Some flexural properties of a nylon denture base polymer

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SUMMARY Nylon denture base material could be a useful alternative to poly (methyl methacrylate) (PMMA) in special circumstances such as patient allergy to the monomer. The aim of this study was to evaluate the flexural properties of a nylon denture base material (Lucitone FRS), a conventional compression-moulded heat-polymerized (Meliodent), a compression-moulded microwave-polymerized (Acron MC) and an injection-moulded microwave-polymerized (Lucitone 199) PMMA polymers. The effect of aldehyde-free, oxygen releasing disinfectant solution (Perform®) on these properties was also investigated. The flexural modulus and the flexural strength were assessed with a three-point bending test. Specimens were stored in water at a temperature of 37 °C for 30 days. For each material, half of the prepared specimens were randomly selected and immersed in the disinfectant 24 h prior to testing. Results were compared statistically at a confidence level of 95%. The result showed that in both the control and disinfected groups, the flexural modulus of nylon was significantly lower than the three PMMA polymers. The flexural strength of nylon was significantly lower than those of Meliodent and Acron MC but was comparable with Lucitone 199. A 24-h immersion in the disinfecting solution increased the rigidity of nylon denture base material.

KEYWORDS: disinfectant, flexural modulus, flexural strength, microwave polymerization, nylon, poly (methyl methacrylate)

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Introduction

The most commonly used denture base resins are based on a monomer, methyl methacrylate, and a polymer, poly (methyl methacrylate) (PMMA), system. Polymerization of heat-cured PMMA is normally carried out in a temperature-controlled water bath for several hours. The use of microwave energy to polymerize PMMA, as first reported by Nishii (1), makes it possible to process acrylic dentures in a very short time (2–4). Some investigators have evaluated the physical properties (5–11) and others have examined the porosity (12–15) and the dimensional accuracy of microwave polymerized PMMA denture base polymers (16, 17).

It is considered relatively easy to process acrylic denture using the conventional compression moulding technique. However, there are many factors in the laboratory procedure that can lead to alteration to the denture teeth occlusion. To overcome the problems of significant increased vertical dimension after processing associated with the technique, injection moulding was developed. Strohaver (18) and Nogueira et al. (19) reported a significant smaller incisal pin opening for complete denture when processed in the injection system, when compared with the compression moulding technique. No appreciable difference in the laboratory working time between the two techniques was found (19). However, the injection moulding technique employed for denture construction saves time because of smaller incisal pin opening and hence less time is spent on post-processing occlusal adjustments.

Acrylic denture has outstanding resistance to the oral environment and to most solvents and UV radiation. However, there is a small risk of toxicity and hypersensitivity to the material as a result of its oxidation products and other components present in the system. Tissue reactions to acrylic resin denture base materials have been reported (20–25). Some potential alternative materials to PMMA used in such cases are such as polycarbonate and nylon.
Nylon is a generic name for certain types of thermoplastic polymers belonging to the class known as polyamides. These polyamides are produced by the condensation reactions between a diamine and a dibasic acid. The use of nylon as a denture base material has been described in the literatures in the 1950s (26–28). Although nylon was not recommended for general use at that time, it was used in special circumstances such as repeated denture fracture (27, 29) and for the construction of orthodontic appliance (30). Some of the disadvantages reported in the early form of nylon included the tendency of the base colour of the material to deteriorate, stain, high water sorption and the development of a rough surface after a short period of time. The inherent flexibility of nylon was later improved and the stiffness increased by the use of short glass fibre reinforcement (31, 32).

To date, no research has assessed the lately developed nylon-based flexible resin system for denture base construction. Lucitone FRS is a flexible and monomer-free thermoplastic dental polymer. The manufacturer recommends its use for the fabrication of temporary partial dentures or small to medium-size full removable dentures as well as occlusal splints and night-guards. The finishing can be performed using normal procedure for PMMA denture base. The manufacturer claims that it could provide comfort to the patient, as it is lightweight. The manufacturer provides a warranty against breakages for 5 years when fabricated in accordance with the instructions.

The use of effective infection control procedures in the dental surgery and laboratory is designed to prevent cross-infection between dentist, dental surgery staffs, dental technicians and the patients. Both the dentist and dental technician are at risk of contracting infections when adjusting or repairing prostheses that have not been disinfected. The use of chlorhexidine scrub followed by a 3-min contact time with chlorine dioxide disinfection solution was shown to be effective in disinfecting contaminated surfaces of acrylic resin dentures (33). Lin et al. (34) investigated the effectiveness of this solution and found that a short period of disinfection reduced but did not eliminate viable microorganisms on the prostheses.

