Zirconia-reinforced dental restorations
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CHAPTER 1

Introduction
1.1 Introduction

Dental restoration is a broad term for any material or prosthesis that restores or replaces lost tooth structure, teeth or oral tissues [1]. Dental restorations are either fixed or removable and termed as ‘fixed restorations’ or ‘removable dental prosthesis (RDP)’. Fixed restorations can be further subdivided into ‘crowns’ (surrounding the remaining tooth structure) and fixed dental prosthesis (FDP), formerly fixed partial denture (FPD), also referred to as ‘bridges’. A FDP is a common method to replace a short span of missing teeth, using adjacent teeth or implants as their supporting abutments [2, 3].

From a review of Zitzmann in 2007, approximately half of the adult population (≥ 16 years) in most European countries have had some type of prosthetic dental restoration [3]. The frequency of fixed restorations including crowns and FDPs (bridges) ranged from 3.6% (in Denmark) to as high as 34% in Switzerland and 45% in Sweden. If just a few teeth are absent, these are generally replaced with short span FDPs because of their convenience and better performance than RDP. The prevalence of fixed restorations is influenced by age, socio-economic status, education and incomes [3].

![Figure 1.1](image_url)

**Figure 1.1** Prevalence of the different dental restorations in different EU countries of patients between 15—74 years old and studies from 1991—2002. RPD = removable partial denture and CD = complete denture. (Data adapted from Zitzmann et al., 2007 [3])

The major purpose of dental restorations is to maintain oral function, by replacing missing teeth and contiguous tissues with artificial substitutes. Meanwhile,
the comfort, appearance and health of the patient should also be considered during the replacement. In addition, restorations are also indicated to treat teeth with severely damaged clinical crowns, or when aesthetic alterations in tooth colour and/or form and contour are required [3].

The primary trigger for reconstructive treatment is normally the patients’ perception of whether or not they need treatment. Generally, this is based on how great the impact of any functional, aesthetic or psychological impairment is to them [3]. However, a dentist should also assess the balance between improvement and harm by inserting restorations made from alloplastic materials. In this way, a dentist should focus on the consequences of a reduction of oral tissues as well as the outcomes of modifying such conditions, instead of emphasizing only the technical aspects of the discipline. As a result, biocompatibility, aesthetics, reliability, longevity and stability are central to the search for optimal dental restorative materials [2, 4]. The fabrication of fixed dental restorations is also developing with the increasing demand for aesthetics from both dentists and the patients.

1.2 History of fabrication of FDP

Traditionally, a dental restoration is usually fabricated in dental labs by dental technicians or in the dental clinic by the dentist himself. Since the restoration is mainly manually fabricated, the quality of the dental restoration is largely influenced by the skills of dental technicians or dentists, making it technique sensitive. There are several representative materials or techniques in the history of fabrication of dental restorations.

In modern dentistry, the first widely used material to fabricate fixed restorations are the casting metals. Casting was introduced in dentistry in the 1890s. In the literature, credit is given to Dr. Swasey (1890), who introduced the technique to make a solid casting gold inlay. He used a gold foil to adapt to the inlay cavity form and contour of the tooth, and then the foil was removed, invested, and filled with 20-karat gold, resulting in a solid gold inlay. Although casting metals by the lost-wax process had been widely used in industry and art for many years, wax was introduced in dentistry for the first time in 1891 by Martin, as to make gold inlays. However, it is William H. Taggart who described a lost wax technique which can truly be said to have revolutionized restorative and prosthetic dentistry. On January 15th, 1907, Dr. Taggart presented a paper entitled ‘A New and Accurate Method of Casting Gold
Inlays’ for the New York Odontological Society, in which he discussed his casting technique and machine. Taggart’s success was mostly due to his improved casting machine, since his casting technique was not original; the idea of using wax to form the pattern was from Martin (1891), and using pressure to cast the alloy was that of Philbrook (1896) [5, 6].

On account of the expensive price of gold, in the late 1920s, cobalt-chromium alloys were first introduced for casting dentures by the lost wax method, followed by a series of investigations on nonprecious or base metal alloys, which do not contain gold, silver, platinum or palladium [5, 7]. Nickel and cobalt are the primary metals in most commercially available nonprecious alloys, with chromium being the next most predominant metal, since chromium increases nickel or cobalt alloys’ resistance to oxidation and also enhances solid solution hardening [7]. The proponents of nonprecious alloys suggested that these alloys would be cheaper and that dentures made of these materials would have better physical properties (in particular greater strength) than those made of gold alloys [8]. Although increases in the cost of gold have made cobalt-chromium alloys more formidable rivals for denture work amongst dental practitioners and patients, gold alloys are still the materials of choice for metal-based inlays, crowns and FDPs [5].

