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Engineering long term clinical success of advanced ceramic prostheses

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Abstract Biocompatability and, in some applications, esthetics make all-ceramic prostheses compelling choices but despite significant improvements in materials properties and toughening mechanisms, these still have significant failure rates. Factors that contribute to the degradation in strength and survival include material selection and prosthesis design which set the upper limit for performance. However, fabrication operations introduce damage that can be exacerbated by environmental conditions and clinical function. Using all-ceramic dental crowns as an example, experimentally derived models provide insight into the relationships between materials properties and initial critical loads to failure. Analysis of fabrication operations suggests strategies to minimize damage. Environmental conditions can create viscoplastic flow of supporting components which can contribute additional stress within the prosthesis. Fatigue is a particularly challenging problem, not only providing the energy to propagate existing damage but, when combined with the wet environment, can create new damage modes. While much is known, the influence of these new damage modes has not been completely elucidated. The role of complex prosthesis geometry and its interaction with other factors on damage initiation and propagation has yet to be well characterized.

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1 Background and objective

Ceramics are an appealing material choice for prostheses primarily because of their superb biocompatibility and, in at least some applications, esthetics. A primary goal of developments for these materials with biomedical applications has been to deliver long term excellent clinical performance. Innovations in ceramic science have created structural materials with increasing strength, nearly matching the fracture strength of metals. The inherent brittle nature of this class of materials has motivated advances in toughening mechanisms. The financial drivers for improvement are high. Orthopedic implants are estimated to have a world-wide market exceeding \$4.3 billion per year [12], replacing knees, hips, fingers, and spinal processes. Over 300,000 total knee replacement surgeries are performed each year in the United States [1]. Advanced ceramics are also used for dental prostheses, including implants, crowns, and bridges. Dental implants account for about \$1 billion of revenues each year and that market is estimated to be growing at 18 percent/year and dental crowns generate over \$2 billion each year in revenues with 20% of the units being all ceramic [2]. The aging population will likely drive the demand for prostheses of all types even higher.

Despite the advances in the materials and widespread utilization, ceramic prostheses have not always performed as predicted or desired. Ceramic hip implants from one manufacturer were recalled in 2001 because of greater than anticipated failure rates [52]. Despite increases in material strength, dental crowns continue to fail at a rate of approximately 3% each year [8] with highest fracture rates on posterior crowns and bridges where stresses are greatest.

The specific aim of this manuscript is to identify factors that must be engineered to design both initial and long term clinically successful prostheses. Factors related to initial

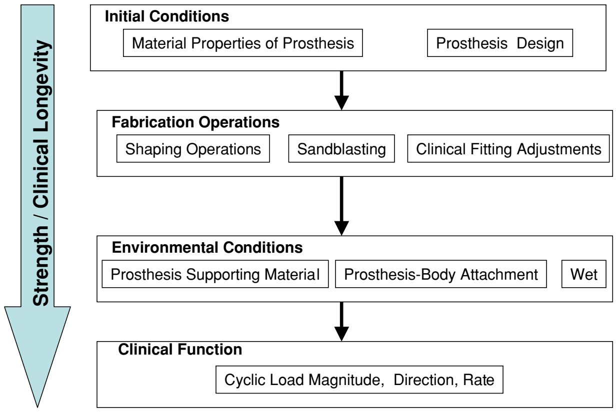


Fig. 1 Influences on prosthesis clinical survival

conditions (material selection and prosthesis design), fabrication operations (shaping, sandblasting, and clinical adjustments or manipulations), environmental conditions (prosthesis supporting material, prosthesis-body interface, wet), and clinical function (load magnitude, rate, and direction) on clinical survivability will be reviewed. Each set of conditions, summarized in Fig. 1, influences the strength and survival probability of the prosthesis.

All ceramic posterior dental crowns will be used as the application for discussing these factors that influence prosthesis strength and survival. Often the geometry is more complex, loads higher, and fit constraints more demanding for dental crowns compared with other prostheses. A ceramic crown (Fig. 2), typically 1.5 to 2 mm thick, can be fabricated from a monolithic material or be a layered ceramic structure. The brittle crown is held to the tooth by a layer (20–200 μm) of adhesive material with a relatively low elastic modulus. The supporting tooth structure can vary from natural tooth structure (dentin) to a very stiff supporting restoration. In function, the crown-cement-tooth system is subjected to cyclic loading with magnitude of approximately 100–700 N [13, 22] at the rate of about 1.5 Hz in a wet environment. Recent findings related to damage initiation and propagation in all-ceramic posterior dental crowns will be summarized.

