The shear bond strength of low fusing porcelain to zirconium oxide substructure compared to metal substructure. A literature review.

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Abstract: Statement of the problem: A major problem of the all-ceramic restorations that made it less reliable than metal-ceramic restorations is the mechanical integrity and the bond strength of the veneering porcelain to the ceramic substructure. Purpose: This review traces the historic evolution of the different metal and all ceramic systems, in terms of mechanical properties and bond strength of substructure to veneering porcelain. Materials and methods: literature covering almost 40 years of studies investigating the bond strength of the metal-ceramic and all-ceramic restorations was reviewed. A MEDLINE search was used to locate the articles. **Conclusion**: The search for the ideal all-ceramic material that has a good clinical performance and longevity will continue. This is very important from a clinical point of view as the selection of new materials should be based on a strong evidence to support the clinical application ceramics, Dicor, In-Ceram, Empress, Lava, Procera, Alumina, Zirconia.

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A literature review

Metal-ceramic restorations have been used successfully in dentistry for almost fifty years. Their good clinical fit, strength and durability resulted in their wide spread acceptance. Over that period of time, substantial improvements were achieved in metal alloys as well as the veneering porcelains. On the other hand, concerns regarding the biocompatibility of metals as well as aesthetic considerations started to rise.[1, 2]

Due to the increased demand for highly aesthetic and metal-free restorations, the all-ceramic restorations were introduced. When they were first introduced, the conventional porcelains that were used for the fabrication of the all-ceramic restorations had a high failure rate due to their insufficient physical properties. Hence, many attempts has been made to improve those properties.[3] The past decade witnessed a significant increase in the use of all-ceramic restorations as the strength became more predictable.[4] A major problem of the all-ceramic restorations that made it less reliable than metal-ceramic restorations is the mechanical integrity and the bond strength of the veneering porcelain to the ceramic substructure. The mechanical properties of core materials and veneering porcelains should be similar to achieve a durable bond. As for metal-ceramic bonds, previous studies reported that the shear bond strength is equal to the shear strength of the veneering porcelain. Therefore, the bond between the core and the veneering porcelain is very important for the success of the all-ceramic restorations. "Without documented evidence of the strength of the bond between the core and veneering porcelain, the profession must rely on manufacturers' claims to judge which material is best for patients".[5]

The search for the ideal all-ceramic material that has a good clinical performance and longevity will continue. This is very important from a clinical point of view as the selection of new materials should be based on a strong evidence to support the clinical application.[6]

Bond strength tests

Due to the complexity of the bonding mechanisms of dental materials, many test designs were developed to evaluate these mechanisms. Those tests can be classified according to the resultant nature of the stresses: shear, tension, shear-tension, flexure and torsion.[7]

The shear strength is "the maximum stress that a material can withstand before failure in a shear mode of loading", it is particularly important in the study of two material interfaces. There are different shear test designs: pull-through, push-through, planer interface, oblique and conical interface shear tests. Mechanically, the pull and push through are similar. In the push-through test, axial load is applied to push one material through another. The shear strength (τ) is calculated by: Shear strength (τ) = F / πdh. Where F is the compressive force applied, d is the diameter of the punch and h is the thickness of the specimen. [8]

Shell and Nielsen pull-through test consisted of discs of porcelain surrounding metal wires. Cohesive ceramic failure was observed when load was applied, so they modified the samples by tapering the ceramic gradually from the wire.[9, 10] Anthony et al further modified the samples to confine the failure to the bond by embedding the rod-ceramic portion of the sample in dental stone.[11] Asgar and Giday reported that embedding the samples in dental stone caused variations in the results unless a separator is applied to prevent error from friction between the metal rod and the dental stone. [12]

In the planer shear test, the ceramic and the metal are held in a jig and loaded in opposite directions until shear failure at the interface occurs. The interface can be either rectangular or circular. Failure of the ceramic-metal bond can occur in six possible locations. The highest strength specimen will display a cohesive fracture within the porcelain when tested. Fracture through the oxide layer and interfacial fracture are commonly observed with poor bonding. Fracture within the oxide layer is common especially in base metal alloys, whereas interfacial fracture is observed with metals that are resistant to forming surface oxides, such as pure gold or platinum. [13]