Metal Ceramics

Despite the development of dental castings between 1920s and 1950s [6, 8], the full-metal-cast restorations were getting less and less popular since the 1960s. This is because of the increasing demand for aesthetic solutions in dentistry [6, 9]. Aesthetics is a subjective judgment, and some individuals, including many dentists, still considered even in the 1990s visible gold restorations as aesthetically acceptable [9], at least in posterior teeth. However, silver, palladium, base-metal alloys, including titanium, were considered unaesthetic by most people [9].

In the 1960s, researchers were looking for further improvements in fixed restorations and their main concern was to improve aesthetics. This time credit is usually given to Dr. Weinstein, who was the first to produce a commercially successful dental porcelain fused gold alloy, and who obtained two U.S. patents ‘Fused Porcelain to Metal Teeth’ [10] and ‘Porcelain Covered Metal-Reinforced Teeth’ [11]. This was the first time that the metal ceramic crown was introduced in dentistry.
The metal ceramic restoration, which was also mentioned as the porcelain fused to metal (PFM) restoration, has a metal core or frame, which is veneered with porcelain in a high temperature oven. The metal provides strong compressive and tensile strength, and the porcelain gives a white tooth-like appearance. Porcelains for metal ceramic restorations contain a vitreous phase, also called glassy matrix, which represents 75 to 85% by volume and are reinforced by various crystalline phases [12]. Most porcelains for metal ceramic restorations contain from 15 to 25 vol.% leucite as their major crystalline phase [13, 14]. Leucite is a potassium alumino-silicate with a high thermal expansion coefficient (TEC) [15]. The choice of the crystalline phase in porcelain for metal ceramic restorations was initially dictated by the need for matching the TEC. The TEC of the porcelain should be close to that of the metallic infrastructure, in order to avoid the development of tensile stresses within the porcelain during cooling.
Different metal alloys, including precious and base metals, can be used as the frames of metal ceramic restorations, in order to meet the mechanical and economic requirements. Meanwhile, the colour of metal ceramic restorations, depends on the colour of the veneering porcelain, and can be adjusted to match the adjacent teeth. Metal ceramic restorations became more and more popular, as they have good mechanical properties, wide economic variety and to somewhat satisfactory aesthetic results, along with a clinically acceptable quality of their marginal and internal adaptation [16-20]. Furthermore, the metal ceramic restoration have been proven to have favourable long-term prospects [2, 21], and has represented the ‘gold standard’ for years in prosthetic dentistry [22]. By the meta-analysis, the 5-year survival rate of metal ceramic restoration is 95.6% for single crown [19, 23], and is 94.4% for FDPs [18, 24].

Nevertheless, the technical procedures of investing wax patterns and casting precious metal alloys are technique sensitive. Many technical variables and different operative steps and firing cycles also make the final quality of the restorations highly technique-sensitive [22]. The use of metals as core also requires the use of masking porcelains. The underlying metal greyish shade of these frameworks is likely to introduce a significant limitation for the aesthetic result due to the absence of translucency. The limitation of these metal ceramic restorations is that they can only absorb or reflect light, while dental tissues are kind of translucent [25].

Figure 1.3  Picture of a PFM and full ceramic (zirconia) FDP.

Nowadays, there is an increased concern about possible health side-effect of dental restorations. Especially attention is paid to the toxic and allergic reaction mainly
caused by corrosion of base-metal [7, 26-29], or even some precious alloys like for example palladium [30, 31]. As long as the demand for biocompatibility increases, metal-free materials are becoming more and more popular.

**Full Ceramics**

Full ceramics or all ceramics are one of the most popular metal-free and tooth-coloured or aesthetic restorative materials. The first time porcelain emerging in dentistry was in 1774, being used to make a complete denture [21]. Early in the nineteenth century, natural human teeth were replaced by porcelain in various forms [5, 22]. However, the all ceramics were not popular at that time, because they were too brittle. When McLean and Hughes described in 1965 a reinforcement of dental porcelain with ceramic oxides [32], stronger and more reliable dental porcelains were invented. Due to increasing concerns about the aesthetics and biocompatibility, the use of all-ceramic restorations has become popular since three decades ago [2].

