THE EFFECT OF GLASS FIBER POLYSULPHONE COMPOSITE REINFORCEMENT ON FLEXURAL STRENGTH OF TWO DENTURE-BASE POLYMERS

MARTHA CAROLINA HENNING

BChD PRETORIA

DIP ODONT PRETORIA

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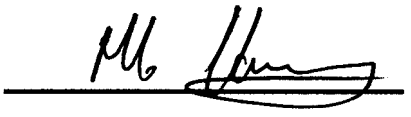
DECLARATION

I, Martha Carolina Henning, hereby declare that the work on which this dissertation is based, is original and neither the whole work nor any part of it has been, is being, or is to be submitter for another degree at this or any other University



M C HENNING

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SUMMARY

Statement of the Problem Despite its popularity as a denture-base material, poly-methylmethacrylate exhibits inherently low resistance to impact and fatigue failure. This is reflected in the analyses of the prevalence of denture fracture which suggests that prosthesis failure remains an unsolved problem. Consequently, various approaches to improve the physical and mechanical properties of denture-base polymers have been proposed. These include incorporation of solid metal forms and various types of fibers to provide reinforcement to fracture prone areas. In recent years there has been considerable interest in glass fiber reinforcement of polymeric denture resins. Partial fiber reinforcement using glass fiber has been shown to improve the mechanical properties of removable prostheses. However, adequate impregnation of fibers often requires the use of excessive methyl methacrylate monomer which has a deleterious effect on the mechanical properties of the fiber composite material.

Purpose The aim of the present study was to determine the effect of glass fiber polysulphone composite reinforcement on the flexural characteristics of two commonly used denture-base polymers.

Materials and methods Prefabricated E-glass fiber polysulphone composite rods, ± 3mm in diameter, with a continuous, unidirectional, non-silanized fiber concentration of ± 55 vol %, were employed as strengtheners. The reinforcement was incorporated axial to the neutral axis in standardized cylindrical heat polymerizing conventional and high impact resin test specimens 6mm in diameter and 28mm in length. The two reinforced pattern groups (n = 10) were compared with unreinforced resin control groups. A three point loading test was performed in air after storage of the specimens in water at 37°C for a period of 8 weeks. The following values were

measured : flexural modulus and flexural strength. The obtained data were subjected to relevant statistical analysis.

Results The flexural modulus of the glass fiber polysulphone reinforcement was 14,106 MPa and the flexural strength 546.6 MPa. The flexural modulus of conventional denture-base resin was 1746 MPa, reinforcement increased it to 2328 MPa, and the flexural strength increased from 164 MPa to 209 MPa. The flexural modulus of high impact polymer was 1684 MPa and reinforcement increased it to 2067 MPa. The flexural strength was increased from 171 MPa to 242 MPa with reinforcement. Statistical analysis using t test showed that reinforcement affected the flexural modulus and flexural strength of polymer brands ($p < : 05$).

Conclusion Novel glass fiber polysulphone composite reinforcement may considerably enhance the flexural properties of multiphase denture-base polymers.

SAMEVATTING

Probleemstelling Polymetiel metakrilaat geniet voorkeur as kunsgebitbasis materiaal. 'n Inherente lae weerstand teen impak en vermoeienissterkte dra egter by tot die soeke na 'n basismateriaal met ideale eienskappe. Hierdie word weerspieel in 'n analise aangaande die voorkoms van kunsgebit frakture, wat duidelike bewys lewer dat kunsgebit frakture 'n onopgeloste probleem bly. Verskeie weë is gevolg in 'n poging tot die verbetering van die fiesiese en meganiese eienskappe behorend tot kunsgebitbasis polymere. Insluitende was die inkorporasie van soliede metaalvorms en verskeie tipes vesels wat tot die versterking van bepaalde swak areas in die gebitbasis moes dien.

Navorsing, in die versterking van kunsgebitbasis polymere, met veselglas, is in die laaste 50 jaar intensief beoefen. Verwyderbare prosteses het gebaat by gedeeltelike veselversterking. Alhoewel genoegsame impregnasie van glasvesels 'n voorvereiste is vir adhesie tussen vesels en polymer materiaal, lei 'n oortollige hoeveelheid metiel metakrilaat diwels tot verswakking van die meganiese eienskappe van vesselkomposiet materiale.

Doel Die doel van hierdie studie is die bepaling van die effek wat glasvesel versterkte polysulfoon komposiet op die buigsterkte eienskappe van twee kunsgebit basis polymere het.

Materiale en metodes Voorafvervaardige E-glass polysulfoon komposiet stafies, ± 3mm in deursnee, met 'n aaneenlopende, nie gesilaniseerde veesel konsentrasie van ± 55 vol %, is as versterker gebruik. Die versterker is aksiaal tot die neutrale as geplaas in gestandariseerde, silindries hitte ekuurde donvensionele en hoë impak materiaal. Toetsstafies was 6mm in deursnee en 28mm lank. Twee versterkte groepe (n = 10) is vergelyk met twee onversterkte groepe n = 10. 'n Drie punt stress toets is gedoen inlug na berging in water vir 8 weke by 37°C. Die volgende aspekte is geevalueer :

modulus van elastisiteit en fraktuursterkte. Die resultate verkry is onderwerp aan stitistiese analise by wyse van die t toets.

Resultate Die E-modulus van glasveselversterkte polisulfoon was 14.106 MPa en die fraktuursterkte 546.6 MPa. Die E-modulus van konvensionele akrielbasis gebit was 1746 MPa. Hierdie resultate verbeter tot 2328 MPa. Fraktuursterkte verbeter van 164 MPa na 209 MPa. Die E-modulus van hoë impak polimer was 1684 MPa en dit verbeter tot 2067 MPa. Die frakuursterkte verbeter van 171 MPa tot 242 MPa. Die resultate was onderwerp aan statisteise analise en 'n statistiese betekenisvolle resultaat is verkry ($p < : 05$).

Gevolgtrekking Glasvesel versterkte polisulfoon komposiet versterker mag die fraktuur eienskappe van polimetiel metakrilaat gebit basis polimer aansienlik verbeter.

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CHAPTER 1

1. INTRODUCTION

To most of our patients, loss of the natural dentition is a matter of great concern and the replacement with artificial substitutes, such as complete dentures, to preserve and reinstate a comparable functional result, is vital to the continuance of normal life.

One of the problems encountered in the provision of such prostheses is whether the constraint of strength and design is able to accommodate the functional and parafunctional demands of the oral cavity.

The material most commonly employed in construction of complete dentures, is the acrylic resin, poly methyl methacrylate, introduced in 1936 (Darbar, Huggett & Harrisson). Despite its popularity the material, although adequate in satisfying aesthetic demands, is not ideal in fulfilling the mechanical requirements of such appliances. This is reflected in the unsolved problem of denture fracture. An analysis of the incidence of denture fracture conducted by Hargreaves in 1969 suggests that sixty eight per cent of prostheses break within three years after fabrication. With the advent of osseointegration in clinical dentistry, the incidence of prosthesis fracture is expected to increase as the functional forces have been shown to increase significantly with implants supported prostheses.

Various methods have been suggested to improve the mechanical properties of denture base polymers. The development of fiber composite materials in industry has inspired a new approach to improve the performance of the polymers used in prosthetic dentistry. One such approach is the inclusion of fiber e–glass polysulphone composite.

The purpose of this study was to determine the effect of prefabricated e-glass fiber composite reinforcement on the flexural modulus and flexural strength of both conventional (Vertex) and high impact (Lucitone) heat polymerizing denture-base polymers

1.2 LITERATURE REVIEW

Due to excellent aesthetic properties, ease of manipulation and relative low cost, poly methyl methacrylate (PMMA) has been the material of choice for the construction of complete dentures since the Second World War. During this stage a shortage in rubber was experienced, which terminated the use of Vulcanite (rubber) and PMMA acquired its highly acclaimed position till the present day.

PMMA is composed of polymers with, complex molecules of high molecular weight. These molecules form through a process of polymerization, which by definition is a repetitive intermolecular reaction that is functionally capable of proceeding indefinitely. During the polymerization process the volatile monomer (liquid) is converted to a solid. An initiator (an organic peroxide), present in the polymer (powder), is decomposed into active free radicals. Decomposition of initiator through heat occurs in heat-cured resins and chemically in cold cured resins. The active free radicals react with the double bonds of the methylmethacrylate monomer molecules. Activated monomers react with additional monomer units and a growing polymer chain is formed (Craig 1985).

Heat must be limited to an acceptable level during polymerization, as excessive heat causes bubbles, which lead to porosity. Polymerization shrinkage is an additional strength-reducing factor linked to monomer. The

weight per unit volume/density of polymer is 25% greater than that of monomer. Usage of a powder to liquid ratio of 3:1 minimizes the problem, as increasing heat and shrinkage are directly related to monomer (Craig, 1985).

