Dental Ceramics for Restoration and Metal Veneering

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KEYWORDS

- Dental ceramics • All-ceramic restorations • Metal-ceramic restorations • Porcelain
- Glass-ceramics • Zirconia • Ceramic-polymer interpenetrating network

KEY POINTS

- A facile understanding of the development, composition, microstructure, properties, and indications of various classes of ceramic dental materials.
- Knowledge of the rationale behind the choice and usage of dental ceramics to maximize esthetics and durability.
- Successful ceramic restorations depend on the balancing of multiple factors.

INTRODUCTION

According to the American College of Prosthodontists, 178 million people in the United States, which represents 55% of the US population, are missing at least 1 tooth and this number is expected to grow over the next 2 decades because of an aging population. Teeth play a critically important role in human life because loss of function reduces people's ability to eat a balanced diet, with negative consequences for systemic health. Loss of esthetics can also negatively affect social function. Both function and esthetics can be restored with dental crowns and fixed dental prostheses (FDPs). Ceramics have become increasingly popular as restorative materials because of their esthetics, inertness, and biocompatibility. Of the crowns and fixed prostheses currently produced in the United States, 80.2% are all-ceramic restorations, 16.9% are porcelain fused to metal (PFM), 2.2% are full-cast, and 0.7% are resin-based composite (RBC).[1] Demands for more esthetic and metal-free restorations, as well as soaring metal prices, are likely to increase further the number of all-ceramic prostheses.[2]

However, a major clinical concern is that ceramics are brittle and subject to fracture.[3,4] The financial drivers for developing fracture-resistant and esthetic ceramics are high: the European crown and FDP market approached $2 billion in 2007[5]; the global crown and FDP market was estimated to be $25 billion in 2010 and more than $30 billion in 2015.[6] This article provides an overview of the background and the current knowledge base associated with dental ceramics for restoration and metal veneering, including a historical review of the development of ceramic restorations and their limitations. It also includes a summary of the current state of the art of porcelain, glass-ceramics, and polycrystalline ceramics. In addition, materials design considerations for dental prostheses are discussed.

THE HISTORY OF DENTAL CERAMICS

Shortly after the introduction of porcelain into Europe in the early eighteenth century, Alexis Duchateau, a Parisian apothecary, introduced ceramics to dentistry when he successfully replaced his ivory dentures with porcelain. With the help of a Parisian dentist, Nicholas Dubois de Chemant, Duchateau, working in concert with a new, high-technology porcelain manufacturer in 1774, created a complete set of porcelain dentures. They must have been very well made because they lasted Duchateau the rest of his life. The development of porcelain dentures was revolutionary in terms of esthetics and oral hygiene, and was recognized as such by Edward Jenner (developer of the smallpox vaccine) and the Faculty of Medicine Paris: they "…united the qualities of beauty, solidity and comfort with the exigencies of hygiene.[7]" Because the then-popular ivory-based or wood-based dentures, often using cadaver teeth, were all porous, they absorbed oral fluids and eventually became badly stained and highly un-hygienic. Also, these early porcelain dentures were dysfunctional because patients had to remove them in order to eat. In addition, those complete porcelain dentures were only intended for edentulous patients, requiring the removal of the remaining teeth from patients' mouths, which was a painful procedure before the discovery of anesthesia by Horace Wells in the middle of the nineteenth century.

Porcelain inlays, onlays, and crowns were introduced by Charles Land[8] in 1886, which ultimately led to the creation of esthetic and functional ceramic restorations. However, the original dental porcelain contained a high feldspathic glass content and was extremely brittle and weak ($\sigma \sim 60$ MPa; σ stands for strength).[9,10] Therefore, despite the esthetic advantage, the early porcelain restorations were not widely applied in dentistry.[11] Dental ceramics have become increasingly popular as restorative materials because of improvements in strength and the increased goodness of fit with development of pressing and computer-assisted design (CAD)/computer-assisted manufacturing (CAM) processes. The timeline of the development of dental ceramics from the inception of initial porcelain materials to modern ceramic compositions, along with processing technologies, is shown in **Fig. 1**. The main compositions and pertinent mechanical properties of various dental ceramic materials, representative of major material classes and developments, are shown in **Table 1**.

Since Weinstein and colleagues[12,13] solved the problem of the coefficient of thermal expansion (CTE) mismatch between the porcelain veneer and metal framework in 1962, great improvements have been made in PFM systems. Until very recently, it was estimated that 70% to 80% of fixed prostheses produced in the United States were PFM (Ivoclar Vivadent, 3M ESPE, Jensen Dental, Marotta Dental Studio, and Glidewell Laboratories, personal communication, 2011). In contrast, the dental community has long recognized that to realize the full potential of dental prostheses, all-ceramic restorations are necessary. Several strategies have been developed to
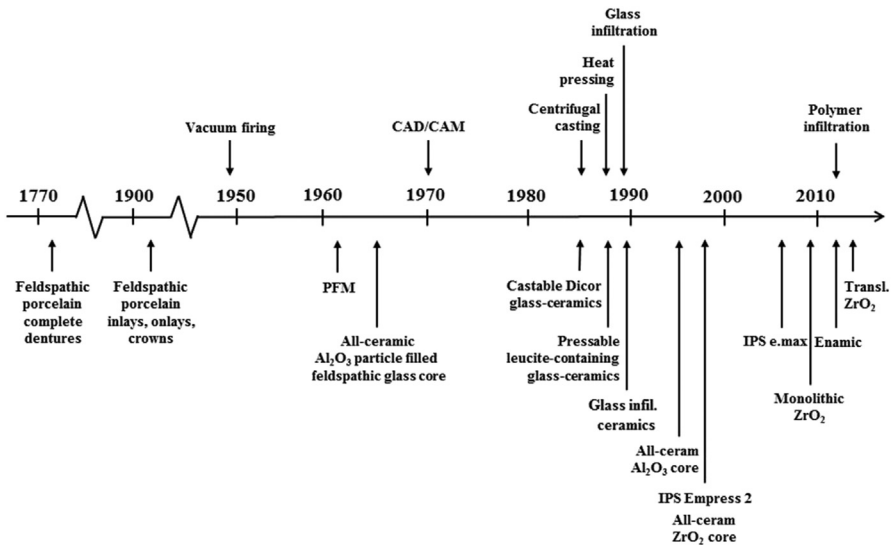
Fig. 1. The timeline of the development of dental ceramics and their processing technologies.

improve the strength and fit of dental ceramics over the past 50 years. Other improvements in longevity have involved the use of high elastic moduli cores and buildup materials and cements to protect single crowns against bulk fracture.