Dentures should be carefully cleaned and disinfected after the adjustment procedure. The British Dental Association Advisory Service (35) advocates sodium hypochlorite at a concentration of 10 000 parts per million available chlorine, for disinfections of prostheses. However, sodium hypochloride is known to stain (36) or whiten plastic components of the prostheses (37, 38). A recent addition to the recommended disinfection materials for prosthetic appliances is a preparation based on active oxygen (Perform®). The main compositions of Perform® are potassium peroxomonosulphate, sodium benzoate and tartaric acid. It is supplied in a sealed sachet in the form of granules and upon mixing with water it remains effective for up to 30 h. The manufacturer states that there are no aldehyde and chlorine components in the solution. It is also claimed to be biodegradable with excellent disinfectant properties. The effectiveness of the disinfectant against certain bacteria has been investigated (39). No studies have evaluated the effect of this disinfectant on the flexural properties on denture base polymers. The flexural three-point bending test is useful in comparing denture base materials as it simulates the type of stress that is applied to the denture during mastication (40), although fatigue properties are clinically more relevant (41). The bending test not only determines the strength but it also gives an indication of the rigidity of the material.

The aim of this study was to compare some flexural properties of a nylon denture base material to other PMMA-based denture polymers with various methods of processing and polymerization modes. The effect of a 24-h immersion in the oxygen-releasing disinfectant solution (Perform®) on these properties was also investigated.

**Materials and methods**

The denture base materials investigated in this study are presented in Table 1. Nylon denture base was compared with some conventional PMMA polymers. Gypsum moulds were prepared by placing master perspex blanks measuring $68 \times 50 \times 4$ mm into their respective flasks. A powder:liquid ratio of 100 g of stone to 30 mL of water was used to prepare the moulds for all the specimens. After it hardened, the perspex blanks were removed and a separating agent was applied to the mould.

The conventional compression moulding technique using metal flasks was employed to prepare Meliodent specimens while Acron MC specimens were prepared in fibre-reinforced plastic flasks.

*Schulke & Mayr GmbH, Norderstedt, Germany.

Specimens for Lucitone 199 and Lucitone FRS were processed using an injection system (Microbase™ Injection System†) provided by the manufacturer. In this system, the high pressure of the injection process accurately regulates the inflowing amount of material into the closed dental flask. The powder and liquid of Lucitone 199 was mixed in the conventional manner and a plastic injection cartridge was used to carry the mixture for injection. The specially designed fibre-reinforced plastic flask has an opening that permits the mixture to flow into it through sprue channels. The injection process was carried out in the injection unit for 1 min at a pressure of 2 bars. The flask was left on the bench for 15 min and then transferred into the microwave oven for polymerization at 400 W for 15 min. Pressure was maintained within the flask system during polymerization by the metal repressing device.

Nylon material was supplied as a single component in a cartridge form. As the nylon was being melted in a furnace, which had been pre-heated to a temperature of 302 °C, the stone mould was exposed under the heat lamps. The mould was uniformly heated for 17 min to a temperature between 65 and 70 °C. The flask halves were assembled with brackets and together with the cartridge containing melted nylon; they were placed on to the injection unit. The injection moulding pressure was maintained at a pressure of 5 bars for 1 min and immediately after that, the assembly was removed and disengaged. The dental flask was bench-cooled for 5 min before deflasking. The blanks were removed from the moulds and the sprues were removed with a cut-off disc.

Table 1. Denture base polymers tested in this study

<table>
<thead>
<tr>
<th>Material</th>
<th>Type</th>
<th>Powder: liquid ratio</th>
<th>Processing method employed</th>
<th>Manufacturer</th>
</tr>
</thead>
<tbody>
<tr>
<td>Meliodent PMMA</td>
<td>23-4 g:10 mL</td>
<td>Compression-moulded technique; heat-polymerized for 7 h at 70 °C and 1 h at 100 °C</td>
<td>Bayer Dental, Newbury, UK</td>
<td></td>
</tr>
<tr>
<td>Acron MC PMMA</td>
<td>30 cm³:9 mL</td>
<td>Compression-moulded technique; microwave-polymerized at 500 W for 3 min</td>
<td>GC International Corp., Tokyo, Japan</td>
<td></td>
</tr>
<tr>
<td>Lucitone 199 PMMA</td>
<td>21 g:10 mL</td>
<td>Injection-moulded technique using a pressure of 2 bar; microwave-polymerized at 400 W for 15 min</td>
<td>Dentsply/DeTrey, Dreieich, Germany</td>
<td></td>
</tr>
<tr>
<td>Lucitone FRS Nylon</td>
<td>Single component</td>
<td>Injection-moulded technique using a pressure of 5 bar; pre-heat in furnace to 302 °C for 15 min</td>
<td>Dentsply/Trubyte, York, PA, USA</td>
<td></td>
</tr>
</tbody>
</table>

PMMA, poly (methyl methacrylate).

