CRITICAL REVIEWS IN ORAL BIOLOGY & MEDICINE

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Performance of Dental Ceramics: Challenges for Improvements

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ABSTRACT

The clinical success of modern dental ceramics depends on an array of factors, ranging from initial physical properties of the material itself, to the fabrication and clinical procedures that inevitably damage these brittle materials, and the oral environment. Understanding the influence of these factors on clinical performance has engaged the dental, ceramics, and engineering communities alike. The objective of this review is to first summarize clinical, experimental, and analytic results reported in the recent literature. Additionally, it seeks to address how this new information adds insight into predictive test procedures and reveals challenges for future improvements.

KEY WORDS: dental ceramics, clinical success, fracture modes, testing protocols.

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INTRODUCTION

Excellent esthetics for crowns and bridges is possible with modern all-ceramic restorations. Yet their clinical survival is only now coming close to the "gold standard" of metal-ceramic restorations (MCRs). Understanding the factors influencing dental ceramics performance has drawn on the expertise of the dental, ceramics, and engineering communities. This review summarizes recent literature relating to ceramics used for full-coverage crowns and bridges. In doing so, it becomes evident that, despite improvements in material properties and excellent esthetics, the performance of all-ceramic restorations still fails to match that of their MCR counterparts. To shed light on this difference, this review addresses: how fractures initiate and propagate in all-ceramic restorations; the role that physical properties, in both pristine and damaged materials, play in clinical lifetimes of restorations; the influence of CAD/CAM shaping in a restoration's lifetime; how laboratory and clinical procedures can compromise restoration function; assessment of the quality of the veneer-core interface; implications of residual stresses resulting from mismatch between the coefficients of thermal expansion of core and veneer materials; the effects of restoration design features on clinical outcomes; and innovative shaping processes. The new information from the recent literature forms an important platform for defining predictors of future performance as well as revealing as-yet-unanswered challenges for future improvement.

CLINICAL PERFORMANCE OF MODERN ALL-CERAMIC RESTORATIONS

Performance of all-ceramic restorations is routinely measured against that of their esthetic predecessor, the MCRs, which are accepted as the "gold standard". However, measuring performance is not straightforward, and data concerning survival are not always consistent across studies. It is known that tooth position, patient factors (including gender, age, frequency of treatment, tooth history, recurrent caries, and periodontal factors), and dentist factors (including age, gender, and country of qualification) can all affect survival rates (Burke and Lucarotti, 2009; Malament and Socransky, 2010) – but these are not always reported. In addition, ceramics are continuously being improved, and handling conditions during fabrication and/or insertion can differ broadly, making it difficult to ensure the equivalence of success even within a single class of ceramic.

Not surprisingly, despite all these complications, clinical survival is a subject of intense interest and has been the focus of recent reviews. Meta-analysis of all-ceramic restorations fabricated from various types of materials (Pjetursson *et al*., 2007) found that for all positions in the mouth, densely sintered alumina crowns (Procera, Nobel Biocare, Göteborg, Sweden) had a five-year survival rate of 96.4%, quite similar to that of leucite-reinforced glass ceramic (95.4% for Empress, Ivoclar Vivadent, Schaan, Liechtenstein) and infiltrated glass ceramic (94.5% for InCeram, Vita Zahnfabrik, Bäd Säckingen, Germany), yet dramatically different from that of tetrasilicic fluormica glass ceramic (87.5% for Dicor, Corning, Ithaca, NY, USA). By comparison, 95.5% of lithium disilicate (IPS Empress 2, Ivoclar Vivadent) crowns survived for 10 years (Valenti and Valenti, 2009). Survival of posterior crowns is lower than for those on anterior teeth (*e.g*., 84.4% *vs*. 94.5% for Empress and 90.4% *vs*. 94.5% for InCeram) (Pjertursson *et al*., 2007). At 5 years, the rate of metal-ceramic crown survival was 95.6% (Pjetursson *et al*., 2007). It is interesting to note that one of the major failure modes of the metal-ceramic restorations is veneer fracture (Kinsel and Lin, 2009), and risks of fracture are higher for implant-supported restorations opposing other implant-supported restorations than on similar tooth-supported restorations (Kinsel and Lin, 2009), perhaps because higher loads can occur with implant-associated proprioception loss.

A recent systematic review (Wittneben *et al*., 2009) of publications appearing between 1985 and 2007 evaluated five-year survival of single-tooth restorations fabricated by computer-aided design and manufacturing (CAD/CAM) systems. The overall survival rate for the 16 studies meeting the inclusion criteria was 91.6% (95% confidence interval of 88.2-94.1%), which is not unlike the 93.3% (95% confidence interval 91.1-95.0%) reported for a combination of CAD/CAM- and conventionally produced methods (Pjetursson *et al*., 2007). But with the CAD/CAM systems, there were distinct differences between survival rates of different classes of ceramics. Interestingly, all but one of these studies focused on posterior tooth restorations. Failure rates for glass-ceramic restorations (18.18%) far exceeded those of feldspathic porcelains and alumina restorations (1.19%) (Wittneben *et al*., 2009). The restorations followed in these studies were fabricated by one of two CAD/CAM systems (Cerec 1 and Cerec 2 by Sirona Dental Systems, GmbH, Bensheim, Germany) and one copy-milling system (Celay, Vita Zahnfabrik, Bad Säckingen, Germany). Survival of restorations produced by these systems was equivalent.

Zirconia, specifically yttria-stabilized zirconia polycrystal (Y-TZP), an exceptionally strong ceramic, was introduced for use as a core in an attempt to eliminate bulk fracture of restorations common in those fabricated from other ceramic material. While exceptional on this front, with core failures only rarely occurring (*e.g*., 1 in 30 and 0 in 25, respectively) (Cehreli *et al*., 2009; Ortorp *et al*., 2009), esthetics veneers on this core material are prone to fracture. The extent of this problem is difficult to assess, in part because this type of problem is inconsistently reported in the literature as fracture or chipping. Explicit definition of the extent of the chipping is rarely provided, making determination of failure rates challenging. In some cases, the chipping is sufficiently expansive to require replacement, in others the need for replacement is left to the judgment of the clinician, and in still others it is repaired, polished away, or left. Unfortunately, the lack of consistency creates a reasonable degree of ambiguity about the extent of the problem.

An additional challenge revolves around the breadth of materials classified as zirconias, with strengths ranging from 786 MPa (not greatly different from those of glass-infiltrated alumina at 687 MPa) to over 1440 MPa (Thompson and Rekow, 2008), as well as implications of degree of densification during shaping (partially sintered, shaped then fully densified *vs*. fully densified at the time of shaping) and doping constituents (Denry and Kelly, 2008).

Few recent studies have reported survival of single-tooth zirconia restorations, but instead focus on performance of zirconia-based bridges (described below). Peer-reviewed studies of survival rates of all-ceramic single-unit crowns fabricated from different materials and by different approaches with different observations periods have recently been summarized (Della Bona and Kelly, 2008). In addition, performance of crowns fabricated from an experimental monolithic shrinkage-free $ZrSiO_4$-ceramic (HPC, high-performance ceramic; Everest HPC, KaVo, Biberach, Germany) was described in a randomized controlled trial (Encke *et al*., 2009), and the 12-month survival rate of 95.1% matched that of gold crowns. This material, with reportedly lower cost than other zirconias, can be used as a monolith, not requiring an esthetic veneer. Disappointingly, the authors did not address the quality of the esthetics that could be achieved.

The long span and high loads of posterior bridges pose high demands on all ceramic materials, which experience slow crack growth and can lose strength over time when exposed to repeated loading in a wet environment (fatigue in the oral cavity). Zirconia has become the material of choice for frameworks of all-ceramic bridges, and framework failure rates are much lower than those reported for glass-infiltrated alumina frameworks at 0-6% for zirconia at 3-5 years (Edelhoff *et al*., 2008; Molin and Karlsson, 2008; Tinschert *et al*., 2008; Silva *et al*., 2010a) *vs*. 10-12% for infiltrated alumina (Vult von Steyern *et al*., 2001; Olsson *et al*., 2003). But fracture of the zirconia framework cannot be entirely excluded (Taskonak *et al*., 2008b; Aboushelib *et al*., 2009; Sailer *et al*., 2009b). Bulk fractures of these frameworks, when they do occur, generally involve connectors of prostheses of 4 or more units or with second molar abutments (Denry and Kelly, 2008).

Zirconia-core bridges are vulnerable to the same veneer fracture as single crowns. Reports of incidence, fraught with descriptive inconsistencies, range from 8 to 50% (Sailer *et al*., 2007b; Della Bona and Kelly, 2008; Denry and Kelly, 2008; Tinschert *et al*., 2008). While a prospective study of zirconia-core bridges reported only a single major chip in 27 posterior bridges (Schmitt *et al*., 2009), a randomized controlled clinical trial reported more troubling results (Sailer *et al*., 2009a), although the survival rates of zirconia and MCR posterior bridges were the same at 3 years. Both types had chipping, but the chip size was much greater with zirconia cores, creating more unacceptable defects (minor chips in 19.4% of the MCR *vs*. 25% of the zirconia-based and extended veneer fractures only in the zirconia-based restorations).