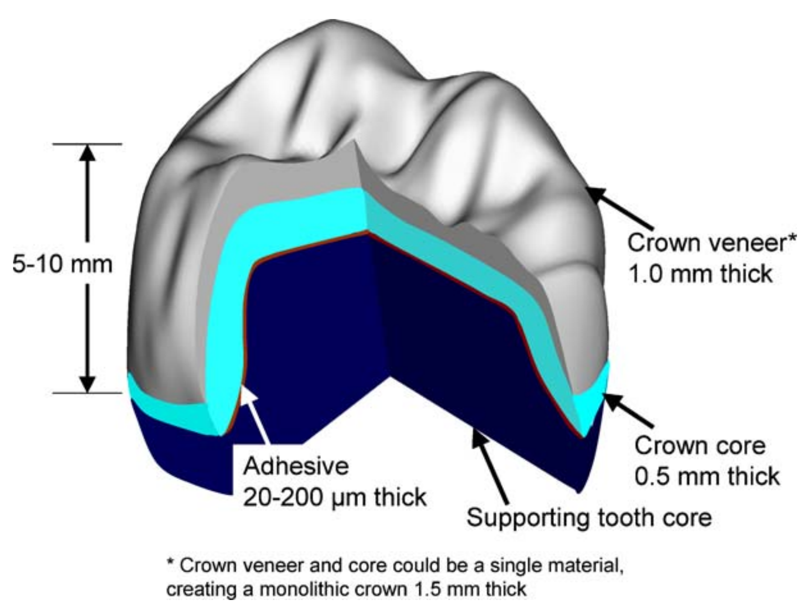
2 Initial conditions

Initial material selection undoubtedly has a significant influence on prosthesis performance. Material strength is a necessary condition for prosthesis survival but, as will be discussed, is not the only property that influences initial prosthesis strength. Equally important to long term success of a prosthesis is its design. Together, these factors set the stage for clinical performance and survival of any prosthesis.

2.1 Prosthesis material properties-initial prosthesis strength relationships

Initial strength of the material selected for a prosthesis is critically important. However, it is a necessary but not sufficient selection criterion. A modern leucite-reinforced glass ceramic (IPS-Empress, Ivoclar, Schaan, Lichtenstein) used in dental crowns has a fracture strength of 95 MPa [24] compared with the 70 MPa fracture strength of porcelains [3] yet clinical success of the glass ceramic crowns has been excellent [43, 44] while all-porcelain crowns had unacceptably high failure rates and are no longer in use. In a prosthesis, the underlying fracture mechanisms controlling damage initiation are more important than initial strength.

Fig. 2 Typical dental crown design and components of the crown-adhesive-tooth system