One well-grounded approach to strengthening porcelain is to add uniformly dispersed filler particles to the glass matrix, a technique referred to as dispersion strengthening. One of the most successful particle fillers used in dental ceramics is leucite, a crystalline mineral possessing an index of refraction similar to that of feldspathic glasses.[14] Commercial dental ceramics containing leucite as a strengthener include IPS Empress ($\sigma \sim$ 138 MPa) (Ivoclar Vivadent, Schaan, Liechtenstein) and Finesse All-Ceramic ($\sigma \sim$ 125 MPa) (Dentsply International). Particle strengthening can also be achieved by heat-treating the glass to facilitate the precipitation and subsequent growth of crystallites within the glass, a process termed ceraming. Dental ceramics produced using the ceraming process are called glass-ceramics. Several commercial products, such as Dicor ($\sigma \sim$ 229 MPa) (Dentsply International), IPS Empress II ($\sigma \sim$ 350 MPa) (Ivoclar Vivadent), and more recently IPS e.max Press ($\sigma \sim$ 400 MPa) and IPS e.max CAD ($\sigma \sim$ 480 MPa) (Ivoclar Vivadent) are in this category. The leucite-strengthened porcelains and the glass-ceramics are translucent, so single-layer (monolithic) restorations can be made from these materials. The drawback is that only moderate strength increases can be achieved via the particle strengthening techniques. Therefore, monolithic ceramic restorations experience high failure rates, ranging from 4% to 6% for Dicor molar crowns[15,16] and 3% to 4% per year for IPS Empress crowns.[17,18]

The traditional approach to the fracture problem of monolithic glass-ceramic restorations is to use a layer structure with esthetic but weak porcelain veneers fused onto strong but opaque ceramic cores. The history of the development of higher-strength ceramic cores involves an increase in crystalline content (from $\sim$40 vol % to 99.9 vol%) accompanied by a reduction in glass content. The first successful strengthened core ceramic was made of feldspathic glass filled with $\sim$40 vol% (vol

Table 1
Properties of various dental ceramic materials

Material	Crystalline Phase (vol%)	Modulus E (GPa)	Hardness H (GPa)	Toughness T (MPa·m$^{1/2}$)	Strength σ (MPa)
Porcelain					
Feldspathic ceramic (Vita Mark II)	Albite (<20)	72	6.2	1.2	122
Veneer for ceramic (Lava Ceram)	Leucite (6)	80	5.2	1.1	85
Veneer for metal (d.SIGN)	Leucite/apatite (25)	68	5.9	1.1	104
Glass-Ceramic					
Mica glass-ceramic (Dicor MGC)	Fluormica (70)	69	6.0	1.2	229
Leucite glass-ceramic (IPS Empress CAD)	Leucite (35–45)	65	6.2	1.3	140
Lithium Disilicate–Ceramic					
(IPS Empress 2)	Lithium disilicate (65)	96	5.5–6.3	2.9–3.2	306–420
(IPS e.max CAD)	Lithium disilicate (70)	95	5.8	2.3	480
(IPS e.max Press)	Lithium disilicate (70)	95	5.8	2.8	400
Ceramic-Glass Interpenetrating Network					
Glass-infiltrated spinel	Spinel (68)	185	—	2.5	350
Glass-infiltrated alumina	Alumina (68)	274	11.8	3.6	548
Glass-infiltrated zirconia	Zirconia-toughened alumina (67)	245	13.1	3.5	700
Polycrystalline Ceramic					
Alumina (dense, fine grain)	Alumina (>99)	372	19.6	3.1	572
Zirconia (Lava Plus)	3 mol% Y-TZP (>99)	210	14.0	4.0	1200
Zirconia (Zpex smile)	Cubic/tetragonal zirconia (>99)	210	13.4	2.4	485
Ceramic-Resin Interpenetrating Network					
Resin-infiltrated porcelain (Enamic)	Feldspathic ceramic (75)	30	1.7	1.3	159
Tooth					
Dentin	Hydroxyapatite (50)	18	0.6	3.1	34–98
Enamel	Hydroxyapatite (95)	94	3.2	0.8	12–42

%, percentage by volume) of alumina particles.[19] The alumina fillers increased the flexural strength of the ceramic to ~120 MPa with a trade-off in translucency; hence veneering was required. In 1983, Coors Biomedical (Golden, CO) developed Cerestore all-ceramic restorations with a ceramic core containing ~60 wt% of Al_2O_3, 9 wt% of MgO, a barium aluminosilicate glass at 13 wt% (wt%, percentage by weight), and enough silicone (12 wt%) and kaolin clay (4 wt%) to impart sufficient plasticity for transfer molding at 160°C.[20] It was reported that the alumina reacted with magnesia to form magnesium aluminate spinel, expanding to become net shaped. It is highly unlikely that this reaction occurred given the low firing temperature of 1300°C and short firing time. Subsequent analysis showed that the net-shape ability occurred because of oxidation of the silicone base releasing gaseous products and leading to the crown blowing up like a loaf of bread contained within its mold.[21] However, following universal problems with fractured restorations, the manufacturer withdrew the system. A similar product from the same era, the Hi-Ceram restorative system (Vita Zahnfabrik, Bad Säckingen, Germany) with its core material containing around the same amount of alumina as the Cerestore core, also failed to meet the requirements for posterior restorations.[22] The Hi-Ceram system was replaced by In-Ceram (Vita Zahnfabrik) in 1990. The In-Ceram restoration had a core that was fabricated by lightly sintering an alumina powder compact and then infiltrating the still-porous alumina matrix with a low-viscosity glass containing lanthanum, which reduced viscosity and increased the index of refraction of the infiltration glass. In contrast with Hi-Ceram, in which ~60 vol% alumina particles were added to a glass matrix, In-Ceram alumina was derived from high-temperature glass infiltration of an alumina scaffold, resulting in an alumina-glass interpenetrating network structure. The final product contained ~70 vol% of alumina and had a flexural strength of ~550 MPa.[23] Products along the same line are In-Ceram spinel and In-Ceram zirconia (toughened alumina). The former has a higher translucency but lower strength, whereas the latter has a higher strength but lower translucency, relative to In-Ceram alumina. In 1993, Procera (Nobel Biocare, Göteborg, Sweden) presented a new all-ceramic restoration concept[24] in which the fully dense core material contained 99.9 vol% alumina and displayed a flexural strength of 572 MPa. Several years later, even stronger yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) ceramic was introduced to dentistry as a core material with a flexural strength more than 1200 MPa.

