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RESEARCH AND EDUCATION

Mechanical performance of implant-supported posterior crowns

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Implant-supported crowns are commonly used to replace missing teeth. These restorations have a good clinical survival rate and long-term patient satisfaction.[1,2] However, implant-supported restorations show a lower success rate than tooth-supported restorations, mainly because of mechanical complications.[3-5] Natural teeth are connected to the periodontal ligament, while osseointegrated implants are connected directly to the bone. As a consequence, the periodontal ligament cannot function as an elastic buffer, which offers tactile sensitivity.[6-8] This is an important reason why fracture of dental porcelain (chipping) is a common problem in implant-supported crowns.

ABSTRACT

Statement of problem. The fracture of implant-supported restorations, especially of the veneering layer, is a common problem in dentistry. Monolithic ceramic or resin restorations might help solve this problem.

Purpose. The purpose of this study was to obtain additional insight into the risk of fracture of implant-supported restorations.

Material and methods. Identical crowns (n=10) of 10 different ceramic and composite resin materials were cemented on conventional abutments on implant replicas embedded in polymethyl methacrylate blocks. The specimens were subjected to compressive load in a universal testing machine to record initial load to failure (ILF). Additionally, the flexural strength (FS), compressive strength (CS), and elastic modulus (E) of the investigated materials were determined. These results were used in a finite element analysis model of a composite resin and a lithium disilicate crown.

Results. Anatomic contour zirconia (Lava Plus) crowns had the highest ILF (6065 N), followed by lithium disilicate (IPS e.max) (2788 N) and the composite resin materials (Protemp 4, Majesty Flow, Telio CAD, Estenia C&B, Lava Ultimate, VITA Enamic) (2386 to 1935 N). Veneered zirconia (Lava) crowns showed the lowest ILF (1477 N). The highest FS, CS, and E were found for Lava Plus and IPS e.max. No direct relationship was found between ILF and the FS, CS, or E. The finite element analysis showed stresses that did not exceed the FS or CS of IPS e.max. The surface roughness of these crowns might have caused initial failure at relatively low stresses.

Conclusions. In this laboratory study, monolithic implant-supported crowns showed a higher ILF than conventional veneered ceramic crowns. Monolithic ceramic restorations might perform better than composite resin crowns. (J Prosthet Dent 2015;■:■-■)

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Clinical Implications

Anatomic contour ceramic crowns can withstand higher forces in the oral cavity than conventional veneered ceramic crowns. Although composite resin crowns still do not match the monolithic ceramics, the resins with a low elastic modulus offer promising results.

Crowns with a zirconia core show more chipping than metal ceramic crowns.[4,9] A decrease in chipping is expected with better understanding of this material, improved framework design,[10,11] veneering techniques,[12] and slower cooling.[13] Another way to prevent chipping of the veneering porcelain is to avoid the problems related to this weak veneering layer (flexural strength 30 to 100 MPa)[14] with monolithic restorations made entirely of zirconia. Beuer et al[15] showed in a laboratory study that anatomic contoured zirconia crowns demonstrated higher resistance to static loading tests than veneered zirconia crowns. Monolithic lithium disilicate restorations also show good results in laboratory studies on implants as well as in clinical studies on teeth.[16-18]

In contrast to ceramics, resin-based restorative materials are more elastic and appear to compensate for the absence of the damping effect of the periodontal ligament.[8,19,20] Composite resins have been widely and successfully used as a direct restorative material and also show promising clinical results for indirect restorations on teeth.[21-23] Within their limitations, laboratory studies show that composite resin applied as a veneering material on a metal or zirconia substructure leads to results similar to those of veneering porcelain.[24,25] Although these studies are promising, the mechanical properties of composite resin still do not match those of currently available ceramics.[26-29] Clinically, ceramic crowns perform better than composite resin crowns.[30]

Composite resin crowns are often based on bis-acryl (Bis-GMA) or polymethyl methacrylate (PMMA).[31] PMMA-based resins appear to be more color stable than autopolymerizing bis-acryl composite resins like Protemp 4 (3M ESPE).[32] Highly filled composite resins like Estenia C&B (Kuraray Dental) show the highest resistance to wear.[33] Manufacturers of new resin-based materials developed especially for computer-aided design and computer-aided manufacturing (CAD/CAM) application claim to combine the features of resin and ceramics, examples of which are the particle-filled composite Lava Ultimate (3M ESPE) and the interpenetrating phase composite VITA Enamic (VITA Zahnfabrik).