All specimens were cut and finished to the dimensions of 64 × 10 × 2.5 mm as specified in the International Standard Organization (42) for the testing of denture base materials. The surfaces of the specimens were wet polished on polishing machine (Struer Rotopole‡) with silicon carbide paper discs of 600 and 1000 grit sizes. Ten specimens were prepared for each material and they were stored in distilled water in an incubator at the temperature of 37 °C for 30 days. A longer duration of immersion than that specified by the International Standards Organization (42) was selected because of the variability in the time needed for various denture polymers to reach their equilibrium flexural properties (43). Major changes in the flexural properties of most polymers occurred within the first month of storage in water (44). For each material, half of the specimens were randomly selected and transferred for immersion in a disinfecting solution (Perform®) 24 h before testing. Forty grams of Perform® powder was added to 8 L of water to give the solution a 2% concentration. Disinfected specimens were rinsed thoroughly with distilled water prior to testing. The other half of the specimens tested as controls was immersed in water during the same disinfection time.

A flexural three-point bending test was carried out in a water bath at 37 °C, on an Instron testing machine (Instron Universal Testing machine, model 4466§). The dimensions of each specimen were entered into the program for computation. The distance between the two supporting wedges was 50 mm apart and the crosshead speed was set at 5 mm min⁻¹. The flexural modulus (E) was determined by calibrating the

†Dentsply/DeTrey, Dreieich, Germany.
‡Struers, Copenhagen, Denmark.
§Instron Inc., High Wycombe, UK.
machine and the values automatically computed from the equation: \( E = FL^3/4ybd^3 \), where \( y \) is the deflection corresponding to load \( F \) at a point in the straight-line portion of the load–deflection curve, \( L \) is the length between the jigs, \( b \) is the width and \( d \) is the thickness of the specimen. The flexural strength (\( S \)) was calculated using the equation: \( S = 3PL/2bd^2 \), where \( P \) is the load at fracture and in the case of nylon, it was the maximum load recorded from the load–deflection curve.

Statistical analysis was made using a one-way analysis of variance, the Scheffe test for post hoc comparisons, and the \( t \)-test at a significant level of \( P = 0.05 \).

### Results

The mean flexural modulus values, standard deviations and coefficient of variations for the denture base materials, of control and disinfected groups were shown in Table 2. In both groups, there was a significant difference in the flexural modulus value between nylon and the three PMMA polymers (\( P < 0.05 \)). Individual \( t \)-test result showed that the flexural modulus was not significantly different between the control and disinfected groups except for nylon materials (Table 3).

The mean flexural strength, standard deviation and coefficient of variations for the denture base materials, of control and disinfected groups were shown in Table 4. Nylon specimens in both groups deflected beyond the capacity of the transverse test machine and came off the rollers of the jig. Flexural yield strength value was taken because the specimens did not break during the test. All specimens of the other three denture base materials fractured during the test. For both the control and disinfected groups, there was a statistically significant difference in the flexural strength values between nylon and the two compression-moulded PMMA polymers (Meliodent and Acron Mc) (\( P < 0.05 \)). No significant difference was observed between nylon polymer and the injection-moulded

### Table 2. Flexural modulus (MPa) of denture base materials

<table>
<thead>
<tr>
<th>Materials</th>
<th>Control</th>
<th></th>
<th></th>
<th>Disinfected</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (MPa)</td>
<td>SD</td>
<td>CoV</td>
<td>Mean (MPa)</td>
<td>SD</td>
<td>CoV</td>
</tr>
<tr>
<td>Acron MC</td>
<td>362.6±3</td>
<td>362.1±3</td>
<td>10±3</td>
<td>3363±9</td>
<td>362.1±3</td>
<td>10±8</td>
</tr>
<tr>
<td>Lucitone 199</td>
<td>3197±9</td>
<td>192±5</td>
<td>6±0</td>
<td>3214.7</td>
<td>107±2</td>
<td>3±3</td>
</tr>
<tr>
<td>Meliodent</td>
<td>3159±3</td>
<td>152±3</td>
<td>8±9</td>
<td>3196±3</td>
<td>275±7</td>
<td>8±9</td>
</tr>
<tr>
<td>Lucitone FRS</td>
<td>1714±4</td>
<td>152±3</td>
<td>8±9</td>
<td>1936±8</td>
<td>780±4</td>
<td>4±0</td>
</tr>
</tbody>
</table>

