IPS E.MAX

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Abstract

Executive Summary (Ips E.Max Press; Ips E.Max Cad - Ivoclar Vivadent Scientific Documentation, Guess (2010), Silva (2011), Gehrt (2012), Wolfart (2009)

IPS E.max (Ivoclar Vivadent) was introduced in 2005 as an improved ceramic material compared to IPS Empress 2. It consists of a lithium-disilicate pressed glass ceramic, but its physical properties and translucency are improved through a different firing process. E.max comes in two different forms - E.max Press and E.max CAD [1].

E.max Press - the microstructure of IPS E.max Press consists of lithium disilicate crystals (approx. 70%), Li2Si2O5, embedded in a glassy matrix. Lithium disilicate is the main crystal phase and consists of needle-like crystals. The crystals measure 3 to 6 μm in length. In general, the properties of E.max Press are very slightly superior to that of E.max CAD due to larger and longer crystals. This is because of the differing firing temperatures for the two (820oC for E.max Press, and 850oC for E.max CAD). E.max CAD has a different firing temperature as it needs to be machined prior to firing.

E.max CAD - Blocks are presses in a partially crystalline state (called the blue state). Partially crystallized IPS e.max CAD consists of 40% lithium metasilicate crystals (Li_2SiO_3), which are embedded in a glassy phase. The grain size of the platelet-shaped crystals is in the range of 0.2 to 1.0 μm. End-crystallized IPS e.max CAD (fired at 850°C for about 20 - 30mins) consists of approx. 70% fine-grain lithium disilicate crystals ($Li_2Si_2O_5$).

Keywords: IPS E.Max; Ivoclar Vivadent; lithium disilicate

Introduction

E.max is thought to resist fracture due to Crack Deflection- this is due to different coefficient of thermal expansion between the glass matrix and crystal. This dispersion of the crack away from where it was originally travelling to causes dispersion of its energy. The strength of lithium disilicate, and most ceramics, comes from the bonding of the material to the tooth.

Lithium disilicate glass- ceramics, the strongest and toughest of the glass-ceramics available, have moderate flexural strength (360 - 440 MPa) and fracture toughness (2.5 - 3MPam$^{1/2}$) [2].

Monolithic E.max crowns have been able to withstand chewing stimulation loading of up to 2500N before bulk fracture. Thinner and veneered E.max anatomical crowns have achieved similar failure loads as that of metal-ceramic crowns (around 1300 - 1500N). Similar

findings have been found when monolithic FPDs were assess with metal ceramic FDPs - Guess [3], Silva [4].

In terms of wear, there is a general consensus in the literature that E.max material shows a lower rate of wear than other ceramics and cause less wear to opposing structures when wear tests have been conducted *in-vitro*.

With regards to longevity, there are no published reviews as yet. However, clinical studies - though having short mean observation periods - have shown to be promising.

Gehrt [5] conducted a prospective study to evaluate the clinical outcome of anterior and posterior crowns made of a lithium-disilicate glass (n = 104 - 82 anterior; 22 posterior). Observation time ranged from 34 - 109.7 months, with a mean observation time of 79.5 months. The cumulative survival rate according to Kaplan-Meier was 97.4 % after 5 years and 94.8 % after 8 years.

Wolfart (2012) evaluated the clinical outcomes of 36 FPDs made from lithium disilicate glass-ceramic with 84% in posterior sectors and 16% in anterior sectors. A 7% fracture rate was observed over an 8-year period.

This is a material that has gained widespread clinical use due to its superior aesthetics and reported strength. With further studies, this material can become a material that will give an excellent as well as predictable long-term result for both the clinician and the patient.

Glass ceramics were first developed by Corning Glass Works in the late 1950s. Over time, manufacturers started adding filler particles to the base glass composition to improve mechanical properties, such as strength and thermal expansion and contraction behavior. Crystalline filler particles can be added mechanically to the glass, for example by mixing together crystalline and glass powders before firing. In a more recent approach, the filler particles are grown inside the glass object (prosthesis or pellet for pressing into a mold) after the object has been formed. After forming, the glass object is given a special heat treatment causing the precipitation and growth of crystallites within the glass. Because these fillers are derived chemically from atoms of the glass itself, it stands to reason that the composition of the remaining glass is altered as well during this process.

In order to be able to extend the use of resin-bonded ceramic restorations and possibly use them for bridge construction, a glass ceramic based on a SiO_2-Li_2O (lithium disilicate) system was been developed and introduced into the market in 1998 (Empress II, Ivoclar-Vivadent). To increase the strength, thermal expansion and contraction behavior of ceramics, manufacturers have added crystalline filler particles.

The crystalline phase that forms is a lithium disilicate ($Li_2Si_2O_5$) and makes up about 70% of the volume of the glass ceramic. Lithium disilicate has an unusual microstructure, in that it consists of many small interlocking plate-like crystals that are randomly oriented. This is ideal from the point of view of strength because the needle-like crystals cause cracks to deflect, branch or blunt; thus, the propagation of cracks through this material is arrested by the lithium disilicate crystals, providing a substantial increase in the flexural strength.

IPS e.max Press (Ivoclar Vivadent) was introduced in 2005 as an improved press-ceramic material compared to IPS Empress 2. It also consists of a lithium-disilicate pressed glass ceramic, but its physical properties and translucency are improved through a different firing process. E.max comes in two different forms - E.max Press and E.max CAD. E.max is the commercial name of the material that is now being used. Lithium-disicilate refers to the material composition and includes Empress 2 as well as E.max Press and E.max CAD.

E.max PRESS

The IPS e.max Press material consists of a lithium disilicate pressed glass ceramic. The chemical basis of the material is the same as the chemical basis of IPS Empress 2 ($2SiO_2$-Li_2O), but properties are changed by a different firing process. In comparison with IPS Empress 2,

the two glass ceramic materials exhibit substantially improved physical properties and greater translucency. E.max Press comes in four levels of opacity - HT (High translucent), LT (Low translucent), MO (Medium opaque), HO (High opaque).

