Comparison of the Dimensional Accuracy of Injection-Molded Denture Base Materials to that of Conventional Pressure-Pack Acrylic Resin

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Purpose: **This study compared the linear dimensional changes of 3 injection-molded denture base materials to that of conventionally processed polymethylmethacrylate (PMMA) resin.**

Materials and Methods: **An impression of an aluminum maxillary edentulous arch was made with a condensation silicone impression material (Denture Elasticon) to fabricate a gypsum master cast that was replicated as a silicone mold. A maxillary complete denture with acrylic teeth was waxed to full contour on the master cast and replicated to make 40 wax dentures. ERA attachments cast in metal (Rexillium) with indices milled into the centers were waxed into 3 positions in each denture for recording dimensional measurements of the wax denture. Ten dentures were allocated to each of 4 groups; Group 1 was processed using conventionally processed PMMA (Microlon), Group 2 used injection-molded PMMA (SR-Ivocap), Group 3 employed injection-molded nylon (Valplast), and Group 4 used injection-molded styrene (Northern). All processed specimens were stored at room temperature (25°C, ambient humidity) for 1 week (while still on the master cast) before anteroposterior and cross-arch measurements were made using the ERA reference points with a digital caliper. After separation from the master cast and following water storage at 37°C for 7 days additional measurements were made.**

Results: **An analysis of the results showed that the effect of processing was not the same for the 3 dimensions studied, regardless of which dimension was considered ($p < 0.0001$). The pattern of dimensional changes associated with the material type was not the same between the wax and processing stages as it was for the change between the processing and decasting stages ($p < 0.0001$).**

Conclusions: **Processing the denture base materials produced unequal deformation in different dimensions (anterior-posterior and cross-arch). Each material tested also responded differently to the processing stages.**

INDEX WORDS: acrylic resins, denture bases, complete dentures, dental prosthesis, dental materials

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Introduction

Polymethylmethacrylate (PMMA) has become the most commonly used material for denture bases since its introduction in 1937.[1] The advantages of PMMA include excellent esthetic properties, adequate strength, low water sorption, low solubility, lack of toxicity, facility of repair, and construction by a simple molding and processing technique. However, despite its widespread popularity, the inaccuracies inherent in the use of PMMA as a denture base material include dimensional change during processing, frequently due to polymerization shrinkage.[1]

Polymerization shrinkage has 2 main effects. The shrinkage distorts the palate of a maxillary denture resulting in an inaccurate fit to the supporting tissues,[2] and it affects the position of the teeth on maxillary and mandibular dentures, and thus the final occlusion of the dentures.[3]

Woelfel and Paffenbarger[4] have shown that the greatest distortion occurs in the cross-arch region when the denture is deflasked. This has been attributed to the release of internal stresses developed during processing, and the difference in the coefficient of thermal expansion between stone cast and acrylic resin.

Acrylic resins have been modified to improve not only their physical and chemical properties but also their working properties to aid the laboratory in the processing of complete dentures. One example is the introduction of injection molding. Injection molding allows directional control of the polymerization process through the flask design. A constant flow of new material from the sprue compensates for the polymerization shrinkage.

Various injection-molded denture base materials and processing techniques are now available, each claiming to produce more accurate denture bases. The SR-Ivocap system (Ivoclar AG, Schaan, Liechtenstein) is an injectable PMMA. There have been various studies carried out on the material, and the literature in general supports the claim that the SR-lvocap system has less linear dimensional change than conventional PMMA.[5,6]

Other materials used for denture bases include nylon and styrene. Nylon was developed as a result of the classic research by Carruthers and associates of the Du Pont Chemical Co.[7] It was initially studied as a denture base material in the 1950s.[7,8] The material had many problems, such as warpage, water sorption, discoloration, surface roughness, bacterial contamination, and difficulty in polishing. There was a scarcity of research on dental applications for nylon until 1971, when Hargreaves[9] evaluated a different polymer of nylon and developed guidelines for optimum properties of nylon fabricated for dental use. MacGregor et al[10] compared the dimensional changes of nylon-12 to injection-molded PMMA and conventionally processed PMMA and the changes found were clinically insignificant.

Styrene was also studied to some extent in the 1950s,[11] but very little research has been conducted on it as a denture base material in recent years.

Recently, other nylons and styrenes have become available to the dental profession, but there are no studies available to support the manufacturers' claims that these newer materials are more dimensionally accurate than conventionally processed PMMA. The purpose of this study was to compare the linear dimensional accuracy of 3 chemically different injection-molded denture base materials to that of conventional pressure-pack acrylic resin.

