T.C.

YENIYUZYIL UNIVERSITY

HEALTH SCIENCES INSTITUTE

DEPARTMENT OF PROSTHODONTICS

MASTER THESIS

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Supervisor

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ISTANBUL

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THE EFFECTS OF FIVE REPAIR SYSTEMS ON SHEAR BOND STRENGTH BETWEEN COMPOSITE AND TWO DIFFERENT INFRA STRUCTURES AFTER THERMOCYCLING

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İSTANBUL YENİYÜZYIL ÜNİVERSİTESİ

SAĞLIK BİLİMLERİ ENSTİTÜSÜ

Protetik Diş Tedavisi Anabilim Dalı Yüksek Lisans Tezli Programı
çerçevesinde yürütülmüş olan bu çalışma Yüksek Lisans tezi olarak kabul
edilmiştir.

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Abstract:

Purpose: The purpose of this in vitro study was to evaluate the effect of five ceramic repair systems on shear bond strength between composite and two infrastructures (zirconia and nickle chromium alloy) and to determine the mode of bond failure after thermocycling.

Materials and Methods: Disc-like specimens (N=100) were made of two dental ceramic infrastructure materials (zirconia, nickle chromium alloy). All the specimens were polished with silicon carbide paper, prepared for bonding by surface modification with airborne particle abrasion. Each infrastructure groups were divided into five subgroups (n=10). They were bonded using five repair systems: 1. Bisco intraoral repair kit, 2. Clearfil repair system, 3. Ceramic repair system(ivoclar), 4. Ultradent ceramic repair system and 5. single bond adhesive. All specimens were repaired with hybrid composite. Bonded specimens were stored in 37°C distilled water for 24 h and were thermocycled at 5–55°C for 1200 cycles with a 30-sec dwell time and 5-sec transfer time. Shear bond strengths were determined with a mechanical testing device. And mode of failure was recorded for each group.

Results: Infrastructure groups displayed the following values in megapascals: Zr (clearfil) =18,61±5,37, Zr (bisco) = 18,91±4.33, Zr (ultradent)= 6,63±1,5. Zr (ivocar) = 15,24±5,30, Zr(singlebond)=3,63±0,62, N-Cr(clearfil)=17,37±4,46, Ni- Cr(bisco)= 20,93±,2,63Ni-Cr(ultradent)=8,17±2,48, Ni-Cr (ivocar)= 17,37±4,46, Ni-Cr (single bond)= 10,33±3,55. Every repair system have no significant difference on repairing both Zr and Ni-Cr except single bond. Single bond was weekest for repairing Zr samples. The three repair systems (ivoclar, clearfil and bisco) had both mixed and adhesive failures. For single bond and ultradent groups all the spicemens were failed adhesively. The mode of failure result also supports that the three repair systems (ivoclar, clearfil and bisco) were better than single bond and ultradent repair systems.

Conclusion: The three repair systems (ivoclar, clearfil and bisco) were more effective than the other two systems for repairing both infrastructures and their mode of failures were both mixed and adhesive.

Dedication

I dedicate this thesis to my father Mohamed, my mother Fouzia, my whole family, my friends, my Country Libya. I wish to thank you for your support in the difficult and easier times. It is impossible to find the right words to thank you. With all my heart and love, I say, thank you very much.

Acknowledgment

I would like to express my deep gratitude and respect to Associate Professor Funda Erol, my supervisor, for her generous help, valuable advice during my research period. I wish to thank her for her patience, precious time and moral support. Her interest in my work, constant encouragement and confidence were essential to accomplish this work. These few words stated above could not express my deep respect to Dr Funda, her suggestions and creative ideas will forever never be forgotten. Thank you very much.

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1. Introduction and aims:

Prosthodontic treatments have conventionally desire to rehabilitate the impairment of function such as chewing, speech and swallowing while giving esthetics that achieve present day foundation for fascination. The necessitate for excellent esthetics is conditioned both by social pressure and the concerns of the dental employment. Only a few decades ago, some type of dental restoration for example partial coverage crowns and fenestrated crowns, were reported as esthetic restorations and spreading out of these restorations stays high. However, nowadays the description of esthetic restoration relates to ceramic restorations.

Ceramic used for produce dental restorations with many types of substructures are extensively recognized and employed in prosthodontic for oral reconstruction and there is a considerable studies in the literature have demonstrated the repair of porcelain fused to metal restorations, there is no information about the repair of substructures that involve metal alloy (non-precious), zirconia (1). Recent studies have tested whether fixed partial denture with zirconia frameworks exhibited similar survival rates and technical and biological outcomes to those with metal frameworks after a few years of function (1).

Despite the fact that the fractures of dental restorations do not unquestionably mean that the restoration is failed, the remaking of restorations it is cost money and takes long time and therefore stills a clinical challenge. Porcelain fracture is usually regarded an emergency treatment and the repairing procedure can produce uncontrolled challenges to the dentist. Due to the nature of the porcelain making, new porcelain cannot be added to an existent failed dental restoration intra-orally. The hand manufacturing of metal frameworks and porcelain veneers is taking long time and demands a very experienced technician. Remaking of a failed dental restoration is not necessarily the most practicable solving for the problem due to the apparently consequential costs and the complex nature of the dental restoration. Besides the economic and technical factors, it was mentioned that the cracks of fractured area might become a place for microorganisms and plaque accumulation and accompanied by staining (2).

The most commonly reported repairable clinical complications are minor chip-off fractures of the veneering ceramic. While a conventional approach is the replacement of all-ceramic restorations, when the restoration is not completely damaged, it can be repaired intraorally. Replacement of a damaged all-ceramic restoration may cause trauma to the supporting abutments and dental tissues, in addition, this procedure is not economical (2). Many commercially intraoral ceramic repair systems have been developed for ceramic restorations; however, studies indicate that ceramic repair systems are not a permanent solution because of their lack of high bond strength (2,3)

The bond strength between the repairing material and fractured restoration needs to be strong and durable. If the bond strength of ceramic repair systems can be demonstrated to be clinically demonstrated acceptable, producers and clinicians will be able to avoid wasting the time, material, and money required to make a new restoration (2,3).

In this study we used five different ceramic repair system to repair two infrastructures (zirconium and nickle chromium) with composite restoration as a chairside repair method for that the aims of this study is:

1-Compare the effect of five different repair system on shear bond strength between composite and zirconium and composite and nickle chromium alloy.

2- Mode of failure of bond between the composite and the two infrastructures.

2. Review of Literature

2.1. Metal ceramic restoration alloys:

Porcelain and ceramic materials have been used for fabricating esthetic dental restorations since the early 1800s. The first released reviews describing the positive use of porcelain fused to metal appeared in the mid-1950s (4,5). Good clinical performance has been supported by longitudinal studies that reported that up to 88.7% of metal-ceramic crowns and 80.2% of metal-ceramic fixed partial dentures (FPDs) were still in function after 10 years (6).

2.1.1. Requirements of alloys for metal-ceramic:

In spite of their chemical structure, the alloys mentioned on this part share at the least three common characteristics 1- they have the advantage to bond to dental porcelain, 2- they show coefficients of thermal contraction suitable with those of dental porcelain, 3- their solidus temperature is sufficiently high to withstand softening throughout the porcelain sintering procedure (7).

2.1.1.1. Porcelain bonding to metals:

The addition of a small quantity of base metal to noble and high noble alloys encourages oxide formation on the surface which promotes chemical bonding between the porcelain and alloy. The bond strength of base metal alloys to porcelain as checked in vitro have not generally been demonstrated to be higher ranking or lower ranking to those of noble metal alloys. Moreover, clinical studies not established a change on a failure incidence between metal-ceramic restorations made from base metal alloys and those made from high noble or noble metal alloys. Some research shows that frequency of porcelain firing may badly affect the bond between porcelain and metal infrastructure. Concerning the laboratory technician to make the best metal oxide features, the manufacturer's instructions must be followed exactly. The employ of cheaper, lower-purity aluminum oxide abrasives by dental technicians can contaminate the metal surface and subsequently impact the unity of the metal-ceramic adherence zone (7).

2.1.1.2. Coefficient of thermal contraction:

The thermal expansion and contraction measures of base metal alloys are typically like to those of noble metal alloys. The thermal contraction differential between metal alloys and dental porcelains might, under certain conditions, lead to high degrees of stress in porcelain which can cause cracking of porcelain or delayed fracture (7).

2.1.1.3. Solidus temperature:

When an alloy is heated up close to its solidus temperature may get susceptible to flow under its own mass (creep). The degree of creep can be increased by the size of the prosthesis and the number of firings that are demanded for porcelain veneering. All metal-ceramic alloys must have a solidus temperature that is significantly higher than the sintering temperature of the porcelain so as to decrease creep deformation (7).

Table (1) Alloy Classification by Noble Metal Content—American Dental Association 1984) (7).

Alloy Type	Total Noble Metal Content
High noble (HN)	Must contain ≥40% Au and ≥60% by weight of noble metal elements.
Noble (N)	Must contain ≥25% by weight of noble metal elements
Predominantly base metal (PB)	Contains <25% by weight of noble metal elements

2.1.2. Noble-metal-ceramic alloys:

Noble metal alloy is described by their corrosion resistant even under extreme conditions that occur in oral cavity (8). It is essential to integrate various components in gold to fabricate alloys with suitable properties. Incorporation of platinum raises the hardness and elasticity of and increase the melting temperature of

the alloy. When Palladium became too expensive, Platinum was used in place of Palladium, Copper in an adequate quantity relative to the gold component imparts the alloy heat-treatable, Silver is efficient in neutralizing the reddish color of copper and an ability to render good mechanical properties, has superior tarnish/ corrosion resistance and is relatively biocompatible. Zinc works as an oxygen removal throughout melting and casting of high noble and noble alloys. Iridium or ruthenium is added in small amounts as a grain refiner since smaller grains improve yield strength (7). Porcelain discoloration is not obstacle due to the fact that alloys contain small or no silver. The main disadvantages of the high-gold alloys besides the high price are low elastic. This element is also troublesome for fixed partial dentures and suggest to use a substitute alloys for these situations (8).

2.1.2.1 Gold-platinum-palladium alloys:

This group of alloys developed the first successful metal ceramic restorations. Platinum raised their melting temperature. Rhenium is added to some alloys as a grain refiner to enhanced hardness. Iron, which is added to form a bonding oxide, also raised the proportional limit and strength of the alloy. These alloys have enough elastic modulus, strength hardness, and elongation but are low in sag resistant. Therefore, uses of these alloys should be restrained to three-unit fixed partial denture (7).

2.1.2.2. Gold–palladium–silver (Au–Pd–Ag) alloys:

This type of alloys were the first replacement systems for higher –gold content group, brought in 1970 as Will-Ceram W (Williams), and may be continue to be in the market until now. The incorporation of significant amounts of silver (10% to 15%) and a relatively large amount of palladium content (20% to 30%) may lower the price of these alloys compared with the larger-gold content group (5). Their superior tarnish and corrosion resistance and relative freedom from technique sensitivity associated with porcelain bonding and thermal contraction to their long-term success (7). The primary weakness of gold–palladium–silver (au–pd–ag) alloys is their tendency to cause porcelain colour changes due to its silver content (8).

2.1.2.3. Gold-palladium alloys:

This silver-free alloy has a smaller coefficient thermal contraction than Au-Pd-Ag or Pd-Ag alloys. The Au-Pd alloys are considered almost ideal compared with other noble metal alloys, since these and their surface oxide is virtually indiscernibl. The esthetic quality of metal-ceramic prostheses made with Au-Pd alloys is corresponding to that made from Au-Pt-Pd alloys (7).

2.1.2.4. Palladium-Silver Alloys:

This group of alloys was introduced to the dental market in the 1970s (5). The elastic modulus of palladium-silver alloy is the most wanted from the of all the precious metal alloy, these alloy solder good and have very good sag resistant from all of the precious metal alloys. And tendency to the porcelain bond is superior. Mechanical properties of these alloys are bigger to even the most expensive noble metal alloys (8). The primary weakness of this group is tendency to change colour of porcelain to green which happens in a bigger degree in this group than in alloys have lower silver content (8).

2.2.1.5. Palladium–copper (Pd–Cu) alloy:

This type of alloys is a comparatively new generation, first brought in the dental market in 1982 as an alternative. These alloys are commonly composed of 70% to 80% palladium and comprise a little quantity or no gold, up to 15% by weight of copper, and approximately 9% of gallium. Such bigger amounts of copper evidently do not induce ceramic colour changing problems in alloys rich in palladium (8). This alloys do not cast as easily as palladium-silver alloys, but they are acceptable. The sag resistance is not very good as in the palladium-silver alloys, and they are contraindicated for long-span fixed partial dentures (8).

2.1.2.6. Palladium–cobalt (Pd–Co) alloys:

This type of alloys, made from 88% palladium and 4% to 5% cobalt by weight (8). The chief advantage of Palladium–cobalt alloys is a high coefficient of thermal expansion that is useful with certain porcelain systems. The primary

disadvantage of these alloys is their ability to form a dark oxide layer, which can alter porcelain esthetics. Moreover, it has been described that Palladium–cobalt alloys have low bond strength to porcelain than Palladium–copper alloys (9).

2.1.2.7. Palladium–Gallium (Pd–Ga) Alloys:

This type of alloys has greater high temperature and more strength and lighter surface oxide layer than other high-palladium alloys (9).

2.1.3. Base metal alloy:

Introduced in 1975 and at one time the largest-selling alloy in the United States (8). The replacement of gold by palladium increase the melting range but decrease the thermal expansion coefficient of the alloy. Silver, on the other hand, lowers the melting range but raises the thermal expansion. A good balance of the two components keeps a reasonably low casting temperature and a compatible thermal expansion coefficient. The thermal compatibility of these alloys is generally good except with certain low-expansion porcelains (7).

2.1.3.1. Nickel–chromium (Ni–Cr) alloys:

Nickel–chromium alloys are consisting mainly from   nickel and chromium; however, beside that contain many minor alloying metals. All Ni–Cr alloys have the same composition and physical properties, but may have differences from each other in corrosion resistance. Titanium and aluminum are incorporated in small amounts to form strengthening precipitates of Ni3Al Ti3Al; iron, tungsten, and vanadium are incorporated for hardening (9). The addition of beryllium increases fluidity and improves the castability of the alloy. Beryllium also controls surface oxidation lead to less technique-sensitive porcelain-metal bonds. Generally, these bonds are acceptable when the alloy contains beryllium but are doubtful when beryllium is absent (8). Nickel-chromium alloys have a very good sag resistance considered to be superior to all noble metal alloys. This feature, beside the raised stiffness and high tensile strength, give explanation for uses of these alloys in fixed partial dentures. The flexibility of a fixed partial denture framework made form nickel-chromium alloys is less than half that of a framework of the same dimensions made from a

high-gold alloy. Nickel-chromium alloys have the biggest elastic moduli of all dental alloys, which minimize flexibility to a significant degree. In perspective, allergies to nickel in only 5% to 8% of the population (8).

2.1.3.2. Cobalt–chromium (Co–Cr) alloys:

Cobalt is the main component of cobalt-based metal-ceramic. The castability of Co-Cr is in the same range as that of Ni-Cr without beryllium. It has been proposed that Co alloys be further categorized into two subgroups: those that contain ruthenium and those that are ruthenium-free. Cobalt–chromium alloys are the most common base-metal replacement for patients known to be allergic to nickel (9). With the exception of titanium alloys, the Co–Cr have the highest melting ranges alloys of the casting alloys. In part, this makes it difficult to control these alloys in the laboratory (9).

2.1.4. Bonding behaviour of PFM (Porcelain Fused to Metal):

Bond strength is defined is the bond developed between substrates, in this case metal and porcelain after porcelain fused to metallic framework. The factors that contribute to a strong bond between metal framework and porcelain are following considered:

2.1.4.1. Chemical:

Non-precious metals as an alloy component produce metallic oxides at the surface of the framework, which enables a chemical bond to develop with the porcelain. To create a reliable bond an oxide layer needs to be uniform both in thickness and composition. An alloy that forms too much surface oxide after oxidation can lead to lower bond strengths. Therefore, it is important to remove some of the oxide layer thickness by blasting the alloy with Al2O3 particles following oxidation. An oxidation/reaction layer is developed at the interface on metals during ceramic firing. (10,11,12).

2.1.4.2. Mechanical:

Reliable mechanical bonding is very important for good clinical performance of metal-ceramic restorations. Removing casting oxides as well as properly sandblasting the alloy with 100 - 250 μm aluminium oxide particles has become the standard approach to create such a reliable mechanical bond (10,12).

2.1.4.3. Coeffiecient of thermal expansion:

Coeffiecient of thermal expansion compatibility between porcelain and underling framework is necessary to achieve a reliable bond. After producing a homogenous oxide layer and sandblasting, if the coeffiecient thermal expansion of alloy is not compatible with the coeffiecient of thermal expansion of the porcelain it will generate significant stresses that may distrubt the bond either fracture or debonding of porcelain. As a result of many investigations it is advantageously preferred to have the CTE of the porcelain slightly lower (approximately 10%) than that of the metal framework to ensure surface compressive stresses develop during cooling (12,13).

2.2. Zirconium:

2.2.1. History:

The name describing Zirconium derived from Arabic word "Zargon" that means "golden in colour". Zirconium dioxide (ZrO2) was incidentally detected by a German chemist called Martin Heinrich Klaprot. The first recommended application of Zirconium as a ceramic biomaterial was recorded for total hip replacements (THR). Afterwards, focus was more upon the evolution of zirconia yttrium ceramics incorporation usually distinguished as Tetragonal Zirconia polycrystals (TZP). Because of its beneficial mechanical and dimensional stability, such as mechanical strength and toughness, TZP applied as practical employment in space shuttle, automobiles, cutting tools, and combustion engines in 1990s (14). Zirconium was employed as endodontic posts and as abutments for implant. (14).

Zirconium (Zr) is a metal oxide and colour of zirconium is greyish. The melting temperature of zirconium is 1852 °C and a density of 6.49 g/cm³.The boiling temperature of zirconium is 3580 °C, the crystal of zirconium is hexagonal in shape. Zr is in nature cannot be in a pure condition. It can be detected in conjugation with silicate oxide with the mineral called Zircon (ZrO2 × SiO2) or as a free oxide (ZrO2) with the mineral called Baddeleyite.Due to impurities of various metal ingredient of zirconium that change colour, and radionuclides constituents like urania and thoria which made them radioactive zirconium and it minerals cannot be applied in dentistry as primary materials without purification (15). Mechanical properties of zirconium are identical to mechanical properties of metals and its colour is same to tooth colour. Its mechanical properties are higher than any dental ceramic, for this reasons zirconia can be used for posterior fixed partial dentures and lead to a significant decrease in core thickness (16).

2.2.2 Phases of zirconia:

Zirconia have three different crystalline shape: monoclinic (m), cubic (c) and tetragonal (t) at room temperature. Pure zirconia has a monoclinic structure and it is unchanging until 1170°C. Tetragonal zirconia is formed from 1170°C to2370°C, and at temperatures over 2370°C until the melting point (2680°C) cubic zirconia is created. Throughout cooling immediate reversal of changing occur. Moreover, the t–m transformation combined by 4-5% volume expansion lead to creation of high compressive stresses in the material (16).

(a)   (b)   (c)
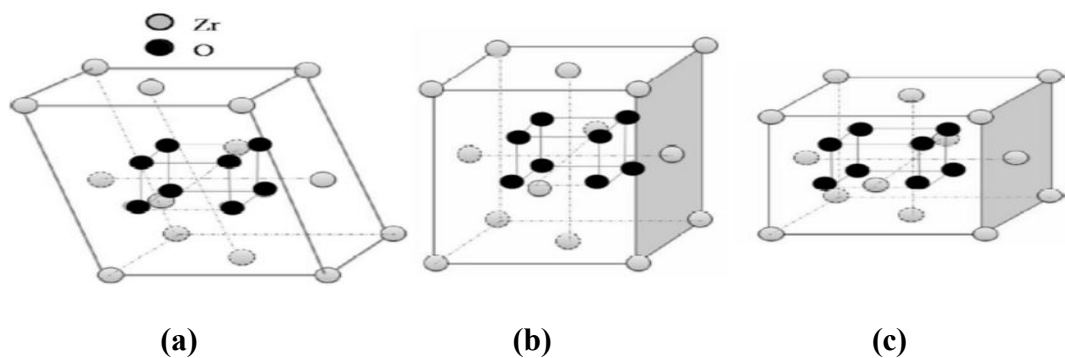
Figure (1) Types of polymorphs of zirconia: (a) monoclinic phase; (b) tetragonal phase; (c) cubic phase (14).

