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Relationship between elastic and mechanical properties of dental ceramics and their index of brittleness

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Abstract

The purpose of the study was to verify the effects of a number of materials' parameters (crystalline content; Young's modulus, E; biaxial flexure strength, σ_i; Vickers hardness, VH; fracture toughness, K_{Ic}; fracture surface energy, γ_f; and index of brittleness, B) on the brittleness of dental ceramics. Five commercial dental ceramics with different contents of glass phase and crystalline particles were studied: a vitreous porcelain (VM7/V), a porcelain with 16 vol% leucite particles (d.Sign/D), a glass-ceramic with 29 vol% leucite particles (Empress/E1), a glass-ceramic with 58 vol% lithium-disilicate needle-like particles (Empress 2/E2), and a glass-infiltrated alumina composite with 65 vol% crystals (In-Ceram Alumina/IC). Discs were constructed according to manufacturers' instructions, ground and polished to final dimensions (12 mm × 1.1 mm). Elastic constants were determined by ultrasonic pulse-echo method. σ_i was determined by piston-on-3-balls method in inert condition. VH was determined using 19.6 N load and K_{Ic} was determined by indentation strength method. γ_f was calculated from the Griffith–Irwin relation and B by the ratio of HV to K_{Ic}. IC and E2 showed higher values of σ_i, E, K_{Ic} and γ_f, and lower values of B compared to leucite-based glass-ceramic and porcelains. Positive correlations were observed for σ_i versus K_{Ic}, and K_{Ic} versus $E^{1/2}$, however, E did not show relationship with HV and B. The increase of crystalline phase content is beneficial to decrease the brittleness of dental ceramics by means of both an increase in fracture surface energy and a lowering in index of brittleness.

Keywords: C. Mechanical properties; C. Fracture; C. Toughness; E. Biomedical applications; Ceramics

1. Introduction

The development of new all-ceramic restorative materials has been triggered by the search for esthetics, biocompatibility and strength. Despite the great clinical success of metal-ceramic crowns and fixed partial dentures, the increasing strength of core-veneered all-ceramic materials has also made possible the construction of metal-free restorations with more reliability. Now several all-ceramic systems are available for the construction of all-ceramic restorations. Among these, the most popular examples are leucite-reinforced porcelains, glass-ceramics, glass-infiltrated ceramic composites, and polycrystalline ceramics [1–8]. However, it is well recognized that the main drawback of all-ceramic restorations is their inherent brittleness and inability to undergo plastic deformation, resulting in high fracture rates in clinical trials [9–12].

To understand the fracture behavior of all-ceramic restorations, fracture mechanics principles are particularly important and can shed some light onto the fracture process and its causes. Since all-ceramic components contain intrinsic and extrinsic flaws, it is important to know what flaw size can be tolerated in a structure and what the expected lifetime of a defect-containing structure will be. The first fundamental and significant work on defects acting like stress concentrators was published in 1913 by Inglis [13], who demonstrated that sharp cracks are much more deleterious than blunt ones.

In the 1920s, Griffith developed his world famous energy-balance approach, establishing a relationship between fracture

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strength and crack size [14]. For ductile materials, Griffith's original work had to be later modified by Irwin [15]. Irwin suggested that the Griffith's equation should be rewritten to add the elastic deformation energy involved in the fracture process. Instead of developing an explicit relationship between the energy-consuming parameters, Irwin chose the parameter G (named after Griffith), which is the strain energy release rate.

While the Griffith's approach provides great understanding of the fracture process, an alternative method that studies directly the stress fields near the crack tip has gained attention in the materials science field. In fracture mechanics, the cracks can be characterized in terms of the stress intensity factor (K), which quantifies the stress field around a crack in a predominantly elastic material. In 1958, Irwin associated the concepts of Inglis and Griffith and showed that the strain energy release rate (G) is a function of the stress intensity factor (K) [16].

Rearranging the Griffith's equation, the stress intensity factor at the crack tip (K_{I}, I denotes uniaxial tensile or opening mode), which is also called K_{Ic} (critical stress intensity factor), is related to the stress at fracture in mode I according to the equation:

$$K_{\mathrm{Ic}} = \sigma_{\mathrm{f}} Y \sqrt{a} \tag{1}$$

where Y is a dimensionless constant that depends on the loading mode, the shape and dimensions of the material and the geometry and length of the crack, and a is the length of the crack from which the fracture propagates. K_{Ic} is a constant for a given material and is commonly referred to as fracture toughness. Eq. (1) can also be considered a failure criterion, since the brittle fracture of a material will occur when the product of the stress applied by the square root of a crack size is equal to or greater than the material's fracture toughness value.