Anusavice *et al.* introduced the oblique shear test. The samples consisted of two metal blocks with porcelain fired in-between. The findings were questionable because of the difficulty in establishing an oblique sliding movement between the metal blocks.[14] Sced and Mclean described the conical interface shear test, by developing a conical metal surface surrounded by porcelain. Residual stresses were the problem.[15]

Many tests have been used to determine the bond strength between ceramics and metals but none is universally accepted because of lack of correlation between the test data and the clinical failures. In addition, it was possible that each test may not be measuring the strength of the adherence zone, but the mechanical properties of dental porcelain alone.[3]

Anusavice *et al.* analyzed eleven porcelain fused to metal bond tests for interfacial shear stress distribution using finite element stress analysis: the pull shear test, push shear test, conical interface shear stress, tension/shear test, three-point flexure test (under compression and tension), four-point flexure test, semi-circular arch flexure test, parallel shear test and modified parallel shear test (loaded at interface or distributed loading). Stress concentration effects were significant in ten of the eleven tests. A high probability of tensile failure within porcelain or the interfacial region was found in eight of the eleven tests analyzed.[16]

Kelly reviewed the traditional load to failure test and compared it to the actual clinical failure of all-ceramic restorations. The traditional fracture tests were inappropriate, because they did not resemble the failure mechanisms seen in retrieved clinical specimens. They did not create appropriate stress states and did not cause failure from clinically relevant flaws. The article described how to simulate the different intraoral conditions such as clinical contact stresses, cyclic loading, wet environment and crack systems. this will result in more clinically relevant tests and more reliable results.[17]

Metal-ceramic restorations

The 1950s and 1960s witnessed the development of metal-ceramic restorations: their alloys, veneering porcelains and investments. Important properties included ease of handling, thermal compatibility, high bond strength, sag resistance and low costs. Since then, remarkable advances in both the alloys and the veneering porcelains has taken place.

The most common mechanical failure of metal-ceramic restorations is porcelain debonding from the metal. Many factors control metal ceramic adhesion: chemical bonding, mechanical interlocking and residual stresses. The chemical bonding is mainly attributed to the formation of surface oxides on the metal e.g. Noble metals, which are resistant to oxidizing, must have other elements added to form surface oxides. The rough surface topography of the metal helps the ceramic to mechanically interlock with the metal and provide an increased surface area for chemical bonds to form. High residual stresses between the metal and ceramic can lead to failure if they have different thermal expansion coefficients. A difference of $0.5\text{x } 10^{-6}$ / ºC in thermal expansion between the metal and porcelain causes the metal to contract more than the ceramic during cooling, which puts the ceramic under slight residual compression making it less sensitive to tensile forces. Wetting angle, the contact angle between the porcelain and metal, is important for the formation and the strength of ceramic-metal bonding. Low contact angle is an indication of good wetting.[13]

Anthony et al tested the shear-bond strength of porcelain and high fusing gold alloys. The interfacial shear bond strengths was found to be as great as 20,000 psi. Chemical depletion from of the surface oxides and gold coating decreased the bond strength by 30% and 84%, respectively.[11] Hamerink *et al.* evaluated the porcelain to metal bond using a push-shear test and they reported values between 8,500 to 17,000 psi.[18] Mackert *et al.* strongly supported the oxide layer theory of porcelain-metal bonding in dental alloy systems.[19] Lubovich *et al.* tested the shear bond strength of precious, semiprecious, and nonprecious ceramic-metal alloys with two porcelains using a pull-through shear test. That test was found to be a reliable and reproducible method of testing the metal-ceramic shear bond strength. Results showed that the various alloys tested demonstrated significantly different bond strengths. [20] Another study compared the shear bond strength in porcelain-metal restorations using the push test and the pull test. The push test showed higher shear bond strength values than the pull test method. A bond strength of 4,500 to 6,000 psi between the dental stone and metal rod was eliminated in a waxed-rod test and it was believed to be true shear bond strength values. The shear bond strengths between porcelain and gold was 12,000 psi in push tests and 9,700 psi in pull tests.[21] Then they reported that the metal-ceramic restorations should be cooled at a slow rate, so that they have approximately the same cooling rate. The push-test shear bond strengths were always higher than pull-test shear bond strengths but both tests were equally valid for evaluating bond strengths in porcelain-metal restorations.[22]