2.2.3. Stabilized zirconia:

Stabilize the tetragonal and/or cubic phases of zirconia by incorporation of various dissimilar oxides to zirconia like Magnesia (MgO), Yttria (Y2O3), Calcia (CaO), and Ceria (CeO), these oxides permit the development of multiphase materials as Partially Stabilized Zirconia (PSZ) commonly consists of cubic zirconia as major phase and monoclinic and tetragonal zirconia as the minor phase in it mircosturcure (15).

When zirconium consists mainly from transformable t-zirconia grains it is called as Tetragonal zirconia polycrystals (TZP). For dental applications zirconia stabilized with Y2o3 is used because of the its optimum properties. Yttrium-Oxide Partially Stabilised Zirconia (y-PSZ) it composed from100% small metastable tetragonal material grains (Y-TZP) after the incorporation of 2 to 3 % yttrium oxide (y2O3) as a stabilizing agent (16).

2.2.4. Phase transformation toughening:

Starting of crack on surface of zirconia include metastable tetragonal (Y-TZP) it is submitted to tensile stress intensity at the crack crest leads to conversion of metastable t- $ZrO2$ to the monoclinic crystalline phase and the volume of the crystals increase by 3% lead to restrained of increased crystals by the surrounding ones, guides to a favourable compressive stress which acts on the tip of the crack, and stop it is   spread (16,18).
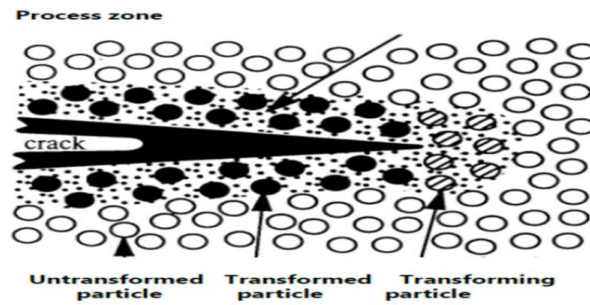
Figure (2) Phase transformation toughening of zirconia (16).

2.2.5. Mechanical Properties of zirconium:

Zirconia Mechanical properties higher than mechanical properties of any ceramics used in dentistry (16). Fracture toughness of Zirconia is between 6 and 10 MPa and it is almost twice aluminium oxide ceramics, good fracture toughness of zirconium related to transformational toughening mechanism, which impart zirconia its extraordinary mechanical properties (19). Zirconia have superior wear manner but it is lower than porcelains (20).

2.2.6. Biocompatibility of zirconia:

Zirconia has been studied widely. Many studies regarding it is biocompatibility has been done in vivo and vitro confirm the high biocompatibility of Y-TZP.The utilizing of extremely refined powders zirconia that have been filtered from it is radioactive content. Zirconia has no cellular or systemic deleterious responses to the material were described. Concerning periodontal health, none of the studies reported no changes regarding biological health of the soft and hard tissues surrounding crowns or bridges made from zirconia (15). Bone osteo integration around Zirconia implants as same as around titanium implant. Zirconia have less bacterial colonization in tissue than titanium (16).

2.2.7. Radioactivity of zirconium:

Zirconia includes little quantities of radionuclides as uranium-radium (226Ra) and thorium (228Th) actinide series. After filtering procedure, zirconia have a minimum radioactivity (< 100 Gyh-1) can be achieved and it is under the European

radiation border for human body. The radioactivity of zirconium is same to alumina ceramics and Co-Cr alloys (16).

2.2.8. Esthetics properties and light transmission of zirconium:

Yttria-stabilized zirconia demonstrates elevated refractive index ranged from (2.1 to 2.2), coefficient of zirconium light absorption is minimum, and in the visible and infrared spectrum zirconium has elevated opacity. The increased opacity of zirconia is very practicable in esthetically requiring clinical indications for masking black teeth (16). Translucency of zirconia restoration (crowns or bridges) is described to be lower than that of EMax ceramics restoration, In-Ceram Zirconia has a comparatively lower translucency similar to porcelain fused to metal crowns when the evaluated contrast ratio method used (16). Radiopacity of zirconia to considered to be high similar to radiopacity of metal alloys leads to enhance the radiographic examination of marginal integrity and recurrent caries for zirconium restoration (16).

2.2.9. Aging of zirconium:

Low water degradation zirconia was first discovered 1981 by Kobayashi where in a humid environment, spontaneous changing from the tetragonal to the monoclinic phase developed in zirconia grains at lower temperatures ranged from 150°C to 400°C. The delay changing of tetragonal crystals to the unchanging monoclinic phase begins at the surface in separated grains by a stress corrosion mechanism. The transformation of one grain is accompanied by an elevation in volume that induce stresses on the circumventing grains, leads to formation of microcracks on zirconium surface which imparts a route for the water to diffuse inside the specimen. As a result of water diffusion degradation of zirconium surface becomes worse and the transformation progress from one grain to another. Ageing of zirconium is linked to roughening which will guide for increase of wear and microcracking, as result of wearing grain will pull-out and collection of particle debris and slowly growing of the crack guiding to early failure of zirconium

restoration. This phenomenon can have e detrimental consequences to zirconium mechanical properties (18,22).
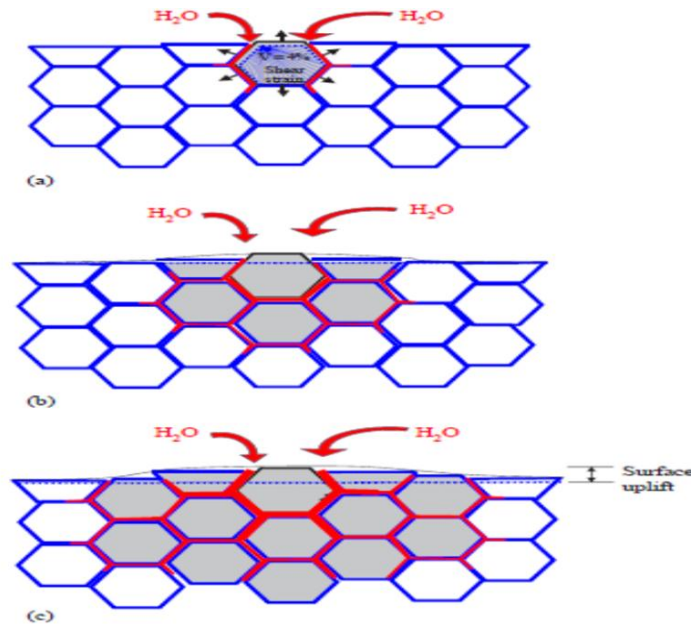
Figure (3) Low-temperature degradation of zirconia (19).

2.2.10. Different types of zirconia ceramics available for dental applications:

In spite of the fact that numerous kind of zirconia systems are at the present time available just three systems are employed for the present time in dentistry. These

are yttrium stabilised tetragonal zirconia polycrystals (3Y-TZP), magnesium partially stabilised polycrystals zirconia (Mg-PSZ): and zirconia-toughened alumina (ZTA).

2.2.10.1. Yttrium cation-doped tetragonal zirconia polycrystals (3Y-TZP):

This type of zirconium is the predominately used in dentistry. It made up from transformable t-Zr grains stabilized by the incorporation of 3% yttrium-oxide (16). Femoral heads is made from Y-TZP in total hip substitute prostheses from 1990 but its application in orthopedic surgery decreased by more than 90%, because of a series incompetence that happened. In 2001 3Y-TZP is achievable for using in dentistry for example manufacturing of dental crowns and bridges. The crowns and

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bridges are fabricated by two ways, the first way is milling of restoration through soft machine from presintered blanks after that the restoration is sintering at elevated temperature, the second way through milling of restoration by hard machine of fully sintered blocks (23).The microstructures Y-TZP is containing tiny grains it is diameter ranged from (0.2 to 0.5) mm depending on the sintering temperature which hinders the phenomenon of un stable structural of zirconia in the presence of saliva, slowing the development of subcritical cracks leads to very low porosity and high density of zirconium restoration (16). The 3Y-TZP mechanical properties powerfully related to size of zirconium grains. Over the decisive size of grain, 3Y-TZP has low stability and grater liability to spontaneous t→m transformation whereas a little grain sizes (<1m) are accompanied with a minimal transformation grade. Furthermore, beneath a definite (0.2m) size of zirconium grain, the transformation of zirconium grains is not likely to occur, but it leads to low fracture toughness of zirconium (23). At the present time availability of 3Y-TZP for soft machining of dental restorations use final sintering temperatures ranging from 1350 to 1550 ◦C related to the manufacturer instruction. (23). The fracture toughness of tetragonal 3Y-TZP is about 8 to 10.3 MP (21).

2.2.10.2. Zirconia toughened alumina:

Zirconia-toughened alumina is composed of 70% to 90% alumina and 10% to 20% zirconia by weight. The core of In-Ceram Zirconia earlier to glass infiltration, by weight is approximately 62% alumina, 20% zirconia, and 18% glass. Similar to the toughening of Y-TZP, ZTA is toughened by a stress-induced transformation mechanism (18). In-Ceram Zirconia can be fabricated through two ways first way with slip casting and second way through soft milling machining. The early sintering procedures happens at 1100 ◦C for two hours, preceding to the sintering procedures the porous ceramic composite being glass-infiltrated, the quantity of porosity is considered to be larger than sintered 3Y-TZP and the percentage of porosity is ranged from 8 to 11%. This may give explanation concerning lower mechanical properties of zirconia toughened alumina when compared to 3Y-TZP dental ceramics. Zirconia toughened alumina fabricated by soft milling machining is

considered to   demonstrate good mechanical properties compared to the slip-cast method due to uniform processing (23).

2.2.10.3. Partially stabilized zirconia (Mg-PSZ):

In spite of the fact that a lot of researches has been done concerning the possibility for employ partially stabilized zirconia (mg-psz) in biomedical field, this material is not successful because of porosity, related to the large grain size (30–60m) that can cause wear (20). MgO is incorporated in material as stabilizer in concentrations a little than that need for full c-ZrO2 stabilization. The amount of MgO in general ranges from 8 to 10 %. (16).

2.2.11. The bond between zirconia and veneering ceramics:

At present, there are two commonly used methods of securing ceramic onto zirconia frames: the layering technique and the press technique. In the layering technique, porcelain powder is applied onto the zirconia frame before firing. In the press technique, the lost wax method it is employed to create the restoration. The ceramic ingot is fired up and after that pushed by the pressure into a wax-formed void. The layering technique is usually used for PFM crowns. It results in excellent esthetics . but several firings are required in order to reproduce the desired colour. For both the layering technique and the press technique, the coefficient of thermal expansion of the porcelain used for veneering is the same or a little bite lower than that of zirconia. Because of the huge difference in the coefficient of thermal expansion between a zirconia frame and veneering ceramic will cause expansion between a zirconia frame and veneering ceramic will cause residual stress on the crown, thus resulting in reduced reliability of the restoration (25).

2.2.12. Mechanism and evaluation of integration between zirconia and veneering ceramics:

There is no clear evidence demonstrating the presence of chemical bonding between zirconia and veneering ceramics, although there is one report suggesting

such a bond. It is thus assumed that mechanical bonding plays the major role in the zirconia-to-porcelain integration of zirconia based restorations (26). There is an international standard (ISO9693) used for evaluate the bond strength between metal and ceramic using a bending test, and PFM restorations in clinical use are required to have a bond strength of 25 MPa or more (27). Although there have not been many reports (27). Concerning the evaluation of zirconia-to-porcelain integration using a bending test (ISO9693), all of those reported that the bond strength was 25 MPa or more. In experiments where the bond strength between metal and porcelain and that between zirconia and porcelain were compared, many studies established that the bond strength between metal and porcelain is greater than that between zirconia and porcelain (28).

2.2.13. Factors affecting bond strength between zirconia and veneering ceramics:

2.2.13.1. Veneering ceramic:

It is known that the strength of the bond between zirconia and veneering ceramic varies greatly with the type of veneering ceramic used (29). This is probably because different veneering ceramics have different coefficients of thermal expansion, causing a mismatch in the coefficient of thermal expansion between zirconia and the veneering ceramic being used (30).

2.2.13.2. Zirconia:

Sandblasting is the most widely-used surface treatment method in dentistry. For porcelain-veneered zirconia restorations, the purpose of sandblasting is to produce irregularities on the zirconia to enhance the mechanical bonding between zirconia and veneering ceramic. It has in fact been reported that sandblasting produces changes in the surface topography and surface roughness of zirconia.

2.3. Direct composite resins:

Many composite restorations have been formulated in recently with a large number of shades, translucencies, opacities, with advanced placement techniques,

make the manufacturing of restorations that dependably simulate the polychromatic variations and optical characteristics present in natural teeth (31).

2.3.1. Composite resin formulations:

An organic resin matrix, inorganic filler particles, and a coupling agent is the main component of composite resin restoration. The monomers, initiator systems, stabilizers, and pigments made the resin matrix, while the inorganic filler consists of some form of glass filler particles., most composites are fabricated using either bisphenol-A-diglycidylmethacrylate (bis-GMA) or urethane dimethacrylate (UDMA) matrix polymers and different kinds of filler particles, such as glass, quartz, and colloidal silica (32).

2.3.2. Classification of composite resins:

The most regularly utilized categorization takes mainly the average particle size and distribution of a given composite's filler phase. Most composite resins can be classified in four groups.

2.3.2.1. Microfilled composite resins:

Its brought in early 1980s. They are mainly filled 35% to 50% by weight with 0.02-μm to 0.04-μm silicon dioxide filler particles. The principal features of these type of composites restoration are the high polish surface that can be preserved over time and awesome enamel-like translucency. For these reasons the microfilled composite indicated for the restoration of anterior teeth and cervical abfraction lesions, and it not indicated in heavy stress-bearing areas because they often often lead to marginal chipping and bulk fracture (33). In general, their physical properties are less than hybrid composites because of minimum filler content; the exception is their compressive strength, which can be relatively high. These type of composite restorations have smaller module of elasticity, greater water sorption, large coefficients of thermal expansion, and smaller fracture toughness, more polymerization shrinkage, smaller tensile strength, (32).

2.3.2.2. Hybrid composite resins:

This type of composite restoration contains un homogeneous aggregate of filler particles. They are commonly filled 70% to 80% by weight with 0.04-μm and 1-μm to 5-μm filler particles. The particle size of this type of composites is usually > 1 μm. This mixture of fillers gives very good physical properties and high polishability of it is surface when compared with macrofilled composites. The main disadvantage of hybrid composite it is loss of it is gloss (34). The demand for a highly polishable composite resin with optimum physical properties for apply in the posterior and anterior teeth regions, fabricators formulated microhybrid composite restoration. They decreased the particle size, ranging from 0.04 μm to 1 μm. Through addition smaller particles, microhybrid composites can be polished and handled better than their hybrid composite restorations. Also the microhybrid composites are harder than most microfilled composites. The indication of mircrohybrid composite restoration in both anterior and posterior teeth. So the microhybrid composites can be used as all-purpose composite resins. Concerning consistency, the large percentage of microhybrid composite are medium viscosity; nevertheless, some other types composite can be classified as 1- high-viscosity (packable composites) 2-low-viscosity (flowable composites) (33).

High-viscosity composite resins (condensable) have a large amount of load (above of 80% in volume), which makes them more resistant and easier to apply because of their excellent sculptability. The indication of this composite resins for the reestablishment of proximal contacts in class 2 cavity restorations and for the exact making of occlusal anatomy features (34). The clinical results of these type of composites is like to regular-viscosity microhybrid composites, so the choice of this type of material must be based on personal preferences (35). The flowable composite resins present much less load (about 50% in volume), have lower mechanical properties, and have more polymerization shrinkage (36). A lot of studies have established that the employ of flowable composites does not make the marginal sealing of adhesive restorations good (37,38). This type of composite indicated in areas of difficult access and irregular cavity preparations due to it is easy insertion and adaptation of increments (39). These composite resins are also useful in the restoration of highly conservative preparations such as applying of preventive

composite restorations, repairing margins of old composite resin restorations, cementation porcelain veneers, and applying of temporary pedodontic restorations (34,40).

2.3.2.3. Nanofilled composite resins:

This type of composite restoration was newly brought in dental market and they composed of nanomers (5 nm to 75 nm particles) and "nanocluster" as the fillers. Nanoclusters are agglomerates (0.6 μm to 1.4 μm) of primary zirconia/silica nanoparticles (5 nm to 20 nm in size) mixed together at points of contact, and the leading to porous structure is infiltrated with silane (41). The nanofilled composites have the same mechanical and physical properties of microhybrid composite, but about gloss retention they perform significantly better (41).

2.3.2.4. Macrofilled composite resins:

This type of composite resins still available on the market and their use is very limited because of very low mechanical properties lead to very bad clinical performance (roughness, staining, wear, and discoloration (41).

2.4. Chipping of veneering ceramics in zirconium dioxide fixed dental prosthesis:

2.4.1. Classification of chipping ceramic from veneering zirconium fixed partial denture:

The chipping of porcelain-veneered can be classified by seriousness and the amount of treatment required for repair according to Heintze and Rousson as:
• Grade 1: Small surface chipping Can be treated by polishing the restoration surface.

• Grade 2: Moderate surface chipping. Can be treated by resin composite repair system.

• Grade 3: Severe chipping lead to display the zirconia core. Treatment: remake the restoration again (42).

Raigrodski, Anusavice and Heintze proved that the most familiar types of zirconia-based fixed dental prostheses chipping are Grades 1 and 2, which do not include restoration impairment (43).

2.4.2. Possible reasons for chipping veneering ceramic in zirconium prosthesis:

2.4.2.1. Mechanical properties of veneering materials:

One of the main causative agents for chipping is the minimum mechanical properties of veneering ceramics compared to different ceramics employed for frameworks. The fracture toughness of veneering ceramic is low as 1mpa, whereas the fracture toughness of zirconia 10 MPa due to the phase transformation toughening mechanism (44). The veneering ceramics has a low fracture toughness because of their microstructure that is consists of mainly a weak glassy matrix and low amounts of leucite crystals, ranging from 5 to 30 %, related to the commercial brand (45). Glass composition of the ceramic matrix is extremely subjected to subcritical crack growth (which happens as a result of corrosive action of water in conjunction with tensile stresses intensified around the crack crest (46-49).

2.4.2.2. Framework design:

Another agent that affect chipping are planning of the framework of crows and bridges and the proportion of veneering porcelain thickness to the framework thickness. The optimum thickness copings is 0.5 millimeter in case is dos not account for single anatomical crown or multi units fixed partial denture dimensions, which result in a wide differences in thicknesses of ceramic used for veneering and changes in the ratio of infrastructure thickness to the veneering porcelain thickness. Some scientist established that variation of thickness related to veneering porcelain influence the strength and crack initiation of veneered porcelain (50). Scientists generally agree that good anatomically designed of copings are better to reduce chipping (51,52). Several investigators reported that a consistent veneering porcelain thickness resulted in a more even distribution of residual stresses in the material (53,54, 55).

2.4.2.3. Veneering method:

More research is needed to decide the better method to produce the veneering porcelain for zirconia frameworks. Many in vitro studies demonstrate good results with hand-layered veneering method than for pressing method over the frameworks (56,53). Some authors established good results by applying computer-aided manufacturing to create veneering zirconia framework with porcelain (57).

