‘COMPARATIVE EVALUATION OF FLEXURAL STRENGTH OF HEAT POLYMERIZED POLYMETHYL METHACRYLATE DENTURE BASE MATERIAL REINFORCED WITH SILANIZED ALUMINIUM OXIDE NANOPARTICLES AND SILANIZED TETRAGONAL ZIRCONIUM OXIDE NANOPARTICLES – AN *IN VITRO* STUDY.’

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List of Abbreviations Used

No.	Abbreviations	Full form
1	n	Number of samples in each group
2	N	Newton
3	p value	Probability of happening of an event
4	S.D.	Standard deviation
5	ANOVA	Analysis of variance
6	Mpa	Mega Pascal
7	^{0}C	Degree Celsius
8	^{0}	Degree
9	mm	Millimeter
10	i.e.	that is
11	PMMA	Polymethyl methacrylate
12	μ	Micron
13	Al_2O_3	Aluminium oxide
14	TiO_2	Titanium dioxide
15	ZrO_2	Zirconium oxide
16	SiO_2	Silica
17	TMSPM	Trimethoxysilylpropylmethacrylate
18	Rpm	Rotations per minute
19	mins	Minutes
20	gms	Grams
21	Nps	Nanoparticles

Introduction

The loss of teeth by an accident or disease has plagued mankind throughout the ages and the means of replacing missing teeth structures by artificial materials continues to justify a large part of the application of material sciences. [1]

Edentulism is the result of loss of all permanent teeth. Tooth loss is regarded as mutilating, terminal outcome of a multifactorial process involving a varied biologic process (caries, periodontal diseases, pulpal pathology, trauma, oral cancer) as well as non-biological factors related to dental procedures (access to care, patient's preferences, treatment options etc.).[2]

Over a past few decades, there has been a steady decline in the rates of tooth loss, more than one-third (33.1 %) of those aged ≥ 65 years are edentulous. The

number of edentulous people are expected to increase as a result of the strong increase in the aging population.[3] Over the next two decades the need and demand for complete dentures will surge tremendously as this generation matures into the upper age groups.[4]

The treatment options for edentulous patients vary from a wide range from conventional complete dentures to fixed implant-supported restorations. In most of the cases conventional complete dentures will remain the treatment of choice due to medical and financial reasons. As the missing tooth structure is replaced by artificial materials it is very crucial that there has to be continuous research and development in the field of dental materials.[1]

The first dental prosthesis was believed to have been constructed in Egypt about 2500 BC. Dentures are believed to have surfaced as a mode of treatment for replacing missing teeth around 700 BC.[5] In 8th century the Japanese mastered the art of woodcarving. Dentures carved from a single piece of wood were readily available and inexpensive.[1] Natural teeth were secured with the help of screws. George Washington had a set of dentures which were fabricated from wood.[6]

The leap of further development in the field of dental materials was slow until 17th century when modern dentistry had been said to begin with **Pierre Fauchard (1678-1761)**. He fabricated dentures from bone by measuring individual arches and shaped the bone to fit the arches. Dentures fabricated from ivory were stable in the oral environment and offered better aesthetics but they were not readily available and were expensive. [7]

Although the art of firing porcelain was practiced in China in the 9th and 10th century, during the 18th century, gold and porcelain were the material of choice for denture base materials. It was not until 1774 that **Alexis Duchateau**, a Parisian apothecary, dissatisfied with his own stained hippopotamus ivory denture, teamed up with Parisian dentist **Nicholas Dubois De Chemant** to fabricate a baked porcelain complete denture in a single block.[6] The advantages were that it could be shaped easily, ensured intimate contact with the underlying tissues, it was stable, had minimal water sorption, smooth surfaces after glazing, less porosity, low solubility and could be tinted. Its shortcomings were brittleness and difficulty in grinding and polishing. The first porcelain denture with artificial teeth was fabricated by **Loomis (1854**).[1]

During the 19th century, Tortoise Shell (1850), Gutta Percha (1851), Vulcanite (1851), Cheoplastic (1856), Rose Pearl (1860), Aluminium (1867), Celluloid (1870) were used as denture base materials. Vulcanite remained the principal denture base material for the next 75 years. [1]

In the 20th century various materials like Bakelite (1909), Stainless steel (1921), Cobalt Chromium (1930), Acrylic Resin (1937), Self-cure Acrylic Resin, Epoxy Resin (1951), Polystyrene (1951), Nylon (1955), Polycarbonates (1967), High impact acrylic (1967), Visible L.C (1947), Pure Titanium (1998) were utilized for fabrication of dentures. [1]

During the latter part of the nineteenth century, polymers entered the field of denture base materials. The PMMA and its copolymers introduced by **Dr. Walter Wright (1937)** continues to be the most popular non-metallic material till date and is being used in 95% dentures.[4] PMMA has been the material of choice because of its

biocompatibility, stability in the oral environment, superior esthetics, favorable working characteristics, processing ease, accurate fit, and inexpensive equipments required for fabrication process.[8,9] However, because of certain drawbacks like residual monomer allergy, poor mechanical strength, low fatigue strength, brittleness, poor thermal conduction and low hardness, porosity, crazing, warpage, poor adhesion to metal and porcelain, it is still far from ideal in fulfilling the mechanical requirements of the denture base material. [10]

Studies have shown that 68 % of the complete dentures fabricated, fractured within the first three years.[11] The fracture of acrylic resin dentures is an unresolved problem in removable prosthodontics. The midline fracture of a maxillary denture is most common and is often the result of flexural fatigue and deep incisal notching at the labial frenum.[12]

Smith [8,14] studied the practical situation with respect to the fracture of dentures and concluded that there are two types of failures, one that is outside the mouth which was caused by impact forces e.g. accidental dropping of the denture while cleaning, insertion and removal. Second, inside the mouth, usually in function; this is probably due to a fatigue phenomenon, i.e. a low and repetitive stress rate.

Denture base material should have sufficient strength to withstand fracture while in service. Different physical properties can be used to assess the strength of denture base materials. The most common test is flexural strength i.e. the amount of force needed to deform the material to fracture or yield irreversibly. [9]

Given the function of a denture base in a removable prosthesis, high flexural strength, flexural modulus, and a large yield point distance would help resist torsional forces in function, leading to a longer clinical service life. There are three ways to improve the mechanical properties of PMMA: replacing PMMA with an alternative material; chemically modifying it; and reinforcing the PMMA with other materials.[8]

Improvement of mechanical properties of denture base materials were tried to be achieved by adding a polyfunctional crosslinking agent such as polyethylene glycol di-methacrylate or by incorporating a rubber phase.[16]

Recently, much attention has been directed towards the incorporation of inorganic nanoparticles into PMMA to improve its properties. The properties of polymer nanocomposites depend on the type of incorporating nanoparticles, their size and shape as well as the concentration and interaction with the polymer matrix. [18]

Aluminium oxide nanoparticles possess strong ionic inter-atomic bonding, have high hardness, good thermal properties, decrease warpage, make the material radio-opaque and inhibit growth of bacteria over the denture surface. They can exist in several crystalline phases, which all revert to the most stable hexagonal alpha phase at elevated temperature. [8,14,20] **Jasmin BS and Ismail JJ (2014)** concluded that addition of silanized Al_2O_3 nanoparticles increased the values of flexural strength with 1wt% significantly compared to control groups, then the flexural strength began to decrease with 3 wt% in which the value of flexural strength was less than control. [21]

Zirconium oxide (ZrO_2), commonly referred to as zirconia, possesses strong ionic inter-atomic bonding, giving rise to its desirable material characteristics.[18] Zirconium oxide fillers (ZrO_2) can be used because of their excellent biocompatibility and also because they are white in color, they less likely alter the esthetics. The nano-filler particles of zirconium oxide (ZrO_2) yields a better dispersion, eliminate aggregation and improve its compatibility with organic polymer.[16] Zirconium oxide (ZrO_2) exists in three crystalline phases i.e. Monoclinic, cubic & tetragonal.[22] It has shown to reduce the polymerization shrinkage of polymethyl methacrylate denture base material, decrease its warpage, make the material radio-opaque and inhibit the growth of bacteria over the denture surface.[19] **Soodad A. and Intisar J (2016)** from their pilot study concluded that addition of 3% by weight of silanized zirconium oxide nanofillers had the greatest values of flexural strength, impact strength and hardness.[23]

Surface modification of an inorganic particle with an organic substance is a useful way to reduce its surface energy, increase dispersion, homogeneity and thus improve the properties of the polymer nanocomposites, increasing its compatibility with polymer matrix. [18] Silanes have the ability to bond inorganic particles to organic matrix resulting in improved mixing, better bonding and increased matrix strength.[24]

As limited amount of data is available in literature regarding the effect of incorporation of silanized tetragonal zirconium oxide nanoparticles and silanized aluminium oxide nanoparticles on flexural strength of heat polymerized PMMA, the purpose of this study is to evaluate and compare the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with silanized aluminium oxide nanoparticles and silanized tetragonal zirconium oxide nanoparticles.

Aim and Objectives

Aim:

To evaluate and compare the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles.

Objectives:

1. To evaluate the flexural strength of heat polymerized polymethyl methacrylate denture base material without reinforcement.

2. To evaluate the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles.

3. To evaluate the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 3% silanized tetragonal zirconium oxide nanoparticles.

4. To compare the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles, 3% silanized tetragonal zirconium oxide nanoparticles and heat polymerized polymethyl methacrylate denture base material without reinforcement.

Review of Literature

Polymethyl Methacrylate (PMMA), introduced by **Dr. Walter Wright** in 1937, is still one of the most widely used material in the field of prosthetic dentistry. Despite its popularity which satisfies aesthetics, simple processing and easy repair, its major drawback as a denture base material is low impact and flexural strength. This material is not ideal in every aspect and it is not one single desirable property but the combination of different properties that accounts for its popularity and wide usage.[9,10]

The clinical problems often experienced due to low flexural strength of heat polymerized acrylic resin denture base material has led to many scientific studies to improve mechanical properties of polymethyl methacrylate.

Studies have shown that 68% of the dentures that were fabricated, fractured within the first three years of its daily use.[11] Many trials have been made to enhance the mechanical properties of denture base materials. Ceramic materials are biocompatible and also improve the mechanical properties.[9] Many of these materials have obtained good results in improving the fracture resistance of heat polymerized acrylic resin denture base (PMMA). The prime interest has been the reinforcement of polymers used in dentistry with the metal-composite

In **1774 Alexis Duchateau**, a Parisian apothecary, dissatisfied with his own stained hippopotamus ivory denture and inspired to use porcelain for denture fabrication. He teamed up with Parisian dentist **Nicholas Dubois De Chemant.**[6] Chemant's denture was popular until the introduction of individually baked porcelain teeth in **1808** by an **Italian dentist Giuseppangeio Fonzi**. In this, teeth were attached to the denture base by a small platinum hook. This pin was soldered to a gold denture base. It was one of the most important events in the history of dentistry.[1] **Loomis in 1854** fabricated the first porcelain denture with artificial teeth.[1]

During the latter part of the 19th century, polymers entered the field of denture base materials. **Charles Goodyear (1839)** developed the art of producing rubber and in 1851 his brother Nelson Goodyear invented a process for making hard rubber called vulcanite. Despite its displeasing appearance vulcanite dentures fitted the ridges of the patient more exactly, so that dentures could be worn with comfort.

In **1850 CF Harrington** introduce a **tortoise shell base** that was first the thermoplastic denture base material.[1] In **1851 Edwin Truman** made a **base of gutta percha**. However, the material was unstable and its use required complicated equipment.[1]

In **1856 Alfred A Blandy** made cheoplastic dentures by using low fusing alloy of silver, bismuth and antimony. He embedded a wax model of the denture in plaster of Paris and after melting the wax, he poured the metal compound. Although this metal denture was never accepted, moulding and pouring technique was adopted for manufacturing of vulcanite dentures.[1]

Smith (1957)[8] investigated reinforcement of PMMA by mixing discrete glass fibers with the dough or by lamination with glass cloth and found that reinforcement of fibers did not improved the tensile strength. He also concluded that to strengthen polymer resins by glass fibers, the adhesion of polymer matrix and the fibers should be good because untreated fibers act as inclusion bodies and inhibits the homogenous mixture of acrylic resin and weaken the resin, in spite of strengthening it.

Charles H Land in 1890 made porcelain dentures with platinum bases known as continuous gum dentures. **Alexander Gutowski in 1962** from West Germany made dentures from one piece of porcelain.[1]

D.C. Smith (1961)[14] stated that one of the most practical deficiencies of a denture is fracture. Fracture can occur both; inside as well as outside the mouth. The most common are the midline fractures which act as fatigue fractures. It was found that, whether the denture fractures from accidental or masticatory cause, the strength of the denture has been inadequate in each case. It was also observed that the strength of a denture depends on the shape, residual stress, mechanical properties of the material and condition of loading. It is seen that the incisal notch in a maxillary denture acts as a crack initiator.

Grantt AA and Greener EH (1967)[20] evaluated the flexural property of polymethyl methacrylate by incorporating sapphire (Al_2O_3) whiskers of various diameters. Two types of sapphire whiskers were used; one sapphire whisker of 1-10µ in diameter, silanized and non-silanized was incorporated in a concentration of 8.3% by weight, the other type consisted of sapphire mixed grade fibers incorporated at 10%, 11% and 27% by weight of sapphire. Addition of sapphire whiskers to heat cured polymethyl methacrylate improved the physical properties. With large additions of mixed grade and smaller additions of graded sapphire whiskers the ultimate bending strength was approximately doubled and 25 % changes in the modulus and resilience were noted. The addition of silane increases the surface activity of whiskers permitting better transfer of stresses from polymethyl methacrylate to the whisker. They concluded that an enhancement of the flexural strength of denture base polymethyl methacrylate was possible through the whisker reinforcement with sapphire fibers.