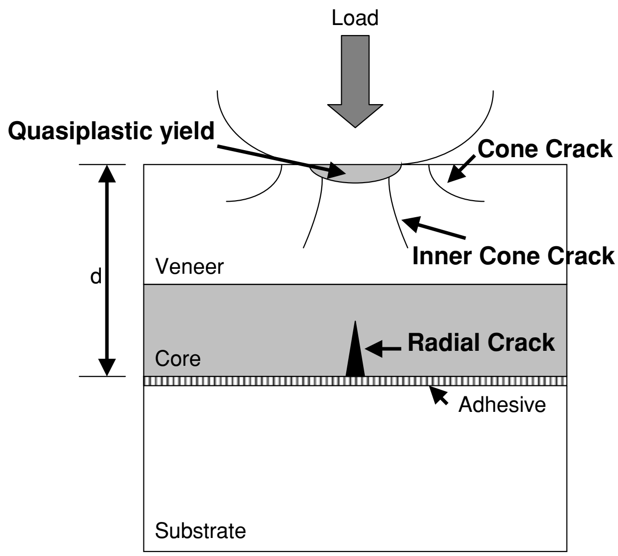


Fig. 3 Fracture mechanisms

In any brittle material of 1–2 mm cross-sectional thickness, failure (summarized in Fig. 3) can initiate beneath the point of load application on the surface, in the form of cone cracks or quasiplastic yield, or from the tensile surface of the prosthesis (dental crown in this discussion) in the form of radial fracture. Critical loads to initial damage for these three failure modes for six modern dental ceramics are given in Table 1. For monolithic layers of ceramic supported by a compliant polycarbonate substrate ($E = 2.3$ GPa), critical loads for cone cracks resulting from single-cycle loading is given by the relationship: $P_{\text{cone}} = A\ (T_c^2/E)r$ where A is a dimensionless coefficient (determined experimentally to be 8.6×10^3 for the ceramics investigated and listed in Table 1), T_c is the toughness of the ceramic (commonly termed K_{1c} in the engineering fracture community), E is the elastic modulus of the ceramic, and r is the radius of the loading indenter [35]. Toughness is the controlling parameter in this equation because the cone crack first develops as a surface ring and then propagates into the bulk of the materials.

For quasiplasticity, also a compressive surface phenomenon, the critical load is given by $P_{\text{quasiplasticity}} = D\ H_c\ (H_c/E)^2 r^2$ where D is a dimensionless coefficient (determined experimentally to be 0.85 for the ceramics investigated and listed in Table 1) and H_c is the indentation hardness (load/projected area, Vickers indentation) of the ceramic. Quasiplasticity is a yield process that determines the intensity of the shear stress responsible for creating the damage, controlled by hardness [35]. In all classes of materials investigated with the exception of some porcelains, critical load to quasiplastic yield is lower than critical load for cone cracks [36]. Note that critical load to *both* cone cracks or quasiplastic yield depends on the radius of the indenter. In the dental case, the contact radius on the tooth is typically on the order of 2–10 mm [26, 30] depending on the amount of wear on the cusp tip of the opposing tooth.

Radial cracks, initiate spontaneously from a starting flaw in the inner ceramic surface at the interface between the ceramic and the underlying more compliant substrate when the maximum tensile stress at the surface approaches the bulk flexure strength of the material. The critical load for radial fracture is given by $P_{\text{radial}} = B\ \sigma_F\ \mathrm{d}2'\log(E_c/E_s)$ where B is a dimensionless constant (determined experimentally to be 2.0 for the ceramics investigated and listed in Table 1), σ_F is the flexural strength, d is the ceramic thickness, E_c is the elastic modulus of the ceramic and E_s is the elastic modulus of the substrate [35, 50]. For the materials investigated, critical load to initiation of radial cracks is greatest in zirconia and then decrease progressively for aluminas, glass ceramics, and porcelains [11, 16–18, 36, 37, 40, 41, 45, 50].

For five classes (porcelain, glass ceramics, aluminas, and zirconias) of monolithic ceramics subjected to single cycle loading, radial cracks predominate when thickness of the ceramic is less than approximately 1 mm. This is a critical factor in dental crown design because the crown thickness is dictated by the amount of tooth structure the dentist is able to remove, preparing the tooth for the crown. Conventional dental wisdom dictates that ceramic crowns be at least 1.5 mm thick. However, the complex geometry on posterior teeth often results in lower thicknesses, particularly in the central region of the occlusal surface. Unfortunately, this is one area directly loaded during chewing functions. As a consequence, all-ceramic layered structures have been created using a strong and stiff ceramic core layer (which are generally not esthetic) veneered with a weaker but esthetic porcelain. Typically the core is approximately 0.5 mm thick and the veneering porcelain is approximately 1.0 mm thick. In these layers, the critical load to radial fracture can be predicted by a relationship similar to the one for monolithic ceramics using $P_{\text{radial}} = \mathrm{B}\ (E_{\text{eff}}/E_{\text{core}})\ d^{2'}\ \log(E_{ff}/E_s))$ where an equivalent value is substituted for the thickness ($d = d_{\text{core}} + d_{\text{veneer}}$) and $E_{\text{eff}}/E_{\text{core}}$ is given by $\{(1+\alpha\beta^3)/(1+\beta)^3 + (3\alpha\beta)/(1+\alpha\beta)(1+\beta)\}$ with $\alpha = (E_{\text{veneer}}/E_{\text{core}})$ and $\beta = d_{\text{veneer}}/d_{\text{core}}$ [18, 36, 39].