The purpose of this study was to evaluate the risk of mechanical failure of implant-supported restorations by means of fracture and chipping. A broad spectrum of possible materials was subjected to a variety of laboratory tests. The flexural and compressive strength and the elastic modulus of the materials were determined, and crowns on implant abutments were subjected to a static loading to determine the initial load to failure. Additionally, these mechanical properties were used in a finite element analysis (FEA) model for a better understanding of the failure mode and strength of the total structure.

MATERIAL AND METHODS

The initial load to failure (ILF), flexural strength (FS), elastic modulus (E), and compressive strength (CS) were determined. Additionally, FEA was performed to determine the failure modes of the crowns. Materials used in this study are summarized in Table 1.

Ninety identically shaped crowns of a mandibular left first molar were made on a prefabricated titanium abutment with an external connection for a 5-mm diameter implant (Esthetic abutment; Nobel Biocare) (Figs. 1, 2). The first crown was designed in the dental laboratory, and the occlusal pattern was adjusted to the shape of the metal antagonist used in the ILF test.

The crowns of groups Telio CAD (TC), Lava Ultimate (LU), VITA Enamic (VE), IPS e.max (EM), veneered Lava (LV) substructure, and Lava Plus (LP) were digitally designed as identical copies of the first crown. The LU, LV, and LP were milled (Lava CNC 500; 3M ESPE). After milling, the zirconia LV and LP substructure were sintered in a furnace (Lava Therm; 3M ESPE). The LV crowns were veneered by pressing a homogenous layer of fluorapatite glass-ceramic (e.max ZirPress; Ivoclar Vivadent) on the substructure. All crowns had an identical occlusal pattern. TC, VE, and EM were milled (RXD-5; Röders GmbH), and the EM crowns were sintered and glazed in a porcelain furnace (Programat CS; Ivoclar Vivadent) according to the e.max sintering program.

Protemp 4 (PT), Clearfil Majesty Flow (MF), and Estenia C&B (ES) crowns were made by duplicating the shape of the CAD/CAM crown by using a mold of polyvinyl siloxane putty (Flexitime; Heraeus Kulzer) combined with a light-body impression material (Flexitime Correct Flow; Heraeus Kulzer). The occlusal part of the mold was made of translucent polyvinyl siloxane impression material (Memosil 2; Heraeus Kulzer) for photopolymerization. The PT crowns were chemically polymerized in this mold. The MF and ES crowns were photopolymerized for 20 seconds by blue light. Afterward, the ES crowns were heat polymerized (100°C for 15 minutes) and photopolymerized (Heat-Curing-110; Toesco and Lumamat 100; Ivoclar Vivadent) and polished according to the manufacturer's manual.

For the ILF test, 10 prefabricated abutments were screwed on a 5 mm diameter stainless steel implant replica (Brånenmark System WP; Nobel Biocare). The replicas were embedded in identical positions in a block of PMMA (Vertex Dental) by a duplication method using

