CHAPTER 1

INTRODUCTION AND LITERATURE REVIEW

1.1 INTRODUCTION AND LITERATURE REVIEW

Various materials have been used in the past to construct dentures, including cellulose products, phenol formaldehyde, vinyl resins and vulcanite. However, there were a variety of problems related to their use as denture base materials.

Cellulose products suffered from warpage in the mouth, and from a taste of camphor due to its use as a plasticiser. Camphor leached out of the denture, and caused blistering, staining and loss of colour within a few months (Van Noort, 2002). Phenol formaldehyde (Bakelite) was proven to be too difficult to process and also lost its colour in the mouth (Van Noort, 2002). Vinyl resins were found to have a low resistance fracture, and failures were common, most likely as a result of fatigue (Van Noort, 2002).

Vulcanite was the first material to be used for the mass production of dentures, but its aesthetic qualities were poor and it has now been replaced by acrylic resins (Van Noort R., 2002). Acrylic resin (polymethyl methacrylate) is now the material of choice.This material has the required aesthetic qualities, is inexpensive and easy to process. In addition to its use in the construction of complete denture, this material is also used for a wide range of other applications, such as the construction of custom trays for making impressions, denture repairs, soft liners and artificial teeth for removable dentures.

1.2 Requirements of a denture base polymers include:

- Natural appearance
- High strength, stiffness, hardness and toughness
- Dimensional stability
- Absence of odour, taste or toxic products
- Resistance to absorption of oral fluids
- Ability to retain polymers, porcelain and metals
- Ease of repair
- Acceptable shelf life
- Ease of manipulation
- Low density
- Accurate reproduction of surface detail
- Resistance to bacterial growth
- Acceptable thermal conductivity
- Radiopaque properties
- Ease of cleaning
- Economy of use

1.3 Types of denture base polymers

1.3.1 Heat-polymerized denture base polymer

Heat-polymerized denture base polymer is the most widely used denture base material today. The polymer is normally supplied as a powder and liquid. The major component of the powder is beads of polymethylmethacrylate (PMMA). Benzoyl peroxide is normally the initiator that is added to PMMA. PMMA is a clear, glass-like polymer. To produce a more ‘lifelike’ denture base, pigments are incorporated in the PMMA.

The major component of the liquid is methylmethacrylate (MMA) monomer. This is a clear, colourless, low-viscosity liquid with a boiling point of 100.8˚C (Dogan et al., 1995). The liquid normally contains some cross-linking agent, such as ethylene glycoldimethacrylate (EDGMA) (Hill, 1981). The main purpose of adding EDGMA is to provide craze resistance. The inhibitor is used to prolong the shelf-life of the liquid component. In the absence of the inhibitor, polymerization of the monomer and cross-linking agent would occur slowly, even at room temperature and below, because of the random occurance of free radicals within the liquid. The inhibitor, which is commonly used is a derivative of hydroquinone.

The proper polymer-to-monomer ratio is of considerable importance to the final structure of the resin (Jerelimov et al,1985). If the ratio is too high, not all the polymer will be wetted by monomer, and the cured acrylic resin will be granular. If the ratio is too low, there will be excessive shrinkage during polymerization with possible porosity of the polymerized acrylic resin (Jerelimov et al, 1985). Excessive residual monomer can adversely affect the mechanical properties of denture base polymers (Lamb, 1983).

There are four stages that can be identified during the interaction of the powder and liquid (Phillips, 1994);

Stage 1. The polymer gradually settles into the monomer to form a somewhat fluid, incoherent mass.

Stage 2. The monomer attacks the polymer. This process is accomplished by the penetration of the monomer into the polymer; the layer of polymer that is penetrated, sloughs off and either goes into solution or is dispersed in the monomer. This stage is characterized by a stringiness and adhesiveness if the mixture is touched or pulled apart.

Stage 3. As the monomer diffuses into the polymer, and the mass becomes more saturated with polymer in solution, it becomes smooth and doughlike. It is no longer tacky, and it does not adhere to the walls of the mixing jar. It consists of undissolved polymer particles suspended in a plastic matrix of monomer and dissolved polymer. This stage is often called the dough or gel stage.

Stage 4. The monomer disappears, by evaporation and by further penetration into the polymer. The mass becomes more cohesive and rubberlike. it is no longer completely plastic, and it cannot be molded.