2.2 Prosthesis design

Prosthesis design is multifaceted, driven by the combination of materials properties and the clinical application. In the case of dental crowns, especially posterior crowns, the occlusal surface geometry is complex and must contact the opposing tooth in a prescribed manner throughout the entire range of motion during chewing functions [13]. The amount of tooth structure that can be removed to permit a crown to be placed is driven by the history of the tooth (e.g., fractured, decayed) but is usually limited by physiological design of the tooth

Table 1 Material properties and failure modes for six modern dental ceramics [15, 18, 34, 36]

Material	Brand Name	Function	Strength (MPa)	Elastic modulus (GPa)	Hardness (GPa)	Toughness (MPa·m$^{1/2}$)	Critical load to initial fracture (N)				Supplier
							Radial		Surface*		
							0.5 mm	1.0 mm	qp^{a}	cone	
				Ceramics							
Porcelain	Mark II	Veneer	130	68	6.4	0.92	58	233	443	325	Vita (1)
	Empress	Monolithic crown	120	67	5.6	1.4	185	378	336	800	Ivoclar (2)
Glass ceramic	Dicor	Monolithic crown					185	740	99	480	Corning (3)
	Empress II	Core	320	104	5.5	2.9	186	744	132	2211	Ivoclar (2)
Alumina	InCeram**	Core	550	270	12.3	3.0	143	574	219	911	Vita (1)
Zirconia	InCeram Zirconia	Core	440	245	13.1	3.5	131	523	312	1367	Vita (1)
	Prozyr***	Core	1450	205	12.0	5.4	463	1852	353	3890	Norton (4)
				Adhesives							
Resin	Rely X	Adhesive		4–6 (depends on thickness)							3M/ESPE (5)
Epoxy	RT Cure	Adhesive		3.5	0.4						Master Bond (6)
Zinc Phosphate	Fleck's	Adhesive		10–12 (depends on thickness)							(7) Mizzy, Inc.
				Substrates							
Dentin		Substrate		15–20	0.6	3.1					
Composite	Z-100	Substrate		18	0.8						3M/ESPE (5)
Poly-carbonate	Hyzod	Substrate		2.3	0.3						AIN Plastics (8)
				Indenter							
Tunsten carbide	Kenna-metal	Indenter		614	19						J&L Industrial (9)

*Cone crack or quasiplastic damage initiated; **glass infiltrated; ***Zirconia Y-TZP; ^{a}quasiplastic.
(1) Vita Zahnfabrik, Bad Säckingen, Germany; (2) Ivoclar, Schaan, Lichtenstein; (3) Corning Glass, Corning, NY; (4) Norton, East Granby, CN; (5) 3M/ESPE, St, Paul, Minnesota; (6) Master Bond, Inc., Hackensack, NJ; (7) Mizzy, Inc., Cherry Hill, NJ; (8) AIN Plastics, Inc., Virginia Beach, VA; (9) J&L Industrial, Chicago, IL.

structure which includes a vital nerve and vascular system in the pulp chamber in the middle of the tooth that should be preserved as often as possible. In general, a dental crown (Fig. 2) is approximately 1.5 mm thick at the top surface, 5–10 mm tall and the sides become increasingly thin at the bottom surface (away from the biting surface).

The effect of complex surface geometry on crack initiation and propagation has not been the focus of much attention. Glass plates ($E = 73$ GPa) with elastic modulus similar to that of dental enamel and dental porcelain were curved over spheres with radii from 20 mm to 4 mm to evaluate the effect of geometry [47]. The undersurfaces of the glass were undamaged or sandblasted and the glass was bonded to an epoxy resin substrate. Specimens with convex, concave, and flat surfaces were subjected to a single cycle load applied with a 4-mm radius tungsten carbide sphere. The anticipated classical damage occurred in the flat samples. In convex surfaces, radial cracks were more exaggerated than in the flat specimens. They initiated from the tensile surface beneath the indenter and propagated to the base of the hemispherical surfaces and through the thickness of the glass layer. At high loads, the radial cracks linked at the base, causing separation and dislodgement of triangular glass segments, similar to those seen in clinically failed all ceramic crowns (Fig. 4). In concave surfaces, the radial crack arms appeared to be encompassed by outer ring cracks at the top surface. In general, high contact loads were required to initiate radial cracks in curved surfaces, especially in concave surfaces. Consequently, geometry does influence strength and onset of damage. The effect of the complexity of actual dental crowns has yet to be subjected to the same systematic evaluation.