2.4.2.4. Thermal residual stresses:

This factor is extremely depended on two agents. The first agent is the differences in the coefficient of thermal expansion (CTE) between zirconia and porcelain used as veneer, and second agent is the temperature that produced on the percaline which is used as veneer throughout cooling (58.59). The variation in the CTE between zirconia framework and porcelain leads to different ranges of contraction of both materials. Therefore, tensile or compressive residual stresses are development and propagated un homogenously over the porcelain and zirconia layers. The second factor, the temperature gradient is referred to very low thermal diffusion of veneering porcelain and zirconia. The variation among the temperature of the porcelain surface used as veneer and the zirconia framework more than 200 °C in a sample with thickness 0.7mm (60). Both materials require time for the surface temperature to correspond their inner part. The temperature gradient leads to development of residual stresses along there construction throughout the cooling process (61).

2.4.2.5. Thermal conductivity:

The raised occurrence of chipping of porcelain that used for veneering zirconia restorations is due to low thermal conductivity of the zirconia. Throughout cooling process, the residual stresses originates in the veneering porcelain due to the temperature gradient among the cool external surface and the warm internal surface neighboring the coping. As a consequence of that, tensile stresses spreads deeper inside of the porcelain leads to speed up the crack generation (60,59,63).

2.4.2.6. Phase transition:

The zirconium crystals convert from the tetragonal crystal shape to the extra stable monoclinic crystal shape (64), the transformation procedure happened with 4% raised in volume (65). The feature of phase transition causes a toughening mechanism of zirconia that prevent crack from spreading. At the porcelain zirconia interface, phase transition causes tensile stresses on the deep side of the veneering ceramic, gives a beginning spot for cracks development. However, many research proves that tensile stresses produced by veneering zirconia with porcelain are very low to create a general phase change at the interface among zirconia and porcelain (66,67).

Table (2) Summary of complications and survival rate for zirconia-based restorations (68).

Authors (Year)	Type of restoration	Mean time	Sample Size	Framework complication	Veneer complication	Survival ,rate
Philipp et al.2010	3unit FPDs	1year	8	0	0	100%
Roediger et al.2010	3–4unit FPDs	4years	99	1	13	98.9%
Vigolo et al.2011	Single crowns	5 years	20	0	2	79%
Sorrentino et al2012	Single crowns	5 years	20	0	1	85%
¨Ortorp et al2012	3unit FPDs	5 years	48	0	3	100%
Kern et al 2012	3–4 unit FPDs	5 years	20	3	-	90%
Salido et al.2012	4unit FPDs	4 years	17	3	5	76.5%
Pelaez et al.2012	3unit FPDs	4 years	20	0	2	95%

| Rinke et al.2012 | 3–4unit FPDs | 7 years | 97 | 5 | 23 | 83.4% |

2.5. Metal ceramic restoration fracture reason:

2.5.1. Technical factors:

2.5.1.1. Compatibility between the coefficient of thermal expansion of the metal and thermal expansion of the ceramic.

The stress intensity at the metal-porcelain interface is due to the variation between the coefficient of thermal expansion of the metal and coefficient of thermal expansion porcelain (69). Good bond between porcelain and metal required little differences between coefficient of thermal expansion of porcelain and metal (70), usually a $0.5 \times 10{-6}°C$ difference in the coefficients of thermal expansion of the metal and porcelain is suitable (71).

2.5.1.2. Surface treatment and design of the metal coping:

The fracture resistant of porcelain was seriously decreased when porcelain was fused to metal surface without oxidation layer on it is surface and when incorrect thickness of the porcelain was utilized (72). Some scientist proved that decrease oxidation layer on alloy surface by thirty percent (30%) lead to decrease in the bond strength between metal and ceramic (73). Nevertheless, increase thickness of oxides layer on metal surface lead to increase the risk of metal-porcelain fracture (74). The argument concerning the bond strength between metal and porcelain if it is affected by increasing the roughness of the metal surface or not. Kelly and colleagues believed that more rough metal surfaces may increase stress intensity at metal and porcelain bond (75). Nevertheless, some scientist proves that the bond between ceramic and metal is (66.67%) from total metal ceramic bond is chemical bond and (33.33%) from total metal ceramic bond is van der Wall's force. Therefore, the influence of metal alloys surface roughness on bond strength between metal and porcelain is weak as the authors decreased the significance of mechanical bonding (76). A gently roughened metal surface was wetted more readily, lead to probably raising the bond strength between metal and ceramic. Contours of connector and also

the cross-sectional dimensions of connectors have a powerful influence on the strength and stability of the framework (77). The thickness of the connector should be enough to impart sufficient resistance to occlusal forces (78). Furthermore, the thickness of occluso-gingival surface of the pontic has an influence on the bending of framework of crown or fixed bridges, the bending different directly with the cube of thickness of the occluso-gingival surface of pontic, making the pontic one half as thick will also made it deflect eight times as much (78).

2.5.1.3. Thickness of porcelain:

Weaker the restoration because of the thicker the porcelain due to the direct relationship between the stress concentration at metal and ceramic bond and thickness of the porcelain and brittleness of the porcelain under tension (69). The porcelain behind the interface is commonly under compression stress because of the contraction of metal is more than contraction of porcelain, nevertheless the further the surface of the porcelain is from the interface, the greater the tension (76). Therefore, to decrease creation of microcracks, a properly undifferentiated porcelain thickness is advocated (79).

2.5.1.4. Ceramic build-up and firing technique:

Evans et al. highly advocated reducing air entrapment between the ceramic particles during ceramic application, because porosity does occur can cause ceramic fracture (80). Cracks form in the surface of ceramics through abrasion by dust; such cracks, matched with low fracture toughness, impair the strength of ceramics (81). The rate of cooling and heating during porcelain firing affect the stress concentration at the metal-ceramic interface (69). Repeated firings or a very high oven temperatures have been considered as induces of superficial and deep imperfections or porcelain blistering (82).

2.5.1.6. Thickness and elastic modulus of the metal substructure:

The veneering porcelain support is directly associated to the modulus of elasticity, not colligated to the strength of the core material (83). Alloys with a raised

modulus of elasticity withstand deformation better than alloys with low modulus of elasticity (84).

2.5.1.7. Location of porcelain-metal finish lines:

If partial coverage crown made from metal ceramic is decided to be an option of treatment, the lingual or occlusal finish lines position is important. The metal and porcelain junction must not be placed at centric occlusion contacts in order not to display the porcelain-metal junction to the extra force. The angle of porcelain-metal occlusal junction should be a 90-degree or larger to prevent thin "lips" of metal that may deform during service (82).

2.5.2. Factors related to the dentist

2.5.2.1. Sufficiency and pattern of tooth preparation:

Under reduction of tooth preparation, which guides in too short inter occlusal distance for the metal core and the veneering porcelain, leads to fracture of porcelain (79). Furthermore, sharp line angled preparations rise the possibility for production of micro cracks among the porcelain throughout firing procedure (85).

2.5.2.2. Incorrect registration of occlusion and articulation often causes destructive premature contacts:

Inadequate diagnosis and an improper design are significant factors affecting the long-term success of fixed partial dentures, and the clinical skill of the dentist is extremely important for increasing the longevity of metal-ceramic restorations.

2.5.2.3. Anterio-Posterior Length of Pontic Span:

Long anterior posterior metal infrastructures bend under heavy loads conducting to porcelain fracture (85). A fixed partial denture with two-tooth pontic span will flex eight times as much as a single-tooth pontic fixed partial denture will, if everything else remains un- changed (86). Replacing three posterior teeth with a fixed partial denture rarely has a good prognosis, especially in the mandibular arch. Under such considerations, it is better to make implant-supported prosthesis or removable partial denture (87).

2.5.2.4. Environmental factors:

The water can act chemically at crack tips, diminishing the strength of glasses and ceramics. This phenomenon is termed "chemically assisted crack growth" or "static fatigue" (87). It has been established that silicate bonds in the glassy ceramic matrix are susceptible to hydrolysis by environmental moisture in the presence of mechanical stress. declines of 20% to 30% in metal-ceramic bond strength were detected in moist environments (88). As a consequence, this static fatigue guides to the propagation of fractures along the microcracks causing failure in the restoration (79). Moreover, it is established that common beverages with low pH ranges could also cause fractures in glass-containing dental (89).

2.5.2.5. Inherent material properties:

The mechanical fatigue of ceramics is probably controlled by various factors including crack length, microstructure and fracture toughness. It has been proven that amorphous materials like glasses do not have an ordered crystalline structure as metals, and dislocations of crystalline lattice do not exist in glassy materials; thus, they have no mechanism for yielding without fracture (90).

2.5.2.6. Direction and frequency of applied loads:

Niedermeier et al, proved that occlusion is not registered correctly and articulation is not checked properly, the premature contacts would act as stress bearing zones on the ceramic (91). Llobell et al stated that mastication, parafunction and intraoral occlusal forces produce repetitious dynamic loading; they regarded impact load and fatigue load as cause for intraoral ceramic fracture (92). Anusavice and Zhang also described that high biting forces may cause glass-containing dental restorations to fracture (90). Stress direction is another contributory factor for failure, as sometimes failure happens at sites of comparatively low local stress just due to a large flaw oriented in the stress field and this is ideal for inducing fracture (79). White and Li stated that the sites possible from which failure may start are highly

unpredictable because its depends on flaw size and is corresponding to the stress distribution (94).

Table. (3) Summary of complications and survival rate of metal–ceramic crowns (95).

Study	Year of publication	Total number of crowns	Mean follow up time	Number of failure	Estimated survival after 5 years (%)
Braegger et al.	2007	106	17	28	91,6%
Reitemeier et al.	2006	190	7	7	97.3%
de Backer	2006	1037	10	116	94,6%
Marklund et al	2003	42	5	3	92,4%
Walton	1999	347	7.1	12	97.1%
Jokstad & Mjor	1996	43	10	0	100%

2.6. Adhesion measurement of bond strength:

2.6.1. Shear bond strength test:

This test is defined as two materials are connected through an adhesion agent and putted in shear until failure of bond happens (96). This bond strength is calculated by dividing the maximum applied force leads to failure of the bond by the bonded cross-sectional area (97). Shear bond strength testing is the most widely used as test for bond strength, due to its simple to use, clear test protocol, and rapid production of test results (98). In addition, this test simulates shear stresses, which are considered major causes of bonding failure of restorative materials in the oral cavity (99). However, the shear bond strength tests are the subject of criticism. The biggest disadvantage of shear bond strength tests is that they require large specimen

sizes, which in the case of ceramic materials may result in increased structural flaws that may lead to premature failure of the test specimen before the maximum bond strength levels are reached (100).

2.6.2. Tensile bond strength

Tensile bond strength test employed for measure the quality of bond between metal and ceramic comprises of porcelain attached to one end of a metal rod or among the two terminals of metal rods. Longitudinal tensile force is employed on the long axis of the rod or rods to break porcelain from metal (101). The bond strength is measured by dividing the axial load at bond failure by the cross-sectional bonded area (102). The results of this test is extremely effected by sample geometry and the occurring of non-uniform stress distributions trough out load application (103). Just this test may not give actual data about bond strength of materials (100). Hence, cohesive failure of porcelain, not interfacial bond strength, was actually evaluated (101).

2.6.3. Flexural bond strength test:

Due to a lots of problems related to direct tension testing of brittle materials, according to that fact the bending or flexure tests have been the most widely used test in the ceramics world and are commonly believed to be the most acceptable way for measuring strength (104). Flexural strength can be measured by to ways the first way is a three-point flexure test, and second way is the four-point flexure test. In both ways of the test, elevated load is employed until fracture happens (105).

2.6.4. Three-point bending test:

This test is an uncomplicated, dependable, and sensitive technique for testing strength of ceramics used in dentistry (104). The international standards organization (DIN ISO 9693: 19999) encourage the utilize of the three-point bending test as a Schwickerath crack initiation test for measuring the bond strength among porcelain and metal alloys (106). Kosyfaki 2011 utilized the Schwickerath crack initiation test for measuring the bond strength related to the veneering ceramics to zirconia cores (107).

2.6.5. Four-point bending test:

The test configuration was first introduced by Charalambides et.al in 1989 to determine the adhesion tow material interface (108). This test has been employed to measure the interfacial fracture toughness in terms of strain- energy release rate (G) of different metal-ceramic bonding systems (109,110,111). The bending test is also uncomplicated to perform experimentally. However, there were some technical problems in the sample preparation and producing the pre-crack, which are mentioned for others hoping to employ this method (112).

2.7. Shear bond strength VS Micro tensile bond strength:

Many laboratory tests have been used for measure the quality of bond of zirconia to ceramics. These techniques involve tensile, microtensile, shear bond strength and pull out tests. Shear bond strength test was preferred for this experiment due to it is a commonly employed test and has confirmed to be dependable (113). Literature has shown that macro-shear testing was the most commonly used method of testing only because it does not need furthermore sample processing of fully sintered zirconia when the bonding process is finished. But the mean predicted shear bond strength was lower than that of any of the three other tests, shear bond strength testing has given that non-uniform interfacial stresses may outcome in cohesive failures in the bonding infrastructure, leading to misunderstanding of the result. Moreover, stress concentrations near the loading site lower the calculated shear bond strength below the true failure stress levels (114). Valandro et al. found a higher percentage of adhesive failures when they compared shear to micro-tensile bond strength between the resin cement and high strength ceramics. However, the authors did not observe significant differences in the bond strength results from both tests (100).

3. Materials and Methods

In our study we used 100 disc-shaped porcelain substructure with 10 mm diameter and 3mm thickness .50 of the discs were produced from Nickel Chromium metal alloy (NI-Cr) and 50 from Zirconia (Zr).

This disk was prepared in the pre-sintered blocks using cad/cam system (Cortitec T 350i loader, imes-icore, Germany) and sintered to the final required dimension (10mm in diameter and 3mm thick) in a special high temperature furnace. The specimen dimensions have been produced according to ISO standards for dental ceramics (ISO 6872, 2008).
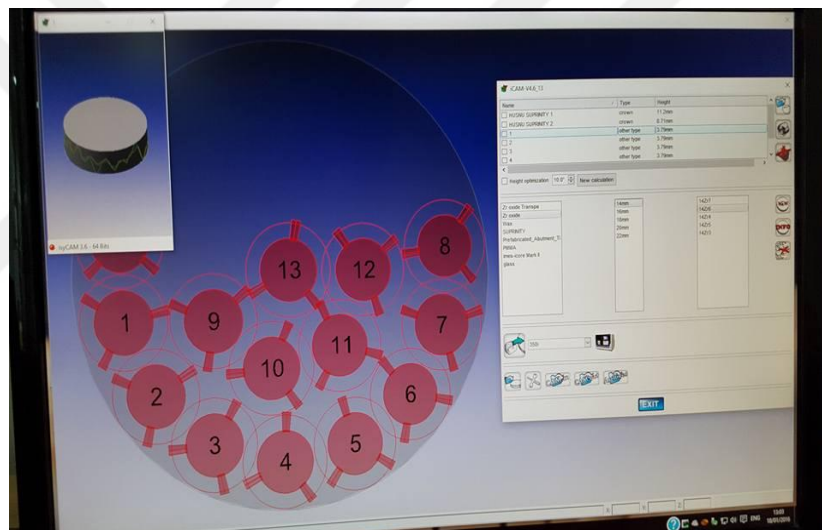
Figure (4) design of zirconium disks with computer.
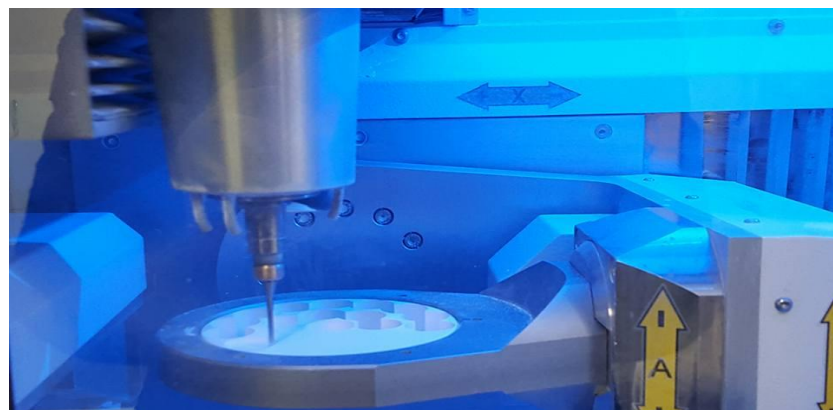
Figure (5) milling of zirconium disk from presintred zirconium blocks
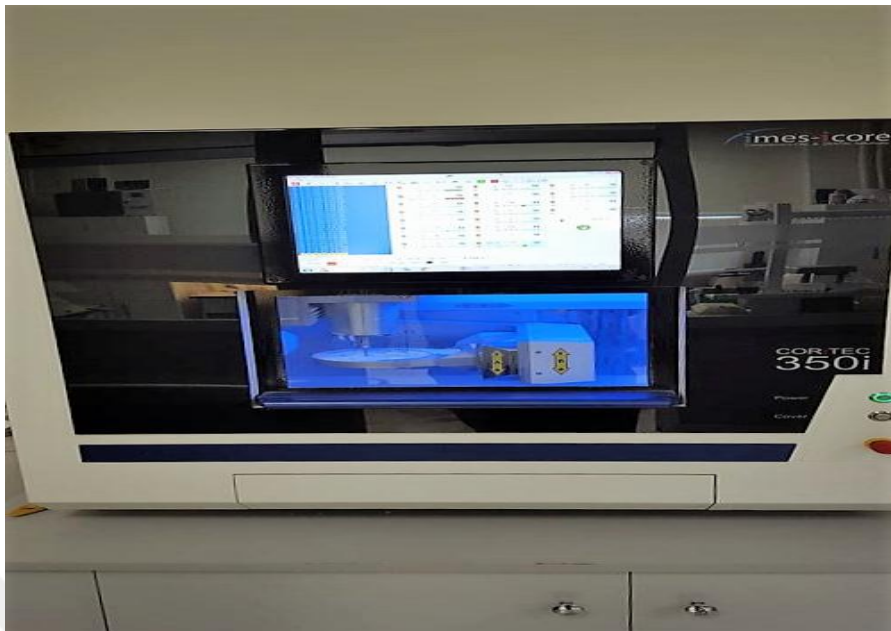
Figure (6) cad cam machine

Figure (7) Special oven used for sintering zirconium disks.

Figure (8) 50 zirconium disk after sintering.

The wax disks were fabricated using cad/cam system from prefabricated wax blocks after that, the wax patterns were made as per the dimensions of metal die using crown wax (Kronenwachs, Bego Germany). Sprue was attached and placed in a silicon crucible former and invested using phosphate bonded investment material (Bellasun, Bego Germany) and casting was done using Ni-Cr alloy in the induction casting machine.
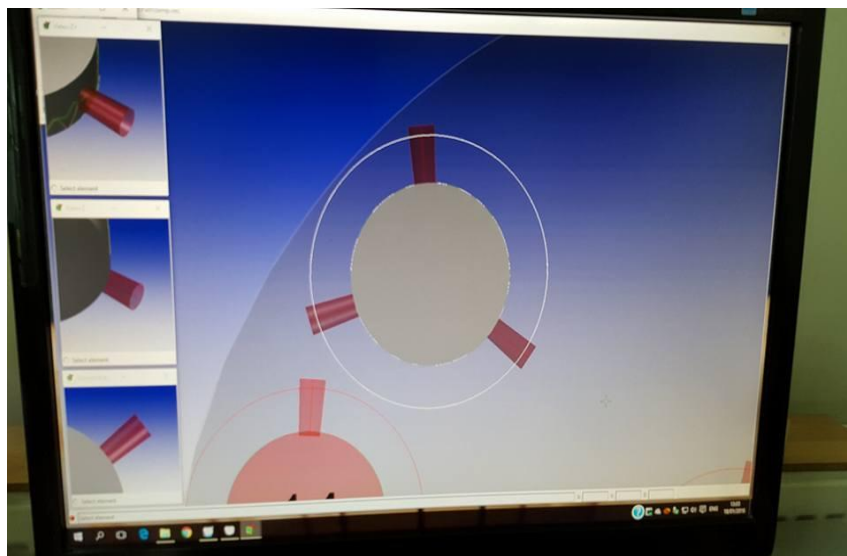
Figure (9) design of wax disk with cad cam system.
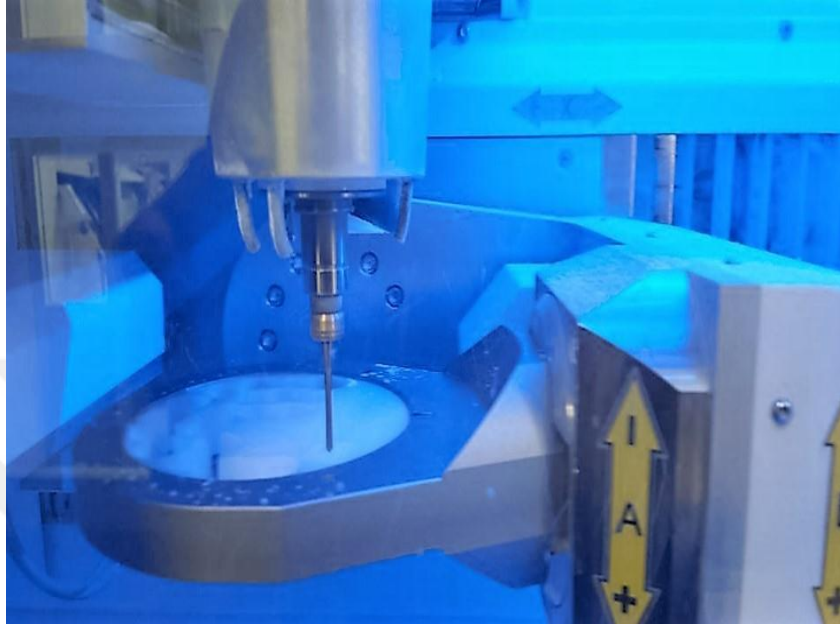
Figure (10) cutting of wax disk with cad cam system

Figure (11) attached the sprue to wax disk

Figure (12) Aattach sprued wax disks to crucible former.