Complex biomechanical properties of the masticatory system, and patient handling are conducive to failure of 68% of poly methyl methacrylate appliances within the first three years after manufacturing (Hargreaves, 1969). Approximately one million denture repairs were done in the UK during 1994 at a cost of 7 million pounds (Darbar *et al,* 1994). The high incidence of denture fracture within the first three years after manufacturing does not occur as a result of deterioration, but appears to be due to stresses produced within the appliance (Hargreaves, 1969).

Stress in a body is the force per unit area, S= F/a. The stress formed on a body is equal in intensity and opposite in direction to the applied external force. Since the stress in a structure varies directly with the force and inversely with the area over which it is applied, it is important to recognize that the area over which the force is exerted is the important factor.

All stresses can be resolved into three basic types; tension, compression and shear.

Tension results in a body when subjected to two sets of forces that are directed away from each other, compression results when two sets of forces are in the same straight line and directed towards each other. Shear is the result of two sets of forces being directed parallel to each other in opposing directions.

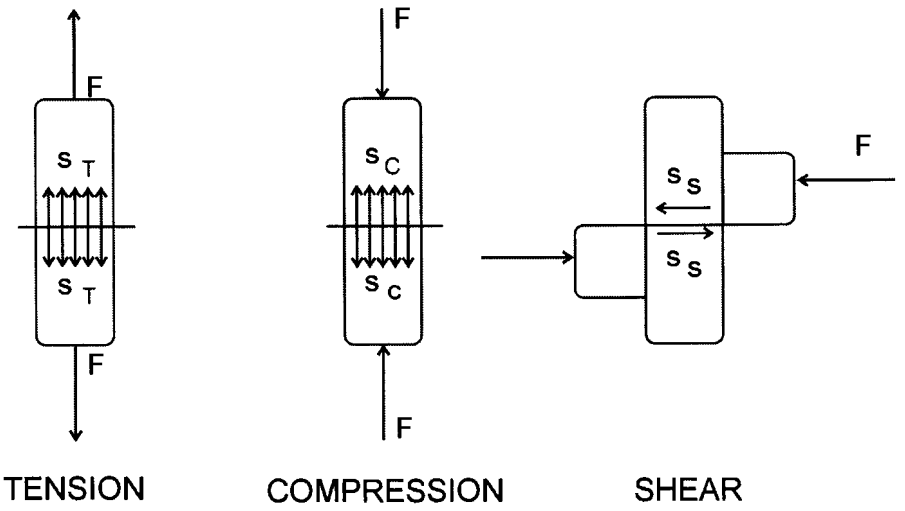

Fig. 1. Illustration of shear, compression and tension

Deformation resulting from compression produces a shortening in the length of a body and tension result in an elongation of the length of a body. Strain is described as the change in length occurring per unit length of the body when a stress is applied. Strain has no unit of measurement, but is represented as:

$$\text{Strain} = \frac{\text{Deformation}}{\text{Original length} - \text{length after application of force}}$$

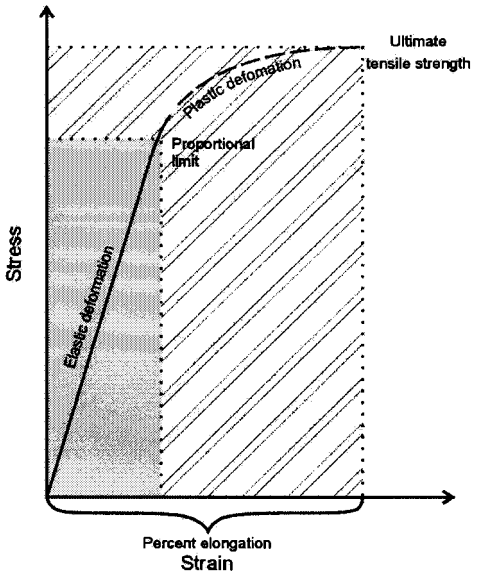

Fig. 2. Illustration of stress versus strain curve

In region of elastic deformation, stress is proportional to strain, and a body will, therefore return to its original length after application of this force.

The proportional limit, represents the highest force at which a body will return to its original length.

Plastic deformation represents the area at which a force has been applied that will cause permanent deformation in a body, the body will not return to the original form.

Ultimate tensile strength represents the maximum stress just before the object breaks.

No permanent deformation will occur in a structure if a stress is applied below the proportional limit. The material is elastic in nature and if the material is stressed, to a value below the proportional limit, the structure will return to its original dimensions.

The region of the stress strain curve beyond the proportional limit is called the plastic region, a material in this region does not return to its original dimensions.

The elastic limit is defined as the maximum stress a material can withstand without permanent deformation. The elastic limit and the proportional limit in a structure represents the same stress. The elastic limit describes the elastic behaviour of a material and the proportional limit deals with the proportionality of stress to strain in a structure.

Yield strength describes the stress at which a material begins to function in a plastic manner. Yield strength is defined as the stress at which a material exhibits a specified limiting deviation from proportionality of stress to strain.

The ultimate tensile or compressive strength is the maximum stress that a material can withstand in tension or compression.

Fatigue failure occurs in the event of an appliance been subjected to alternating stress applications below the proportional limit for such a material. Appliances such as removable partial dentures and complete dentures are subjected to repeated flexure. These stresses can be less than the proportional limit, but failure may eventually occur. For this reason it is important to know

the approximate number of stress cycles that a material will be able to withstand before fracture occurs. This is known as the fatigue limit.

To determine tensile properties of brittle materials. The diametrical compression test can be employed. A disk of the brittle material is compressed diametrically in a testing machine until fracture occurs. The compression stress applied to the specimen introduces a tensile stress in the material in the plane of the force application of the test machine. The formula demonstrates that the tensile stress is directly proportional to the load applied in compression.

$$Q = \frac{2P(\text{load applied})}{\pi \times D \times T \quad (\text{Diameter} \times \text{Thickness})}$$

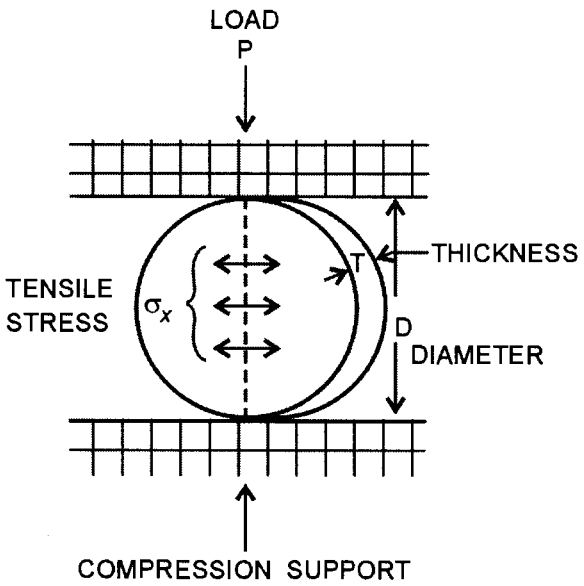

Fig. 3. Illustration of the relationship between tensile and compressive strengths

Certain properties observed in tension can also be observed in compression.

It is possible to obtain an elastic modulus of a material in compression from the ratio of the stress to the strain in the elastic region. Such a value is similar whether the material is in compression or tension. A proportional limit or yield strength can also be observed in compression.

$$\text{Ultimate compressive strength} = \frac{\text{maximum force}}{\text{cross sectional area}}$$

Elastic modulus represents the relative stiffness of a material within the elastic range. Elastic modulus can best be determined from a stress strain curve by calculating the ratio of stress to strain. The intermolecular forces of a material are responsible for the property of elasticity. The stronger the basic attraction forces, the greater are the values of the elastic modulus and the weaker the forces of attraction the less rigid is the material and the lower the elastic modulus.

Toughness represents the energy required to stress the material to the point of fracture.

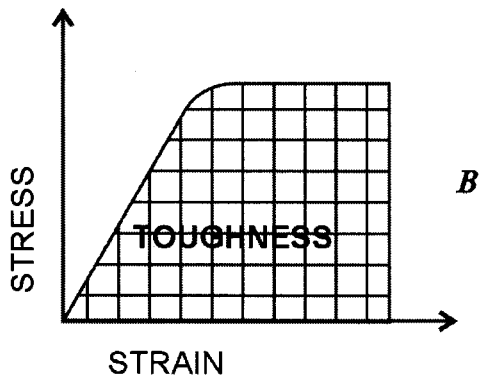

Fig. 4. Schematic presentation of toughness

During loading of a simple beam, which is supported at the ends, with the load applied in the middle, the transverse strength of the material can be obtained. The transverse strength is also known as the modulus of rupture or flexural strength.

