CHAPTER II

LITERATURE REVIEW

2.1 Historical of all ceramic restorations

Metal ceramic restorations are still the most widely used type of ceramic system. Metal ceramic restorations combine the natural esthetics of a brittle material such as porcelain, with the durability and marginal fit of the metal casting. Bonding porcelain to metal supports the porcelain with a metal substructure, increases the strength of the porcelain by reducing ceramic surface defects at the porcelain-metal interface, and possibly places the external surface of porcelain in compression. Although metal ceramic restorations are universally accepted, there are disadvantages such as occasional failures of the veneer. In addition, the metallic coping can distort as a result of porcelain application [17].

However, the demands for more esthetic materials with biocompatible properties is increasing and the rapid development of all ceramic restoration that do not require a metal substructure. Therefore, all ceramic restorations are considered as an alternative of high importance and clinical value. All ceramic restorations without metal have great advantages in their biocompatibility and esthetic aspects is to create a translucent restoration that mimics the appearance of the natural tooth. Ceramic materials are often the restorative of choice because they can replicate the color and shape of the natural dentition.

Dental ceramics are nonmetallic, inorganic materials that consist of a compound of metals (aluminium, calcium, lithium, magnesium, potassium, sodium, tin, titanium, and zirconium) and nonmetals (silicon, boron, fluorine, and oxygen) that may be used as a single structural component, such as when used for a CAD-CAM inlay, or as one of several layers used for the fabrication of a ceramic-based restoration. Conventional dental porcelain is a vitreous ceramic based on a silica (SiO_2) network and potash feldspar ($K_2O{\cdot}Al_2O_3{\cdot}6SiO_2$), soda feldspar ($Na_2O{\cdot}Al_2O_3{\cdot}6SiO_2$) or both. Pigments, opacifiers and glasses are added to control the fusion temperature, sintering temperature, thermal contraction coefficient, and solubility. The feldspars used for dental porcelains are relatively pure and colorless. Therefore, pigments must be added to produce the hues of natural teeth. Most of the ceramics are characterized by their refractory nature, hardness, and chemical inertness. A hardness of a ceramic similar to that of enamel is desirable to minimize the wear of resulting ceramic restorations, and reduce the wear damage that can be produced on enamel by the ceramic restoration. Chemical inertness ensures that the surface of dental restorations does not release potentially harmful elements, and reduces the risk for surface roughening and an increased susceptibility to bacterial adhesion to insure excellent biocompatibility over time. Furthermore, ceramics demonstrate excellent insulating properties, such as low thermal conductivity, low thermal diffusivity, and low electrical conductivity. Their most attractive property is their potential for matching the appearance of natural teeth, offering great esthetic results [18].

On the other hand, the susceptibility of ceramics to brittle fracture is a drawback, particularly when flaws and tensile stresses coexist in the same region of the restoration. The flaw can be a microcrack on the surface (e.g. created during occlusal adjustment with a diamond stone), or it can be a subsurface porosity (e.g. from a processing error during the build-up and baking of the porcelain). When tension stress is applied, small flaws tend to open up and propagate cracks (crack propagation theory) [18].

In order to strengthen dental ceramic, it is essential that a mechanism should exist to prevent crack propagation under low tensile stresses. Current and future research would appear to centre around five approaches to strengthening dental ceramic as follow [13, 19]:

2.2 Strengthening of dental ceramic

2.2.1 Enamelling of metals

The term "porcelain-fused-to-gold (metal)" (PFM) or metal-ceramics have been widely used to describe the bonding of porcelain to metal alloys.The porcelain enamel is reinforced by the metal and is less likely to be placed under tensile forces which can cause brittle fracture.

The mechanisms of the bond between metal and ceramic are classified in several divisions such as mechanical, Chemical, Van der Waal's forces and compressive stresses [14].

For PFM restoration systems, it is preferable to have a positive expansion/contraction mismatch between the metal and veneering porcelain. The CTE of the metal should be slightly higher than that porcelain. It is thought that the greater thermal contraction of the metal during cooling subjects the porcelain layer to a compressive stresses. By contrast, a negative mismatch as a consequence of a greater expansion/contraction of the porcelain would produce tensile stresses in the porcelain layer. Compressive stress increases the strength of the whole restoration because ceramics are able to withstand higher compressive stress than tensile stress due to the fact that compressive stress tends to inhibit the propagation of cracks from the surface by keeping them closed. Nielsen and Tuccillo concluded that metal and porcelain are thermally compatible when their difference in thermal expansion/contraction ($\Delta\alpha$) is less than 0.5 μm/m·K in order to generate safe levels of residual stress. It is important to realize that metals show a ductile behavior under stress, while ceramics are brittle, which means that metals have the ability to deform to an applied stress without fracturing [15].