Dental ceramics usually have a composite structure, always containing a glassy matrix, which is reinforced by various crystalline phases [12]. The nature, amount, and particle size distribution of the crystalline phase directly influences the mechanical and optical properties of the material [33, 34]. The match between the refractive indices of the crystalline phase and glassy matrix is a key factor for controlling the translucency of the porcelain. Similarly, the match between the thermal expansion coefficients of the crystalline phase and glassy matrix is critical in controlling residual thermal stresses within the porcelain [35].

Today, the best aesthetic characteristics are still based on the glass-ceramics. Glass-ceramics have optimal light transmission and high translucency. This results in natural, tooth-like colours which are also in the presence of very light shades [22]. For example, feldspathic ceramics usually provide excellent aesthetic, together with very good biocompatibility. They have also good resistance to compressive forces, but they easily fracture under tensile load [22]. At present, the most popular leucite-reinforced glass-ceramic system is the IPS Empress (Ivoclar Vivadent, Liechtenstein), which is a pressable system. Similar to the lost-wax casting technique, the pressure-injected leucite-reinforced ceramic can be hot-pressed a hollow investment mould by using a pressable-porcelain-oven. The restoration can be then stained and glazed or cut-back and layered with feldspathic ceramic to match the patients’ teeth colour and shape. Due to their low flexural strength (100 – 120 MPa), leucite-reinforced ceramics are only indicated in the anterior region.
Developments in dental ceramics then have led to the introduction of new high-strength ceramic core materials for all-ceramic FDPs [36]. Glass-infiltrated alumina was introduced as a dental framework (core) in 1989. The most popular example of this ceramic is the In-Ceram Alumina system (VITA Zahnfabrik, H. Rauter GmbH & Co. KG, Bad Säckingen, Germany), which was the first restorative system introduced for the fabrication of 3-unit anterior FDPs. The flexural strength of the framework material ranges from 236 to 600 MPa [22, 36]. Later on, a lithium-disilicate glass core material was available on market, known as the Empress 2 or the enhanced IPS e.max Press system (Ivoclar Vivadent, Liechtenstein). The framework can also be fabricated with the lost-wax and heat-pressure technique, demonstrating a flexural strength of 300–400 MPa [22, 36]. It can be veneered with fluoroapatite-based ceramics, and still has a very appealing translucency [22]. Lithium disilicate glass-ceramics have been advised for clinical application in single crowns and 3-units FDPs in the anterior region.

The glass-infiltrated alumina reinforced with 35% partially stabilized zirconia was introduced to the dental CAD/CAM system, known as the In-Ceram Zirconia system (VITA, Germany). Such a composition shows a high flexure strength ranging from 400 to 800 MPa [22]. However, it is reported that the In-Ceram Zirconia core is as opaque as a metal-alloy, in terms of translucency. Therefore, In-Ceram Zirconia is not recommended for fabricating anterior all-ceramic restorations, where the translucency of the all-ceramic core materials is a major factor in enhancing an aesthetic result [22, 36]. According to the manufacturer’s recommendations, it is suitable for 3-unit FDP frameworks in the posterior sites.

The zirconia applied in dentistry is usually stabilized with the addition of yttrium oxide. It is generally known as yttria-stabilized zirconia (YSZ) or yttrium tetragonal zirconia polycrystals (Y-TZP). Y-TZP based materials were initially introduced for biomedical use in orthopaedics for hip prostheses. Until the early 1990s, the indication for Y-TZP expanded to endodontic dowels and implant abutments. Nowadays, Y-TZP is usually the core or framework material. The flexural strength of Y-TZP is quite high, reported from 850 to 1200 MPa which is similar to steel [23, 36].

Depending on the composition of all ceramics, their clinical survival rates varied largely. Survival rates for anterior restorations ranged from 94% to 100% for 3–7 years, which are always quite good. However, the survival rate for the posterior FDPs was always lower as 70% to 88% for 5–10 years, mainly due to chipping of surface porcelain or connector fractures [22-24].
Resin Composites

Due to the increased demand on metal-free materials explained before, resin composites also became of interest. Yet the application of resin composites have expanded continuously from small anterior restorations to large posterior restorations [37].