SD, standard deviation; CoV, coefficient of variation. A one-way analysis of variance demonstrated that there was a significant difference among the materials (\( P < 0.05 \)). The vertical tie bars indicate values, which are not significantly different from one another according to Scheffe test.

### Table 3. Student’s \( t \)-test comparing the mean flexural modulus between control and disinfected groups of denture base polymers

<table>
<thead>
<tr>
<th>Material</th>
<th>( t )-value</th>
<th>( P )-value</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acron MC</td>
<td>-1.188</td>
<td>0.301</td>
<td>NS</td>
</tr>
<tr>
<td>Lucitone 199</td>
<td>0.204</td>
<td>0.848</td>
<td>NS</td>
</tr>
<tr>
<td>Meliodent</td>
<td>-0.434</td>
<td>0.686</td>
<td>NS</td>
</tr>
<tr>
<td>Lucitone FRS</td>
<td>2.993</td>
<td>0.040</td>
<td>S</td>
</tr>
</tbody>
</table>

NS, not statistically significant; S, statistically significant at \( P = 0.05 \).

### Table 4. Flexural strength (MPa) of denture base materials

<table>
<thead>
<tr>
<th>Materials</th>
<th>Control</th>
<th></th>
<th></th>
<th>Disinfected</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (MPa)</td>
<td>SD</td>
<td>CoV</td>
<td>Mean (MPa)</td>
<td>SD</td>
<td>CoV</td>
</tr>
<tr>
<td>Meliodent</td>
<td>70.5±5</td>
<td>5.7±1</td>
<td>8.1±1</td>
<td>71.3±3</td>
<td>2.7±1</td>
<td>3.8±1</td>
</tr>
<tr>
<td>Acron MC</td>
<td>65.9±9</td>
<td>5.2±1</td>
<td>7.9±1</td>
<td>64.3±1</td>
<td>6.6±1</td>
<td>10.2±1</td>
</tr>
<tr>
<td>Lucitone 199</td>
<td>63.7±3</td>
<td>3.3±1</td>
<td>5.2±1</td>
<td>61.7±3</td>
<td>3.3±1</td>
<td>5.3±1</td>
</tr>
<tr>
<td>Lucitone FRS</td>
<td>55.3±3*</td>
<td>3.0±1</td>
<td>5.5±1</td>
<td>53.7±3</td>
<td>3.3±1</td>
<td>6.1±1</td>
</tr>
</tbody>
</table>

SD, standard deviation; CoV, coefficient of variation. A one-way analysis of variance demonstrated that there was a significant difference among the materials (\( P < 0.05 \)). The vertical tie bars indicate values, which are not significantly different from one another according to Scheffe test.

*Flexural yield strength.

### Table 5. Student’s \( t \)-test comparing the flexural strength between control and disinfected groups of materials

<table>
<thead>
<tr>
<th>Material</th>
<th>( t )-value</th>
<th>( P )-value</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Meliodent</td>
<td>0.216</td>
<td>0.840</td>
<td>NS</td>
</tr>
<tr>
<td>Acron Mc</td>
<td>-0.432</td>
<td>0.688</td>
<td>NS</td>
</tr>
<tr>
<td>Lucitone 199</td>
<td>-1.050</td>
<td>0.353</td>
<td>NS</td>
</tr>
<tr>
<td>Lucitone FRS</td>
<td>-0.715</td>
<td>0.514</td>
<td>NS</td>
</tr>
</tbody>
</table>

NS, not statistically significance at \( P = 0.05 \).
microwave-polymerized PMMA polymer (Lucitone 199), and between the latter and the compression-moulded microwave-polymerized PMMA polymer (Acron MC). For all the materials, individual $t$-test showed that the flexural strength values between the control and disinfected groups was not significantly different (Table 5).