2.2.2 Dispersion strengthening of glasses

Glassy materials such as dental porcelain may be strengthened by dispersing ceramic crystals of high strength and elasticity in the glass matrix. Types of reinforcing crystals such as leucite, alumina, lithia disilicate, magnesia-alumina spinel or zirconia

2.2.3 Controlled crystallization of glasses

Controlled crystallization of glass depends upon the fact that glass, at ordinary temperatures, is a super-cooled liquid which does not crystallize on cooling from a melt. It can be made to crystallise by heating to a suitable temperature with crystal seed or nuclei present. The glass is then converted to a dense mass of very tiny interlocking crystals. Titanium dioxide is an effective nucleating agent and the starting glass must be homogeneous with qualities like optical glass [13].

2.2.4 Production of pre-stressed surface layers in dental porcelain via ion-exchange

When smaller sodium ions (Na^+) lying in the surface of dental porcelain are exposed to surface contact with liquids containing larger metallic cations such as potassium (K^+), the sodium ions may be exchanged with certain types of metallic cations. The larger potassium ions result in crowding of atoms at the surface of the porcelain and a pre-stressed surface layer is produced. This surface compression resulted in an increase in strength of the glazed or abraded condition of the porcelain [13].

2.2.5 Enamelling of high strength crystalline ceramics

Metal sub-structures are by nature unaesthetic and costly to produce, with the result that considerable interest is always shown by the clinician in all ceramic systems [13, 19]. Combining the strength of ceramic cores and superior esthetics of a weaker veneer ceramic can result in a reliable and more biocompatible restoration [3].

For an all-ceramic system, the thermal compatibility between ceramic substrate and veneering porcelain is based on the same principle of thermal expansion /contraction mismatch. Now, in place of the metal at room temperature, the high-strength ceramic core is subjected to a tensile stress while the weak veneering porcelain is still subjected to compressive stress. In contrast to metal, this condition might have a negative effect on the ceramic core, as the tensile strength of the brittle ceramic is much lower than its compressive strength. More likely, in an ideal all-ceramic system both veneering porcelain and ceramic core should not have significant mismatch in their thermal contraction values [15].

Core-veneered all ceramic restorations have gained popularity in the last few years. Combining the strength of ceramic cores and the esthetics of veneer ceramics and using layering techniques allows dental technicians to build an esthetic restoration with an individual character. The veneer consists of a glass and a crystalline phase of fluoroapatite, aluminum oxide or leucite. The core ceramic is commonly composed of crystalline nepheline or lithium disilicate in a glass matrix, or zirconia (zirconium dioxide : ZrO_2) [18].

2.3 Zirconia ceramic

Zirconia is the metal dioxide (ZrO_2). It was used for a long time, blended with rare earth oxides, as pigments for ceramics. The first biomedical application of Zirconia, was carried out in 1969 by Helmer and Driskell, while the first use of zirconia in orthopedics was introduced by Christel to manufacture ball heads for total hip replacements. Its application over the years was further expanded in dentistry; including the fabrication of brackets in orthodontics, post and core systems and ceramic implants/implant abutments offering improved esthetic alternatives[18].

Zirconium dioxide (ZrO_2) was introduced as core or framework material for all ceramic restorations because of its good chemical and dimensional stability, high mechanical strength and toughness, and Young's modulus similar to that of stainless steel. Fracture toughness is a very important physical property since it represents the ability of a material to resist crack growth. Clinically, lots of subcritical loads are applied on the materials by chewing, leading to the growth of subcritical cracks. Therefore, materials with higher fracture toughness are more ideal clinically, since it takes more energy to cause crack growth.

Zirconia ceramic is the material is polymorph occurring in three forms: monoclinic (m), tetragonal (t) and cubic(c). Pure zirconia is monoclinic in room temperature. This phase is stable up to 1170°C. Above this temperature it transforms into tetragonal and then into a cubic phase at 2370°C. When ZrO_2 is heated above 1170°C, the transformation from the monoclinic to the tetragonal phase is associated with a 5% volume decrease. Reversely, during cooling, the transformation from the tetragonal to the monoclinic phase is associated with a 3% volume expansion. These phase transformations, however, induce stresses which result in crack formations. The inhibition of these transformations can be achieved by adding stabilizing oxides such as CaO, MgO or Y_2O_3, which allow the existence of tetragonal-phase particles at room temperature. When sufficient stress develops in the tetragonal structure and a crack in the area begins to propagate, the tetragonal grains transform to monoclinic grains. The associated volume expansion results in compressive stresses at the edge of the crack front and extra energy is required for the crack to propagate further[18].