Table 1. Material composition and indication according to manufacturer's data

Code	Material	Composition[a]	Manufacturer	Indication
PT	Protemp 4	Bis-GMA, UDMA, TEGDMA, Bis-EMA, 50 nm silanized amorphous silica	3M ESPE	Interim direct in restorations
MF	Clearfil Majesty Flow	Hydrophobic aromatic dimethacrylate, TEGDMA, 3 μm, silanated barium glass + 20 nm silanated colloidal silica	Kuraray Dental	Flowable direct restorations
TC	Telio CAD	PMMA	Ivoclar Vivadent	Interim indirect restorations
ES	Estenia C&B	Bis-GMA, UDMA, decandiol dimethacrylate, 2 μm, surface treated alumina, silanated glass ceramics	Kuraray Dental	Indirect restorations
LU	Lava Ultimate	20 nm silica filler, 4 to 11 nm zirconia filler, silica-zirconia filler	3M ESPE	Indirect restorations
VE	VITA Enamic	Silicon dioxide (60%), aluminum oxide, sodium oxide, UDMA, TEGDMA	VITA Zahnfabrik	Indirect restorations
EM	IPS e.max CAD	0.2 to 2 nm lithium disilicate glass ceramic	Ivoclar Vivadent	Indirect restorations
LV	Lava frame and CAD/CAM layered veneer of IPS e.max ZirPress	Tetragonal polycrystalline zirconia, partially stabilized with 3 mol% yttria, fluorapatite glass-ceramic	3M ESPE, Ivoclar Vivadent	Indirect restorations
LP	Lava Plus	Tetragonal polycrystalline zirconia, partially stabilized with 3 mol% yttria	3M ESPE	Indirect restorations

[a]Bis-GMA, bis-phenyl glycidylmethacrylate; UDMA, urethane dimethacrylate; TEGDMA, triethylene glycol dimethacrylate; Bis-EMA, ethoxylated bis-phenol-A-dimethacrylate; EBPADMA, ethoxylated bis-phenol-A-dimethacrylate; PMMA, polymethyl methacrylate.

Figure 1. Arrangement used for ILF test.

a polyvinyl siloxane putty mold. The distance from most coronal resin to the top of the implant replica was 3 mm. All abutments were roughened above the restoration outline by airborne-particle abrasion with aluminum oxide (50 μm). According to the manufacturer's instructions, the abutments were screwed on the replicas with a torque of 35 Ncm, and the screw access hole was sealed by a cotton ball and temporary filling material (Cavit; 3M ESPE). The crowns were cemented with an adhesive resin cement (RelyX Ultimate; 3M ESPE), and the internal surfaces of the crowns were pretreated with bonding agent (Scotchbond Universal; 3M ESPE) according to the manufacturer's instructions. After the crown had been placed, the cement excess was removed, and the cement was photopolymerized for 20 seconds by blue light on the buccal, occlusal, and lingual surfaces. All specimens were kept at 100% humidity and 37°C for 24 hours. To test ILF, the specimens were placed in a universal testing machine (Instron 6022; Instron) with a load cell of 10 kN at a crosshead speed of 1 mm per minute. A stainless steel ball with a diameter of 12 mm was used to apply the load on 4 cusps of the crowns (Fig. 1). The failure load was recorded in newton. After testing, all cement and crown remnants were removed from the abutments and photographed to analyze the origin of the fracture.[34,35] The abutments were cleaned with ethanol and prepared for another test group.

The FS and the E were determined by using a 3-point bending test according to ISO standard 4049:2009[36] on 20 specimens of each material. In a mold of 2.0×2.0×25.0 mm, the PT specimens were made by chemical polymerization. ES and MF specimens were systemically photopolymerized on both sides for 20 seconds each time, with 50% overlap. ES specimens were subsequently photopolymerized and heat polymerized. Specimens of LU, TC, and VE were cut from a prefabricated CAD/CAM block. LP specimens were cut out 20% larger from the presintered CAD/CAM block and were then sintered. These specimens had a size of 1.0×1.0×12.0 mm. After setting, all specimens were kept wet at 37°C for 24 hours. According to the specimen size, the 3-point bending test was carried out at a span of 20.0 or 10.0 mm in a universal testing machine (Instron 6022; Instron), with a load cell of 500 N and at a speed of 1 mm per minute. The FS and E were calculated according to the following equations:

$$FS = \frac{3\mathrm{Fl}}{2\mathrm{bh}^2}$$

$$E = \frac{\Delta \mathrm{F}}{\Delta \mathrm{z}} \times \frac{\mathrm{l}^3}{4\mathrm{bh}^3},$$

where F is the recorded force (N), l is the length of the specimen, b is the width of the specimen, h is the height