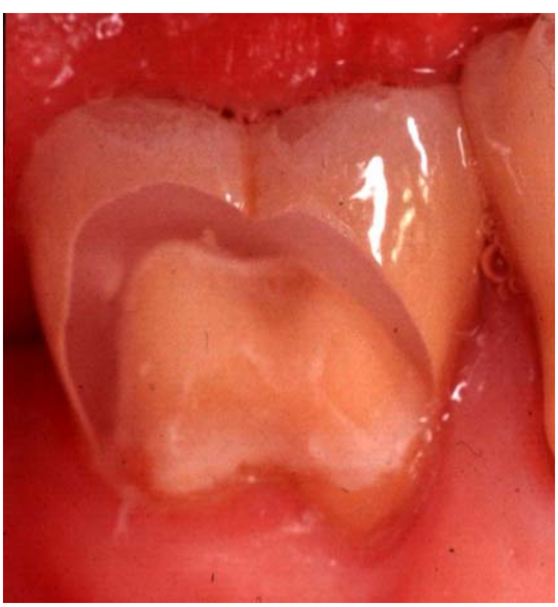


Fig. 4 Typical posterior all-ceramic dental crown fracture (complements of Dr. Kenneth Malament)

3 Fabrication operations influencing strength

Shaping operations of milling or grinding necessarily introduce damage in a prosthesis as do post-shaping modifications. In the case of dental crowns, post-shaping modifications can include sandblasting the inner surface to improve the surface for adhesion to hold the crown to the tooth, as well as modifications of the inner and outer surfaces for accurate clinical fit and function.

3.1 Shaping-induced damage

While initial crown material selection and thickness is critical, shaping-induced damage can remarkably reduce the material strength. Trilayers of 1.0 mm thick glass ($E = 70$ GPa) on 0.5 mm thick sapphire ($E = 417$ GPa) on 10 mm-thick polycarbonate ($E = 2.3$ GPa) with damage at varying surfaces were loaded with a 3.18 mm radius tungsten carbide sphere [36]. When the outer surface of the glass was abraded (simulating a post-grinding shaping operation), the critical load to failure was 700 N and a cone crack developed at the surface. When the underside of the glass (in contact with the sapphire) was abraded, load to failure was similar at 800 N but a radial crack rather than a cone crack developed in the glass. However, when the much stronger sapphire was abraded at the inner surface (equivalent to the cementation surface of a crown), the load to fracture was only 430 N and the initial damage was in the form of radial cracks from the tensile surface of the stronger sapphire! The load in the bilayer was carried by the high elastic modulus substrate.

The size of flaws introduced by processing techniques or fabrication damage can influence the strength of the ceramic. Controlled flaws, introduced by Vickers indentation at prescribed loads in the adhesive surface of ceramic monoliths and bilayer structures supported on a polycarbonate substrate, substantially reduced the critical load required to induce a radial crack [33]. This effect is magnified in multicycle fatigue loading as is discussed below.

Shaping operations, either machining or grinding, damage the prosthesis being created. The extent of the damage depends on the combination of the material, the machining parameters, and the cutting tool. Fracture strength of the material after machining can be reduced by as much as 50% in some materials [9, 49]. Analysis of edge chipping as a measure of machinability of modern dental materials (3 grain sizes of machinable glass ceramics, porcelains, aluminas, and zirconia) yielded a parametric equation predicting the degree of damage from milling relative to the depth of cut (d) and feed rate (f) of the machine tool and the hardness (H) and toughness (K_{1c}) of the material: $w = 0.984 \times 10^{-3}\ d^{0.514} f^{0.254} (H/K_{1c})\ r^{1/2}$. Surprisingly, the degree of edge chipping did not significantly improve or worsen when machining environments changed (e.g., in water-, alcohol-, or glycerin-based fluids) [21]. This suggests that a chemo-mechanical mechanism could be involved and, perhaps, used to advantage to improve cutting speeds without increasing damage. The cutting tool itself is also important; tools with hardness at least 3 times that of the material being shaped create smoother surfaces [48].