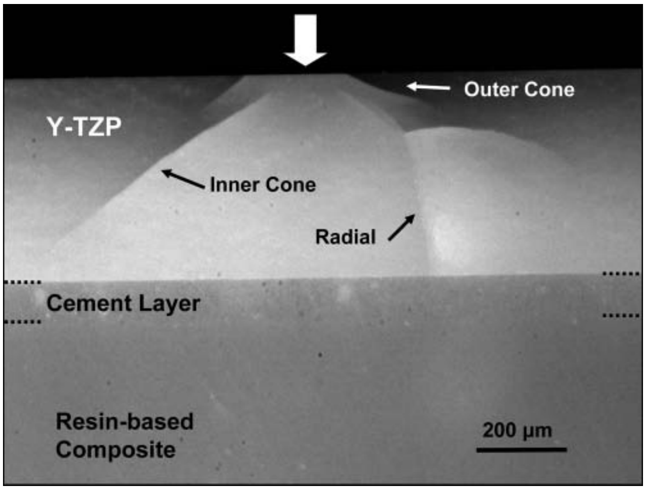

Figure 1. Cross-section of mouth-motion uniaxially loaded 0.5-mm-thick Y-TZP core following 110,000 cycles at 325 N with a 1.54-mm-diameter WC indenter in water. A tensile radial fracture from the as-machined and bonded (Panavia-21) intaglio surface intersects with a propagating inner cone crack, deflecting this inner cone. This indicates that the radial crack was present before the inner cone. Note the lack of propagation of the outer cone crack (from the study by Guess *et al.*, 2010).

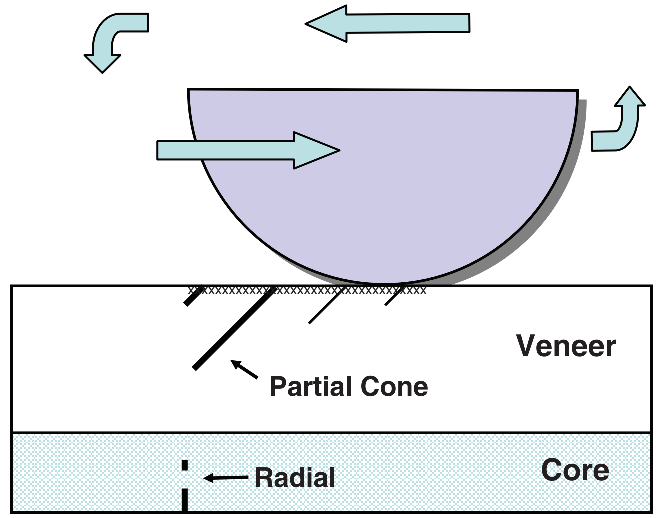

Figure 2. Schematic of possible crack evolution in veneered high-modulus core ceramics (alumina or zirconia). Friction-assisted partial cone cracks and surface wear develop beneath the indenter. One partial cone crack propagates toward the veneer core interface. Radial cracking from the intaglio surface of the core can develop. In sliding contact studies, radial cracks develop in nearly all of the alumina core specimens and rarely in less stiff zirconia cores (Santana *et al.*, 2009).

At recent professional meetings, clinicians have shown restorations fabricated from monolithic zirconia. By the use of different colors and unspecified shading techniques, esthetics in the few cases shown seems to be acceptable. To date, no laboratory or clinical reports relating to these monolithic zirconia restorations have appeared in the literature.

FRACTURE INITIATION AND PROPAGATION

Failures in all-ceramic restorations can initiate from several different sites on the surface, at interfaces, or within the material. They have been previously described extensively and recently summarized (Bhowmick *et al*., 2007; Thompson and Rekow, 2008). Remarkably, in dental ceramics, the first crack to initiate seems not to be the one that propagates and ultimately causes the restoration to fail (here defined as requiring clinical replacement). In laboratory studies with cyclic loading in water, the first crack to appear in nearly all dental ceramics is an outer cone crack, developing on the outer surface of the restoration in the stress field created by the loaded indenter (cusp). At subcritical cyclic loads, it does not progress to failure. Instead, other failure modes develop that ultimately lead to failure. In glasses (like feldspathic porcelains) and aluminas, this failure mode is usually a radial fracture, initiating from the cementation surface (tensile zone) of the core and propagating through the entire crown, leading to bulk fracture. In zirconia, radial fractures rarely occur. Instead of a radial crack, the second crack that develops could be an inner cone crack beneath the indenter/cusp. Inner cone cracks develop during loading as water becomes trapped in the cracks created by the expanding compressive field beneath the indenter. With each subsequent loading cycle, the water is driven deeper and deeper into the specimen (hydraulically assisted crack growth) (Kim *et al*., 2008), ultimately creating a fracture in the veneer. All 3 of these failure modes may be operational in a monolithic ceramic (Fig. 1).

Adding occlusion-like sliding (Kim *et al*., 2008; Zhang *et al*., 2008) concentrates the inner cone cracks into one large crack, oriented perpendicular to the sliding direction and penetrating deep into the ceramic (Guess *et al*., 2009; Santana *et al*., 2009). These "partial cone" cracks initiate and propagate at orders of magnitude fewer cycles than without the sliding component at similar loads (Kim *et al*., 2007; Kim *et al*., 2008). These hydraulically and friction-assisted partial cone cracks develop and, with continued subcritical loading in water, propagate (Bhowmick *et al*., 2007) (Fig. 2). Partial cone cracks initiated at similar numbers of cycles and loads in veneered alumina (Procera All-Ceram, Nobel Biocare) and zirconia (Lava Frame, 3M/ESPE, St. Paul, MN, USA) but propagated faster in alumina-based specimens (Santana *et al*., 2009).

Chipping

Veneer fracture is the leading cause of structural failure of zirconia-core restorations. Chips often originate in the wear facets (Scherrer *et al*., 2006) seen on veneered lithium disilicate, alumina, and zirconia (Della Bona *et al*., 2008; Etman and Woolford, 2010). In a randomized clinical study of three types of veneered cores (lithium disilicate IPS e.max Press, Ivoclar Vivadent; alumina, Procera AllCeram; and metal, Simidur S2, Weiland Dental, Pforzheim, Germany), cracks, initiating from the contact area, became evident long before fracture (Etman and Woolford, 2010).

Circumstances increasing the probability of chipping are described below, along with influences of fabrication on restoration success.

Slow Crack Growth

Dental ceramics, particularly porcelains, are vulnerable to slow crack growth. At low continuous or cyclic loads, especially in a humid environment (even at ambient conditions), a crack slowly but continuously grows in length, degrading the strength of the ceramic (Lawn, 1993). Cyclic loading in a humid environment permits crack propagation at stress levels in some cases of less than 50% of the initial material strength (Salazar Marocho *et al*., 2010).

The mechanism responsible for this loss of strength in dental ceramics is the combined effect of stress corrosion by water molecules at the crack tip and mechanical degradation of the polycrystalline dental ceramics (Freiman *et al*., 2009; Salazar Marocho *et al*., 2010). Of the non-zirconia core materials, glass-infiltrated alumina (InCeram Alumina) is the most susceptible, followed by lithium disilicate (IPS Empress 2) (Gonzaga *et al*., 2009). Of the veneering materials, leucite glass ceramic (Empress), leucite low-fusing porcelain (d.Sign, Ivoclar Vivadent), and an experimental high-fusing porcelain for alumina frameworks (VM7, Vita Zahnfabrik) are equally susceptible (Gonzaga *et al*., 2009). It is noteworthy that there are non-dental ceramics that increase their strength in aqueous environments; their reaction with water blunts sharp flaws (Taskonak *et al*., 2008a).

Y-TZPs are particularly vulnerable to this slow crack growth (Studart *et al*., 2007a-d; Salazar Marocho *et al*., 2010). For these, water depletes the yttria which had been incorporated to increase toughness (Taskonak *et al*., 2008a). In multi-phase alumina-zirconia-glass composites (In-Ceram Zirconia), the glass phase is the dominant factor controlling slow crack growth (Taskonak *et al*., 2008a; Salazar Marocho *et al*., 2010).

Leucite content has been thought to reduce slow crack growth, but in a study of 7 dental porcelains, this was shown not to be the case (Cesar *et al*., 2008). Leucite's failure to hinder slow crack growth may be related to tensile stresses in the glass matrix around the leucite particles, increasing the matrix inter-atomic spacing and weakening the inter-atomic bonding, making the region more sensitive to the effects of water (Michalske and Bunker, 1984).

Materials with high resistance to fast crack propagation (measured by K_{1c}) do not necessarily have higher resistance to slow crack growth. Fast cracking occurs at supersonic speeds, and in that case, water is unable to reach the crack trip as it propagates (Cesar *et al*., 2008). Ion exchange, substituting smaller sodium atoms for larger potassium ions in the porcelain outer layer, can reduce porcelain's vulnerability to slow crack growth (Rosa *et al*., 2009). This technique creates a thin (approximately 100 μm) compressive outer layer in the veneer. That, in turn, increases average flexural strength [by as much as 126% (Rosa *et al*., 2009)] and decreases slow crack growth [the stress corrosion coefficient (N) increased from 24.1 ± 2.5 to 36.7 ± 7.3 (Rosa *et al*., 2009)]. Unfortunately, the value of these improvements was tempered by a loss of reliability as measured by greater variation in strength.

Determining Damage Modes Related to Failure

Determining what caused a restoration to fail is complicated. The order in which competing failure modes develop can be difficult to determine, even in the laboratory, since most clinically relevant materials are relatively opaque. Establishing clinical timing of failures is even more difficult, since the patient or clinician may not be aware that a crack has initiated or even propagated until it has evolved to the point of failure. Even after a dramatic fracture has occurred, portions of the restoration may be lost or a patient may choose not to seek treatment for some time, so that fractographic clues are lost.

Analysis of clinical failures by fractographic techniques is gaining momentum, despite difficulty in securing pertinent evidence (Scherrer *et al*., 2007, 2008). Fractographic analysis of 19 clinically failed veneered zirconia crowns revealed that 10 failed by chipping of the veneer, leaving an intact core with cracks originating from the occlusal surface and propagating to the core-veneer interface, while another 6 originated at the core-veneer interface (Aboushelib *et al*., 2009). Associated stresses were calculated to be 31 ± 8 MPa and 23 ± 6 MPa, respectively, an extremely low value but similar to that reported for the posterior regions of the mouth (Lohbauer *et al*., 2008). This is one of the few analyses indicating that the chips involve the core-veneer interface; usually the failures are judged to be cohesive within the veneer.

Similar analysis of 17 clinically failed zirconia-core bridges (10 three-unit, 5 four-unit, and 2 five-unit) found that 8 had veneer fracture and 7 had connector fracture (Aboushelib *et al*., 2009). Stresses calculated as the cause of the veneer fracture dropped as the number of units increased (24 ± 9 MPa in three-unit but only 12 ± 1 MPa in four-unit bridges). Stresses relating to connector fracture increased with increasing numbers of units (656 ± 119 MPa for three-unit, 932 ± 90 MPa for four-unit, and 1368 ± 204 MPa for five-unit bridges). Single-cycle loading to failure of replicates of the bridges did mimic the connector failures but did not mimic the fractures in the veneer.

Another fractographic analysis of four-unit veneered zirconia core bridges (Cercon, Degudent, Hanau, Germany) ceramics retrieved from a clinical study (Taskonak *et al*., 2008b) determined that veneer fractures, occurring in 4 of the 5 bridges, originated at the gingival surface of the connector (associated stress was calculated to average 343 ± 3 MPa). The fractures were stopped at the veneer-core interface, and some included propagation along the interface, creating evidence of delamination. In all of the bridges, a second fracture initiation site was observed within the core layer, leading to ultimate failure. The calculated stress at failure for this second site fracture was 461 ± 49 MPa.

New Modeling to Understand Fracture Initiation and Propagation

Numerical analysis holds great promise for understanding the complex interplay between physical properties, competing damage modes, and the complex geometry of all-ceramic restorations. Three approaches were described in the recent literature.

Damage Initiation and Propagation with R-T^{2D}

A new numerical analysis system, R-T^{2D} code (Kou *et al*., 2007), initially developed to analyze fractures in rocks, can model non-homogeneous materials, including composites and materials with pre-existing weaknesses, including flaws. It permits step-by-step examination of initiation and propagation of different kinds of cracks, and the fragments that form during fracture in response to mechanical loads. Fracture patterns created in a three-unit zirconia bridge framework supported by stainless steel abutments analyzed with this software were similar to those created in a laboratory test of physical specimens.

Radial Cracks with Boundary Elements

Critical load for the initiation of radial crack damage is highly dependent on the flaw state within the material (Rudas and Bush, 2007). When loaded, structures can have both tensile and compressive zones within each layer, and these combinations can become complex in curved geometries like those found on posterior restorations. Cracks, once initiated, will propagate through the tensile region but be arrested in the compressive region. With boundary element analysis of dome-shaped structures subjected to both concentrated and distributed loads, the relationship of curvature to damage resistance and tolerance was elucidated (Rudas and Bush, 2007). In their findings, small dome radii led to greater susceptibility to catastrophic radial crack growth (smaller loads to propagate the crack are needed) from surface flaws. This, however, is contrary to the behavior of the critical load to initiate growth of intrinsic flaws. While reduced radius of curvature may lead to improved damage resistance (resistance to flaw growth), it reduces damage tolerance (ability to contain cracks).

Effects of Friction between Indenter and Specimen

Hertzian contact theory, widely used in most modeling, disregards the effect of friction during indentation (Jelagin and Larsson, 2008). Experimentally verified modeling/theory shows that: (1) difference in friction between the specimen and the indenter can significantly alter the load to initiate cone cracks; (2) the effect of friction has profound influence on the maximum surface tensile stress, and critical loads to initiate fracture were more than doubled compared with glass-to-glass contact when friction was taken into account; (3) tensile stress distribution changes dramatically, and the location of maximum stress shifts farther from the indenter contact area when friction is included; (4) friction results closely agree with experimental observations; (5) both the magnitude and shape distribution of tensile stress change when friction is integrated into the model; and (6) cone cracks with characteristic shape routinely form at unloading. This last point is quite surprising and has not been addressed in investigations of crack propagation in dental materials. With steel and tungsten indenters on float glass and one-time loads in nearly every single specimen (70 of 71 with steel indenters and 72 of 80 with tungsten indenters), more than 50% of the cone cracks formed within the first 20% of the *un*loading cycle. This is created by dissimilar elastic contacts that develop during unloading, whereby the frictional tractions change signs over part of the contact area, creating protective areas during loading and giving rise to damage peak tensile stresses when the load is removed. How this modeling is affected with repeated loading and by hydraulic pumping in cracks remains to be determined; such studies are important in consideration of clinical behavior.

PHYSICAL PROPERTIES OF MATERIALS

Physical properties of materials are important in determining the success of all-ceramic restorations, but alone they do not fully explain clinical behavior. Physical properties are usually given for materials in their pristine state, but as the literature published during this review period reveals, these properties may change, sometimes dramatically, when exposed to various environments. In later sections, we see that their properties can be even further diminished by various shaping, laboratory, and clinical procedures.

Initial Properties

Ceramics across an array of classes are currently being used for all-ceramic crowns and bridges. The classifications and physical properties have been described in detail elsewhere (Guazzato *et al*., 2002, 2004a,b; Thompson *et al*., 2007; Kelly, 2008; Thompson and Rekow, 2008). Here the latest literature relating to core and veneer material properties, thermal and chemical degradation, new raw material processing techniques, and innovations for improved damage tolerance is summarized.

Core Materials

Presently, 3 core materials seem to predominate in the literature: lithium disilicate, glass-infiltrated materials including aluminas and zirconias, and zirconia. Details of the microstructure and physical properties for the infiltrated materials and zirconia have been comprehensively described (Denry and Kelly, 2008; Kelly and Denry, 2008). How physical properties and toughening mechanisms (damage resistance) can be tailored during fabrication of the raw materials has been detailed (Kelly and Denry, 2008).

Homogeneity within materials is commonplace, as reported for lithium disilicate (IPS Empress 2) (Mitov *et al*., 2008) as well as infiltrated alumina and zirconia core materials (InCeram Alumina and InCeram Zirconia) (Salazar Marocho *et al*., 2010). Moreover, the close match of Weibull moduli of the infiltrated materials suggested great similarity in flaw state and stress distribution in the parameters tested (Salazar Marocho *et al*., 2010). It is noteworthy, however, that in at least one study (Mitov *et al*., 2008), the Weibull modulus measured experimentally, in this case that for lithium disilicate, was less than ½ that given in the manufacturer's literature, suggesting variability in flaws or residual stress states in the test specimens. Furthermore, materials from different manufacturers may have different physical characteristics (Della Bona *et al*., 2008).