Lorenzana *et al.* evaluated the flexural and shear bond strength of ceramic-four palladium based alloys, and compared them to the bond strength of ceramic-type IV gold alloy. Similar bond strengths were found but Palladium alloys containing up to 2% gold were the best.[23]

Chung *et al.* investigated the effect of the presence of a titanium-aluminum nitride film on the flexural bond strength between a dental porcelain and two nickel-based dental alloys using a three-point bending test. Bond strengths ranged from 159.0 +/- 11.7 N to 278.0 +/- 12.3 N. The (Ti,Al)N film provided an appropriate oxide layer for porcelain bonding and subsequently increased the flexural bond strength.[24]

Wagner et al investigated variables affecting the shear bond strength between porcelain and a palladium alloy. Aluminum oxide coating, preoxidation and surface roughening were found to improve the bond strength.[25] When the effect of seven different alloy surface treatments on the tensile bond strength of the porcelain-nickel-chromium alloy was compared, it was found that oxidation of the alloy prior to porcelain application increased the bond strength, while excessive grinding and steam cleaning of the metal surface decreased the bond strength.[26] Textured opaque porcelains have been introduced to improve esthetics of metal ceramic restorations. When their shear bond strength to six different alloys was compared to that of conventional opaque porcelain, no significant differences in was found.[27] The effect of plating nickel-chromium alloys with tin and chromium on porcelain shear bond strength was tested, and tin plating was found to increase the bond strength of porcelain to a nickel-chromium alloy while chromium plating did not.[28]

The effect of repeated firing on shear bond strength of different opaque porcelain-metal combinations was evaluated. No significant reduction in bond strength occurred during firing of the opaque porcelain to the metal in compatible porcelain-metal combinations.[29] Schweitzer *et al.* compared the debonding/crack initiation strength (D/CIS) of a low-fusing pressable leucite-based glass ceramic (PC) fused to metal to a feldspathic porcelain (FP) fused to metal using a three point bending test. Specimens were cast in either a base metal nickel-chromium alloy (BA) or a noble metal palladium-silver alloy (NA). The mean D/CISs, measured in MPa: NA-FP 32.56 (4.62), NA-PC 30.23 (5.06), BA-FP 30.98 (4.41), and BA-PC 31.81 (3.48). It was found that the D/CIS of PC to metal was not significantly different from that of feldspathic porcelain fused to metal.[30] Another study evaluated the effects of different bonding agents, surface roughness, and acid pickling on the bond strength of ultralow-fusing porcelain fused to gold alloy using a 3-point bending test. It was found that the type of the bonding agent, surface roughness, and acid pickling positively influenced the bond strength. [31] When examining the effect one and two layer opaque porcelain application on the flexural bond strength to three silver-palladium alloys, no significant differences were found.[32]

The effect of using gold as an intermediate layer was tested on the flexural bond strength of porcelain to palladium alloy. The toughness bond strength of the porcelain-gold-metal system was much higher than that of the porcelain-metal system.[33] Another study suggested that the mechanism of action of the gold coat on bond strength is because of its effect on the oxide layer on the surface of the base metal alloy. It was found that a thin gold coat increased the porcelain-metal bond strength and resulted in cohesive failure in the porcelain, while a thicker gold coat decreased the bond strength and resulted in adhesive failure.[34]

Shimoe *et al.* compared the relative shear bond strengths of ceramic and composite to metal (gold alloy) bonds before and after extensive thermocycling. The metal-ceramics exhibited the highest shear bond strength (40.5 MPa and 28.5 MPa respectively).[35]

During its development, titanium was found to be incompatible with conventional dental porcelains due to the weak bond strength due to titanium's high oxidative nature.[36] Kimura *et al.* investigated the oxidation effects on the porcelain-titanium interface tension-shear bond strength. Unlike the conventional ceramic-gold alloy system the recommended oxidation procedure was not suitable for porcelain-pure titanium restoration, since the excess thick layer of TiO2 weakened the bond strength of porcelain-titanium.[37]