Hargreaves AS (1969)[13] studied the prevalence of dentures fractures. She conducted a survey for 6 months and stated that during that period, there were 113 denture repairs cases and 68% of the dentures fabricated were fractured within the first three years. Women tend to have fractured the denture during eating. She found that habits such as pipe smoking, nail biting or pencil chewing did not play a significant part in denture fracture. This survey showed that upper dentures lost more teeth and fractured in the midline during mastication whereas lower dentures encountered a midline fracture after being dropped. Low thermal conductivity, inferior mechanical strength, brittleness, high coefficient of thermal expansion and relatively low modulus of elasticity makes it more prone to failure during the clinical service.[25]

Fracture of dentures in clinical service has been a concern and several attempts have been made to improve flexural and impact strength of PMMA by giving maximum bulk to the material in the region's most heavily stressed, by copolymerization and cross-linking, reinforcement with carbon fibers.[26]

Chevitarese, O., Craig, R. G., & Peyton, F. A. (1962) investigated the properties of the acrylic resins, the styrene resins, the vinyl-acrylic resins and concluded that the transverse tests established that all four materials were satisfactory with respect to transverse deflection. The styrene type was the stiffest, the vinyl-acrylic resin the toughest, and the epoxy resin the most brittle. [27]

Schreiber CK (1971)[28] studied the effect of carbon fibers to reinforce polymethyl methacrylate by using untreated carbon fibers and untreated chopped carbon fibers. Bundles of fibers were wet with monomer and incorporated into the polymer to form a thin shit within the matrix. The results showed greatest transverse strength by the reinforcement of surface treated carbon fibers which exceeded acrylic by 50%, while adverse effects were shown by untreated carbon fibers.

Berry HH and Funk OJ (1971)[29] stated that midline fracture of the denture is a commonly encountered problem. Breakage may be due to difficulty in cleaning, coughing which pushes the denture out of the mouth, lack of denture base material at the midline, greater than average biting force and dropping the denture accidentally. Breakage is more commonly seen in neuropsychiatric patients, especially those having neuromuscular disorders such as Huntington's chorea, hemiparalysis, muscular dystrophy and Parkinson's disease. They incorporated vitallium as a denture strengthener. This reinforcement used was designed to retain all the qualities of the

acrylic resin denture in addition to adding the needed strength to prevent denture breakage.

Shreiber CK (1971)[28] evaluated and compared the transverse strength of acrylic resin denture base materials, with and without carbon fiber reinforcement. Carbon fibers were in the form of untreated fibers, untreated chopped carbon fibers and surface treated carbon fibers. It was concluded that polymethyl methacrylate reinforced with surface treated carbon fibers had the greatest increase in transverse strength that exceeded the control group by 50%.

Beyli MS and Fraunhofer JA (1981)[12] analysed the causes of fracture of acrylic resin denture. A survey was conducted where 20 laboratories specialized in denture construction and repair were consulted. The result of the survey showed that the ratio of upper to lower denture fracture was about 2:1 with the most common causes of fracture appearing to be poor fit and lack of balanced occlusion. It was also found that the incisal notch was the most important causative factor in midline fractures and crack initiates at the tip of the notch where there is high local stress concentration. Various methods to prevent denture fracture i.e. good processing technique, high strength polymers, denture with metal plate for patients with heavy occlusions, palatal relief in anterior portion of palate, increasing the bulk of base material palatal to incisors and reinforcement in the anterior part of the palate of the denture, have been suggested.

Carroll CE, Fraunhofer JA (1984)[30] conducted a study to determine the effect of the use of commonly available materials to reinforce autopolymerizing acrylic resin. The acrylic resin was reinforced with flat, braided, two-strand brass wire

(Great Lake Ortho Products, Inc., Buffalo, N.Y.) and one of four diameters of orthodontic wires (Unitek Corp., Monrovia, Calif.): 0.016, 0.025, 0.036, and 0.051 inches (0.41, 0.64, 0.91, and 1.30 mm). Statistically significant increase in strength were not obtained consistently until wires of 0.025-inch diameter were used. Clinically significant increase in strength may not be obtained until a wire of at least 0.036-inch diameter was used. The 0.051-inch diameter wire imparted increase in transverse strength that was clinically significant.

Grave, Chandler & Wolfaardt (1985)[10] compared the transverse strength of samples of cross linked acrylic resin with samples containing various percentages of aramid fibers. All of the reinforced specimens were significantly weaker. A possible explanation is the failure of adhesion between the fiber and the matrix resulting in the layers of fiber separating the matrix into layers of narrow cross-section. Contrary to this finding **Berrong, Weed & Young (1990)** reported a significant improvement of impact strength with a fiber content up to 2%. **Mullarky (1985)** also reported an increase in the strength and fatigue resistance of acrylic resin appliances reinforced with unidirectional aramid fiber.

Yazdanie N and Mahood M (1985)[31] investigated the transverse strength of acrylic resin reinforced with varying amounts of carbon fiber in two different lay-ups. The fibers used were strands and woven mat. Both had been silane coated to improve their bonding to acrylic resin. The strands were in the form of long filaments and woven mat was cut according to the size of the test pieces. The orientation of strands was along the long axis of the specimens, whereas the fibers of the mats were parallel with and at right angles to the long axis and sandwiched between thin sheets of acrylic

resin dough. They concluded that carbon fiber acrylic resin composites are stronger and stiffer than unfilled acrylic resin and strands are more efficient strengtheners than woven mats.

Gutteridge DL (1988)[32] studied the effect of including ultra-high-modulus polyethylene fiber on the impact strength of acrylic resin. He concluded that when concentration of fiber used was 1% there was significant increase in impact strength, significant increase was observed up to the concentration of 3% but there was no beneficial effect on impact strength beyond this concentration.

Sejpal SP and Sood VK (1989)[17] studied the effect of metal fillers on thermal conductivity, tensile strength, compressive strength and radio-opacity of PMMA. Silver, copper and aluminium particles of 10 microns were used as fillers. They were added to PMMA in percentages of 5, 10, 15, 20 and 25. The study concluded that with the increase in metal fillers concentration there was a progressive increase in compressive strength of polymethyl methacrylate and a decrease in tensile strength. The maximum increase in thermal conductivity was seen with aluminium, silver and copper fillers in 25% by volume concentration.

Berrong JM, Weed RM and Young JM (1990)[33] conducted a study on the fracture resistance of polymethyl methacrylate denture base resin reinforced with kevlar fibers embedded in the ratio of 0% (control), 0.5%, 1% and 2% by weight of PMMA resin specimen. All the specimens were subjected to impact testing. The result showed that all reinforced sample groups had greater fracture resistance than the unreinforced control group. They concluded that the use of kevlar fibers up to 2% as reinforcement might increase resistance of the resin denture base.

Solnit GS (1991)[34] studied the effect of methyl methacrylate reinforcement with silane-treated and untreated glass fibers. Cloth form glass fibers, loose form yellow glass fibers and loose form white glass fibers were used. They were soaked in a silane coupling agent for 5 minutes and air dried before incorporation. This study suggest that glass fibers can be pre-treated with a silane coupling agent to obtain a chemical bond between the fibers and the acrylic resin. Samples with untreated fibers tested weaker than samples without fibers. Samples with silane-treated tested stronger but the difference in strength was not statistically significant.

Vallittu PK and Lasilla VP (1992)[35,36] studied the effect of different types of commonly used metal wires and glass fibers, as well as carbon and aramid fibers on the fracture resistance of polymethyl methacrylate. Metal strengtheners used were remanium's spring hard clasp wire, semi-circular wire, braided wire plate, stainless steel mesh and they were further divided into: glossy and sandblasted. Fibers were divided into 2 groups untreated and silanized. They concluded that all metal strengtheners increased the fracture resistance except stainless steel mesh. They found that the unsilanized glass fibers slightly weakened the test specimens. However, the weakening was not statistically significant when compared to the silanized glass fibers, which had a significant strengthening effect. They also studied the effect of surface roughness of various metal wires on the fracture resistance of the acrylic test specimens. Metal strengtheners used were semicircle wire (1.0 x 2.0mm and 1.25 x 2.50mm), braided wire plate (0.8x 2.4mm) and spring hard clasp wire (1.0 mm). The sandblasting was done using aluminium oxide particles with grain sizes 50 µm and 250µm, and the air pressure applied was 5.5 bar. Roughening of wires was done by grinding them with a heatless stone and with a 0.6 mm separating disc. All the wires

were cleaned in water with an ultrasonic cleaning device. The semi-circular wire had maximum effect on the fracture resistance of the specimens. The best results were obtained by sandblasting procedure. The resistance was not influenced by the grain size of the sand (50 µm or 250 µm).

Ladizesky NH, Cheng YY, Chow TW, Ward IM (1993)[37] evaluated 3 mechanical properties i.e. flexural strength, flexural modulus and impact strength of acrylic resins reinforced with woven highly drawn linear polyethylene fibers (HDLPE) and they concluded that incorporation of the polyethylene fibers in woven form, substantial improvements in impact strength was obtained. They concluded that when chopped high performance polyethylene fibers were incorporated 30% by volume in acrylic denture base resins, it resulted in substantial improvement in several important properties like flexural stiffness and impact strength that were higher; Water sorption, polymerization shrinkage, dimensional changes during water immersion significantly decreased, and it also removed the weakening effect of anatomical features such as the frenal notch.

Vallittu PK (1993)[38] studied the effects of two different silane compounds on the adhesion between different fibers and acrylic resin. The fibers used were glass, carbon and aramid fibers and each type of fibers were either untreated or treated with silane agents. The fibers were studied by a scanning electron microscope (SEM) to establish the adhesion between the fibers and acrylic resin and the fracture resistance of the specimens were tested. The results showed that the adhesion between the fibers and acrylic resin improved with silanization of glass and aramid fibers. SEM photographs taken confirmed these findings.

He also studied the effect of metal wire bonding to acrylic resin on the fracture resistance of an acrylic denture base material construction. Two different bonding methods were tested: Silicoating (group B19) and Eudicolle (group B18). The fracture resistance of the test specimens reinforced with sandblasted metal wires were higher than the resistance of the control specimens. The bonding method of Silicoater increased the resistance significantly whereas the bonding compound of Eudiclle did not increase the fracture resistance when compared with unsilanized strengtheners. The different positions of the wires had no effect on the fracture resistance.[39]

Marie M et al (1994)[40] evaluated four physical and mechanical properties: thermal conductivity, impact strength, compressive strength and warpage by adding tin or aluminium powder with particle size of 10μm to heat cure acrylic resin in a concentration of 5% by volume. The addition of 5% by volume of both the metal powders to polymethyl methacrylate (PMMA) improved the four tested properties. However, aluminium powder was superior to tin powder in improving thermal conductivity of PMMA and decreasing its warpage. Tin powder was superior to aluminium powder in improving the impact and compressive strength of PMMA. The use of these metal-filled resin is recommended in the areas where it is not displayed because both metal powders caused undesirable discoloration to the heat cure acrylic resin.

Vallittu PK, Vojtkova H, Lassila V (1995)[41] compared the impact strength of heat-cured acrylic resin specimens reinforced with 1.0-mm-diameter steel wire and continuous E-glass fibers. When compared to unreinforced specimen it was found that

both types of reinforcements increased the impact strength of the resin. They concluded that concentrations of glass fiber greater than 25 wt.% yield better impact strength than steel wire 1.0 mm in diameter.

Vallittu PK (1997)[42] conducted a study wherein the clinical usefulness of continuous E-glass partial fiber reinforcement of acrylic resin removable dentures was evaluated. Twelve removable complete dentures and ten removable partial dentures with a history of recurrent fractures were selected for this study. The partial fiber reinforcement was incorporated into the denture at the time of repair and the dentures were evaluated at an average of 13 months after the insertion of the fibers. He stated that when continuous, unidirectional glass partial fiber reinforcement is used in dentures, the fibers should be oriented at a 90-degree angle to the potential fracture line and placed as near as possible to the denture margin, which is prone to fracture. Reinforcement of other weak regions of the denture is recommended to prevent the occurrence of new fractures.

Stipho HD (1998)[43] investigated the transverse strength, maximum deflection, and modulus of elasticity of repaired acrylic resin joints reinforced with different concentrations of glass fibers to the weight of the powder/ liquid mix (0%, 1%, 2%, 5%, 10%, and 15%). Transverse strength, maximum deflection and the stiffness of all joints were significantly lower after the repair. Among the groups tested, the specimens treated with 1% glass fiber displayed the highest transverse strength before and after the repair. Modulus of elasticity of the repaired 1% fiber concentration units was improved by approximately 25% when compared to specimens which were repaired but untreated with glass fiber (0% fiber).