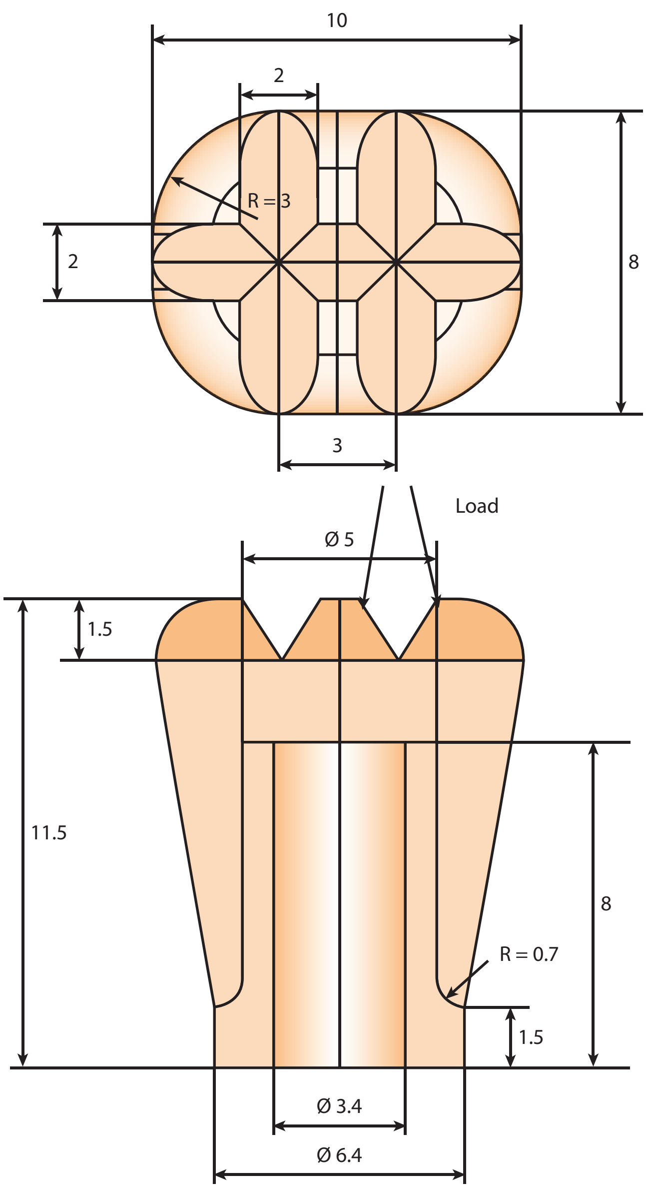

Figure 2. Digital design of crown on abutment as used in static loading test for FEA (dimensions in mm).

of the specimen, ΔF is the difference in normal force between 2 load points, and Δz is the difference in deflection values at the respective load points.

As shown in earlier studies, the strength of ceramics is compromised by surface roughness,[37,38] and most cracks of fractured crowns originate from the internal surface. Because it is difficult to polish or glaze the intaglio, we also measured the FS of 20 EM specimens roughened by air abrasion. The surface roughness of both specimens was determined by profilometry (SJ-400 Profilometer; Mitutoyo).

The CS was determined by testing 10 cylinder specimens of each material with a height of 5 mm and a diameter of 3.5 mm. The specimens of PT, MF, and ES were made in a mold and respectively chemically or light and heat polymerized for 20 seconds from all sides. The LU, TC, VE, and EM specimens were drilled out of the blocks with a water-cooled, hollow-cylinder drill. The EM specimens were sintered and glazed in a porcelain furnace (Programat CS; Ivoclar Vivadent). After measuring each diameter, the cylinders were vertically loaded with a flat node in a universal testing machine (Instron 6022; Instron) with a load cell of 10 kN at a speed of 1 mm per minute. The CS was calculated according to the following equation:

$$CS = \frac{F}{\pi r^2}$$

One-way ANOVA with post hoc least significant difference analysis (PASW Statistics 20.0; SPSS, IBM) was used to analyze differences in ILF, FS, E, and CS (α=.05).