Materials and Methods

Conventional pressure-packed PMMA (Microlon, Dentsply, York, PA, USA) was compared to injection-molded base materials including: PMMA (SR-Ivocap, Ivoclar A.G. Schaan, Liechtenstein), nylon (Valplast, Valplast Int. Corp., NY, USA), and styrene (Northern, Rapid Injection Systems, NY, USA).

For the fabrication of the test specimens, an aluminum master cast (55U, Columbia Dental Corporation, NY, USA) simulating a maxillary edentulous arch with a relatively flat palate was used. This cast was chosen based on the published findings that a flatter palate produced larger openings in the posterior palatal seal area when evaluating different processing techniques.[12] An impression of the master cast was made using a condensation silicone impression material (Denture Elasticon, Kerr Manufacturing Co., Romulus, MI, USA) in a custom tray (Triad, Dentsply International, York, PA, USA). The impression was then boxed and poured in type III gypsum (Microstone, Whip Mix Corp, Louisville, KY, USA) using a ratio of 100 g gypsum to 30 ml water as recommended by the manufacturer. Once the stone was set, the gypsum master cast was recovered from the impression and trimmed; 3 notches were carved into the land area, one in the anterior, and one on each posterior side. The notches served as reference points for accurate repositioning of the

master cast in an impression mold for the duplication of the wax dentures.

A silicone mold (Perma-Flex Mold Co., Columbus, OH, USA) of the gypsum master cast was fabricated. The master cast mold was used to make 40 duplicate master casts in type III gypsum on which the complete maxillary dentures could be waxed and processed.

The casts were numbered and randomly assigned to the 4 test groups. The reference points for the linear measurements consisted of ERA attachments (Sterngold, Attleboro, MA, USA) cast in a base metal alloy (Rexillium, Generic Pentron, USA). A depression was machined in the center of each attachment (Anderson Precision Machining, Inc., Iowa City, IA, USA), to intimately fit the metal measuring point present on digital calipers (Mitutoyo Corporation, Japan).

To fabricate the initial denture, 2 layers of base plate wax were adapted on a gypsum master cast. Anterior prosthetic teeth (Dentsply International, York, PA, USA) were arranged to basic esthetic guidelines, monoplane posterior denture teeth (Dentsply International, York, PA, USA) were arranged on a flat plane, and the denture was completely waxed and festooned. Prosthetic teeth were included in the processed resins, because when no teeth are present in a heat-cured denture base, noticeably less distortion occurs.[13] Additionally, the use of prosthetic denture teeth more closely simulated the patient's clinical situation. After completing the waxed denture, 3 prepared ERA attachments were waxed at positions A, B, and C, (Fig 1). A silicone putty mold of the completed denture was then made. This denture replication technique, previously discussed by Lindquist et al,[14] provided uniform vertical positioning of the prostheses, giving the test dentures uniform thickness, uniform tooth position, and similar ERA attachment locations.

Using the silicone mold, 40 maxillary wax dentures with teeth were fabricated on the previously prepared and indexed casts. The completed wax dentures were stored at room temperature (25°C, ambient humidity) for 1 week on the casts, for stress relief within the wax. ERA attachments were added by making a small depression in the A, B, and C positions; they were secured with base plate wax. All specimens were festooned, and measurements were made at the wax stage, prior to proceeding to denture base processing.

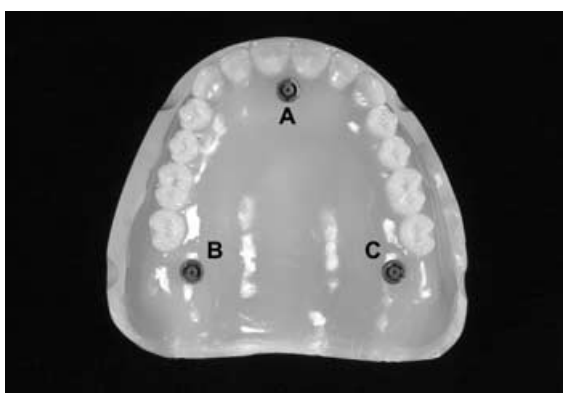

Figure 1. The wax denture was completed and 3 prepared ERA attachments were waxed at positions A, B, and C.

The conventional PMMA specimens were fabricated using a conventional flasking and pressure-pack technique and a 9-hour water bath processing at 74°C. For the PMMA injection-molded resin, the dentures were flasked according to the manufacturer's instructions using the Ivocap flask. Premeasured capsules of resin and monomer (20 g polymer, 30 ml monomer) were triturated for 5 minutes before injecting into the flask. Hydraulic pressure was maintained for 5 minutes before placing the assembly into boiling water (100°C) for 35 minutes. The assembly was then removed and placed in cold water for 20 minutes before deflasking the denture.