Gilbert *et al.* tested the effectiveness of a new bonding agent on the shear and flexural bond strength of porcelain to milled titanium. It was found that when a titanium bonding agent was used, the bond strengths were significantly increased. [38] In another study, the shear bond strength of low-fusing porcelain to Ti-75 alloy was evaluated and compared to that of conventional porcelain fused to metal restorations. There was no significant difference between the shear bonding strength of both combinations, since the shear bond strengths of porcelain-Ti-75 alloy and conventional porcelain-metal restorations were 47.38 +/- 7.95 MPa and 48.50 +/- 7.60 MPa, respectively. [39] A study by Bergman *et al.* compared the metal-ceramic bond strength of a titanium copings-low fusing ceramic to those of a conventional noble alloy-medium fusing ceramic system. They found comparable bond strength in both systems.[40] The bond strength of commercially pure titanium was evaluated when subjected to porcelain firing under three-point bending mode with different surface treatments: sandblasting, mono- and triple-layered nitridation and mono-layered chrome-doped nitridation. It was concluded that mono-layered nitridation and mono-layered application of chrome-doped nitridation on both sandblasted and non-sandblasted surfaces were the most promising conditions for a successful Titanium-Porcelain System.[41] Another investigation found that surface treatment with Hydrochloric acid is a promising alternative to sandblasting of the titanium substrate in the titanium-porcelain system, since the bonds strength achieved were comparable to that of conventional metal-ceramic alloy system.[36] When the flexural and tensile bond strength were compared in Ti-porcelain and NiCr-porcelain, it was found that bond compatibility between titanium and porcelain was comparable with the NiCr-porcelain system.[34] The bond strength of porcelain to experimental cast titanium alloys (Ti-Cr, Ti-Pd, Ti-Ag and Ti-Cu) was compared to that of commercially pure titanium and a high noble gold alloy using a three-point bend test. There was a significant difference in bond strength only between the Ti-Pd and Ti-Ag alloys. The bond strengths for all the experimental alloys ranged from 29.4 to 37.2 MPa, which are above the minimum value required by the ISO specification (25 MPa).[42]

All Ceramic Restorations

In metal-ceramic restorations, the conventional feldspathic porcelains had a very low flexural strength of 60-70 MPa, which is one of the main reasons for the use of a metal substructure. [43] All-ceramic systems were developed to eliminate the metal substructure to provide better aesthetics, biocompatibility and durability. [44] In 1885, Land introduced the first jacket crown system. In 1964, McLean and Hughes suggested porcelain reinforcement with 50% aluminum oxide, which increased the flexural strength up to 100-130 MPa.[45] The all-ceramic systems became very popular among dentists and patients. Unfortunately, ceramics had a higher failure than metal-ceramic restorations. They were brittle, had low tensile strength and were prone to degradation in a moist environment.[44]

During the 1980s new systems were developed such as Cerostore, Dicor, Hi-Ceram, Cerapearl and Optec.[44] Numerous attempts have been made in order to improve the physical properties including the incorporation of high strength cores such as alumina and zirconia, or by the improved number, distribution and size of the crystals in the glass matrix. This resulted in the development of the In-Ceram system, Empress system, Procera system etc.

Ceramics can be divided into three general categories:

1. Predominantly glass: Dental ceramics that best mimic the optical properties of enamel and dentin have a high glass content. Manufacturers use small amounts of filler particles to control optical effects such as opalescence, color and opacity.

2. Particle-filled glass:Manufacturers add filler particles to the base glass composition to improve mechanical properties, such as strength and thermal expansion and contraction behavior. These fillers usually are crystalline, but they also can be particles of high-melting glasses that are stable at the firing temperatures of the ceramic. Often, it is these filler particles that are dissolved during etching to create micromechanical retentive features enabling bonding. Particles can be added mechanically during manufacturing as powder or be precipitated within the starting glass by special nucleation and growth heating treatments; in the second case, such materials are termed "glass-ceramics."

3. Polycrystalline**:** Polycrystalline ceramics contain no glass; all of the atoms are packed into regular crystalline arrays through which it is much more difficult to drive a crack than it is atoms in the less dense and irregular network found in glasses. Hence, polycrystalline ceramics generally are much tougher and stronger than glass-based ceramics. Well-fitting prostheses made from polycrystalline ceramics were not practical before the availability of computer-aided manufacturing.