F**igure (13) Mechanical mixing of phosphate bonded investment material**

Figure (14) Special oven used for molten wax disk.

Figure (15) Induction casting machine.

Figure (16) 50 nickle chromium disks

All discs were polished by a polishing machine (Tegrapol-11; Struers, Ballerup, Germany) with wet silicon carbide paper, grinding with 600- and 1000-grit under cool water for about 1 minute.

The 100 disk divided to two groups 50 Zr and 50 Ni-Cr, after that each group divided to 5 subgroups according to the 5 repair systems

Group (1): Zirconium discs repaired with clearfil repair system.

Group (2): Zirconium discs repaired with bisco repair system.

Group (3): Zirconium discs repaired with ultradent repair system.

Group (4): Zirconium discs repaired with single bond.

Group (5): Zirconium discs repaired with ivoclar repair system.

Group (6): Nickle chromium discs repaired with clearfil repair system.

Group (7): Nickle chromium discs repaired with bisco repair system.

Group (8): Nickle chromium discs repaired with ultradent repair system.

Group (9): Nickle chromium discs repaired with single bond.

Group (10): Nickle chromium discs repaired with ivoclar repair system.
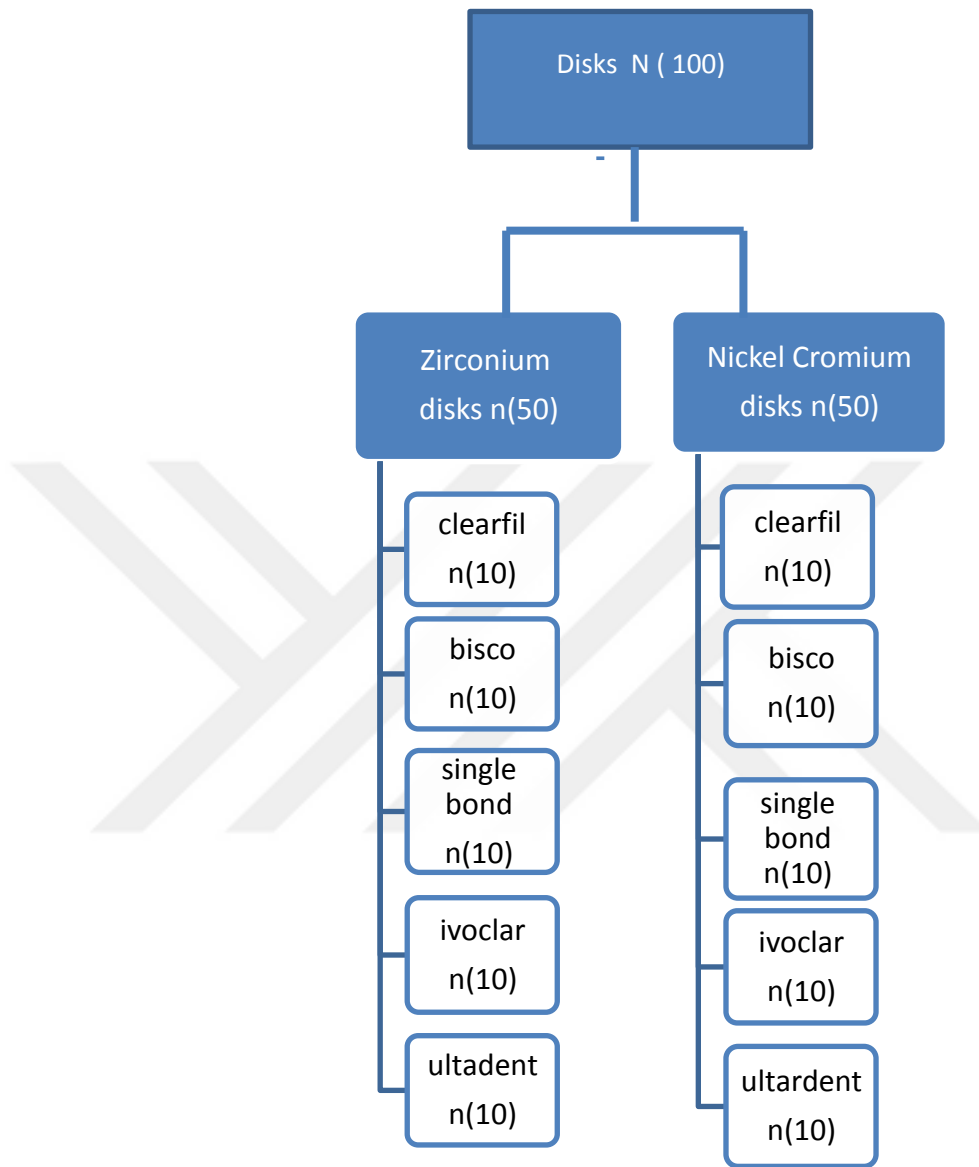
Figure)17) Flow-chart showing experimental groups

Table (4) Material utilized in our experiment

Brand	Manufacturer	LOT Number
Zirconium	H.C.STARCK GERMANY	50574292 50575967
Nickle chromium alloy	Eisenbacher Dental Waren Edgmbh GERMANY	H14-8
Clearfill Repair Kit (CL)	Kuraray Medical Inc, Okayama, Japan	000016
Ultradent Repair Kit (UL)	Ultradent PRODUCTION,INC. USA	BBFC4
Ceramic Repair system Kit	Ivoclar vivadent, Liechtenstein	T42712
Intra Oral Repair KIT	Bisco,Inc USA	
Single Bond Universal adhesive	3M ESPE, St Paul, Minn., U.S.A.	604724
Filtek Z250 (shade C2)	3M ESPE, St Paul, Minn., U.S.A.	(3)N566178.(3)N545065 (3)N507264.(3)N616236

All zirconium and nickle chromium discs was treated by airborne-particle abrasion with alumina (sandblasting) with 50 μm particle size aluminum oxide for 10 seconds at a pressure of 40 psi and from distance about 10 m

Figure (18) Sand blasting machine used in this study

Table (5): Repairing procedures with clearfil repair system for zirconium and nickel chromium disks.

Repair system	Application procedures
Clearfil for zirconium disk	1-Apply the mixture of clrarfil se bond and clearfil percaline bond on zirconium disk for 5 second, after that dried with mild air. 2. Use the bonding agent after that air dried softly, and photo-polymerization for10seconds.
Clearfil for nickle chromium disks	1-Applay alloy primer and dry. 2-Applay the mixture of clrarfil se bond and clearfil percaline bond on zirconium disk for 5 second, after that dry with mild air. 3. The bonding agent, air drying gently, and photo-polymerization for10 seconds. 4-Applay clearfil st opaquer and light cured for 40 second

Figure (19) clearfil repair system

Table (6): Repairing procedures with bisco repair system for zirconium and nickle chrome disks.

Repair system (bisco)	Application procedures
Repair for zirconium disk	1. put1-2 coats of Z Prime after that use the air syringe to dry the surface for 3-5 second. 2.put porcelain bonding resin after light cured for about 10 seconds
Repair for nickle chromium disks	1-. put 1-2 coats of Z Prime and dried with an air syringe for 3-5 second 2-shake OPAQUER base and catalyst well before dispening. dispene one drop of base and catalyst onto amixing pad,and mix with a brush tip .apply a thin coat of opaquer onl to metal surface.and light cure for 5 second 3-. Put porcelain bonding resin after that light cured (for 10 seconds).

Figure (20) Bisco intra oral repair system

Table (7) Repairing procedures with ceramic repair system (ivoclaer vivadent) for zirconium and nickle chrome disks:

Repair system (ivoclar)	Application procedures
Repair for zirconium disk	. 1-By using the brush to spread the Monobond Plus on the exposed zirconium and left to react for 60 seconds. after that dryied with oil free air. 2-Spread a thin layer of Heliobond to the surface area which require to be repaired. Take out any excess with compressed air. Then light cure for 10 second.
Repair for nickle chromium disks	1-By using the brush brush to spread the Monobond Plus to the exposed nickle chromium and left to react for 60 seconds. after that died with oil free air. 2-Spread a thin layer of Heliobond to the entire surface area which requires to be repaired. Take out any excess with compressed air. then light cure for 10 second 3- Put IPS Empress Direct Opaque, Then, light-cure for 20 seconds

Figure (21) ceramic repair system (ivoclar vivadent)

Table (8) Repairing   procedures with ultradent ceramic repair system for zirconium and nickle chromium disks:

Repair system (ultra dent)	Application procedures
Repair for zirconium disk	1-Apply Hydrofluoric acid on metal surface for 90 second rainse and dry the surface. 2-Spred the Silane over the zirconium disc and leave it to evaporate for about one minute. After one minute,we can dry it used air spray if needed. 3- .Apply a puddle coat of Peak® Universal Bond with Inspiral Brush – softly spread it for about 10 sec. 5.Dryness can be done by air syringe for about 10 seconds using ¼ to ½ air pressure, Prep should appear shiny, after that the surface cured with light for about 10 second.
Repair for nickle chromium disks	1-Apply Hydrofluoric acid on metal surface for 90 second. rainse and dry the surface. 2-Spread the s Silane and leave it to evaporate for about one minute. After one minute, can be dried with air if needed. 3-Spread a puddle coat of Peak Universal Bond with Inspiral Brush –softly agitate for about 10 sec. 5. Dryness can be done for10 seconds using ¼ to ½ air pressure, the surface should appear shiny, after that should be cured with light for 10 second. 4-Exposed metal, put a thin layer from PermaFlo Dentin Opaquer over metal disks.

Figure (22) Ultra dent porcelain repair system

Table (9) Repairing procedure with single bond adhesive for zirconium and nickle chromium disk:

Repairing procedures with single bond	Application procedures
For zirconium and nickle chromium disk	1-Apaly on surface of zirconium and nickle chromium disks with brush and leave it for 20 second 2-Dry with air foe 5 second 3-Light cure for 10 second

Fıgure (23) Single bond universal adhesive

Repairing was done used hybrid composite resin and it incrementally packed with hand instrument on the surface treated zirconia and nickle chromium discs using a specially designed epoxy glass mold and light cured incrementally in three layers not excessed than 2 mm. Each layer was light cured with a halogen photo polymerization unit for about twenty second**.**

The bonding process was done by the one operator during the experiments. After polymerization repaired samples assembly were removed from the mold and the adhesive interface light curing once again from the five aspects of the block (upper and lateral) for about 20 s per side.
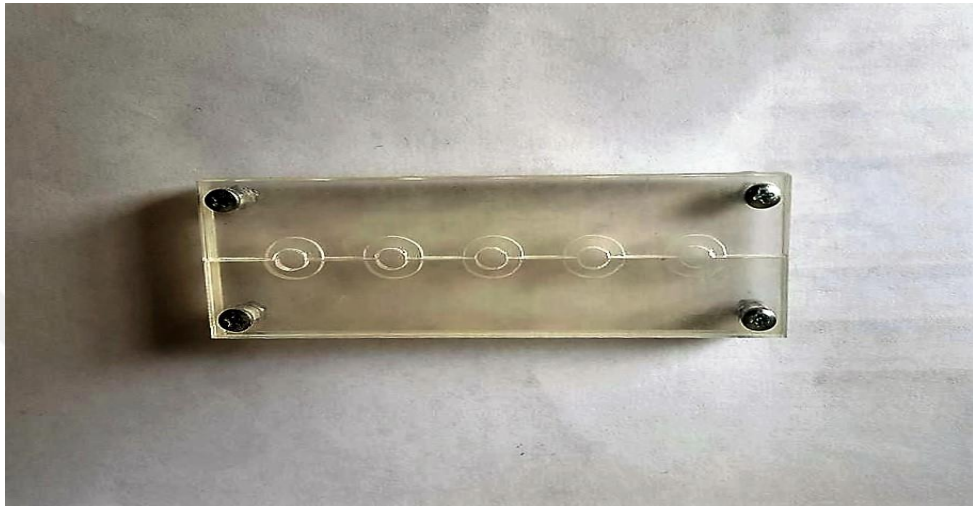
Figure (24) Epoxy glass mold used for repairing disk with composite

Figure (25) 3M (Filtick) Z250 hybrid composite

The samples were put in distilled water at 37°C for one day and then subjected to the thermal cycling (1200 cycles, between 5°C and 55°C, with a dwell time of 20 seconds at each temperature, and a transfer time from one bath to the others of 10 seconds).

Figure (26) Thermocycling machine

All samples were fixed by chemical cured acrylic resin in a steel mold after that all samples are ready for the shear bond test.

Figure (27) Fixed all Ni-Cr disc in steel mod using acrylic resin
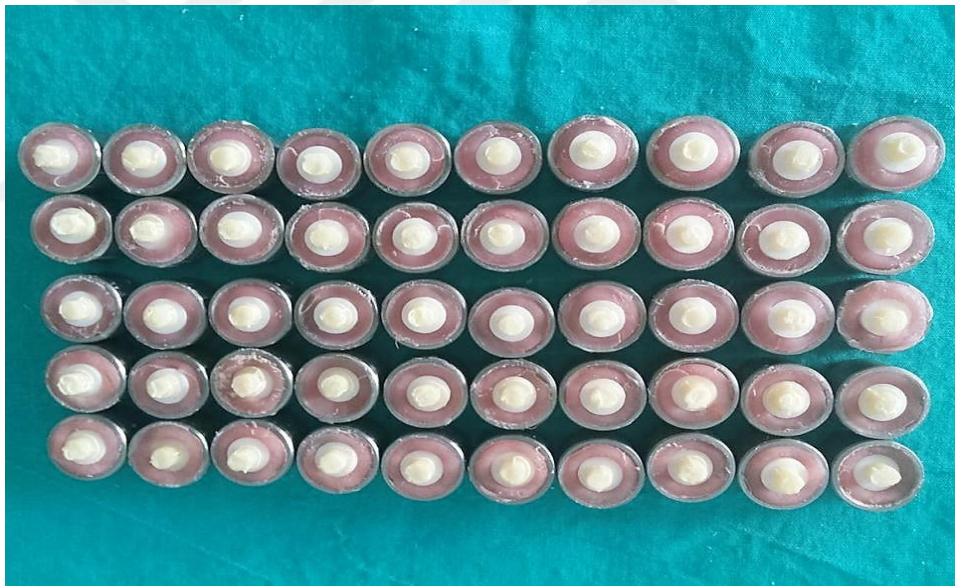
Figure (28) Fixed all Zr disc in steel mod using acrylic resin

Shear bond testing of all groups was carried out by using of a universal testing machine at a crosshead speed of 1mm/min. A knife-edge blade apparatus was employed to direction parallel to shearing force as near as possible to the interface between the zirconia and the composite, nickel chromium and composite disc. The

shear debonding forces were registered in Newton (N). The failure loads (N) were divided by the bonding areas (mm2), and then the shear debonding forces were changed into MPa.
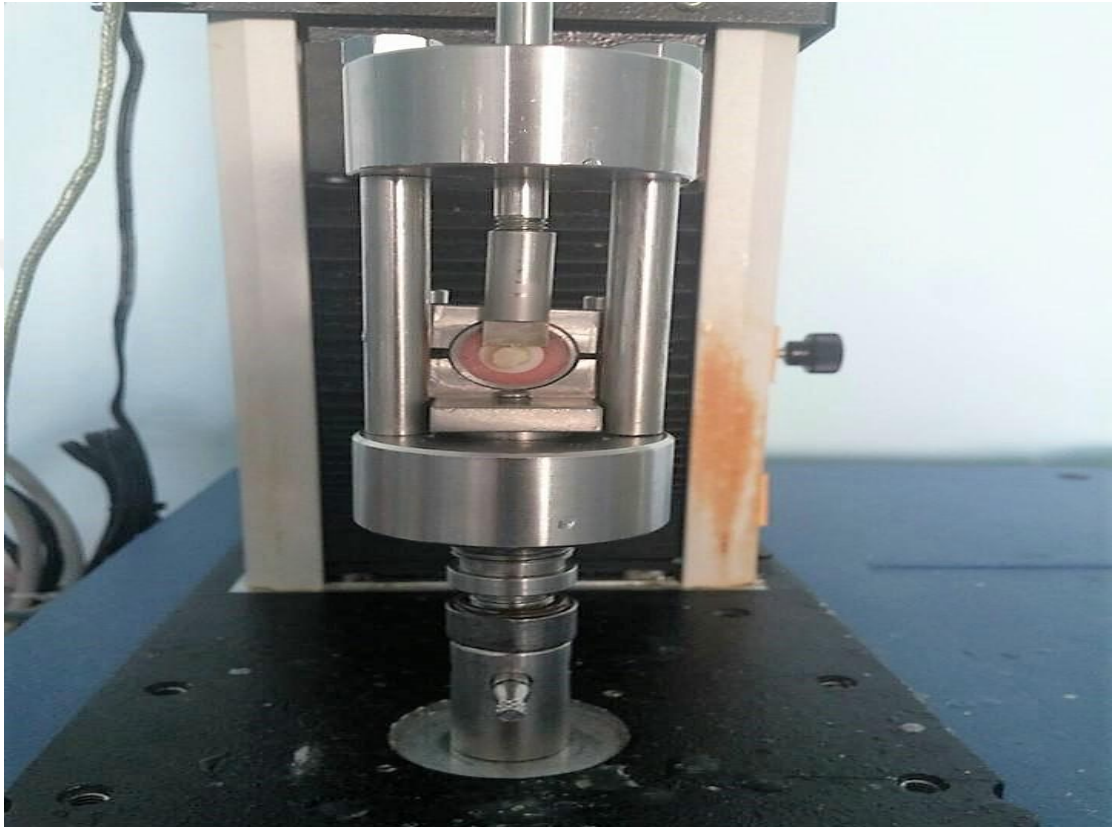
Figure (29) Universal testing machine

The failure of bond related to the fractured surfaces were visually inspected using a stereomicroscope at 20 magnifications to check the dominant failure mode classification areas that presented a resin-free zirconium nickel chromium discs' surfaces were grouped as adhesively failed portions, and areas that were covered by resin composite were classified as cohesively failed portions the areas containing both adhesive and cohesive phases were classified as mixed failure.

Figure (30) Stereomicroscope

4. Statistical method

Statistical calculations were performed with (Number Cruncher Statistical System) 2007 Statistical Software (Utah, USA) program for Windows. Besides standard descriptive statistical calculations (mean and standard deviation), one-way ANOVA was used in the comparison of groups, post Hoc Tukey multiple comparison test was utilized in the comparison of subgroups, Unpaired t test was used in the comparison of two groups of Shear bond strength values, and Chi square test was performed during the evaluation qualitative data. Statistical significance level was established at p<0,05.