For the nylon specimens the dentures were flasked according to the manufacturer's instructions with a bifurcated sprue positioned at the posterior edge of the palate. After the boil out, a cylinder containing the nylon was plasticized for 11 minutes at 550°F before injecting into the flask. The levers of the press were turned rapidly to apply firm pressure until the springs of the press were fully compressed. The pressure was maintained for 3 minutes. The pressure was then relieved and the flask was allowed to bench cool for at least 20 minutes before opening.

The styrene-waxed denture specimens were sent to a commercial laboratory (Lafayette Dental Laboratory, Lafayette, IN) that was equipped to process the dentures utilizing this material. A commercial laboratory was utilized since the special processing equipment, flasks, and infrared heating ovens were not available at our facility. The processing technique was the same as that for the nylon with the exception that the flasks were heated in an infrared oven for 25 minutes prior to injecting the material.

Three dimensions (A–B, B–C, and C–A) were measured on the denture base for the evaluation of linear dimensional change (Fig 1). Each dimension was measured 3 times by one of the authors (AP) at each protocol interval using a digital caliper with a resolution of 0.01 $\pm$ 0.02 mm, and the average of 3 measurements were recorded as the distance between the points.

Measurements were made at the wax stage immediately before investing, after processing on the master cast (24 hours after breakout), and after decasting and storage in water at 37°C for 1 week (Council on Dental Materials and Devices, ADA Specifications, No. 12). The measurements at the wax stage were used as the baseline readings,

and all values were calculated with these measurements as the starting point.

Changes in the measured distances were calculated for the 3 dimensions (A–B, B–C, and C–A) for each material between the 3 pairs of stages: comparison of the mean measure at the processing stage versus wax stage, comparison of mean measurement at the decasting stage versus processing stage, and overall change, i.e., comparison of mean measurements at the decasting stage versus wax stage. In each case, the difference taken was between the measurement of the later stage and the measurement at the earlier stage, so that a negative change score reflects a diminution of the measured dimension, representing linear shrinkage, and a positive change score represents expansion.

Multivariate linear models were used to assess the changes in the measured dimensions (A–B, B–C, and C–A), while taking into account the correlated nature of this measurement and the 3 pairs of stages. The effect of the material on dimensional change was assessed using Wilks' lambda and the associated F test statistic derived from Wilks' lambda. If the overall multivariate test for the 3 dimensions was found to be significant, comparisons between the pairs of materials for each dimensional measure were accomplished using the Tukey adjustment for multiple comparisons, in conjunction with an overall 5% level of type 1 error. The Wilks' criterion and the associated F test were also used to assess effects of time (uniformity of change from wax to processing stages, and change from processing to decasting stages), of any differences in change among the 3 dimensions measured, and of the interaction between time and material effects.

Results

Descriptive statistics for the changes in measured distances for the 3 dimensions for each material between the 3 pairs of stages can be seen in Table 1.

Statistical evaluation of the results of the change from the wax to the processing stage yielded a Wilks' lambda = 0.084, $p < 0.0001$, indicating evidence of a multivariate effect, and suggesting that there were differences between the types of denture base materials (Fig 2). The analysis of change for each dimension separately by univariate analysis resulted in change at each of the 3 dimensions, and each was highly significant ($p < 0.0001$ for each of the dimensions). The injection-molded PMMA had smaller decremental change in the A–B and C–A dimension (mean of –0.001 and 0.031, respectively) than conventionally processed PMMA, styrene, or nylon (mean changes ranging from –0.148 to –0.249). For the B–C dimension, mean changes did not differ for injection-molded PMMA and nylon materials (0.086 and 0.030), and both were positive, indicating a small increase in mean from the wax to the processing stage.