Phillip B. Messersmith, Ales Obrez, and Sara Lindberg (1998)[44] evaluated the thermal diffusivity of new acrylic resin composite**.** Thermal diffusivity is one material property that has been cited as being important in determining gustatory response. Denture base acrylic resins have low thermal diffusivity compared with denture base metal alloys. Sapphire (Al_2O_3) whiskers were added to conventional denture base acrylic resin during processing to achieve loadings of 9.35% and 15% by volume. Thermal diffusivities of the sapphire-containing composites were found to be significantly higher than the unmodified acrylic resin. Thermal diffusivity was found to increase in proportion to the volume percentage of sapphire filler, which suggested that the high aspect ratio ceramic particles formed a pathway for heat conduction through the insulating polymer matrix.

Jagger DC and Harrison A (1999)[10] studied the transverse strength of reinforcement of acrylic resin with chopped polymethyl methacrylate fibers. 0.75 mm in diameter and 5 mm in length of polymethyl methacrylate fibers were added to the denture base material in 0%, 5%, 15%, 20% & 25% by weight. The study concluded that randomly arranged chopped polymethyl methacrylate fibers have no advantage on acrylic resin when compared to unmodified polymer in terms of strength and chopped polymethyl methacrylate cannot be mentioned as reinforcing agent for denture base material.

Vallittu PK (1999)[45] evaluated the flexural properties of acrylic resin polymers reinforced with unidirectional and woven glass fibers. Heat-curing and autopolymerizing denture base polymers were reinforced with continuous unidirectional and woven pre-impregnated glass fibers and he concluded that the

increase in transverse strength and flexural modulus of the polymers was seen with Stick reinforcement, increased strain at fracture of the polymers was seen with Stick Net reinforcement & SEM examination revealed that fibers of Stick and Stick Net were well impregnated with the resin of polymer matrix.

Kanie T, Fujii K, Arikawa H and Inoue K (2000)[15] studied the reinforcing effect of woven glass fibers on deflection, flexural strength, flexural modulus and impact strength of acrylic denture base polymer. The flexural strength and deflection was significantly higher in specimens reinforced with silanized glass fibers of 1 mm thickness than those of unreinforced specimens. Also, the impact strength was significantly higher in specimens reinforced with silanized glass fiber of 2 mm thickness than that of unreinforced specimens.

Gulay Uzun, Nur Hersek and Teoman Tinçer (1999)[46] measured the effect of 5 fiber strengtheners on the fracture resistance of denture base acrylic resin material. Impact strength, transverse strength, deflection and elastic modulus values of a heat-polymerized denture base resin (Trevalon) reinforced with glass, carbon, thin kevlar, thick kevlar, and polyethylene fibers in woven form were studied. Polyethylene and glass-reinforced acrylic resin specimens were significantly more resistant to impact strength. Fiber reinforcement had no significant effect on the transverse strength, polyethylene reinforcement significantly raised deflection value. Carbon, thick Kevlar, and polyethylene-reinforced specimens showed significantly higher elastic modulus values.

Jagger D et al (2001)[47] studied the effect of the addition of surface treated chopped and continuous poly (methyl methacrylate) fibers on some properties of

acrylic resin. Properties like transverse and impact strengths were investigated. The fibers were added in three arrangements (i) a single unidirectional layer (longitudinal in the direction of the specimen), (ii) two unidirectional layers and (iii) two layers in cross ply (an inferior layer at 0° and a superior layer at 90° to the direction of the length of the specimen). The addition of surface treated chopped or continuous fibers to acrylic resin did not improve the transverse or impact strengths over unmodified specimens, therefore cannot be recommended as a method of reinforcement.

Foo H et al (2001)[48] evaluated the effect of unidirectional poly-aramid fiber reinforcement on the transverse strength of intact and repaired heat-polymerized denture base acrylic resins. The denture base resins tested were Acron MC (microwave polymerized resin), Lucitone 199 (butyl rubber reinforced high impact resin) and Microlon (conventional denture base resin). They concluded that poly-aramid reinforcement significantly increased the transverse strength of intact heat-polymerized PMMA resins when compared to the control group. Use of poly-aramid reinforcement in repair of unreinforced PMMA and poly-aramid reinforced PMMA did not significantly increase transverse strength.

Chen S, Liang W, Yen PS (2001)[49] evaluated the mechanical properties of acrylic resin reinforced with three types of fiber-polyester fibers (PE), kevlar fibers (KF), and glass fibers (GF) cut into 2, 4 and 6 mm lengths and incorporated at concentrations of 1, 2 and 3% (w/w). Polyester fiber and kevlar fiber both improved the mechanical properties, but the polyester fiber was superior in aesthetics. There was significant increase in the impact strength of acrylic resin reinforced with 3% (w/w) of 6 mm polyester fiber when compared to other formulations. In this study

they concluded that the optimum formulation to reinforce acrylic resin for improved strength is by incorporation of 3% and 6 mm length polyester fibers.

Aydin C, Yilmaz H, Çaglar A (2002)[50] investigated the effect of a glass fiber reinforcement system on the flexural strength of three different denture base resins. Three denture base resins selected for testing: (1) heat-cured resin (QC 20 De Trey/ Dentsply), (2) an autopolymerized resin (Vertex, Dentimex), and (3) a light-activated resin (Triad, Dentsply). Stick (S) and Stick Net (SN) (Stick Tech) were used to reinforce the three denture base polymers. The results showed that different denture base resins when reinforced with the glass fiber may be a useful means of improving the flexural strength thereby improving the performance of dental acrylic resins.

T Kanie et al (2003)[15] studied the flexural properties of denture base polymers reinforced with glass cloth–urethane polymer composite. The silanized glass cloth was sandwiched between two pieces of polyethylene film and pressed to form a sheet of 0.3 mm in thickness, which was light-cured and prepared using four different surface conditions: with or without the polyethylene film and with or without a bonding agent. The results showed that baseline flexural strengths of the self-curing resins, heat-curing resins, and light-curing resins were 76.2, 68.6, and 55.6 MPa, respectively, and these values were increased to 271.7, 216.4, and 266.5 MPa by the reinforcement sheet.

Zappini, G., Kammann, A., & Wachter, W. (2003)[51] studied the fracture toughness of denture base resins and to compare the results with impact strength and flexural strength measurements and concluded that specimen geometry and testing configuration influenced the impact strength measurements. The fracture toughness

method seems to be more suitable than impact strength measurements to demonstrate the effects of resin modifications.

Kim SH, Watts DC (2004)[52] studied the effect of glass fiber reinforcement on the impact strength of high impact acrylic resin maxillary dentures. They drew the following conclusion: at crack initiation and at complete fracture, the impact strength and crack propagation energy of high impact acrylic maxillary complete denture reinforced with woven E-glass fibers was higher than that of the unreinforced denture.

Matinlinna J et Al (2004)[53] reviewed silanes and their clinical application in dentistry. They stated silanes, hybrid organic-inorganic compounds, function as mediators and through dual reactivity promote adhesion between dissimilar, inorganic and organic matrices. They are called primers, coupling agents, or sizes, depending on their function and substrates. Methacryloxypropyltrimethoxysilane (or 3-trimethoxysilylpropyl methacrylate [MPS]) is most commonly applied silane in dental laboratories and chairside. MPS is used to optimize and promote the adhesion, through chemical and physical coupling, between metal-composite, ceramic-composite and composite- composite.

Franklin P, Wood DJ and Bubb NL (2005)[54] evaluated the effect of adding glass flake to denture base acrylic powder on the fracture toughness of the set material. Glass flake was added in 5, 10 or 20% w/w to Trevalon denture base powder. They concluded that improvement in fracture toughness of a denture base acrylic material using glass flake is an extremely promising result but other mechanical properties would require testing before glass flake can be recommended as a reinforcing material.

Tacir IH, JD Kama, M Zortuk, S Eskimez (2006)[55] compared the fracture resistance of unreinforced and glass fiber reinforced acrylic resin polymers prepared under both conventional heat-curing and microwave curing techniques. They concluded that fracture resistance and impact strength of heat polymerized acrylic resin improved with glass fiber reinforcement. It may be possible to apply these results to distal extension partial and complete denture bases.

Vojdani M, Khaledi AAR. (2006)[56] conducted a study in which they observed that the transverse strength of heat polymerized denture base resin was considerably enhanced by incorporating either metal wires or glass fibers. S fiber reinforcement increased the transverse strength of PMMA upto 50%, StickNet SN fiber reinforcement increased the transverse strength upto 30% and reinforcing with wire increased the transverse strength of PMMA upto 14%. Moreover, the flexural strength of specimens reinforced with continuous unidirectional glass fibers was significantly higher than that of metal wire or woven fiber reinforcements.

Goyal S (2006)[24] reviewed the use of silanes in dentistry. She said that one reactive group of the silane (eg. methoxy, ethoxy and silanolic hydroxy groups) reacts with various inorganic materials such as glass, metals, silica, sand to form a chemical bond with the surface of the inorganic material, while the other of the reactive groups (e.g., vinyl, epoxy, methacryl, amino and mercapto groups) is reactive with various kinds of organic materials or synthetic resins to form a chemical bond. As a result of possessing these two types of reactive groups, silanes are capable of providing chemical bonding between an organic material and an inorganic material. This unique property of silanes is utilized for the surface treatment of glass fiber products,

performance improvement of fiber-reinforced plastics by the direct admixture to the synthetic resin, improvement of paints and other coating materials, and adhesives, modification of surface properties of inorganic fillers, surface priming of various substrate materials. When used as a coupling agent, silanes bind organic polymers to mineral or siliceous fillers, resulting in improved mixing, better bonding of pigment or fillers to resins, increased matrix strength, decreased water intake of composite and minimize wear. She stressed that silanization of E-glass and aramid fibers enhances the adhesion between the fibers and organic acrylic resin system in a denture material and long-range intraoral stain protection has been accomplished in a denture, when its surface was modified with a fluorocarbon chain containing silane.

Dogan O, Bolayir G, Keskin S, Dogan A, Bek B, Boztug A (2007)[57] evaluated the changes in impact resistance of a denture base resin reinforced with five different types of fibers. E-glass, polyester, rayon, nylon 6, and nylon 6,6 fibers were cut into 2, 4, and 6 mm lengths and added into the resin at a concentration of 3% by weight. The results indicated that impact energy tended to increase with fiber length, and that the highest value was recorded for rayon fiber-reinforced specimens of 6 mm length. E-glass fiber reinforcement produced relatively stable, high values for each length.

Nakamura M, Takahashi H, Hayakawa I (2007)[58] evaluated flexural strengths and moduli by reinforcing denture base resins with high short-rod fiber. A commercial PMMA (AC; average particle size, 150μm) and an industrial PMMA powder (MB; average particle size, 4μm) was used. Short-rod glass fibers were mixed with two powders at a mass ratio of 0 to 50%. The flexural strength of MB

composites increased significantly at fiber contents exceeding 40%. The flexural moduli were significantly greater for AC and MB composites at fiber contents exceeding 20% than those of control group respectively.

Orhan MD, Giray B, Selda K, Arife D, Bu¨ lent B (2008)[59] conducted a study to determine whether some flexural properties of a denture base resin material could be improved through reinforcement with aesthetic fibers i.e. glass, rayon, polyester, nylon 6 and nylon 6,6 fibers at 3% concentration by weight and in 2, 4, and 6 mm length. It was found that specimens reinforced with nylon 6,6 fibers of 6 mm length showed the highest flexural strength. Young's modulus and maximum load suggested that such reinforcement made resin resistant to fracture.

Ellakwa AE, Morsy MA, El-Sheikh AM (2008)[9] conducted a study to evaluate the effect on the flexural strength and thermal diffusivity of heat-polymerized acrylic resin on adding 0%, 5%, 10%, 15%, and 20% by weight aluminium oxide powder (Al_2O_3). The study concluded that incorporating Al_2O_3 powder from 5% to 20% by weight into conventional heat-polymerized denture base resin resulted in an increase in both flexural strength and thermal diffusivity over control samples. The flexural strength increased significantly after incorporation of 10% aluminium oxide powder. The highest mean flexural strength was seen with addition of 15% by weight of aluminium oxide powder. Thus, increasing the flexural strength and heat transfer characteristics of the acrylic resin base material could result in more patient satisfaction.

Dagar SR, Pakhan AJ, Thombare RU, Motwani BK (2008)[60] evaluated the flexural and impact strength of commercially available heat polymerizing PMMA

denture base resin reinforced with glass and nylon fibers. It was concluded that fiber-reinforced specimens were more resistant to impact and flexural fatigue than conventional PMMA specimens. When compared with nylon fiber reinforcement it was found that glass fiber reinforcement considerably improves both impact and flexural strengths of denture base resin. Silane-impregnated glass fiber reinforcement suits best to increase the flexural and impact strengths of heat-polymerized PMMA denture base resin.

Ayad NM, Badawi MF, Fatah AA (2008)[61] evaluated the effect of reinforcing high-impact acrylic resin (Metrocryl HI) with zirconia powder in two different concentrations (5% and 15%) on the impact strength, transverse strength, water sorption, surface hardness and solubility. They concluded that the addition of zirconia resulted in a highly significant increase in transverse strength of high-impact acrylic resin when compared to control samples by a factor of 29% and 76% in a concentration of 5% and 15% respectively.

Vojvodic et al (2009)[62] studied the effect of different glass fibers ("dental" and "industrial" origin) on flexural strength values of dental base polymers. Flexural strength of the control specimens was 91.76 MPa, while there was a rise of flexural strength values (103.10-163.88 MPa) in specimens reinforced with glass fibers irrespective of type ("dental" or "industrial") fibers used but due to better investment to benefit ratio "industrial" glass fibers could be recommended for dental laboratory use.