5. Results:

Table (10) Shear bond strengths (MPa) for all groups.

	Repair system	Minimum	Maximum	Mean	Std. Deviation
Zirconium	Single bond adhesive	2,73	4,80	3,63	0,62
	Bisco repair system	14,29	27,65	18,91	4,33
	Clearfil repair system	9,33	28,37	18,61	5,37
	Ceramic repiar system ivoclar	8,76	23,51	15,24	5,30
	Ultradent repair system	4,99	9,87	6,63	1,50
	Single bond adhesive	5,64	15,21	10,33	3,55
	Bisco repair system	16,92	24,89	20,93	2,63
	Clearfil repair system	10,17	26,90	17,37	4,46
	Ceramic repiar system ivoclar	10,62	29,50	16,13	5,94
Ni-Cr	Ultradent repair system	6,75	13,71	8,17	2,48

The mean SBS values for zirconium group repaired with clearfil intra oral repair kits was 18,61, zirconium group repaired with bisco repair system was18,91, zirconium group repaired with ivoclar repair system was 15,24 , zirconium group repaired with ultradent was 6,63 and for zirconium group repaired with Single bond adhesive was 3,63.

The mean SBS values for Ni-C r group repaired with clearfil intra oral repair kits was 17,37 Ni-Cr group repaired with bisco repair system was 20,93, Ni-Cr group

repaired with ivoclar repair system was 16,13, Ni-Cr repaired with ultradent was 8,17 and for Ni-Cr repaired with Single bond adhesive was 10,33.

Table (11) Comparison shear bond strengths between the groups (MPa)

Repair system	Zr n:10	Ni-Cr n:10	P*
Single bond adhesive	3,63±0,62	10,33±3,55	0,0001
Bisco repair system	18,91±4,33	20,93±2,63	0,225
Clearfil repair system	18,61±5,37	17,37±4,46	0,581
Ceramic repiar system ivoclar	15,24±5,3	16,13±5,94	0,729
Ultradent repair system	6,63±1,5	8,17±2,48	0,111
p+	0,0001	0,0001	

*Unpaired t test +One-Way ANOVA

There is a significant difference between shear bond strength of two nickle chromium and zirconium groups repaired with single bond adhesive (p=0,0001). There is no significant difference between shear bond strength of two nickle chromium and zirconium groups repaired with bisco repair system (P=0,225). There is no significant difference between shear bond strength of two nickle chromium and zirconium groups repaired clerafil repair system (p=0,581). There is no significant difference between shear bond strength of two nickle chromium and zirconium groups repaired ultadent repair system(p=0,111). There is no significant difference between shear bond strength of two nickle chromium and zirconium groups repaired ivoclar vivadent repair system(p=0,729).

Table (12) Comparison shear bond strengths between the repair system

Tukey Multiple Comparisons Test	Zirconium	Ni-Cr
Single bond adhesive / Bisco repair system	**0,0001**	**0,0001**
Single bond adhesive / Clearfil repair system	**0,0001**	**0,003**
Single bond adhesive / Ceramic repiar system ivoclar	**0,0001**	**0,019**
Single bond adhesive / Ultradent repair system	0,447	0,750
Bisco repair system / Clearfil repair system	0,999	0,293
Bisco repair system / Ceramic repiar system ivoclar	0,249	0,075
Bisco repair system / Ultradent repair system	**0,0001**	**0,0001**
Clearfil repair system / Ceramic repiar system ivoclar	0,331	0,958
Clearfil repair system / Ultradent repair system	**0,0001**	**0,0001**
Ceramic repiar system ivoclar / Ultradent repair system	**0,0001**	**0,0001**

There is a significant difference between shear bond strength of two repaired zirconium groups with bisco repair system and single bond adhesive system (p=0,0001). There is a significant difference between shear bond strength of two repaired Ni-Cr groups with bisco repair system and single bond adhesive (p=0,0001). There is a significant difference between shear bond strength of two repaired zirconium groups with single bond adhesive and clearfil repair system(p=0,0001).

There is no significant difference between shear bond strength of two repaired Ni-Cr groups with with single bond adhesive and clearfil repair system (p=0,003). There is a significant difference between shear bond strength of two repaired zirconium groups with single bond adhesive and ceramic repair system ivoclar (p=0,0001). There is no significant difference between shear bond strength of two repaired Ni-Cr with single bond adhesive and ceramic repair system ivoclar (p=0,019). There is no significant difference between shear bond strength of two repaired zirconium groups with single bond adhesive and ultradent repair system (p=0,447). There is no significant difference between shear bond strength of two repaired Ni-Cr with single bond adhesive and ultradent repair system (p=0,750). There is no significant difference between shear bond strength of two repaired zirconium groups with bisco repair system and clearfil repair system (p=0,999). There is no significant difference between shear bond strength of two repaired Ni-Cr groups with bisco repair system and clearfil repair system(p=0,293). There is no significant difference between shear bond strength of two repaired zirconium groups with bisco repair system and ceramic repair system ivoclar (p= 0,249). There is no significant difference between shear bond strength of two repaired Ni-Cr groups with bisco repair system and ceramic repair system ivoclar (p= 0,075). There is a significant difference between shear bond strength of two repaired zirconium groups with bisco repair system and ultadent repair system (p= 0,0001). There is a significant difference between shear bond strength of two repaired Ni-Cr groups with bisco repair system and ultadent repair system (p= 0,0001). There is no significant difference between shear bond strength of two repaired zirconium groups with clearfil and ceramic repair system ivoclar (p=0,331). There is no significant difference between shear bond strength of two repaired Ni-Cr groups with clearfil and ceramic repair system ivoclar (p=0,958). There is a significant difference between shear bond strength of two repaired zirconium groups with clearfil and ultradent repair system (p=0,0001). There is a significant difference between shear bond strength of two repaired Ni-Cr groups with clearfil and ultradent repair system (p=0,0001). There is a significant difference between shear bond strength of two repaired zirconium groups with ceramic repair system ivoclar and ultradent repair system (p=0,0001). There is a significant difference between shear bond strength of two repaired zirconium groups with

ceramic repair system ivoclar and ultradent repair system (p=0,0001). There is a significant difference between shear bond strength of two repaired Ni-Cr groups with ceramic repair system ivoclar and ultradent repair system (p=0,0001).

Figure (31) Bar-chart for Shear bond strength values of all groups

Table (13): Bond failure between the groups.

		Zirconium		Ni-Cr		Total		p
Single bond adhesive	**Adhesive Failure**	10	100,00%	10	100,00%	20	100,00%	
Ultradent repair system	**Adhesive Failure**	10	100,00%	10	100,00%	20	100,00%	
Clearfil repair system	**Adhesive Failure**	6	60,00%	7	70,00%	13	65,00%	
	Mixed Failure	4	40,00%	3	30,00%	7	35,00%	0,500
Altra Dent repair system	**Adhesive Failure**	3	30,00%	7	70,00%	10	50,00%	
	Mixed Failure	7	70,00%	3	30,00%	10	50,00%	0,089
Ceramic repiar system ivoclar	**Adhesive Failure**	6	60,00%	7	70,00%	13	65,00%	
	Mixed Failure	4	40,00%	3	30,00%	7	35,00%	0,500

There is no significant difference between bond failure of two repaired zirconium and groups with bisco repair system.

There is no significant difference between bond failure of two repaired zirconium groups with clearfil repair system. There is no significant difference between bond failure of two repaired zirconium and groups with ceramic repair system ivoclar.
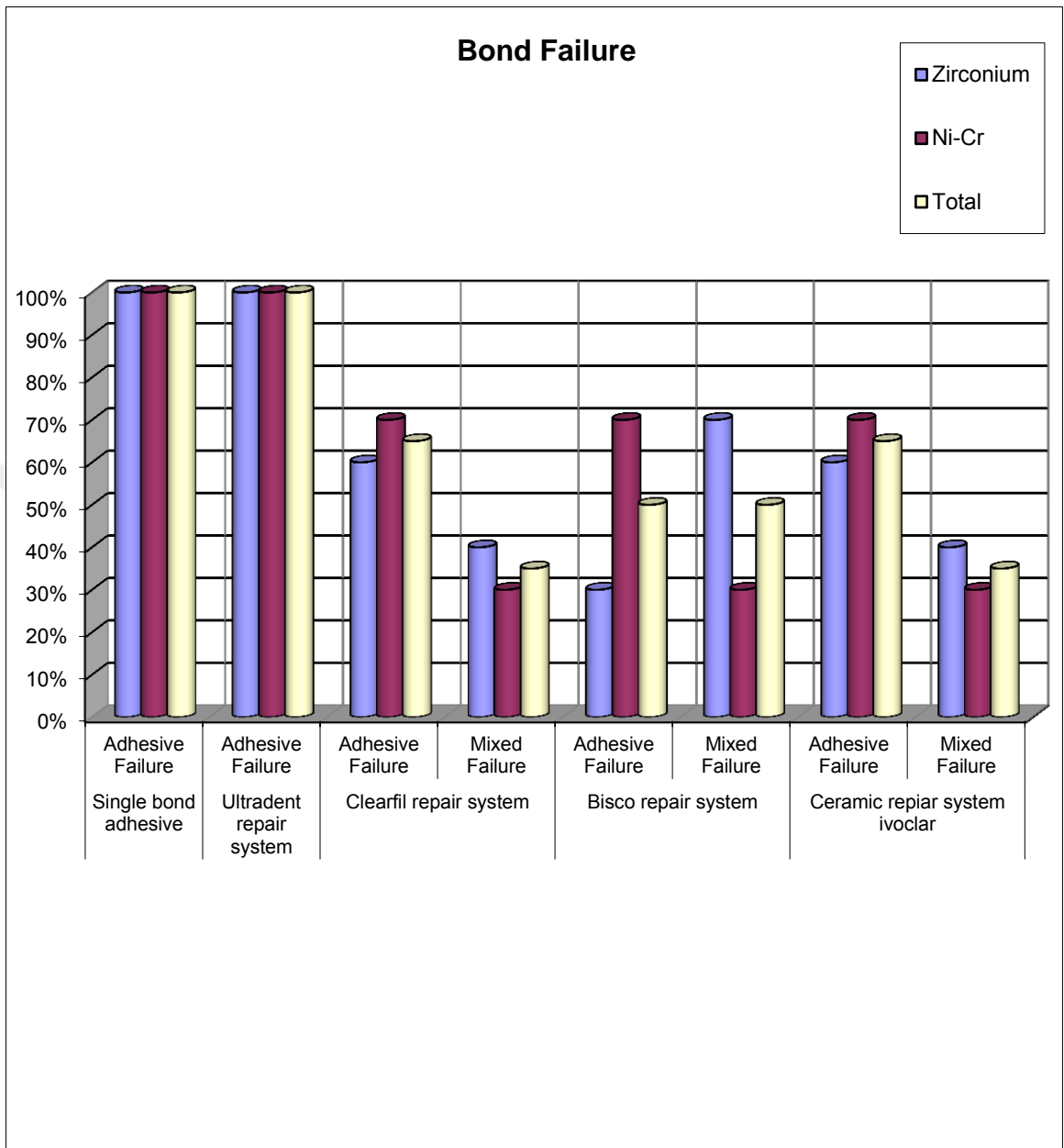
Figure (32) Bar-chart for bond mode failure of all groups .

4. Discussion:

Porcelain fused metal restorations are considered as a good optional procedure in fixed prosthodontics not only because of mechanical strength, as well as esthetic qualities imparted by porcelain material. The modern day dental practices utilize base metal alloys as a metal infrastructure (115). Also metal ceramic fixed restorations usually employed in fixed prosthodontics due to their superior biocompatibility and excellent esthetic characters (79). Porcelain failures have been often reported due to fracture of either as material itself or exposing the metal substrate or completely debonding exposing of porcelain (116).

Porcelain material veneered to metal coping has the ability to fracture because of many factors including, impact force and fatigue load, high occlusal forces, incompatible thermal expansion coefficients between the porcelain and metal infrastructure, employ of metal alloys with low modulus of elasticity, seating force throughout testing of insertion or cementation, wrong tooth preparation especially cervical part improper design, irregular laboratory work, microdefects within the material, and trauma (117,118).These fractures may be either as fracture in porcelain (cohesive fracture) only or fracture with both porcelain and metal exposed or fracture with substantial metal exposure (adhesive fracture) (116).

Coornaert et al. (1984), revealed that the fractures in metal ceramic crowns was found to be about 5% in 10years of function (116), Karlsson (1986) mentioned that a 93% success rate for fixed bridge restorations made from metal ceramic during a 10-year period (120). (Kerschbaum, Seth & Teeuwen, 1997), they reviewed, 1219 three- unit fixed bridges and 1618 single crowns in the anterior region, the result showed that after 10 years in function, 88,7 %of the metal–ceramic crowns and 80,2% of the metal ceramic bridges were survived (121).

Zirconia is the best and most favorite dental material used in dentistry due to good mechanical properties such as elevated fracture toughness and natural appearance (122, 123). Thus, zirconia has wide clinical uses, particularly as framework in anterior and posterior region. The adequate bonding possibility

between_veneering porcelain and zirconia framework is essential for long term functioning of all restorations. Nevertheless, some authors proved that the failure of zirconia restoration occurs almost among zirconia and veneering porcelain (124,125). Some authors mentioned that the fracture rates of veneer ceramics were 13% after 3 years in function (126). Some authors proved that the fracture rates of 25% of zirconia chipping of the veneer ceramics after 31 months (124). The classification of chipping of porcelain-veneered zirconia restorations. According to Heintze and Roussont it can be grouped by intensity and the treatment demanded for repair to three grades. Grade 1 describe the small surface chipping and it is treatment done by polishing the restoration surface, Grade describe the moderate surface chipping and it is treatment done by repairing the restoration with resin composite repair system, Grade 3: Severe veneer ceramic chipping exposing the zirconia core and it is treatment done by remake the restoration (42). Factors that leads to enhance the hazard of chipping of porcelain-veneered zirconia-based restorations are: Residual stress induced by incompatibility in the coefficient of thermal expansion (CTE) between the zirconia framework and the porcelain used for veneer, Poor zirconia framework wettability by the porcelain used for veneer, which consequence of inadequate engagement among zirconia and porcelain and inadequate micromechanical interlocking between zirconia framework and porcelain used for veneer , inadequate zirconia infrastructure wettability with the porcelain veneered which consequence of inadequate engagement between materials and poor micromechanical interlocking and fabrication defects (42).

Intraoral repair of the chipped and fractured porcelain used as a veneer with composite restoration very notable method for repair. More novel product, adhesive systems require various treatment steps and for porcelain repair with composite resins. Resin composite is the substitute material used for repairing ceramic restorations because this method is easy and inexpensive compared to other methods (127). The resin composite used in this study can be used for anterior and posterior teeth and for ceramic repair with porcelain repair kits (128,129). Repairing dental restorations must be selected over remaking a new one, because it is cheaper than remaking. Repair of fractured and chipped porcelain on the crown by surface

treatments has89%, after a 3-year of success producing this an satisfactory selection for proper cases (130). For this reason we choose intra oral repair technique with composite for our study to compare different repair system and their effects on infra structure.

There are various ceramic repair systems usable on the dental store that are established on different conditioning mechanism, it is harder for dentists to choose the better that give good results. In our research, five repair systems established on various conditioning ideas were chosen. Adhesion basics nowadays commonly depended on the union of physical and chemical bonding.

Composite resins are normally applied for the repair chipping and fracture porcelain, in working order to resist the functional force, the bond between the composite as repair material and the chipping veneered framework must be adequate. The repair material must have a minimum coefficient of thermal expansion and minimum polymerization shrinkage. Gregory and Moss,1990 established that using of larger particle size composite resins or hybrid composite resins as repair gives a high bond strength than using microfilled composite resins for repairing chipped porcelain (128). (Lutz & Phillips, 1983) recommended to of the hybrid composite resins for repairing purposes (132). Both studies done by Simonsen and Stangel proved that using of hybrid composite for repairing chipped porcelain leads to increase bond strength and decrease stress compared with using a microfilled type of composite resin (133,134) . Wearing problems and surface changes are related to use of the microfilled composite resin which could be decreased by using a hybrid composite resin, it is also commended to be applied where fatigue loading is of consideration (Creugers et al., 1992; Llobell et al.,1992) (135,136). For these reasons we choose hybrid composite as repair material four our study.

Erdemir et al. 2014 proved that using self-adhering flowable composite resin as repair for chipped porcelain gives weak bond strength irrespective of the surface treatment is used (137). Capaet et al. proved that flowable composite does not appear to increase the bond strength of intra-oral ceramic repair systems (138). For this reason, we don't use flowable composite and for standardizations we use hybrid

composite. Narasimha Jayanthi and V. Vinod they used glass molds for fabrication of cylindrical composite sample (139). In our study we also used a similar glass molds for repairing procedure.

Airborne-particle abrasion (APA) can be used as a surface treatment for metal alloys and ceramics (140) also can be uses as surface treatment for hard dental tissues (enamel dentin) (141) and also can be used for roughening the surface of zirconia for intensifying mechanical interlock and increase the contact area needed for bonding (142,143). The various parameters in airborne-particle abrasion with alumina the first parameter is the grain size which starts from 25 to 250 mm, the second parameter is the propulsion pressure starts from 0.05 to 0.45 MPa, the third parameter is the distance from the nozzle to the specimen is ranged from 5 to 20 mm and fourth parameter is the time of parameters in airborne-particle abrasion is ranged from 5 to 30 seconds (144.145).

Treatment of zirconia surfaces by air abrasion to create the micromechanical retention can be done by small grains ranged from 25 mm, 50 mm or larger grains (110 mm) and was not significantly different between the two sizes (146.147). Despite the different surface roughness produced (148). Airborne-particle abrasion is used for maximize surface energy and decrease the organic contamination. The application of airborne-particle bring up two principal concerns: the ability of production of microcracks on the surface after completing the procedure and the development of phase transformation from tetragonal crystal to monoclinic crystal (t/m) at the surface and subsurface, which leads to reducing the mechanical properties of the material (149,150) ,to prevent that the manufacturers advice heating of zirconia after airborne-particle abrasion to reverse the (m/t) conversion (151) or applying of airborne-particle abrasion before the final sintering procedure (152.153). Some manufacturers advice to application APA with alumina grain not more than 50 mm (154,155). Significant phase conversion (t/m) appears to be caused by aggressive APA increasing the monoclinic phase (156). In our study we used alumina particle 50 mm.

Sandblasting gives micromechanical retention. Airborne-particle abrasion treatment when performed on the alloy surface, microscope was showed cleaned and roughened surface with adequate wetting by resins and also gives stronger composite-alloy bonds (157).

Some authors believed that using grooves or undercuts as mechanical retention to hold the composite with porcelain or metal alloys. Leads to microleakage and finally weak bond due to humidity of intra-oral conditions. Wood et al., proved that the employ of fine and coarse diamond burs raises crack development and spreading through the ceramic which leads to failure of bond (158). Because of that we didn't choose the bur as surface treatment of our thesis.

Because zirconia is too hard it need to grinding with coarse diamond bur which it is grain size range from 120 to 200 mm (159). Although the roughness with a coarse-grained diamond burs had been tested, giving a rougher surface more than other techniques used for a surface treatment and also leads to improving bond strength but was not satisfactory due to it is a very aggressive method leads to make microcracks and produce damage to zirconia surfaces (160,161). Grinding circumstances also create the impression that grinding with a 91-mm diamond wheel in wet environment did not seriously decrease the flexural strength (162). Testing of grinding on three different zirconia materials with 100-mm diamond rotary instruments demonstrates that only one type of zirconia it roughness has been elevated significantly (163).