Despite significant improvements in the performance of dental ceramics, the structural stability of all-ceramic systems remains less reliable than that of PFM systems, in which only nonbiological complications are considered.[25] Clinical studies have revealed that the primary cause of failure for lithium disilicate and alumina restorations are fracture in both veneer and framework, whereas that for zirconia-based restorations is cohesive fracture of the veneering porcelain.[26] In an effort to circumvent the problem of veneer chipping and fracture, translucent glass-ceramic materials and, more recently, so-called cubic zirconias have been developed for monolithic restoration applications. However, these translucent ceramic materials are considerably weaker than the traditional dental tetragonal zirconia (Y-TZP), and thus cannot be used to replace the strong but more opaque Y-TZP.

THE STATE-OF-THE-ART DENTAL CERAMICS
Porcelain

Dental ceramics that best mimic the optical properties of natural teeth are predominantly glassy materials, which derive principally from feldspar-quartz-kaolin triaxial

porcelain compositions.[21,27] Many technological advances have contributed to the use of porcelain in fixed prosthodontics, such as the development of the vacuum firing technology in 1949, the invention of the high-speed handpiece, the discovery of elastomeric impression materials, and the advent of pressing and CAD/CAM technologies in the 1980s.[28] From a materialistic viewpoint, porcelain compositions have evolved from the original hard-paste Meissen porcelain, which contained a high clay content and thus required a high firing temperature, to the modern soft-paste porcelains that are composed of mostly feldspar with no kaolin or quartz and possess excellent translucency. However, dental porcelains with the most desirable esthetics also tend to have the lowest strength and resistance to crack propagation, which severely limits their clinical indications.[29–32] One major breakthrough came in 1962, when the Weinsteins, with the help of Koenig[27], developed a leucite-containing porcelain composition that could be fired directly onto common dental alloys.[21] Leucite is a rock-forming mineral that is composed of potassium aluminosilicate. At room temperature, leucite possesses a tetragonal structure. However, the crystal structure undergoes a tetragonal to cubic phase transformation at 625°C. This phase transformation is accompanied by a volume expansion of 1.2%, resulting in a high CTE ($20–25 \times 10^{-6}/°C$).[33] In contrast, feldspar glass has a low CTE ($\sim 8 \times 10^{-6}/°C$). Therefore, by varying the proportions of leucite and feldspar glass, porcelain frits with average CTEs matching those ($12–14 \times 10^{-6}/°C$) of dental alloys can be produced. A matching CTE between porcelain veneer and metal alloy coping prevents the development of deleterious thermal stresses on cooling from firing temperatures. Dental manufactures have also discovered that having the porcelain with a slightly lower CTE than the metal (typically differing from $<1 \times 10^{-6}/°C$) can place the porcelain in slight compression, thus increasing the fracture resistance of the restoration. The leucite content for tailoring the CTE of porcelain can vary from several weight percent when coupled with ceramic frameworks to 17 to 25 wt% when matched with common metal alloys. Leucite is also an effective material for the dispersion strengthening of feldspar glass, because a large amount of leucite (up to 35–50 wt%) can be incorporated without significantly compromising its translucency because the refractive index of leucite (n = 1.51) is very close to that of the feldspar glass (n = 1.52–1.53). In addition, owing to preferential etching of leucite crystals relative to the glass matrix, the leucite-containing feldspar glasses can be acid etched to create micromechanical features for resin bonding, thus making the restorations more fracture resistant. The microstructures of several commercial leucite-containing feldspathic ceramics used as veneers for ceramics and metals, as well as dispersion-strengthened monolithic glass-ceramics, are shown in **Fig. 2**.

Leucite feldspathic porcelain materials remain as some of the most esthetic and widely used dental ceramics. Their clinical indications include inlays, onlays, partial crowns, and crowns, as well as veneers for ceramics and metals. Clinical studies have shown that feldspathic porcelain restorations have excellent long-term success rates when bonded to and supported by primarily enamel structures. For example, the survival rate of inlays and onlays is 92% at 8 years,[34] veneers 94% at 12 years,[35] and crowns 95% at 11 years.[36] These findings suggest that this class of materials is ideal for cases in which a significant amount of healthy tooth structure and enamel remain.[29]