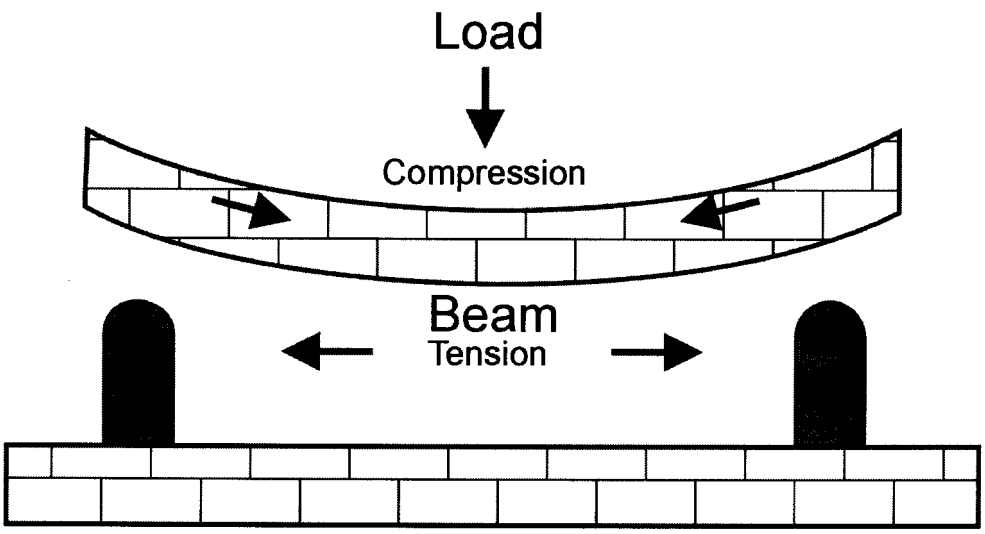

Fig. 5. Tension and compression forces applied on a beam

1.2 CAUSES OF DENTURE FRACTURE

The causes of numerous failures of polymer appliances have been investigated macro-and microscopically. Polymers subjected to a tensile stress, which is applied along the long axis of the polymer chains, moduli of elasticity which are surprisingly high and in the region of the modulus of elasticity of certain alloys. In the event of the applied stress being perpendicular to the long axis of the polymer chains, the modulus of elasticity is inherently low.

Microscopically, the reason for this discrepancy is related to nature of the chemical structure of the polymer material. The bond between atoms within polymer chains are high, but the bond between atoms of adjacent chains are weak. The low strength observed in polymers compared to metals is due to the weak bonds between polymer chains.

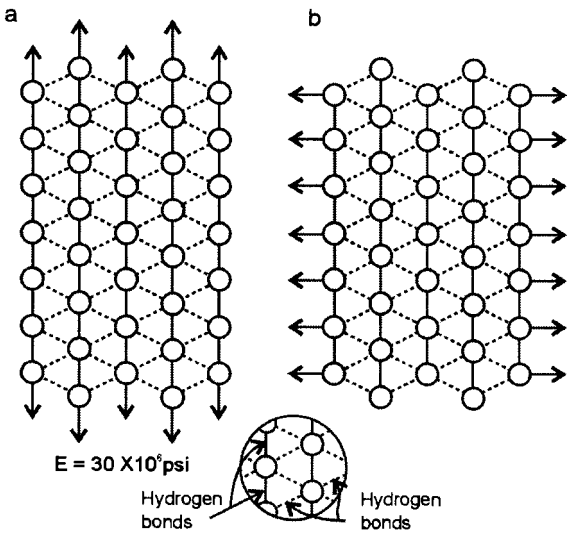

Fig. 6. Illustration of polymer bonds. Strong bonds exist between atoms within a polymer chain (a), weak hydrogen bonds exist between polymer chains (b).

On macroscopic level, denture fracture is the result of initiation and propagation of a crack which can be caused by different factors, such as changes in contour, pinholes, inclusions and deep scratches. Deep notching in the midline of the labial frenum results in numerous failures due to high stress locations at the tip of the notch. The most heavily stressed area in the denture base resin, is the area lingual to the incisors where tensile stresses are observed (Darbar *et al*, 1994).

1.2.1 Processing stresses

Internal or residual stress can be induced in a material by rapid cooling from an elevated temperature (Matthews & Wain, 1956). Inhibition of natural

dimensional change will lead to stresses in the structure with distortion or warpage if such stresses are released.

Friction between the mold walls and soft resin, which are tensile in character, may inhibit normal shrinkage of resin. Other factors, which contribute to processing stresses, are water absorption, variations in thickness of denture base resin and localized polymerization shrinkage, seen around porcelain teeth. No beneficial results have been obtained with techniques that could eliminate processing shrinkage. Dimensional changes seem to be in the order of 0.1mm-0.2mm, during processing.

1.2.2 Crazing

Warpage may be of no clinical importance, but relaxation of surface stresses may lead to crazing or cracks, which are of great detriment to the denture base. Crazing may consist of small cracks that are of microscopic dimension or large enough to be visible with the naked eye. These cracks may initiate fracture. Crazing most commonly is visible as a foggy area on the denture surface. Crazing may also occur as a result of weak solvents, such as alcohol. This can be eliminated by cross-linking of the polymer. Crazing, due to mechanical stress, appear under tensile stress with the crack at right angles to the tensile stress, which are indicative of a separation of the polymer chains or groups of chains.

1.3 Biting forces

Dental prostheses are required to withstand certain forces in service and during fabrication. Bite force measurements have been used as an indicator of masticatory function. In healthy subjects the maximum bite force in the molar region averages 300 to 500 N (Bates, Stafford & Harrison, 1976). Removable

prostheses are associated with a reduction in bite force, which is most pronounced in complete dentures where only about 14 percent of the force is exerted (Craig, 1985). The introduction of implant supported prostheses results in functional forces similar to those recorded in patients with natural teeth (Lindqvist & Carlsson, 1985). Distribution of biting force between anterior and posterior differs significantly. Ramfjord & Ash in 1971 showed that the biting force on anterior teeth is one eighth of that generated on molar teeth. The reduced forces on anterior teeth presumably allows them to provide guidance in excursions.

Parafunctional habits that generate horizontal forces have been shown to cause a high incidence of bone resorption (Hobo, Ichida & Garcia, 1989) and fatigue failure of prostheses due to loss of support (Beyli *et al*, 1981).

Excessive biting force contributes to fatigue failure of dental prostheses. Fractures occur mostly in the maxillary midline. Women exhibits smaller arches with less bulk of material, which predisposes the denture base to failure in female patients, during function. 40% of failure in PMMA prosthesis are caused by fatigue failure (Smith , 1962).

1.4 Degenerative changes

A prime requisite when constructing an oral prosthesis is the maintenance of the supporting tissue in a physiological condition. Despite the best prosthodontic techniques, underlying soft tissue often undergo degenerative changes, which are often exacerbated by poor health and poor nutritional status of the patient (Darbar *et al*, 1994).

Deformation of denture base during function will affect both the supporting tissues and the denture base itself. Denture base deformation is considered to be a contributing factor in residual ridge reduction (Lambrecht & Kydd, 1962), with denture bases exhibiting a lateral deformation some 8.5 times greater than that of metal bases (Koivuma, 1958). Deformation, however is affected by the anatomy of the supporting tissues, with high ridge bases exhibiting tortional deformation and compression occurring with flat ridges, (Koivuma, 1958).

Elasticity in a denture base causes the strain of masticatory forces to be borne by a limited part of the ridge, close to the point of application. Compression and tension are the two forces found to have significant effect on denture base deformation. Lateral tension exists in the posterior part of the denture, which is in direct proportion to the applied force. Thickness, size and contour of the denture palatal area can modify the amount of extension. Photo stress analysis indicate compressive stress occur toward the tissue surface with greatest stress beneath the teeth and on the ridges than towards the palate (Beyli *et al,* 1981)

Swallowing and bruxism produce the same deformation pattern. Deformation in the labial flange is primarily compression. Torsional components are also present. Lowest stress is measured in the midline, which increases from anterior to posterior region over the ridge with maximal stress in the molar region (Johnson, 1965).

Deep notching in the midline of the labial frenum results in numerous failures due to a high location of stress at the tip of the notch. The most heavily stressed area, during mastication, is the area lingual to the incisors. For this

reason this area should be constructed with the maximum permissible cross-sectional area (Neihart, 1988 and Polyzois, 1995).

1.5 Fracture Mechanisms

Anselm- Wiskott *et al,* (1995) demonstrated that stress intensity is associated with the shape of the crack in a specific location with respect to the loading geometry. Toughness of a material can be defined as the resistance to rapid crack propagation and is independent of the size of the initiating crack. On the contrary, strength is dependant on the size of the initiating crack and is therefore limited by the size of processing defects (Ruffino, 1985).

Crack propagation can proceed transgranular or inter granular. In Poly-methyl- methacrylate appliances, fractography depends on molecular weight of polymer and strain rate. Fractures are initiated within defects and propagate on a circular front generating a relatively smooth and highly reflecting mirror region. As crack increases in size, it generates enough energy to form secondary cracks. These cracks interact with primary cracks to produce steps on the fracture surface that look like conic sections. Eventually the energy associated with the accelerating crack is sufficient to cut out a rough area known as a rib band. Deceleration takes place with smooth area formation and acceleration forms a second rib band. This process continues until catastrophic failure of the PMMA occurs (Kusy and Turner, 1974).