The uses hydrofluoric acid by percentage ranged from 2.5 to 10 for about 60 seconds it is not difficult way to treat the fractured surface as chairside method; nevertheless, it is used only to treat silicate based dental ceramic. Etching is not effective to use with metal or oxide dental ceramics and has a low silicate content (< 15 percent volume) due to no currently available acid is able to break down the metallic bonds and also destroy the very strong bonds of oxide based dental ceramics (162) . There is no significant elevation in the bond strength between zirconia and resin cements after using (HF) as surface treatment (164,165). Same process employed for etching the metal wings of resin bonded bridge was tested on zirconia

surface and it produces a rougher surface. An experimental hot hydrochloric acid (HCl) solution significantly elevated the roughness of zirconia surface (166,167). Matsumara et al, proved that using of acid as surface treatment is only used in dental practice, just for removing the smears from the ceramic surface (168). Some authors proved that using hydrofluoric acid for 5 minutes leads to increase incidence of cohesive failures (169). The survival of bond formed between composite and ceramic with chemical agents as surface treatment was considerably lower than that roughened with aluminium oxide (Al2O3) air abrasion, or a combination with hydrofluoric acid (170). The risks of using hydrofluoric acid as surface treatment are very clear. Even its very effective method for surface treatment, hydrofluoric acid produces inexorable damage to human tissue and it is advised to found more reasonable repair alternative methods (171). Creugers et al studied the influence of using 37% phosphoric acid as surface treatment, due to of the low survival rate the using of phosphoric acid as surface treatment was not preferred, especially in occlusal repair of porcelain fused to metal crowns (172).

Using of Nd:YAG laser enhanced the surface roughness and bond strength but the point of employment cause a silver spot and using of Nd:YAG laser cause greater increasing the monoclinic phase at the surface ranged from 26.5% and 30.5% from total crystals (173-176). For this reason, we choose APA for surface treatment for our study.

Many studies proved that the shear bond (SB) strength of different materials as metal alloys or zirconia with their veneering materials, like porcelain or composite. Al-Doham et al, said that the SB test is the best test for the porcelain bonding systems; many authors proved that the SB test is indicated for ceramic made us chose it for our study as well (63,64). Shear bond test is performed in static state; load is applied when the test specimen is stationery unlike dynamic tests where the specimen is in dynamic state. Static tests are classified into macro-tests which the bond area is more than 3 mm ^{2} and micro-tests is less than 3 mm^2 bond area, in our study size of bonded area was 5 mm^2 the test is macro shear bond test (177). In a shear bond test, two materials are combined together through an adhesive agent and

loaded in shear machine until fracture happens. It was the most widely used test (178). As no further specimen treatment is needed after the bonding process (177).

Very few studies evaluated the influence of area on "macro-bond" strengths. Sano et al. reported that for specimens with rectangular bonding areas between 0.25-11.65 mm², tensile bond strength to dentin was shown to decrease as bonding area increased, following a logarithmic function (179) A similar trend was noticed in shear bond strengths where smaller surface areas had significantly higher values when compared with those of larger areas (180).

Storage conditions include distilled water, saline, 0.05% saturated solution of thymol, 0.5% chloramine-T, 2% gluteraldehyde, and 10% formalin solutions were studied as storage media for bond strength tests (181,182). In our study we choose distilled water as storage media before the test

All study regarding repairing procedure mentioned that the height of composite ranged from 2 to 5 mm, total studies showed that the crosshead speed of shear bond strength testing machine is 0.5 or 1.0mm/min being the most common values (46 and 41%, respectively), also in agreement with a recent survey (183,184). In our study composite height was 4mm and cross head speed 1.0 mm.

Thermal cycling has been employed as a method to reproduce clinical conditions. Mair established that oral temperature ranged from −4°C-0°C when eating ice cream to 60°C-65°C when eating a hot cheese sandwich (185). The vitality of the bond strength values to survive under different stresses present in oral environment is significant for clinical predictability of dental materials shelf live. Commonly, dental materials are submitted to different kind of stresses like mechanical, thermal and chemical stresses in the mouth throughout service in the mouth. Both of thermocycling and water storage are used in in vitro studies as a common method for testing dental materials to demonstrate their suitability for in vivo conditions. Testing the samples by thermocycling speeds up the diffusion of water among changing the temperature produce stress at the interface of the two materials due to different coefficients of thermal expansion of both materials (186).

Majority of studies showed that thermocycling significantly reduced the shear bond strengths of dentin (187,188). Miyazaki et al. found reduction in the mean enamel bond strength after subjection to thermocycling, while dentin bond strengths significantly decreased after 30,000 thermal cycles (189,190). Cochran et al., said that majority of the studies including repair process with dissimilar thermocycling times but in general the consensus of thermocycling method was that reducing in bond strength as it weakens structure of composite (191).

A short thermal cycling regimen of 500 cycles is commended by the ISO TR 11450 standard 2003 (192). The number of cycles was decided according to previous studies that said that 6000 thermal cycles are equal to 5 years of clinical function (193,194). So 1200 thermal cycles used in our study are equal to 1 years of clinical function.

Concerning mechanical cycling the amount of load exerted while chewing and swallowing varies between 70-150 N (195) and cyclic compression is major stress that happens in the mouth (196). Majority of in vitro studies used monotonic tests for example tensile, compression, shear, or flexural strength to examine the properties of dental material (197). As these tests cannot produce cumulative damage (fatigue) that happens in the mouth, studies with fatigue tests are advocated to get better clinical relevance (198). The very common storage method for testing samples is water for different time intervals equal to three months. This perhaps at room temperature or at nearly mouth temperature. Also artificial saliva and sodium hypochlorite can be used for ageing of the samples. Aging have not any influence on the tensile bond strength of composite to dentin (198) and six months of aging in water reduce shear bond strength (199). Celik et al. compared different aging method as thermocycling water storage and mechanical fatigue, the result was thermocycling as aging method seems to be the best aging method for testing the quality of the bond. From all aging methods used in their study, the remarkable reduction in bond strength was detected after thermocycling (200). For this reason, we choose thermocycling as aging method for our study.

Some authors have recommended a mandatory shear bond strength about 20 MPa for permanent success of the repaired restorations, due to the level of adhesion between a composite filling materials to acid etched enamel and should be used for reference (201,202,203). Regarding the results of our study all groups below shear bond strength of 20 MPa except bisco repair system. ISO 10477 is recommended shear bond 5 MPa for polymer-based crown and bridge veneering materials (204). Regarding the results of our study all tested repair systems fulfilled the ISO requirement of 5 MPa except single bond adhesive system for zirconium group shear bond strength was 3,63 MPa. The shear bond strength of composite resin to porcelain surfaces tested in most of the previous publications ranged from 7 to 18 MPa (205,206,207). Regarding the results of the present study all tested repair systems shear bond strength was in these range except single bond was 3.63 MPa and ultradent repair systems was 6,63 MPa used for repairing zirconium and bisco repair sustem for metal. For orthodontic bonding of brackets with artificial crowns, shear bond strength of 6 to 10 MPa is advocated (208). Regarding the results of the present study all tested repair systems are seven of the ten groups exceeded 10 MPa, two group (utradent repair for zirconia 6,63 MPa and for nickle chromium 8,17 MPa within the range and one group single bond 3,63 MPa system below this range. There is an international standard (ISO9693) for the method of evaluating the bond strength between metal and porcelain using a bending test, and PFM restoration, zirconia porcaline restoration in clinical use are required to have a bond strength of 25 MPa or more (27). Regarding the results of our study all tested repair systems shear bond strength was below 25 MPa. In spite of the fact that no prognosis can be made regarding the clinical longevity of repaired restorations, Matsumura et al. reported 10 MPa as a minimal shear bond strength for the accomplishment of clinically acceptable results (209). Regarding the results of the present study all tested repair systems are seven of the ten groups exceeded 10 MPa, two group (utradent repair for zirconia 6,63 MPa and for nickle chromium 8,17 MPa, single bond 3,63 MPa system below this 10 MPa.

Kocaagaoglu et al ,2015 evaluated shear bond strengths between two porcelain repair kits and various ceramic substructure material after thermocycling

for 1200 cycle main shear bond strength of zirconium group using clearfil repair system was 8.80 MPa and for metal was 19.75 MPa (210). We evaluate shear bond strengths among five porcelain repair systems and two porcelain substructure material after thermocycling for 1200 cycle This result of metal group repaired with clearfil repair system is similar to our study result 17,37 1 MPa, the result of zirconium group repaired with clreafil repair system was 18,61 MPa. The difference may be due to using of rotary cutting instrument 30 μm is not enough to make roughness to zirconia surface. Becouse zirconia has a superior hardness and needed to be grinded with coarse diamond rotary instruments (120 to 200 mm grain size) (159).

Goncalo et al, tested the influence of mechanical surface treatment and chemical primer application on the composite shear bond strength to zirconia, the result of sand blasted monobond plus treated group after thermocycling for 500 cycle was 15.3 MPa and z prime plus was 23.2 MPa (211). This results show similarities with our study result. Zirconium group repaired with bisco repair system which include Zprime plus and bond the main shear bond strength was 18.91 MPa, and main shear bond strength for zirconium group repaired with ceramic repair system which include monobond plus and heliobond was 15.24 MPa. These scores are similar with our study.

Han et al, studied the influence of three intraoral ceramic repair kits on the bond strength between composite resin and zirconia and main shear bond strength for ceramic repair system (ivoclar) was 38.,21 MPa, cojet repair system was 7.80 MPa and Signum zirconia bond was   8.98 MPa (212). Concerning to the main shear bond strength of ceramic ivoclar repair system is so weak especially without thermocycling, this result is opposite to result of our study ivoclear was (15,24) MPa The difference may be due to using of rotary cutting instrument 30 μm is not enough to make roughness to zirconia surface because zirconia has a superior hardness and needed to be grinded with coarse diamond rotary instruments (120 to 200 mm grain size) (156).

Washa et al 2015 they said that single bond adhesive is claimed to have unique chemistry containing silane coupling agent and MDP in /addition to other component which allows the adhesive to chemical bond to glass ceramic surface without using a separate ceramic prime and they test the bond of composite to lithium disilicate using single bond adhesive without surface treatment and main tensile strength was 2.5 MPa, with diamond bur as surface treatment the main tensile strength was 2.4 MPa, and with air abraded with cojet powder as surface treatment the main tensile bond was 2.7 MPa (213). There is agreement with our study result regarding using single bond adhesive for zirconia disks before bonded with composite the main shear bond strength was 3,63 MPa.

Hanaa et al, investigated the repair potential of CAD/CAM (computer‑aided design/computer‑aided manufacturing) ceramic and composite blocks using a silane‑containing bonding agent with different repair protocols, concerning the first ceramic groups ceramic repaired with single bond adhesive and diamond stone as surface treatment the micro shear bond strength was 6.34 MPa , second ceramic group repaired with single bond adhesive and diamond stone as surface treatment and addition saline agent micro shear bond was 6.72 MPa, third ceramic group repaired with single bond adhesive and HF acid as surface treatment the micro shear bond strength was 5.72 MPa, the fourth ceramic group was treated with HF as surface treatment followed by application of the silane coupling agent and single bond adhesive the micro shear bond strength for this group was 24.45 MPa, the fifth ceramic group treated with silica coated alumina particle and single bond adhesive the micro shear bond strength was 7.14 MPa, the sixth group treated with silica coated alumina and saline treatment and single bond adhesive the micro shear bond strength of this group was 20.18 MPa (213). In agreement with our study result concerning single bond adhesive alone with any surface treatment without additional saline or metal prime lead to weak bond like our study. The main shear bond strength for zirconium group treated with single bond was3.63 MPa, according to single bond adhesive instruction permit to you another metal prime so in ceramic groups that use addition saline with single bond adhesive get good bond.

The result of the ultradent repair system using to repaired zirconium disc with composite was not compared with literature. However, no previous studies exist about this bond. The main shear bond strength was 6,63 MPa which is weak and this may can be explained that saline does not bond to zirconia well. Also the company of ultradent also don't advice to repair zirconia with ultradent repair kit. Our results also supported this advice.

Sidharth et al, evaluated the effect of three intra oral repair system on ceramic and metal infrastructure on shear bond strength between infrastructure and composite as a repair method they found the main shear bond strength with ceramic repair kit and air abrasion with 50 alumina was 18.61 and main shear bond strength for metal group repaired with clearfil repair kit was 14.98 MPa (215). This result is similar with our study main shear strength for nickle chrome group repaired with ceramic repair system was 16,13 MPa and main shear bond for nickle chromium group repaired with clearfil repair system was 17,37 MPa. The scores are very similar with our study becouse we used the same surface treatment and same repair kits.

Gasthi et al, they evaluated the shear bond strength and microleakage of two repair systems for porcelain fused metal, one of two repair system was ceramic repair system ivoclar used for repairing nikle chromium discs with composite the main shear bond strength was 24.70 MPa (216). This result has disagreement with our study the main shear bond strength for nickle chromium group repaired ceramic repair system was 16,13 MPa in our study. The difference may be because of thermocycling, our study all samples were thermocycled for 1200 cycle and their samples were thermocycled just for 300 cycle.

There is no significant difference between clerafil, biseco and ceramic repair system used for repairing zirconia and Ni-Cr infrastructure properly due to the three repair system contain bonding agents/primers (both commercial and experimental products) containing organo-phosphate monomers, such as 10- Methacryloxydecyl Dihydrogen Phosphate (MDP), have been developed for improving bond strengths of resin materials to silica-free Zirconia surface (217,218). Studies have shown the commercial phosphate-monomer-containing-zirconia primers, such as

Metal/Zirconia Primer (Ivoclar Vivadent), Monobond Plus (Ivoclar Vivadent), Clerafil Ceramic Primer (Kuraray), Signum Zirconia Bond (Heraeus), AZ Primer (Shofu) and ZPrime Plus (Bisco) significantly improved the initial and long-term resin bond strengths to zirconia ceramics (219,220,222,223,223,225). Moreover, Pott, Stiesch, and Eisenburger, concluded that pretreatment of zirconia with MDP-containing adhesive systems can lead to sufficient adhesion between different types of composite and ceramic surface, even after artificial aging (226). Another agreement with our study was that the surface treatment of a combination of air-abrasion and phosphate-monomer-containing primers improved the durability of Zirconia-resin bond strengths (227,228,229).

Three is significant difference between three repair system (bisco, ultradent, clearfil) and other two repair system (ultradent, single bond adhesive) proparrly due to ultradent and single bond adhesive depend on saline as surface treatment. Silanes are, commonly used for coupling with silica-based ceramics via the formation of chemical covalent bond (Si-O-Si), to achieve a chemical bonding between resin and Zirconia ceramics have silica-free surface and possess relatively non-polar surface. It is more chemically stable than silica-based ceramics, so traditional silane chemistry is not usually effective for Zirconia (3).

An evaluation of mode of failure of specimens is important in demonstration the quality of the bond to treated zirconium and nickle chromium surfaces and composite resins. In this experiment, it was noted that the tested specimens exhibited adhesive failure for ultradent and single bond repaired specimens that means single bond adhesive and ultradent repaired specimens get weak bond with composite. There is agreement with Aljehani et al. They said that adhesive failure does not occur in presence of good bond between compatible ceramic core and veneer material (225) .

None of the repair methods resulted in cohesive failures in the zirconia or nickle chromium specimens. This may be due to effect of thermocycling on bond between zirconia and composite and nickle chromium alloy and composite.

No significant differince between mode of bond failure related to three repair system bisco and ultradent and clearfil repair systems may be related to all the three repair system containing MDP which increase bond between metal oxide like zirconia and resin composite

One limitation of this study is that disc-shaped specimens were used instead of complete dental restorations. The medium used to perform thermal and mechanical cycling tests was distilled water, no saliva was used. The bond strength of a composite material is sensitive to chemical or mechanical influences in intraoral conditions. Another limitation is that no chewing simulator was used. Also themocycling performed just for 1200 cycle which is below the recommended number of cycles (5000) according to the ISO for metal–resin adhesion test.

5. Conclusion:

1. The three repair systems (ivoclar, clearfil and bisco) can be used effectively for repairing chipped veneering porcelain for both infrastructures. Zirconium and Ni-Cr. This may be because these repair systems include alloy primers which contain MDP and used separately.

2. All repair systems have no significant difference on repairing both Zirconium and Ni-Cr except single bond. Single bond was the weakest for repairing Zirconium restorations.

3. The three repair systems (ivoclar, clearfil and bisco) had both mixed and adhesive failures and there were no significant differences between them. For Single bond and ultradent groups all the specimens were failed adhesively. The mode of failure result also supports that the three repair systems (ivoclar, clearfil and bisco) were better than single bond and ultradent repair systems.

6. Reference:

1- Abd Wahab, Bakar WZ, Husein A. 2011. Different surface preparation techniques of porcelain repaired with composite resin and fracture resistance. J Conserv Dent 14:387–390.

2- Ozcan M. Evaluation of alternative intra-oral repair techniques for fractured ceramic-fused-to-metal restorations. Journal of Oral Rehabilitation 2003 30; 194–203.

3- Kimmich M, Stappert CF . Intraoral treatment of veneering porcelain chipping of fixed dental restorations: A review and clinical application JADA 2013;144(1):31-44.

4- Brecker SC: Porcelain bakes to gold—a new medium in prosthodontics. J Prosthet Dent 1956;6: 801-810.

5-Johnston JF, Dykema RW, Cunningham DM: The use and construction of gold crowns with a fused porcelain veneer—a progress report. J Prosthet Dent 1956;6: 811 -818.

6- Ozcan M: Fracture reasons in ceramic-fused-to-metal restorations. J Oral Rehabil 2003; 30:265-269.

7- Kenneth J. Anusavice, Chiayi Shen, H. Ralph Rawls .Philpis science of dental material ,12 edition (2013, Chapter 16 dental casting alloys and metal joining.

8-, William J. O'Brien.Dental materials and their selection fourth edition, Chapter 14 Alloys for porcelain-fused-to-metal restorations.

9- Roberts HW, Berzins DW, Moore BK, Charlton DG, Metal-Ceramic Alloys in Dentistry: A Review, Journal of Prosthodontics 18 (2009) 188–194.

10- Mc Lean JW. Ceramics in clinical dentistry. Br Dent J 1988; 164: 187-194

11- Mc Cabe JF, Wall AWG. Applied dental materials. Blackwell Science 1998; 8th.

12- Shell JS, Neilsen JP. Study of bond strength of dental porcelain fired to metal. JDent Res. 1966; 45: 32–36.

13- Malhotra ML, Maickel LB. Shear bond strength of porcelain-fused-to alloys of varying noble metal contents. J Prosthet Dent. 1980; 44: 405–412.

14- Prashanti Eachempati, Zirconia: Creating a new perspective to dentistry, Research and reviews: Journal of material sciencese-ISSN:2321-6212.

15- Saridag S, Tak O, Alniacik G, Basic properties and types of zirconia,World J Stomatol 2013 August 20; 2(3): 40-47ISSN 2218-6263.

16-Seyed A A, Suma K, Mangala D, Ranjana M, Zirconia: properties and application —Areview Pakistan Oral & Dental Journal Vol 34, No. 1 (March 2014).

17- Y.Ji, X.D. Zhang, X.C. Wang, Z.C. Che , X.M. Yu ,H.Z. Yang. Zirconia bioceramics as all-ceramics crowns, Rev.Adv.Mater. Sci. 34 (2013) 72-78.

18- Philpis science of dental material ,12 Edition (2013), Kenneth J. Anusavice, Chiayi Shen, H. Ralph Rawls,

19-H, Filser F, Loeffel O, Schumacher M, Gauckler LJ, Hammerle CH, Strength and reliability of four-unit all-ceramic posterior bridges. Dent Mater. 2005 Oct; 21(10):930-7.

20- Khamverdi, Z. Moshiri, Z. Zirconia: an up-to-date literature review, DJH 2012; Vol.4, No.1.

21- Pereira GK, Venturini AB, Silvestri T, Dapieve KS, Montagner AF, Soares FZ, Valandro LF, Low-temperature degradation of Y-TZP ceramics: A systematic review and meta-analysis, J Mech Behav Biomed Mater,2015 Mar;55:151-63.

22- Chevalier J r m, what future for zirconia as a biomaterial? .biomaterials 27 (2006) 535–543.

23- Denry I, Kelly JR. State of the art of zirconia for dental applications, Dental materiales 2008 Mar;24 (3):299-307.