The microstructure of IPS e.max Press consists of lithium disilicate crystals (approx. 70%), $Li_2Si_2O_5$, embedded in a glassy matrix. Lithium disilicate is the main crystal phase and consists of needle-like crystals. The crystals measure 3 to 6 μm in length. In general, the properties of E.max Press are very slightly superior to that of E.max CAD due to larger and longer crystals. This is because of the differing firing temperatures for the two (980°C for E.max Press, and 850°C for E.max CAD). E.max CAD has a different firing temperature as it needs to be machined prior to firing.

Figure 1: *Microstructure of IPS e.max Press (SEM, etched with HF vapour for 30 s).*

Physical Properties

Physical Property	Value	Investigator
Fracture toughness	2.5 - 3.0Mpam$^{1/2}$	in-house (Ivoclar Vivadent AG, Schaan)
Modulus of elasticity	95 ± 5 GPa	in-house (Ivoclar Vivadent AG, Schaan)
Modulus of elasticity	91.0 GPa	Albakry., *et al*. [2]
Modulus of elasticity	94.4 GPa	Lohbauer
Modulus of elasticity	96.0 GPa	Anusavice
Poisson's ratio u	0.23	Albakry., *et al*. [2]
Vickers hardness [HV10]	5900 ± 100 MPa	in-house (Ivoclar Vivadent AG, Schaan)
Hardness	5.5 GPa	Albakry., *et al*. [3]
Density	2.5 ± 0.1 g/cm^3	in-house (Ivoclar Vivadent AG, Schaan)

Table 1: *Physical properties.*

Flexural strength

Flexural strength of IPS e.max Press (various methods)

Flexural strength values largely depend on the methods used to measure them. Figure 2 provides an overview of the flexural strength values measured with different methods.
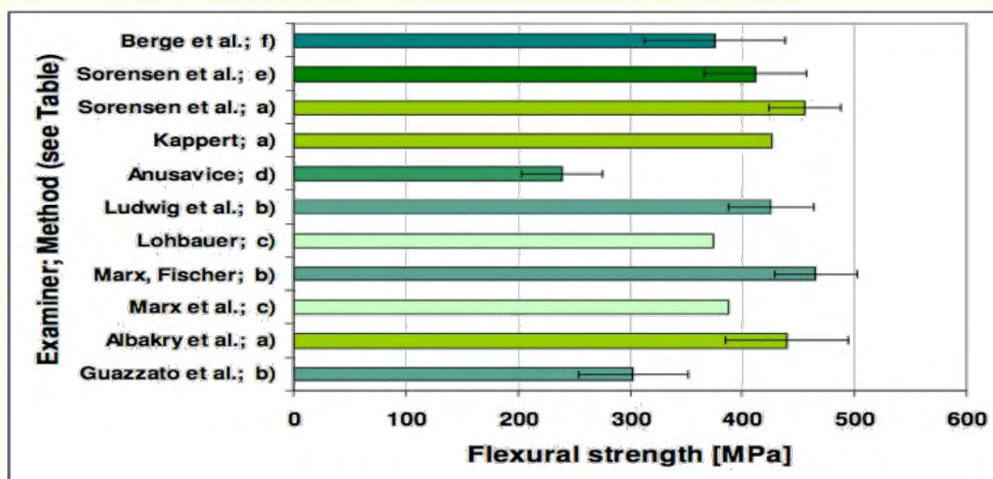

Figure 2: *Flexural strength values measured for IPS e.max Press using different methods (see also Table 2).*

Investigator	Flexural strength [MPa]	Measuring method:
Berge., *et al.* [4]; f)	375.7	Biaxial flexural strength ISO 6872; test in H_2O
Sorensen., *et al.* [5]; e)	411.6	Biaxial flexural strength (wet test)
Sorensen., *et al.* [5]; a)	455.5	Biaxial flexural strength
Kappert; a)	426	Biaxial flexural strength
Anusavice [6]; d)	239	4-point flexural strength after 48 hours of storage in H_2O
Ludwig., *et al.* [7]; b)	426	3-point flexural strength
Lohbauer c)	374.4	Weibull strength σ 63.21%; 4-point flexural strength DIN EN 843-1
Marx, Fischer; b)	466	3-point flexural strength
Marx., *et al.* [8]; c)	388	Weibull strength σ 63.21%; 4-point flexural strength DIN EN 843-1
Albakry., *et al.* [2]; a)	440	Biaxial flexural strength
Guazzato., *et al.* [9]; b)	303	3-point flexural strength

Table 2: *Values and measuring methods shown in Figure 2.*

E.max CAD

Blocks are presses in a partially crystalline state. Partial crystallization ensures that the blocks can be processed in a crystalline intermediate phase, which enables fast machining with CAD/CAM systems - call the "blue state" (as the material appears a translucent blue colour due to different oxidation state of the polyvalent colouring elements). The partial crystallization process leads to a formation of lithium metasilicate crystals Li_2SiO_3.

Following the milling procedure, the restorations are tempered and thus reach the fully crystallized state. In the course of this process, lithium disilicate crystals ($Li_2Si_2O_5$) are formed, which impart the ceramic object with the desired high strength.

Partially crystallized IPS e.max CAD consists of 40% lithium metasilicate crystals (Li_2SiO^3), which are embedded in a glassy phase. The grain size of the platelet-shaped crystals is in the range of 0.2 to 1.0 μm.