Elastic modulus and fracture toughness are inherent properties of great importance for dental ceramics. Elastic modulus represents the stiffness of a material within the elastic range when tensile or compressive forces are applied [17] and is also an indication of the amount of reversible deformation that will occur in a structure when a load is applied to it. At the atomic scale, elastic strain can be expressed as a measure of the resistance to pulling apart adjacent atoms, expressed as small changes in inter-atomic spacing caused by stretching of bonds [18]. The fracture toughness may be defined as the measure of a material's ability to absorb energy from elastic deformation, in relation to the level of tensile stress that can be achieved near the crack tip before the initiation of catastrophic fracture [19]. Dental ceramics, as are all brittle materials, are unable to absorb appreciable quantities of elastic strain energy prior to fracture and fracture toughness can be considered a measure of the strain-energy absorbing ability of a brittle material [19].

All dental ceramics tend to fail at a critical strain of approximately 0.1% [20] and, for this reason, it has been argued that any increase in strength and toughness can only be achieved by an increase in the elastic modulus [21]. Several studies have already determined the physical and mechanical properties of commercial dental ceramics, such as elastic modulus and fracture toughness [1,2,21,22]. However, to the authors' knowledge, the relationship between these two properties has never been explored in dental materials, in spite of the fact that according to the Griffith–Irwin equation ($K_{\mathrm{Ic}} = \sqrt{2E\gamma_{\mathrm{s}}}$, in plane stress – where E is the Young's modulus, and γ_{s} is the surface energy per unit area), they are supposed to be positively correlated. For polycrystalline or multi-phase materials this relationship can be more complex, since elastic modulus can vary on micro-scale level at the crack front, and alternative strain-energy consumption mechanisms (deflection, crack bridging, micro-cracking, phase transformation) can be activated. To take into account the effects of toughening mechanisms, γ_{s} is usually replaced by γ_{f}, fracture surface energy (energy to create unit surface area) [23].

K_{Ic} and γ_{f} are considered suitable brittleness parameters when the loading is purely elastic (e.g. fracture in tension or thermal shock fracture), but when contact stresses are involved (as in scratch, impact, wear, erosion and machining events) other brittleness parameters which compare deformation to fracture processes may be more useful [24]. The ratio of hardness to toughness, H/K_{c}, determined under sharp contact like a Vickers indenter has been proposed as a simple brittleness parameter, B, since H is a measure of resistance to deformation and K_{c} is the resistance to fracture [25]. It seems this parameter has not been applied yet to evaluate dental ceramics.

Therefore, the objective of the present study was to determine the effects of a number of materials' parameters, such as crystalline content, Young's modulus (E), biaxial flexure strength (σ_i), Vickers hardness (VH), fracture toughness (K_{Ic}), fracture surface energy (γ_{f};) and index of brittleness (B) on the brittleness of dental ceramics. The main hypothesis tested was that strong correlations would be found between the parameters analyzed.

2. Materials and methods

The dental ceramics used in this study are described in Table 1. Materials were selected to provide different types of clinically relevant microstructures. Fifteen disks (12 mm in diameter and 2 mm thick) of each material were produced according to each manufacturer's instructions. Porcelains were prepared by the vibration–condensation method and sintered in a dental porcelain furnace (Keramat I, Knebel, Porto Alegre, Brazil) following the firing schedules recommended by the manufacturers. Glass-ceramics were processed by the heat-press technique using a specific oven (EP 600, Ivoclar Vivadent, Schaan, Liechtenstein). In-Ceram Alumina composite was processed by a lanthanum-silicate glass infiltrating a porous partially sintered alumina preform made by slip casting. The sintering of alumina preform and the glass infiltration cycles were carried out in a special furnace (InCeramat II, Vita Zahnfabrik, Bad Sackingen, Germany). All disks were machined to reduce thickness to 1.3 mm, following the guidelines in ASTM C 1161 [26]. Then, one of the disk surfaces was mirror polished using a polishing machine (Ecomet 3, Buehler, Lake Bluff, IL, USA) with diamond