Several factors come into play during the processing of the material. They include temperature of the water bath, boiling temperature of the MMA monomer, and the temperature at which maximal polymerization could occur (Virendra, 2004). The monomer boils at about the same temperature as water, i.e at 100°C (Combe, 1992). It is necessary to heat the denture flask at a sufficiently high temperature but at which no boiling of the monomer and its vaporation could occur. Once most of the monomer is partially polymerized into a non-liquid state, the flask can be heated at a higher temperature to achieve the maximum amount polymerization possible without boiling the monomer (Virendra, 2004). Harrison and Huggett,(1984) recommended a polymerization cycle of 7 hour at 70°C plus 1 hour at 100°C as the optimal cycle to produce polymer with high mechanical properties. This cycle also will produce an optimal level of polymerization and avoid the risk of porosity even in thicker section (Harrison and Huggett, 1992; Huggett et al.,1987). Heating at the lower temperature converts most of the monomer into low molecular weight polymers without causing the monomer to boil. The subsequent heating at a higher temperature produces maximal polymerization (Virendra, 2004).

1.3.2 Light-polymerized poly-urethane based polymer

Light-polymerized denture base polymer was introduced to the dental profession with various applications in fixed and removable prosthodonthics, maxillofacial prosthetics, implantology, and orthodontics (Fellman, 1989). The first light-polymerized denture base polymer, marketed under trade name of Triad (Dentsply international,York, Pennsylvania), is similar to light-polymerized composite resin restorative material (Ogle et al.,1986).

A composite material is a product which consists of at least two distinct phases normally formed by blending together components having different structures and properties (Phillips, 1994). The purpose of this blending is to produce a material having properties that could not be achieved from any of the individual components alone. The two main components of composite restorative materials are the resin phase and the filler (McCabe, 1998). The beneficial properties contributed by the filler are rigidity, hardness, strength and high modulus of elasticity (McCabe, 1998).

The nature of the composite resin as a restorative material may alter slightly from one product to another, although, essentially, all composite resins contain a modified methacrylate or acrylate. One of the most commonly used is urethane dimethacrylate. Fillers which are commonly used include quartz, fused silica and many type of glass, including aluminosilicates and borosilicates (McCabe, 1998). There has been a trend towards the use of fillers with smaller particle size, such as microfine silica. The small particle size of the filler produces a massive increase in surface area for a given volume of filler.

Light activated materials are generally supplied as a single paste which contains monomers, co-monomers, filler and an initiator which is unstable in the presence of light. The use of UV-activated materials has diminished greatly because of the possible dangers of long-term exposure to ultraviolet radiation. The current composite resins are activated by light in the visible spectrum, in a range of 400-500 nm wavelength of blue visible intense collimated light (Al-Mulla et al., 1988). Before the material is exposed to the light source, only a minimal increase in viscosity takes place. After being exposed to the light source, polymerization is often very rapid. Sufficient intensity and wavelength of the light are required for the camphoroquinone activation responsible for initiating polymerization (Blankenau et al., 1983). Care must be taken in the storage of unused materials because exposure to sunlight or operating light may be sufficient to activate a slow initiation process that causes the material to thicken and become unworkable.

The pattern of polymerization is dictated by the fact that activation is first achieved in the surface layers of material where the light intensity is greatest (McCabe, 1998). Because a certain level of intensity is required to cause activation, it follows that light polymerized materials have a limited depth of cure. The depth of cure of composite resin has been reported to depend on the composition of the composite resin, light-source parameter, exposure time and the storage time (Sobrinho et al., 2000). The maximal intensity of the light radiation beam is at the surface of the resin. As the light penetrates into the resin it will lose intensity as a result of the scattering and reflection of light by the fillers (Craig and Powers, 1997).

The depth of polymerization has often been measured indirectly by the measurement of the hardness of the material at specific depths (Cook, 1980). There are several other methods that have been used to evaluate the depth of polymerization. The optical microscope could be used to determine the demarcation line between polymerized and unpolymerized resin in the composite specimens (Murray, 1981). The direct approach for evaluating the depth of polymerization is infra-red spectroscopy (IR). This method determines the percentage of the degree of conversion of carbon-carbon double-bonds into single bonds during the polymerization reaction (Ruyter and Øysaed, 1982).