End-crystallized IPS e.max CAD (fired at 850°C for about 20 - 30 mins) consists of approx. 70% fine-grain lithium disilicate crystals ($Li_2Si_2O_5$), which are embedded in a glassy matrix. By etching with hydrofluoric acid vapour, the glassy phase is dissolved and the lithium disilicate crystals become visible.
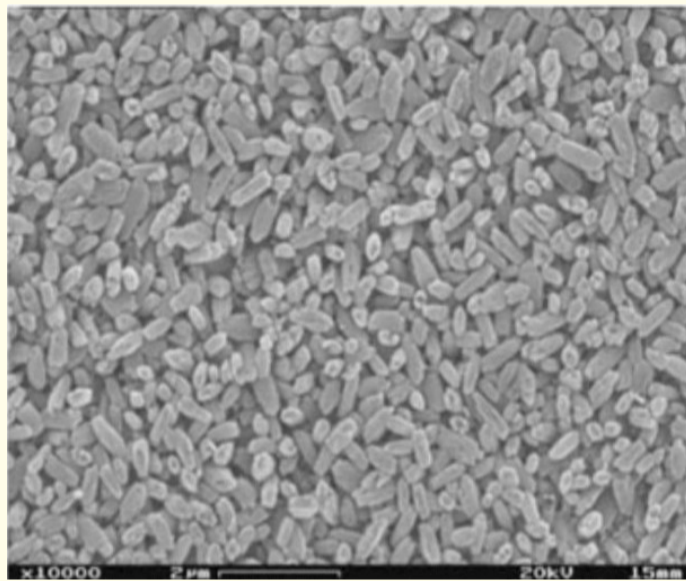

Figure 3: Fully crystallized IPS e.max CAD (SEM, etched with 0.5% HF vapour for 30 seconds).

Figure 4: Bridge framework in the partially crystallized state.

Physical Properties

Physical Properties	Partially crystallized state	Fully crystallized state
Biaxial strength (ISO 6872)	130 ± 30 MPa	360 ± 60 MPa
Fracture toughness (SEVNB)	0.9 – 1.1 MPa m$^{\frac{1}{2}}$	2.0 – 2.5 MPa m$^{\frac{1}{2}}$
Vickers hardness	5400 ± 100 MPa	5800 ± 100 MPa
Modulus of elasticity		95 ± 5 GPa
CTE (100-500°C)		10.45 ± 0.25 10^{-6}/K^{-1}
Density		2.5 ± 0.1 g/cm^3
Linear shrinkage during the tempering process	0.2%	
Chemical solubility	100 – 160 µg/cm^2	30 – 50 µg/cm^2

Table 3: *Physical properties (Ivoclar Vivadent, Schaan, 2005).*

As lithium disilicate glass-ceramic (LS2) and zirconium oxide (IPS e.max ZirCAD) feature a very similar coefficient of thermal expansion, the same layering ceramic (IPS e.max Ceram) can be used in conjunction with all the components of the IPS e.max system.

Colour

Natural esthetic rendition is a primary aim after restoring teeth with full-coverage crowns. Therefore, an ideal dental material for the fabrication of crowns would allow the control of substrate color and translucency. Traditional metal-ceramic crowns exhibit a lack of light exchange with the surrounding soft tissues caused by the reflection of their metal frameworks and their opaque layers. As a result, they often present a compromised esthetic appearance compared to natural teeth.

One of the many advantages of E.max is its optical refractive index - the lithium disilicate crystal's refractive index is similar to that of the glass matrix, thus different translucency (four levels) levels can be achieved and different shades can also be achieved by changing the glass matrix colour (adding pigment to the glass matrix). This property allows the high strength of the material to be maintained as more crystals can be added to the matrix without affecting the translucency. This allows the material to achieve a fracture toughness of 360 - 400 MPa, yet still maintain the same degree of translucency as some traditional feldspathic ceramics.

The color of the ceramic restoration can also be modified to match that of the natural tooth by layering it with veneering ceramic or by custom staining and glazing. Although ceramic systems improve color and translucency of the restorations, a perfect color result cannot be ensured. Dentin constitutes the bulk of a tooth and is largely responsible for its color. Ceramics that are more translucent allow more light to enter and scatter, which means that the underlying tooth has a significant influence over the resultant color.

In general, the optical behavior of a ceramic restoration is determined by the combination of the underlying tooth structure color, the thickness of the ceramic layers, and the color of the cement.

Chaiyabutr [6] evaluated the cumulative effect of the tooth abutment color, cement color, and ceramic thickness on the optical resultant color of a CAD/CAM lithium disilicate crown. Four prepared abutment tooth colors (light, medium light, medium dark, and dark), 2 cement (Variolink II) colors (translucent and opaque), and 4 ceramic thickness values (1.0 mm, 1.5 mm, 2.0 mm, and 2.5 mm) were used. The color of each combination was measured using a spectrophotometer, and the average values of the color difference (ΔE) were calculated. The results showed that the underlying tooth abutment color, cement color, and ceramic thickness significantly influenced

the resulting optical color of the crown. Changing the underlying color of the abutment tooth from a lighter to a darker color resulted in increased ΔE values. On dark-colored abutment teeth, the crowns with a ceramic thickness of 1.0 mm, cemented using translucent cement or opaque cement, and the crowns with a ceramic thickness of 1.5 mm, cemented with translucent cement, were within a clinically unacceptable color change range.

It is known that ceramic opacity is increased with increasing thickness. As the thickness of ceramic increases, the diffused reflection effects of the underlying abutment tooth diminish, and the majority of diffused reflection occurs in the ceramic crown.

Wear

With the increased demand of aesthetic dentistry and the development of newer ceramics, the wear studies of ceramics have become a widely researched area.

In a recently published study, Rosentritt [7] investigated the two-body wear of different ceramics. Two-body wear tests were performed in a chewing simulator with steatite and enamel antagonists, respectively. Specimens were loaded with a vertical load of 50 N for 1.2×10^5 cycles. Human enamel was used as a reference. Specimen samples included zirconia ceramics, veneering porcelains, glass-infiltrated ceramics and lithium disilicate ceramics. Veneering and lithium disilicate ceramics were glazed before testing. The results found that wear of specimens and antagonists was strongly material dependent. No visible wear was found on zirconia and glass-infiltrated ceramics. Porcelain and lithium disilicate ceramic showed a comparable or lower wear than the enamel reference.