24- Tang X, Nakamura T, Usami H, Wakabayashi K, Yatani H. Effects of multiple firings on the mechanical properties and microstructure of veneering ceramics for zirconia frameworks. J DENT 2012 May;40(5):372-80.

25- Luiz N. Baratieri d, Peter Greil e, Ulrich Lohbauera, Thermal-induced residual stresses affect the lifetime of zirconia–veneer crowns, dental materials 2 9 (2 0 1 3) 181–190.

26- Fischer J, Grohmann P, Stawarczyk B. Effect of Zirconia Surface Treatments on the Shear Strength of Zirconia/Veneering Ceramic Composites. Dental Materials Journal 2008; 27(3): 448－454.

27- ISO 9693.Metal-ceramic dental restorative systems. Geneva: International Organization for Standardization; 1999.

28- Yamaguchi H, Ino S, Hamano N, Okada S, Teranaka T. Examination of bond strength and mechanical properties of Y-TZP zirconia ceramics with different surface modification Dent Mater J. 2012; 31(3):472-80.

29- Cömlekoğlu ME, Dündar M, Ozcan M, Güngör MA, Gökçe B, Artunç C. Evaluation of bond strength of various margin ceramics to a zirconia ceramic. J Dent 2008 Oct;36(10):822-7.

30- Gerd G, Michael J, Lothar B, Friedrich-W B, Meike S, Philipp K. Effect of thermal expansion mismatch on the Y-TZP/veneer interfacial adhesion determined by strain energy release rate Journal of Prosthodontic Research 56 (2012) 93–101.

31- Baratieri LN, Araujo Jr EM, Monteiro Jr S. Composite Restorations in Anterior Teeth: Fundamentals and Possibilities. Hanover Park, IL; Quintessence Publishing: 2005.

32- Ferracane JL. Current trends in dental composites.Crit Rev Oral Biol Med. 1995;6 (4): 302-318.

33- Ritter AV. Direct resin-based composites: current recommendations for optimal clinical results. Compend Contin *Educ Dent*. 2005;26(7): 481-490.

34- Strassler HE. Product advances with direct-placement composite resins: Current state-of-the-art.Contemporary Esthetics. 2006;10(2):16-19.

35- Loguercio AD, Reis A, Hernandez PA, 3-year clinical evaluation of posterior packable composite resin - restorations. J Oral Rehabil. 2006;33(2):144-151.

36- Labella R, Lambrechts P, Van Meerbeek B, Vanherle G. . Polymerization shrinkage and elasticity of flowable composites and filled adhesives. Dent Mater. 1999;15(2): 128-137.

37- Lindberg A, van Dijken JW, Horstedt P. In vivo interfacial adaptation of class II resin composite restorations with and without a flowable resin composite liner. Clin Oral Investig. 2005;9 (2):77-83.

38- Baratieri LN, Canabarro S, Lopes GC, Ritter AV . Effect of resin viscosity and enamel beveling on the clinical performance of Class V composite restorations: three-year results. Oper Dent. 2003;28(5):482-487.

39- Fortin D, Vargas MA. The spectrum of composites: new techniques and materials. J Am Dent Assoc. 2000;131(Suppl):26S-30S.

40- Andersson-W, Sunnegardh-G K. Flowable resin composite as a class II restorative in primary molars: A two-year clinical evaluation. Acta Odontol Scand.2006;64(6):334-340.

41- Hervás-G A, Martínez-L MA, Cabanes-V J, Barjau-E A, Fos-G P. Composite resins.A review of the materials and clinical indications. Med Oral Patol Oral Cir Bucal. 2006;11(2): E215-E220.

42- Heintze SD, Rousson V. Survival of zirconia and metal supported fixed dental prostheses: a systematic review. Int J Prosthodont.2010;23:493-502.

43- Anusavice KJ. Standardizing failure, success, and survival decisions in clinical studies of ceramic and metal-ceramic fixed dental prostheses. Dent Mater.2011;28:102-11.

44- Wang G, Zhang S, Bian C,Kong H.Fracture mechanics analyses of ceramic/veneer interface under mixed-mode loading. J Mech Behav BiomedMater.2014;39:119 –28.

45- Deany IL. Recent advances in ceramics for dentistry. Crit Rev Oral Biol Med. 1996;7(2):134–43.

46- Cesar PF, Soki FN, Yoshimura HN, Gonzaga CC, Styopkin V. Influence of leucite content on slow crackgrowthof dental porcaline.Dent Mater. 2008;24(8):1114–22.

47-. Wiederhorn SM, Dretzke A, Rodel J. Crack growth in soda-lime silicate glass near the static fatigue limit. J Am Ceram Soc. 2002;85(9):5.

48- Morena R, Beaudreau GM, Lockwood PE, Evans AL, Fairhurst CW. Fatigue of dental ceramics in a simulated oral environment. J Dent Res. 1986;65(7):993–7.

49- Gonzaga CC, Yoshimura HN, Cesar PF, Miranda WG. Subcritical crack growth in porcelains, glass-ceramics, and glass-infiltrated alumina composite for dental restorations. J Mater Sci Mater Med 2009;20(5):1017–24.

50- Wakabayashi N, Anusavice KJ. Crack initiation modes in bilayered alumina/porcelain disks as a function of core/veneer thickness ratio and supporting substrate stiffness. J Dent Res 2000;79(6): 1398-1404.

51- Marchack BW, Futatsuki Y, Marchack CB, White SN. Customization of milled zirconia copings for all-ceramic crowns: a clin clinical report. J Prosthet Dent 2008;99(3):169-173.

52- Silva NR, Bonfante EA, Rafferty BT, et al. Modified Y-TZP core design improves all-ceramic crown reliability (published online ahead of print Nov. 5, 2010). J Dent Res 2011;90(1):104-108.

53- Baldassarri M, Zhang Y, Thompson VP, Rekow ED, Stappert CF. Reliability and failure modes of implant-supported zirconium fixed dental prostheses related to veneering techniques J Dent 2011;39(7):489 498

54- Kokubo Y, Tsumita M, Kano T, Fukushima S. The influence of Kokubo Y, Tsumita M, Kano T, Fukushima S. The influence of zirconia coping designs on the fracture load of all-ceramic molar crown Dent Mater J 2011;30(3):281-285.

55- Rosentritt M, Steiger D, Behr M, Handel G, Kolbeck C. Influence of substructure design and spacer settings on the in vitro performance of molar zirconia crowns. J Dent 2009;37(12):978-983.

56- Stappert CF, Baldassarri M, Zhang Y, Hänssler F, Rekow ED Thompson VP. Reliability and fatigue failure modes of implant supported aluminum-oxide fixed dental prostheses, Clin Oral Implants Res 2012;23(10): 1173-1180.

57- Beuer F, Schweiger J, Eichberger M, Kappert HF, Gernet WEdelhoff D. High-strength CAD/CAM-fabricated veneering materials intered to zirconia copings: a new fabrication mode for all-ceramic restorations, DentMater 2009;25(1):121-128.

58- Tholey MJ, Swain MV, Thiel N. Thermal gradients and residual stresses in veneered Y-TZP frameworks. Dent Mater. 2011; 27(11): 1102–10.

59- Lenz J, Schweizerhof K, Rong Q. Thermal stresses in ceramometallic crowns: firing in layers. Chin J Dent Res. 2002; 5(5):19.

60- Swain MV.Unstable cracking (chipping) of veneering porcelain on all-ceramic dental crowns and fixed partial dentures. Acta Biomater2009;5(5):1668–77.

61- Meira JBC, Reis BR, Tanaka CB, Ballester RY, Cesar PF, Versluis A, et al. Residual stresses in Y-TZP crowns due to changes in the thermal contraction coefficient of veneers. Dent Mater. 2013; 29(5): 594–601.

62- Baldassarri M, Stappert CF, Wolff MS, Thompson VP, Zhang Y. Residual stresses in porcelain-veneered zirconia prostheses. Dent Mater 2012;28(8):873-879.

63- Choi JE, Waddell JN,Swain MV. Pressed ceramics onto zirconia, part 2: indentation fracture and influence of cooling rate on residual stresses.Dent Mater 2011;27(11):1111-1118.

64- Schelling PK, Phillpot SR, Wolf D.Mechanism of the cubic-totetragonal phase transition in zirconia and yttria-stabilized zirconia by molecular-dynamics simulation. J Am Ceram Soc 2001; 84(7):1609-1619.

65- Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. Biomaterials 1999;20(1):1-25.

66- DE Kler M, De Jager N, Meegdes M, van der Zel JM. Influence of thermal expansion mismatch and fatigue loading on phase changes in porcelain veneered Y-TZP zirconia discs. J Oral Rehabil 2007;34(11):841-847.

67- Denry IL, Holloway JA. Microstructural and crystallographic surface changes after grinding zirconia-based dental ceramics J Biomed Mater Res B Appl Biomater 2006;76(2):440-448.

68- Miyazaki T, Nakamura T, Matsumura H, Ban S, Kobayashi T .Current status of zirconia restoration, Journal of Prosthodontic Research 57 (2013) 236–261.

69- Tuccillo, J.J. and Nielsen, J.P. (1972) Shear stress measurements at a dental porcelain-gold bond interface. Journal of Dental Research, 51, 626-633.

70- Pang, I.C., Gilbert, J.L., Chai, J, Eugene P. L. (1995) Bonding characteristics of low-fusing porcelain bonded to pure ti- tanium and palladium-copper alloy. Journal of Prosthetic Dentistry, **73**, 17-25.

71-Craig, R.G. (1993) Restorative dental materials.9th ed. Mosby, St. Louis, 492.

72- Warpeha, W.S. and Goodkind, R.J. (1976) Design and technique variables affecting fracture resistance of metal- ceramic restorations. Journal of Prosthetic Dentistry, 35, 291-298.

73- Anthony, D.H., Burnett, A.P., Smith, D.L., et al. (1970) Shear test for measuring bonding in cast gold alloy-por- celain composites. Journal of Dental Research, 49, 27-74. Craig, R.G. and Powers, J.M. (2002) Restorative dental materials.11th edition. Mosby, St. Louis, 580.

75- Kelly, M., Asgar, K. and O'Brien, W.J. (1969) Tensile strength determination of the interface between porcelain fused to gold. Journal of Biomedical Materials Research, 3, 403-408.

76- Shell, J.S. and Nielsen, J.P. (1962) Study of the bond between gold alloys and porcelain. Journal of Dental Re- search, 41, 1424-1437.

77- Riley, E.J. (1977) Ceramo-metal restoration. State of the science. Dental Clinics of North America, 21, 669-682.

78- Shillingburg, H.T., Hobo, S., Whitsett, L.D., Jacobi, R. and Brackett, S.E. (1997) Fundamentals of fixed prost- hodontics, 3rd edition, Quintessence Publishing Co, Chi- cago, 89-94.

79- Benetti p,PelogiaF,ValandroLF,Bottino MA,Bona AD The effect of porcelain thickness and surface linear application on fracture behavior of a ceramic system. Dent mater.2011sep:27(9):984-53.

80- Evans, D., Barghi, N., Malloy C, et al. (1990) The influence of condensation method on porosity and body por- celain. Journal of Prosthetic Dentistry, 63, 380-389.

81- Ritter, J.E. (1995) Predicting lifetimes of materials and material structures. Dental Materials, 11, 142-146.

82- Barreto, M.T. (1984) Failure in ceramometal fixed resto- rations. Journal of Prosthetic Dentistry, 51, 186-189.

83- Campbell, S.D. (1989) Acomparative strength study of metal ceramic and all-ceramic esthetic materials: Modulus of rupture.Journal of Prosthetic Dentistry, 62, 476- 479.

84- Suansuwan, S. and Swain, M. (1999) New approach for evaluating metal porcelain interfacial bonding. International Journal of Prosthodontics, 12, 547-552.

85- Burke, F.J. (1996) Fracture resistance of teeth restored with dentin-bonded crowns: the effect of increased tooth preparation. Quintessence International, 27, 115-121.

86- Rosenstiel, S.F., Land, M.F. and Fujimoto, J. (2001) Con- temporary fixed prosthodontics, 3rd edition. Mosby, St Louis, 73.

87- Charles, R.J. (1958) Static fatigue of glass. II. Journal of Applied Physics, 29, 1554-1560.

88- Sherill, C.A. and O'brein, W.J. (1974) Transverse strength of aluminous and feldspathic porcelain. Journal of Dental Research, 53, 683-690.

89-Anusavice, K.J. and Zhang, N.Z. (1998) Chemical dura- bility of Dicor and fluorocanasite-based glass-ceramics. Journal of Dental Research, 77, 1553-1559.

90- Ban, S. and Anusavice, K.J. (1990) Influence of test method on failure stress of brittle dental materials. Journal of Dental Research, 69, 1791-1799.

91- Niedermeier, W., Proano, F.P., O¨ zcan, M., Mayer, B., Nergiz, I. & PFEIFFER, P. (1998) Enorale Reparaturen mit tribochemischem Verbund. Zahna¨rztliche Mitteilungen, 16, 54

92- Llobell, A., Nichlls, J.I., Kois, J.C.,Daly CH. (1992) Fatigue life of porcelain repair systems. International Journal of Prosthodontics, 5, 205-213.

93- Anusavice, K.J. and Zhang, N.Z. (1998) Chemical durability of Dicor and fluorocanasite-based glass-ceramics. Journal of Dental Research, 77, 1553-1559.

94- White, S.N., Li, Z.C., Yu, Z., et al. (1997) Relationship between static, chemical and cyclic mechanical fatigue in feldspathic porcelain. Dental Materials, 13, 103-110.

95- Bragger, U., Aeschlimann, S., Burgin, W., et al. (2001) Biological and technical complications and failures with fixed partial dentures (FPD) on implants and teeth after four to five years of function. Clinical Oral Implants Re-search, 12, 26-34.

96- Hammad I.A, Talic Y.F. (1996). "Designs of bond strength tests for metal-ceramic complexes: review of the literature." J Prosthet Dent 75(6): 602-608.

97- Versluis A., Tantbirojn D.,Douglas W.H. (1997). "Why do shear bond tests pull out dentin?" J DentRes 76(6): 1298-1307.

98- Hadavi F., Hey J.H., Ambrose E.R., Louie P.W, Shinkewski D.J. (1993). "The effect of dentinprimer on the shear bond strength between composite resin and enamel." Oper Dent 18(2): 61-65.

99- Blatz M.B., Bergler M., Ozer F., Holst S., Phark J.H, Chiche G.J. (2010)."Bond strength ofdifferent veneering ceramics to zirconia and their susceptibility to thermocycling." Am JDent 23(4): 213-216.

100- Dundar M., Ozcan M., Gokce B., Comlekoglu E., Leite F Valandro L.F. (2007). "Comparison oftwo bond strength testing methodologies for bilayered all-ceramics." Dent Mater 23(5): 630-636.

101- Kelly M., Asgar K.,O'Brien W.J. (1969). "Tensile strength determination of the interface between porcelain fused to gold." J Biomed Mater Res 3(3): 403-408.

102- Aboushelib M.N., de Kler M., van der Zel J.M.,Feilzer A.J. (2009).Microtensile bond strength and impact energy of fracture of CAD-veneered zirconia restorations." J Prosthodont 18(3): 211216.

103- Azer S.S., Drummond J.L., Campbell S.D.,El Moneim Zaki A. (2001). "Influence of core buildupmaterial on the fatigue strength of an all-ceramic crown." J Prosthet Dent 86(6): 624-631.

104- Seghi R.R., Daher T.,Caputo A. (1990). "Relative flexural strength of dental restorative ceramics." Dent Mater 6(3): 181-184.

105- Fischer J., Stawarczyk B., Hammerle C.H. (2008 b). "Flexural strength of veneering ceramics forzirconia." J Dent 36(5): 316-321.

106- Lenz J.,Kessel S. (1998). "Thermal stresses in metal-ceramic specimens for the ISO crack initiationtest (three-point flexure bond test)." Dent Mater 14(4): 277-280.

107- Kosyfaki P. (2011). Bond strength of veneering ceramics to zirconia, Freiburg university.Thesis.

108- Charalambides P.G., Lund J., Evans A.G., McMeeking R.M. (1989)."A Test Specimen for Determining the Fracture Resistance of Bimaterial Interfaces. " Journal of Applied Mechanics 56(1): 77-82.

109- Suansuwan N.,Swain M.V. (1999). "New approach for evaluating metal-porcelain interfacialbonding." Int J Prosthodont 12(6): 547-552.

110- Suansuwan N.,Swain M.V. (2001). "Determination of elastic properties of metal alloys and dentalporcelains." J Oral Rehabil 28(2): 133-139.

111- Suansuwan N.,Swain M.V. (2003). "Adhesion of porcelain to titanium and a titanium alloy." J Dent31(7): 509-518.

112- Suansuwan N. (2001). Application of Strain Energy Release Rate to Characterise the Adhesion at Ceramic-Metal Interfaces. Department of Biomaterials Science.The University of Sydney Doctor of Philosophy.

113- Ozcan M, Bernasconi M. Adhesion to zirconia used for dental restorations: asystematic review and meta-analysis. J Adhes Dent 2015;17(1):7-26.

114- Inokoshi M, De Munck J, Minakuchi S, Van Meerbeek B. Meta-analysis of bonding effectiveness to zirconia ceramics. J Dent Res 2014;93(4):329-34.

115- Dos Santos JG, Fonseca RG, Adabo GL, dos Santos Cruz CA,Shear bond strength of metal-ceramic repair systems. J Prosthet Dent2006; 96:165-73.

116- Chung KH, Hwang YC. Bonding strengths of porcelain repair systemswith various surface treatments. J Prosthet Dent 1997; 78:267-74.

117- Shadid R M, Sadaqah N R,Abu nabaa LA,Alomari W M.Porcelain fracture of metal-ceramic tooth supported and implant-supported restorations :Areview.Open Journal of stomatology ,2013,3,411-418

118- Berry T, Barghi N, Chung K. Eff ect of water storage on the silanization in porcelain repair strength. J Oral Rehabil 1999; 26:459-63.

119- Coornaert,J . Adrians,P. & De Boever, j. (1984) Long-term clinical study of porcelain-fused-to-gold restorations. Journal of Prosthetic Dentistry, 51, 338.

120- Karlsson, S. (1986) Clinical evaluation of fixed bridges 10 years following insertion.Journal of Oral Rehabilitation, 13, 423.

121- Kerschbaum, Th, Seth,M. & Teeuwen,U. (1997) Verweildauer von Kunststoff-und Metal–Keramisch verblendeten Kronen und Bru¨ cken. Deutsche Zahna¨rztliche Zeitschriften, 52, 404.

122- Vagkopoulou T, Koutayas SO, Koidis P, Strub JR. Zirconia in dentistry: part 1. Discovering the nature of an upcoming bioceramic.Eur J Esthet Dent 2009;4:2–23.

123- Pittayachawan P, McDonald A, Young A, Knowles JC. Flexural strength, fatigue life, and stress-induced phase transformation study of Y-TZP dental ceramic. J Biomed Mater Res B Appl Biomater 2009; 88:366–77.

124- Raigrodski AJ, Chiche GJ, Potiket N, Hochstedler JL, Mohamed SE, Billiot S, Mercante DE. The efficacy of posterior three-unit zirconium-oxide-based ceramic fixed partial dental prostheses: a prospective clinical pilot study. J Prosthet Dent 2006;96: 237–44.

125- Kirmali O, Akin H, Ozdemir AK. Shear bond strength of treatments. Photomed Laser Surg 2013; 31:261–8.

126- Sailer I, Fehér A, Filser F, Lüthy H, Gauckler LJ, Schärer P. Prospective clinical study of zirconia posterior fixed partial dentures: 3-year follow-up. Quintessence Int 2006; 37:685–93.

127- Noel LG, Mitchell WC: Porcelain veneer repair of prostheses. Gen Dent 1997; 45:182-185.

128- Filho AM, Vieira LC, Araújo E, Monteiro Júnior S.: Effect of different ceramic surface treatments on resin microtensile bond strength. JProsthodont 2004; 13:28-35.

129- Güler AU, Yilmaz F, Yenisey M, Güler E, Ural C .Effect of acid etching time and a self-etching adhesive on the shear bond strength of composite resin to porcelain. J Adhes Dent 2006; 8:21-25.