Three-dimensional FEA simplified models of the test parameters for the abutment with crown were created. Two models were made: in the first model, the adhesive interface between the abutment and the crown was designed to be strong enough to resist the shear stresses in the interface abutment/crown (model 1), and in the second model, the surface in the interface between the abutment and the crown was modeled as contact surfaces with a friction coefficient of 0.45 (model 2). Using the symmetry, half models were made in order to facilitate the boundary conditions, with the nodes in the centric plane allowed sliding in the surface only. The finite element modeling was carried out with software (FEMAP 10.1.1; Siemens PLM software), and the analysis was done with software (NX Nastran; Siemens PLM Software). The dimensions of the abutment with crown are provided in Figure 2. The models were composed of 57 395 parabolic tetrahedron solid elements. To test 2 different types of materials, the material properties of Protemp 4 (PT) and IPS e.max (EM) were used in this analysis to model the crown. The calculated E from the 3-point bending test (Table 2) and a Poisson ratio of 0.3 was used.[39] An E of 107 GPa was used for titanium.[39] The nodes at the bottom of the abutment were fixed, and no movement was allowed in any direction. The crown was loaded on the nodes on the occlusal surface, simulating the contact points of the metal antagonist, used in the initial failure load test (Fig. 1). Calculations of the principal and normal stresses were made at the load of fracture values found in the ILF test of this study.

RESULTS

The highest ILF was observed for the LP crowns, followed by EM (Table 2). PT and MF crowns had a significantly higher fracture load than VE, ES, TC, and LU, which did not differ significantly among each other. The lowest ILF was found for LV. Additionally, the LV

Table 2. Mean initial load to failure (ILF), compressive strength (CS), flexural strength (FS), and elastic modulus (E)

Characteristic	ILF (N)	CS (MPa)	FS (MPa)	E (GPa)
PT	2386 (405)^{bc}	312 (93)^{cde}	79 (8)^{ef}	1.7 (0.2)^{efg}
MF	2334 (397)^{cd}	316 (52)^{cde}	140 (10)^{de}	7.9 (0.5)^{defg}
TC	1978 (161)^{cd}	431 (25)^{b}	136 (14)^{de}	2.9 (0.2)^{efg}
ES	1998 (405)^{cd}	300 (65)^{de}	114 (20)^{def}	14.0 (1.6)^{de}
LU	1935 (217)^{c}	516 (30)^{a}	198 (31)^{c}	13.1 (1.8)^{de}
VE	2171 (307)^{cd}	370 (88)^{ce}	131 (15)^{de}	26.3 (3.1)^{c}
EM (smooth)	2788 (488)^{bc}	-	301 (76)^{b,*}	69.7 (14.3)^{b}
LV	1477 (487)^{e}	-	-	-
LP	6065 (966)^{a}	-	1235 (151)^{a}	113.1 (14.4)^{a}
EM (rough)	-	-	249 (36)*	-

Lowercase superscripted letters reflect groups with significant difference.
*Statistically significant difference with other EM.

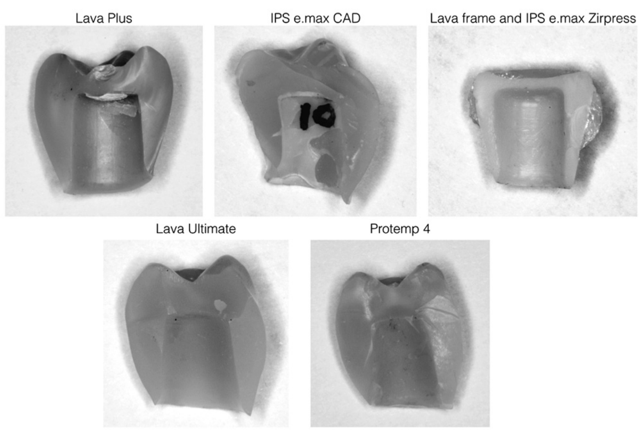

Figure 3. Remnants of fractured specimens after ILF test. Mist and hackle area in occlusal part of fractured LP and PT crowns demonstrate that fracture originated from fissure.