Elshereksi NW et al (2009)[63] investigated the effect of barium titanate ($BaTiO_3$) filler incorporation on the fracture toughness properties of denture base

Poly(Methyl Methacrylate). The $BaTiO_3$ was treated by a silane coupling agent, 3-trimethoxysilylpropyl methacrylate (γ-MPS) before incorporating into PMMA. The samples were tested for fracture toughness before and after soaking for 28 days in simulated body fluid (SBF). They concluded that the dry samples provided higher fracture toughness than the wet samples.

Abdulhamed AN, Mohammed AM (2010)[64] evaluated the thermal conductivity, impact and tensile strength of alumina reinforced heat cure acrylic resin. Alumina powder was added to PMMA powder by weight in three different percentages 5%, 7.5% and 10%. The study concluded that the addition of Al_2O_3 powder to acrylic resin improves the thermal conductivity, decreases both tensile and impact strength values. Further there was an increase in surface hardness. Water sorption and solubility were decreased while surface roughness was not affected with small percentages of alumina.

Arora N, Jain V, Chawla A, Mathur VP (2011)[65] studied the effect of adding silver filler particles and sapphire (aluminium oxide) on the flexural strength, thermal diffusivity and water sorption of polymethyl methacrylate (PMMA) resin. Silver filler particles and sapphire (aluminium oxide) were added in 25% by weight of acrylic resin. The study concluded that as compared to silver fillers, sapphire fillers were better for the reinforcement of polymethyl methacrylate resin. This is because they have low density, are highly esthetic and bring about an improvement in the mechanical properties (flexural strength and fatigue strength) and thermal properties (thermal diffusivity) of polymethylmethacrylate (PMMA) resin.

Ihab NS, Moudhaffar M (2011)[66] studied the effect of addition of modified nano-zirconium oxide (ZrO_2) particles in various percentages on strength and radio-opacity of heat cured acrylic denture base material. The nanoparticles were coated with a layer of trimethoxysilypropylmethacrylate (TMSPM). They were sonicated in monomer (MMA) in different percentages 2%, 3%, 5% and 7% by weight. It was found that maximum increase in transverse strength, impact strength, and radio-opacity was observed in denture base reinforced with 5wt% of nano-ZrO_2.

Alhareb AO, Ahmad ZA (2011)[67] evaluated the effect of 5wt% of Al_2O_3/ZrO_2 (80:20 ratio) reinforcement on the fracture toughness, flexural, and tensile properties of PMMA denture base. They concluded that the PMMA reinforced with Al_2O_3/ZrO_2 improved the tensile modulus, fracture toughness and flexural properties of the denture base material.

Saritha MK, Shadakshari S, Nandeeshwar DB (2012)[8] conducted an *in vitro* study to investigate the flexural strength of conventional heat polymerized denture base resin reinforced with 5%, 10% and 15% by wt. of aluminium oxide powder. They concluded that incorporation of 10% (group-C) and 15% (group-D) by wt. aluminium oxide powder to heat cure denture base resin significantly increased the flexural strength of denture base resin. The highest flexural strength was found with incorporation of 15% by wt aluminium oxide powder to heat cure denture base resin.

Yadav P, Mittal R, Sood VK, Garg R (2012)[68] studied the effect of incorporating metal filler particles on different strengths and thermal conductivity of polymethyl methacrylate (PMMA). The study was carried out in two parts. Part 1 was

an in vitro investigation regarding the effect of incorporating metal fillers (aluminium and silver) in the concentrations of 10%, 20%, and 30%, by volume on the tensile, compressive and flexural strength of PMMA. Part 2 of the study comprised of clinical evaluation of the thermal perception by 10 edentulous patients. Each patient was given two sets of complete dentures, one fabricated with unfilled PMMA and another with 20% aluminium particle filled PMMA on the palatal portion of the maxillary denture. They concluded that compressive strength increased progressively on increasing the filler concentration for both silver and aluminium filled PMMA. At 30% concentration silane-treated metalized PMMA showed reduction in tensile and flexural strength and incorporating metal filler particles led to an appreciable increase in thermal perception by the participants of this study.

Ihab NS, Hassanen KA, Ali N.A (2012)[18] evaluated impact strength, tensile strength and color stability of heat polymerized denture base reinforced with silanated and non-silanated zirconium oxide (ZrO_2) nanofillers. Silanization was done by coating with a layer of trimethoxysilypropylmethacrylate (TMSPM) and nanoparticles were sonicated in monomer (MMA) in two percentages of 3% and 5% by weight of polymer. The maximum increase in impact strength was observed in PMMA reinforced with 5%wt of silanated ZrO_2 nano-fillers. Silanized ZrO_2 nano-fillers were effective in improving impact strength while it was not effective in improving the tensile strength. Also, significant color differences were seen between control group and specimens incorporated with zirconium oxide nano-fillers in different immersion solutions.

M Vojdani et al (2012)[69] studied the effect of aluminium oxide reinforcement on the flexural strength, surface hardness and roughness of heat polymerized acrylic

resin. 0.5%, 1%, 2.5% and 5% of aluminium oxide was measured and flexural strength was assessed with a three-point bending test using a universal testing machine. The results showed 2.5% of Al_2O_3 significantly increased the flexural strength compared to the control group. The Vickers hardness significantly increased after incorporation of 2.5 and 5% Al_2O_3. No significant difference was detected in surface roughness levels between the reinforced and control groups.

Chaijareenont P (2012)[70] evaluated different amounts of 3 methacryloxypropyltrimethoxysilane (MPS) coated alumina filler particles on flexural strength of alumina reinforced polymethyl methacrylate (PMMA) denture base. Ten mass % of alumina filler silanized with 0, 0.1, 0.2, and 0.4 mass% of MPS was blended with PMMA. They concluded that the 0.1 mass% MPS silanized group had significantly higher flexural strength than the control group.

Asar NV, Albayrak H, Korkmaz T, Turkyilmaz I (2013)[71] evaluated the effect of addition of different types and percentages of metal oxides on the mechanical and physical properties of heat cured PMMA. They concluded that addition of Al_2O_3, TiO_2 and ZrO_2 fillers resulted in significant increase in impact strength and fracture toughness and significant decrease in water sorption and solubility. Therefore, reinforcing heat cure PMMA with certain amounts of metal oxides may be useful in preventing fracture of denture bases and unwanted physical changes resulting from oral fluids clinically.

Sodagar A et al (2013)[72] evaluated the effect of addition of TiO_2 and SiO_2 nano-particles on flexural strength of polymethyl methacrylate acrylic resins. They were divided into seven groups: control group and AR containing nano TiO_2 (2), SiO_2

(2) and TiO_2 (2) with SiO_2 (2) in two concentration of 1% and 0.5%. The maximum mean flexural strength (43.5 MPa) was seen with the control group. Incorporation of nano-particles into polymethyl methacrylate acrylic resin caused these particles to agglomerate and aggregate, adversely affecting the flexural strength of the final products and this effect was directly associated with the concentration of nano-particles.

Atla J et al (2013)[19] studied the effect of adding 5% to 20% by weight aluminium oxide powder (Al_2O_3) on thermal diffusivity of heat–polymerized acrylic resin. Incorporating Al_2O_3 powder of 20% by weight into conventional heat polymerized denture base resin, resulted in an increased thermal diffusivity, which ensured a better perception of temperature changes and which ultimately led to enhanced patient satisfaction

Ahmed MA, Ebrahim MI (2014)[16] studied the effect of addition of zirconium oxide (ZrO_2) nano-fillers powder in different concentrations (1.5%, 3%, 5% and 7%) on the flexural strength, fracture toughness, and hardness of heat-polymerized acrylic resin and concluded that the addition of Zirconium oxide nanofillers to PMMA significantly increased the flexural strength, fracture toughness and hardness. Also, according to this study the maximum mechanical properties were observed in heat-polymerized acrylic resin reinforced with 7% of zirconium oxide nanofillers.

Jasim BS and Ismail IJ (2014)[21] study was to evaluate the effect of addition of surface treated Aluminium oxide nanofillers on some properties of heat cured (PMMA) where silanized (Al_2O_3) nanoparticles was added to PMMA powder by weight in three different percentages 1wt%, 2wt% and 3wt% and mixed by probe

ultrasonication machine. A highly significant increase in transverse strength was observed with the addition of (Al_2O_3) nanoparticles to (PMMA) at the percentage of 1wt%, the value was 117.72 Mpa and significant increase at 2wt%; while a significant reduction occurred in transverse strength at the percentage of 3% the value was 90.110 Mpa. They concluded that addition of Al_2O_3 nanoparticles to acrylic resin improves the thermal properties and transverse strength of acrylic resin at the same time this addition decreases water sorption and solubility.

Girish Nazirkar et al (2014)[73] evaluated and compared the effect of different concentration of TiO_2 NP on the flexural strength of PMMA resins. Specimens made from heat polymerizing resin (DPI) without NP were used as a control group (Group A). The two experimental groups, (Group B and Group C) had 0.5 and 1 % concentration of TiO_2 NP respectively. They concluded that the maximum mean flexural strength (90.65 MPa) belonged to the control group; and acrylic resin with 1 % TiO_2 NP demonstrated the minimum mean flexural strength (76.38 MPa). Addition of TiO_2 NP into acrylic resin can adversely affect the flexural strength of the final product and is directly proportional to the concentration of NP. They concluded that addition of metal oxides increases the flexural strength of a denture base material.

Amedi-Rad F et al (2014)[74] compared of the thermal conductivity, compressive strength, and tensile strength of the acrylic base of complete dentures with those of acrylic reinforced with nanosilver. PMMA was mixed with 5 weight% nanosilver. They found out that the thermal conductivity and compressive strength of PMMA reinforced with nanosilver were significantly higher than the unmodified PMMA, while the tensile strength decreased significantly after the incorporation of

nanosilver. They recommended the use of this material in the palatal area of maxillary acrylic resin dentures.

Zhang XY et al (2014)[75] studied the hybrid effects of ZrO_2 nanoparticles (nano- ZrO_2) and aluminium borate whiskers (ABWs) on flexural strength and surface hardness of denture base resin, polymethyl methacrylate (PMMA). ZrO_2 nanoparticles of an average granularity of 90 nm and an average surface area of 7±2 m^2/g and Aluminium borate whiskers of 5–30 μm length and a diameter of less than 1.5 nm, with a surface area of 2.0–2.5 m^2/g were used. Both nano-ZrO_2 and ABWs were modified by silane coupling agent i.e. Z-6030 before being mixed with PMMA. They concluded that the flexural strength was maximum when 2 wt% of nano-ZrO_2 was mixed with ABWs at a ZrO_2/ABW ratio of 1:2, amounting to an increase of 52% when compared with pure PMMA.

Yu W et al (2014)[76] evaluated the reinforcement of denture base PMMA with ZrO_2 nanotubes and compared it with ZrO_2 nanoparticles. ZrO_2 nanotubes were prepared by anodization; they were pre-stirred with the denture base PMMA powder by a mechanical blender and mixed with methyl methacrylate liquid to fabricate reinforced composites. Silane coupling reagent (A-151, triethoxyvinylsilane) was used for silanizing ZrO_2 nanotubes and ZrO_2 nanoparticles. The results indicated that ZrO_2 nanotubes had a better reinforcement effect than ZrO_2 nanoparticles, and surface-treatment lowered the reinforcement effect of the ZrO_2 nanotubes which itself was significantly different from that of the ZrO_2 nanoparticles. The flexural strength of the composite was maximized when 2.0 wt.% untreated ZrO_2 nanotubes were added.

Anne G et al (2015)[77] conducted a study to investigate the effect of adding 5-20% aluminium oxide (Al_2O_3) powder by weight on the flexural strength of heat-polymerized acrylic resin. The specimens of the remaining were reinforced with Al_2O_3 powder to achieve loadings of 5%, 10%, 15% and 20% by weight. The study concluded that Al_2O_3 fillers are potential components to be added in denture bases to provide increased flexural strength.

Asopa V et al (2015)[78] evaluated and compared the transverse strength, impact strength, surface hardness and water sorption of 10% and 20% zirconia (ZrO_2) reinforced high impact acrylic resin. They concluded that there was increase in transverse strength on addition of zirconium oxide as a filler in the high impact acrylic resin. On the other hand, impact strength and surface hardness of the zirconia reinforced specimens were found to have relatively lesser values when compared to control specimens. Water sorption of the zirconia reinforced specimens was found to increase but was within the limit of ADA Specifications No. 12.

Kareem S, Moudhaffer M (2015)[79] studied the effect of silanized zirconium silicate nanopowder reinforcement on some mechanical and physical properties of heat cured poly (Methyl Methacrylate) denture base material. Silanization was done by coating zirconium silicate nanoparticles with a layer of trimethoxysilylpropylmethacrylate (TMSPM). Addition of silanized zirconium silicate nano fillers was done in two groups -1% and 1.5% by weight of polymer. They concluded that the maximum increase in impact strength, transverse strength, and surface hardness was observed in denture base nanocomposites containing 1.5% zirconium silicate nanoparticles.