The first polymeric tooth-coloured restorative materials applied in dentistry were based on polymethylmethacrylate (PMMA). Resin-based materials entered the dental market in the 1930s [38]. These materials were more aesthetic than amalgam restorations, but still suffered variety of problems. The main problems of the early PMMA restorations were poor colour stability, high polymerization shrinkage, a lack of adhesion to tooth structure, and a large coefficient of thermal expansion [38]. In the 1950s the resin materials became resin composites due to addition of silica fillers to the matrix of resins of the 1930s. This improved mechanical properties and aesthetics. However most problems remained, e.g. a lack of bonding to tooth structure and still significant polymerization shrinkage [39].

With the introduction of the silica fillers, a new problem occurred. As the silica filler particles were not bonded to the polymer matrix, the filler particles easily dislodged from the matrix and significant clinical wear was observed. This problem was solved by the incorporation of a coupling agent. The most commonly added coupling agent is organosilane, such as γ-methacryloxypropyl trimethoxysilane or vinyl triethoxysilane. The coupling agent can provide a covalent bond (−Si−O−Si−) on the filler surface, and can also form covalent bonds with the resin matrix during polymerization. Thereby the filler particles and the resin matrix are bonded together through the coupling agent, so that the more flexible polymer matrix can transfer stresses to the more rigid filler particles, resulting in improved mechanical properties and wear resistance [39].

The polymerization shrinkage was reduced in a new formulation stimulated by Bowen in 1962, which increased the popularity of resin composites. The basic idea was to decrease the polymerization shrinkage by using high molecular weight monomers instead of the small methylmethacrylate monomers. The resulting monomer is presently well-known as Bisphenol A-glycidyl methacrylate or Bis-GMA. The negative effect of using high molecular monomers is that the viscosity of these monomers is so high that it is difficult to incorporate reasonable amounts of filler particles in the composite. To overcome this problem, triethylene glycol dimethacrylate (TEGDMA), a low viscosity monomer was added. At present, the combination of Bis-GMA and TEGDMA is still the most popular monomer.
combination for dental resin composites [40]. Other composite formulations with different low viscosity monomers have also been prepared with, for example, urethane dimethacrylate resin (UDMA), methyl methacrylate (MMA), or ethylene glycol dimethacrylate (EDMA). When polymerized, the resin matrix forms covalent bonds between polymer chains, known as cross-linking. The cross-linking process improved the properties of the matrix phase, resulting in better mechanical and physical properties of resin composites [39].

Since the commercial introduction of resin composite for restoring anterior teeth in the early 1960s [40, 41], their characteristics have been improved remarkably. During the last 25 years, resin composites have become stronger and more resistant to wear, making it possible not only as a filling material or a temporary restorative material, but also to make permanent crowns and short dental FDPs [42]. However, resin composites still have several shortcomings, such as water uptake, polymerization shrinkage resulting in accompanying stress, lack of wear resistance and colour stability, as well as mechanical, chemical and biological degradation [39, 43-50].

Reinforced Resin Composites

With the rapid development in ceramics, the resin composites also developed rapidly in recent decades, mainly in its reinforce mechanisms. One remarkable advance is the development in the filler particles, given that the filler content and filler size have been shown important to determine and regulate the physical and mechanical properties, such as strength, modulus, and colour [51].

The initially ground composite fillers were limited in size ranging from 0.1 to 100 μm. This was due to the inability to grind and sieve quartz, glass, borosilicate or ceramic particles [52]. Then, smaller particles have been prepared through hydrolysis or precipitation in a process which is also called fumed or pyrolitic silica. This can result in formation of ‘micro-fillers’ with particle sizes in the range of nano particles, e.g. from 0.06 to 0.1 μm. New and alternative routes are sol-gel processes, which use precursors that agglomerate to form particles also in the range from nano to micrometers [53]. In principle, according to manufacturers, nano-composites have, due to their enormous surface to volume ratio, the potential to exhibit excellent mechanical and physical properties.

Zirconia particles currently find use, in combination with silica, as fillers in dental resin composites in order to improve the mechanical properties of cured resin, especially when the zirconia-silica fillers were silanated on the silica parts [54-56]. It showed that polymerization shrinkage of resin composites can be decreased by
addition of metastable tetragonal zirconia filler particles, which expand upon exposure to water molecules [57]. Besides, the concentration of filler particles will also influence the physical and mechanical properties of resin composites [51, 56].