The change from the processing to the deflasking stage yielded a Wilks' lambda = 0.016, $p < 0.0001$, suggesting a difference in denture base materials based upon multivariate assessment (Fig 3). Univariate analysis results were significant for all 3 dimensions ($p < 0.0001$ for A–B and A–C, $p = 0.016$ for B–C). In the A–B dimension, styrene showed a small positive (expansion) change (0.073), whereas the other mean changes were negative and ranged from –0.055 to –0.126, indicating shrinkage. For the C–A dimension, the mean change for styrene was positive in nature and the smallest in magnitude (0.036). In the B–C dimension all 4 materials showed

Table 1. Descriptive Statistics for Changes Between Stages in Denture Base Distances in mm for Three Dimensions and Four Materials

	Processing-Wax			*Deflasking-Processing*			*Deflasking-Wax*		
Material	*A–B*	*B–C*	*C–A*	*A–B*	*B–C*	*C–A*	*A–B*	*B–C*	*C–A*
Microlon (N = 10)	−0.148 (0.081)	−0.119 (0.079)	−0.181 (0.083)	−0.055 (0.016)	−0.175 (0.060)	−0.158 (0.314)	−0.204 (0.073)	−0.294 (0.075)	−0.339 (0.300)
Nylon (N = 10)	−0.249 (0.107)	0.030 (0.048)	−0.212 (0.023)	−0.126 (0.099)	−0.849 (0.080)	−0.189 (0.029)	−0.374 (0.053)	−0.819 (0.089)	−0.402 (0.035)
SR-Ivocap (N = 10)	−0.001 (0.123)	0.086 (0.057)	−0.031 (0.063)	−0.104 (0.053)	−0.299 (0.063)	−0.114 (0.035)	−0.106 (0.107)	−0.213 (0.059)	−0.144 (0.048)
Styrene (N = 10)	−0.181 (0.038)	−0.111 (0.037)	−0.150 (0.036)	0.073 (0.031)	−0.010 (0.043)	0.036 (0.025)	−0.108 (0.033)	−0.121 (0.047)	−0.114 (0.046)

The values represent means (standard deviation values are given in parentheses).

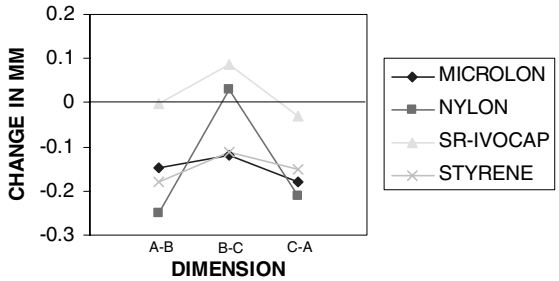

Figure 2. Mean change between wax and processing stages by dimension for each material.

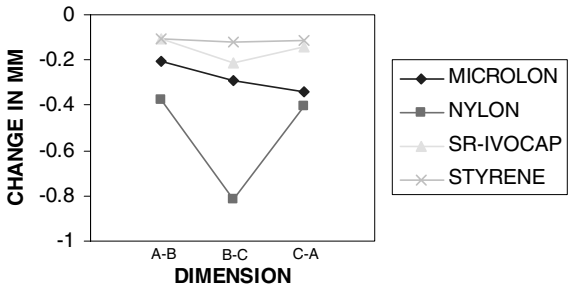

Figure 4. Mean change between wax and deflasking stages by dimension for each material.

shrinkage, the smallest mean shrinkage was associated with styrene (–0.010), and the largest with nylon (−0.849).

Statistical analysis of the results of the overall change from wax to decasting stage yielded a Wilks' lambda = 0.038, $p < 0.0001$, indicating that the type of denture base material affected dimensional measurements based upon multivariate assessment of the overall change scores for all 3 dimensions (Fig 4). There was evidence for material effects in the A–B and B–C dimension ($p < 0.0001$) and in C–A dimension ($p = 0.0002$). In each dimension (A–B, B–C, and C–A) shrinkage occurred with each material.

There was evidence of interaction among the factors in this experiment ($p < 0.0001$). The effect of transitions through fabrication stages was not similar in the 3 dimensions studied, regardless of which change measure was measured. Also, the pattern of deformation associated with material type was not the same for the change between the wax and processing stages as it was for the change between the processing and decasting stages ($p < 0.0001$).

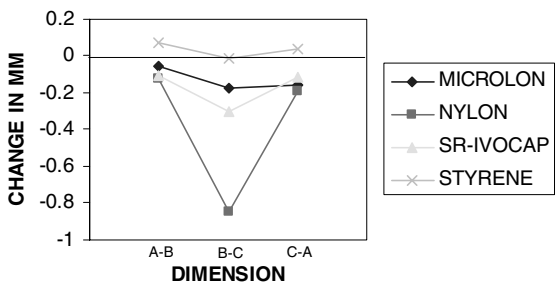

Figure 3. Mean change between processing and deflasking stages by dimension for each material.

Discussion

Overall there was considerable interaction among the factors, i.e., material, transition stage, and dimension in terms of their impact on deformation. An evaluation of the statistical analysis indicated that material effects were quite marked, showing less deformation for injection-molded PMMA during the change occurring between the wax and processing steps (Fig 2). Another trend was less deformation of styrene in terms of change measured between the processing and decasting stage, regardless of dimension measured (Fig 3).