It was reported that the compressive strength for Dicor was 828 MPa, modulus of rupture 152 MPa, Modulus of elasticity 70.3 GPa and microhardness 362 Kg/mm^2.[5] Although the reported flexural strength values for In-Ceram Alumina ranged from 236 to 578 MPa for the core material, those of the veneering porcelain were only about 90 Mpa. This low strength affected the core-veneer interface bond strength.[46]

Chu *et al.* investigated the effect of different surface roughness on the flexural bond strength of In-Ceram core/Vitadur Alpha self-glazed veneering porcelain. The samples evaluated were either self glazed, polished after self glazing or reglazed. It was found that reglazing significantly reduced the surface roughness, improved the surface texture and enhanced the flexural strength of the materials tested. [46]

Castable glass ceramics e.g. Dicor experienced ceramic shrinkage due to the casting and ceramming procedure. To overcome this problem, heat pressing technique was developed in 1983. [41] Leucite has been incorporated in porcelain used in metal-ceramic restorations to increase its coefficient of thermal expansion to become more compatible to that of the metal. Recently, Lucite was added to all ceramics as a reinforcing material. IPS Empress is a Leucite reinforced heat pressed ceramic. Heat pressing was reported to increase the flexural strength of the restoration by producing better dispersion of the Lucite particles in the glass matrix and promoting increased crystallization during firing. [47, 48]

Castellani *et al.* compared the fracture resistance of Dicor, Hi-Ceram and In-Ceram to metal ceramic restorations. They found that the In-Ceram specimens showed a significantly higher resistance to fracture than the other two all ceramic systems, but it was not significantly different from the metal ceramic ones. While the metal ceramic restorations exhibited cracking in the ceramic layer, the all-ceramic ones demonstrated an overall fracture. [49]

Giordano *et al.* compared the flexural strength of In-Ceram to that conventional feldspathic porcelain and Dicor ceramic. The In-Ceram core material had more than twice the flexural strength of feldspathic and Dicor ceramic, it was about 236.15 MPa ± 21.94 MPa. [50]

Wagner and Chu evaluated the biaxial flexural strength and fracture toughness of Procera AllCeram, In-Ceram and Empress. Their average flexural strength was found to be 687 MPa, 352 MPa and 134 MPa respectively. Both Procera AllCeram and In-Ceram had similar fracture toughness (4.48 MPa . $m^{1/2}$ and 4.49 MPa . $m^{1/2}$ respectively) which was significantly higher than that of Empress (1.74 MPa . $m^{1/2}$). [51]

Ohyama *et al.* investigated the effect of cyclic loading on the biaxial flexural strength of In-Ceram and IPS-Empress. The results suggested that although the In-Ceram system had higher flexural strength, it was more sensitive to flaws and more susceptible to fatigue fracture. IPS-Empress was less affected by fatigue. [52]

Chai *et al.* investigated the probability of fracture of four systems of anterior all-ceramic crowns when compressive load is applied. The 4 systems were: In-Ceram, In-Ceram CAD-CIM, IPS-Empress and Procera. There was no significant difference in the probability of fracture among the 4 systems studied. [4] Seghi *et al.* compared the fracture toughness of different ceramic materials including fluormica-, alumina-, Leucite- and zirconia-reinforced glasses. Alumina-reinforced materials resulted in the highest fracture toughness values. [53]

Rizkalla *et al.*compared the flexural strength, modulous of elasticity and true hardness of Vita In-Ceram alumina core and Vita In-Ceram matrix glass with the standard aluminous porcelain (Hi-Ceram and Vitadur), Vitadur N and Dicor glass and glass-ceramic. Vita In-Ceram alumina and IPS Empress 2 exhibited significantly higher flexural strength than aluminous porcelains and IPS Empress. The modulous of elasticity and true hardness of Vita In-Ceram alumina core were significantly higher than the other ceramic core materials. [54]

Curtis et al evaluated the effect of pre-cementation surface modification (alumina particle abrasion and surface grinding) on the flexural strength of a Y-TZP Lava dental ceramic in wet and dry conditions. It was concluded that with the alumina abrasion, a resultant combination of the reduced surface roughness and formation of compressive stress was identified, which increased the reliability of the bi-axial flexure strength. The presence of water did not really affect the performance of the Y-TZP ceramic. Coarse grinding significantly reduced the bi-axial flexure strength.[55]