Resin composites can be further reinforced with all kinds of fibers or nano-fibers [58]. Elastic properties and flexural strengths of fiber-reinforced resin composites (FRCs) differ widely, depending on the fiber type (carbon, glass, or high density polyethylene), fiber design (single fiber bundles, woven fibers, or braided fibers) and to a lesser extent the resin chemistry [59]. Some FRC materials are provided dry, to be resin-impregnated during prosthesis fabrication, and many are provided as pre-impregnated fiber bundles. Fabrication differences have very definitely influence to clinical behaviour for FRC- FDPs [60], experienced skills are always required to achieve a good clinical performance.

The fibers inside the composites can improve their fracture resistance to withstand higher oral forces, so that they are gradually being adopted for use in long-term restorations, such as FDPs [60-70]. Because of diversity in composite and fiber types and their fabrication techniques, clinical survival rates of such FRC restorations differed a lot from 72% to 95% for at least 3 years. Veneer debonding or delaminating and fractures are the main reasons for decreased survival rates [60].

1.3 Industrially Manufactured Materials

A number of industrially manufactured materials are available for CAD/CAM systems, including aesthetic and high-strength ceramics, resin composites. These CAD/CAM materials can be used for temporary and permanent restorations. For example, the most recent materials for CAD/CAM chairside temporary restorations are acrylic temporary materials and the resin nano ceramics can be used a permanent material and offer a glossy surface finish. Resin based CAD/CAM materials, like Paradigm MZ100 (3M ESPE), were introduced in the year of 2000 [71]. It was reported that the CAD/CAM resin composites may show less crack propagation under fatigue forces than some ceramics [72]. They may even provide better fracture resistance for non-retentive occlusal veneers in posterior teeth than some ceramics [73]. Lava Ultimate CAD/CAM Restorative (3M ESPE) is another new material for CAD/CAM technique. As introduced by its manufacturer, this new material is called Resin Nano Ceramic (RNC), which is claimed to be unique in durability and function. However, from the view of dental materials, this material is still one kind of zirconia reinforced resin composite, which is supposed to combine the properties of composite
and ceramic. Like a glass ceramic, the material has excellent polish retention for lasting aesthetics. Different from ceramic, the material is not brittle and is fracture resistant. Due to its recent introduction to the market, few studies have examined factors that influence its properties.

Furthermore, there are industrially fabricated feldspathic glass-ceramic materials which are marketed as aesthetic ceramics. Examples are Vivablocs Mark II (Vident, Canada) and CEREC Blocs (Sirona, Germany). There are also industrially fabricated leucite-reinforced glass-ceramic materials, such as IPS Empress CAD (Ivoclar Vivadent). These aesthetic ceramics are indicated for anterior restorations because of their excellent optical properties like colour and translucency, while on the other side they are not as weak as glass porcelain [71]. Besides the pressable lithium disilicate, it is also available for the CAD/CAM technique, known as the IPS e.max CAD (Ivoclar Vivadent, Liechtenstein), which was introduced in 2006. The flexural strength values were reported from 360 to 400 MPa [71], which are two to three times stronger than previous mentioned aesthetic ceramics. The increased strength enables it to apply either etch and adhesively cementation or conventional cementation technique. Lithium disilicate was initially developed as a substructure material that offered greater translucency compared with other high-strength ceramic core materials. Lithium disilicate is now also frequently selected in chairside CAD/CAM systems due to its strength, while its machinability is still reliable. A wide variety of shades and translucencies are available to cover the whole range of tooth colours [71].

1.4 Bonding

The application of all-ceramics as a restorative material has increased substantially in the last three decades. Among other things, this trend can be attributed to the development of ‘adhesive prosthodontics’, or the ability to effectively bond metal-free ceramic restorations to tooth structure using adhesive techniques and cements. An example of the use of the adhesive techniques are veneers where a strong adhesive bond between porcelain and enamel appeared successful over a period of at least 10 years [2].