Arora et al (2017)[65] evaluated the flexural strength, hardness, and impact strength of heat cured high impact denture base resins with different polymer/monomer ratios. The samples were divided into five groups based on different powder/liquid ratios (g/ml). The polymer/monomer ratio in Group 1 (Ratio 2.2:1) was the manufacturer's recommended ratio and was used as control. In Group 2, the ratio was 2.7:1, in Group 3, the ratio was 3.2:1, in Group 4, the ratio was 1.9:1, and Group 5 the ratio was 1.6:1. The results showed that the flexural strength values and VHN values showed a similar trend. The values decreased significantly as the ratio was increased or decreased from the control group. He concluded that for reinforcing resins or high impact resins, the manufacturer's recommended polymer/monomer mixing ratio should be used to obtain the appropriate strength of the material.

Gad et al (2017)[80] reviewed the effect of fibers, fillers, and nanofillers addition on polymethyl methacrylate (PMMA) properties and he concluded that there was a significant improvement in the mechanical properties of PMMA when reinforced with glass fibers. Natural fibers (OPEFB) and vegetable fibers can be used, but further studies are needed. Also, Addition of nanoparticles and nanotubes to denture base materials results in enhancement of mechanical properties of PMMA, depending on the application and manipulation. It was also concluded that silane coupling agent improves the bonding between fillers and the resin matrix and thus bring about an improvement in mechanical properties of the resins. Hybrid fiber, hybrid fillers, or hybrid fiber and filler are the newest reinforcement system and they may considerably enhance the properties of PMMA.

Gad M. et al (2019)[81] evaluated the hybrid reinforcement effects of zirconium oxide nanoparticles (nano-ZrO_2) and glass fibers (GFs) at different ratios and percentages on the flexural and impact strengths of a polymethylmethacrylate (PMMA) denture base. It was seen that the addition of nano-ZrO_2, GFs or nano-ZrO_2 + GFs to PMMA denture base materials improves the flexural strength, also the addition of nano-ZrO_2 or nano-ZrO_2 + GFs to PMMA denture base materials improves the impact strength and 95% PMMA+ 2.5% nano-ZrO_2 + 2.5% GF composites have the best balance of flexural strength and impact strength, and this ratio is recommended as the hybrid reinforcement for denture base materials.

Materials and Method

Flexural failure of denture base resins is considered as the dominant means of clinical failure.[82] Consequently, the definitive flexural strength of a material reflects its potential to oppose catastrophic failure under a flexural load. High flexural strength is fundamental to the long-term success of dentures. Modifications in the composition of conventional acrylic resin denture base material can be done to attain this purpose.

Lately, plentiful attention has been directed towards the incorporation of inorganic nanoparticles into PMMA to improve its properties. The properties of polymer nanocomposites depend on the type of incorporating nanoparticles, their size, shape, as well as the concentration and interaction with the polymer matrix.

Nanoparticles are surface treated with silane coupling agent and inserted into PMMA.[83]

Hence, this *in-vitro* study was done to evaluate and compare the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles.

Materials and methods have been divided under the following heads:

I. Materials

II. Armamentarium and equipments

III. Method

I. Materials (PLATE I)

SR. NO.	MATERIALS	MANUFACTURER	BATCH NO.
1	Heat polymerized acrylic resin (Fig 1)	DPI Heat Cure™, (Dental products of India Ltd)	8174
2	Die stone (Fig 2)	Ultrarock; Kalabhai Karson Pvt Ltd, India	190603
3	Aluminium oxide nanoparticles (Fig 3)	Reinste	A-AlO-132
4	Tetragonal zirconium oxide nanoparticles (Fig 4)	Sigma Aldrich	DCZrOAK41
5	Silane coupling agent (3-Trimethoxypropylsilylmethacrylate TMPSM) (Fig 5)	Sigma Aldrich	M0725
6	Toluene (Fig 6)	MERCK	1811610317
7	Cold mould seal (separating medium) (Fig 16)	PYRAX	CMS-085

II. Armamentarium and equipment (PLATE II & III)

1. High accuracy balance (Fig 7)
2. Ultrasonicator (Fig 8)
3. Magnetic stirrer (Fig 9)
4. Vacuum rotary evaporator (Fig 10)
5. Acrylizer with thermostat (Fig 11)
6. Universal testing machine (Fig 12)
7. Rubber bowls and plaster spatula (Fig 13)
8. Varsity flasks and clamps (Fig 13)
9. Sand paper (No. 120) (Fig 14)
10. Camel hair brush (Fig 14)
11. Glass Beaker
12. Sterile Syringe
13. Mixing spatula
14. Petroleum jelly (Fig 14)
15. Porcelain jar and Dapen dish (Fig 14)
16. Vernier caliper (Fig 15)
17. Brass metal dies (Fig 17)
18. Para-film (Fig 18)
19. Hydraulic bench press (Fig 19)
20. Distilled water (Fig 20)

III. Methodology (PLATE IV)

The basic methodology consisted of –

a. Die preparation

b. Silanization of tetragonal zirconium oxide nanoparticles (Fig 21)

c. Silanization of aluminium oxide nanoparticles (Fig 22)

d. Preparation of gypsum moulds for fabrication of samples (Fig 23)

e. Preparation of heat polymerized polymethyl methacrylate denture base samples (Group C)

f. Preparation of heat polymerized polymethyl methacrylate denture base samples reinforced with 1% silanized aluminium oxide nanoparticles. (Group A)

g. Preparation of heat polymerized polymethyl methacrylate denture base samples reinforced with 3% silanized tetragonal zirconium oxide nanoparticles. (Group Z)

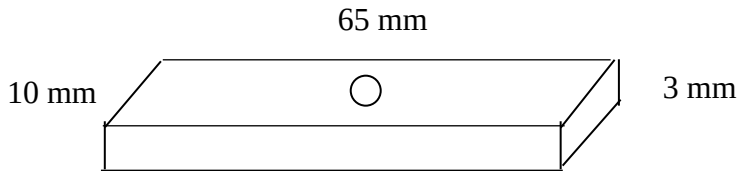
h. Testing of samples for flexural strength. (Fig 24)

A total of 90 samples were prepared with each group having 30 samples.

Group C	The control group; heat polymerized polymethyl methacrylate denture base material without reinforcement. (n=30)
Group A	Heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles. (n=30)
Group Z	Heat polymerized polymethyl methacrylate denture base material reinforced with 3% silanized tetragonal zirconium oxide nanoparticles. (n=30)

a. Die preparation

Metal dies were fabricated to prepare moulds for the fabrication of heat polymerized polymethyl methacrylate denture base material samples. Three brass metal dies of dimension 65 mm in length, 10 mm in width, and 3 mm in height (65×10×3) were fabricated. (ISO 1567 standard).[9,69]



These fabricated metal dies had a threaded hole at the centre. These holes were of 5 mm in diameter and 3 mm in depth. Screws were used to engage these threaded holes to facilitate easy removal of dies from the stone mould.

b. Silanization of aluminium oxide nanoparticles

4 gms of aluminium oxide nanoparticles were added to 100 ml of toluene in a glass beaker and sonicated for 20 mins at room temperature. After sonication, magnetic stirrer was placed in the beaker and kept on the stirring machine at room temperature for 30 mins. After this 0.2 ml of silane coupling agent (3-methacryloxy propyl trimethoxy silane) was added slowly, drop by drop under rapid stirring for homogenous mixing of silane coupling agent with aluminium oxide nanoparticles and toluene solvent.[66,67]

Parafilm was used to cover the beaker and left standing for 2 days. Toluene solvent was then removed by rotatory evaporator under the vacuum at 60℃ and rotation of 150 rpm for 30 mins. The silanized aluminium oxide nanoparticles were dried in a vacuum oven at 60ºC for 10 hrs.[18]

c. Silanization of tetragonal zirconium oxide nanoparticles:

30 gms of tetragonal zirconium oxide nanoparticles was added to 200 ml of pure toluene in a glass beaker and sonicated for 20 mins at ambient temperature. After sonication, magnetic stirrer was placed in the beaker and kept on the stirring machine at room temperature for 30 mins. After this 1.5 gms of silane coupling agent (3-methacryloxy propyl trimethoxy silane) was added slowly, drop by drop under rapid stirring for homogenous mixing of silane coupling agent with aluminium oxide nanoparticles and toluene solvent.[18]

Parafilm was used to cover the beaker and left standing for 2 days. Toluene solvent was then removed by rotatory evaporator under the vacuum at 60℃ and rotation of 150 rpm for 30 mins. The silanized tetragonal zirconium oxide nanoparticles were dried in a vacuum oven at 60ºC for 10 hrs.[16]

d. Preparation of gypsum mould for fabrication of samples

Preformed brass metal dies were used to prepare gypsum moulds. Before investing them, the threaded holes on the dies were blocked with carding wax. A thin layer of petroleum jelly was applied on three metal dies which were then invested in the lower half of the varsity flask. Die stone was used for base flasking and care was taken to embed half the thickness of the metal die in it.[84] Once the investment material had set, a thin layer of petroleum jelly was applied to the metal dies and to the investment material and then the counter flasking was done. The flasks were closed to ensure metal to metal contact between the base of the flask and its counterpart. After the investment material had set (1 hour)[85] the flasks were opened and the carding wax

within the holes was removed. The dies were engaged with a screw and gently teased out. The moulds were then immersed in hot water to remove any traces of petroleum jelly, wax and also to facilitate application of separating medium. These mould thus obtained were used for the fabrication of heat polymerized acrylic resin denture base material samples (PMMA).

e. Preparation of heat polymerized polymethyl methacrylate denture base samples without reinforcement (Group C)

30 samples were prepared using conventional heat polymerized denture base material (PMMA). Monomer and polymer were mixed in ratio of 1:2.5 by weight as per manufacturer's recommendations.[86] The materials were weighed using electronic balance of high accuracy. 7.5 gms of polymer powder and 3 ml of monomer was used for preparing 3 samples. Packing was done at dough stage, following which trial closure was performed. Final closure was done under a hydraulic bench press at a pressure of 3000 psi for 3 mins (according to the manufacturer). The flask was clamped and maintained under pressure for 1 hour.[85] It was then immersed in water in an acrylizer at room temperature. The temperature was raised slowly up to 74^0C and was held for 2 hours. The temperature was then raised to 100^0C and was maintained for 1 hour. After completion of this short curing cycle, the flask was removed from the water bath and allowed to bench cool at room temperature prior to deflasking.[87]

The polymerized samples were carefully removed and samples with defects were discarded. Finishing of the samples was done using sand paper (No. 120). The finished samples were stored in distilled water for 1 week at room temperature.[8,69]

f. Preparation of heat polymerized polymethyl methacrylate denture base material reinforced with silanized aluminium oxide nanoparticles (Group A):

30 samples were prepared using conventional heat polymerized polymethyl methacrylate denture base material reinforced with silanized aluminium oxide nanoparticles. 7.425 gms of polymer, 3 ml of monomer and 0.075 gms of silanized aluminium oxide nanoparticles were used for fabrication of 3 samples.[18.66]

High accuracy balance was used to weigh the material. By using ultra sonicator, silanized aluminium oxide nanoparticles was well dispersed in the monomer, which was done at 120 W, 60 KHz for 3 mins.

Polymer powder was then added gradually to reduce the possibility of particle aggregation. Mixing, flasking, packing and curing was done in the same manner as done for control group (Group C). The finished samples were then stored in distilled water for 1 week at room temperature. [18.66]

g. Preparation of heat polymerized polymethyl methacrylate denture base material reinforced with silanized tetragonal zirconium oxide nanoparticles (Group Z):

30 samples were prepared using conventional heat polymerized polymethyl methacrylate denture base material reinforced with silanized tetragonal zirconium oxide nanoparticles. 7.275 gms of polymer, 3 ml of monomer and 0.225 gms of silanized tetragonal zirconium oxide nanoparticles were used for fabrication of 3 samples. [18.66]

High accuracy balance was used to weigh the material. By using ultra sonicator, silanized tetragonal zirconium oxide nanoparticles were well dispersed in the monomer, which was done at 120 W, 60 KHz for 3 mins.

Polymer powder was then added gradually to reduce the possibility of particle aggregation. Mixing, flasking, packing and curing was done in the same manner as done for control group (Group C). The finished samples were then stored in distilled water for 1 week at room temperature. [18.66]

h. Testing of samples (PLATE IV)

Testing of samples was carried out at metallurgical laboratory. The samples for each group were tested for flexural strength. The flexural three-point bending test is useful in comparing the flexural strength of denture base materials as it simulates the type of stress that is applied to the denture during mastication.

Flexural strength was tested with universal testing machine system, at a 5.0 mm/minute crosshead speed.[8] The samples were supported on the jig separated at a distance of 50 mm. Load was applied at the centre of the sample. Stress- strain curves were recorded on a chart throughout the flexural tests. The maximum load during fracture was determined from the chart and recorded as fracture load in N (Newton) and the flexural strength was calculated in MPa.

