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A review of engineered zirconia surfaces in biomedical applications

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Abstract

Zirconia is widely used for load-bearing functional structures in medicine and dentistry. The quality of engineered zirconia surfaces determines not only the fracture and fatigue behaviour but also the low temperature degradation (ageing sensitivity), bacterial colonization and bonding strength of zirconia devices. This paper reviews the current manufacturing techniques for fabrication of zirconia surfaces in biomedical applications, particularly, in tooth and joint replacements, and influences of the zirconia surface quality on their functional behaviours. It discusses emerging manufacturing techniques and challenges for fabrication of zirconia surfaces in biomedical applications.

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Keywords: Dental restotations; joint repalcement; manufacuture; surfece; zirconia

1. Introduction

Load-bearing surfaces are widely used in medicine and dentistry as medical and dental prostheses, such as joints and crowns/bridges, which are increasingly needed by millions of people in our aging societies. A wide spectrum of materials is used for these functional surfaces, including metals, ceramics, polymers and composites. Among these materials, zirconia combines high strength and fracture toughness, outstanding slow crack growth resistance, low thermal conductivity, high ionic conductivity, attractive biocompatibility and chemical inertness [1–5]. The tetragonal-to-monoclinic phase transformation in zirconia [6] results in a high damage tolerance and enhanced fracture toughness [2]. This transformation toughening mechanism makes zirconia the strongest and most fracture resistant material of all bio-structural ceramics [7–10].

Zirconia has long been used in engineering as ferrules in optic fiber connectors [11–13], environmental filters, and mechanical components [14,15]. Zirconia has been used in surgical implants such as femoral heads for total hip replacements for nearly a half century [16]. Over the last thirty years, zirconia has also been applied in restorative dentistry as dental implants and abutments [17]. Over the last twenty years, zirconia has been used as cores for bi-layered posteriors as dental crowns and bridges [2,4,9,18,19]. Recently, coloured zirconia with improved translucency has been developed to closely match colours of human teeth. This new material has a flexural strength of 900–1400 MPa and a fracture toughness of up to 6 MPa $m^{1/2}$ [20]. Such advantages have led to an exponential increase in the use of zirconia for monolithic crowns and bridges for posterior applications [21].

In the application of zirconia in medical and dental devices, the material must be machined and surface-treated to obtain not only mechanical functions such as wear and fatigue resistances, but also biomedical capabilities such as cell adhesion, bacterial decolonization and bonding strength. This paper reviews current manufacturing techniques for fabrication of zirconia surfaces in biomedical applications, particularly, in tooth and joint replacements, and influences of

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the zirconia surface quality on their functional behaviours. It discusses emerging manufacturing techniques and challenges for fabrication of zirconia materials.

2. Zirconia Materials

2.1. Zirconia microstructures

Most structural zirconia materials are various zirconia containing ceramic alloys, in which zirconia is doped with other oxides, such as magnesium oxide (MgO), yttrium oxide (Y_2O_3), calcium oxide (CaO), and cerium oxide (Ce_2O_3), to form stabilized tetragonal or cubic phases [8]. In medical and dental applications, yttrium-stabilized tetragonal zirconia polycrystal (Y-TZP) is most popular.

Pre-sintered Y-TZP is porous and has low strength. An example of pre-sintered Y-TZP is IPS e.max Zir CAD (Ivoclar Vivadent), which is designed for dental crowns and bridges using chairside dental CAD/CAM systems. The material is 97% tetragonal and 3% monoclinic zirconia, containing approximately 87–95 wt. % ZrO_2, 4–6 wt. % Y_2O_3 as a stabilizer for retention of tetragonal grains to room temperature, 1–5 wt. % HfO_2 as binders, and 0.1–1 wt. % Al_2O_3 as sintering aids to facilitate the densification of zirconia [22,23]. It has a highly isolated or interconnected porous microstructure with a porosity of approximately 47.3–49.3 vol. % [2] and Y-TZP crystals of approximately 300-nm grain size [24].