However, long-term durable adhesion to zirconia ceramics is reported to be difficult [74, 75], because of its surface stability and lack of glass matrix that makes it ‘un-etchable’, as abrasion with alumina is also not very effective. Compared to its popularity due to biocompatibility, acceptable aesthetics and good mechanical properties, the clinical success of zirconia based all-ceramics will be compromised by
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a relatively bad bond strength. As zirconia is a monolithic ceramic not containing a glass phase, conventional hydrofluoric acid etching does not lead to changes of the surface roughness of zirconia, nor does it enhance the bond strength [76, 77]. Without surface pretreatment of silica coating, a single silane coupling agent will not improve the bond strength to zirconia [74, 75, 78, 79], as it lacks the required silica phase for the chemical reaction to form a siloxane network. Although the real function is still not clear, MDP (10-methacryloyloxydecyl dihydrogen phosphate) is assumed to have a chemical reaction to zirconia surface, and can improve the initial bond strength of resin cement to zirconia ceramics. However, this bond strength will always be compromised by artificial aging, such as long term storage in water, indicating the high bond strength based on MDP is not stable in water in the long turn [80-83].

Figure 1.4  SEM image from untreated zirconia (× 12,000) and sandblasted zirconia (× 2000).

To roughen the zirconia surface by the pretreatment of airborne-particle abrasion is still suggested, since it could increase the surface bonding area and possible micro mechanical interlocking [75, 83-86]. However, it can also produce surface defects, which will result in a reduced strength of the zirconia restorations [87]. Moreover, it could also affect the long-term performance of zirconia ceramics [88], due to the surface flaws and the tetragonal to monoclinic phase transformation [87, 89]. Therefore, alternative surface modification methods are required for improving the resin bond strength to zirconia ceramics. Several alternatives to roughen zirconia are reported in literature, such as SIE (selective infiltration etching) technique [90], plasma spraying [79], coating with nano-structured alumina (Al₂O₃) [91], coating with zirconia ceramic powder and pore former [92, 93], coating with silica-based ceramic [75, 79, 85, 94], and coating with silica-like seed layers [95].

Among these techniques, coating with a silica-based layer seems to be easy and more reliable, since the bonding of resin cement to zirconia ceramics may be further
improved by silanization [94, 96]. However, according to previous reports, there are still several weaknesses of the existing silica coating methods. For example, the coating layer could be too thick, hence influencing the clinical fit [96]; the zirconia surface may not be fully covered by the coating layer [79] or the coating layer may not be firmly attached to the zirconia surface [75, 82]. Most of these techniques were still based on air-abrasion [75, 85, 94] which may cause micro defects when blasting with high pressure.

1.5 Failure of restorations

With the development of adhesive dentistry, most of the restorations nowadays are bonded to the tooth structure. As a result, adhesive failures may lead to loss of restorations. Debonding is one of the most common reasons for failure of zirconia restorations [74, 75]. It is also reported that the failure rate due to debonding of zirconia restorations is higher than that of traditional metal ceramics or other all-ceramic restorations [18, 97, 98].

Since bonding is joining two materials together securely with an adhesive substance such as cement or glue, debonding is related to the joining materials, the adhesive substance and the interfaces. There are three modes of failure: adhesive failure, cohesive failure and the mixed failure. An adhesive failure occurs at an interface between two materials due to a tensile or shearing force. A cohesive failure occurs within a material which can be either the joining materials or the adhesive substance. A mixed failure incorporates the above two types of bond failure.

However, there is also some kind of bonding without the use of an adhesive substance, for example, the bonding between the porcelain veneers and the metal or ceramic frameworks. It is reported that the most common failure mode of both metal-ceramic and all-ceramic restorations are delamination, minor or large porcelain fractures [24, 99-101]. The cause of these phenomena is complex. Besides the inherent brittle property, the poor bond strength between the framework and the veneer caused by the coefficients of thermal expansion mismatch can also play a role. Conventional wisdom from metal-ceramic restorations tells us that veneering ceramics should have a slightly lower coefficient of thermal expansion compared with that of the core, creating compressive stresses in the weaker veneering ceramic and thus enhancing the overall strength of the restoration [102]. Unfortunately, application of this principle to all-ceramics has not been equally successful, evidenced by high rates of veneer chipping in all-ceramic bilayered restorations [23].
Ceramics, especially the ones used as veneer porcelain, are brittle materials which are sensitive to flaws during fabricating and functioning. Fractures or chipping can easily initiate from these flaws on different sites, such as the surface, interfaces, or even within the material. Remarkably, in dental ceramics, the first crack usually seems not to be the one that propagates and ultimately causes the restoration to fail (here defined as requiring clinical replacement). Instead, other cracks developing from other sites may grow and ultimately lead to failure. In glasses (like feldspathic porcelains) and alumina, this failure mode is usually a radial fracture, initiating from the cementation interface (tensile zone) and propagating through the entire restoration, leading to bulk fracture. In zirconia, radial fractures rarely occur; instead of a radial crack, the second crack that develops could be an inner cone crack beneath the indenter or cusp [23].