Flexural strength (FS) was calculated using the formula.

$$FS = {}^{3PI}/_{2bd^2}$$

Where, FS = flexural strength (N/mm^2),

P = load at fracture (N),

I = distance between the supporting wedges (mm),

b = width of the sample (mm) &

d = thickness of the sample (mm).[6]

PLATE I

Fig 1: Heat polymerized acrylic resin

Fig 2: Die Stone

Fig 3: Aluminium Oxide Nanoparticles

Fig 4: Tetragonal Zirconium Oxide Nanoparticles

Fig 5: Silane Coupling Agent

Fig 6: Toluene

PLATE II
ARMAMENTARIUM AND EQUIPMENTS

Fig 7: High accuracy balance

Fig 8: Ultrasonicator

Fig 9: Magnetic stirrer

Fig 10: Vacuum rotatory evaporator

Fig 11: Acrylizer with thermostat

Fig 12: Universal testing machine

PLATE III
ARMAMENTARIUM AND EQUIPMENTS

Fig 13: Rubber bowl, plaster spatula, lacron's carver and varsity flask & clamp

Fig 14: Porcelain jar, petroleum jelly, dapen dish, sandpaper (No. 120), camel hair brush, sterile syringe, mixing spatula

Fig 15: Vernier caliper

Fig 16: Separating medium

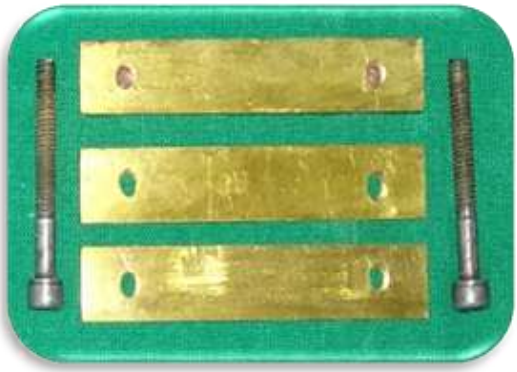


Fig 17: Brass metal dies

Fig 18: Parafilm

Fig 19: Hydraulic bench press

Fig 20: Distilled water plant

PLATE IV
METHODOLOGY

Fig 21: Silanization process of tetragonal zirconium oxide nanoparticles

Fig 22: Silanization process of aluminium oxide nanoparticles

Fig 23: Preparation of gypsum mould to obtain samples

Fig 24: Testing of samples

PLATE V

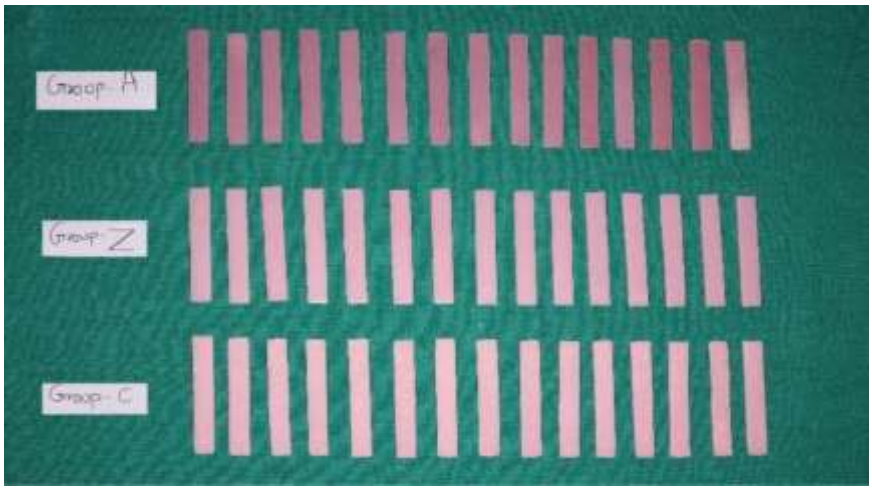


Fig 25: Samples of Group A, Group Z, Group C before testing

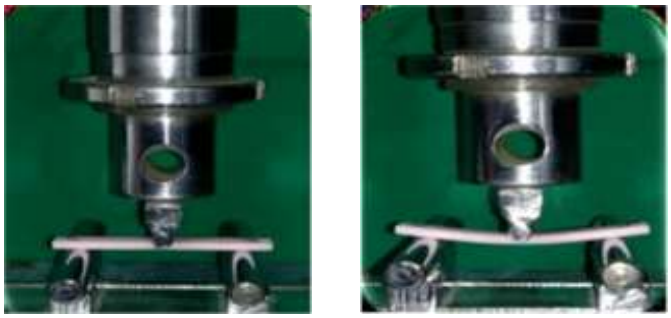


Fig 26: Testing of samples

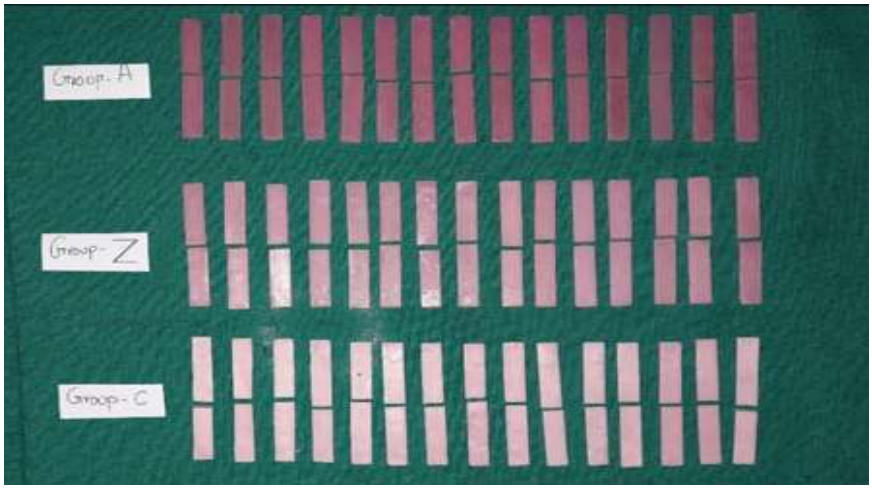


Fig 27: Samples of Group A, Group Z, Group C after testing

Results

In this study the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with silanized aluminium oxide nanoparticles and silanized tetragonal zirconium oxide nanoparticles was evaluated and compared. A total of 90 samples were prepared and were divided into three groups. Each group comprised of 30 samples.

Distribution of samples into groups

Sr no.	Group	Code	n = no. of samples
1	The control group; heat polymerized polymethyl methacrylate denture base samples without reinforcement	Group C	30
2	Heat polymerized polymethyl methacrylate denture base samples reinforced with 1% silanized aluminium oxide nanoparticles.	Group A	30
3	Heat polymerized polymethyl methacrylate denture base samples reinforced with 3% silanized tetragonal zirconium oxide nanoparticles.	Group Z	30
	Total number of samples		90

30 samples of each group were tested for flexural strength. Flexural strength was tested with a universal testing machine at a 5.0 mm/minute crosshead speed. The maximum load was determined from the chart and recorded as a fracture load in N (Newton) and the flexural strength was calculated in MPa and the results were then statistically analyzed.

STATISTICAL ANALYSIS

The data was analysed using SPSS statistical software. The *p* value was taken as significant when less than $p < 0.05$.

The statistical test used for the analysis of the results were:

1. One-way ANNOVA by K. Wallis test
2. Tukey's post-hoc test

The mean, median and standard deviation were calculated by ANOVA (one-way analysis of variance) and pair wise comparison of means was carried out and tested for statistical significance used Tukey's post-hoc test.

- Mean is sum of all observations divided by the no. of observations.
- Median is value of the variable that divides the distribution into two equal parts i.e. 50% observations will lie below and above it.
- Standard Deviation is summarized as the amount of variation (change) in the observation from their average value (mean).
- The formula used for calculating standard deviation:

$$SD = \sqrt{\frac{\Sigma(\bar{x} - x)^2}{n - 1}}$$

Where:

$\overline{X}$	=	Mean
X	=	Values of the variables
Σ	=	Sum of the value
n	=	Number of observations
Min	=	Minimum Value
Max	=	Maximum Value

Table 1 provides the descriptive statistics for flexural strength of test samples in three study groups. The mean of Group C was minimum i.e. 89.34 MPa and the strength ranged between 67.95 to 108.62 MPa. For Group A, the mean flexural strength was 102.52 MPa and ranged between 73.91 to 120.45 MPa. In Group Z, the

mean flexural strength was maximum i.e 109.65 MPa and ranged between 67.95 to 123.15 MPa. A graphical visualization of mean strength along with error bar is given in the **Graph I.**

Table 2(a) and 2(b) reveals that the mean flexural strength across groups differed highly significantly across three groups, as indicated by p-value < 0.001 (F=28.4) using Anova F test. In order to determine which groups contributed to overall significance, a pair-wise comparison of mean flexural strength was performed using Tukey's post hoc test.

Table 3 shows highly significant difference on comparison of mean flexural strength of Group C and Group A ($p < 0.001$) where Group A has higher flexural strength than Group C. Also, there exist high statistical difference ($p < 0.001$) between Group C and Group Z where Group Z also having higher mean flexural strength as compared to Group C. Group Z have statistically significant ($p<0.05$) higher mean flexural strength as compared to Group A.

Discussion

In edentulous patients the use of dental prosthesis is fundamental in order to restore function and aesthetics. There has been unceasing growth in the field of material science for fabricating dentures from using naturally occurring materials to use of synthetic resins.[1]

In the past, vulcanite, celluloid & phenol formaldehyde were the materials used for denture bases.[88] Vulcanite dentures were very popular until the 1940s, until acrylic denture bases replaced them. **Rohm and Hass** (1936) introduced PMMA in sheet form, **Nemours** (1937) in powder form and **Dr. Walter Wright (1937)**, introduced polymethyl methacrylate as a denture base material. Since its introduction, till today it is the most primarily used denture base material because it is economical, has excellent aesthetics, it has an ease of processing and repair.[1,5]

It is a combination of the benefits mentioned above rather than one excellent aspect that accounts for its popularity and wide usage. However, this material is not ideal when looking at the mechanical requirements of a prosthesis. It has certain drawbacks like poor mechanical strength, residual monomer allergy, low fatigue strength, poor conduction of heat, low hardness, thermal shrinkage, porosity, high coefficient of thermal expansion, crazing and warpage.[10] The denture base resin, during function is subjected to various stresses, which include compressive, tensile, shear, and impact stresses.[12] A study by **Johnston et al**[11] concluded that 68% of acrylic resin dentures fractured within the first three years after fabrication.

In 1942, vinyl acrylic copolymer and in 1948 polystyrene, a styrene polymer developed by **Charles Dimmer**, were introduced as denture base materials which had greater flexural strength and high residual stresses.[1] **Faber (1957)** described the technique for fabrication of lower cast metal bases in dentures. According to **Faber, Peyton (1943) and Skinner (1951**), the use of metal base is acceptable as it causes fewer tissue changes.[1] The denture base material should have sufficient strength and resilience to withstand normal masticatory forces. These materials should not creep under masticatory loads for long-term use if good occlusion is to be maintained and potential effects of irritants are to be kept minimum.[89]

Several studies have been conducted with the goal of enhancing the properties of PMMA by using different curing methods and/or incorporating fillers in its composition. Addition of fillers and fibers to PMMA is a commonly used method to improve both the physical and mechanical properties of the material.[80]

Although PMMA has been commonly utilized in the fabrication of removable denture bases, a number of polymeric materials, such as high-density polyethylene (HDPE), polyamide (PA), and poly(*L*-lactide) (PLLA), have been studied for their prosthodontic applications. Furthermore, polystyrene polyvinyl acrylic and light-activated UDMA have also been used in the construction of denture bases. These materials should be durable and strong enough to withstand masticatory forces, especially in patients with parafunctional habits. However, none of these polymers provides the unique combination of physical and aesthetic properties exhibited by PMMA. Despite these excellent properties, improvement in the fracture resistance of PMMA is needed.

Smith (1961)[14] analyzed the practical situation with respect to fracture of dentures and showed that there are two types of failures.

1. Outside the mouth which is caused by impact forces, i.e. a high stress rate and
2. Inside the mouth, usually in function; probably due to a fatigue phenomenon, i.e. a low and repetitive stress.

Fracture of acrylic resin denture base happens because of the fatigue and chemical degradation of denture base material.[10] Any factor that exacerbates deformation of the base or alters its stress distribution will influence the denture to fracture.[12] During chewing, denture base material undergoes flexural deformation. Flexural strength of a denture base resin is considered as the primary property responsible for its clinical failure.[11] For this reason, flexural strength was selected as the most relevant mechanical property to evaluate the strength of denture base resins.

The ultimate flexural strength of a denture base material is directly proportional to its potential to resist catastrophic failure under a flexural load. As a foundation, the acrylic resin materials should exhibit a high proportional limit to resist plastic deformation and also exhibit fatigue resistance to bear repeated masticatory loads.[11,90,91] An acrylic resin capable of sustaining higher flexure in combination with high resistance to cyclic loading may be less prone to clinical failure.[8]

Beyli et al (1981) stated that the ratio of upper denture fracture to the lower denture fracture is 2:1 with the common cause of poor fit and lack of balance in the denture. He suggested that midline fracture of a denture base is caused due to flexural fatigue failure which is a result of cyclic deformation of the base during function. He stated that any factor that exacerbates deformation of the base or alters its stress distribution will predispose the denture to fracture.[12]

Huggett and Harrison conducted a questionnaire survey and reported that 29% of midline fracture occurred due to flexural fatigue failure and mostly in upper dentures. These considerations led to a general conclusion that denture fracture occurs due to flexural fatigue under respective conditions and due to this reason flexural strength test were selected as most relevant to evaluate the strength of denture base resin. Studies have shown that the average values of flexural strength of heat polymerizing acrylic resins are near to 78-92 MPa.[12] The mean flexural strength of conventional heat polymerized polymethyl methacrylate without any reinforcement recorded in the present study was **89.34 MPa** which was comparable with the previous studies.