Guazzato *et al.* compared the mechanical properties of In-Ceram Alumina and In-Ceram Zirconia. The mean biaxial flexural strength was found to be 600 MPa and 620 MPa respectively. The mean fracture toughness was 3.2 MPa . $m^{1/2}$ and 4 MPa . $m^{1/2}$ respectively when measured according to the indentation strength and 2.7 MPa . $m^{1/2}$ and 3 MPa . $m^{1/2}$ when measured according to indentation fracture. No difference in strength was found. In-Ceram zirconia was tougher only according to indentation strength.[56]

Carrier *et al.* evaluated the effect of the presence of excess glass infiltrate on the alumina core on the failure behavior of the all-ceramic restorations. It was found that excess infiltration glass on the core surface will not decrees the biaxial flexural strength of the In-Ceram structures. Microscopic evaluation showed decreased porosity at the core-veneer interfaces in the presence of excess infiltration glass, were most of the failures took place.[57]

Pure zirconia material undergoes phase transformation from a cubic, to a tetragonal, to a monoclinic phase. The cubic phase is only stable at very high temperatures of pure zirconia. The tetragonal phase may be stabilized at room temperature by adding 3-6 wt % Y_2O_3. The monoclinic phase is a low temperature stable phase. The tetragonal to monoclinic phase transformation exhibits volumetric increase of about 3-5 % which helps sealing the microscopic cracks. [2, 58]

Kosmac *et al.* compared the effect of surface grinding and sandblasting on the microstructure, flexural strength and reliability of yttria stabilized tetragonal zirconia (Y-TZP). They found that sandblasting increased the strength of the material and provided the highest amount of the monoclinic phase, while surface grinding lead to substantial strength degradation. [58] Boening et al reported that although the *in-vivo* fit of Procera AllCeram crowns was inferior compared to *in-vitro* studies, it was within the range of clinical acceptance with a marginal opening ranging from 80 to 245 µm. [59]

Studart *et al.* investigated the fracture behavior and strength of three veneer-framework all-ceramic fixed partial dentires: Empress 2/IPS Eris, TZP/Cercon S and Inceram-Zirconia/Vita VM7. It was found that cracks initiated at the veneer surface and propagated the framework. With the tough framework material (TZP/Cercon S and Inceram-Zirconia/Vita VM7), the cracks were deflected at the veneer-frameworkinterface, while weaker frameworks (Empress 2/IPS Eris) did not. They concluded by recommending the tough ones for posterior teeth restorations.[60]

Aboushelib *et al.* investigated the microtensile bond strength between zirconia and different veneering porcelain (pressable and layered ceramics), the effect of a liner material application between the core and veneer was also evaluated. The microtensile strength of Rondo Dentine and Lava Dentine veneer ceramics were significantly higher. Higher percentage of interfacial failure was reported. It was concluded that the liner material should only be used with some layering ceramics but not in with pressable ceramics because it resulted in a decrease in the microtensile bond strength.[61]

White *et al.* investigated the strength of layered zirconia-ceramics. Samples were made of a tetragonal polycrystalline zirconium dioxide partially stabilized with yttria core (Lava System Frame) and feldspathic porcelain (Lava Ceram veneer ceramic). Cracks were found to involve porcelain-zirconia interface, as well as bulk porcelain and zirconia. It was concluded that the layered zirconia-porcelain system tested had significantly higher tensile bond strength than previously reported for other layered all-ceramic systems.[62]

Tenschert *et al.* reported flexural strength and Weibull moduli values for different machined and laboratory-processed dental ceramic materials: Cerec Mark II, 86.3±4.3; Dicor, 70.31±2.2; In-Ceram Alumina, 429. 3±87.2; IPS Empress, 83.9±11.3; Vitadur Alpha Core, 131.0±9.5; Vitadur Alpha Dentin, 60.7±6.8; Vita VMK 68, 82.7±10.0; and Zirconia-TZP, 913.0±50.2. There was no significant difference in the flexure strength values of Cerec Mark II, Dicor, IPS Empress, Vitadur Alpha Dentin, and Vita VMK 68 ceramics. The highest Weibull moduli were reported with Cerec Mark II and Zirconia-TZP ceramics (23.6 and 18.4). Dicor glass-ceramic and In-Ceram Alumina had the lowest values (5.5 and 5.7). Except for In-Ceram Alumina, Vitadur Alpha, it was concluded that machined ceramics are more structurally reliable despite the fact that the CAD-CAM procedures may induce flaws that might have an adverse effect.[63]