Antagonist wear was found to be lower when specimens were made of substructure oxide ceramics (zirconia) instead of veneering porcelain.

In another lab-based study, a comparison of the wear characteristics of 3 types of ceramics - IPSd. SIGN, Empress 1 and E.max was conducted by Heinze [8]. Wear machine were used 120,000 cycles with 5kg weight.

Modified variables were crown configuration - flat and cusped, surface treatment - polished and glazed; and the antagonists were enamel and ceramic.

The authors concluded that it was difficult to review systematically however because of the following variables:

- enamel cannot be standardized, so unable to get standard specimens of this
- pressed ceramic contain voids and other defects which may interferes with wear resistance
- layering of ceramics can incorporate voids
- fabrication steps like sandblasting, polishing, application of glaze, etc, can contribute to wear variables

However, despite this the results showed that there was more material wear and less antagonist wear for ceramic stylus versus enamel stylus, however E.max wore less than all the other ceramics even with ceramic stylus.

As reported by many of the papers on wear, hardness traditionally has been blamed for the accelerated loss of material; however, scientific studies have not demonstrated a strong correlation between the hardness of ceramic and the wear rate of human enamel.

Instead, the wear process appears to be more closely related to the ceramic microstructure, the roughness of contacting surfaces, and environmental influences [9].

Hardness is determined by measuring the resistance of material surface to permanent deformation or penetration by an indenter. Recent evidence suggests that the hardness of a restorative material alone is not a reliable predictor of the wear of opposing enamel [10]. In particular, the relationship of wear to hardness is not valid for materials that are brittle in nature. When ceramic slides against ceramic or enamel, wear does not occur by plastic deformation, as with metals, but by fracture. This type of abrasive wear mechanism has been well addressed by DeLong [11].

Internal porosity and other surface defects, which are produced by an inadequate firing technique, act as stress concentrators and result in greater wear. Glazing and/or polishing ceramic can influence the early stage of the wear process, but the positive effect of a glazed/polished surface is quickly lost when the material is placed in function. The internal characterization of ceramics is recommended because shading materials contain abrasive metal oxides. Application of external stains should be limited to the non-contacting surfaces of esthetic restorations [10].

Fracture Toughness

Fracture toughness is the intrinsic ability of a material to resist crack propagation - that is, the amount of energy required to propagate an existing flaw.

It is independent of flaw distribution - therefore thought to be a more consistent property.

The fracture toughness of advanced ceramics is often measured using an indentation technique. A polished surface of the material under test is indented using a Vickers hardness tester, the cracks which emanate from the corners or the indents are then measured and provide an indication of the toughness of the material.

E.max is thought to resist fracture due to Crack Deflection - this is due to different coefficient of thermal expansion between the glass matrix and crystal. This dispersion of the crack away from where it was originally travelling to causes dispersion of its energy.

Lithium disilicate glass-ceramics, the strongest and toughest of the glass-ceramics available, have moderate flexural strength (360 - 440 MPa) and fracture toughness (2.5 - 3MPam$^{1/2}$) [2].

Drummond [12] evaluated the material properties of several ceramics (leucite reinforced, feldspathic, low fusing feldspathic and lithium disilicate ceramics). The study assess the flexure strength under static and cyclic loading and the fracture toughness under static loading. The results showed that the lithium disilicate ceramic excelled in all parts of the experiment - the mean flexure strength for the other tested ceramics in air and water (without aging or cyclic loading) ranged from 67 to 99 MPa, whilst the lithium disilicate specimen was twice as strong with a mean flexure strength of 191 - 205 MPa. For the mean fracture toughness, the range the other ceramics were 1.1-1.9 MPa/m0.5 with the lithium disilicate ceramic being 2.7 MPa/m0.5. The author believed that increase in fracture toughness (like the observed increase in flexure strength) is most likely due to smaller, more uniform, particle distribution rather than particle composition.

The high fracture toughness of E.max has been attributed to its resistance to damage with clinically applicable loads in its monolithic form. Guess [3] evaluated the fatigue behavior of (CAD/CAM) lithium disilicate crowns. 19 fully anatomically shaped mandibular molar monolithic lithium disilicate crowns (IPS e.max CAD) designed and milled (occlusal reduction 2.0 mm and wall reduction of 1.5 mm). The crowns were cemented on aged dentin-like resin dies with resin cement.

Indenter was slide 0.7 mm lingually down the disto-buccal cusp to mimic occlusal chewing contact.

It was found that when load to failure was conducted, bulk fracture occurred around 2500N.

At up to 1100N @ 1,000,000 cycles, there was no damage to the crowns - which far exceeds the loads that can be clinically applied in the mouth.

This study was repeated by Silva in 2012 [4], however using thinner crowns and thinly veneered E.max crowns, in comparison to a metal ceramic crown to test how the thickness affected the fracture load. Exactly the same methods and loading protocols were used, however the monolithic crowns had a thickness of 1.0 mm, the veneered crowns had a thickness of 2.0 mm with a 0.5 mm veneer and the metal ceramic crown was used as a control.

It was found that the thin monolithic crown and the thinly veneered crown withstood loading up to about 1500N and it was comparable to the metal-ceramic crown, which failed at ~1300N. The veneered crowns failed via chipping of the veneering porcelain. This study suggests that the lithium disilicate crowns can withstand similar loading to that of metal-ceramic crowns.