There are three ways to improve the properties of PMMA: development of an alternative material to PMMA; the chemical modification of PMMA such as by the addition of a rubber graft copolymer and the reinforcement of PMMA with other materials such as carbon fibers, glass fibers and ultra-high modulus polyethylene.[10] Various alternates for polymethyl methacrylate have been familiarized such as polystyrene, polyvinyl acrylic, polyamides (nylons) and light activated urethane dimethacrylate resins. These materials exhibited desirable properties but none have been proven superior to polymethyl methacrylate (PMMA).[8]

As the most important property of a denture base resin is its fracture resistance, efforts have been made to enhance the mechanical properties of acrylic resin by giving maximum bulk to the material in the region most heavily stressed by copolymerization and cross linking, reinforcement with glass fibers, aluminium and sapphire whiskers, polycarbonates, carbon fibers, and the addition of metal strengtheners. These reinforcing methods were done to improve the mechanical properties of the denture base resins and to prevent the problem of denture fractures.[31,92]

The chemical alteration of polymethyl methacrylate through the addition of rubber in the form of butadiene styrene has been studied. Modifications of the chemical structure, by adding cross-linking agents or copolymerization with rubber resulted in significant increase in impact strength. However, stiffness, fatigue resistance, and transverse strength were reduced.[69]

The fiber-reinforced plastics are commonly used in many fields of industry because of their good mechanical properties, which can be tailored to specific needs.

Reinforcement of dental resin with short or long fibers have been described in the literature for nearly half a century. Several different types of fibers have been used, with varying results but fiber reinforcement has never been adapted to routine clinical practice. Effective fiber reinforcement is dependent on many variables, including the type of the fibers, the percentage of fibers in the matrix, the modulus and distribution of the fibers, fiber length, orientation, forms, and interfacial bond.[8] Although the incorporation of fibers produced reassuring results, this method has various problems including tissue irritation, increased fabrication time, difficulties in handling, the need for precise orientation, and placement or bonding of the fibers within the resin.[69]

Various other fibers that have been tested and used comprise of kevlar fibers[15,33], glass fibers[52,44,75] and ultra-high molecular weight polyethylene fibers.[32] Commercially, reinforcement of polymers with long, continuous fibers has been established as an effective means of developing engineering materials. However, dental applications of fiber-reinforced resin require a unique balance of properties like biocompatibility, esthetics and stability in the oral environment.[8]

Larson et al (1971)[28] have concluded that the use of carbon fibers improves the strength of denture bases. It has improved the mechanical properties of the matrix because of its inherent high strength and optimal combination of the carbon fibers and matrix. Mainly, carbon fibers have been used to enhance fatigue and impact strength. Despite good mechanical properties, cytotoxicity of carbon fibers is problematic.

In 1959, **Feynman** introduced the concept of nanotechnology. Since then, nanotechnology has been widely used in many applications, including medical sciences where it plays an important role in diagnosis, treatment planning and

regenerative medicine. A nanomaterial is an object, in which at least one of its dimensions is at the nanometer scale (approximately 1 to 100 nm). Nanomaterials are categorized according to dimension – those with all 3 dimensions less than 100 nm [nanoparticles (Nps) and quantum dots]; those that have 2 dimensions less than 100 nm (nanotubes, nanofibers, and nanowires); and those that have one dimension less than 100 nm (thin films, layers, and coatings).[93] Recently, researchers have used nanofillers for reinforcement of denture base resins. Size, shape, surface area, concentration and dispersion of nanofillers into resin matrix have an effect on the mechanical properties of the filler/resin composite. Nanoparticles of Alumina, titanium (TiO_2), silver, zirconia (ZrO_2), gold, platinum, silicon dioxide (SiO_2) are among the fillers that have been incorporated to enhance the mechanical properties of denture base resins.[80]

Mullarky (1985) and Berrong, Weed and Young (1990)[33] studied the reinforcement of acrylic resin with aramid fibers. They were successful in enhancing the fatigue resistance of the aramid fiber reinforced acrylic resin denture base material. But the incorporation of aramid fibers produced similar problems with respect to color. The yellow appearance of the fibers was difficult to mask within the denture, necessitating thick layers of acrylic resin that added significantly to the bulk of the denture.

Korkmaz et al (2005) [94] suggested that the small size of filler particles ensures proper processing. The average particle size of PMMA beads used in denture base resins is around 100 μm. The particle size of Ag nanoparticles (80-100 nm) used is much smaller than that of powder resin particles. Silver nanoparticles fill the

interstitial spaces of polymer particles to give a heterogenous mixture and does not force the displacement of the segments of polymer chain. Also, low percentage of nanoparticles should be used to ensure that they will be embedded in resin. Due to high surface area of these nanoparticles, the applied stress is transformed from the matrix onto the silver nanoparticles, resulting in an enhancement of the mechanical properties.

Recent development of composite materials of great strength and low mass has made significant contribution in the field of dental material science. The incorporation of the ceramic nano-filler into the more flexible and lower thermal resistance polymer improves its stiffness and thermal stability.[95] Ceramic fillers were used for reinforcements as opposed to metal fillers because of its lower filler density.[8]

Zirconium oxide, commonly referred to as zirconia (ZrO_2), possesses strong ionic inter atomic bonding, giving rise to its desirable material characteristics. Zirconium oxide (ZrO_2) exists in three crystalline phases i.e. monoclinic, cubic & tetragonal and the most stable phase is monoclinic phase.

The effect of crystallite size on the phase formation both in pure zirconia and its solid solutions was studied carefully and it was demonstrated that the diminishing of the crystallite size in the material leads to the stabilization of high symmetric zirconia based solid solutions.[96]

ZrO_2 possesses strong ionic interatomic bonding, giving rise to its desirable material characteristics, that is, hardness and strength. Incorporation of Zirconia nanofillers to acrylic resin was found to improve its mechanical properties. In addition

to that, ZrO_2 was used because it has an excellent biocompatibility and it is white in color which is less likely to alter any esthetics. The Nano-filler particles were used in this study as it yields a better dispersion, eliminate aggregation and improve its compatibility with organic polymer.[97,98] Proper percentage range of zirconium oxide Nano-fillers (Percentages of 3% by weight) was selected because percentages above 7% leads to massive changes in the color of acrylic.[18] It is noted that the concentration of ZrO_2 (3%wt) lead to the maximum value of fracture toughness. There is no significant improvement in fracture toughness values of the modified acrylic resin at the concentrations of ZrO_2 above that limit (5%wt and 7%wt). It is probably due to complete saturation of the polymer matrix with the ZrO_2 particles.[99]

The size of its particle is another important factor as larger particle size decreases the tensile strength because they settle down when mixed with monomer.[14] The small metal oxide particles fill the interstitial polymer particles to give a heterogeneous mixture. Thus, it was more advantageous to use a smaller particle sized filler.

Many attempts, however, to strengthen acrylic resin in this way failed because stress concentrations occurred around embedded materials, and the net effect of embedding fibers or metals is actually to weaken the polymer. This is often due to poor adhesion between the fiber, metal inserts and acrylic resin matrix.[9]

Although acrylic resin is popular for denture construction, there is a common problem of lack of perception of temperature of food and beverages.[52] Interestingly, many of the ceramic materials have thermal conductivities approaching or even exceeding that of some of the metals.[9]

H. H. Chandler et al (1971)[55,56] suggested that one of the deficiencies of the available denture base resins is their radiolucency. Thus, there are chances of ingestion or aspiration of either broken or portions of ill-fitting complete dentures by the patient. **Sehjpal and Sood (1989)**[17] stated that reinforced PMMA with metal oxide fillers like silver, copper, aluminium and zirconium not only increase the strength but also provides radio-opacity to the heat polymerized denture base material.

Ihab (2011)[50] gave a method for homogenous dispersion of filler particles in polymer matrix. It plays a major role in the mechanical properties of particulate-filled polymer composites. The reason for increase in flexural strength with addition of aluminium oxide powder was credited to the proper distribution of alumina spheres within denture base powder which acts as potential fillers in the resin matrix.

Arora et al (2011) [80] recently reviewed the effect of addition of alumina and reported a positive impact on the properties of acrylic resin. Addition of alumina powder to acrylic resin improved its thermal conductivity and, accordingly, patient satisfaction was expected to increase. Also, reinforcing PMMA with aluminium increased the flexural strength, impact strength, tensile strength, compressive strength, and surface hardness of the resin. Warpage decreased significantly after addition of aluminium to PMMA. On the other hand, some studies found that adding aluminium decreases both the impact and tensile strength of PMMA. Treating aluminium oxide particles with a coupling agent increased the flexural properties of acrylic resin. Also, silane-treated aluminium particles significantly increased the compressive, tensile, and flexural strength and the wear resistance of reinforced denture base resin.[80]

Safi et al (2017)[80] concluded that adding Al_2O_3 nanoparticles to PMMA increases its thermal stability compared with that of pure PMMA. Addition of silanized Al_2O_3 nanoparticles to acrylic resin improved the thermal properties and flexural strength of acrylic resin, and at the same time this addition decreased the water sorption and solubility. Although **Kul E, Aladağ Lİ, Yeşildal R**. indicated the addition of Al_2O_3 to PMMA significantly increased thermal conductivity, the flexural strength values of PMMA were not significantly different.[80]

Pisaisit et al (2012)[21,70] added silanized and nonsilanized alumina to PMMA, and by SEM image they found that there was a gap between nonsilanized alumina particles and resin matrix which could explain reduction in mechanical properties.

In this study, the addition of silanized Al_2O_3 nanoparticles showed the mean value of flexural strength was **102.52 MPa** which was comparable with the study conducted by **Jasmin BS, Ismail IJ** [21] who showed that there is a significant increase in the value of flexural strength with incorporation of 1 wt% Al_2O_3 nanoparticles compared to control groups.

Abdulkareem M, Hatim N (2016)[100] concluded that microwave radiation of PMMA powder and the addition of Al_2O_3 and Ag nanoparticles are effective in increasing the flexural strength of denture base resins. Increase in the flexural strength may be explained based on the property of transformation toughening. When sufficient stress develops and microcracks begin to spread, a transformation phenomenon of nanoparticles occurs, which depletes the energy of crack propagation. Hence, proper distribution of the nanofiller within the matrix can stop or deflect the cracks. Another possible reason for increase in flexural strength of the denture base

with the addition of nanoparticles was due to transfer of stress from more flexible polymer to the higher modulus, more rigid and stiffer nanoparticles.

A study by **Ihab et al**[66] concluded that an increase in the transverse strength occurred with addition of 2-5 wt% ZrO_2 nanoparticles due to good distribution of the very fine size of nanoparticles. But increasing the percentage of modified nano-ZrO_2 to 7 wt% lowered the impact strength and transverse strength due to agglomeration of nano-ZrO_2. Hence an optimum of 3% was chosen for this study. A high surface energy is usually displayed by the inorganic filler particles because of their hydrophilic ionic nature. But due to the difference in surface energy, the hydrophobic polymer does not wet or interact with the filler particles.[71] Therefore, it is important to modify the filler surface for better dispersion and improve surface wetting, thereby improving the physical properties of the composites.[101] Hence in this study, tetragonal zirconium oxide nanoparticles were treated with trimethoxysilylpropylmethacrylate (TMSPM) to improve adhesion of nanoparticles to the resin matrix.[79]

The test samples utilized in this study were prepared according to ISO 1567 standardization with the dimensions of 65 × 10 × 3 mm from heat polymerizing acrylic resin.[9,69] A digital caliper was utilised to measure the thickness and width of each sample.[102] Samples were tested for flexural strength using a three point-bending test with a universal testing machine. Distance between the supports was 50 mm. A load was applied to the specimens by a centrally located rod until fracture struck. The chart recorder on the testing machine produced a complete force versus deflection history of each test. The force and deflection at fracture were obtained from the data of machine.[92]

The average values of flexural strength of heat polymerizing acrylic resins are near to 78-92 MPa.[12] Therefore, the ultimate goal of this study was to evaluate and compare the flexural strength of heat polymerized acrylic resin denture base material reinforced with 3% silanized tetragonal zirconium oxide nanoparticles and 1% silanized aluminium oxide nanoparticles. The results showed that the mean flexural strength of **Group A (102.52 MPa)** and **Group Z (109.65 MPa)** was greater than **Group C (89.34 MPa).** This study shows that flexural strength can be increased by reinforcement with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles. The reason for the increase in flexural strength with addition of 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles was attributed to the proper distribution of particles within the denture base powder and these act as potential fillers in the resin matrix. There was a significant statistical difference between the reinforced PMMA with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles seen.

Clinical significance

When the entire spectrum of this study is analyzed, it becomes evident that the heat polymerized acrylic dentures reinforced with silanized aluminium oxide nanoparticles and silanized tetragonal zirconium oxide nanoparticles increases the flexural strength of the denture base material and thus, reduces the probability of occurrence of fracture. It also increases the thermal diffusivity of the denture base material, which enhances the patient's perception to hot and cold, hence improving the adaptability of the patient to the denture. This in turn, aids in better comfort and

contentment with the prosthesis in place. In addition to this, it imparts radio-opacity to the material so that any fractured remnants can be detected radiographically.

Limitations of the study

The study was designed and carried out with utmost accuracy, however certain limitations encountered in the study can be enlisted as follows:

In the oral cavity, reinforced denture base is exposed to various forces of varying magnitudes acting in different directions. The same situation could not be simulated in this in vitro study.