130- Hickel R, Brushaver K, Ilie N: Repair of restorations–criteria for decision making and clinical recommendations. Dent Mater ,2013;29:28-50.

131- Gregory, W.A. & Moss,S.M. (1990) Effects of heterogeneous layers of composite and time on composite repair of porcelain.Operative Dentistry, 15, 18.

132- Lutz, F. & Phillips, R.W. (1983) A classification and evaluation of composite resin systems ..Journal of Prosthetic Dentistry, 50, 480.

133- Simonsen, R.J. & Calamia, J.R. (1983) Tensile bond strength of etched porcelain. Journal of Dental Research, 62, 297.

134- Stangel, I., Nathanson, D. & HSU, C.S. (1987) Shear strength of the composite bond to etched porcelain. Journal of Dental Research, 66, 1460.

135- Creugers, N., Snoek, P. & KA¨ Yser, A.F. (1992) An experimental porcelain repair system evaluated under controlled clinical conditions. Journal of Prosthetic Dentistry, 68,724.

136- Llobell, A., Nicholls, J.I., Kois, J.C. & Daly, C.H. (1992) Fatigue life of porcelain repair systems.International Journal of Prosthodontics, 5, 205.

137- Ugur E, Hande S S , Erkan S , Meltem M E, Sevda Oz, Taner Y, and Esra Y , Shear bond strength of a new self-adhering flowable composite resin for lithium disilicate-reinfored CAD/CAM ceramic material, J Adv Prosthodontic 2014;6:434-43.

138- Nuray C, Ahmet U, Ender K, Senih C Effect of Flowable Composite Intermediate Layer on Bond Strength of Intra-Oral Porcelain Repair: An In-Vitro Study, OHDMBSC - Vol. VIII - No. 2 - June, 2009.

139- Narasimha J, Vinod V, Comparative Evaluation of Compressive Strength and Flexural Strength of Conventional Core Materials with Nanohybrid Composite Resin Core Material an in Vitro Study. J Indian Prosthodont Soc (July-Sept 2013) 13(3):281–289.

140- Nadabalung D, Powers J, Conelly M. Comparison of bond strengths of three denture base resins to treated Ni-Cr-Be alloy. J Prosthet Dent 1998; 80354-61.

141- Hegde V, Katavkar R. A new dimension to conservative dentistry: Airabrasion. J Conserv Dent 2010; 13:4-8.

142- Kern M, Wegner S. Bonding to zirconia ceramic: adhesion methods andtheir durability. Dent Mater 1998; 14:64-71.

143- Della Bonna A, Borba M, Benetti P, Cecchetti D. Effect of surface treatment on the bond strength of a zirconia-reinforced ceramic to composite resin. Braz Oral Res 2007; 21:10-5.

144- Akgungor G, Sen G, Aydin M. Influence of different surface treatments on the short-term bond strength and durability between a zirconia post and a composite resin core material.J Prosthet Dent 2008; 99: 388-99.

145- Yang B, Barloi A, Kern M. Influence of air abrasion on zirconia ceramic bonding using an adhesive composite resin. Dent Mater 2010; 26:44-50.

146- Gomes A, Oyagüe R, Lynch C, Montero J, Albaladejo A. Influence of sandblasting granulometry and resin cement composition on micro tensile bond strength to zirconia ceramic for dental prosthetic frameworks. J Dent 2013; 41:31-41.

147- Valentino TA, Borges GA, Borges LH, Platt JA, Correr-Sobrinho L. Influence of glazed zirconia on dual cure luting agent bond strength. Oper Dent2012; 37: 181-7.

148- Tsuo Y, Yoshida K, Atsuta M. Effects of alumina-blasting and adhesive primers on bonding between resin luting agent and zirconia ceramics. DentMater J 2006; 25:669-74.

149- Thomson J, Stoner B, Piascik J, Smith R. Adhesion/cementation to zirconia and other non-silicate ceramics: where are we now? Dent Mater 2011; 2771-82.

150- Karakoca S, Yilmaz H. Influence of surface treatments on surface roughness phase transformation, and biaxial flexural strength of Y-TZP ceramics. J Biomed Mater Res Part B: App Biomat 2009; 91B:930-7.

151- Kitayama S, Nikaido T, Maruoka R, Zhu L, Ikeda M, Watanabe A, Foxton RM, Miura H, Tagami J. Effect of an internal coating technique on tensile bond strengths of resin cements to zirconia ceramics. Dent Mater J 2009; 28:446-53.

152- Moon JE, Kim SH, Lee JB, Ha SR, Choi YS. The effect of preparation order on the crystal structure of yttria-stabilized tetragonal zirconia polycrystal and the shear bond strength of dental resin cements. Dent Mater 2011;27: 651-63.

153- Monaco C, Tucci A, Esposito L, Scotti R. Microstructural changes produced by abrading YTZP in presintered and sintered conditions. J Dent 2013;41: 121-6.

154- Magne P, Paranhos M, Burnett L. New zirconia primer improves bond strength of resin-based cements. Dent Mater 2010; 26:345-52.

155- Scherrer SS, Cattani-Lorente M, Vittecoq E, De Mestral F, Griggs JA, Wiskott HW. Fatigue behavior in water of Y-TZP zirconia ceramics after abrasion with 30 mm silica-coated alumina particles. Dent Mater 2010;27: e28-42.

156- Chintapalli R, Marro F, Pique E, Anglada M. Phase transformation and subsurface damage in 3Y-TZP after sandblasting. Dent Mater 2013;29: 566-72.

157- Schneider, W., Powers, J.M. & T Pierpoint, H.P. (1992). Bond strength of composites to etched and silica coated porcelain fusing alloys. Dental Materials,8, 211.

158- Wood, M., Litkowski, L.J., Thompson, V.P. & Church, T. (1992) Repair of porcelain ∕ metal restoration with resin bonded overcasting.Journal of Esthetic Dentistry, 4, 110.

159- Ohkuma K, K.azama M, Ogura H. The grinding efficiency by diamond points developed for yttria partially stabilized zirconia. Dent Mater J 2011; 30:511-6.

160- Derand P, Derand T. Bond strength of luting cements to zirconium oxide ceramics. Int J Prosthodont 2000; 13:131-

161- Curtis R, Wright A, Fleming G. The influence of surface modification techniques on the performance of a Y-TZP dental ceramic. J Dent 2006; 34: 195-206.

162- Kim H, Lim H, Park Y, Vang M. Effect of zirconia surface treatments on the shear bond strength of veneering ceramic. J Prosthet Dent 2011; 105: 315-22.

163- Karakoca S, Yilmaz H. Influence of surface treatments on surface roughness transformation, and biaxial flexural strength of Y-TZP ceramics. J Biomed Mater Res Part B: App Biomat 2009;91B:930-7.

164- Della Bonna A, Anusavice K, Hood J. Effect of ceramic surface treatment on tensile bond strength to a resin cement. Int J Prosthodont 2002;15: 248-53.

165- Ural C, Kulunk T, Kulunk S, Kurt M. The effect of laser treatment on bonding between surface and resin cement. Acta Odontol Scand 2010;68: 354-9.

166- Casucci A, Osorio E, Osorio R. Influence of different surface treatments on surface zirconia frameworks. J Dent 2009; 27:891-7.

167- Abu-Eittah M. Assessment of different surface treatments effects on surface-roughness of zirconia and its shear strength to human dentin. Life Sci J2012; 9:1792-803.

168- Matsumara, H., Kawahara, M., Tanaka, T. & Atsuta, M. (1989) A new porcelain repair system with silane coupler, ferric chloride, and adhesive opaque resin. Journal of Dental Research,68, 813.

169- Wolf, D.M., Powers, J.M. &O'keefe, K.L. (1992) Bond strength of composite to porcelain treated with new porcelain repair agents.Dental Materials, 8 158.

170- Thurmond, J.W., Barkmeier, W. & Wilwerding, T.M. (1994) Effect of porcelain surface treatments on bond strengths of composite resin bonded to porcelain.Journal of Prosthetic Dentistry, 72, 355.

171- Chung, K.H. & Hwang, Y.C. (1997) Bonding strengths of porcelain repair systems with various surface treatments.Journal of Prosthetic Dentistry, 78, 267.

172- Creugers, N., Snoek, P. & Kayser, A.F. (1992) Anexperimental porcelain repair system evaluated under controlled clinical conditions.Journal of Prosthetic Dentistry, 68, 724.

173- Akyil M, Uzun I, Bayindir F. Bond strength of resin cement to yttrium stabilized tetragonal zirconia ceramic treated with air abrasion, silica coating and laser irradiation. Photomed Laser Surg 2010; 28:801-8.

174- Usumez A, Hamdemirci N, Koroglu BY, Simsek I, Parlar O, Sari T. Bond strength of resin cement to zirconia with different surface treatments. Laser Med Sci 2013; 28:259-66.

175- Paranhos M, Burnett L, Magne P. Effect of Nd:YAG and CO2 laser treatment the resin bond to zirconia ceramic.Quint Int 2011;42:79-89.

176- Van Meerbeek B, Peumans M, Poitevin A, Mine A, Van Ende A, Neves A, et al. Relationship between bond-strength tests and clinical outcomes. Dent Mater. 2010; 26: e100–21.

177- Burke FJ, Hussain A, Nolan L, Fleming GJ. Methods used in dentine bonding tests: An analysis of 102 investigations on bond strength. Eur J Prosthodont Rest Dent. 2008;16: 158–65.

178- Burke FJ, Hussain A, Nolan L, Fleming GJ. Methods used in dentine bonding tests: An analysis of 102 investigations on bond strength. Eur J Prosthodont Rest Dent. 2008;16: 158–65.

179- El-Askay FM, Nassif MS, Andrade AM, Reis A, Loguercio AD Effect of surface area and air drying distance on shear strength of etch-and-rinse adhesive. Braz Oral Res 2012; 26:418 –23.

180- Salz U, Bo ck T. Testing adhesion of direct restoratives to dental hard tissue - A review. J Adhes Dent 010;12: 343–71.

181- Perdigao J. Dentin bonding - Variables related to the clinical situation and the substrate treatment. Dent Mater. 2010;26: e24–37.

182- Al-Salehi SK, Burke FJ. Methods used in dentin bonding tests: an analysis of 50 investigations on bond strength. Quintessence Int 1997; 28:717–23.

183- Burke FJ, Hussain A, Nolan L, Fleming GJ. Methods used indentine bonding tests: an analysis of 102 investigations onbond strength. Eur J Prosthodont Restor Dent2008; 16:158–65.

184- Khoroushi M, Rafiei E. Effect of thermocycling and water storage on bond longevity of two self-etch adhesives. Gen Dent. 2013;61: 39–44.

185- Mair L, Padipatvuthikul P. Variables related to materials and preparing for bond strength testing irrespective of the test protocol.Dent Mater. 2010;26: e17–23.

186- Chang CY, Kuo J S, Lin YS, Chang YH, Fracture resistance and failure modes of CEREC endo-crowns and conventional post and core- supported CEREC crowns Journal of dental sciences 4(3):110-117.

187- Sirisha K, Rambabu T , Shankar Y K, Ravikumar P. Validity of bond strength tests: A critical review: Part I, J Conserv Dent. 2014 Jul-Aug; 17(4): 305–311.

188- Helvatjoglu-Antoniades M, Koliniotou-Kubia E, Dionyssopoulos P.The effect of thermal cycling on the bovine dentine shear bond strength of current adhesive systems.J Oral Rehabil. 2004;31: 911–7.

189- Miyazaki M, Sato M, Onose H, Moore BK. Influence of thermal cycling on dentin bond strength of two-step bonding systems. Am J Dent. 1998; 11:118–22.

190- Miyazaki M, Sato M, Onose H. Durability of enamel bond strength of simplified bonding systems. Oper Dent. 2000; 25:75–80.

191- Cochran, M., Carlson, T., Moore, B., Richmond, N. & Brackett, W. (1988) Tensile bond strengths of five porcelain repair systems Operative Dentistry, 13, 112

192- Dental materials - testing of adhesion to tooth structure. Second ed. Switzerland: 2003. Technical specification ISO/TS 11405.

193- A. Leibrock, Degenhart M, Behr M, Rosentritt M, Handel G, In vitro study of the effect of thermo- and load-cycling on the bond strength of porcelain repair systems J. Oral Rehabil., 26 [2] 130–137 (1999).

194- Fischer J, Zbären C, Stawarczyk B, Hämmerle CH, The effect of thermal cycling on metal-ceramic bond strength, JDent.,37 [7] 549–553 (2009).

195- Anderson DJ. Measurement of stress in mastication. I. J Dent Res, 1956;35:664–70.

196- Baran G, Boberick K,McCool J. Fatigue of restorative materials. Crit Rev Oral Biol Med.2001; 12:350–60.

197- Scherrer SS, Wiskott AH, Coto-Hunziker V, Belser UC. Monotonic flexure and fatigue strength of composites for provisional and definitive restorations. J Prosthet Dent.2003;89:579–88.

198- Mair L, Padipatvuthikul P. Variables related to materials and preparing for bond strength testing irrespective of the test protocol. Dent Mater. 2010;26: e17–23.

199- Takemori T, Chigira H, Itoh K, Hisamitsu H, Wakumoto S. Factors affecting tensile bond strength of composite to dentin. Dent Mater. 1993; 9: 136–8.

200- Celik G, Ismatullaev A, Sari A, and Usumez A, Comparison of the Effectiveness of Bonding Composite to Zirconia as a Repair Method, Int. J. Appl. Ceram. Technol., 1–7 (2015).

201- Haselton DR1, Diaz-Arnold AM, Dunne JT Jr Shear bond strengths of 2 intraoral porcelain repair systems to porcelain or metal substrates. J Prosthet Dent. 2001 Nov;86 (5):526-31.

202- Kato H1, Matsumura H, Tanaka T, Atsuta M., Bond strength and durability of porcelain bonding systemsJ Prosthet Dent. 1996 Feb; 75(2):163-8.

203- Behr.M, Rosentritt .M , Groger .G and Handel .G, Adhesive bond of veneering composites on various metal surfaces using silicoating, titanium-coating or functional monomers.J. Dent 31, 33–42 (2003).

204- Proano, P., Pfeiffer, P., Nergiz, I., Niedermeier, W. Shear bond strength of repair resin using an intraoral tribochemic.al coating on ceramometal, ceramic and resin surfaces. J Adhesion Sci Technol. 1998;12 :1121–1135.

205- Kupiec KA, Wuertz KM, Barkmeier WW, Wilwerding TM., Evaluation of porcelain surface treatments and agents for composite-to-porcelain repair, J Prosthet Dent. 1996 Aug;76(2):119-24.

206- Chung KH, Hwang YC. Bonding strengths of porcelain repair systems with various surface treatments. J Prosthet Dent. 1997 Sep;78(3):267-74.

207- Frankenberger R, Krämer N, Sindel J, Repair strength of etched vs silica-coated metal-ceramic and all-ceramic restorations, Oper Dent. 2000 May-Jun;25(3):209-15.

208- Barcelo S. H. F, Medina H.R, Torres S.L.A, Herrera L.M.S, Pedrero A G Fand Gonzalez R.O, Evaluation of bond strength of metal brackets by a resin to ceramic surfaces J. Clin. Dent. **17**, 5–9 (2006).

, Evaluation of bond strength of metal brackets by a resin to ceramic surfaces J. Clin. Dent. **17**, 5–9 (2006).

208- Schmage P, Nergiz I, Herrmann W, Ozcan M. Influence of various surface conditioning   methods on the bond strength of metal brackets to ceramic surfaces, AmJOrthodDentofacialOrthop.2003May;123(5):540-6.

209- Matsumura H, Yanagida H, Tanoue N, Atsuta M, Shimoe S. Shear bond strength of resin composite veneering material to gold alloy with varying metal surface preparations, J Prosthet Dent. 2001 Sep;86(3):315-9.

210- Kocaagaoglu H H , Gurbulak A, An Assessment of Shear Bod Strength Between Ceramic Repair Systems and Different Ceramic Infrastructures, scanning vol. 37, 4 (2015)

211- Barragan G, Chasqueira F, -Oliveira, Jaime Portugal SA, Ceramic Repair: Influence of Chemical and Mechanical Surface Conditioning on Adhesion to Zirconia. OHDM - Vol. 13 - No. 2 - June, 2014.

212- Han IN, Kang DW, Chung CH, Choe HC, and Kyoung M. Effect of various intraoral repair systems on the shear bond strength of composite resin to zirconia. J Adv Prosthodont. 2013 Aug;5(3):248-255.-

213- WASHA M M . GHALLAB OH. influence of different surface treatment on microshear bond strength of repair resin composite to two cad/cam esthetic restorative

materials.Tanta Dntal Journal 12(2015) 178-184.

214- Zaghloul H, Elkassas DW, and Haridy MF. Effect of incorporation of silane in the bonding agent on the repair potential of machinable esthetic blocks. Eur J Dent. 2014 Jan-Mar; 8(1): 44–52.

215- Sidharth J, Hari P, Sharad G, and Akshaya B.To Evaluate the Effect of Various Surface Treatments on the Shear Bond Strength of Three Different Intraoral Ceramic Repair Systems: An In Vitro Study. J Indian Prosthodont Soc. 2013 Sep; 13(3): 315–320.

216. Gasthi A, Chakravarthy AK, Kumar BM, Evaluation of shear bond strength and microleakage of two repair systems for porcelain fused metal restoration. 2015, IP: 78.180.223.235.

217.Tanaka, R., A. Fujishima, Y. Shibata, A. Manabe and Y. Miyazaki, 2008.Cooperation of phosphate monomer and silica modification on Zirconia. J Dent. Res., 87: 666-670.

218. Magne, P., M.P.G. Paranhos and L.H. Burnett, 2010. New Zirconia primer improves bond strength of resin-based cements. Dent. Mater., 26: 345-352.

219. Cura, C., M. Ozcan, G. Isik and A. Saracoglu, 2011. Comparison of alternative adhesive cementation concepts for Zirconia ceramic: Glaze layer vs Zirconia primer. J. Adhes. Dent., 14: 75-82.

220. Koizumi, H., D. Nakayama, F. Komine, M.B. Blatz and H. Matsumura, 2012. Bonding of resin-based luting cements to Zirconia with and without the use of ceramic priming agents. J. Adhes. Dent2012 Aug;14(4):385-92.

221- Attia, A. and M. Kern, 2011a.Long-term resin bonding to Zirconia ceramic with a new universal primer. J. Prosthet. Dent., 106: 319-327.

222- Ural, C., T. Kulunk, S. Kulunk, M. Kurt and S. Baba2010.Determination of resin bond strength to, Zirconia ceramic surface using different primers.Acta. Odontol. Scand., 69: 48-53.

223- Kitayama, S., T. Nikaido, R. Takahashi, L. Zhu and M. Ikeda., 2010b. Effect of primer treatment onbonding of resin cements to Zirconia ceramic.Dent. Mater.,26: 426-432.

224- Lehmann, F. and M. Kern, 2009. Durability of resin bonding to Zirconia ceramic using different primers. J Adhes Dent. 2009 Dec;11(6):479-83.

225- P.C. Pott, M. Stiesch, M. Eisenburger, Influence of artificial aging on the shear bond strength of zirconia- composite interfaces after pretreatment with new 10-MDP adhesive systems. JDMT, Volume 5, Number 1, March 2016.

226- Lindgren, J., J. Smeds and G. Sjogren, 2008. The effect of surface treatments and aging in water on bond strength to Zirconia. Oper. Dent., 33: 675-681.

227- Kern, M., A. Barloi and B. Yang, 2009. Surface conditioning influences Zirconia ceramic bonding. J. Dent. Res., 88: 817-822.

228- Akgungor, G., D. Sen and M. Aydin, 2008. Influence of different surface treatments on the short-term bond strength and durability between a Zirconia post and a composite resin core material. J. Prosthet. Dent., 99: 388-399.

229-Thompson, J.Y, B.R. Stoner, J.R. Piascik and R. Smith, 2011. Adhesion/cementation to Zirconia and other non-silicate ceramics: Where are we now? Dent. Mater., 27: 71-82.: