From porcelain-fused-to-metal to zirconia: Clinical and experimental considerations

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Abstract

Objective. The interest of dental research in metal-free restorations has been rising in the last 20 years following the introduction of innovative all-ceramic materials in the daily practice. In particular, high strength ceramics and related CAD/CAM techniques have widely increased the clinical indications of metal-free prostheses, showing more favourable mechanical characteristics compared to the early ceramic materials.

The purpose of the present paper is providing a brief review on the all-ceramic dental materials, evaluating pros and cons in the light of the most recent scientific results and of the authors’ clinical experience.

Materials. A structured review of the literature was given on the basis of medical and engineering papers published in the last decades on the use of dental ceramics and zirconia in particular. The experimental and clinical findings of the most relevant researches were reported.

Results. Zirconia is one of the most promising restorative materials, because it yields very favourable mechanical properties and reasonable esthetic. Several in vitro and in vivo investigations reported suitable strength and mechanical performances of zirconia, compatible with clinical serviceability as a framework material for both single crowns and short-span fixed partial dentures. However, clinical results are not comparable, at the moment, with conventional metal–ceramic restorations, neither is there sufficient long-term data for validating the clinical potential of zirconia in the long run.

Significance. The use of zirconia frameworks for long-span fixed partial dentures or for implant-supported restorations is currently under evaluation and further in vivo, long-term clinical studies will be needed to provide scientific evidence for drawing solid guidelines.

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1. Introduction to the review

In the last decades, since the development of porcelain-fused-to-metal (PFM) procedures in the early sixties, metal–ceramic restorations have represented the “gold standard” for years in prosthetic dentistry, thanks to their good mechanical properties and to somewhat satisfactory esthetic results, along with a clinically acceptable quality of their marginal and internal adaptation [1-7]. The predictability and consistency of positive clinical results, validated by long-term scientific evidence, the ease and accuracy of the conventional casting procedures, as well as the findings of rare adverse reactions to precious alloys have made PFM crowns and bridges more and more popular and widespread over time.
Nevertheless, the technical procedures of investing wax patterns and casting precious metal alloys involve many technical variables and a considerable number of operative steps and firing cycles, making the final quality of the restorations highly technique-sensitive. Moreover, the metal framework and the layer of opaque porcelain needed for masking the underlaying metal grayish shade are likely to introduce a significant limitation for the esthetic result due to the absence of translucency, especially when a clear tooth color is to be reproduced: in fact, metal–ceramic restorations can only absorb or reflect light, while dental tissues show a high degree of translucency [8]. Furthermore, from an economic standpoint, the cost of precious metals has markedly risen over the years [9].

Since the early introduction of the porcelain jacket single crowns into the dental practice, at the end of the XIX century, dental ceramics have been considered among the most promising restorative materials, because of noticeable prosthetic advantages: esthetic appearance, chromatic stability, biocompatibility, low plaque retention and fluids absorption, high hardness, wear resistance, low thermal conductivity, chemical inertness. However, early dental ceramics encountered a undeniable degree of resistance in the clinicians’ perspective of an extensive, routine use due to limiting factors for an acceptable restoration longevity, mainly associated to mechanical shortcomings: brittleness, low tensile strength and fracture toughness, ease of crack propagation, poor marginal fit, difficulty of repair.

In the last 30 years, the growing patients’ demand for highly esthetic and natural–appearing restorations has led to the development of new all-ceramic materials, whose mechanical characteristics have been dramatically improved, in order to provide suitable longevity and limitation of the technical problems. In a few cases, however, the increasing industrial pressure on the one hand and growing enthusiasm for attractive esthetic outcomes on the other have led to an early introduction on the market of unreliable, not sufficiently tested products, resulting in commercial and clinical disasters.

To date, a large number of studies and scientific data have been produced in order to investigate the mechanical properties of dental ceramics, mainly aimed at getting an evidence-based validation of the metal-free materials and of the related manufacturing systems.

The purpose of the present paper is providing a brief review on the all-ceramic dental materials and their related production techniques, evaluating pros and cons in the light of the most recent scientific results and of the personal prosthodontic experience of the authors.

2. Literature

2.1. Glass- and alumina-based dental ceramics

An ideal all-ceramic dental material should exhibit excellent esthetic characteristics, like translucency, natural tooth color, outstanding light transmission and, at the same time, optimal mechanical properties, like flexural strength ($\sigma$), fracture toughness and limited crack propagation at the functional and parafunctional load conditions, in order to ensure lifetime serviceability. Unfortunately, today, none of the available dental ceramics fulfills all of these requirements at the same time, neither is any of them suitable for every different clinical situation both in anterior and posterior regions. A comprehensive review on this topic was provided in 2007 by Conrad et al. [10].

First of all, it has to be pointed out that not all of the dental ceramics show equally favorable esthetic features: to date, an inverse proportion between strength (i.e., the mechanical performance) and optical properties (i.e., the esthetic appearance) still seems to be a rule.

Feldspathic ceramics usually provide excellent esthetic, together with very good biocompatibility and mechanical resistance to compressive forces, but, unfortunately, they easily fracture under shear loads, owing to low tensile strength. A first, successful attempt at strengthening feldspathic porcelain was made by McLean and Hughes [11] in the mid-sixties, who dramatically reinforced dental porcelain with the addition of up to 50% aluminum oxide powder.

Today, the best esthetic characteristics are still displayed by the class of the glass–ceramics: optimal light transmission, high translucency and natural, tooth-like colors also in the presence of very light shades. Early glass–ceramics, like Dicor (Dentsply, USA), no longer on the market, attained only limited success because of the modest survival rates [12]; on the contrary, leucite-reinforced glass–ceramic (IPS Empress—Ivoclar Vivadent, Lichtenstein) have been highly appreciated for more than 20 years thanks to their outstanding esthetic performances [13–16]. Inside the Empress ceramic ingots, leucite crystals, that measure only a few microns, are embedded in a glass matrix; the wax pattern of the restoration is invested and burnout, then, at high temperature the glass–ceramic is pressed into the mold (“hot-pressed”). Due to their low values of flexural strength (~100–120 MPa) [17], leucite-reinforced ceramics are only indicated in the anterior region, where esthetic is paramount, both for single crowns (SCs) and laminate veneers. In a long-term study (11 years), a remarkable survival rate of 98.9% was evidenced with IPS Empress anterior SCs, however such a value dropped to 84.4% in the posterior region [18]. With respect to the veneers, IPS Empress yielded a success rate of 98.8% after 6 years [19], equivalent to the positive results reported for the veneers made of feldspathic ceramics [91–94% at 12 years] [20,21].

A significant improvement in clinical performance was introduced by lithium disilicate glass–ceramics, veneered with fluorapatite-based ceramics, like IPS Empress 2 (Ivoclar Vivadent, Lichtenstein), showing higher flexural strength (~350 MPa) than the precedent ones and, at the same time, very appealing translucency, much more suitable than in zirconia-based ceramics [8,22]. For such promising characteristics, lithium disilicate glass–ceramics have been advised for clinical use in SCs (molar region excluded) and 3-units fixed partial dentures (FPDs) in the anterior region; in the last years, moreover, both their mechanical and optical properties have been enhanced, with the development of IPS e.max Press (Ivoclar Vivadent, Lichtenstein), thanks to some technical improvements in the production process [23]. Favorable survival rates for SCs have been reported, from 95% [24] to 100% [25] at 5 years; less encouraging was the survival data related to the IPS Empress 2 FPDs, with very different results at 2 years (50% survival rate according to Taskonaz and Sert-
guz [26] compared to 93% reported by Esquivel-Upshaw et al. [27] but quite poor results at 5 years (70%) [25].

Another noticeable improvement in the mechanical properties of all-ceramic restorations was offered by the so-called "glass-infiltrated high-strength ceramic core systems", developed for the first time in the late eighties with In-Ceram Alumina, followed, after some years, by In-Ceram Spinell and In-Ceram Zirconia (VITA Zahnfabrik, Germany). All of these oxide-ceramics allow the realization of highly stable frameworks for SCs or three-unit bridges with one pontic, based on the so-called "slip-casting technique": a semi-liquid mixture containing up to 80 wt% of metal-oxides, like Al$_2$O$_3$ (In-Ceram Alumina), MgAl$_2$O$_4$ (In-Ceram Spinell) or Al$_2$O$_3$+ZrO$_2$ (In-Ceram Zirconia), is sintered to a refractory die, so creating a porous, oxide-ceramic core that undergoes a further firing cycle for lanthanum glass infiltration. Thanks to such a process, the framework flexural strength and load-bearing capacity are remarkably enhanced: the infiltrated glass fills the minute spaces and voids that might initiate cracks and induce excessive stress concentrations in the core structure [28]. Eventually, the esthetic veneering material (i.e. a feldspathic ceramic) is layered on the core surface. In-Ceram Alumina (described flexural strength between 350 and 500 MPa [17,22,29–32]) has been on the market for more than 20 years, well accepted by clinicians not only for its fairly good mechanical properties but also for a natural esthetic appearance. Survival rates shown by many studies on In-Ceram Alumina, with observational periods comprised between 3 and 7 years, ranged from 94% to 99.1% for SCs, with a higher amount of complications in the posterior sites (crown fractures or chippings) [29,33–36]. A similar trend was evidenced for the In-Ceram Alumina posterior bridges when compared to the anterior 3-unit FPDs: a survival rate of 83% in the posterior sites, mainly influenced by connector fractures, was evidenced by Sorensen et al. in a three years study, raising to 100% in the anterior regions [37]. Other studies with longer observational periods (between 5 and 10 years) showed, for such 3-unit FPDs, survival rates ranging from 73.9% at 5 years [38] to 88% at 10 years [39]. In-Ceram Spinell is characterized by a lower mechanical strength ($\sigma$ = 350–400 MPa) than the other glass-infiltrated ceramics, but shows better optical properties, like high translucency and optimal light diffusion. Its use has been limited to the anterior crowns, with fairly good survival rates (from 97.5% at 5 years [40]. In-Ceram Zirconia is a glass-infiltrated zirconia-toughened alumina (ZTA), in which, for the first time, zirconium oxide was used as in a dental ceramic; as reported below, thanks to its metastable nature, zirconia is a high performance ceramic material [41]. High strength cores are composed of 67 wt% of aluminum oxide +33% of 12 mol% cerium-partially stabilized zirconium oxide [16], so that zirconia crystals (grain size $<1 \mu$m) are embedded in an alumina matrix (larger grains $<2-6 \mu$m, high elastic modulus) in such a composition that yields the highest tenacity and flexure strength inside this class of ceramics ($\sigma$ = 400–800 MPa): microcracks may trigger the so-called "transformation toughening" of zirconia (see below), so that a crack tip is more often seen to propagate through the alumina matrix surrounding the transformed crystals [42,43]. ZTA can be manufactured according to two different processes: soft machining or slip-casting. The latter presents the advantage of a more limited shrinkage but, at the same time, higher porosity and poorer mechanical properties than yttrium partially stabilized tetragonal zirconia polycrystal (Y-TZP) [43–46], the strongest and most commonly used zirconia-based ceramic. Moreover, stabilization by cerium oxide provides better thermal stability and resistance to Low Temperature Degradation (LTD) than Y-TZP [47–50]. According to the manufacturer's indications, ZTA is suitable for 3-unit FPD frameworks with one pontic in the posterior sites (reported survival rate at 3 years = 94.5% [51]), while, on the other hand, it exhibits intense opacity and low translucency [22]; although providing an efficient masking power in presence of dark colors or discolored teeth, ZTA is not appropriate for situations in which a high esthetic result is paramount.

Besides slip-casting for the glass-infiltration technique, another widespread and successful production system for alumina-based restorations is CAD/CAM industrial manufacturing of densely sintered, high-purity alumina (Procera AllCeram, Nobel Biocare AB, Goteborg Sweden) introduced in the early nineties and extensively utilized for both single-unit restorations and 3-unit anterior FPDs so far. The Procera AllCeram core is realized by compacting, with an industrial process performed at a centralized manufacturing plant in Sweden, high purity aluminum oxide against an enlarged, refractory die of the prepared tooth obtained through a scan- sion by the dental technician; eventually, the coping is milled in the outer aspect and then sintered to full density. In the end, the resulting, leucite-free porcelain framework, containing about 99.9% alumina in a polycrystalline state, is veneered with low-fusing feldspathic ceramic. AllCeram cores are characterized by a higher flexural strength than glass-infiltrated pre-sintered alumina [52] that, in addition to the pure and homogeneous structure of aluminum oxide and to the accuracy of the sinterization process, can explain the very good mechanical performance and resistance to fracture, still maintaining a fair translucency and opalescence [14,53–55]. As to the marginal fit, gaps ranging between 60 and 80 $\mu$m were detected, demonstrating a suitable prosthetic precision of fit [56,57] and also from the clinical standpoint marginal integrity was reported to be excellent or acceptable [54,55,58].

In a recent study, the use of luting resin cement with Procera AllCeram alumina SCs significantly showed a lower amount of mechanical complications in a sample of 209 tooth- and implant-supported SCs, with cumulative survival and success rates of 95.2% and 90.9%, respectively, after 6 years of function [55]. Other authors observed lower microleakage when aluminum oxide blasting plus silane treatment was performed before the luting phase [57,59,60].

Various studies in the last years reported a quite good clinical serviceability of polycrystalline, densely sintered alumina SCs. Cumulative survival rates of 97.7% and 93.5% and cumulative success rates of 97.7% and 92.2% after 5 and 10 years, respectively were recorded [58]. A recent retrospective clinical study on 86 alumina SCs reported success rates, in anterior esthetic sites, of 100% (supported by natural tooth) and 98.3% (implant-supported) over 4 years of function; the total crown success rate was 98.8% [54]. In another clinical investigation, the cumulative survival rate was 100% in the anterior region and 98.8% in the posterior region after 5 and 7 years of service;
clinical success was achieved irrespective of the tooth position [61].

From such data, it can be inferred that Procera AllCeram alumina crowns can provide reliable, high strength clinical solutions for anterior sites restorations, both supported by natural teeth and by osseointegrated implants, with a preferential indication when high load conditions are foreseen and the more translucent and esthetic glass–ceramics would be likely to fail. On the contrary, some criticism arose about the clinical use of laminate veneers made of high density polycrystalline alumina (Procera AllCeram, Nobel Biocare AB, Goteborg Sweden). A 3D finite element analysis (FEA) study based on a very careful and up-to-date modeling technique was conducted to evaluate the biomechanical behavior of feldspathic versus alumina porcelain veneers [62]; its results demonstrated different performances in terms of elastic deformations and stress distributions, feldspathic ceramics better simulating the biomechanics of sound enamel. On the contrary, alumina significantly withstooded deformations inside the veneer structure but, at the same time, high stress concentrations at the adhesive interfaces were generated; these might negatively influence the clinical performances of both the side of bonding agent and of resin cement, with high risk of bonding failure [62]. Another drawback for an extensive use of such veneers is that densely sintered alumina is a non-etchable material and the adhesion between this polycrystalline ceramic and dental tissues remains controversial, to date; compared with the high predictability of the adhesion between feldspathic ceramics and resin cements, this aspect represents a further noticeable limitation that, along with a poorer esthetic outcome and a scarceness of scientific support, does not justify to replace materials like glass/pressed ceramics for laminate veneers [63].

2.2. Zirconia

Since it was introduced in Dentistry, the polycrystalline zirconium dioxide (zirconia) resulted particularly attractive in prosthodontics, due to its excellent mechanical properties and improved natural-looking appearance compared to metal–ceramics [43].

Due to the increasing interest in dental applications of such an innovative material, a brief review of the main related topics is presented here, with no reference to cementation and to the optical/chromatic properties, in that these issues will be treated in different articles.

Zirconia is a non-etchable material and the adhesion between this polycrystalline ceramic and dental tissues remains controversial, to date; compared with the high predictability of the adhesion between feldspathic ceramics and resin cements, this aspect represents a further noticeable limitation that, along with a poorer esthetic outcome and a scarceness of scientific support, does not justify to replace materials like glass/pressed ceramics for laminate veneers [63].

Creating high compressive stresses in the material: as a matter of fact, until the twenties, in spite of the worldwide abundance of zirconia in very large quantities, such a material could not be utilized as a refractory for brick manufacturing due to the onset of severe cracks, as reported by Lughi and Sergio [64] in their elegant and updated review on the mechanical properties of zirconia. In the thirties, such a drawback was clarified and interpreted as a potential, outstanding property of the zirconia [41,72]: when alloyed with other "cubic" oxides like MgO, CaO, Y2O3 and CeO2 (so-called "stabilizers"), the phase transformation could be prevented, so retaining the zirconia crystals in their tetragonal or cubic shape at room temperature, in a thermodynamically metastable state. This property is the main reason why the biomedical research over the last years has been increasingly focusing on such a material, in that it can induce a remarkable increase in fracture toughness of the material by hindering (but not preventing [10,16,73]) the propagation of a crack; in fact, at a possible crack tip, tensile stress concentration generates the transformation from metastable (t) ZrO2 to the (m) crystalline phase. The consequent volume increase of the crystals, constrained by the surrounding ones, results in a favorable compressive stress, that acts as a crack-limit [64]. Such a mechanism has been defined "transformation toughening" or "phase transformation toughening (PTT) [71,74] and, along with the grain size of such a material, can explain why zirconia presents the highest flexural strength and fracture toughness among all the other ceramics.

At room temperature, the transformation from tetragonal to monoclinic is a one-way process. This means that, once it takes place, the crack-hindering effect cannot be exploited for limiting further fractures, "like a used match cannot be lit again" [64]. Heating the material at a temperature between 900 °C and 1000 °C for a short time, the process can be reversed [75,76]; in this case, the phase transition from monoclinic back to tetragonal form, rather than making crystals available again for further transformation and crack repair, generates a relaxation of the advantageous compressive stress at the surface, reducing the material toughness. From this point of view, the high temperature thermal process of veneering zirconia with feldspathic ceramic should be taken into account as a possible risk of such a detrimental reverse transformation [43].

The grain size dramatically influence the mechanical behavior of zirconia, in that higher temperatures and longer sintering times produce larger grain sizes [77]. The critical crystal size is approximately 1 μm: above such dimension, zirconia is more prone to spontaneous PTT due to lower stability, whereas a smaller grain size makes zirconia less susceptible to this phenomenon [78], although below 0.2 μm PTT does not happen and zirconia fracture toughness decreases [79]. Consequently, the sintering conditions are paramount since they influence the crystal size, strongly affecting the mechanical properties and the stability of zirconia [80] and have to be strictly controlled in the whole production process.

Although being many types of zirconia-based ceramics available, to date three zirconia-containing systems have been more or less extensively used for dental applications. Two of them are by-phasic materials: the previously mentioned "glass-infiltrated zirconia-toughened alumina" (ZTA) and the "magnesium partially stabilized zirconia" (Mg-PSZ); the third,
most used, is the “yttria partially stabilized tetragonal zirconia polycrystal” (3Y-TZP), a mono-phasic material [43,74].

2.3 Magnesium partially stabilized zirconia (Mg-PSZ)

The microstructure of Mg-PSZ consists of clusters of tetragonal crystals within a cubic stabilized zirconia matrix. The added stabilizer is MgO (8–10 mol%). As regards dental applications, with some exceptions (Denzir-M—Dentronic AB), such a material has not been extensively used, neither has it encountered large popularity due to its remarkable porosity, large grain size (30–60 μm), low stability, tendency to framework wear [73], and overall poor mechanical properties, especially when compared to 3Y-TZP [43,48].

2.4 Yttrium partially stabilized tetragonal zirconia polycrystal (3Y-TZP)

This type of zirconia is made of transformable, t-shaped grains stabilized by the addition of 3 mol% yttrium-oxide (Y₂O₃). Such a polycrystalline material exhibits low porosity and high density [81]; at the moment it is the most popular and frequently used form of zirconia commercially available for dental applications.

2.5 Mechanical properties of zirconia

The mechanical performances of zirconia were extensively investigated on both SCs and 3- and 4-unit FPDs, with variable reported data, due to a noticeable difference of experimental conditions and measurements. Mechanical properties of zirconia were proved to be higher than those of all other ceramics for dental use, with a fracture toughness of 6–10 MPa/m¹/₂, a flexural strength of 900–1200 MPa and a compression resistance of 2000 MPa [43,73,82]. An average load-bearing capacity of 755 N was reported for zirconia restorations [83]. Fracture loads ranging between 706 N [83], 2000 N [84] and 4100 N [76] were reported; all of the studies demonstrated that in dental restorations zirconia yields higher fracture loads than alumina or lithium disilicate. A recent in vitro investigation on zirconia FPDs evidenced failure loads ranging between 379 and 501 MPa, thus higher than average human biting force, confirming a satisfactory serviceability of such frameworks [85].

2.6 “Aging” of zirconia

The low temperature degradation (LTD), or “aging”, of zirconia is a well known process, strictly related to the PTT, of which represents the other side of a same coin: it consists in a spontaneous, slow transformation of the crystals from the tetragonal phase to the stabler monoclinic phase in absence of any mechanical stress. This phenomenon decreases the physical properties of the material and exposes zirconia frameworks at the risk of spontaneous catastrophic failure [86]. Mechanical stresses and wetness accelerate zirconia LTD; other factors affecting such a process are: grain size, temperature, vapor, surface defects of the material, type, percentage and distribution of stabilizing oxides and processing techniques [64,87].

Although LTD is to be considered as a risk factor for mechanical prosthetic failures, to date such a direct relationship has not been demonstrated by scientific evidence in the clinical service [43,88]. Even though the long-term effects of LTD on zirconia in dental restorations have not been completely investigated yet, aging is regarded as likely to induce detrimental changes in the mechanical behavior of the material, like microcracking, strength decrease, enhanced wear rates with release of zirconia grains in the surrounding environment [89,90], as well as surface roughening, with further degradation of mechanical and esthetic properties [43,86,91].

An in vitro research was aimed at comparing failure loads of standardized zirconia 3-unit FPDs before and after exposure to an artificial aging process by means of a mastication simulator, corresponding to a 5-years of clinical function (about 1,200,000 cycles of thermomechanical fatigue in liquid environment). Such a treatment reduced the failure loads of all test samples, with significant differences due to different fabrication techniques for each system, but such a reduction ranged within clinically acceptable values; in fact, all test specimens showed minimum failure loads higher than 1000N, both before and after fatigue loading, thus widely exceeding average masticatory loads [92]. In the same paper, a cautionary warning was addressed to clinicians about the risks of intentionally leaving a Y-TZP framework without ceramic veneering at the level of the gingival side of the FFDs, as suggested by other researchers, in order to enhance the core strength [46]; this would expose the framework zirconia to the intraoral salivary environment, increasing, at the same time, the potential for plaque retention and reducing resistance to low temperature degradation and service life [92]. Such an issue is still debated and controversial; in any case, further investigations will be necessary to elucidate the relationship between aging of zirconia and long-term survival of the products [88].

2.7 Manufacturing procedures

CAD/CAM zirconia dental frameworks can be produced according to two different techniques: “soft machining” of pre-sintered blanks or “hard machining” of fully sintered blanks [43,93,94].

The soft machining process is the most diffused manufacturing system for 3Y-TZP, based on milling of pre-sintered blanks that are fully sintered at a final stage. Such zirconia blanks, at the so-called “green state”, are produced by compacting zirconia powders (in presence of a binder that will be eliminated in the following pre-sinterization step) through a cold, isostatic pressing process; this results in a very narrow pore size (20–30 nm) and quite homogeneous distribution of the components inside the blank [43,80]. Processing at a proper pre-sintering temperature of zirconia is a crucial factor since this parameter affects hardness, machinability and roughness of the blanks. From the manufacturers’ point of view about the choice of the most convenient production technique, hardness and machinability act as opposite factors: an adequate hardness is necessary to manipulate the 3Y-TZP blanks safely, but, if excessive, it is detrimental to a proper machinability. Moreover, higher pre-sintering temperatures create rougher blank surfaces [43,80].

After scanning a stone die of the supporting abutment(s) (or directly the wax pattern of the crown/FPD), a virtual, enlarged framework is designed by sophisticated CAD softwares. Then,
through a CAM milling procedure, a framework with enlarged, accurately controlled dimension is machined out of the blank. At the end, the sinterization is completed at high temperature: the zirconia framework acquires its final mechanical properties in that it undergoes a linear volume shrinkage of about 25%, so regaining its proper dimensions. Such processing is known to produce very stable cores containing a significant amount of tetragonal zirconia with surfaces virtually free from monoclinic phase [43]. Nevertheless, a certain amount of cubic zirconia may be present due to an uneven distribution of yttrium oxide. The cubic phase is richer in stabilizing oxides than the surrounding tetragonal crystals and this may negatively influence the stability of the material [77].

Frameworks can be colored either adding minimum amounts of metal oxides to the zirconia powder or, after machining, by soaking the core in solutions of metal salts (like cerium, bismuth or iron); the framework coloration seems neither to induce PTT nor to decrease the mechanical performance of the restorations [43,95].

Soft-machining is the preferred process by the majority of the manufacturers, like Procera Zirconia (Nobel Biocare AB, Gotteborg, Sweden), Lava (3 M ESPE, Seefeld, Germany) and Cercon (Dentply Degudent, Hanau, Germany).

In the hard machining technique, on the other hand, the 3Y–TZP blocks are previously densely sintered through a process called “hot isostatic pressing”; at high temperatures (1400–1500 °C) and high pressure in inert gas environment, very hard, dense and homogeneous blocks of fully sintered zirconia are produced [96], out of which the frameworks are shaped to the proper, desired form and to the right, final dimension by using powerful and resistant milling machines with diamond abrasives. Hard-machining of Hi-pressed (“HiPed”) Zirconia is utilized by Denzir (Decim AB, Skelleftea, Sweden) and DC-Zirkon (DCS Dental AG, Allschwill, Germany).

The issue of which technique is suitable to get the better outcomes still remains a controversial topic. The major drawback of soft-machining is the problem of matching the sintering shrinkage of the framework to the enlargement amount programmed by the software as precisely as possible [16]. In any case, some in vitro investigations have confirmed high fracture toughness and flexural strength with different production techniques, using both hot and cold isostatic pressed zirconia blanks [45,46].

It is clear that, compared to the soft-machining, the hard-milling procedure is more time consuming and requires cutting devices that have to be very tough and resistant to wear; the fully sintered 3Y-TZP blocks are much harder and less machinable of both fully sintered zirconia and densely sintered alumina blocks, making milling time much longer and the production procedure more expensive [64]. From an operative point of view, moreover, milling zirconia blanks at thin sections is very difficult and can lead to unpredictable results [97]. Finally, it has been demonstrated that grinding such blocks introduces various kinds of surface microcrack and defects [98], both of the “brittle” and the “ductile” type, depending on various factors, such as the grain size of the diamond burs or the rotation speed: fine burs determine a more “ductile” damage compared to the “brittle” fractures due to the coarse ones, while high speed grinding procedure allows to reduce the applied force to the block and minimize the dimension and depth of the surface defects [98]. In any case, there is high level of evidence that all surface treatments creating stress, like grinding, sandblasting or indentations on the zirconia surface determine some degree of (t) > (m) transformation before clinical use [64,99] being detrimental to the long-term serviceability of zirconia restorations [100–102]. Surface grinding can determine deep defects that reduce toughness [99,103], decrease the strength [104,105], and the consequent exposure of the processing flaws to wetness may have further detrimental effects [87]; the resultant alteration of phase integrity is reported to increase the susceptibility of the material to aging [106]. Frameworks produced by hard machining exhibit a considerable amount of monoclinic zirconia, associated with higher susceptibility to LTD and surface microcracking, resulting in a less stable material [77]. In any case, since there is no standardization of the treatments utilized, it is very difficult to compare the results of the studies focused on the surface treatments of zirconia [43]. Soft machining procedures provide predictable stability of the framework, as long as its surface is not damaged after sintering (e.g. by an occlusal adjustment). To date, the surface state after processing is still a controversial issue, particularly after hard machining, although there is wide agreement on the fact that microcracking due to processing flaws or occlusal adjustments is among the main causes of fatigue damage and failure [43].

Residual stress, like that arising when zirconia is fired at high temperature and then rapidly cooled down or when a ceramic material with different coefficient of thermal expansion (CTE) is used for veneering, was found to be a more critical factor than final surface roughness in inducing LTD [106]. Furthermore, the presence of large cubic phases is detrimental to the resistance of zirconia to LTD and aging [77].

Scientific interest has been rising over the last years toward a possible use of cerium oxide as a stabilizer for dental applications at higher concentration than yttria (8 mol%) [107]. Under similar conditions and thermo-cycling, ceria partially stabilized zirconia (Ce–TZP) showed better thermal stability and resistance to LTD than Y–TZP and, furthermore, before the fracture point, it exhibited the highest bending capacity among the ceramic materials [47]. The major drawback that has mainly limited a use of such a material in the dental practice is its basic yellow-brownish color, along with a marked mutability over time, with the tendency to a dark gray shade after exposition to reducing substances (like glucose or lactose) [64]. Further investigation is needed to identify possible future utilization of mixtures between alumina, ceria and yttria to get better and better outcomes [64].

As regards the framework thickness, almost all manufacturers agree in considering 0.5 mm the minimum thickness for copings, in order to prevent core deformation [43,108]. It is a well accepted concept that framework thickness and shape should be optimized and individualized to achieve an even thickness of veneering ceramic [48] as well as a suitable support for it.

CTE compatibility between the zirconia-based framework and the veneering ceramic is a very critical factor from the mechanical point of view. This topic will be treated at a later point.
Another important aspect determining the mechanical properties of zirconia-based FPDs is connector shape and size. In some clinical trials, fractures of zirconia FPDs have been shown to be associated with insufficient connector height [109–112], since such prosthetic component, connecting retainers and pontic(s), represents a locus minoris resistentiae under load [113]. Flexural strength has to be high enough to withstand occlusal loads, since connectors are under applied tensile stress, so the dimensions of connectors are paramount factors for the long-term success of zirconia FPDs; however, they are limited in height by the presence of the periodontal soft tissues [109,113]. Notwithstanding the lack of strong scientific evidence about the ideal connector size, some in vitro analyzes recommended minimum diameters of about 3.0–6.0 mm for 3-unit, 4.0–6.0 mm for 4-unit and 5.0–6.0 mm for 5-unit zirconia FPDs [114] and these are the recommended figures by most manufacturers. Moreover, another in vitro study suggested that the radius of curvature at the gingival embrasure of the connector strongly affects the fracture resistance of all-ceramic FPDs; therefore, the gingival embrasure should have a radius as wide as possible [113].

2.8. Precision of fit

The precision of the zirconia-based restorations is dependent on various factors, like differences in manufacturing systems, individual characteristics of the prosthesis (e.g. span length, framework configuration), effect of veneering and influence of aging [82]. As to soft-machined 3Y-TZP restorations, the precise numerical compensation required by such a system for the enlargement ratio of the model is a paramount factor, strictly dependent also on the composition and homogeneity of pre-sintered zirconia blanks that should be consistent and precise [82].

For SCs, milled dense zirconia copings showed high accuracy of fit, ranging between 0 and 74 μm [45,46,97,103,115,116]. As regards zirconia FPDs, in recent literature various in vitro and in vivo studies investigated the precision of fit of such restorations, although different experimental designs and evaluation procedures made the data comparison very hard to do: for example, some studies did not cement the specimens while others used different luting agents; furthermore, some procedures involved impression taking whereas other investigations employed scanning techniques. In these studies, the absolute marginal gaps ranged between 9.0 and 148.8 μm, with an average value of 73.8 μm. Higher discrepancies were detected at the internal gap (i.e. the internal distance measured between the coping and the abutment), ranging between 68.8 and 215 μm in the occlusal direction and between 52.3 and 192.2 μm in the axial direction [82].

An in vitro investigation [117] reported lower values of absolute marginal opening when a feather-edge was realized during the tooth preparation (± 10 μm), while the detected values were higher for mini-chamfer (114 ± 11 μm), shoulder (114 ± 16 μm) and chamfer finish line (144 ± 14 μm); however, due to the limits of the in vitro experimental design and to the mechanical limitations of feather edge preparations, such a finish line design was not recommended for clinical applications of zirconia SCs by the same authors.

A relationship between the extension of zirconia FPD and the marginal fit has been demonstrated, in that the larger the FPD span, the higher the detected marginal discrepancies [97,118]. Moreover, also the shape of the framework exhibits a relationship to marginal fit: a straight framework design, i.e. with pontics positioned in the straight line with retainers, resulted in more accurate margins than curved framework configurations [119]. Some authors advise that, in case of complex prosthetic geometry, post-sintered, hard machining is the most predictable manufacturing system [82].

As to veneering process, its final effect on the overall fit still remains controversial.

About manufacturing techniques, two studies demonstrated less marginal discrepancies with post-sintered machining [120,121], while another investigation did not record any difference between hard- and soft-milling [122]. Moreover, two studies proved that CAD/CAM systems resulted in lower marginal discrepancies [111,120], whereas other two investigations did not highlight any significant difference between CAD/CAM, in which the design step is performed by software, and CAM only, where such a procedure is realized by scanning both the internal and the external aspects of a wax-modeled physical framework [119,121].

Interestingly, a fatigue load simulation demonstrated that thermal and masticatory stimulation did not influence the marginal fit of zirconia FPDs. At the same time, aging seemed not to affect the long-term stability of zirconia fit and marginal integrity [122].

It can be concluded that most of the currently available zirconia-manufacturing systems provide clinically acceptable marginal and internal gaps; however, remarkable variations were evidenced using different systems and materials. For an extended, detailed analysis of the research data on zirconia-based restorations fit, refer to the review by Abduo et al. [82].

2.9. Clinical and experimental studies on zirconia

As pointed out, in the last decade an increasing interest in the zirconia as a dental material for SCs and FPDs has led to the realization of several clinical trials focused on defining success/survival rates of such restorations. Most of these studies investigated the clinical results with posterior FPDs [111,112,123–139], while only a few were about SCs [140,141] and implant abutments [124,142,143]; only one regarded implant-supported FPDs [132].

The choice of zirconia as a core material for SCs, both in the anterior and in the posterior sites, has been increasing over time with clinical results that seem quite comparable to metal–ceramic single restorations, although clinical trials are very few, to date. Zirconia SCs showed a success rate of 93% after a 2 years observation period, with a favorable soft tissue response, in a limited sample size of 15 Cercon crowns (Dentsply Degudent, Hanau, Germany) [140]. Another investigation with a longer observational period (3 years), performed on 204 Procera zirconia SCs delivered in a private practice, showed a survival rate of 93%; in this study, 16% of complications were recorded (6% loss of retention, 2.5% extraction of abutment teeth, 5% persistent pain, 2% porcelain chipping) [141].
On the contrary, a larger amount of data regarding zirconia posterior FPDs is emerging from various studies that showed quite favorable clinical results; not differently from metal–ceramic bridges, failures have been reported, both related to biologic complications, like secondary caries, and to technical problems, such as fracture of the bridge or chipping of the veneering ceramic [6,124,144]. Comprehensive systematic reviews of the literature on the survival rates of all-ceramic SCs and FPDs in comparison with metal–ceramic restorations have been published [5,6], reporting, after 5 years of observation, favorable survival rates (95.6%) for metal–ceramic prostheses, to be compared to a figure of 93.3% for all-ceramic restorations, among which zirconia-based prostheses showed the best clinical performances and resulted as the most reliable all-ceramic systems. Zirconia was affected only by cracking or chipping of veneering ceramic, whereas other all-ceramic restorations showed some framework fractures. It has been pointed out that it is very difficult, and in some cases impossible, to make a scientifically valid comparison between the results of different clinical studies conducted on the various zirconia-based FPDs, for the diversity in the research methods, evaluation parameters, production techniques and observation periods. After 2 years of function, a recent clinical short-term research reported 100% survival rate [126]. After 3 years of clinical service, almost all of the studies reported very good clinical outcomes for zirconia-based FPDs, with failure rates between 0% and 4.8% [111,124], showing a promising reliability of such restorations [123,130,145]. In a clinical study on 18 teeth, one failure was reported, due to a radicular fracture [145]. At 4 years of use, the reported failure rates were comprised within 4% and 6% [138,139]. To date, in such studies the longest observational period was 5 years [6,128]; 3–5-unit posterior zirconia FPDs on natural teeth were evaluated, with the only exception of a 3-unit FPD in the anterior site, replacing a lateral incisor [128]. After 5 years of clinical service, the overall survival rate ranged between 74% [6] and 100% [125,128].

In conclusion, positive success rates have been recorded in most investigations and a suitable serviceability of zirconia-based FPDs seems to be expected in the medium-term.

The reported mechanical complications related to such restorations are framework fractures, chipping of veneering ceramic and loss of retention.

Zirconia framework fractures were reported in only 5 studies, 3 on FPDs [6,111,112], 1 on SCs [140] and 3 on inlay-retained FPDs [134]. The incidence of core fractures ranged between 3% [112] and 10% [134] and such data seem to indicate a strong relationship to the design of the prostheses, in that the highest incidence was recorded with inlay-retained FPDs. It can be inferred that bulk fractures should be considered very infrequent events; when they occur, connectors of multi-unit FPDs or second molar abutments are mainly interested.

Conversely, crazing or chipping (superficial cohesive fractures) of the veneering porcelain were reported by the majority of the studies as the most frequent complications affecting zirconia-based prostheses, mainly at level of posterior teeth and independently from the type of restoration. According to Sailer et al. [129], after 5 years of clinical service, 3–5-unit posterior zirconia FPDs supported by natural teeth exhibited secondary caries as the most common cause of (biologic) failure, affecting 21.7% of the restorations, whereas porcelain chipping occurred in 15.2% of the prostheses. In a study conducted on implant-supported FPDs, a much higher incidence of porcelain chipping (54% after 1 year of clinical service) was reported [132]. So far, the incidence of chipping reported in the studies in zirconia-based restorations ranges from 0% to 54% after 1 or 2 years of observation [123,127,132,146], in any case less favorable than the figures referred to PFM restorations (gold-based alloys) in the scientific literature: 98% of intact porcelain over 5 years [147] and 4–6% of ceramic-related failures recorded over 10 years [148]. The estimated risk per year of ceramic chipping for all-ceramic restorations is 2.92% [6]. Nonetheless, it is quite arduous to come to conclusions based on the comparison of such data, due to evident differences in machining systems, failure evaluation criteria, follow-up period and clinical experience of the operators. It has also to be highlighted that, in the majority of the cases, chippings did not hinder function, being repairable [123,127,132,137]; sometimes the problem was easily solved by intraorally polishing the restoration surface [112,137], while, in another study, repair was carried out with composite resin [131]. Only in very few cases of major chipping or when serious esthetic problems arose, the restorations needed a total replacement [6,141].

Although minor cohesive fracture of veneering ceramic being the most frequent typology of failure reported in the majority of the cited clinical studies, exposure of the underlying zirconia core was rarely observed and, in any case, it is very hard to detect by the naked eye. Chipping of veneering porcelain occurred also in non-load bearing areas [6,112], even though second molars and connectors of mandibular posterior FPDs were the preferential sites for such mechanical complications, probably due to the more intense biting forces [123,133,141].

The causes of porcelain chipping may be material-related to some extent; on the other hand, factors may be also dependent on the prosthetic design, such as core-porcelain thickness ratio and framework architecture. An incorrect shaping of the framework does not provide adequate, uniform support to the veneering ceramic and this could play a critical role in porcelain chipping. As well as in metal–ceramics, the zirconia-based FPD frameworks should be shaped in order to ensure an optimal support to veneering porcelain, that should be mainly subjected to compressive loads, limiting detrimental tensile stresses. In metal–ceramics, this is conventionally achieved by wax-modeling the complete anatomic contour of the restoration and, later, creating the correct spacing for veneering ceramics by means of the controlled, so-called “cut-back” technique; in this way, the proper ceramic thickness for both an optimal esthetic and mechanical performance can be achieved, avoiding excessive porcelain thickness (>2 mm) [9]. The shape of zirconia frameworks should be customized by dental technicians according to such requisites, rather than modeled according to the concept of uniform thickness of zirconia.