Strength of the denture base can vary considerably, depending on factors such as composition of the resin, processing technique, environmental influences and patient factors. Consequently, various approaches to improve the physical and mechanical properties of denture-base resins have been suggested. These include chemical modification of the denture base material by inclusion of

cross-linking agents or by grafting rubber-like substances to the polymer chains to produce the high impact resins.

In conventional heat polymerizing resins, the best physical and mechanical properties are obtained by using high molecular weight distributions, low plasticizer content together with cross-linking agents. The monomer phase contains cross-linking agents capable of defusing into the polymer beads. These agents reduce the tendency of the denture-base to craze under stress, accelerates the increase in the curing systems molecular weight and combats the effect of oxygen inhibition (Craig, 1985). Excessive amounts of cross-linking agents cause PMMA to become brittle. The principal cross-linking agents used commercially are ethylene glycol dimethacrylate and 1-4 butylene glycol dimethacrylate.

High impact denture-base resins were developed to improve the mechanical properties of conventional PMMA. These materials are rubber-modified acrylic resin where a styrene-butadiene rubber is co polymerized with methylmethacrylate. The monomers used contain little or no cross-linking agents and the inclusion of rubber dispersed within the polymer beads during suspension polymerization, produces a craze inhibiting effect. (Rodford, 1986). The modification has been shown to increase the impact strength considerably, and is accompanied by some increase in transverse deflection. It also features decreased stiffness, water absorption and a low free monomer content as compared with conventional resins (Rodford, 1990).

Although the high impact resins have been shown to improve fracture resistance, some doubt exists as to their fracture toughness (Stafford, 1968 and Niehart & Flinton, 1988).

1.6 Denture-base polymer reinforcement

Over the years various approaches to strengthen acrylic resin prostheses have been suggested including incorporating solid metal forms and various types of fibers to provide reinforcement to fracture prone areas.

Stainless steel wires have been the most popular type of strengthener. The strengthening effect of metal wires incorporated into PMMA vary greatly. A 1,3mm diameter circular wire produced an appreciable increase in transverse strength (Carrol and von Fraunhofer, 1984). Ruffino in 1984 suggested that the incorporation of two braided plats, several millimeters apart, should produce significant resistance to flexure resulting in a reduction of denture-base fractures.

Semicircular steel wires have produced comparatively good results as a strengthener (Vallittu and Lassila, 1992). The shape of the semicircular wire relative to the direction of the loading force did not affect the strength of PMMA. There have also been some interesting findings by Vallittu 1993, revealing that the position (tension, middle or compression side) of the steel wire or mesh did not affect the transverse strength of the test specimens.

Metal strengtheners with macroscopic retention have shown a slight increase in strength and strengtheners with microscopic retention, such as sandblasted surfaces, have demonstrated a clear increase in strength (Vallittu and Lassila, 1992).

The presence of a chemical bond between the resin and metal, such as silicoating, enhanced the strength of PMMA-metal composite when the metal

was located on the tension side of the specimen (Kawano, Miyamoto and Tada, 1990).

The effect of metal inclusions on the transverse strength of PMMA has not been shown to be of practical value. It could be assumed that, during cyclic loading of the composite material, any enhancement of strength increase would be even less. This is due to the small interface layer present between the PMMA and the metal wire (Vallittu, 1995).

Ultra-high modulus polymethylene (UHMPE) fibers which display natural colours, inertness and acceptable biocompatibility, have demonstrated strengthening of PMMA (Ladizesky, Ho and Chow, 1992). Although these fibers possess low surface energy and poor wettability, it was reasoned that plasma-treatment would etch the fibers and improve the mechanical bond with the matrix material. However, plasma-treated fibers showed only slight improvement in fracture resistance of PMMA (Clarke, Ladizesky and Chow, 1992). The study by Ladizesky et al, 1992 revealed an interesting phenomenon in the stress strain curves. When reinforcing fibers where plasma-treated, a slight roughness was observed in the curve, whereas, untreated fibers produced smooth curves. This phenomenon has been related to fiber peel in plasma-treated fibers.

Kevlar, a synthetic aramid polymer fiber, and carbon fibers have demonstrated improved flexural and impact strengths when used to reinforce PMMA (Berong, Weed & Young, 1990) and (Schreiber, 1971). Limited use of both these types of fibers is mainly due to undesirable aesthetic properties.

In recent years considerable attention has been given to glass-fiber reinforcement of denture-base polymers.

The mechanical properties obtained from glass-fiber reinforcement of PMMA depend on fiber structure, orientation, volume fraction and adhesion between fiber and polymer matrix.

Glass fibers have been investigated in various forms such as continuous, woven and chopped fiber rovings. Of all these forms, partial fiber reinforcement using continuous electrical glass fiber placed at right angles to the direction of load application, has been reported to enhance the mechanical characteristics of removable prostheses "in vitro" (Vallittu, 1996) and "in vivo" (Vallittu, 1997).

Continuous unidirectional fibers demonstrated improved strength and stiffness for the composite, only in one direction, namely, in the direction of the fibers (Vallittu & Docent, 1999). Therefore the reinforcing effect of these fibers are anisotropic in contrast to woven fibers, which demonstrate reinforcement in two directions and are orthotropic. The reinforcement effect of woven fibers was however significantly reduced.

The optimal volume % fiber appears to lie between 50 and 60% (Vallittu, *et al*, 1994). Higher concentrations of fibers within the matrix material appear to cause clumping together of fibers thereby reducing reinforcement.

Adequate pre-impregnation of the fibers is a prerequisite for bonding of the fibers to the polymer matrix. Heat-cured denture-base polymers form high viscous dough with poor wetting abilities (Vallittu *et al*, 1999). Fabrication of the conventional fiber/PMMA matrix system therefore requires the use of excessive amounts of methyl methacrylate monomer for adequate impregnation of the fiber rovings. This has been shown to increase the

polymerization shrinkage of PMMA this affecting the dimensional accuracy of the appliance (Vallittu, 1996) and contributing to internal void formation within the fiber/PMMA matrix (Vallittu, 1995).

To overcome the problem of impregnating the fiber rovings with dental PMMA, a new technique is required. One solution might be the development of a product in which the fibers are impregnated by the manufacturer.

Water storage has revealed a decrease of approximately 20% in the tensile strength of E-glass fiber composites. It has been reported that elements such as boron oxide (B_2O_2), which are added to E-glass fibers to decrease the high calcium oxide (CaO) content, may increase the hydrolitic degradation and negatively influence the polymer-fiber system (Vallittu, 1999). Polymers with a relatively high water absorption may increase the hydrolysis of the polyoxilane network, increase the leaching of the glass-fiber surface and also promote plasticization of the matrix. Voids, which are produced during polymerization shrinkage, contribute to water absorption through capillary effect, can damage the fiber-matrix interface (Vallittu *et al,* 1994).

There is evidence from dynamic "in vitro" tests that glass fiber reinforcement increased the fatigue resistance of dental appliances up to 100 times compared with fatigue resistance of unreinforced prostheses (Vallittu, 1996). This finding is not surprising because fiber composites are reputed to have high fatigue strength and have therefore been used for years in the aircraft industry (Ouring, 1995).

1.7 Repair of fractured acrylic resin

The ultimate goal of denture repair is to restore the original strength of the denture and avoid further fracture.

The highest transverse strength obtained from repaired dentures is about 80% of the original strength when heat-cured resin is used (McCroice & Anderson, 1960) and (Polyzois, 1995). Documented values using autopolymerizing PMMA range between 35-75% of the original strength (Morandias *et al*, 1982). Furthermore, the fracture of repaired devices often occurs at the junction of the new and the old material rather than through the center of the repair. This finding clearly indicates that the interface of the old and new materials is the location of stress concentration.

Fragments of a broken denture show an increase in internal stress. This increase is due to the breaking of molecular bonds at the point of fracture. Alteration of the molecular forces within each fragment results in distortion of the fractured sections, which often causes difficulty in accurately reassembling the broken pieces.

Denture repairs involve joining two parts of the fractured denture with a denture repair material. The success of the repair relies on the phenomenon of adhesive bonding. Correct surface preparation of the two sites to be joined predisposes a strong bonding interface which improves the repair strength and stress concentrations.

Roughening the surface to be adhered is standard method to enhance bonding and apparently creates more surface area to improve van der Waal's force attraction (Cagle, 1973). The effect of repair surface design on transverse

strength of repaired PMMA has been documented by Ward *et al*, 1992. They found the transverse bend strengths of the rounded and 45 degree bevel joints with a 1.5mm space to be superior to other induced profiles.

Roughening the surface however introduces microvoids and overhanging grooves. Because monomer is not a powerful solvent of PMMA, painting or immersing the surface will not efficiently remove the debris and create a particle-free surface for bonding. Immersion of roughened surface in chloroform, a powerful solvent of PMMA, for a period of 5 seconds revealed removal of microdebris and elimination of overhanging grooves, this creating a cleaner and more efficient site for effective bonding (Shen *et al*, 1984). Since polymer beads are dissolved using this procedure, it is suggested that the monomer from the repair material may form a penetrating network across the interface into the parts to be joined.

CHAPTER 2

2. MATERIALS AND METHODS

The materials and methods employed during this study will be documented in sections which are broadly co-incident with the experimental sequence adopted for the investigation.

2.1 Preparative procedures

2.1.1 Reinforcement rods

Unsized (without silane coupling agent) continuous E-glass fiber polysulphone composite rods[1] were used as reinforcement for the denture-base resins to be evaluated. The rods had a diameter of 3 ± 0.1mm and a fiber concentration of 55 ± 3 vol/%. The fabricated rods, 150mm in length, were cut into 28mm sections with a diamond disc for incorporation into the test samples. The particulars of the materials used in this investigation are listed in Table 1 and 2.

1. **Fibertek Developments (Pty) Ltd.**
   Irene, South Africa

Table 1 Materials used in this investigation

Type of Material	Code	Manufacturer	Batch No
Glass Fiber E-Glass Fiber P336B	GF	AFI Johannesburg, South Africa	586.09
Polysulphone Ultrason-NaturB S2010	PSu	BASF Ludwigshafen, Germany	17-8147
Reinforcement Rods Glass-fiber Polysulphone CompositeComposite	GF/Psu	Fibertek Developments (Pty) Ltd. Irene, South Africa	
Denture Base Polymer Vertex, Rapid Simplified, Clear Powder Liquid	PMMA (V)	Dentimex B.V Xeist, Netherlands	 GK 294PO1 GK 3811L14
Lucitone 199 Powder Liquid	PMMA (L)	Dentsply International In York, PA USA	980617

Table 2 Physical properties of the reinforcement material

Product	Size (μm)	Strength (MPa)	E-Modulus (MPa)	Elongation (%)
GF	10	3 400	73 000	2.8
PSu (Ultrason)		80	2 600	5.7

Polysulphone was the thermoplastic material employed in the manufacturing of strengtheners. It is an amorphous, thermoplastic poly-condensation resin with the following chemical structure.

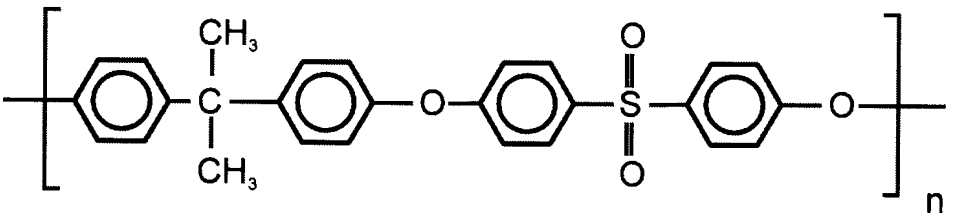

Fig. 7. Illustration of the structural formula of polysulphone

Polysulphone is a polymer with great strength, rigidity and ductility with a glass transition temperature (Tg) of approximately 270°C.

2.1.2 Unreinforced samples

The conventional heat polymerizing denture-base resin Vertex (PMMA-V) and the high impact resin Lucitone 199 (PMMA-L) were used in this study. The PMMA-V contained the cross-linking agent EGDNA which has been shown to minimize the effect of water absorption on the flexural strength of PMMA. A powder (Liquid P/L) ration of 3:1 parts by volume was mixed for 30 seconds and allowed a doughing time of ± 15 minutes before packing in a custom-made aluminium alloy split mould. A P-L ratio of 3.2:1 parts by volume was used for PMMA-L, allowed a doughing time of ± 9 minutes and packed in the split mould.

The split mould was assembled and final packing performed in a hydraulic flask press at 100Kpa/cm3. the united sections of the mould were bolted together, secured in a flask clamp and polymerized in a water bath. A curing cycle of 20 minutes at 100°C, and 180 minutes at 63°C and 30 minutes at 100°C followed by a slow cool down to room temperature was adopted for PMMA-V and PMMA-L respectively.

The cylindrical test samples were moulded to a standard length of 26mm and diameter of 6mm in the pattern sections of the mould. After demoulding, each sample was measured for dimensional accuracy with Mitutoyo Electric Dial Calipers[2] and stored in distilled water at 37°C or 8 weeks before testing. Ten unreinforced samples were fabricated for each of the two different denture-base polymers. These samples were employed as controls.

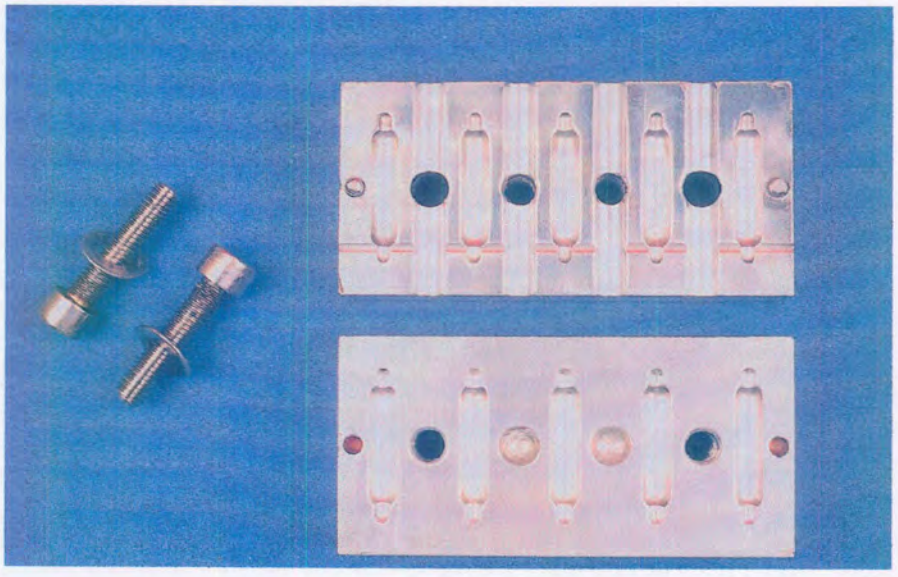

Fig. 8. Aluminum alloy split mould employed for fabrication of resin samples

2 **Mitutoyo Corp., Tokyo, Japan**

2.1.3 Reinforced samples

Identical technical procedures as used in the fabrication of the unreinforced specimens were employed to fabricate the reinforced test samples. The GF/PSu reinforcing rods (28 x 3mmθ) were immersed in chloroform for a period of 5 seconds prior to being positioned longitudinally across the resin filled sections of the mould. The rods were localized 0,5mm axial from the bottom of the tension side of the specimens. Accurate placement of the rods were facilitated by localizing slots (2 x 3mm) machined on the extremities of the pattern sections of the split mould. The protruding ends of the reinforcement rods were marked on the tension (bottom) side with red Dura Lay autopolymerizing resin. This was performed to ensure correct localization of the reinforcement in the split-mould prior to polymerization. These markings were also used for correct orientation of the samples on the transverse bending test jig used for flexural evaluation.

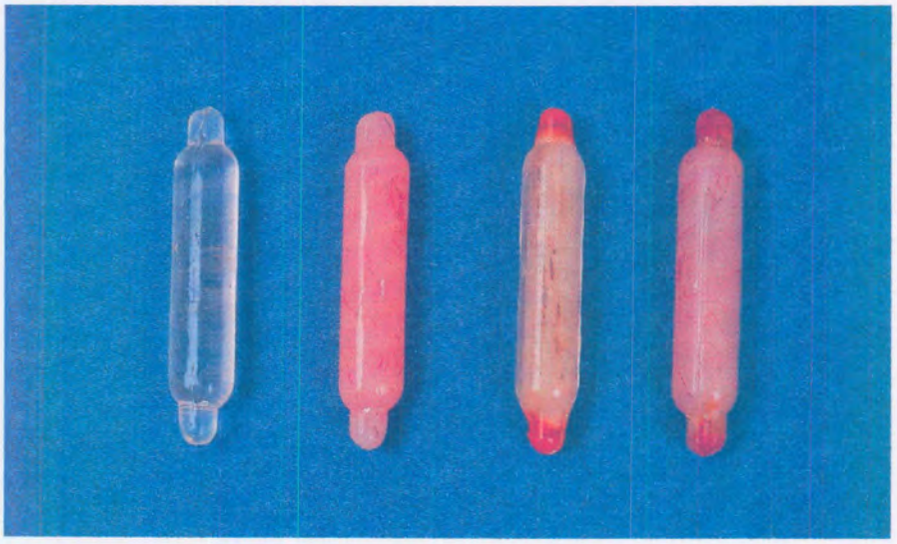

Fig. 9. Fabricated unreinforced and reinforced test samples

Ten samples were fabricated from each of the two denture-base polymers.

These specimens were also measured for dimensional accuracy and stored in distilled water at 37°C for a period of 8 weeks before testing.

2.2 Mechanical testing procedure

A three-point loading test was conducted in air on all the specimens. The Zwick testing Machine Model Z010[3] was employed to record the stress-strain relationship in transverse bending, operating at a cross-head speed of 5mm/min. The round indenter (2mmθ) engaged the correctly positioned test specimens midpoint between the round supports with a span-distance of 20mm on the Bencor Multi T System[4] test jig. The specimens were subjected to transverse testing under three-point loading until initial fracture occurred. During loading, the cylindrical specimens behave as freely supported beams which are deflected at a constant rate. The flexural behaviour or stress-strain relationships were recorded by the computerized chart recorder. Graphic representations of the flexural characteristics were obtained and analyzed.

3 **Zwick, ULM, Germany**

4 **Danville Eng. Inc., San Raman, USA**

Fig. 10. Illustration of Zwick testing machine

The following properties of all the test specimens were determined:

1. Flexural modulus (FM)
2. Flexural strength (FS)

The flexural modulus was derived from value substitution in the following formula (ISO 3597, 1993)

$$FM = \frac{4L^3F}{3\pi 4\delta}$$

where F is the applied load (N) at the highest point of the load deflection curve, L is the span length (20mm), d is the diameter of the test specimens

and δ the mid-point deflection of the specimen corresponding to F at a point in the linear section of the load versus displacement graph.

The flexural strength was calculated from the following formula (ISO 3597, 1993)

$$FS = \frac{8FL}{\pi d^3}$$

Both FM and FS values were expressed in MPa.

Statistical Analysis

The flexural moduli and flexural strength for the conventional (Vertex) versus the high impact (Lucitone 199) denture-base base polymers were statistically compared by t test at $p<0.05$. Significance level. The differences between the SF/PSu composite reinforcement and the test samples were analyzed.

CHAPTER 3

RESULTS

The results of the flexural modulus and flexural strength tests are listed in Tables 4 to 9 and summarized in Table 3. These values are also illustrated in bar graphs in Figure 11 and 12. the relevant statistical analyses are given in Tables 10 and Table 11.

The prefabricated glass fiber polysulphone composite reinforcement produced the highest values for flexural modulus (14 106 MPa) and flexural strength (546,6 MPa). All the reinforced specimens exhibited significant reduction in FM and FS values as compared with those obtained from the strengthener before incorporation into the test samples. The reinforcement appears to be negatively influenced by both the denture-base polymers. Due to the huge differences noted, it was deemed unnecessary to conduct a statistical analysis on these values as depicted in Table 3.

The flexural modulus of conventional denture-base polymer was 1746 MPa and glass fiber polysulphone composite reinforcement increased it to 2328 MPa. The flexural strength increased from 164 MPa to 209 MPa.

Conversely, the flexural modulus of high impact denture-base resin improved from 1684 MPa to 2067 MPa. The flexural strength of this resin was 171 MPa and reinforcement increased this value to 242 MPa.

By comparison, as summarized in Tables 3 and illustrated graphically in bar graph format in Fig. 11 and Fig. 12, the highest flexural modulus values were

experimentally obtained for the conventional reinforced resin whereas the flexural strength of the reinforced high impact resin exceeded that of the conventional polymer.

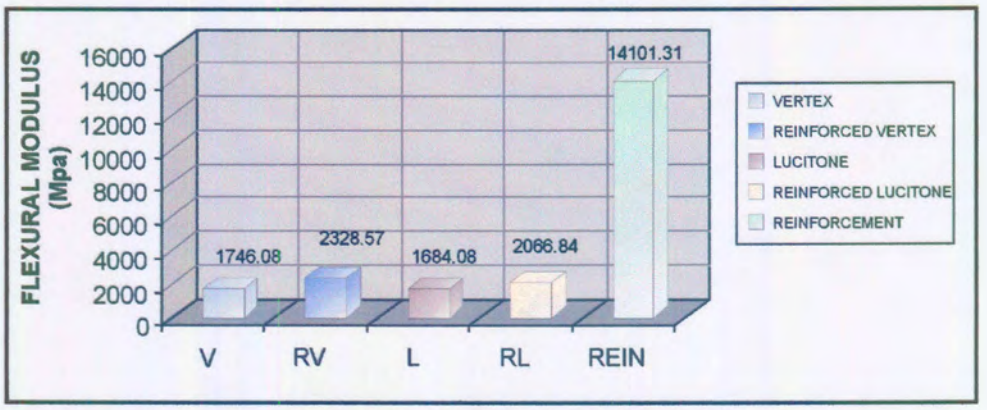

Fig. 11. Graphic illustration of comparative flexural modulus values

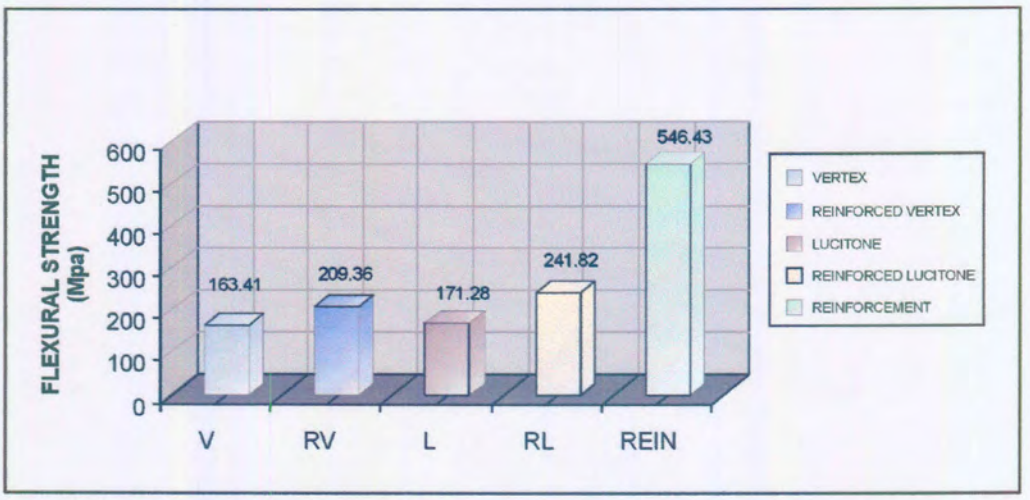

Fig. 12. Graphical illustration of comparative flexural strength values

Table 3. Average values for flexural modulus and flexural strength

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
Conventional PMMA		
Vertex	1 746.1 (394.12)	163.4 (37.01)
Reinforced Vertex	2 328.6 (625.12)	209.4 (59.48)
High Impact PMMA		
Lucitone 199	1 684.1 (243.14)	171.4 (24.73)
Reinforced Lucitone 199	2 066.9 (412.43)	241.8 (48.25)
GF/PSu Strengthener	14 106.0 (3614.5)	546.6 (140.02)

Table 4. Flexural modulus and flexural strength values of fiber reinforced composite

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
E-glass fiber reinforcement	14570.90	564.14
	15681.31	607.43
	12066.31	467.41
Average Values	**14106.31**	**546.43**

Table 5. Flexural modulus and flexural strength values of unreinforced conventional PMMA

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
Conventional unreinforced PMMA Vertex	1652.59	154.68
	1757.28	164.48
	1802.37	168.70
	1800.48	168.52
	1980.61	185.38
	1586.49	148.37
	1788.38	167.39
	1662.80	155.63
	1777.27	166.35
	1652.59	154.68
Average Values	**1746.08**	**163.41**

Table 6. Flexural modulus and flexural strength values of reinforced conventional PMMA

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
Reinforced conventional PMMA Vertex	2076.32	186.86
	2238.65	201.47
	2565.85	230.92
	2701.44	243.13
	2040.62	183.65
	2253.80	202.84
	2244.32	201.98
	2321.62	207.15
	2277.27	204.77
	2565.85	230.92
Average Values	**2328.57**	**209.36**

Table 7. Flexural modulus and flexural strength values of unreinforced high impact PMMA

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
	1774.87	180.50
	1640.89	166.87
Unreinforced high impact PMMA Lucitone	1814.88	184.57
	1703.89	173.28
	1589.77	161.67
	1571.74	159.84
	1682.53	171.17
	1656.66	168.48
	1718.25	174.74
	1687.33	171.60
Average Values	**1684.08**	**171.26**

Table 8. Flexural modulus and flexural strength values of reinforced high impact PMMA

Design	Flexural Modulus (MPa)	Flexural Strength (MPa)
Reinforced high impact PMMA Lucitone	1965.28	229.93
	2083.46	243.76
	1895.90	221.82
	1899.98	222.29
	2308.33	270.07
	2304.66	269.64
	2073.72	243.21
	2012.19	235.42
	2064.97	241.59
	2055.96	240.54
Average Values	**2066.94**	**241.82**

Table 9. Average deflection values of different materials

Design	Deflection (mm)
E-glass reinforcement	0,81
Conventional PMMA – Vertex	1,04
Conventional reinforced PMMA Vertex	0,98
High impact unreinforced PMMA Lucitone	1.30
High impact reinforced PMMA Lucitone	1.13

Table 10 Mean and standard deviations of E-modulus values of different materials evaluated with statistical comparison between samples using t test

Material	Mean ∀ SD	Not Significant *
Vertex Clear	1746.09 ∀ 112.5 →	
Vertex Reinforced	2328.57 ∀ 216.45	p = 0.186
Lucitone	1684.08 ∀ 75.32 →	
Lucitone Reinforced	2066.95 ∀ 143.67	
Glass Fiber Reinforcement	14106.32 ∀ 872.82	

Comparison for all the other materials indicated that the E-modulus was significantly different at $p<0.05$.