Although several opacities and translucencies have been developed for pressable glass ceramic systems, ceramic cores are generally veneered with weaker porcelain to achieve optimized esthetics. The lower intrinsic strength of veneering porcelain may still determine the longevity in spite of a strong substrate, as the flexural resistance of a bi-layered structure is dependent upon the veneering external layer of the structure. Zao [13] found that the fracture loads of full anatomic monolithic crowns were significantly higher than veneered crowns. Similarly, to the findings of the above mentioned studies by Guess [3] and Silva [4], monolithic crowns mainly fail via bulk fracture, whereas veneered crowns predominately fail via cohesive veneer and ceramic interface failure or solely cohesive veneer failure [13]. With the superior properties of monolithic E.max and the different degrees of translucency that can be achieved, it is possible to achieve an acceptable aesthetic outcome by using just monolithic E.max and utilizing surface staining. This can help reduce the incidence of failure of the restoration via chipping of the porcelain - which is often one of the most common reasons for failure in the longevity studies.

Ma [2] used FEA to study the load-bearing capacity of a monolithic lithium disilicate occlusal onlay when compared to zirconia (which has been documented to have a much higher fracture toughness).

Ceramic occlusal onlays of various thicknesses cemented to either enamel or dentin were considered. Occlusal load was applied through an enamel-like deformable indenter or a control rigid indenter.

When bonded to enamel (supported by dentin), the load-bearing capacity of lithium disilicate can approach 75% of that of zirconia, despite the flexural strength of lithium disilicate (400 MPa) being merely 40% of zirconia (1000MPa). When bonded to dentin (with the enamel completely removed), the load-bearing capacity of lithium disilicate is about 57% of zirconia, still significantly higher than the anticipated value based on its strength. Both ceramics show slightly higher load-bearing capacity when loaded with a deformable indenter (enamel, glass-ceramic, or porcelain) rather than a rigid indenter.

It should be noted, however, that this test did not use anatomical specimens - rather it utilized disk shapes. The behavior of the ceramic when differentials exist in terms of thickness and supporting structure, may result in a different fracture pattern and rate than those seen in the study. Nevertheless, this study gives way to a general outline of the fracture behavior of lithium disilicate when compared to zirconia and also the increased performance of lithium disilicate when adhesively bonded.

With the advances of bonding, it has been found to significantly increase the fracture toughness of ceramic crowns, to a point that is much higher than any clinical loads that can be generated functionally in the mouth.

Groten [14] looked at how bonding affected the fracture resistance of 120 pressed feldspathic ceramic crowns. The crowns were either luted with zinc phosphate/GIC or adhesively cemented with resin cement. The specimens were placed in dry storage for 48 hours before being subjected to fracture resistance test with a universal testing machine (progressive load at 45 degree angle until failure). The author noted significant increase (1.5 - 2.0x) in fracture resistance when the fitting surface of the crown was etched and silanated and a resin cement was used.

Similarly, Scherrer [15] used ceramic crowns bonded to dentine (with resin cement) on third molar teeth. Some were adhesively bonded, some luted with zinc phosphate. A spherical indenter was used to test the fracture toughness of the tooth. Intact teeth had highest fracture resistance than ceramic crowns. The adhesively bonded ones displayed higher fracture resistance than the luted ones.

One of the other indicators of the performance of a ceramic is the Weibull's modulus. This is related to the distribution of physical flaws present in the surface or body of the brittle specimen since brittle failure processes originate at these weak points. When flaws are consistent and evenly distributed, samples will behave more uniformly than when flaws are clustered inconsistently. This must be taken into account when describing the strength of the material, so strength is best represented as a distribution of values rather than as one specific value.

Consider strength measurements made on many small samples of a brittle ceramic material. If the measurements show little variation from sample to sample, the calculated Weibull modulus will be high and a single strength value would serve as a good description of the sample-to-sample performance. It may be concluded that its physical flaws, whether inherent to the material itself or resulting from the manufacturing process, are distributed uniformly throughout the material. If the measurements show high variation, the calculated Weibull modulus will be low; this reveals that flaws are clustered inconsistently and the measured strength will be generally weak and variable. Products made from components of low Weibull modulus will exhibit low reliability and their strengths will be broadly distributed.

Marx [16] found that the Weibull strength of IPS e.max Press demonstrated to be higher than that of IPS Empress 2.
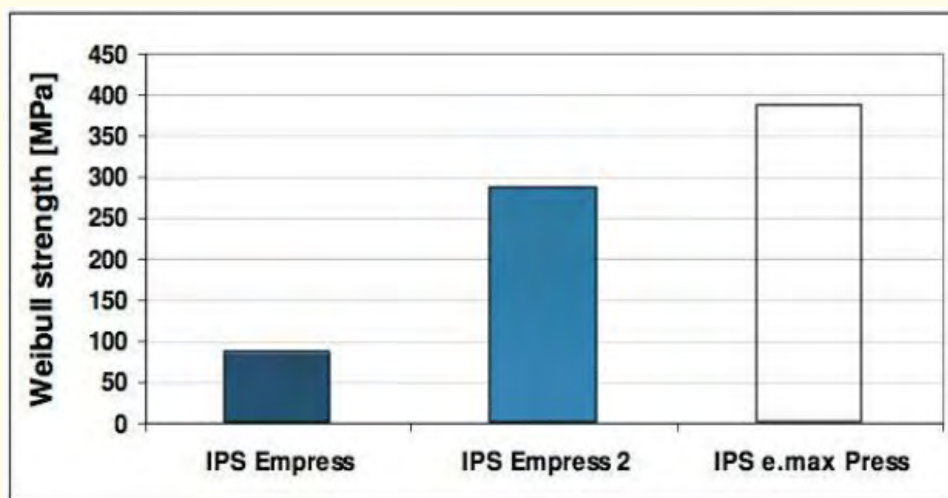
Marginal Fit