Delamination of veneering porcelain from underlying ceramic substrates has been reported for all-ceramic restorations. Whether this phenomenon is due to a week bond between the core and the veneering porcelain or a fracture of the veneering porcelain was unknown. [5] Smith *et al.* investigated the In-Vitro fracture behavior of ceramic and metal – ceramic restorations. Failure in both systems was due to interfacial stresses occurring at the core-veneer interface. They stated that delamination was the primary factor in the fracture process during failure. [64]

In another study the shear bond strength of four individual veneering ceramics-three feldspathic and one fluoroapatite-to their corresponding core ceramics was evaluated: leucite-reinforced ceramic (Evopress); low leucite-reinforced ceramic (Finesse); glass infiltrated alumina (In-Ceram Alumina); and lithium disilicate (Empress 2) respectively. The shear bond strength in the Empress 2 system was significantly higher (41 ± 8 MPa) than those of the Finesse (28 ± 4 MPa), In-Ceram Alumina (26 ± 4 MPa) and Evopress (23 ± 3 MPa) systems. Thermocycling was found generally to decrease the bond strength. Although the failure mode was mainly adhesive at the core veneer interface for In-Ceram Alumina, predominantly cohesive fractures in the core materials were observed in the Empress 2, Finesse, and Evopress systems. Bilayered ceramic specimens exhibited complex failure modes that could be attributed to differences in the flexural strengths between the two ceramics, as well as to the differences in their thermal expansion coefficient. Although the thickness of the core ceramic was standard for all groups, it was reported that small variations could affect the strength of the restorations. Fluorapetite veneering ceramic demonstrated higher bond strength to lithium disilicate ceramic than the leucite-glass ceramic/feldspathic ceramic or glass-infiltrated alumina/feldspathic core-veneer ceramic combination did. After thermocycling core-veneer bond strength was affected the most in lithium disilicate/fluorapatite comdinations.[65]

Al-Dohan *et al.* conducted a study to investigate the shear strength of the substructure and veneering porcelain interface in all-ceramic systems: IPS-Empress2 with Eris (IE), Procera AllCeram with AllCeram (PA), Procera AllZircon with CZR (PZ), and DC-Zircon with Vita D (DC). A metal ceramic (MC) combination was tested as a control group. The mean shear strengths (±SD) in MPa were MC control 30.16 ± 5.88; IE bonded to Eris 30.86 ± 6.47; PZ bonded to CZR 28.03 ± 5.03; DC bonded to Vita D 27.90 ± 4.79; and PA bonded to AllCeram 22.40 ± 2.40. IE, PZ, and DC were not significantly different from the MC control. Microscopic examination showed that adhesive failure, or complete delamination, did not occur between the compatible ceramic core and veneering materials. Failure primarily occurred near the interface with residual veneering porcelain remaining on the core. IE with Eris exhibited cohesive failure in both the core and the veneer. The bond strengths of 3 of the tested all-ceramic materials (IE, PZ, and DC) were not significantly different from the control (MC) group.[5]

Another study evaluated the core-veneer bond strength of the components of two CAD-CAM ceramics; Cercon and Vita Mark II and one pressable system; (IPS)Empress 2. Standardized core specimens were fabricated according to the manufacturer's instructions, or polished with siliconcarbide paper. They were veneered with either its manufacturer's veneering porcelain or with a porcelain with higher thermal expansion coefficient (TEC). Specimens were subjected to the microtensile bond strength test. It was found that the core materials were significantly stronger than the veneering materials. Polishing the cores did not have an effect on the bond strength. Experimental veneer with higher TEC resulted in massive fractures in both the core and veneering material. They stated that the bond strength between the core material and the veneering porcelain is one of the weakest links of layered all-ceramic restorations and has a significant role in their success. To exploit fully the high strength of zirconium oxide cores, further research is needed to improve its bond with its corresponding veneering material.[66]

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