As previously seen, surface damages can represent a starting point for the onset of fractures. Recent fractographic examinations [111,112,124] demonstrated that chipping of the veneering ceramic may origin from occlusal roughness, that can be the consequence of possible incorporation of air bub-
bles during the powder buildup for veneering porcelain or the result of damages originating from occlusal adjustments, likely exposing the underlying zirconia core. Spontaneous delamination, i.e. interfacial adhesive failure separating the zirconia core from the veneering ceramic, also has been considered as a possible modality of failure, but it can only be evidenced by a microscopic examination [134] and, in any case, most authors are quite skeptical about a likely occurrence of delamination, since the bond strength between zirconia and dedicated veneering ceramics is higher than the cohesive force of the porcelain itself [49,50,149–152].

It is clear that ceramic veneer cracking is a multifactorial phenomenon and that only some of its possible causes have been distinctly highlighted: among the others, differences in CTEs between framework and ceramic, firing shrinkage of porcelain, areas of porosities, flaws on veneering, poor wetting by veneering material on core, improper framework support, overloading and fatigue [49,50,153–155]. Even though the nature of the bonding between zirconia cores and veneering porcelain has not been completely clarified and the compatibility parameters have not been definitely characterized yet [151], CTE seems to play a crucial role in such a phenomenon. Dedicated ceramics have been developed for zirconia in order to reduce mechanical problems, but further investigations on the bond between zirconia core and veneering ceramic are needed. Nowadays, almost all of zirconia-based systems offer veneering ceramics specifically developed to exhibit CTEs compatible with zirconia frameworks. In agreement with a principle that has been widely applied by dental technicians in manufacturing metal–ceramic restorations, a little, controlled mismatch of such parameters seems to be advisable, in terms of a lower veneering ceramic CTE than the zirconia CTE, in order to place porcelain under compression and in terms of a lower veneering ceramic CTE than the zirconia CTE, in order to place porcelain under compression and reduce the risk of crack development by increasing the bonding strength to the framework [82,156]. At the same time, tempering residual stresses are to be controlled, to prevent the onset of detrimental tensile forces and to limit porcelain chipping risks; this is obtained by lowering the cooling rate after the final firing or glazing of porcelain [155]. Presumably, other interacting factors can be involved in the compatibility issue between zirconia and veneering ceramics, but not all of them have been sufficiently studied yet: tensile stresses concentrated at the zirconia–ceramic interface [64]; chemical processes, like the harmful dissolution of refractory materials (like zirconia) induced by silicate glasses contained in the veneering ceramics [157]; surface phase changes, as those due to a depletion of stabilizing oxides determined by functional wear or ceramic adjustments, leading to destabilization of the (t)—zirconia [158,159].

Loss of retention has been reported to be another possible technical complication with all kinds of luting agents, particularly in FPDs; anyway, in the zirconia clinical trials, all of the debonded restorations were eventually recemented successfully [96]. The estimated risk per year of loss of retention for all-ceramic prostheses has been reported to be around 0.47% [6]. Differently, catastrophic, not repairable fractures probably caused by debonding of the restorations were observed with some inlay-retained FPDs cemented with resin luting agents [134].

None of the cited clinical trials took bruxism into account, more often such a parafunction figuring among the exclusion criteria: consequently, since they were not considered in any clinical investigation, parafunctional activities should be regarded as a potential limitation for zirconia-based restorations [96].

Although it has been demonstrated that, among the metal-free restorative materials, zirconia exhibits the highest mechanical properties, from a clinical standpoint, zirconia bridges maximum span remains controversial: 5-unit is the maximum serviceable extension for zirconia FPDs supported by a piece of scientific evidence [132]. As regards more extended prostheses, further investigations with longer observational times are needed to draw solid guidelines, even though some manufacturers suggest and support the technology for fabricating full-arch restorations.

It has to be noticed that in a few of the analyzed clinical trials, some restorations presented cantilevers; notwithstanding none of them was affected by framework failure, the authors themselves did not recommend the cantilevered design for zirconia FPDs, particularly in distal segments [134].

According to in vitro cyclic fatigue tests, Y-TZP was reported to have a lifetime revision comparable to that of metal–ceramic prostheses, with a predicted serviceability longer than 20 years [114]. Nevertheless, two main drawbacks were noticed to be more frequent for zirconia in comparison to metal–ceramic restorations: chipping of veneering ceramic and accelerated aging, so further clinical studies with longer observational period will be necessary to thoroughly investigate the clinical behavior and reliability of zirconia in the long term.

As a last note, it has to be taken into account that, in the last years, a conspicuous interest in the clinical utilization of zirconia in implant-borne restoration has been growing. Most of the implant manufacturer’s brands provide customized zirconia abutments for their platforms, either produced by CAD/CAM techniques or milled from fully sintered blocks. The few published studies show promising results and favorable hard and soft tissue responses after 4 years of service; no abutment fracture was reported but two screw loosenings were observed [142]. Another study recorded 100% survival rate of zirconia single implant abutments after 3 years of follow-up [143]. However, as stated in a recent consensus report [160], the mean follow-up time of ceramic abutments in the available clinical reports is 3.7 years, while for metal abutments is 4.8 years; so far, data are not sufficient to define the indications and performance limits for such abutments. Similarly, as regards the application of CAD/CAM technologies for fabricating implant-supported all-ceramic restorations, especially FPDs or large fixed, full-arch prostheses, available data are not sufficient to compare at a convenient level of evidence such production systems with the conventional ones.

3. Summary

The trend toward an increasingly extended use of all-ceramic SCs and FPDs is an undeniable reality in Fixed Prosthdon-
tics. After the development era, dental ceramics introduced in the last 20 years exhibit different, favorable and promising esthetic and mechanical properties. At the moment, there is no one ceramic material that equally excels in all of these characteristics. The choice of one specific typology of ceramic, rather than on the latest fashion, should be based on a careful evaluation of the very advantages and disadvantages of the material related to the specific dental application, always referring to clinical data with a proper level of scientific evidence and paying attention to the real esthetic needs of the patient.

Further investigations regarding bonding to veneering ceramic, cementation procedures, aging and wear and, above all, long-term clinical performance of zirconia will be needed to define potential and limitations of such an innovative, promising and intriguing restorative material.

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