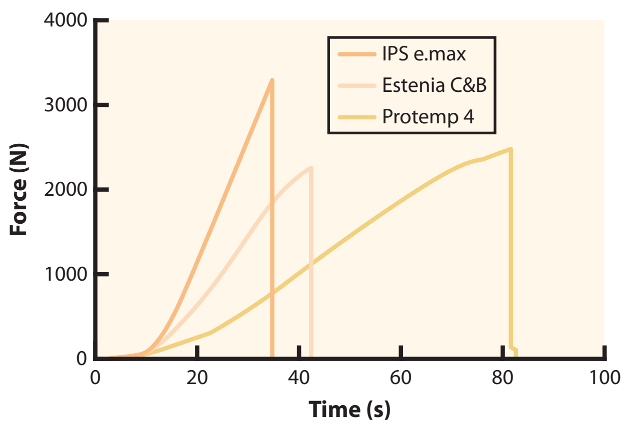

Figure 4. Stress-time curve of ILF experiment.

crowns showed a different fracture pattern than the other crowns. In these specimens, only a part of the ceramic veneer broke off the zirconia core, whereas the other crowns mostly showed a fracture of the abutment. Only the LU and VE crowns showed some minor chipping before cusp fracture. This cusp fracture was used in the results. Remnants of specimens, fractured after ILF test are shown in Figure 3, while the stress-time curve of the 3 different materials is shown in Figure 4. No fracture of the abutments or abutment screws occurred.

In addition to the ILF, the mechanical properties of the materials were also measured. The FS of anatomic contour zirconia was significantly higher than that of the other materials (Table 2), followed by lithium disilicate, whose FS was also significantly higher than that of the composite resins. No significant difference was found among LU, MF, TC, VE, and ES. These materials had a higher FS than PT, which on its own does not significantly differ from ES. A significant decrease of the FS was found with an increased surface roughness for EM. The roughness for the internal surface of EM crowns was Ra=4.3 μm. The glazed specimens for the FS test were Ra=0.08 μm and for the roughened specimens Ra=1.07 μm.

The E was calculated from these results and used for the FEA. LU had the highest CS, followed by TC. The CS of LP and EM could not be tested because the CS of these materials exceeded the capacity of the universal testing machine.

Table 3 shows the vertical shear stresses (solid Y normal stresses) in the model with fixed connection between the abutment and crown (Fig. 5). Table 3 also summarizes the maximum compressive stress (solid minimal principal stress) and the maximum tensile stress (solid maximum principal stress) in the model with contact surfaces. The maximum shear stresses in the titanium (solid Y normal stress) in the vertical interface between the abutment and the crown for the PT crown with fixed connection is 315 MPa. This stress would be comparable with a shear stress of 36 MPa in the (not modeled) cement layer.

Figure 6 shows the solid minimum principal stress for the EM crown with contact surfaces in the interface abutment/crown. The negative values represent the

Table 3. Results of finite element analysis

Model	Description	Protemp	E_{max}
Model 1	Solid Y normal stress (MPa)	640	315
Model 2	Solid minimum principal stress (MPa)	-247	-260
	Solid maximum principal stress (MPa)	71	80

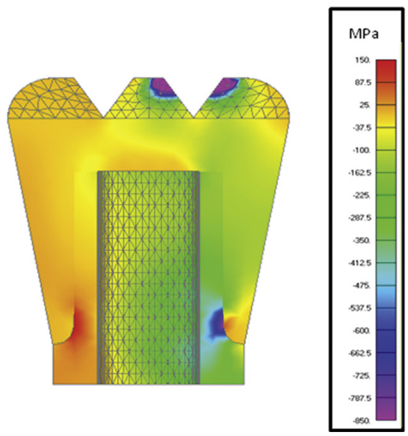

Figure 5. Solid Y normal (vertical) stress on lithium disilicate crown fixed to titanium abutment with peak in cervical area (model 1).