Whilst there is no consensus on what is an acceptable marginal gap, some authors such as Christensen [17] reported the clinically detectable range for sub- gingival margins to be 34 - 119 μm and 2 - 5 μm for supragingival margins. Subsequently, McLean [18] suggested that 120 μm should be the limit for clinically acceptable marginal discrepancies. Poor marginal adaptation can result in cement dissolution, microleakage, increased plaque retention, and secondary decay.
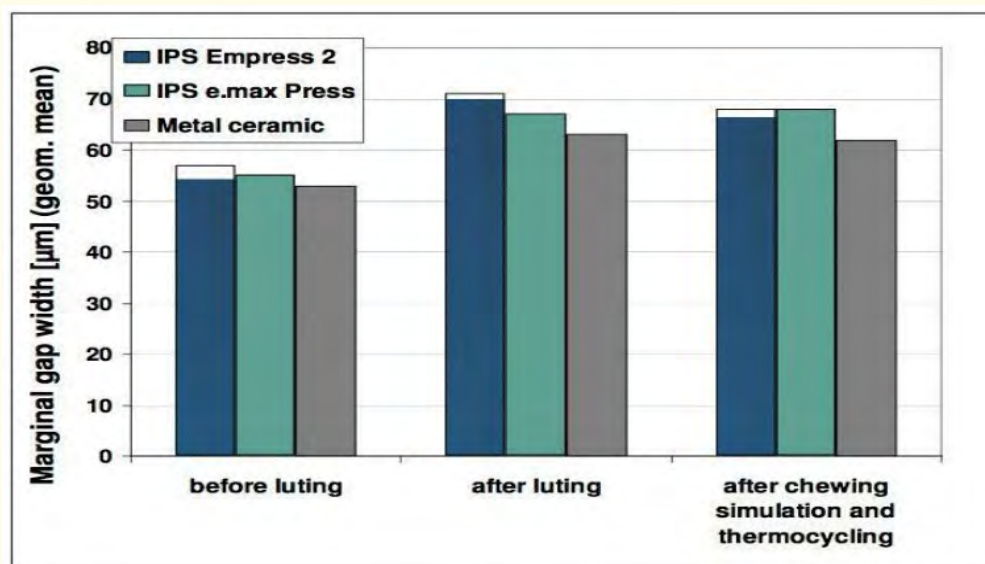
Stappert [19,20] measured the marginal gap widths in three-unit bridges before and after cementation and after thermomechanical loading. IPS Empress 2, IPS e.max Press and metal-ceramic bridges as a control group were examined. The bridges were adhesively cemented with Variolink II and thermomechanical loading was performed in a chewing simulator.

The results show that E.max had comparable marginal gaps after luting and had smaller marginal gap than the other specimen after the chewing simulator and thermocycling had been carried out.

Although indirect ceramic restorations undergo complex three-dimensional changes during their fabrication process, fit analyses have usually been restricted to a single dimension.

In a literature review by Conrad [21] it was found that lithium disilicate crowns achieved a marginal gap of around 65 - 120 um. The author states that marginal gaps observed with glass-ceramic restorations may be dependent upon the mechanical properties of the luting cement to resist functional forces. Most of the literature, however, reports marginal discrepancies in the range of clinical acceptability recommended by Christensen and McLean.

Longevity

No reviews as yet. However, clinical studies - though having short mean observation periods - have shown to be promising. Gehrt [5]

conducted a prospective study to evaluate the clinical outcome of anterior and posterior crowns made of a lithium-disilicate glass (n = 104 - 82 anterior; 22 posterior).

Observation time ranged from 34 - 109.7 months, with a mean observation time of 79.5 months.

The cumulative survival rate according to Kaplan-Meier was 97.4 % after 5 years and 94.8 % after 8 years. The location of the crown in the mouth was not statistically significant.

Most common failure was chipping of veneering material.

Wolfart (2009) conducted a clinical study with lithium disilicate ceramic FPD's (n = 33; 3-unit - mostly posterior region).

The mean observation period of the 33 FDPs was 86 months (range: 67-98 months): two FDPs in two patients had to be replaced (6%) because of fractures and chipping of the veneering material was found in two FDPs (6%).

The 8-year survival rate according to Kaplan-Meier was 93%.

Kern [22] observed 36 three-unit FDPs made from monolithic lithium disilicate ceramic (IPS e.max Press, Ivoclar Vivadent) without any cantilever pontics. Both anterior and posterior FDPs were used, even though the manufacturer recommends this material for anterior FPDs only.

The mean observation period was 121 months. FDPs' survival rate (survival being defined as remaining in place either with or without complications) was:

- 100% after 5 years and 87.9% after 10 years

Their success rate (success being defined as remaining unchanged and free of complications) was:

- 91.1% after 5 years and 69.8% after 10 years

Conrad (2007) reported longevity of ceramic restorations to range from:

- 88 - 100% after 2-5 years in service
- 84 - 97% after 5-14 years in service

Discrepancy in the classification of failures and variability of the materials and systems available for all-ceramic restorations present a challenge to combining data from several studies.

E.MAX for Fixed Partial Denture Fabrication

Whilst the longevity and clinical performance of single unit crowns have been established and research, the use of ceramics in fixed partial denture design has always been a controversial area. Despite the excellent esthetics of all-ceramic FDPs, skepticism remains regarding their strength and long- term serviceability. Evidence is lacking with respect to how connector design affects these new ceramic materials. It is also un- clear how connector design and fabrication technique affect the strength of these materials.
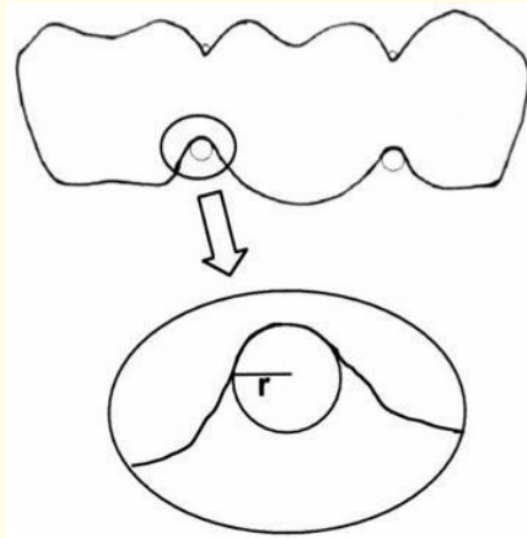
Figure 7: *Schematic illustration of radius of curvature (r) at embrasure area.*