Triad material (Dentsply, USA) is a visible light-polymerized denture base polymer that was initially introduced in the 70s. It was supplied in a sheet of 2 mm in thickness and rope form. The material is packaged in light-shielded envelopes to prevent premature polymerization (Alsawaf et al., 1991). The system was originally decided to eliminate the need for waxing, flasking, boiling out, packing, pressing, and heat-processing required for conventional denture construction (Khan et al., 1987). The material is thermoplastic; at room temperature it has the consistency of glazing putty and can be readily shaped by finger pressure and with instruments. When a lower viscosity is desired, it can be heated in its package in a water bath before opening. This procedure allows control of the material’s consistency for various application procedures, with acceptable flowability (Alsawaf et al., 1991).

There were many studies that examined the flexural strength of light-polymerized denture base materials. Ogle et al.,(1986) compared flexural strength of a light-polymerized denture base polymer, a conventional heat-polymerized polymer and an orthodontic chemically-polymerized acrylic resin. They found that light-polymerized denture base polymer recorded higher flexural strength compared with chemically-polymerized acrylic resin, and lower flexural strength compared with heat-polymerized resin. It was also observed that light-polymerized denture base polymer recorded higher modulus of elasticity (greater stiffness) compared with either heat-polymerized or chemically-polymerized acrylic resins (Ogle et al., Al- Mulla et al., 1988 and Lewinstein et al., 1995).

Khan et al., (1987) studied the staining characteristics, transverse strength and microhardness of visible light-polymerized denture base material (Triad). They reported that Triad material had superior transverse strength and microhardness characteristics compared with heat-polymerized acrylic resin.

Dixon and Breeding (1991) reported that incorporation of polyethylene fibers increased the mean flexural strengths of heat-polymerized polymer. On the other hand, Triad resin recorded significantly increased flexural strength after reinforcement with polyethylene fibers. If Triad denture resin were used to fabricate removable prosthesis, it would be beneficial to incorporate aesthetic polyethylene fiber into resin .

Various benefits that have been claimed when using light-polymerized denture base polymers can be listed as:

1. Ease of fabrication, manipulation, and availability (Alsawaf et al.,1991)
2. Time saving
3. Excellent handling characteristics and convenience (Polyzois,1990 and Barron et al.,1992)
4. Negligible exposure to organic solvents, minimizing the potential for allergic reactions (Barron et al., 1993)
5. Minimal laboratory equipment and laboratory time
6. Fewer patient appointments (Barron et al.,1992)

1.3.3 Chemically-polymerized denture base polymer

This material is often referred to as “cold-curing”, “self-curing”, or “autopolymerizing” resin. The composition of this material is similar to that of heat-polymerized materials, except that the liquid contains an activator, such as dimethyl-p-toluidine (McCabe,1998). The function of the activator is to react with the benzoyl peroxide (initiator) in the powder to create free radicals that can initiate polymerization of the monomer. The polymerization can be completed at room temperature and the degree of polymerization by using a chemical activator is not as high as that achieved by activation with heat (Phillips, 1991). The rate of polymerization is influenced by the particle size of the polymer. The smaller the particle size, the more rapid is the polymerization (Phillpis, 1991).

Processing of the resins could be conducted by compression molding in a flask, where initial hardening of the resin occurs within 30 minutes of flask closure. The working time for the chemically-polymerized materials is invariably shorter than for heat-polymerized materials (Phillips, 1991).

The resins could also be used to produce dentures by using a pour or fluid resin technique. The polymer powders formulated for this technique usually have very fine particles (Phillips, 1991). This fine particle size is neccessary to ensure fluidity of the mix. The dough is poured into the mold through an opening, or vent, and held in a pressure chamber at room temperature until polymerized (Phillips and Moore, 1994)

In comparison with heat-polymerized resins, chemically-polymerized resins have a higher residual monomer content (McCabe 1998). Polymerization in those resins is never as complete as in heat-polymerized resins. Residual monomer has the potential to cause irritation, imflammation, and an allergic response of the oral mucosa (Jorge et al., 2003). Chemically-polymerized denture base polymers were claimed to be the most cytotoxic of the denture base polymer. Water storage may reduce the level of residual monomer,resulting in decrease cytotoxicity of these acrylic resin denture base materials (Jorge et al., 2003).