amount of compressive stress, which is concentrated in the simulated contact points between the crown and the metal loading ball and on top of the abutment. The positive values of Figure 7 show the solid maximum principal stress for the EM crown with contact surfaces in the interface abutment/crown, which represents tensile stresses. The highest tensile stress is found in the fissure. The PT crown was also calculated with FEA. These results showed a similar stress distribution to the EM crown.

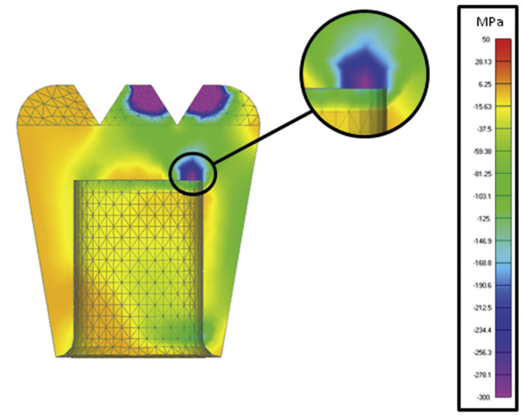

Figure 6. Solid minimum principal stress (compressive stress) on lithium disilicate crown with contact surfaces in interface crown/abutment (model 2). Highest stresses were found on top of abutment.

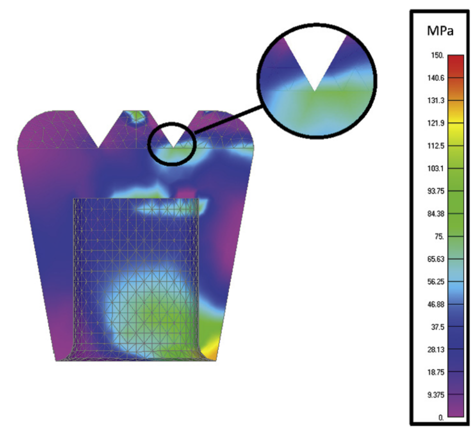

Figure 7. Solid maximum principal stress (tensile stress) on lithium disilicate crown with contact surfaces in interface crown/abutment (model 2). Highest stresses were found in fissure and beside top of abutment.

DISCUSSION

The results of this study show clear differences in the mechanical failure properties of the several ceramic and composite resin implant-supported crowns. Anatomic contour zirconia crowns (LP) have the highest fracture load, FS, and E. Because of these properties, the risk of fracture might be expected to be limited. In contrast, the forces withstood by the restoration might be passed toward other parts of the implant restoration, and this could create more stress dissipation around the cervical region of the abutment and abutment-implant junction.[20] When this stress exceeds a certain threshold, it might lead to bone loss around the cervical region and finally even to loss of the implant. In this study, no damage was seen on the abutment, abutment screw, or the implant itself after ILF testing.

The lithium disilicate (EM) crowns performed second best, with a failure load less than half of the LP crowns. They were followed by MF and PT, which is remarkable because these 2 materials are not used for definitive indirect restorations. PT performed well in terms of ILF, despite a significantly lower FS and E compared to the other resins tested in this study. As shown in Figure 4, high elasticity leads to more plastic deformation and more energy absorption, which probably explains the high ILF. This effect is less obvious with the other composite resin crowns (ES, LU, TC, and VE), which also shows significantly lower E compared to the ceramic crowns, but not as low as PT, except for TC. Still, they absorb more energy than ceramics and therefore lead to more damping of occlusal forces, as found in earlier studies.[19] This phenomenon is illustrated by the observation that the FS of EM is more than twice as high as that of most tested composites, while the ILF shows a relatively small increase. According to these results, PT seems to be the best composite resin material for implant-supported crown applications. The disadvantage of this material is its limited resistance to wear, which makes it inappropriate for more definitive restorations.[33] Composite resins did not exceed monolithic ceramic crowns in ILF. Additionally, the mechanical properties of composite resins are negatively influenced by water sorption when exposed to the oral cavity.[26] However, they do have the advantage of being easier to repair when broken.