Oh (2002) evaluated the effect of connector design on the fracture resistance of all-ceramic FPDs (using lithium disilicate). Two wax carvings with radius of curvature at their tips of 0.90 and 0.25 mm were used. The results showed that as the radius at the gingival embrasure increased from 0.25 to 0.90 mm, the mean failure load increased by 140%.
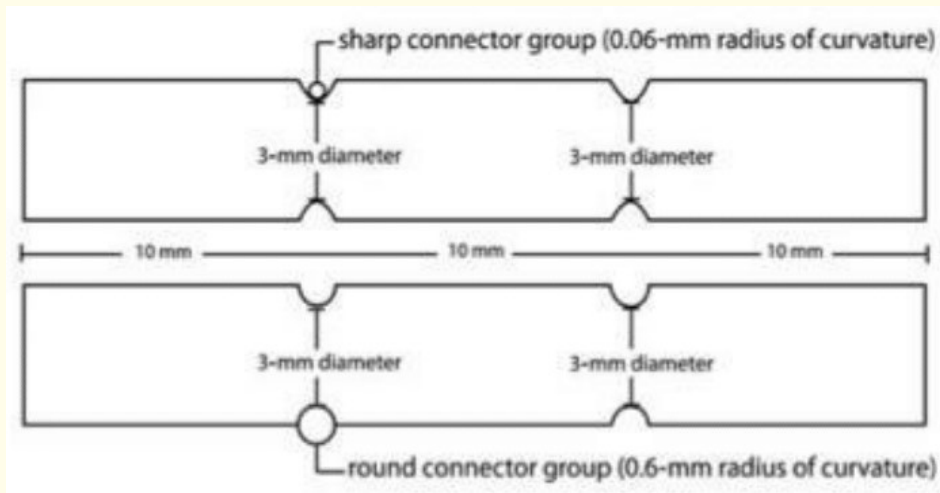
Similarly, Plengsombut [23] investigated the effect of two connector designs on the fracture loads of lithium disilicate FPD's (E.max Press and E.max CAD). Two connector designs, round (0.60 ± 0.01-mm radius of curvature) and sharp (0.06 ± 0.001- mm radius of curva-

ture), with a 3.00 ± 0.05-mm cross-section for each connector, were studied. When loaded it was found that the round connector design was able to withstand higher loading than the sharp connector design.

SEM subjective assessment of the fractured specimens revealed that the fracture initiated from the gingival surface (tensile) of the connector toward the pontic (central loading point). This fracture pattern can be explained by the physical properties of ceramic materials that enables them to withstand compressive forces better than tensile forces.

Zheng [24] used finite element modeling to reveal differences in tensile and compressive stresses between different pontic preparation configurations and core materials. It was found that in general, gold alloy provided the most even stress distribution at the connector and pontic when compared with lithium disilicate and larger connector configurations, provided less stress at the connector and pontic area.

Schultheis [25] assess the fracture loads and failure modes of lithium disilicate FPDs compared to metal-ceramic FPDs. Lithium-disilicate ceramic FPDs (IPS-e.max-CAD) were milled in full-anatomic FDP dimensions (posterior 3-unit to replace a premolar) - both monolithic and as a framework for veneering. The connector size for monolithic was 4x6 mm and for the veneered was 4x4mm. Metal-ceramic FDPs served as control. Single-load-to-failure tests were performed before and after mouth-motion fatigue. The results are as shown:

Material	Mean fracture load (N) before mouth-motion fatigue	Mean fracture load (N) after mouth-motion fatigue	Failure mode
Monolithic lithium disilicate	1298	1900	Via bulk fracture within the connector
Veneered lithium disilicate	817	699	Via veneer fracture within the connector
Metal-ceramic	1966	1818	Via ceramic veneer fracture exposing the metal core

The results showed that posterior monolithic CAD/CAM fabricated lithium-disilicate FPDs were shown to be fracture resistant with failure load results comparable to the metal-ceramic. Monolithic CAD/CAM generated lithium-disilicate FDPs with the investigated connector dimensions revealed high failure loads after fatigue and can be considered for selected posterior FDP indications. Bi-layer CAD/CAM generated lithium-disilicate FDPs were susceptible to low-load fracture failure. Metal-ceramic restorations have an inherent stress absorbing mechanism in the metal substructure that limits crack propagation, which explains the superior performance of this system compared to the bi-layer all-ceramic system. Also, the larger connector size resulted in a higher mean tolerance of loading. Low-strength veneering materials are prone to fail at low loads during the evolution of complex tensile fields in function. It is well known that the load to cause bulk fracture increases as the square of the thickness increases [26]. FEA studies additionally confirmed, that deleterious tensile stresses are significantly lower with full-anatomic FDP restorations (336 MPa) as compared to reduced framework designs (670 MPa) [26].

Soula-Ruiz [27] assess the 10-year longevity of E.max FPD's and found fracture failure rate was 28.6% after 10 years; a high percentage corresponded to connector fractures and occurred during the first 5 years. The authors concluded that lithium disilicate glass-ceramic FPDs present a higher risk of fracture than standard therapies (metal-ceramic).

However, a recent prospective study by Wolfart (2012) evaluated the clinical outcomes of 36 FPDs made from lithium disilicate glass-ceramic with 84% in posterior sectors and 16% in anterior sectors. A 7% fracture rate was observed over an 8-year period.