Examination by a Scanning electron microscopy (SEM) of the samples to evaluate the adhesion of silanized tetragonal zirconium oxide nanoparticles to the surface of PMMA and the adhesion of Silanized Aluminium oxide nanoparticles to the surface of PMMA was not performed.

Scope for further studies

Fatigue testing of these materials under dynamic loading using the denture base configurations in simulated oral conditions, using saliva or its substitutes is an area for further research.

Further research is needed to evaluate the effect of aging on the new reinforced denture base material before clinical application.

Other physical and mechanical properties like thermal diffusivity, hardness, abrasion resistance, color stability and disinfectant property can be studied.

The heat polymerized acrylic dentures can be reinforced with even further different size nanoparticles and various physical and mechanical properties can be evaluated.

Summary

The heat cure denture base resins are widely used for their excellent properties such as ease of handling, aesthetics and also polishing.[8,9] Conversely, to maintain the durability of the denture, mechanical strength is not sufficient.[10] The fracture of acrylic resin denture is a usual incident.[11]

This study was conducted to evaluate and compare the flexural strength of heat polymerized polymethyl methacrylate denture base material reinforced with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles with that of conventional heat polymerized polymethyl methacrylate denture base material. The samples tested were fabricated according to ADA specification no. 12 with 30 samples in each group.

Flexural strength was tested using Star system universal testing machine at a crosshead speed of 5 mm/min. The findings were statistically analyzed and the flexural strength was calculated in MPa.

Results of the study shows that the mean flexural strength of Group A was maximum which was **102.52 MPa.** For **Group Z**, the mean strength was **109.65 Mpa, and** for **Group C** the mean strength was **89.34 Mpa.** Statistical analysis shows highly significant difference on comparison of mean flexural strength of Group C and Group A ($p < 0.001$) where Group A has higher flexural strength than Group C. Also, there exist high statistical difference ($p < 0.001$) between Group C and Group Z where Group Z also having higher mean flexural strength as compared to Group C. Group Z have statistically significant ($p<0.05$) higher mean flexural strength as compared to Group A.

Thus, the study results concluded that reinforcement of acrylic resin with 1% silanized aluminium oxide nanoparticles shows a highly significant increase in the flexural strength when compared with the unmodified acrylic resin, 3% silanized tetragonal zirconium oxide nanoparticles shows a highly significant increase in the flexural strength when compared with the unmodified acrylic resin and there was a significant statistical difference between PMMA reinforced with 1% silanized aluminium oxide nanoparticles and 3% silanized tetragonal zirconium oxide nanoparticles.

Conclusion

Within the limitations of this study following conclusions were drawn:

1. Samples with reinforcement increase the flexural strength.
2. Reinforcement with 1% silanized aluminium oxide nanoparticles showed highly significant increase in flexural strength.
3. Reinforcement with 3% silanized tetragonal zirconium oxide nanoparticles showed highly significant increase in flexural strength.
4. There was a statistical significance between the heat polymerized acrylic resin denture base samples reinforced with 1% silanized aluminium oxide nanoparticles (Group A) and 3% silanized tetragonal zirconium oxide nanoparticles. (Group Z).

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Table 1

Descriptive statistics of flexural strength of heat polymerized polymethyl methacrylate denture base material without reinforcement (Group C), reinforced with silanized aluminium oxide nanoparticles (Group A) and silanized tetragonal zirconium oxide nanoparticles (Group Z) respectively

	Mean	**Median**	**S.D**	**Range**
Group C (Control) (n=30)	89.34	88.80	10.93	67.95 – 108.62
Group A (Aluminium Oxide) (n=30)	102.52	105.06	12.0	73.91 – 120.45
Group Z (Zirconium Oxide) (n=30)	109.65	110.52	8.52	67.95 – 123.15

Table 2(a)

Comparative statistics of flexural strength of heat polymerized polymethyl methacrylate denture base material without reinforcement (Group C), reinforced with silanized aluminium oxide nanoparticles (Group A) and silanized tetragonal zirconium oxide nanoparticles (Group Z) respectively

	Mean	S.D	ANOVA F Test	p value, Significance
Group C (Control) (n=30)	89.34	10.93	F = 28.40	p < 0.001 **
Group A (Aluminium Oxide) (n=30)	102.52	12.0		
Group Z (Tetragonal Zirconium Oxide) (n=30)	109.65	8.52		

p >0.05 – not significant * p<0.05 – significant difference **p<0.001 – highly significant

Table 2(b)

Detailed statistics of One –way analysis of variance for flexural strength across three groups

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	6371.5	2	3185.7	28.40	p <0.001**
Within Groups	9759.1	87	112.174		

p >0.05 – not significant * p<0.05 – significant difference **p<0.001 – highly significant

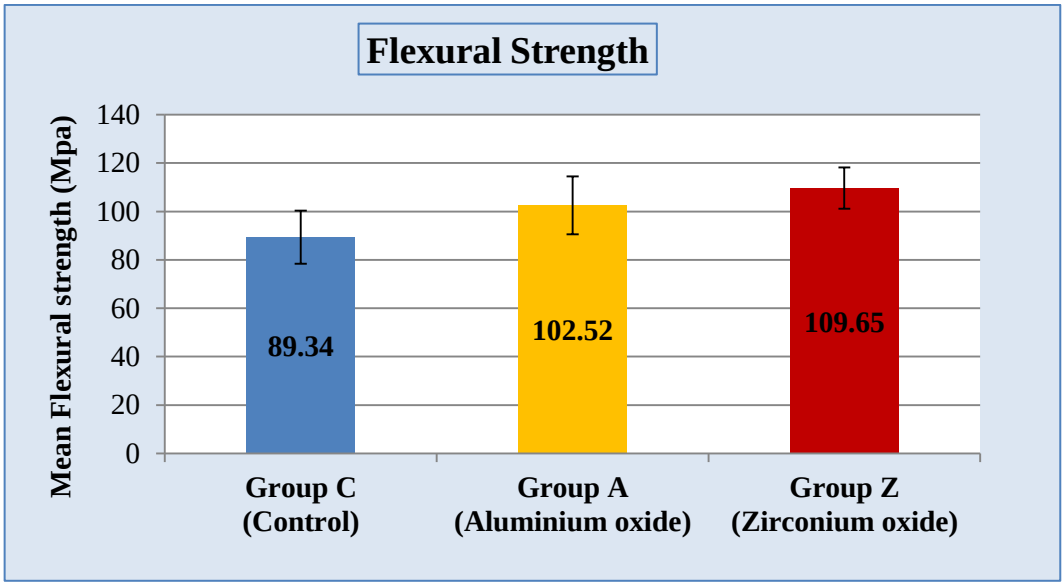
Table 3

Pair wise comparison of flexural strength between groups using Tukey's post – hoc test

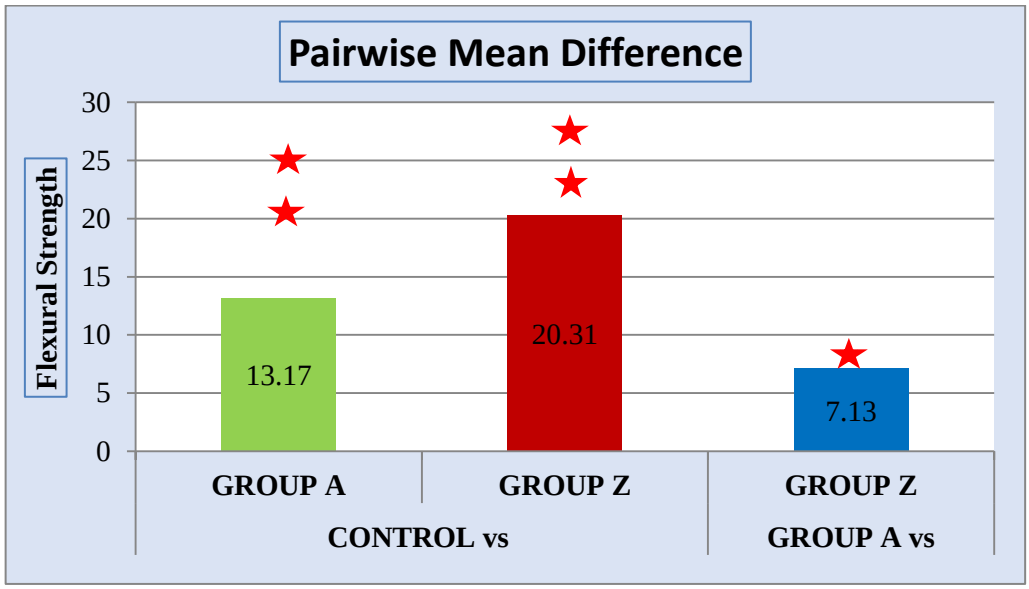
Comparison	Mean Difference	p value, Significance
Control vs Group A	13.17	p <0.001**
Control vs Group Z	20.31	p <0.001**
Group A vs Group Z	7.13	p = 0.029*

p >0.05 – not significant * p<0.05 – significant difference **p<0.001 – highly significant

Graphs



Graph 1: Bar chart with error bars showing means flexural strength according to the study groups



Graph 2: Comparison of mean flexural strength between groups

Annexure

Master Chart

Name of Test : Flexural Strength

Machine Specifications : Universal testing machine (computerized, software based)

Company : Acme Engineers, India.

Model : UNITEST 10, System Accuracy of the machine: ±1%,

Cross head speed : 5mm /minutes.

Sample Dimensions : Thickness = 3.0 mm, Width = 10.0 mm, Length = 65 mm

GROUP C (Control) : Heat polymerized Acrylic Resin(PMMA) without reinforcement								
Sr. No.	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**		**Sr. No.**	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**
1	No. 1	98.60	83.15		16	No. 16	113.38	93.82
2	No. 2	123.45	102.87		17	No. 17	105.45	86.83
3	No. 3	115.65	96.83		18	No. 18	92.23	77.82
4	No. 4	107.60	89.66		19	No. 19	81.53	67.95
5	No. 5	113.80	94.50		20	No. 20	96.58	79.82
6	No. 6	128.20	106.00		21	No. 21	104.75	83.62
7	No. 7	124.04	103.40		22	No. 22	95.95	79.95
8	No. 8	130.35	108.62		23	No. 23	113.17	93.29
9	No. 9	119.85	99.87		24	No. 24	101.85	83.04
10	No. 10	116.32	97.75		25	No. 25	88.45	73.33
11	No. 11	126.51	105.58		26	No. 26	111.40	91.41
12	No. 12	89.85	74.72		27	No. 27	102.54	85.58
13	No. 13	95.45	79.54		28	No. 28	95.95	78.92
14	No. 14	117.42	97.89		29	No. 29	93.26	77.82
15	No. 15	104.76	87.95		30	No. 30	117.10	98.73
Average					**Average**			**89.34**

GROUP A (Silanized): Heat polymerized Acrylic Resin (PMMA) reinforced with 1% silanized aluminium oxide nanoparticles								
Sr. No.	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**		**Sr. No.**	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**
1	No. 1	132.68	110.91		16	No. 16	126.35	106.29
2	No. 2	124.42	104.30		17	No. 17	134.67	112.02
3	No. 3	117.75	98.83		18	No. 18	113.15	94.15
4	No. 4	144.65	120.20		19	No. 19	133.25	111.45
5	No. 5	136.35	113.45		20	No. 20	129.25	109.21
6	No. 6	93.84	78.70		21	No. 21	129.65	109.87
7	No. 7	132.00	110.66		22	No. 22	139.09	115.72
8	No. 8	104.78	87.23		23	No. 23	89.32	73.91
9	No. 9	123.85	103.54		24	No. 24	104.25	87.15
10	No. 10	124.07	102.42		25	No. 25	121.32	101.62
11	No. 11	126.64	105.83		26	No. 26	131.95	110.29
12	No. 12	144.35	120.45		27	No. 27	120.60	100.66
13	No. 13	135.55	112.93		28	No. 28	111.57	92.96
14	No. 14	128.58	107.26		29	No. 29	96.42	80.38
15	No. 15	120.88	100.60		30	No. 30	111.72	92.58
Average					**Average**			**102.52**

GROUP Z (Silanized): Heat polymerized Acrylic Resin (PMMA) reinforced with 3% silanized tetragonal zirconium oxide nanoparticles							
Sr. No.	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**	**Sr. No.**	**Sample No.**	**Flexural Load (N)**	**Flexural Strength (MPa)**
1	No. 1	132.68	110.91	16	No. 16	126.35	106.29
2	No. 2	124.42	104.3	17	No. 17	134.67	112.02
3	No.3	117.75	98.83	18	No. 18	146.15	123.15
4	No. 4	144.65	120.2	19	No. 19	133.25	111.45
5	No. 5	139.35	116.45	20	No. 20	142.25	119.21
6	No. 6	106.84	89.03	21	No. 21	134.65	112.87
7	No. 7	132	110.66	22	No. 22	141.09	117.72
8	No. 8	112.78	93.23	23	No. 23	139.88	116.65
9	No. 9	123.85	103.54	24	No. 24	139.25	117.15
10	No. 10	124.07	102.42	25	No. 25	121.32	101.62
11	No. 11	126.64	105.83	26	No. 26	131.95	110.29
12	No. 12	146.35	122.45	27	No. 27	121.6	101.66
13	No. 13	135.55	112.93	28	No. 28	120.57	100.96
14	No. 14	128.58	107.26	29	No. 29	131.42	110.38
15	No. 15	144.88	121.6	30	No. 30	129.72	108.58
Average				**Average**			**109.65**