The PFM technology has made it possible to fabricate more structurally demanding dental restorations, such as crowns and FDPs. PFM restorations are ideal for cases in which minimal to no tooth structures remain[29] and splinted restorations are required.[37] The esthetic qualities of PFM are at their best when a high-gold-content framework material (eg, Captek) is used.[29] However, the trade-off is that the low modulus of the high-gold framework provides little support to the porcelain veneer, resulting in a greater tendency for veneer fracture and chipping.[38]
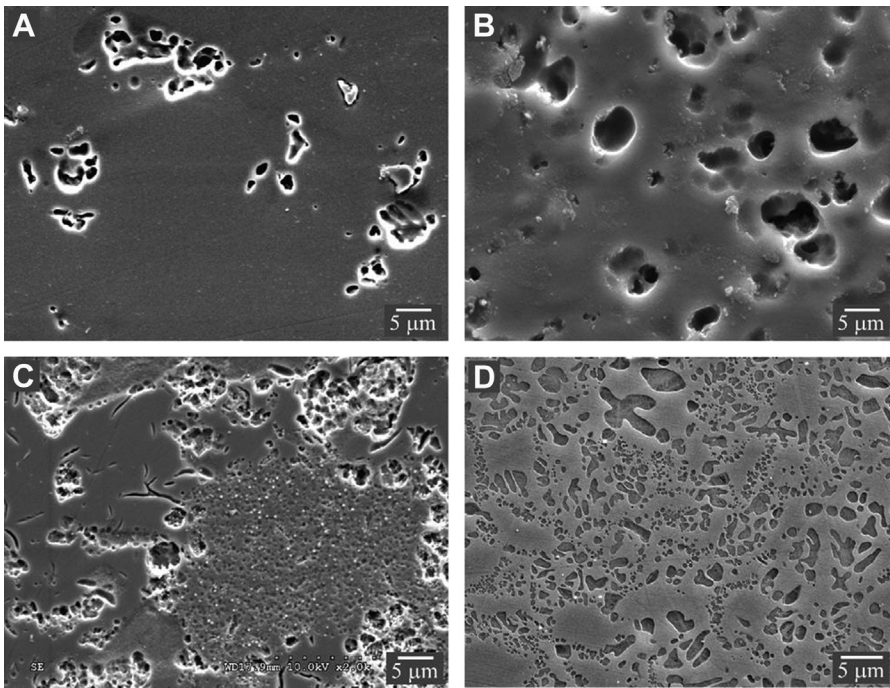
Fig. 2. Microstructures of leucite-containing feldspathic ceramics. Images were taken using secondary electrons in a scanning electron microscope (SEM). Feldspathic overlay porcelains for zirconia: (*A*) Lava Ceram and (*B*) Vita VM9. Porcelain overlay for metal: (*C*) d.SIGN. A dispersion strengthened glass-ceramic: (*D*) Empress CAD. Acid-etched surface revealing craters once occupied by leucite crystals and microcracks in the glassy matrix. Note: the leucite content increases from porcelain veneers for ceramic to metal to dispersion-strengthened glass-ceramic.

Glass-Ceramics

Glass-ceramics are much stronger and tougher but also have lower translucency relative to porcelain. The strengthening and toughening of glass-ceramics are achieved by a ceraming process, in which crystals are precipitated under controlled heat treatments from homogeneous glass through the nucleation and growth processes. The material Dicor was the first glass-ceramic material used for the fabrication of dental restorations. It consisted of fluormica crystals in the form of individual sheets or plates embedded in a glass matrix. Its microstructure, analogous to a house of cards, provides an interlocking mechanism for strengthening. However, because of its poor mechanical performance in clinical applications, Dicor was withdrawn from the market. Some current leucite-reinforced glasses are also produced via the ceraming process. However, currently the most widely used and, arguably, the strongest and toughest dental glass-ceramics are made with lithium disilicate reinforcement.

The first dental lithium disilicate ceramic was fabricated from a base glass composition (SiO_2-Li_2O-Al_2O_3-K_2O-P_2O_5-ZnO-La_2O_3) plus some additives for color and fluorescence. A homogeneous base glass ingot, containing a limited amount of lithium metasilicate, was heated until it reached a viscous state, and then pressed into a mold. Through a judiciously controlled heat treatment, a glass-ceramic containing ~70 vol% of elongated lithium disilicate crystals could be precipitated from the base

glass to produce an interlocked microstructure. The resulting material possessed a flexural strength of 350 MPa and fracture toughness 2.9 MPa m$^{1/2}$, which were more than twice those of leucite-based glass-ceramics. The material was commercialized for dental framework use and marketed under the trade name IPS Empress 2. However, this material had high clinical failure rates at 9% to 50% after 24 to 60 months, with a higher tendency of framework fracture in the connector area of short-span posterior FDPs.[39–41] These findings indicate insufficient flexural strength of the IPS Empress 2 framework for multiunit prostheses. Subsequently, a new and improved lithium disilicate glass-ceramic (IPS e.max) with a much higher flexural strength (400–480 MPa) was developed. The improvements were made through the refinement of the base glass composition as well as by improving the quality of the initial glass ingot (with fewer defects and pores). Compared with the base glass for IPS Empress 2, the new glass composition (SiO_2-Li_2O-Al_2O_3-K_2O-P_2O_5-ZrO_2) contained up to 4 wt% ZrO_2 additives, whereas it had diminished ZnO and La_2O_3 contents (<0.1 wt%).

The IPS e.max glass-ceramics come in 2 forms, Press and CAD (**Fig. 3**), reflecting differences in processing conditions.[42,43] The IPS e.max Press ingots are heat-pressed at 920°C for 20 minutes. The IPS e.max CAD ingots are first heat treated to form the intermediate lithium metasilicate glass-ceramics, which are easier to machine to shape. These glass-ceramics are then heated to 840°C for 7 minutes, during which the lithium metasilicate glass-ceramic is transformed to a chemically more stable and esthetically pleasing lithium disilicate glass-ceramic. Lithium disilicate Press and CAD have a glass matrix containing ~70% elongated, needlelike crystals. In the Press grade the crystallites are ~4 μm long and ~0.6 μm wide and aligned perpendicular to the external surfaces, whereas in the CAD grade the crystallites are ~1 μm long and ~0.4 μm wide and more randomly oriented. The Press grade has slightly higher toughness because of the greater impedance to crack propagation by the larger grains (ie, crystals). However, it also has slightly lower strength because these same grains introduce larger starting flaws into the structure (see **Table 1**). Lithium disilicate glass-ceramics are indicated for veneers, anterior crowns, and posterior inlays and onlays. However, when fabricated to monolithic restorations and luted with resin cements, they are also suitable for single-unit, full-coverage crowns for molar teeth. In addition, the large elongated grains in lithium disilicate Press are thought to improve the fracture toughness by crack bridging and deflection, especially in the connector areas of an FDP, in which elongated crystals are preferentially oriented parallel to
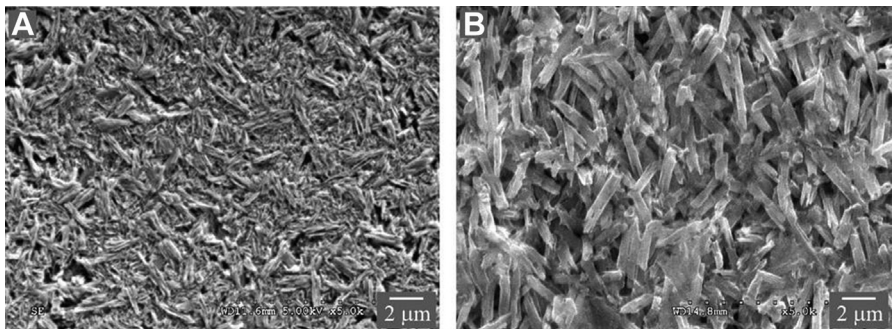

Fig. 3. Microstructures of lithium disilicate glass-ceramics. (A) CAD and (B) Press. Images were taken on an acid-etched surface using secondary electrons in an SEM, revealing elongated lithium disilicate crystallites. Note in the Press material (B), the preferential orientation of the coarse elongated lithium disilicate crystallites.

the tensile surface. Such a logs-on-the-river structure can effectively improve the fracture resistance of the restoration. Long-term clinical data support the use of lithium disilicates as single restorations anywhere in the mouth[44] and as short-span FDPs in the anterior region.[45]

Polycrystalline Ceramics

Recent advances have created stronger and tougher ceramics, predominantly Y-TZP (Fig. 4). However, Y-TZP has severe clinical deficiencies owing to its low translucency. The opacity of zirconia becomes a problem, especially when placing an anterior crown or short-span FDPs in the presence of natural teeth. In that case, the reflectance and light scattering do not appear natural. In order to create space for a porcelain veneer thick enough to cover an opaque zirconia core and to match the optical properties of the adjacent natural dentition, a substantial reduction of existing tooth structure is required. In addition, clinical research and practice have revealed that although zirconia frameworks are very fracture resistant, chipping[46–53] and delamination[54,55] of the porcelain veneer are frequent problems. In 25 clinical trials on a variety of brands and makes of zirconia-based crowns and FDPs, chips and delaminations were consistently reported at 6% to 10% in 3 to 5 years in single crowns and 20% to 32% in 5 to 10 years in FDPs.[51–53,56–75] In contrast, crowns and FDPs with metal frameworks revealed substantially lower fracture rates, ranging from 2.7% to 6% up to 15 years.[76–79] One of the primary reasons for the poor clinical performance of porcelain-veneered zirconia bilayer prostheses is the low thermal conductivity of the zirconia core relative to the metal coping, which could result in a large temperature gradient in the porcelain veneer on cooling, and thus residual thermal stresses become locked into the material system.[80] Although it is evident that the high chipping/fracture rate is caused predominantly by these residual stresses, a comprehensive knowledge of the governing material (elastic modulus and CTE), design (veneer/core thickness ratio), and processing (cooling rate) parameters remains largely absent.[80–84] Thus, this continues to be an active research area.

In an effort to avoid veneer chipping and delamination, monolithic zirconia is often used in full-arch restorations, posterior crowns, and FDPs.[85–88] In all these cases, the opacity of Y-TZP zirconia remains a serious issue, although the white, opaque, monolithic Y-TZP restorations may be suitable for bleached teeth.
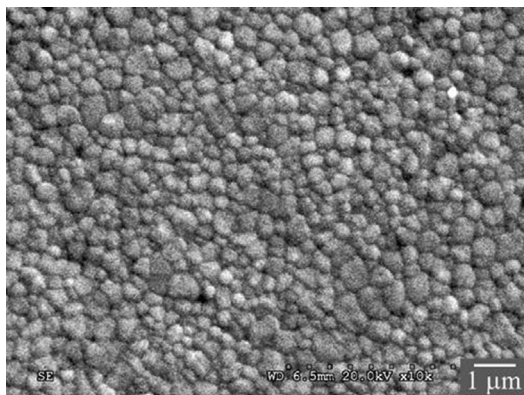

Fig. 4. Scanning electron micrograph, showing a typical fine-grained microstructure of high-strength dental zirconias (Y-TZP). Specimen surface was polished and thermally etched.

After a decade of research and development, progress has been made in improving the translucency of Y-TZP by reducing porosity, decreasing grain size, and eliminating any alumina added as a sintering aid.[89] However, close examinations have revealed that, unless they are thin (ie, <0.5 mm), so-called commercial translucent Y-TZP restorative materials remain largely opaque.[90] Eliminating porosity and impurities alone is not sufficient to significantly improve the translucency of Y-TZP. Tetragonal zirconia is birefringent, meaning that the index of refraction is anisotropic in different crystallographic directions.[89,91] This property causes reflection and refraction at grain boundaries, thus reducing light transmittance. Theory predicts that to make a Y-TZP ceramic sufficiently translucent while preserving strength, a sub–100-nm grain size is necessary, so that light may penetrate without substantial scattering.[89,91–93] However, it is technologically challenging to achieve densification without substantial grain growth beyond the critical 100-nm size.

The current approach to this problem is to introduce an optically isotropic cubic zirconia phase into an ordinarily tetragonal material (eg, DDcubeX^2 by Dental Direkt Materials and Zpex Smile by Tosoh Corporation). However, biphasic tetragonal/cubic zirconia is weaker and more brittle compared with its tetragonal counterpart. For instance, the flexural strength and fracture toughness of Zpex Smile (609 MPa and 2.4 MPa m$^{1/2}$) are only slightly more than one-half of that of Y-TZP. They are more like a dental alumina material (Procera alumina, Nobel Biocare),[33,94] and are also subject to low-temperature degradation. In general, increasing yttria content leads to a larger amount of cubic phase and thus greater translucency. The trade-off is that strength and toughness diminish as the cubic content increases, which has led to the development of several translucent dental zirconia materials containing various amounts of cubic phase. For example, the Katana ultratranslucent zirconia material has a flexural strength of 557 MPa, whereas their supertranslucent and high-translucent zirconias have flexural strengths of 748 and 1125 MPa, respectively. These translucent zirconia pucks also feature multilayered color with a lighter shade in the occlusal one-third thickness and a darker shade at the gingival one-third, sandwiching 2 thinner transition layers. However, the mechanical integrity of these multilayered structures has yet to be evaluated.

New Classes of Materials

The current esthetic and highly fracture-resistant restorative materials are either high-crystalline ceramics or heavily particle-filled resin composites. The elastic properties of these materials are not compatible with enamel or dentin substrates. Therefore, there is a greater tendency for restoration fracture to occur when a much stiffer ceramic material is used, and for underlying tooth fracture to occur when a low-modulus resin composite material is used.[95] In addition, the current advent of great interest in minimally invasive dentistry and chairside 1-visit restorations has resulted in the widespread usage of CAD/CAM technology. Ceramic restorative materials are susceptible to machining damage, especially when the restoration or part of the restoration is thin (eg, marginal chipping).[96,97]

Recently, a new class of material, ceramic-polymer interpenetrating network (CPIN) material (Vita Enamic), has been developed. The impetus for developing the CPIN material is to tailor the material properties, such as elastic modulus, strength, toughness, and hardness, through judicious control of its composition and microstructure. The Enamic material consists of 86 wt% (75 vol%) of a feldspathic ceramic matrix into which an organic phase of dimethacrylate resin containing urethane dimethacrylate and triethylene glycol dimethacrylate is infiltrated.[98] The fabrication process of this material involves 2 steps: first, a porous presintered ceramic network is produced

and conditioned by a coupling agent; then, the network structure is infiltrated with the monomers by capillary action.[99,100] The resulting microstructure has a hybrid structure with interpenetrating networks of ceramic and polymer (Fig. 5), mimicking the interlocking of prism bands in natural teeth. The flexural strength, elastic modulus, hardness, and fracture toughness of the Enamic material have been evaluated by several investigators,[99,100] and has similar properties to natural tooth structures (see Table 1). Compared with ceramic restorative materials, Enamic has reliable millability and edge stability in terms of its ability to be fast milled into thin (<0.5 mm) restorations with excellent precision.[101] A full-contour posterior crown takes a little more than 5 minutes to mill, and eliminates the need for postmilling firing. The material is also easy to adjust and polish. Thus, it is an ideal material for chairside 1-visit restorations.

The three-dimensional (3D) interconnected dual-network structure of CPIN differs from the RBC materials, in which only the resin matrix is continuous. The most recent generation of laboratory-fabricated millable RBC blocks (eg, Lava Ultimate from 3M and Cerasmart from GC) are heavily particle-filled resins cured at a high temperature and pressure. The filler particles in Lava Ultimate are composed of dispersed silica (∼20 nm) and zirconia (4–11 nm) nanoparticles, as well as silica/zirconia nanoparticle clusters (0.6–10 μm) (Fig. 6). The rationale behind the usage of nanoclusters is that, compared with the traditional hard micrometer-sized filler particles, the nanoparticle clusters (analogous to a bunch of grapes) may not be as effective in terms of crack deflection and strengthening, but they are very effective for polish retention. The large nanoclusters break down to nanoparticles on mastication, leading to a smooth wear surface. However, the nanoclusters inevitably consist of defects and voids, which can soak up oral fluids, resulting in the discoloration and degradation of the RBC. Although the filler loading (80–90 wt% or 65–77 vol%) in the millable RBCs is similar to that of CPIN, their elastic properties and fracture behavior are different. In the case of CPIN materials, the interconnectivity of the ceramic phase provides stiffness and hardness that are necessary for the resistance to plastic deformation and wear. In contrast, the ductile polymer network is able to effectively distribute stresses in all directions.[102] As a result, the 3D interpenetrating dual-network materials possess enhanced resistances to a variety of breakdown phenomena, including contact and flexural damage as well as fatigue crack growth and wear.[98,101–104]
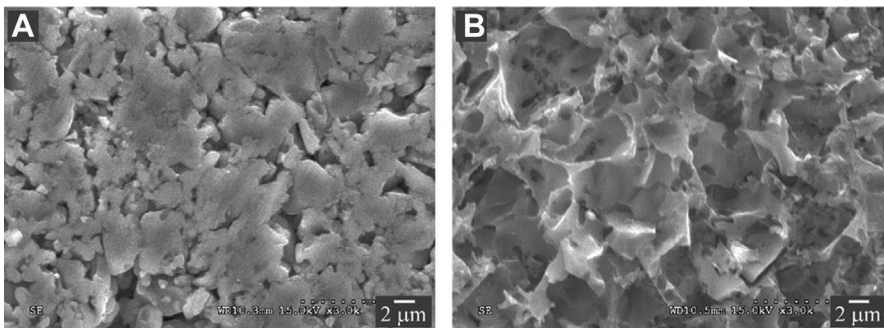

Fig. 5. Microstructure of Vita Enamic observed using secondary electrons in an SEM. (*A*) A polished and then thermally etched surface, revealing a ceramic network structure consisting of ∼25 vol% porosity following selective removal of the polymer phase. (*B*) A polished and then acid-etched surface, showing the polymer network after selective removal of the surface ceramic material.
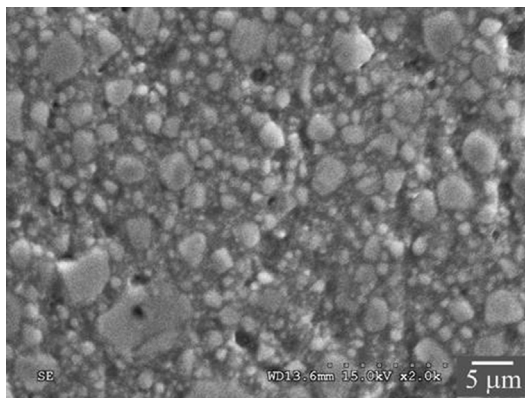
Fig. 6. Scanning electron micrograph of an RBC, Lava Ultimate. The material surface was polished down to 1 μm before imaging.

The CPIN material also differs from another interpenetrating network material (ie, In-Ceram alumina) in which alumina powders consisting of both coarse and fine particles are slip cast to ~70% density. The cast objects are sintered at 1000°C to 1200°C to facilitate the formation of necks between the individual particles, and to prevent significant shrinkage of the components. This effect is achieved by the presence of the coarse grains, which prevent contraction and result in an interconnected porous structure throughout the object. The porous structure is then infiltrated with a low-viscosity lanthanum-containing glass at 950°C to 1000°C, during which infiltrating glass completely wets the alumina scaffold under the influence of capillary forces. The resultant material consisted of a 3D alumina (~70 vol%) and glass interpenetrating network structure. However, because both alumina and glass are brittle materials, only limited toughening mechanics (ie, crack deflection) may be achieved and no significant stress distribution can occur.

It seems desirable to develop a new restorative material that combines the elastic modulus of RBC, which is much lower than that of dentin and even more so than enamel, with the long-lasting esthetics of ceramics. This new CPIN material may offer a unique biomimetic alternative to traditional composites and ceramics. Clinically, Vita Enamic is suitable for single-tooth restorations such as inlays, onlays, veneers, and crowns, including implant-supported crowns and posterior restorations. There are no credible clinical data available concerning the longevity of Enamic restorations at this time. However, laboratory studies have shown that Enamic has excellent resistance to wear and fatigue damage relative to traditional ceramic restorative materials.[98,101]

MATERIALS DESIGN CONSIDERATIONS

Because the clinical performance, in particular the fracture resistance, of dental restorations is influenced by a host of variables, the restoration design and materials selection involve balancing several factors that are considered later. In addition, for reader convenience, some of the commonly observed clinical fracture modes are sketched in Fig. 7.

Material Properties

Fracture in ceramics is governed by toughness and strength, and to a lesser extent by elastic modulus and hardness.[105] For crownlike structures, increasing strength simply
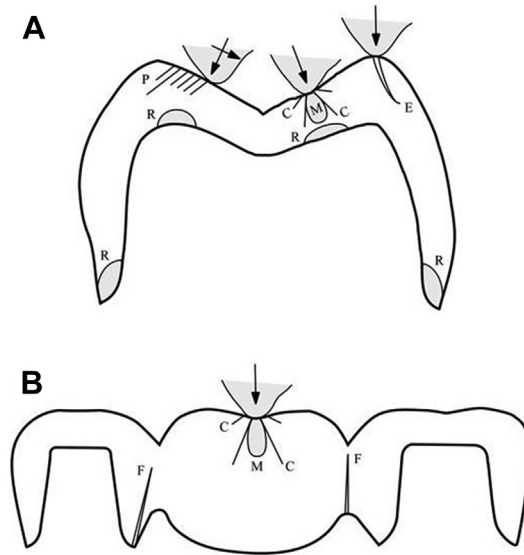
Fig. 7. Various fracture modes in all-ceramic (*A*) crown and (*B*) FDP structures: axisymmetric cone (C) and median (M) cracks; partial cone (P) cracks; edge chipping (E) cracks; radial (R) cracks at cementation surfaces; flexure (F) cracks at connectors. Linear-trace cracks (C, P, E, F) extend out of the plane of diagram, shaded (R, M) cracks extend within the plane of diagram. *The arrows* indicates the directions of load. (*Modified from* Zhang Y, Sailer I, Lawn BR. Fatigue of dental ceramics. J Dent 2013;41(12):1136; with permission.)

increases the resistance to crack initiation in these structures, whereas increasing toughness increases the resistance to crack propagation.[106–109] In many clinical trials covering numerous ceramic systems, fracture toughness of the core ceramic tracks well with clinical success. This fact was taken into consideration when designing a new ceramic classification system based on known clinical indications now in the international standard ISO (International Organization for Standardization) 6872. In addition, strength may be more relevant to FDP structures, in which failure can occur by slow crack growth from a surface flaw, usually on the gingival side of connectors (see **Fig. 7**B). A higher modulus reduces layer flexure on a dentin base, decreasing the failure trends for flexural radial fracture (see **Fig. 7**A).[3,110] Increased hardness diminishes the susceptibility to quasiplastic deformation (contact-induced plastic deformation in brittle materials, which is a precursor of median cracks) and wear at the top surface, and therefore suppresses contact damage (see **Fig. 7**A). Note that zirconia has higher toughness and strength than alumina but lower modulus and hardness. Zirconia is also subject to other forms of long-term degradation; for example, "aging" from hydrothermal degradation associated with phase transformations.[111–114] Porcelains are most vulnerable to damage, whereas glass-ceramics such as lithium disilicate occupy a middle ground. Accordingly, choice of material is a compromise, and requires a fundamental materials science understanding.