**Discussion**

This study compares the flexural strength and the flexural modulus of a range of denture base materials including a nylon polymer. Nylon exhibited the lowest flexural modulus of 1714 MPa when not disinfected while the disinfected specimens had a value of 1937 MPa. The low flexural modulus exhibited by nylon means that it is less rigid than the conventional PMMA polymers. This is not surprising as it is indicated in certain clinical situations where flexibility is desired. Nylon is promoted as a denture base material on the basis of its flexibility, which allows it to engage certain degree of undercuts for retention without these undercuts being blocked. At the same time nylon has also been claimed to be a good alternative in denture patients who have sensitivity or allergy to methyl methacrylate monomer (32).

The result of this study was in agreement with those of Stafford et al. (45) and MacGregor et al. (32), in that nylon was found to be more flexible than PMMA denture base polymers. At water saturation, Hargreaves (31) quoted the flexural modulus of nylon to be 20% lower than those of PMMA polymers and this figure was slightly low than that observed in our study.

In term of the flexural strength, the two compression-moulded PMMA-based polymers (Acron MC and Meliodent) exhibited comparable strength regardless of the different polymerization modes employed. This finding was in agreement with the studies of Memon et al. (11) and Alkhatib et al. (13), where they did not establish any significant difference in strength between the compression-moulded, heat-polymerized PMMA and specially designed, microwave-polymerized PMMA polymers. In this study, nylon was shown to have a lower flexural strength than the two compression-moulded PMMA polymers, which was in agreement with a previous report (46).

Lucitone 199 exhibited lower flexural strength than Meliodent. Both are conventional heat-polymerized PMMA polymers. However in this study, the former was processed using the injection-moulding technique with microwave polymerization. Various curing cycles had been suggested for microwave curing of conventional PMMA but there were always problems associated with porosity especially in thicker areas of the specimens (5, 7, 8, 13, 15). Increased formation of porosity as a result of microwave irradiation could have been a possible explanation for the observed inferior strength of Lucitone 199. This is further supported by previous studies where a reduced flexural strength was associated with an increase in the porosity (47, 48). The results of the study suggest that where strength was concerned, specifically designed acrylic resins for microwave polymerization should be used when processing dentures using microwave.

As there can be cross-contamination between dental surgery and the laboratory, dental prostheses should be disinfected before being sent to the laboratory for repairs and additions. It has also been recommended that the work that comes from the laboratory be disinfected before it is placed in the patient’s mouth. Many disinfectants solutions have been tested for their effect on the flexural properties of denture base polymers. In the present study, the oxygen-based disinfectant (Perform®) was chosen because it is aldehyde-free and its effectiveness to eliminate certain bacteria (39). It was found that a 24-h immersion in the solution did not have any significant effect on the flexural properties of any of the PMMA denture base polymers. The manufacturer recommended an immersion time of 10 min at a concentration of 2%. However, no reference was available as to the effect of this protocol on the mechanical properties of the denture base materials. Prolonged immersion for 24 h as performed in this investigation provided a safety margin as disinfections of prostheses are sometimes carried out for longer periods of time than those recommended. This often occurs when there is a delay in the prostheses being transported to the laboratory or when the dental technician inadvertently leaves them immersed overnight before adjustment. To date, no studies have investigated the effect of an oxygen-releasing disinfectant on denture base materials therefore no direct comparison can be made. However, Shen et al. (49), Asad et al. (50) and Polyzois et al. (51) found that the flexural properties of PMMA was not affected by long-term immersion in a number of other disinfectants available in the market.
Nylon demonstrated a slight increase in the modulus after 24-h immersion in the disinfectant, with no effect, however, in its strength. The change in the rigidity may defeat the intended purpose of selecting the material for that particular clinical situation. As a longer immersion than that recommended was employed in this study, further investigation needs to be conducted to determine the maximum length of immersion time that will not affect the rigidity of nylon.

**Conclusion**

Nylon exhibited a significantly lower flexural modulus than the three PMMA polymers. The flexural strength of nylon was significantly lower compared with both the compression-moulded, heat- and microwave-polymerized PMMA polymers. The flexural strength of nylon was comparable with the injection-moulded, microwave-polymerized PMMA denture base material.

A 24-h immersion in an aldehyde-free, oxygen-based disinfecting solution (Perform®) did not affect the flexural strengths and flexural moduli of all the PMMA polymers. There was a significant increase in the flexural modulus of nylon when treated with the disinfectant but no change was observed in its flexural strength.

**References**


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