Higher levels of residual monomer also can results in inferior mechanical properties and can compromise biocompatibility of the denture bases. Vallittu (1998) stated that increasing the polymerization temperature for chemically-polymerized polymer from 30˚ C to 60˚C decreased the residual monomer content of the polymer from an average of 4.6w% to 3.3w%. The material exhibited higher solubility and inferior color stability, because of oxidation of the amine accelerator (Phillips, 1991), and creep rates were usually high, especially under increased stresses.

1.4 Mechanical properties of the denture base polymer

Mechanical properties are defined by the law of mechanics, that is, the physical science that deals with energy and forces and their effects on bodies.

1.4.1 Microhardness

Microhardness is defined as the resistance to permanent surface indentation or penetration. It provides a possible indication of the abrasive resistance of the material. Denture base materials should have sufficient abrasion resistance to prevent excessive wear of the material by abrassive denture cleansers or foods.

Many methods of hardness measurements have been used, but the most realistic approach to assessment of the hardness of a material is by measurement of its resistance to indentation. The value of hardness, often referred to as the hardness number, depends on the method used for its evaluation. Generally, low values of hardness number indicate a soft material and a high value indicates a hard material (McCabe, 1998).

There are various methods used for hardness evaluation which include Vickers, Knoop, Brinell and Rockwell. Vickers and Knoop methods both involve the use of diamond pyramid indentors. In the case of Vickers hardness, the diamond pyramid has a square base, whilst for Knoop hardness, one axis of the diamond pyramid is much larger than the other (Fig.1.1). The Brinell hardness test involves the use of a steel ball indenter that produces an indentation of circular cross-section. The hardness is a function of the diameter of the circle for the Brinell hardness test, and for Vickers and Knoop hardness tests, it is the distance across the diagonal axes for Vickers and Knoop indenters. In the case of Rockwell hardness, a direct measurement of the depth of penetration of a conical diamond indenter is made.

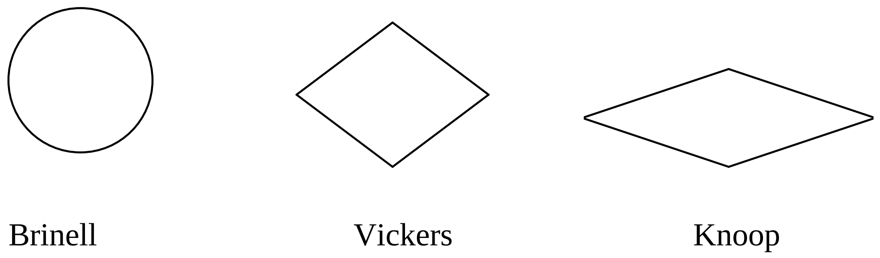

Figure 1.1 Shapes of indentations produced by three types of hardness test.

Measurements are normally made by using a microscope because the indentations are often too small to be seen with the naked eye. After the indenting force has been removed, the hardness value is related to the degree of permanent deformation produced in the surface of the test material by the indenter under a given load.

Fraunhofer and Suchatlampong (1975), reported that water produces softening of the surface of denture base material. This decrease in surface hardness indicates that water either combines with or, more probabaly, enters into the amorphous outer layer of the acrylic resin surface, which suggests a form of chemical reaction that cannot be prevented (Fraunhofer and Suchatlampong, 1975).

There were also studies that showed reduction in surface hardness of denture base polymer as a result of exposure to some disinfecting solutions such as chlorhexidine and glutaraldehyde (Asad et al., 1993). This reduction could be accounted for by the slow absorption of disinfecting chemicals into the resin that resulted in some structural change in the polymer (Asad et al., 1993). Harrison et al.,(1978) used hardness value when investigationg the abrasion resistance of denture base polymer with different amounts of cross-linking agent.

The use of a ultraviolet light-activated coating was believed to improve the surface hardness (Szabo, 1986) and at the same time save time by avoiding a final polishing of the denture base. With the exposure to ultraviolet light for 15 minutes, the glazed surface was harder than the unglazed surface.

1.4.2 Flexural strength and flexural modulus

Strength can be referred to as the ability of the prosthesis to resist applied forces (load) without fracture or excessive deformation (McCabe, 1998). It is important to consider strength of a denture base material before its use in a laboratory or clinical practice, to ensure that the prosthesis will serve its function safely, effectively and for a reasonable period.

Flexural strength, transverse strength or modulus of rupture, as this property is variously called, is essentially a strength test of a bar of material, supported at each end under a static load subjected to three-point flexure. This test is also referred to as a three-point transverse strength test. This test is particularly applicable to materials which are used for removable complete dentures, because a similar type of stress is applied to maxillary dentures during mastication. The test indicates the ability of the denture material to withstand this stress, and it also shows the amount of flexion of the denture that may occur. It is desirable that a denture should not alter its shape under biting force as this will damage the supporting bone and mucosa.

The transverse strength of acrylic resin depends on several factors, such as polymerization cycle, level of residual monomer, water sorption, amount of cross-linking, and internal porosity of the polymer (Harrison, & Huggett, 1992, Jerilomov et al., 1985, Takahashi et al., 1999).

Residual monomer resulting from incomplete conversion of monomers into polymers can adversely affect the transverse strength of denture base resin (Jerilimov et al., 1985 and Austin and Basker, 1982). This incomplete conversion could be a reason why chemically-polymerized denture base polymer has lower flexural strength compared with heat-polymerized denture base polymer (Ruyter and Svendsen, 1980). A polymerization cycle of 7 hours at 70˚C followed by a terminal boil is the optimal cycle resulting in decreased level of residual monomer (Harrison and Huggett, 1992). Short-cut polymerization cycles are undesirable and result in significantly raised levels of residual monomer. Rapid curing may be advantageous in saving time, but it will increase the possibility of generating gaseous porosity, thereby reducing the strength of denture bases.

Various cross-linking agents have been used as a component of PMMA denture base materials. The cross-linking agent ethyene glycol dimethacrylate (EGDMA), which provides a short cross-linking chain between adjoining linear molecules, is commonly added to currently available denture base materials (Hill, 1981). Caycik and Jagger,(1992) investigated the effect of cross-linking chain length on the flexural strength of a dough-molded PMMA resin and reported that increasing the chain length will decrease transverse strength. They also reported that the addition of the cross-linking agent, EGDMA, in concentrations from 0 to 60% produced no significant difference in the transverse strengths of a dough-molded PMMA polymer.

Polishing technique also was believed to affect the flexural strength of denture bases (Orsi and Andrade, 2004). Polishing may be accomplished either by conventional mechanical procedures, including the use of abrasive pastes with rotating felt cones and rag wheels, or by chemical procedures consisting of immersion in a monomer-based polishing fluid. The latter methods produced a coating on the resin that imparted a smooth shiny surface to the prosthesis. Chemical polishing resulted in significantly lower transverse strength values compared with mechanical polishing (Orsi and Andrade, 2004). The presence of the polishing layer on the surface of the material may act as a physical barrier that blocks the diffusion of the residual monomer. The residual monomer will act as a plasticizer (Dogan et al., 1995) that, when present in high levels, may negatively affect the mechanical properties of the resin.

Modulus of elasticity may be defined as the relative stiffness or rigidity of a material (McCabe,1998). The modulus of elasticity is a constant and is unaffected by the the amount of elastic or plastic stress that can be induced in the material. Thus, it is independent of the ductility of a material, and it is not a measure of its strength. Modulus of elasticity is expressed in gigapascals(GPa) or megapascals (Mpa).

Visible-light-polymerized polymer was found to have a higher elastic modulus (greater stiffness) compared with either heat-polymerized or chemically-polymerized denture base polymers (Ogle et al., 1986). Jagger and Harrison,(1999) reported that when the amount of PMMA fibres in acrylic resin was increased, there was a decrease in the modulus of rupture and a decrease in the modulus of elasticity. Memon et al. (2001) made a comparison of the flexural moduli of two PMMA-based polymers and they reported that heat-polymerized polymer was less stiff. This result can be explained by a relatively higher residual methyl-methacrylate monomer content.

The aim of this study was to investigate some properties of a light-polymerized urethane dimethacrylate (Eclipse), and to compare it with a conventional heat-polymerized and chemically-polymerized denture base polymer. The properties that were evaluated are surface hardness, flexural sttrength and flexural modulus.