Table 11. Mean and standard deviations of flexural strength values of different materials evaluated with statistical comparison between samples using t test

Material	Mean ∀ SD	Not Significant *
Vertex Clear	163.418 ∀ 10.549	→
Vertex Reinforced	209.369 ∀ 19.501	p = 0.92
Lucitone	171.272 ∀ 7.661	→
Lucitone Reinforced	241.827 ∀ 16.810	
Glass Fiber Reinforcement	546.327 ∀ 33.795	

Comparison for all the other materials indicated that the flexural strength was significantly different at $p<0.05$.

Flexural modulus

FM is defined as the slope of the stress/strain curve at any given point. Fig. 11 indicates that the increase in mean FM values of the reinforced Vertex (582 MPa) and Lucitone (382 MPa) was significant ($P \leq .05$). Lucitone however demonstrated slightly lower FM value. No statistically significant differences were obtained from the unreinforced samples ($p = 0.186$).

Flexural Strength

To calculate flexural strength values, the loads at the points of fracture were used. The conditions for testing allowed fracture of the specimens to occur with less deflection to reduce the deviation from simple beam theory. This was achieved by reducing the span-do-depth ratio.

In the FS evaluation as shown in Fig. 12, the mean improvement in FS for the reinforced Vertex (40 MPa) and Lucitone (70MPa) was significant ($p \leq .05$). The FS of Lucitone was slightly higher than that of Vertex and the unreinforced samples showed no significant differences.

Lucitone exhibited higher FS values in reinforced and control specimens than the conventional heat polymerized PMMA. In contrast, the FM values obtained form Vertex exceeded those produced by the high impact denture-base resin Lucitone.

The three-point bending test was carried out at a crosshead speed of 5mm/min. This corresponds to a strain rate of 3% per minute. Fracture was initiated on the tension side directly below the loading nose of the testing machine and propagated towards the compression side of the test specimens.

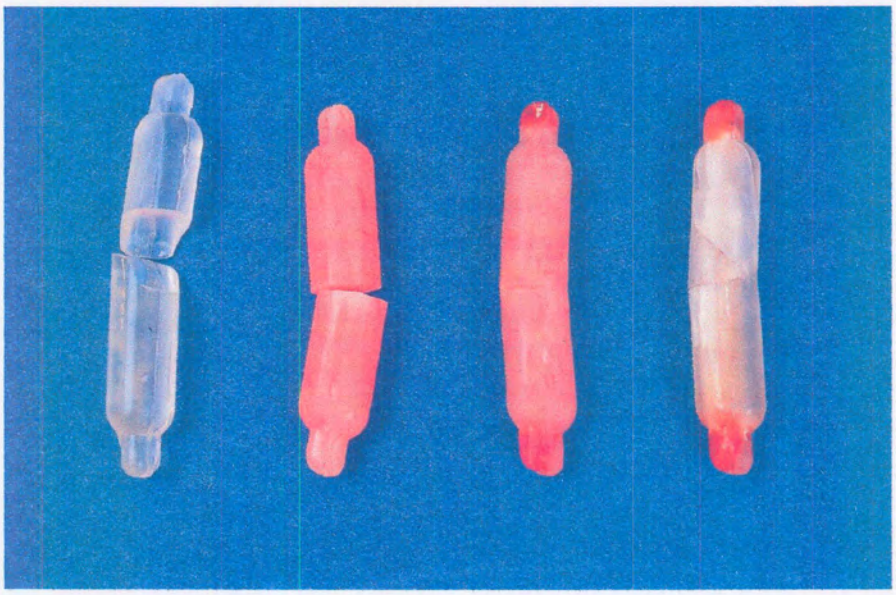

Fig. 13. Fractured unreinforced and reinforced resin samples

Macroscopically, fiber reinforced test samples of the high impact and conventional resins fractured in a ductile manner. On the contrary, all the unreinforced samples, both fractured in a brittle manner. Some evidence of delamination and fiber pull out was observed in the fracture line of both types of denture-base resins.

CHAPTER 4

DISCUSSION

It is known that the eventual fracture of a prosthesis occurs due to crack nucleation and propagation from areas of high stress concentration. Brittle solids, such as PMMA, are generally weak because they contain structural features which act as stress concentrations. These features usually present as defects on the surface and/or within the material. When subjected to loading, the relative inability of PMMA to deform plastically is intimately associated with the fracture process. Brittle fracture occurs at the limit of elastic deformation because crack initiation and propagation are easier to achieve than plastic deformation. These characteristics promote conditions which are conducive to catastrophic failure of unreinforced PMMA at relatively low fracture loads due to fatigue of the material.

The results of the unreinforced specimens indicate that the flexural modulus of conventional PMMA exceeded that of the high impact resin. During the fabrication of high impact resins, rubber is co-polymerized with methylmethacrylate. These grafted inclusions have been shown to introduce some increase in transverse deflection in this material as compared with the relatively brittle conventional cross-linked denture-base resins (Rodford, 1986).

It has been documented that high impact resins are sensitive to surface defects. The presence of these defects are capable of reducing the transverse strength of these polymers to a level below that of conventional heat polymerized PMMA (Rodford, 1990). In this evaluation, the flexural strength values for high impact resins were found to be marginally superior to those of conventional PMMA.

Since recognizing the fact that metal inclusions do not actually behave as reinforcements of polymeric structures, fiber reinforced composites (FRC) are being developed to enhance the mechanical properties of polymers.

The E-glass fibers used in FRC are basically classified according to whether the fibers are preimpregnated or non-preimpregnated. Both these systems have been shown to improve the strength of PMMA. However, there is a lack, of experimental evidence to support the use of glass fiber polysulphone composite as partial reinforcement for denture-base polymers.

This study demonstrated the effect of GF/PSu composite reinforcement on the flexural properties of multiphase polymers using three point flexural strength testing as a method of comparing polymer performance.

Hooke in 1676 discovered that, for small loads, a bar deforms elastically in a mode proportionally to the stress applied.

Accepted engineering beam bending theory (Wang & Salmon, 1965) states that when a beam is loaded midspan between two supporting points, the applied load induces two zones. In the superior layers, the material is placed under compression whereas tensile stresses develop in the inferior zone as shown in Fig. 14. The transition between compression and tension is gradual and occurs along the neutral axis. This is illustrated in Fig. 15.

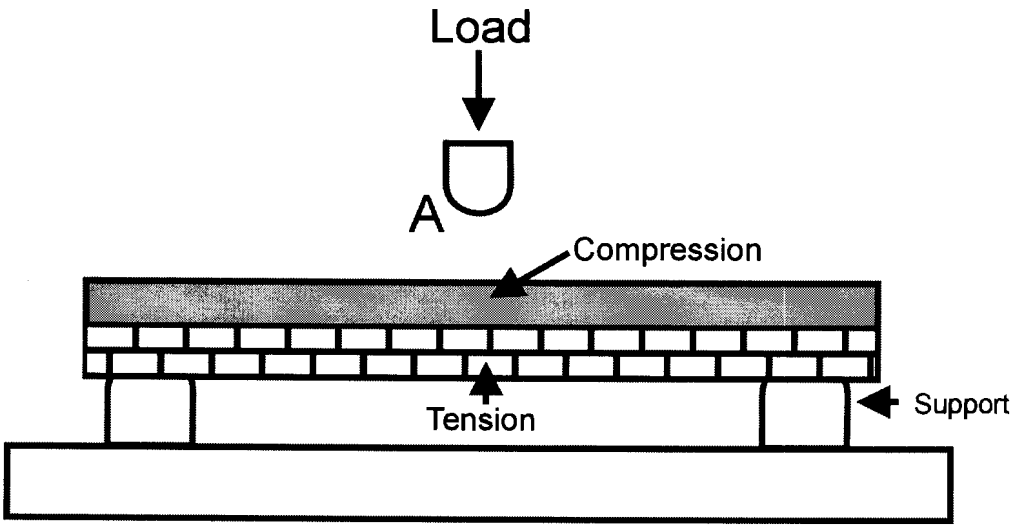

Fig. 14. Centrally loaded end supported beam

The resulting stress distributions across the beam cross section are shown in Fig. 15.