2.3.1 Yttrium partially stabilized Tetragonal Zirconia Polycrystals (Y-TZP)

Zirconia with 3-6 wt% Y_2O_3 added, require to stabilize the zirconia in the tetragonal phase which is otherwise not stable at room temperature, is called yttrium tetragonal zirconia polycrystals (Y-TZP)(20)].

Y-TZP is composed of many very small particles (tetragonal crystallites metastable at room temperature <0.5 µm) (Figure1) and is now available to dentistry through the use of computer-aided design/computer-aided manufacturing (CAD/CAM) technique. Compared to other dental ceramics, this ceramic has better mechanical performance and superior strength and fracture resistance and is used, among other applications, in orthopedics for hip-joint heads. The manufacture of dental restorations using Y-TZP is carried out in two ways: either by milling enlarged restorations out of homogenous ceramic green-body blanks of zirconia, which are then sintered and shrunk to the desired final dimensions, or by milling the restorations directly with the final dimensions out of highly dense sintered prefabricated zirconia blanks, so-called hot isostatic pressed (HIPed) zirconia blanks. Subsequently, the milled frameworks or copies are veneered with feldspar or glass-ceramics suitable for the Y-TZP ceramic used. During the veneering procedure the frameworks and copies are exposed to moisture at relatively high temperatures. Y-TZP has, however, been reported to be unstable over time, because of the spontaneous transformation of the tetragonal phase (t) into the monoclinic phase (m), which can cause mechanical property degradation. This transformation is affected by temperature, vapour, grain size, micro and macro cracking of the material, and type and concentration of stabilizing oxide, hot pressing and HIPing. Since the t→m transformation is influenced by temperature and vapour the possibility that the mechanical properties of the Y-TZP ceramic are affected during veneering cannot be excluded. In addition, it has been suggested that grinding by machining introduces residual compressive stresses on the surface, which influences the mechanical properties of zirconia ceramics and that subsequent heat treatment/veneering relaxes these residual stresses. The bending strength of Yttrium-stabilized tetragonal Zirconia (Y-TZP) ceramics is close to the actually used gold alloys. The development of Y-TZP has allowed for the production of reliable multiple-unit, all ceramic restorations for high-stress areas, such as the posterior region of the mouth. Owing to its high strength and toughness, zirconia may be a universal ceramic restorative material that can be used anywhere in the mouth. There is some confusion as to which manufacturer's zirconia does what and why zirconia might be advantageous[21].

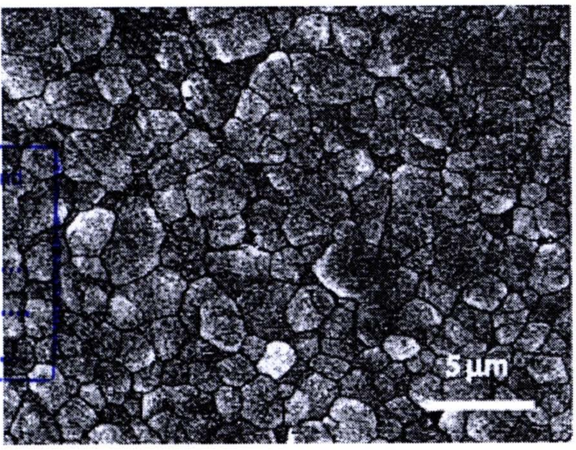

Figure 1 Microstructural feature of Yttrium Tetragonal Zirconia Polycrystals (Y-TZP)[22].

The majority of the Y-TZP–based (CAD-)/CAM systems use CAM of partially sintered Y-TZP blanks (Lava®, 3M Espe Dental AG, Seefeld; Cercon®, DeguDent, Hanau; Cerec InLab, Sirona Dental Systems, Bensheim; Procera® AllZirkon, Nobel Biocare, S-Göteborg). The size of partially-sintered infrastructures is increased during the milling stage to compensate for prospective shrinkage (20-25%) occurring during final sintering. The milling of these blanks is faster and results in less wear and tear to the hardware. With fully sintered blanks, such as DC-Zirkon (DCS-Precident, DCS Dental AG, CH-Allschwill), there is no shrinkage involved in the milling process, but microcracks may be introduced to the infrastructure [12].

