fragments produced were also at odds with clinical reports,[15,16] which report that clinical failures begin from the cementation surface producing one or two fragments; however, a fracture analysis of another clinical study[28] found an occlusoapical direction toward the gingival margins of a crown. In the current study fractographic analysis indicated the fracture origin at the load point for most of the specimens, even though clinical conditions were closely reproduced. The association of the defect found at the load point with the arrest lines, and the characteristics of the hackles and wake-hackles observed in the specimens indicated an occlusal-to-apical propagation direction (Fig 7). Optical and SEM investigation of the propagation direction seemed to be easier to determine than the failure origin, as has been previously reported[20,28] because some specimens did not show clear signs of the crack origin. Fracture markings such as arrest lines and wake-hackles were clearly observed in the polymer crowns, but in ceramic specimens, besides the chipping area, only wake-hackles were present. Few ceramic specimens showed hackles. In the present study, the fracture origin and propagation might be explained by correlating details from FEA and fractographic analysis. Figure 5 shows a mesial view of the restored teeth, and Figure 7 shows the fractography images of a buccal view of the fractured specimens. As a result of the initial compression stress, right below the load point a small area of tensile stress concentration is generated (frame detail, Fig 5) where the material chipping and crack origin also occurred (Fig 7). As a consequence of the progressive loading, a wider tensile stress area between the load points at the buccal and lingual cusps is created and, once reaching the crack origin, the crack propagates in the cervical direction (Fig 7). In clinical situations, repetitive low-level loading may cause a slower propagation of preexisting flaws in the cementation surface toward the outer surface, which may explain the different failure behavior of the restorations.

Assumptions and simplifications required for conducting the in vitro experiments and numerical analysis introduced several limitations in this study. Bovine teeth were used due to the limited availability of human teeth; however, the bovine permanent incisors, besides being easily obtained, improved the standardization of root measures, while previous studies have shown that properties of bovine teeth are similar to human teeth.[29–31] Also, a biological structure was preferred to metal, because bovine roots can better reproduce the actual stress distribution occurring in crowns cemented on natural teeth. Other limitations included a small difference of bonding area among the specimens; the loading forces were purely occlusal and static; the thickness of the luting agent was not controlled; aging techniques were not used; and the 2D FEA had some limitations compared to 3D analysis. In the FEA, the boundary conditions assumed perfect bonding, which can be considered a simplification of reality; however, the general FEA results were validated by the fractographic observations. Although results showed a high incidence of dental involvement, further studies are necessary to investigate whether the same behavior occurs under clinical conditions. Future research should investigate the biomechanical behavior of restorations under the influence of different dowels, cores, and cements using nondestructive tests such as extensometry to quantitatively measure stress and microstrain in the specimens.

Conclusions

Within the limitations of this study, the fracture loads of reinforced ceramic and polymer crowns were significantly lower than those of metal ceramic materials; however, reinforcement appeared to increase fracture load in the ceramic systems, providing sufficient resistance to support normal occlusal forces. Ceramic restorations showed higher incidence of fracture with tooth involvement than metal ceramic restorations. All polymer crown restorations were catastrophic for the tooth when type of fracture was considered. For all groups, the type of fracture corresponded to the stress distributions calculated with FEA and fractographic analysis. The latter indicated the fracture origin at the occlusal load point and the propagation in the occlusal to apical direction, the opposite direction of what has been reported in clinical fractures. Results showed a correlation among the mechanical test, fracture pattern, finite element, and fractographic analyses.

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