High speed grinding is commonly used in CAD/CAM systems to shape dental crowns. It is thought that grinding performance can be enhanced at high speeds [55]. Grinding force, surface roughness, and removal mechanisms were investigated at various grinding speeds for alumina and zirconia and compared with those of silicon carbide and silicon nitride. As long as the maximum chip thickness developed in the grinding process is less the crack penetration in the material, machining occurs below the ductile-brittle transition [5, 6]. The critical maximum undeformed chip thickness d_c at which this transition occurs is given by: $d_c = \beta$ $(E/H)(T/H)^2$ where $\beta = 0.15$, E is the elastic modulus, H is hardness, t is fracture toughness, and assuming that the depth of cut is equal to the machine infeed. For alumina, brittle fracture occurs in grinding at low speeds (40 m/s) but brittle-ductile transition occurs at higher speeds (160 m/s) yielding smoother surfaces. For the machining parameters investigated, the maximum chip thickness for zirconia was always below d_c where grinding was dominated by ductile cutting or quasiplatic diffusion of microdamage driven by shear stress regardless of grinding speed, making it less sensitive to machining damage [18, 46, 54]. Grinding forces are higher for zirconia than alumina and decrease with increasing grinding speed [55].

Most investigations focus on damage on flat samples. However, a dental crown has complex geometry. CAD/CAM systems developed to produce dental crowns are configured specifically for this task and necessarily use cutting tools with smaller diameter than for most industrial applications. Damage introduced by one milling machine (President-DCS, DSC Production AG, Allschwil, Switzerland) configuring the internal surfaces of a dental crown introduced surface flaws and microcracks to a depth as great as 0.1 mm [42]. This damage, on the internal surface of the crown, is particularly troublesome since it is the site of fracture initiation in most clinical crowns [30] and can significantly reduce fatigue life as discussed below.

3.2 Sandblasting

Sandblasting of the interior surface of crowns in commonplace, creating a roughened surface to enhance the adhesion to hold the crown on the tooth [7, 31, 32, 51]. However, sandblasting introduces surface flaws and defects that can compromise the short—and long-term strength of the crown. One mm thick flat samples of dense fine-grain alumina (AD 995, CoorsTek, Golden, CO) and zirconia (Proyr Y-TZP, Norton, Easts Granby, CT) were abraded with 50 μm Al_2O_3 particles for 5 sec at a standoff distance of 10 mm and compressed air pressure of 276 KPa, simulating typical laboratory procedures, and then bonded to polycarbonate substrates [56]. The damage created was substantial, penetrating 4 μm even into the surface in the zirconia and reducing its elastic modulus by over 14%, indicating a significant increase in microcrack density within the damage layer. Initial strength tests do not reflect any degradation in strength [4, 23] but as described below, this damage has a substantial influence on fatigue-related strength degradation.

3.3 Clinical adjustments and manipulations

In the case of dental crowns, clinical adjustments can include grinding the surface to improve the fit between the prosthesis and the opposing tooth. The amount of damage introduced depends on the bur used to make the adjustment, particularly on the effective radius of the "indenter" hitting the material surface. A comprehensive study of the damaging effects of combinations of dental burs and coolants (air vs. water) has not been reported. However, a sharp, pointed indentation, like that in diamond grinding tools, made with as little as an 0.1 N load in air reduces material strength the same amount as a 3000 N load made with a blunt (1.98 mm diameter) indenter [58].

4 Environmental conditions

A prosthesis functions in the biologic environment, supported by remaining tissue in a wet environment. For a dental crown, that environment includes the remaining tooth structure or an endosseous dental implant supporting the crown and the adhesive cement holding it in place.

4.1 Prosthesis supporting material

Dental crowns are supported by remaining tooth structure. This supporting structure can be a variety of materials. In some cases, a great deal of natural tooth structure remains. In that situation, the crown will be supported by 0.5–1.0 mm of dentin ($E = 15$–18 GPa) over the pulp chamber which is filled with fluids and nerves but has open communication through the root apex; structurally the pulp chamber can be considered to be a void. In the other extreme, much of the tooth may have decayed or fractured. Then a root canal procedure may have been completed, removing the organic material from the pulp chamber and filling it with a composite or metal post (E ranging from 50–300 GPa). Or, the tooth may have been lost completely, a high elastic modulus metal implant placed, and a crown fabricated to fit over the implant. For glass infiltrated alumina (InCeram) crowns on molar teeth, 35% of the crowns supported by dentin had failed at 10 years while none of the crowns supported by gold cores had failed [44]. Other materials have less spectacular differences. Finite element analysis of a stylized axi-symmetric tooth-cement-crown model suggests that stress distribution within the crown is influenced slightly by the tooth supporting core [28]. Accumulated damage from fatigue and propagation of inherent flaws and damage from shaping operations probably accounts for the remarkable difference between initial stress distributions and the high failure rates of some crowns.

4.2 Prosthesis-body interface

A prosthesis is stabilized in the body by a number of different mechanisms, ranging from a layer of adhesive to significant amounts of tissue ingrowth. Dental crowns are held to the tooth (or endosseous implant) using adhesive materials with elastic modulus of between 3 and 10 GPa. Ideally the thickness of this adhesive layer is approximately 20–50 μm. A compliant layer beneath a brittle material can result in radial cracks from loads applied to the surface; for crowns, the radial cracks can initiate from the cementation surface when an occlusal load is applied. Theory based on experiments using flat specimens of glass/adhesive/silicon layers [34], suggests that the critical load to radial fracture rises as the adhesive elastic modulus increases but drops for thicker cement layers. For a 1.5 mm thick ceramic ($E = 68$ GPa) on dentin ($E = 16$ GPa), a 10 μm thick layer of cement ($E = 10$ GPa) has a critical load to radial fracture of about 80% of its theoretical value (no cement and perfect bond between ceramic and substrate); a 100 μm thick adhesive layer, however, has a critical load to fracture of only 50% of its theoretical value. At 1000 μm, the critical load to failure drops to 30% of the theoretical value. While a 1000 μm thick adhesive layer not a clinically relevant situation, it does emphasize how dramatically changes in cement layer thickness can affect the strength of a crown-cement-tooth system.

4.3 Wet environment

Dental prostheses are subjected to a wet environment. (The effect of the wet environment relative to fatigue behavior is discussed below.) Specimens of flat glass or porcelain layers cemented to a freshly prepared or air-stored composite base and then exposed to a wet environment *spontaneously* fracture [29]. Clinically, this situation develops when the tooth supporting core is created from a composite material. Water from the mouth will diffuse into the composite, non-uniformly expanding the structure. A brittle ceramic crown cemented to by this expanding substrate will be stressed, perhaps to a critical point of fracture. If the polymer is in place sufficiently long to be saturated before the crown is fabricated and inserted in the patient's mouth, a potentially disastrous situation is avoided. It is anticipated that this mechanism could account for at least some of the failures in ceramic crowns but the delay between composite placement and crown placement as a clinical cause of failure has not been investigated. Analysis confirming experimental results show that for flat layers of clinically relevant materials and thicknesses, highest stresses develop first near but not at the edges and then progressively move toward the center of the sample and the driving force for cracks developing at the cement surface of the brittle ceramic later increase with water diffusion and time until equilibrium is reached (after weeks) [29]. Analysis of the phenomenon on specimens with more complex geometry is underway.

5 Clinical function

A prosthesis, once in place, is subjected to cyclic loading during function, creating the possibility for fatigue to initiate damage and/or exacerbate low level damage that would not otherwise cause failure. A dental crown is subjected to approximately 1500 load cycles each day with loads of 100–700 or more *N*, applied at a rate of 1.5 Hz [13, 22]. As teeth function, they come together, slide across each other, and then separate. The crown-cement-tooth system is subjected to complex loading with full unloading and relaxation. Recently a new fracture pattern has been identified (and is described below), but only in tests involving fatigue in a wet environment, exactly the sort of conditions to which a prosthesis will be subjected.

The interface between the prosthesis and the body may change. In the case of dental crowns, the adhesive layer has been shown to exhibit viscoplastic deformation over time [19, 20, 27, 53]. With creep, crown stresses in the area of the cyclic contact can incrementally rise, increasing the probability of radial fractures occurring in the brittle crown.

The influence on prosthesis strength of each of these mechanisms is discussed.

5.1 Fatigue

Ultimately, success of a prosthesis depends on its long term clinical success. In most cases, clinical usage subjects the prosthesis to loading and unloading. A typical dental crown will be subjected to approximately 1500 load cycles per day with loads as high as 700 N or greater [13, 14]. A five year service life is generally assumed to be a million loading cycles. Any damage that develops will likely be exacerbated by typical service conditions.

Slow crack growth from fatigue degrades strength of ceramic layers on polycarbonate by a factor of 2–4 [57].

Damage from sandblasting, described above, increases the density of microcracks within the damage layer [56]. For both alumina and zirconia, the damage has little effect on the dynamic fatigue (constant stress rate), reducing the strength by less than 10% compared to undamaged control specimens. However, damage intrinsic in a material or created during fabrication operations is magnified in cyclic loading like that of normal occlusal function. Strength reductions of 20% in alumina and 30% in zirconia are seen from cyclic fatigue. Mechanically driven mechanisms in cyclic loading exacerbate the slow crack growth, creating greater damage than found in dynamic loading conditions.

Zirconia materials offer a particular challenge since they are vulnerable to aging and the related $t - m$ phase transformations which weaken the material as reported in hip prostheses [25]. Zirconia-polycarbonate layers indented on the tensile surface (creating damage similar to that introduced by milling or sandblasting) and subjected to cyclic loading had strength reductions greater than those with undamaged surfaces or those subjected to monotonic loading [57]. It should be noted, however, that even with the degradation, the residual strength of the zirconia still exceeded the strength of alumina not subjected to fatigue.

Most prostheses function in a wet environment as is certainly the case for all-ceramic dental crowns. In such an environment, glass and porcelain layers supported by polycarbonate, exhibit a new failure mode, inner cone cracks [59]. In highly brittle materials like glasses and fine-grain polycrystalline ceramics, they can appear *before* radial cracks. Beginning close to the inner radius of damage zone, they are contained within the radius of the cone crack which appears first, and have a steeper angle to the surface than typical cone cracks (55 ± 15 degrees vs. 23 ± 5 degrees for out cone cracks) (Fig. 3). With glass plates bonded to polycarbonate substrates, loaded at the top surface with hard spheres in cyclic loading in water, [10, 59], outer cone cracks developed after a few loading cycles and propagated downward and outward in classical fashion [38]. Shortly thereafter, at about 100–1000 cycles (which, clinically, is almost immediately), inner cone cracks appear below the base of the outer cone and extend downward at a higher velocity than the classical cone crack, extending eventually to the ceramic-substrate interface [59]. No inner cone cracks developed in parallel test in a dry environment, suggesting that the environment plays a significant role in prosthesis survival, particularly relative to esthetic porcelain veneer layers on structural ceramic cores.

5.2 Creep

Dental adhesives are generally polymers, vulnerable to viscoplastic deformation when subjected to cyclic loading. With that flow, the support structure of the crown changes, potentially increasing the stresses in the crown, particularly at the already vulnerable tensile cementation surface. Stress in a pair of brittle ceramic layers (both monolith and bilayer) separated by an adhesive layer sharply increases after approximately 500,000 cycles for loads between 20 and 140 N which are at the lower end of normal chewing forces [28], but only when there are flaws (either intrinsic in the material or introduced by fabrication operations) at the cementation surface. The increased stress causes sub-surface cracks associated with the flaws, ultimately leading to failure. This mechanism is an important and often overlooked factor in fatigue damage.

6 Summary

Despite important advances in material science, performance of all-ceramic prostheses has still not reached its full potential. Initial strength of the ceramics is unquestionably an important starting point, however an array of other factors reduce the strength and potentially limit the clinical life of these prostheses. Fabrication operations, environmental conditions, and fatigue associated with clinical function all degrade the potential for success. Performance of all-ceramic dental crowns is a focus of investigation and serves as an important application for understanding fundamental aspects of damage initiation and propagation in ceramics. Posterior all-ceramic crowns still fail (Fig. 4), leaving the patient and the clinician frustrated. Experimentally derived models have provided insight into the relationship between materials properties and critical loads to damage initiation. Analysis of damage caused by fabrication operations suggests that careful control of machining parameters can minimize damage. But damage will still be created and the environmental conditions in which the prosthesis must function will exacerbate it. Fatigue in a wet environment is particularly troublesome, accelerating drops in strength not obvious with single load cycle tests. Mechanically-driven and water-enhanced damage propagation is clearly a real and troubling contributor to failures. The impact of competing damage modes in fatigue is not yet fully elucidated. Standard tests typically do not challenge the materials with the same complex loading that is expressed clinically. The consequences of the complex geometry of dental crowns on damage initiation and propagation is not fully understood. While significant advances have been made, much remains to be explored.

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