Bulk fracture can also be the main failure mode of a resin composite restoration, while the delaminating between the fibers and the composites can be the most common failure mode for the fiber reinforced composites [61, 103-106]. The connectors of fixed partial prostheses are the most common site to fracture [107-109], and sometimes fractures could also start at the loading point [61, 106, 110].

However, in dental research the failure of restorations can be defined differently [107, 111, 112]. No matter how the failure is defined, it is always related to a stress concentration which is higher than the local strength of the material. Since the excessive stress can cause permanent deformation or local cracks, which grow and finally can lead to a mechanical failure, a complete understanding of the stress distribution in dental restorations is of particular importance. To precisely calculate the stress distribution, finite element analysis (FEA) is usually applied, and it is regarded as the most representative quantitative analysis to model dental restorations [113, 114]. Once the fundamental finite element model has been validated appropriately, it is conveniently to change relative parameters of the model, such as geometries, materials properties and boundary conditions [113].

1.6 Aim of this thesis

Due to the increasing concern about the biocompatibility and the need for better aesthetics, metal-free restorations have become more and more welcomed in dentistry. At present, all-ceramic restorations and the reinforced resin composites are the most popular aesthetic dental materials for indirect application. Among these materials the zirconia ceramics are quite outstanding, because of their appealing mechanical
properties. However, due to their opacity, the aesthetic properties of zirconia ceramics are not so good, and a veneer layer of an aesthetic porcelain is used to optimize the colour of the restoration. Failure occurs mainly within the aesthetic porcelain of the restoration, which can subsequently result in failure of the framework, being one of the most common reasons for clinical failures. On the other hand, debonding is the other common reason for its clinical failure, due to its inert stable chemical properties.

The aim of this thesis is to investigate ways to improve the clinical performance of zirconia and to look for substitutes for the conventional zirconia-veneer all-ceramics. To be specific, in Chapter 2, the sintering procedure of zirconia is investigated. There is little knowledge on the temperature profile required to sinter zirconia. In order to develop new pretreatment procedures for enhanced bonding to zirconia a profound knowledge of the sintering procedure is needed. Therefore, the influence of sintering time and the final temperature on the density, grain size, and biaxial flexural strength of zirconia is investigated.

Alternative surface modification methods are still required for improving the resin bond strength to zirconia ceramics. Several alternatives are reported in literature, such as SIE (selective infiltration etching) technique, plasma spraying, coating with nano-structured alumina (Al$_2$O$_3$), coating with zirconia ceramic powder and pore former, different ways of coating with silica-based. Most of these procedures are complex or involve hazardous materials like HF. In Chapter 3 an attempt is made to improve the bond strength to zirconia, by applying an experimental zirconia-silica coating. The coating is a resin-based zirconia-silica composite which is easy to apply and can in principle be sintered to a zirconia surface. This new method is described and investigated under different priming conditions of resin cement.

By improving the zirconia-resin composite interface combination with new materials becomes feasible. This would allow new ways to make crowns and FDPs on strong zirconia cores and frameworks and the application of more flexible resin composites as veneer. This might be a solution especially in implant supported crowns and FDPs. This type of restoration is nowadays completely rigid, e.g. titanium with PFM or full ceramic, which can result in over-loading of the supporting bone. By substituting the porcelain veneer in these constructions by composite, which is more flexible, a ‘shock absorbing material’ is included, partially taking over the role of the periodontal membrane. This was investigated in Chapter 4, where the conventional veneer porcelain was replaced by resin composite on a zirconia framework. This combination was compared to metal ceramic, all-ceramic, and FRC prosthesis. This was done by in-vitro experiment and FEA calculations.
Recently, Resin Nano Ceramic (RNC) was introduced in the market. This material is an intermediate material between full-zirconia and the existing zirconia based resin composite. RNC is especially indicated for implant supported crown and FDPs as ‘shock absorbing material’, as well as normal crowns and small FDPs. As there is little knowledge of this type of materials, we investigated minimum thickness of this material which was needed when making crowns and compared these findings with lithium-disilicate ceramics. The in-vitro experiment and the FEA calculations are presented in Chapter 5.
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1.7 Reference

Introduction


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