Manufacturers Recommendations

The maximum acceptable pontic width depends on the position, size and state of the teeth, as well as the position of the abutment within the tooth arch. The measurements to determine the bridge pontic width should be taken on the unprepared tooth. - In the anterior region (up to the canine), the bridge pontic width should not exceed 11 mm.

In the premolar region (from the canine up to the 2nd premolar), the bridge pontic width should not exceed 9 mm.

Preparations should adhere to basic ceramic guidelines - that is, no angles or sharp edges - shoulder preparation with rounded inner edges and/or deep chamfer preparation.
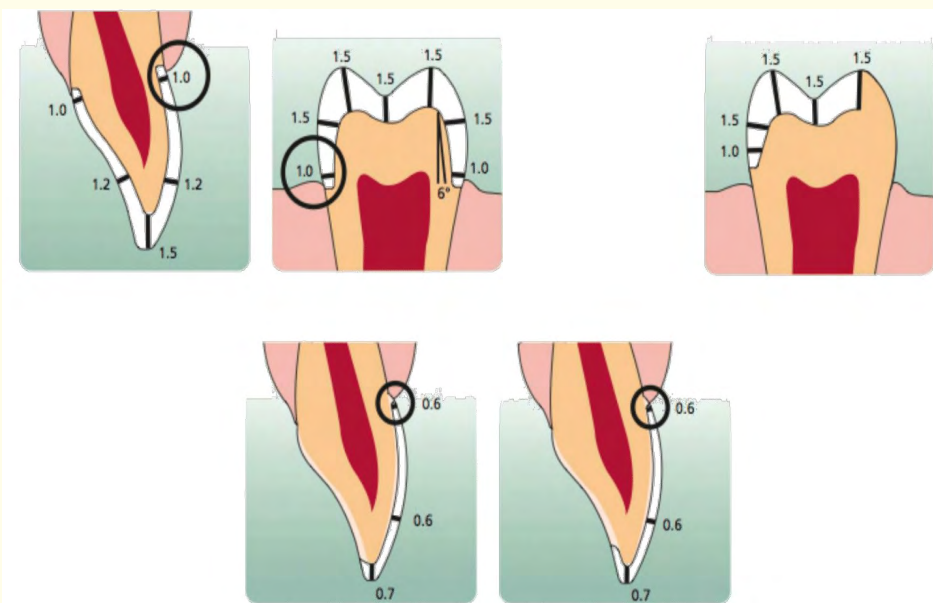
Cementation

Della Bona (2002) assessed microtensile bond strengths of composite resin to lithium disilicate ceramics, and found that silane coating of etched surfaces provided the highest and most durable bond strength values. This has been recommended by the manufacturer for all ceramic preparations. The strength of lithium disilicate, and most ceramics, comes from the bonding of the material to the tooth (as stated previously) [14,15].

Manufacturer recommended steps for adhesively bonding E.max preparations with Variolink II

Adhesive cementation of the crown involves the following steps:

- Conditioning of the restoration:
- Rinse the restoration with water and blow dry with an air syringe.

- Important: Glass-ceramic materials must not be sandblasted!

- Etch the inner aspects of the restoration with the hydro- fluoric acid IPS Ceramic Etching Gel for 20 seconds, thoroughly rinse with water and blow dry with the air syringe.

- Apply the silane Monobond-S to the inner aspects of the restoration for 60 seconds and air-dry.

- Subsequently, apply a thin coat of Heliobond and protect from light until the restoration is seated

- Rinse the preparation with water and blow dry with the air syringe.

- Etch the enamel with phosphoric acid gel (e.g. Total Etch) for 30 seconds. If required, etch the dentin surfaces for 10 - 15 seconds. Thoroughly rinse off phosphoric acid using water spray and blow dry with the air syringe.

- Apply a dentin bonding agent, eg Excite DSC dentin/enamel bonding system (note that Excite DSC combines primer and adhesive in 1 application).

- If using primer, let the primer react on the dentin for 15 seconds.

- Then thoroughly dry with the air syringe.

- Coat enamel and dentin surfaces with adhesive bond using a brush and remove excess with water spray/blow off with the air syringe. Important: Do not polymerize the adhesive, as this could detrimentally affect the fit of the restoration.

- Apply ready-mixed Variolink II luting composite to the inner surfaces of the restoration and/or to the prepared tooth if required (to avoid air entrapments). Seat the restoration.

- Remove gross excess using foam pellets and dental floss.

- Cover margins with glycerine gel (Liquid Strip) to prevent oxygen inhibition.

- Polymerize the seated crown from all aspects using a curing light.

- Occlusal adjustments should be performed using fine diamonds (30 micron).

- Polishing is carried out with ceramic polishing sets (eg diamond-coated ceramic polishers)

Bonding systems that are recommended with Variolink II

- Syntac is a time-tested multi-component adhesive system. An adhesive bond to dentin and enamel is established by consecutively applying Syntac Primer, Syntac Adhesive and Heliobond.

- Excite DSC is a dual-curing single-component adhesive with an innovative applicator.

Ceramics have a rich history in dentistry, playing an integral role in providing high-quality, natural-appearing restorations. From the early days of porcelain jacket crowns to the leucite-reinforced glass ceramics of the 1990s, to new 3rd generation monolithic lithium-disilicate glass-ceramic material, ceramic technologies have evolved and will continue to revolutionize modern-day esthetic dentistry.

All-ceramic systems are no longer experimental or suitable only for specialty practices. Clinical data and years of experience form the basis of the integration of all-ceramic restorations in clinical practice.

However, we have been more inclined to go with technology and use materials that, whilst have very promising *in-vitro* test results, have limited long-term clinical results.

But then it goes to pose the question, when is long enough to wait? If we wait too long, then will this material be surpassed by another material that has come into the market? These are questions that have no answers and often are dictated by personal experience, philosophies and results, in conjunction to the laboratory being used [28-33].

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