Zirconia has become an extremely popular core material. Its mechanical properties are the highest for all dental ceramics

(Denry and Kelly, 2008), permitting the creation of ceramic structures historically not possible. Not surprisingly, they have been the focus of recent in-depth reviews (Manicone *et al.*, 2007; Denry and Kelly, 2008; Kelly and Denry, 2008; Silva *et al.*, 2010a). Zirconia's strength approximates that of steel (Garvie *et al.*, 1975; Manicone *et al.*, 2007). Although zirconia demonstrates strength similar to that of steel, its fracture toughness is ~ 9 MPa·m$^{1/2}$ compared with ~ 40 MPa·m$^{1/2}$ for steel. For at least one material (ICE Zircon, ZirkonZahn, Gais, Italy), strength did not depend on sintering time, though mean grain size increased with time, from 0.77 to 1.05 μm for longer sintering times (Hjerppe *et al.*, 2009). In addition to its exceptional strength, zirconia's intermediate elastic modulus (~ 780 GPa) provides advantages in layered structures by shifting damage and fracture modes into the porcelain veneer layer compared with fully dense alumina (~ 340 GPa) (Kim *et al.*, 2007). Interestingly, hardness, being a combination of elastic modulus and strength, is less for zirconia than for alumina (Lazar *et al.*, 2008).

One of the unique features of zirconia ceramics is their phase transformation. Properly managed in dental ceramics, this can enhance clinical performance. Unalloyed zirconia can assume 3 crystallographic forms, depending on temperature [monoclinic at room temperature to 1170°C, tetragonal from 1170°C to 2370°C, and cubic (2370°C to its melting point)] (Denry and Kelly, 2008). When stabilized with yttria, the high-temperature tetragonal (t) zirconia structure can be retained at room temperature. External stresses can transform the metastable t phase to the more stable monoclinic (m) phase. This t-m transformation has an associated volume expansion of 3-5%, which, in unalloyed zirconias, can lead to catastrophic failure (Denry and Kelly, 2008). However, through the addition of the stabilizing oxides like those in dental ceramics, compressive stresses and microcracks created around the transformed particles effectively oppose opening of the cracks and increase the resistance to crack propagation (Garvie *et al.*, 1975), resulting in the transformation-toughened material with improved strength (Kim *et al.*, 2010a; Tholey *et al.*, 2010).

Lifetimes of transformation-toughened ceramics are lower under cyclic loading than under equivalent static loading (Kelly and Denry, 2008). Under cyclic conditions, crack growth rates can be 7 orders of magnitude higher than for chemically assisted (water-enhanced) crack growth at equivalent crack-tip stress intensities. Interestingly, studies have demonstrated crack growth under cyclic conditions, the arrest of cracks in the same specimen held statically under load, and then resumption of crack growth when cycling resumes (Kelly and Denry, 2008). The apparent threshold for fatigue crack growth can be as low as 50% of the fracture toughness determined by static tests. This is an extremely important observation, since many investigators argue that the static and cyclic tests are equivalent; static tests could potentially grossly overestimate the long-term success of some ceramics.

The mechanical properties of 3 mol% Y-TZP (3Y-TZP) zirconias depend strongly on grain size (Tholey *et al.*, 2010), and grain size is dependent on sintering time (Hjerppe *et al.*, 2009). Above a critical size, zirconia is less stable and more vulnerable to spontaneous t-m transformation than with smaller grains (< 1 μm) (Tholey *et al.*, in press). Moreover, below a certain grain size (approximately 0.2 μm), the stress-induced transformation is not possible, leading to loss in fracture toughness (Tholey *et al.*, in press). The value of fracture toughness measured experimentally is highly dependent upon the notch that is used to initiate a fracture (Fischer *et al.*, 2008a).

The most commonly used dental zirconias, those described above, are yttrium cation-doped tetragonal zirconia polycrystal (Y-TZP) and glass-infiltrated zirconia-toughened alumina (InCeram Zirconia) (Denry and Kelly, 2008). A magnesium cation-doped partially stabilized zirconia (Mg-PSZ; Denzir-M, Dentronic, AB, Skellefteå, Sweden) is available, but porosity and large grain size have limited its success (Denry and Kelly, 2008).

Veneering Materials

Cracks leading to fractures of veneers can severely compromise the esthetics and function of all-ceramic restorations. This has been a particular problem with zirconia-based ceramics. Not surprisingly, then, the literature of the period being reviewed has focused on the physical properties of veneering materials in an effort to determine the variables to best enhance their clinical survival.

The influence of microstructure on the mechanical properties of porcelains compared with those of glass-ceramics and glass-infiltrated alumina has been detailed by Gonzaga *et al.* (2009). Flexural strength of veneering materials generally ranges between 60 and 120 MPa (Fischer *et al.*, 2008; Thompson and Rekow, 2008; Bottino *et al.*, 2009) (compared with > 450 MPa of core materials). The value determined experimentally can vary with test conditions (Fischer *et al.*, 2008; Mitov *et al.*, 2008).

Thermal (low-temperature) Degradation

The metastability of zirconia, significantly contributing to its high strength, also makes it susceptible to aging in the presence of moisture (Chevalier, 2006; Deville *et al.*, 2006; Benzaid *et al.*, 2008; Chevalier *et al.*, 2009a,b; Tholey *et al.*, in press). At relatively low temperatures (150 to 400°C), slow t to m transformations occur, initiating at the surface of polycrystalline zirconia and subsequently progressing into the bulk of the material (Denry and Kelly, 2008; Kelly and Denry, 2008). Transformation of one grain is accompanied by an increase in volume that results in stresses on the surrounding grains and microcracking. Water penetration into these cracks then exacerbates the process of surface degradation, and the transformation progresses from neighbor to neighbor. As the transformation zone grows, the extent of microcracking increases, grains pull out, and, finally, the surface roughens, ultimately leading to surface degradation.

Residual stresses promote low-temperature degradation (Denry and Kelly, 2008). Low thermal degradation in response to surface treatment was evaluated for 5 groups of identically prepared Y-TZP veneered specimens (IPS e-max ZirCAD, IvoclarVivadent) (as prepared, ground with 80-grit abrasive, ground with 120-grit abrasive, ground with 600-grit abrasive, and grit-blasted with 50-µm alumina at 0.5 MPa pressure for 5 sec from 10 mm) (Kim *et al*., 2010a). Microstructures appeared identical before thermal aging, but after aging, all had a grainy appearance, especially along the edges of scratches. All of the damaged surfaces had increased m-phase, which developed early. However, by 10 hrs of aging, the as-received caught up and ultimately surpassed the degree of transformation in all of the damaged specimens (55% m phase in aged as-received *vs*. 30% for grit-blasted, 20% for 80- and 120-grit-abraded, and 15% for 600-grit-abraded). Previous studies suggest that 30% of the initial fatigue strength is lost with 70% m phase present (Zhang *et al*., 2006). Surface treatment and associated residual stresses can have profound effects on hydrothermal degradation behavior.

Chemical Degradation

One of the appeals of ceramics is that they are deemed to be exceptionally stable. However, changes in slow crack growth rates for some porcelains exposed to acidic conditions have been reported (Pinto *et al*., 2008). Some diets, especially South Asian, include frequent consumption of highly acidic foods (Kukiattrakoon *et al*., 2010), and the question has been raised about the influence of such a diet on various ceramics. Different classes of ceramics (feldspathic ceramic, aluminous ceramic, high-leucite ceramic, and fluorapatite ceramic) were exposed to pineapple juice (pH 3.64), green mango juice (pH 2.39), citrate buffer solution (pH 4.99), and 4% acetic acid (pH 2.47) and compared with those stored in de-ionized water. The microhardness decreased significantly for all of the acid-immersed specimens. The decrease occurred quickly (within 24-96 hrs) for specimens stored in acetic acid and citrate buffer, even though the pH of the citrate buffer was not as low as that of other, less harmful, acids. The strength of the materials was also diminished, likely caused by changes in the material compositions as various ions were leached from the surface by the acids. Lithium disilicate was found to be inert in both strong inorganic acid and base (HCl and NaOH) in the glass and glass-ceramic state (ElBatal *et al*., 2009). These tests evaluated constant immersion without regard to *in vivo* shorter exposure times and salivary "clearance", but the results do suggest that all ceramics used in dental restorations may not be as chemically stable as we may have assumed.

New Processing Approaches

Many ceramic processing techniques are used to improve physical and chemical properties of zirconia-based materials. Co-precipitation from metal aqueous solutions has been shown to yield chemical and physical homogeneity in non-dental zirconia applications while being cost-effective and simpler than many alternatives to achieve similar results (Lazar *et al*., 2008). Applying this approach to synthesis of Y-TZP powder specimens created by pressureless sintering produced zirconias with higher toughness than both commercially available alumina (Procera All-Ceram) and zirconia-infiltrated alumina (In-Ceram Zirconia Block created both by dry pressing and slip casting). Co-precipitation approaches could make available even more appealing core materials. However, the current dental zirconias may already be adequate, since few fractures of these cores have been reported. Improvements in the toughness of the veneering materials are needed to substantially enhance clinical performance of zirconia-based all-ceramic restorations.

Graded Structures - Innovations for Damage Tolerance

One approach to improve ceramic clinical performance is to create more damage-tolerant systems. This can be achieved through graded structures, gradually changing the material composition across the core veneer interface and the core intaglio surface. In one set of studies (Zhang and Kim, 2009; Zhang and Ma, 2009; Kim *et al*., 2010b), zirconia was infiltrated with a silicate glass with a matched coefficient of thermal expansion. The percentage of glass changed from 100% to none across a 120-µm interphase. The resulting elastic modulus varied from 125 GPa at the infiltrate surface to 250 GPa at depth (Zhang *et al*., 2009; Zhang and Kim, 2009; Zhang and Ma, 2009; Kim *et al*., 2010b). While there was little change in toughness between infiltrated and non-infiltrated specimens, contact loads required to break bars infiltrated on both the top and bottom surfaces were nearly twice that of non-infiltrated bars of the same dimensions. It is noteworthy that the relative impact of the graded structures was greater for the thin specimens (Zhang and Kim, 2009; Kim *et al*., 2010b). By reducing the modulus in the near-surface regions, much of the stress in the specimen is carried by the stiffer material beneath the surface.

The graded structure eliminates the sharp interface now resulting from traditional core-veneer fabrication, eliminating the potential for delamination between the layers (Zhang and Kim, 2009; Kim *et al*., 2010b). Furthermore, the residual glass at the surfaces encapsulates the zirconia, impeding water absorption and thereby limiting the hydrothermal degradation (Zhang and Kim, 2009) described above. This approach opens promising new possibilities for the creation of thinner dental restorations. We await the results of fatigue studies on the graded ceramics.

INFLUENCE OF CAD/CAM SHAPING

An array of CAD/CAM systems has evolved since Duret introduced the concept in 1971. Miyazaki *et al*. (2009) summarized features of 11 different CAD/CAM systems, describing how digital data are acquired, types of restorations that can be produced, materials that can be shaped with the systems, and whether central machining centers are required.

The brittle nature of ceramics presents a challenge to machining. Most of the literature during the period being reviewed focused on the influences of CAD/CAM machining on zirconia, particularly Y-TZP zirconias. Some CAD/CAM systems (Denzir, Cadesthetics AB, Skellefteå, Sweden; DC-Zircon, DCS Dental AG, Allschwil, Switzerland) machine fully sintered Y-TZP blocks. Due to the hardness and poor machinabililty of fully sintered Y-TZP, the milling system must be extremely strong and stiff. The fine grain size of the fully sintered material leads to very smooth surfaces after machining (Denry and Kelly, 2008), but, not surprisingly, significant t-m transformation is associated with this process, increasing the degree of surface microcracking and the susceptibility to low-temperature degradation (Denry and Kelly, 2008).

The majority of the CAD/CAM systems shape blocks of partially sintered zirconia, eliminating the stress-induced t-m transformation, creating a final surface virtually free of the monoclinic phase, unless grinding adjustments are needed or sandblasting is performed (Denry and Kelly, 2008). Unfortunately, damage inevitably created during CAD/CAM processing is not fully healed by the final sintering process (Kim *et al*., 2010a). Evaluation of as-received CAD/CAM-shaped, then sintered, zirconia (IPS e-max ZirCAD, Ivoclar-Vivadent) specimens revealed a smear layer of flakes and wear debris coupled with extensive microcracking that penetrated 4-6 μm into the surface (Kim *et al*., 2010a).

Machining processes create characteristic trace lines. An important question is what impacts these irregularities, in combination with the microcracking, have on restoration survival. Roughness (R_a, R_p, and R_v) of as-machined zirconia is much greater than that of polished, polished and air-abraded, or ground specimens, yet they are not the weakest (Wang *et al*., 2008), though they were substantially less strong than that of polished specimens (820 *vs*. 1240 MPa) (Wang *et al*., 2008). The machine trace lines creating the roughness have only a few deep or sharp microscopic indentations that serve as fracture initiation sites. When thin (0.5 mm) bonded specimens were subjected to mouth motion Hertzian cyclic fatigue in water, cone cracks developed on the polished top surface beneath the indenter in as-machined and otherwise undamaged zirconia specimens (Guess *et al*., 2010). When the as-machined surfaces were damaged by air abrasion or grit-abrasion, the mode of failure changed. Instead of fractures originating on the top surface, fractures initiated from the bonded bottom surface (radial fractures) of the damaged specimens. The reliability dropped dramatically, from the as-machined and undamaged specimens having a 98% probability of surviving a 200-N load for 100,000 cycles (95% confidence interval of 0.90-0.99) to 40% (CI 0.25-0.56) for the as-machined and damaged specimens (Guess *et al*., 2010).

The strengths of as-machined specimens of 3 classes of zirconias (fully dense before machining, DC Zircon; partially sintered before machining, Lava and Cercon; and zirconia-reinforced alumina, InCeram zirconia) were compared (Chai and Chong, 2009). Of these, the zirconia that was fully dense before machining had the greatest strength, followed by one of the partially sintered at machining (Lava 3M-ESPE), and then the other partially sintered and zirconia-reinforced alumina (which were not statistically different). The superior behavior of the fully dense zirconia could be related to the smoother surface and small grain size of that material (Denry and Kelly, 2008).

Poor marginal adaptation of restorations increases plaque retention, potentially leading to secondary caries and periodontal disease, and, through microleakage, contributes to endodontic inflammation (Beuer *et al*., 2009a). Computer-aided design of restorations has several potentially accuracy-compromising aspects: during data collection, locating the margin in the digital representation, and restoration design. Computer-aided manufacturing, especially of ceramics, also poses some difficult challenges: accuracy of restoration fit related to shaping the complex surfaces, and irregularities in the surface caused by cutting paths, coupled with the 15-30% shrinkage associated with post-machining sintering of the partially sintered blocks (Reich *et al*., 2005; Sailer *et al*., 2007a). Three of the CAD/CAM systems (Etkon, Etkon AG, Graefelfing, Germany; Cerec InLab, Sirona, Bensheim, Germany; and Cercon, DeguDent, Hanau, Germany) investigated were able to produce marginal gaps beneath the conventional 120-μm marginal gap threshold (Beuer *et al*., 2009a). However, there was considerable variation in both the accuracy of the fit by different systems and the technician's time and manipulations required to achieve that fit.

INFLUENCE OF LABORATORY AND CLINICAL PROCEDURES

Laboratory and clinical procedures can influence the strength of all-ceramic restorations, sometimes dramatically. Unavoidable fabrication damage can occur from machining, occlusal adjustment, modification of the shape of the internal crown or abutment surface to remove anomalies that interfere with restoration fit, and air abrasion believed to enhance bonding. Damage induced by these procedures, even when it is microscopic, creates surface flaws that act as stress concentration sites and become sites for crack initiation and dramatically reduces strength and fatigue life (Zhang *et al*., 2006). Damage introduced during laboratory and clinical processes cannot always be eliminated, leading to premature failure (Salazar Marocho *et al*., 2010). Recently addressed processes that can influence clinical performance include veneer application as well as roughening or grinding of zirconia.

Veneer Application

The veneering process often includes wet thick layers of porcelain being applied, dried, and sintered onto the zirconia (Tholey *et al*., in press). This facilitates the t-m conversion with its associated volume dilation creating residual stresses in the zirconia. Applying the veneer without a liquid medium has eliminated this transformation (Tholey *et al*., 2009), strongly emphasizing the role of moisture in the porcelain powder (*vs*. the powder itself), causing the transformation in the zirconia core. The residual stresses associated with the transformation, in addition to those caused by mismatch between core and veneer coefficients of thermal expansion, are thought to contribute to the increased probability of veneer chipping.

A potential for marginal fit distortion develops during veneering, but was found at only a minor level in a series of core-veneer-CAD/CAM systems recently evaluated (Kohorst *et al*., 2010). However, if the misfit creates an interference that is removed by the laboratory technician or clinician, the restoration is damaged, and its long-term survival is potentially compromised.

Roughening Zirconia

Zirconia surfaces are roughened to improve bonding to the adhesive, improve core-veneer bond strength, and improve fit and occlusion. This roughening process, whether by air abrasion or grinding, damages the material, making it more vulnerable to fracture.

Air-borne particle abrasion damages ceramic surfaces, creating sharp scratches, cracks, grain pull-out, and material loss (Wang *et al*., 2008; Guess *et al*., 2010), but the extent of the damage depends on the abrasive conditions. Air-borne particle abrasion with 50-μm alumina oxide particles is less severe than that created with 120-μm particles delivered at the same pressure and offset distance (Wang *et al*., 2008). In zirconia, air abrasion triggers the t-m conversion. With 50-μm alumina oxide, the compressive fields created by this transformation increased the strength of the zirconia (Wang *et al*., 2008), because the surface became smoother than that of the as-received CAD/CAM surfaces, weakly attached surface grains were removed, and other imperfections concentrating stresses that serve as crack initiation sites were eliminated.

This compressive layer, however, is quite thin, and its strengthening effects can easily be counterbalanced by a plastically deformed zone with high incidence of randomly oriented plough marks and grooves (Kim *et al*., 2010a) and high density of microcracks both on and below the surface (Lawn *et al*., 2004). Subsurface cracks, typically 2-4 μm below the surface, can propagate laterally and eventually intersect with the surface (Kim *et al*., 2010a), roughening it and compromising the strength of the specimen.

Sharp and deep defects increase the stress concentrations at their crack tips and are, therefore, more likely to become crack initiation sites. Surface characterization with R_v, which reports the surface depressions measured from the estimated surface, best represents the degradation of the surface as it relates to strength degradation (Wang *et al*., 2008). R_v is highly correlated with fracture strength of both polished and damaged ceramics (correlation coefficient of > 0.9). Neither R_a (which describes average surface roughness as the mean of the elevations and depressions measured from an estimated surface) nor R_p (which represents the average vertical elevations measured from the estimated average surface) is as well correlated. As a reference, surface roughness of only 0.3 μm can be detected by the tip of a patient's tongue (Jones *et al*., 2004).

Survival probability is directly related to the degree of surface damage created (Wang *et al*., 2008). But even the smallest damage can become troublesome in mastication cyclic loading, where small cracks tend to grow until they reach a critical length, resulting in catastrophic failure. As noted above, as-received then damaged 0.5-mm zirconia plate specimens had only a 40% probability of surviving a 200-N load to 100,000 cycles compared with a 98% probability for as-received but not damaged specimens (Guess *et al*., 2010). The damaged specimens failed from radial cracks initiating from the treated cementation surface, whereas the undamaged specimens failed primarily from the top surface by deep-penetrating cone cracks initiated from the beneath the indenter.

Grinding of surfaces induces damage. The damage created with coarse burs [grit size of 106-125 μm, approximately equivalent to that of Blue, Komet, Stuttgart, Germany (Kim *et al*., 2010a)] in glass ceramic (Vita Mk II, Vita) is extensive, propagating as deep as 114 μm beneath the surface at cutting speeds and depths similar to those used clinically (Song *et al*., 2008a,b; Song and Yin, 2009). Furthermore, subsurface damage increases linearly with cutting speed (the speed at which the bur moves across the surface at constant rpm) and depth of cut (Song and Yin, 2009).

Even less aggressive grinding compromises the surface of ceramics. Grinding zirconia (IPS e-max ZirCAD) with 80-, 120-, and 600-grit abrasive created long scratches (Kim *et al*., 2010a). The 120-grit generated cracks that penetrated 2-4 μm into the surface and extended several tens of microns laterally (Kim *et al*., 2010a). Even though there were no significant microcracks observed on 600-grit surfaces (Kim *et al*., 2010a), the reliability of those specimens was low and equivalent to that of those subjected to air-borne abrasion (Guess *et al*., 2010).

It has been theorized that thermal firing after air-borne particle abrasion could "heal" any damage. Unfortunately, this was not found to be the case for zirconias (Wang *et al*., 2008). Thermal firing, either before or after particle abrasion, had no significant effect on flexural strength. Post-abrasion firing did increase the Weibull modulus of polished zirconia specimens, but did not change the modulus for as-received then abraded CAD/CAM-shaped (subsequently sintered) specimens (Wang *et al*., 2008). Previous studies have shown that sandblasting core ceramics can reduce their polished strengths by as much as 30% for zirconia (Zhang *et al*., 2006). Advocates of sandblasting suggest that heat treatment can reverse any damage created. Heat treatment can reverse the t-m transformation and the residual stresses it creates (Denry and Kelly, 2008). However, while thermal treatment of zirconia at 1200°C for 2 hrs induces relaxation of stresses and lower susceptibility to thermal aging to levels even below those of a polished specimen, it unfortunately does not provide a mechanism for healing flaws like those introduced by sandblasting. In zirconia, the flaws remain, serving as probable sites of fracture initiation (Denry and Kelly, 2008). By contrast, cracks have been shown to heal by glass infiltration during thermal treatment of high-purity alumina (Ceralox, SPA-RTP-SB, Sasol, Tucson, AZ, USA) (Fischer *et al*., 2008b).

VENEER/CORE INTERFACE

The integrity of the interface between the core and veneer can influence clinical performance in several ways. As for porcelain fused to metal, fractures of porcelain veneered on zirconia are generally cohesive, remaining within the veneer and only rarely involving the interface, suggesting that the bond itself is adequate (Guess *et al*., 2008). Yet, fractographic analysis of failed clinical specimens indicated that veneer-core interface failures can occur (Aboushelib *et al*., 2009). The bond strengths to zirconia cores (Cercon CeramS on Cercon Base, Vita MV9 on Vita InCeram YZ Cubes, IPS e.max Ceram on DC-Zircon) were lower than those of metal ceramics (Vita VM13 on Degudent U94). One approach thought to improve the bond strength is to sandblast the outer surface of the core before applying the veneer. This was not found to be the case in a study of 3 zirconia-based all-ceramic systems (Guess *et al*., 2008). There was no difference in bond strength between cores that had been roughened (Cercon base/Cercon Ceram S, Vita InCeram Cubes/

Vita VM9, and DC0Zircon/IPS e-max Ceram) and those that had not (InCeram YZ Cubes/Vita Mk 9 and DC-Zircon /IPS e-max Ceram).

CORE AND VENEER COEFFICIENTS OF THERMAL EXPANSION MISMATCH

The majority of all-ceramic restorations rely on veneers to achieve clinically acceptable esthetics on high-strength cores. Conventional wisdom from metal-ceramic restorations informs us that veneering ceramics should have a slightly lower coefficient of thermal expansion (CTE) compared with that of the core, creating compressive stresses in the weaker veneering ceramic and thus enhancing the overall strength of the restoration (Aboushelib *et al*., 2008). Unfortunately, application of this principle to ceramics has not been as successful, evidenced by high rates of veneer chipping in all-ceramic bilayered restorations.

A mismatch between zirconia core and veneer CTE creates stress fields throughout the restoration. High loads to failure can be obtained when the CTE of core and veneer match with radial fractures developing at the bottom of the veneer as the primary failure mode (Aboushelib *et al*., 2008). As the CTE of the veneer becomes increasingly greater than that of the core, the load to cause failure decreases. The combination of loading stresses and tensile pre-stresses at the zirconia-veneer interface caused delamination at less than their theoretical failure load. With veneer CTE less than that of the core, immediate cracking and delamination could occur during the cooling phase of the veneering process (Aboushelib *et al*., 2005). A similar study with veneers on high-fracture-toughness ceria-stabilized zirconia/alumina nanocomposite (Ce-TZP/A) confirmed these findings (Fischer *et al*., 2009).

Interfacial toughness of the bond between the core and veneer is incredibly sensitive to CTE matching. In flat specimens, glass veneers (IPS e.max Ceram) were added to lithium disilicate (IPS e.max Press), zirconia (IPS e.max ZirCAD), and to itself (Anunmana *et al*., 2010). The difference between the CTE of glass and lithium dissilicate is 0.65 ppm/K (9.5 and 10.15 ppm/K, respectively). The mean load to failure and interfacial toughness for glass on glass and glass on lithium disilicate were not significantly different (41.9 ± 9.6 N and 38.9 ± 9.6 N; 0.74 ± 0.17 MPa·$m^{1/2}$ and 0.96 ± 0.11 MPa·$m^{1/2}$, respectively). The difference in CTE between the glass and zirconia was only slightly higher at 1.25 ppm/K (9.5 and 10.75 ppm/K, respectively), but even this slight difference resulted in a dramatic reduction in mean load to failure and interfacial toughness (7.1 ± 4.4 N and 0.13 ± 0.07 MPa·$m^{1/2}$) for the glass on zirconia. This remarkable difference in flat specimens is likely amplified in the cooling of complex geometries like dental restorations.

A laboratory analysis with flat specimens found that resistance to edge chipping was similar for zirconia-based all-ceramic and metal-ceramic restorations (Ceramco PFZ on Cercon, and Ceramco3 on Ultracrown SF Alloy, Dentsply, York, PA, USA) (Quinn *et al*., 2010). This surprising finding suggests that the problem of chipping is more complex than just mismatch between CTE. Swain (2009) presents a fundamental analysis of the causes of residual stress in ceramics that could be associated with veneer chipping. For simplicity, only flat, bilayer geometry is considered, but this comprehensive analysis provides theoretical explanation of thermal expansion mismatch stresses, thermal conductivity of bonded structures, the magnitude of thermal tempering residual stress, and contact-induced fracture of thermally tempered plates. Zirconia's low thermal diffusivity results in higher temperature differences and very high residual stresses compared with those of alumina, even in flat plates.

Crack extension occurs when the driving force of the crack exceeds the toughness of the material (which, for porcelain, is only ~ 1 MPa·$m^{½}$) (Swain, 2009). The predominant factors driving the cracks are cooling rate, coefficients of thermal expansion, and thickness of the porcelain (which predominates as a 5/2 power).

DESIGN FEATURES

Design of a restoration is largely driven by the clinical requirements of the patient and the materials to be used. Within those limitations, however, there exist opportunities for different margin finish lines, pontic designs, and core configurations (*e.g.*, constant-thickness core *vs*. constant-thickness porcelain).

It might be conjectured that margin design can influence fracture resistance of all-ceramic full-coverage restorations. Knife-edge finish lines in zirconia copings supported on stiff metal abutments had a greater single-cycle load to failure than those with chamfer finish lines (1110 ± 175 N *vs*. 697 ± 126 N at 0.5-mm-thick copings and 730 ± 160 N *vs*. 455 ± 79 N for 0.3-mm copings) (Reich *et al*., 2008). Whether these findings would change with clinically realistic low-elastic-modulus abutments remains to be determined. Others (Clausen *et al*., 2010) found no differences with different margin designs.

Pontic design, especially the dimensions of the connector, has been addressed in great detail previously, but has not recently been the focus of much attention. Only a single two-dimensional numerical analysis of fracture initiation and propagation appeared. In it, flattening and lengthening the pontic of a three-unit zirconia-core bridge shifted the crack initiation site from the connector region to the middle of the lower portion of the pontic (Kou *et al*., 2007). While interesting on a theoretical basis, the clinical application of this finding, where design is driven by the opposing occlusion, may be limited.

Investigators and clinicians are exploring different core designs. The hypothesis is that chips and fractures of veneering porcelains develop in areas where the veneer is largely unsupported by the core. A comparison of zirconia crowns (Cercon Base) with constant-thickness core, modified core with some additional veneer support, and "optimized occlusal support" confirmed this hypothesis (Rosentritt *et al*., 2009b). After mild thermal cycling and 1.2 million loading cycles at 50 N, veneers on all the configurations chipped. There were fewer and smaller chips in the modified and "optimized" designs. Surprisingly, more chips occurred in the "optimized" substructure design than in the modified design, suggesting that there is room for optimization (though the numbers of specimens were quite low in this study). Experiments in our laboratory confirm this result (Bonfante *et al*., 2010; Lorenzoni *et al*., 2010): Veneers supported by anatomically

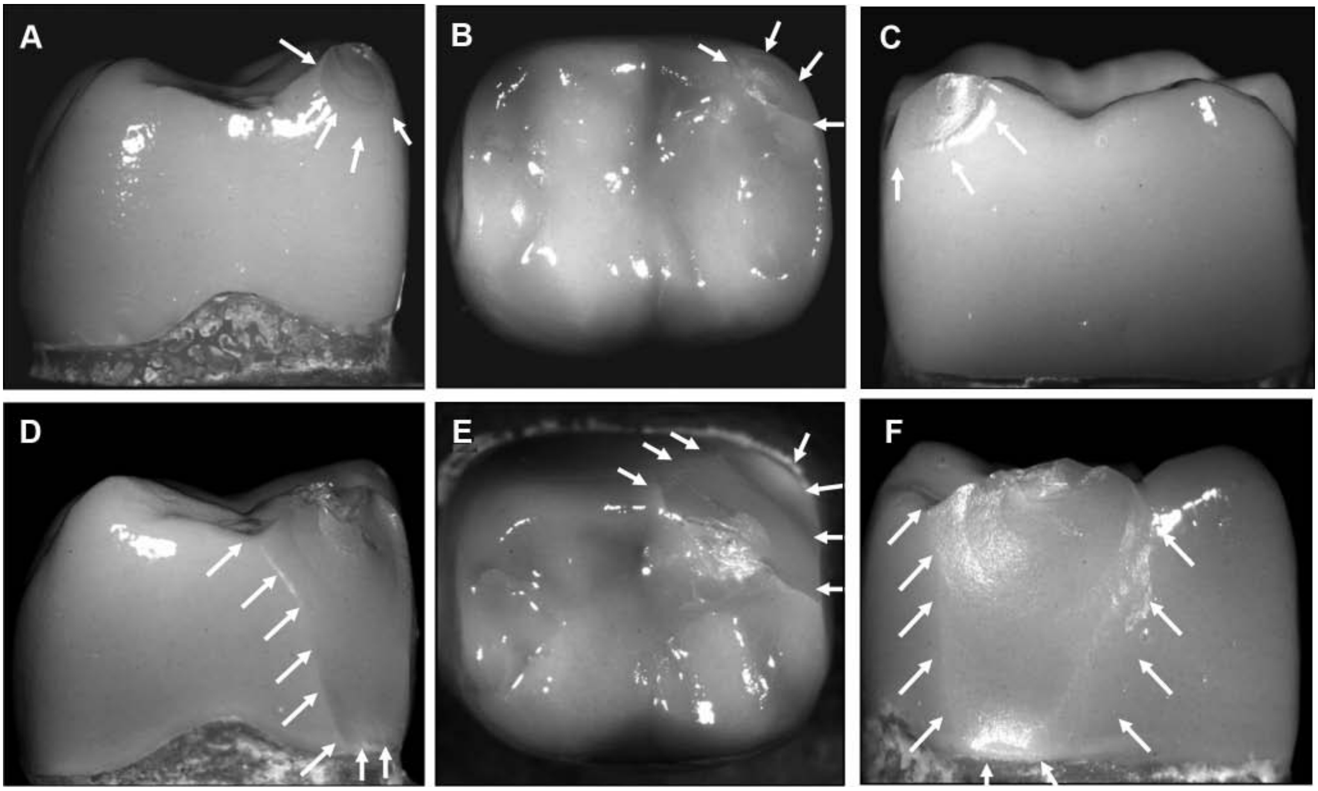

Figure 3. Mouth-motion step stress fatigued veneered Y-TZP crowns exhibiting typical veneer chipping failure mode. In views **A-C**, an approximately uniform 1.0-mm core was veneered with 0.5-1.0 mm porcelain, while in views **D-F**, a 0.5-mm core had a 1.0- to 1.5-mm veneer applied. The anatomic core support resulted in smaller chips which developed at higher loads than for the less supportive thinner core. Note that the veneer-core interface is rarely exposed. These laboratory test results parallel clinical reports.

shaped cores had fatigue-induced chips that were smaller and initiated at higher loads than those supported by constant-thickness cores (Fig. 3). The dental laboratory literature contains at least one reference to this design (Anonymous, 2010), but no manuscripts in the refereed literature were found describing performance in response to fatigue loading.

INTERACTIONS BETWEEN AND AMONG VARIABLES

A host of variables considered individually have been reported to influence restoration performance. Among these are core and veneer materials and thickness, cement modulus and thickness, proximal axial wall height, and loading conditions. However, these may not be independent variables. It is highly likely that there are interactions between and among the variables that could influence the distribution of stresses within a restoration, particularly those in the veneer on zirconia restorations. This possibility was explored by finite element analysis combined with factorial analysis for 64 combinations of these variables (Rafferty *et al*., 2010a,b).

Depending on the combination of variables, the maximum principal stress in the veneer ranged from 248 to 840 MPa. Factors that can be considered as independent variables (main effects) were found to be cement thickness (stress is higher with thicker cement layers) and loading (vertical creates less stress than only vertical plus horizontal components to the load) (Rafferty *et al*., 2010a,b). There were interactions between and among many combinations of variables, including core thickness and cement thickness, cement thickness, and proximal wall height. This factorial approach also uncovered important factors that, surprisingly, did not greatly affect the maximum principal stress within the veneer, including core material (alumina *vs*. zirconia) (Rafferty *et al*., 2010b), confirming previous findings of others (De Jager *et al*., 2006). Interesting differences developed when investigators considered multiple combinations of variables *vs*. assuming that each is an independent variable. For example, when shortening the interproximal wall to reflect typical clinical situations (*vs*. axisymmetric equal length axial walls) was considered as an independent variable, it was shown to have a significant impact on stress distributions within the veneer (Coelho *et al*., 2009a,b). Analysis of multiple variables and their interactions suggests that it does not (Rafferty *et al*., 2010a,b). Analyzing variables only as independent could dramatically limit conclusions that can be drawn and could be misleading. (It should be noted that adhesive cementation is also an important variable, but in-depth discussion of that topic warrants a separate review.)

NEW SHAPING STRATEGIES

Currently, commercial CAM fabrication technologies are just beginning to create separate core and veneer layers that could then be joined. A CAD/CAM-fabricated lithium-disilicate glass-ceramic veneering created and then sintered onto a CAD/CAM-fabricated zirconia coping improved its strength [single-cycle loads to failure increased (6265 ± 2257 N for CAD/CAM-produced veneer *vs*. 3700 ± 1239 N for layered and 3524 ± 1181 N for pressed copings] (Beuer *et al*., 2009b). Importantly, the failure mode shifted away from the veneer; instead, there was a higher frequency of bulk fractures in the CAD/CAM-produced veneered systems. This may reflect the higher flexural strength and elastic modulus of the CAD/CAM veneer (360 MPa, ISP e.max CAD LT) compared with that of the others (110 MPa for IPS e.max ZirPress and 90 MPa for IPS e.max CERAM). There were no failures at the sintered interface. This approach may be a route to speed fabrication times. Commercial software to accomplish this two-part fabrication has yet to be developed. Fatigue tests as predictors of clinical performance have not yet been performed. Based on the fatigue performance of lithium disilicates, however, this approach could be very promising. A similar approach has been introduced commercially fusing a veneer to an anatomic core, but few details of its success are yet available (LAVA digital veneering system [DVS, 3M/ESPE]) (Anonymous, 2009).

CAD/CAM systems remove material from a block to create a restoration. New technologies are emerging that selectively deposit materials to build up a restoration (additive approaches *vs*. CAD/CAM's subtractive approaches). Among these is a technique called "robocasting", wherein "inks" of selected materials are laid down in prescribed patterns to create a restoration (Silva *et al*., 2010b). With this system, complex geometries, including dental restorations, have been created, but interesting challenges remain, including development of suitable materials for the "printing" operation and the design of efficient printing patterns for assembly of complex dental restoration geometry.

PREDICTORS OF FUTURE PERFORMANCE

Great strides have been made in increasing fracture strengths of ceramics (320 MPa for lithium disilicate, 547 MPa for alumina, and 900 MPa for zirconia, depending on source of materials). Increasing only material strength, however, has not been sufficient to dramatically improve clinical survival, though it has influenced fracture modes. Accurate prediction of survival of restorations is multifaceted, and we are just beginning to fully realize the effects of test specimen geometry and loading conditions on outcomes.

Specimen Geometry

Geometric simplifications ease testing demands and costs, but may not be good predictors of future performance. For instance, studies with flat specimens indicated that both high-modulus cement (Lawn *et al*., 2004; Wang *et al*., 2007) and stronger and stiffer cores (Lawn *et al*., 2004) increase the load to fracture initiation. With complex geometry, neither of these significantly affects maximum principal stress in the veneer, a major area of failure (Rafferty *et al*., 2010b). In convex geometries, loads to initiate fractures were equivalent to those of flat specimens, but, once initiated, the fractures propagated much more quickly in the curved specimens (Rudas and Bush, 2007). Core design is emerging as an important factor in the survival and the extent of damage that develops in a restoration.

Loading Conditions

Single-cycle loading to failure is an excellent measure of fracture strength of a restoration-tooth system, but provides little insight into damage initiation and propagation in the oral environment. Anatomically correct veneered zirconia crowns supported by tooth replicas had a single-cycle load to failure of 1227 ± 221 N (Coelho *et al*., 2009a), certainly more than enough to survive maximum bite forces of 70-900 N, depending on tooth, sex, and measurement type (Waltimo and Könönen, 1995; Suputtamongkol *et al*., 2008). But identical specimens subjected to cyclic loading had only a 48% (23-68% at 90% confidence) probability of surviving a 200-N load for 50,000 cycles. Furthermore, rather than bulk fracture seen in the single-cycle loading, with fatigue the specimens failed cohesively within the veneer layer from cracks initiating directly below the sliding path of the indenter (Coelho *et al*., 2009a). This is analogous to chips seen clinically that originate from areas of occlusal adjustment and/or wear facets (Sailer *et al*., 2009b; Etman and Woolford, 2010). Understanding response to fatigue is critically important in predicting clinical behavior of a restoration.

Different loading conditions result in different failures (Aboushelib *et al*., 2009). Clinically failed veneered zirconia restorations (19 crowns and 17 bridges) were reconstructed and reproduced. The reproduced restorations, supported by composite resin dies, were fatigued in water (10,000 cycles to 200 N) then single-cycle-loaded to failure. Both clinical and reproduced bridges failed at the connector (7 of 17), but calculated stresses to failure were higher in the reproduced restorations. Cone cracks were found in all of the reproduced crowns and half of the bridges, but the calculated critical stress causing these cracks was extremely low (28 to 60 MPa), less than typical occlusal forces. It is unlikely that the cone cracks were the cause of the clinical failure. Fatigued specimens fail at subcritical loads (*vs*. fatigued 10,000 cycles and then loaded to single-cycle failure) (Bhowmick *et al*., 2007). Cone cracks appear almost immediately but do not propagate. Instead, other fracture modes ultimately cause the failure: inner and/or partial cone cracks in veneered zirconia. These differences highlight the need for testing configurations to emulate clinical functions as closely as possible.

One fatigue study has demonstrated remarkable degradation in strength (Rosentritt *et al*., 2009a). Fracture strength in veneered zirconia bridges dropped precipitously with increasing numbers of loading cycles (1058 N at 1.2 million cycles *vs*. 533 and 517 after 3.6 million cycles at 50-N and 100-N loads, respectively). Veneers chipped in all of the specimens, becoming more prevalent with higher loads (30% with 50-N load; 70% with 100-N load).

Sliding contact fatigue in water appears to be critical for simulation of occlusal conditions (Coelho *et al*., 2009a). The highly deleterious sliding creates partial cone cracks which developed in anatomically correct laboratory specimens (discussed above) (Kim *et al*., 2007) duplicated the fatigue wake hackle lines found in replicas of clinically failed restorations (Scherrer *et al*., 2008).

Recommendation

In the laboratory, single-cycle load to failure and mouth-motion fatigue of simplified geometries (flat layers or simplified crown-like structures) provide limited data to guide the development of all-ceramic systems (Coelho *et al*., 2009a). This, in turn, necessitates time-consuming clinical trials to develop basic understanding of new restorative materials and/or restoration design. Current evidence suggests that the best predictors of future clinical performance are tests done using: (1) restoration design that represents the anticipated clinical design as closely as possible (*e.g*., full anatomy, interproximal wall length variations, core shape and thickness, veneer thickness); (2) fabrication procedures that closely anticipate laboratory and clinical procedures (*e.g*., sandblasting before cementation with typically used protocols, pressed *vs*. layered veneers, etc.); (3) supporting structures that will be used clinically (*e.g*., implant- *vs*. dentin-supported); and (4) fatigue loading in water with sliding contacts.

CHALLENGES FOR FUTURE IMPROVEMENTS

Challenges remain in both understanding and improving the clinical performance of all-ceramic restorations. These include improved consistency and breadth of information about factors in clinical studies, definition of failures, and laboratory testing procedures. In parallel, developments in numerical analysis, physical properties of materials, and fabrication approaches all hold promise.

The ever-changing restorative materials make it difficult to tease out the major factors that could lead to improvements. The expectation that the remarkable strength of zirconia would eliminate problems with all-ceramic restorations is an excellent case in point. The zirconia itself has performed superbly, but failures shifted to the veneers, despite the use of veneering strategies long practiced with MCRs. Consistency in the definition of failure hampers the researcher's ability to establish the root cause of the problem. Is a small chip a failure? If not, how big and/or where must it be formed to be considered a failure? Agreement on reporting of this phenomenon would be extremely valuable.

The cause of veneer chipping, especially on zirconia cores, is complex. Material factors, including differences in coefficient of thermal expansion and thermal conductivity between veneer and core, likely create residual stresses that predispose a restoration to chipping. But non-material-related factors, such as thickness ratios and core or framework design, also play a significant role.

Few clinical studies have identical restorative materials fabricated by the same technique. Unique requirements of patients, of course, further complicate the challenge of understanding factors that contribute to long-term success of restorations. Few studies include patient control groups or report patient or provider factors. Even in the more recent literature, only two randomized clinical trials were reported (Sailer *et al*., 2009a; Etman and Woolford, 2010). This limited information complicates the making of evidence-based decisions.

Predictive laboratory tests could reduce the need for expensive and time-consuming clinical studies, which sometimes exceed the commercial lifetime of the materials being evaluated. Laboratory tests can now replicate clinical failure modes. Next we must determine whether the predicted failure lifetimes are equivalent to the clinical lifetimes. Laboratory tests likely overestimate clinical lifetimes, and so the question is, by how much? Beyond this, it will be important to determine how and where informed simplifications in testing conditions can be made. For instance, does fatigue loading with sliding on flat specimens create the same failure modes as with complicated geometry? In parallel, numerical analysis approaches will advance, more accurately modeling clinical performance. The value of laboratory tests or numerical analyses that fail to replicate clinical performance (like single-cycle loading to failure) should be seriously questioned.

Advances in materials will continue. Areas of greatest promise are improvements in structural reliability through flaw control and damage tolerance (Denry and Kelly, 2008; Kelly and Denry, 2008).

Fabrication processes and CAD/CAM capabilities creating veneers and cores separately (Beuer *et al*., 2009b) will continue to evolve. Subtractive CAD/CAM approaches, removing material from a block to create a shape, will be complemented by additive approaches, laying down materials only in places where it is needed to create a restoration (Silva *et al*., 2010b). While substantial hurdles remain with each of these approaches, both show promise for the more comprehensive use of technology.

Our understanding of how all-ceramic restorations perform—and fail—has improved dramatically in recent years. The remaining challenges for future advances are exciting and present abundant arenas for future investigations and innovations.

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