Microstructure

Ultimately, material properties are determined by the underlying microstructure.[115] Current dental ceramic technology borrows heavily from the science of materials fabrication, involving a complexity of starting powder preparation, processing

additives, and sintering treatments. Veneering ceramics are generally leucite-containing feldspathic porcelains, with the leucite in the form of crystallites to toughen the structure as well as to create a material thermally compatible with the ceramic framework.[28,116,117] Glass-ceramics are likewise formed by heat-treatment crystallization of glass compositions. The key to superior properties is the choice of constituent starting powders and heat treatments. Lithium disilicates comprise the most recent and most durable of the glass-ceramics.[42,118] Up to 70 vol% needle-like crystallites result in moderately high strengths and toughness by virtue of their crack-containment properties.[119] Alumina ceramics have been prepared in a variety of microstructures, but are now supplanted by zirconias. Zirconia properties are governed by many factors, including transformation phases (which confer toughness) and grain size.[120] Translucent zirconias are fabricated via refinement of processing routes, beginning with ultrafine equiaxed powders with yttrium stabilizer, reduction or elimination of light-scattering sintering aids and porosities, and higher sintering temperatures.[121] Judicious microstructural control holds the key to future dental materials development.

Residual Stresses

Residual stresses can develop in a porcelain veneer from CTE mismatch between the veneer and ceramic framework, and from rapid cooling during processing, especially in frameworks with low thermal diffusivities.[80,83,84,122–128] In some layer structures, thermal stresses may be beneficial; for example, by placing a weak outer porcelain veneer into compression. However, thermal stresses must average out to zero across any layer section, so that compression in one part of a prosthesis must inevitably be counterbalanced by tension elsewhere.[110] Moreover, these stresses are never uniform across the section, so any given layer may experience compression at one surface but tension at the other. Monolithic prostheses are not subject to the same concerns, although even there some stresses can arise from rapid cooling during processing, owing to the presence of substantial thermal gradients. Such stresses can have a profound influence on service lifetime.[110]

Monolithic Versus Veneered Structures

Porcelain-veneered ceramics have superior esthetics but are more vulnerable to fracture, especially chipping. Veneered crowns and FDPs still constitute mainstream dental practice, but are gradually being supplanted by monolithic prostheses fabricated from more resilient ceramics. Full-contour monoliths are much less susceptible to either occlusal surface or cementation fracture damage. The key to the advance of monoliths is improved esthetics. In modern-day zirconias, this is being achieved by fabricating more translucent microstructures or by infiltrating glass into outer surfaces to produce graded structures.[129–135]

Layer Thickness

In accord with intuition, thicker layers provide greater protection against fracture, partly because they diminish flexure and membrane stresses at any given occlusal load (a thickness squared relationship) and partly because they increase the distance cracks have to propagate before encountering a weak internal interface (veneered structures) or opposite surface (monoliths). The influence is strongest for radial cracks at the intaglio surface, with greater fatigue life with increased net layer thickness (see Fig. 7A).[106,107,136] In veneered structures, the critical bite forces to produce flexural radial cracks at the intaglio surface are only mildly sensitive to relative veneer-to-core thickness.[137,138] This allows the veneer/core thickness ratio to be tailored to

optimize the residual stress profile while retaining the flexural strength of the veneered restoration.

Tooth Contact Conditions

Changes in contact geometry primarily affect the ease and extent of occlusal surface damage.[139] Sharper, harder contacts in axial loading distribute the load over smaller areas, increasing local stresses and thereby making it easier to initiate cone cracks (see **Fig. 7**).[139] Such contacts are also likely to promote wear and abrasion damage and to initiate median cracks.[140] However, once these cracks grow away from the contact into the far field, they become less influenced by the nature of the contact.[106,107,109] Radial cracks (especially at the margins) are fairly insensitive to contact conditions. Off-axis contacts can enhance the failure process by initiating partial cones (sliding contacts) or edge chipping (near-edge contacts) (see **Fig. 7**A). From a design aspect, it is advisable to avoid sharp cusps near the edges of crowns, to prevent incurring damage in the first place. Sharp cusps are also more prone to quasiplastic deformation and wear. Contacts with soft materials relative to tooth modulus or hardness (eg, normal food items) or with blunt objects may suppress initiation of occlusal surface damage altogether by spreading the load over a greater area.[141]

Tooth Size and Shape

The geometry of prosthesis, most notably the dispositions of different cuspal shapes and connector configurations, plays a governing role in fracture resistance. Essentially, the greater the curvature (ie, the smaller the radius) of a contacting surface, the lower the bite force to initiate cracks associated with layer flexure.[142] Also, the smaller the crown height, the lower the force to drive longitudinal cracks around a side wall.[143] Clearly, these geometric factors are governed by the spatial restrictions imposed by opposing and adjacent dentition.

Substrate Modulus

The modulus of tooth dentin is about one-fifth that of enamel and an even smaller fraction than that of most ceramics used in crowns and FDPs.[144] A compliant substrate is an additional source of enhanced flexure,[145–148] and hence of radial fracture.[149–152] The modulus of cements or adhesives used to bond the dental prostheses to the underlying tooth structure is a factor of 2 to 5 times lower still, further degrading the load-bearing capacity,[145,148,153,154] and even thin cement layers (eg, <0.1 mm) can substantially enhance crown flexure. The use of high-modulus buildup materials and dental cements seems to be a useful strategy for minimizing flexural fractures.[145]

Surface State

It is evident that some precautions need to be taken in the preparation of prosthesis surfaces to stop cracks forming in the first place. Surface treatments can lead to the introduction of flaws that diminish strength. Aggressive sandblasting procedures with hard, coarse, abrasive particles under high air pressure used to provide greater adhesion at the cementation surfaces of crowns are in this category.[155–157] Likewise, the use of coarse diamond burs to grind down crown cusps in order to adjust the occlusal surfaces enhances the prospect of crack initiation. In contrast, although they compromise the load-bearing capacity of a restoration, prematurely initiated cracks from such damage may arrest within the structure, with little consequent effect on the final fracture condition.[142]

SUMMARY

Ceramic restorations are developed for esthetics, biocompatibility, and chemical durability. The composition, microstructure, and properties of ceramic materials determine the clinical indications of various classes of dental ceramics. Other factors that influence material selection include restoration designs (monolithic or layered structure), layer thickness, residual stresses, tooth contact conditions, tooth size and shape, elastic modulus of the adhesives and substrate (enamel or dentin), and surface state. Successful application of ceramic restorations ultimately depends on material selection, manufacturing technique, and restoration design.

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