The veneered zirconia crowns (LV) showed the lowest ILF due to fracture of the relatively weak ceramic veneering layer.[14] In addition to the strength of the veneering ceramic, the veneering procedure might also have been an influence, as it was done by heat pressing instead of manually to optimize reproducibility. According to previous studies, this method decreases the reliability compared to manual veneering.[11,12] Although the LV crowns showed the lowest fracture load, the zirconia substructure stayed intact and did not show any fatal fracture, while most other crowns did. Therefore, the success of LV crowns is limited, but their clinical survival rate may be higher than those of other materials.

The load applied on all crowns greatly exceeded the average masticatory force of healthy young adults, which makes these results of limited clinical relevance.[7] Even more so, because in addition to occlusal loading, other factors such as fatigue loading, fracture toughness, and degradation by water sorption will be of equal importance in the clinical success of an implant-supported restoration.[5,26] Nonetheless, the ILF is relevant to the comparison of fracture resistance caused by functional loading. The ILF, together with the results of the other tests and the FEA, offers insight into the fracture resistance of several restorative materials.

The results of the 3-point bending test and compressive test show once again that dental restoration materials are more likely to fail because of their lack of FS rather than their lack of CS, as the FS found in this study was approximately one third of the CS. The results are comparable with the values claimed by manufacturers, except for the FS of EM, which is significantly lower (301 ±70 MPa versus 380 to 400 MPa according to the manufacturer). Although the manufacturer's FS was confirmed by Kang et al[27] (408 ±86 MPa), Belli et al[28] found a lower FS (260 ±47 MPa) in a 4-point bending test, and Pollington and van Noort[29] reported 266 ±37 MPa as biaxial FS.

The FEA on the crowns with a fixed connection between the abutment and crown shows a maximum stress at the interface crown/abutment in the cervical region that exceeds the bond strength of the cement (Fig. 5).[40] Therefore, this was not a realistic model. Model 2, with contact surfaces between the abutment and crown, shows compressive stress areas in the simulated contact points between the crown and the metal loading ball and on top of the abutment and a tensile stress area in the fissure of the crown (Figs. 6, 7). This suggests that fracture of the crown occurs from the fissure to the top of the abutment. These findings were confirmed by the fracture patterns on the remnants of the ILF test (Fig. 3). The clear mist and hackle area in the occlusal part of the broken LP and PT crowns demonstrates that the fracture originated from the fissure.[34,35] Additionally, white lines close to the top of the abutment and white spots on the cusps of the remnants indicate permanent plastic deformation caused by compressive stress that did not lead to catastrophic fracture.

It is remarkable that none of the stresses shown in the FEA exceeds the compressive or FS of lithium disilicate, while these crowns do fail at this load during ILF testing. As mentioned earlier, the strength of ceramics is compromised by surface roughness.[37,38] A significantly lower FS was found for the roughened EM specimens (Table 2). This suggests that a surface roughness of Ra=4.3 μm reduces the FS of the material even more and probably reduces the local FS and CS on top of the abutment to the stress found in the FEA.

Finally, the forces applied to the crowns in this study are not representative of those found in the clinical situation. Therefore, all the materials tested should be able to withstand excessive occlusal forces in the mouth. The crowns in the present study were not exposed to the fatigue, wear, and aging of long-term function in the oral cavity.

CONCLUSIONS

Within the limitations of this study, monolithic implant-supported crowns have a higher ILF than conventional veneered ceramic crowns and are expected to show clinically fewer fractures. Monolithic ceramic restorations perform better than composite resin crowns in this study and might even perform better if the internal surface could be smoother. Composite resins with a low E show promising results, and all restoration materials tested should withstand the physiological forces applied in the oral cavity.

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