The resultant compressive (F_1) and tensile (F_2) forces induced in the PMMA beam material, constitute the internal beam resistance to the applied external load. The equation of moment resistance is the product of these two forces multiplied by their individual distances (Y_1 and Y_2) about the neutral axis.

With the introduction of reinforcement in the PMMA beam, a third force (F_3) is induced at a distance (Y_3) about the neutral axis (Fig. 15).

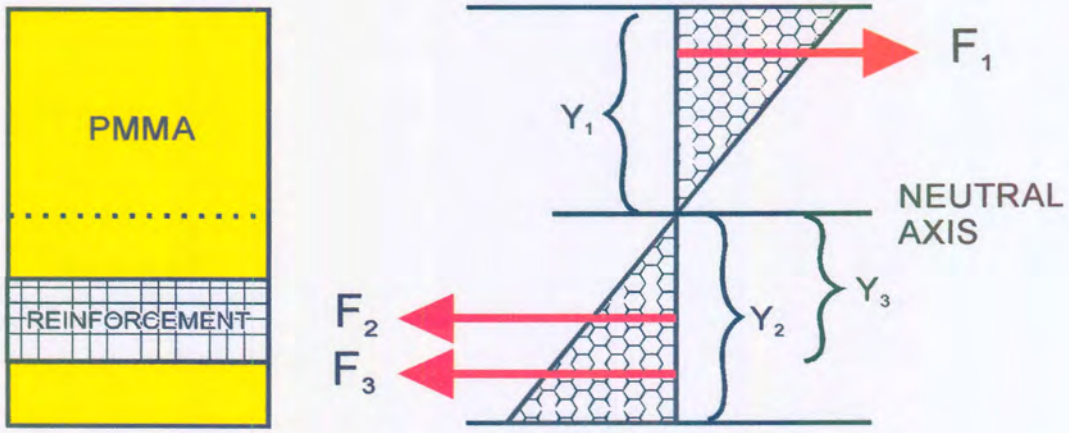

Fig 15. Internal resisting stresses in reinforced PMMA beam

The equation for moment resistance for the reinforced beam can therefore be written as:

Moment resistance = $(F_1)(Y_1) + (F_2)(Y_2) + (F_3)(Y_3)$

From this equation it can be reasoned that reinforcement plays a role in resisting the applied load provided this reinforcement is not at the neutral axis where $Y_3 = 0$. The equation further demonstrates that as the reinforcement distance Y_3 increases, the contribution to reinforcement is also increased.

For this reason the fiber reinforcement rods employed to reinforce the denture-base polymers, were placed 0,5mm from the bottom or tension side of the samples. By adopting this procedure, only a small amount of fibers were located in the neutral axis of the specimens. This distribution facilitated optimal utilization of the reinforcement characteristics of the strengthener within the denture-base polymers.

In the fiber reinforced samples, the flexural moduli and flexural strengths of the two denture-base polymers evaluated, demonstrated different patterns of mechanical improvement. A significant enhancement in flexural modulus, which surpassed that of the high impact was produced by the conventional polymer. The increase in stiffness is due to the reinforcing qualities of the reinforcement which reduces the flexibility of the inherent flexibility of the rubber co-polymers incorporated into high impact resins.

Significant improvements in the flexural strengths of both denture-base polymers were also noted. Despite these improvements, the strength at initial fracture of the high impact PMMA was superior to that of the conventional resin.

Although these findings are basically in agreement with previously reported studies, the reinforcement employed in this study differs from other FRC products, in that the polymer matrix of the fibers in composed of polysulphone instead of monomethacrylate or dimethacrylate polymers, which are commonly used polymers.

The reinforcing efficiency of FRC[s] depends on the component of fiber, fiber orientation, the ratio of fiber to resin and the adhesion between fiber and resin matrix (Agarwal & Broutman, 1980).

Continuous glass fibers reinforcement used in dental applications can be divided into non-preimpregnated and preimpregnated fibers. Preimpregnation can be achieved with light-curing monomer system or with the porous polymer system. In the latter, highly porous polymer pre-impregnated in the fiber forms and interpenetrating polymer network, and in the former, the dimethacrylate monomer in preimpregnation facilitates its use with dimethacrylate resins (Vallittu, 1999).

In this study, non-preimpregnated fibers with fiber concentration of ± 55 vol % were used. Based on the values of fiber volume content and fiber orientation at a 90° angle to the applied load, optimal reinforcement could be expected.

The resistance of a material to crack propagation is characterized by the stress intensity factor Ki. When crack nucleation and propagation occur, Ki has reached its critical level (Kic) and can therefore be equated to the flexural strength of the material (Anselm-Wiskott *et al*, 1995). FRC have been shown to inhibit crack nucleation and propagation. If crack nucleation occurs, propagation is complex. On reaching the fiber matrix/interface, if adhesion is relatively low, delamination is deflected along the interface between the fiber and the matrix, thus enhancing the Kic value. The composite is therefore able to withstand large numbers of such microcracks being formed, before the structure is significantly weakened. Conversely, if the adhesion between the fibers and the matrix is higher than the cohesive strength of the fibers, fracture of the fibers occurs (Ashby & Jones, 1993). In fact, the so-called weak interface theory is based on this phenomenon and has been used in reinforcing composites with a brittle matrix (Cogswell, 1992).

Since the bonding characteristics of the interface between the fibers and the matrix material have been shown to affect the behaviour of the FRC under loading, some consideration should be given to the fiber polysulphone interface used in this investigation.

Conventional PMMA is a multiphase acrylic resin system made from prepolymerized powder beads (predominantly PMMA) and a liquid of monomers such as MMA with a cross-linking agent. Because such a polymer-monomer mixture or dough has a relatively high viscosity, adequate impregnation of

reinforcing fibers with resins has been difficult to achieve (Vallittu, 1994). Impregnation of reinforcing fibers with the resin allows fibers to make contact with the polymer matrix. This is a prerequisite for bonding of the fibers to the matrix, and thus for strength of the composite. Even with low viscous autopolymerizing resin mixture, poorly impregnated regions in the fiber composite namely, voids between the fibers have occurred. This has been shown to considerably decrease the tensile strength and E-modulus of fiber composites (Vallittu, 1998).

Polysulphone is an amorphic thermoplastic resin with a wide variety of properties which allow it to be substituted for thermosetting resins such as PMMA. The degree of impregnation achieved with the fiber polysulphone strengthener was not assessed as it was not part of this investigation. If however, consideration is given to the flexural properties of the reinforcement before inclusion into the denture-base polymers, it is reasonable to assume that the adequate impregnation had been achieved during fabrication of the reinforcement rods. The adhesive characteristics between fiber and polysulphone matrix also requires clarification. Since inclusion of glass-fiber without surface saline coupling treatment weaken the PMMA composite (Vallittu, 1997), it is interesting to note that untreated fibers within the polysulphone has a significant reinforcing effect in this study. This is probably due to the adhesion between the naked fibers and the polysulphone matrix material.

Water absorption also affects the strength of the conventional fiber PMMA composite by affecting the resin, the fibers and their interface. Water storage decreased the tensile strength of PMMA (Stafford & Smith, 1968). The tensile strength of E-glass fiber was decreased by 30% after storage in water for 40 days when compared with the tensile strength of the dry fibers (Eherenstein et

al, 1991) and that of E-glass fiber composite decreased by approximately 20% due to storage in water (Holliday, 1966).

Polysulphone is not prone to water absorption and absorbs about 0.3% water at 23°C. It is therefore unlikely that the detrimental hydrolytic effect of water known as static fatigue, which has a tendency to destroy adhesion at the fiber matrix interface, will be as severe with the polysulphone based fiber composites as that seen with the PMMA composites.

It is recognized that an adequate bond between strengthener and PMMA is a prerequisite for obtaining optimal reinforcement from partial fiber reinforcement of PMMA. As regards the nature of interfacial adaptation between the fully polymerized polysulphone matrix of the reinforcement rod and the denture-base resins, little is known as this aspect was not investigated. However an attempt was made to establish an interpenetrating polymer network between polysulphone and PMMA by dipping the prefabricated reinforcement in chloroform, a powerful solvent of both polysulphone and pmma, prior to inclusion of the reinforcement into the heat polymerizing denture-base resins. This aspect therefore requires further investigation. As it appears to contribute to the discrepancy in flexural properties found between the fiber composite and the reinforced denture-base polymers.

CHAPTER 5

CONCLUSIONS

Subject to the conditions of this study, the following conclusions can be extrapolated from the results:

1. Glass fiber polysulphone composite reinforcement increased the flexural modulus and flexural strength of both conventional and high impact denture-base polymers.

2. Reinforcement of conventional resin produced the highest fiexural modulus.

3. Reinforcement of high impact resin, produced the highest flexural strength.

4. The compatibility of polysulphone with denture-base polymers requires further investigation.

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