Sintered Y-TZP is obtained at temperatures between 900 °C and 1600 °C depending on required microstructures. In general, coarse zirconia microstructures are produced at higher temperatures and longer dwell times [2]. For dental prostheses, sintering is often conducted at 1200 °C–1600 °C [2], resulting in highly compacted zirconia grains of 300 nm or less [2].

2.2. Mechanical properties

The mechanical properties of zirconia are determined by their microstructures and measurement scales [25, 26]. From a machining point of view, the indentation behaviour of zirconia at the micro/nano scales is essential to its manufacturability, because the micro/nano indentation properties are directly associated with material responses to diamond or tungsten carbide abrasive machining processes [24,27–29]. Several studies have focused not only on the indentation hardness and modulus but also on the resistance to plasticity and the resistance to machining-induced cracking based on the Sakai–Nowak model [30]. The resistance to plasticity indicates an independent property from the indenter geometry and represents the plasticity of a material. The resistance to machining-induced cracking is defined as the inverse degree of damage for a unit applied work [30]. These properties can be used to predict the machining behaviour of zirconia materials. Table 1 shows the comparison of the mechanical properties of pre-sintered (IPS emax ZirCAD) and sintered Y-TZP at 1200°C with a holding time of 2 hours [24,27–29]. Sintering decreased the porosity from approximately 47.3–49.3 vol. % to less than 0.5 vol. %. Sintered zirconia is more than 10 times harder and stronger, and approximately 5 times tougher than the pre-sintered state. The much higher resistance to machining-induced cracking for sintered Y-TZP indicates its higher degree of damage tolerance but it is less deformable than pre-sintered Y-TZP [24,27–29].

Table 1. Properties of pre-sintered and sintered Y-TZP materials [24,27–29]

Property	Pre-sintered	Sintered at 1200°C
Porosity (vol. %)	47.3–49.3	< 0.5%
Density (g/cm^3)	3.0–3.21	6.09
Nanohardness (GPa)	1.11±0.34	13.15
Young's modulus (GPa)	29.34±4.93	168.19
Fracture toughness (MPa m$^{1/2}$)	0.8	5.5
Flexural strength (MPa)	50–90	900
Resistance to plasticity (GPa)	3.28±0.98	43.22±9.59
Resistance to machining-induced cracking (J/m^2)	128.90±24.1	400

3. Manufacturing of Zirconia

The selection of white (or soft in dentistry) and hard machining processes for zirconia is based on the microstructure and mechanical properties of the material. Both processes have advantages and disadvantages.

3.1. White machining of pre-sintered zirconia

White machining is used to machine pre-sintered zirconia to obtain complex profiles such as dental crowns and bridges. It is a dry milling process using tungsten carbide milling tools [4,31]. This process enables a rapid and cost-effective generation of complex profiles of zirconia components. However, white machining also produces extensive surface damage on machined surfaces. Fig. 1 shows a scanning electron micrograph of surface fracture, crack and micro-chips produced in pre-sintered Y-TZP during a CNC white machining using a tungsten carbide milling tool [31]. Intragranular and transgranular fractures easily occurred due to weakly interconnected porous structures in pre-sintered state. This machining-induced damage cannot be naturally

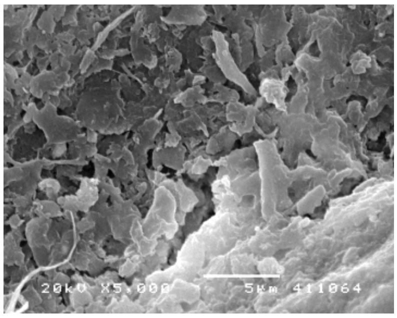

Fig. 1. Surface damage in pre-sintered Y-TZP produced in CNC milling process using a tungsten carbide milling tool [31].