For the all ceramic CAM system Cercon® (DeguDent, Hanau) offers a complete system consisting of an innovative of an innovative processing technology for the framework material zirconia, as well as new veneering ceramics optimally adapted to the framework. With its coordinated components, this system offers the dental technician a high degree of processing reliability. For the practicing dentist and the patient, the clinical study done at the University Dental Clinic of Zurich, successfully running now for the past 2.5 years, offers an assurance of safety. The high-strength, high-toughness zirconia framework can now be veneered using the veneering ceramic Cercon®Ceram especially developed for zirconia. Zirconia has a CTE of approximately 10 μm/m•K, and Cercon®Ceram was specially formulated for this CTE[23].

For all ceramic restorations has the principle of core-veneered compatibility that is similar to, the metal ceramic restorations, porcelain-metal compatibility. In the field of metal ceramic restorations, the manufacturers of dental alloys and porcelains design their products to be closely matched in thermal expansion characteristics. The goal of such a close match in thermal expansion behavior is to minimize residual stresses in the metal ceramic restoration. The effects of thermal expansion behavior on metal ceramic thermal compatibility are best understood when considered from the point of view of the contraction of the porcelain and metal upon cooling. At some point during the cooling of the dental porcelain, the glassy phase changes from a softened state to a rigid material able to support stress. This point is generally termed the set point of the glass. If a difference exists between the thermal contraction of the metal and porcelain, stresses will build up during the cooling cycle. A number of investigators have presented evidence that certain dental porcelains exhibit an increase in the average CTE when subjected to repeated firings. Thus, even if a particular porcelain closely matches a given metal in thermal expansion behavior at the initial firing, changes in the expansion coefficient of this porcelain will cause the development of a mismatch in expansion coefficients[24].

If the CTE (α) of the metal is smaller than that of the porcelain, the metal will contract less than the porcelain when the metal ceramic couple is cooled below the set point. At the set point, where the porcelain begins to support stress, the metal annulus (cross section) will fit precisely inside the porcelain annulus. But on cooling, the metal annulus contracts less than does the porcelain annulus. Thus, at room temperature, the metal annulus is too large for the space inside the porcelain annulus. If in this situation the metal annulus is considered to be compressed elastically and replaced inside the porcelain annulus, and then the compression is released, the porcelain will be subjected to radial compressive stresses (perpendicular to the metal ceramic interface) and tangential tensile stresses (parallel to the metal ceramic interface). In the reverse situation, where the CTE of the metal is larger than that of the porcelain, the tangential cracks will form upon cooling as a result of tensile stresses oriented perpendicular to the external surface of metal ceramic restorations. The stresses that cause such cracking are directly proportional to the difference in thermal expansion between the two materials, as well as the magnitude of the temperature drop during cooling[25].

2.4 Bonding of veneer ceramic to zirconia ceramic core

The bonding mechanism between veneering ceramic to Y-TZP core can allocate to the following :

2.4.1 Micromechanical bond and chemical bond

2.4.2 Firing to effectively wet the surface of zirconia

2.4.3 Heat treatment or veneering produce residual stresses

2.4.4 Sandblasting decrease interfacial failure and increase strength[16].

2.4.5 Fracture resistance (tensile strength) increases as core thickness/ veneer increases[26, 27].

2.4.6 CTE matching of core and veneer ceramic

The cause of inferior core-veneer bond strength is differences in CTE, poor wetting of core by veneer ceramic, firing shrinkage of veneer ceramic, transformation of zirconia crystals at interface (thermal or stress loading) and flaws formation during fabrication[28].

Differences in the CTE between the veneering and substrate (core) material result in the overall change in length for each material to be different. Bonding between the substrate and veneering constrain these differences in length and a CTE mismatch stress is developed. Successful material combinations should have well-matched CTE to avoid the incompatibility stresses that lead to cracking.

Several summaries for the composition of dental porcelain have been written. They cover the composition of feldspathic porcelain as a veneer ceramic in all ceramic and metal-ceramic crowns. The K_2O content was also varied to accommodate the need to match the coefficient of thermal expansion for metal alloys used in dental metal-ceramic techniques. The increase in K_2O content allowed a greater proportion of leucite crystals (CTE 27 μm/m•K) which led to the overall coefficient of thermal expansion rising to something in the order of 13.5 - 15.5 μm/m•K. The feldspathic porcelains used in all ceramic systems have coefficients of thermal expansion ranging from 5.5-7.5 μm/m•K when used over castable glass and alumina based core materials, to 16 μm/m•K when used over the newer pressed leucite systems[29]. With higher KNO_3 additions can increase CTE of dental porcelains which is important to match the CTE of the ceramic core[30].