Distortion occurred in the cross-arch direction for all materials evaluated in this study, which was in agreement with Woelfel and Paffenbarger.[4] All of the dentures evaluated exhibited some degree of shrinkage as a result of processing.

The processing shrinkage was greatest with nylon with 2.5% (0.82 mm) in the cross-arch dimension. This was 2.8 times greater than the conventionally processed PMMA. In this case, the nylon was plasticized at 550°F and injected into a cold flask. The polymerization process was essentially complete in 3 minutes, at which time the pressure from the injection-molding system was relieved. These results were in contrast with those reported by MacGregor.[10] The difference between the 2 studies could be due to the different types of nylon used. MacGregor used nylon-12, whereas in this study, the nylon was a hybrid compound of 4 different nylons. The exact nature of the material is proprietary and not disclosed by the manufacturer.

There may be several reasons for the large distortion of nylon observed in this study. Hargreaves[9] published some guidelines for the

processing of nylon for dental purposes. They were as follows: (1) a dental stone investment, oven-dried overnight at 40–45°C, and used for processing at 40°C, (2) a 10% sodium oleate solution as a separating medium, (3) a minimum of 2 sprues, (4) a relatively cool melt of 225°C, (5) the maximum injection pressure available, within the limits of the investment used, (6) a fast injection speed, and (7) a very slow rate of cooling. The dimensional accuracy of nylon is very technique sensitive during the processing stages. In this study, following the boil out procedure, the stone was allowed to cool completely before injecting the nylon. Nylon is a hydrophilic material, and its dimensional change is affected by water absorption. Even though the injected nylon may have been dry, it was injected into a moist flask, and the high temperatures involved may have evaporated the water from the stone and affected the nylon. The flask was also cold, effectively quenching the nylon, and producing frozen-in stresses within the material leading to a higher shrinkage when the denture is removed from the master cast. Finally, the different coefficients of thermal expansion of the nylon, and the stone may have contributed to stress development as with other materials tested.

The polymerization shrinkage of the styrene was about 0.38% (0.12 mm) in the cross-arch dimension. This was in agreement with a report by Woelfel,[15] but not with Anthony[16] and Peyton.[17] Shrinkage of styrene is almost 7 times less than the nylon and 2.4 times less than the conventionally processed PMMA. Styrene and injection-molded PMMA shrinkage were not statistically different. The accuracy of the styrene can be attributed to the fact that during the plasticizing stage at 500°F, the open flasks are also heated in an infrared oven to a high temperature; therefore, during the injection-molding process, a hot material is injected into a hot flask. This reduces the cooling range and minimizes the stress built up in the denture. This is further proven by the fact that with styrene, most of the distortion occurred during the processing stage and an insignificant amount in the deflasking stage. The low water sorption rate of the styrene may also be the reason for the exceptional dimensional stability of the dentures, which were the most stable dentures evaluated.

The injection-molded PMMA resin shrank about 0.65% (0.21 mm) in the cross-arch dimension. This was significantly better than conventionally processed PMMA, and this accuracy can be attributed to the processing technique. In this case, the resin is injected into a cold mold and held under pressure for 5 minutes before placing the flask (still under pressure) into boiling water for 35 minutes. This is followed by the placement of the flask into cold water for 20 minutes before the pressure is released. During the processing cycle, the polymerization shrinkage is compensated to a certain extent from a reservoir of non-polymerized material from the sprue. This improvement in dimensional stability was in agreement with the report of Anderson.[6] In this study, there were no statistically significant differences between the dimensional changes of styrene and injection-molded PMMA.

The conventionally processed PMMA, which is a heat-activated denture base, shrank about 0.9% (0.29 mm) in the cross-arch dimension. Polymerization of the heat-cured acrylic resin occurs in essentially the same manner as the chemically activated, or self-curing, resin, with the exception that heat, rather than a chemical accelerator, is employed to decompose the initiator in the polymer powder and initiate the reaction. The different amounts of shrinkage between these 2 types of resins may be attributed to the greater amount of cooling required after hardening of heat-cured material.

The dimensional accuracy of the styrene and the injection-molded PMMA was better than the conventional PMMA, and these results are in agreement with Woelfel,[15] Anderson,[6] and Huggett.[18] These differences are unlikely to be clinically significant, but the dimensional change of the nylon is considered clinically significant, and may have an effect on the final fit of the denture.

Although the linear dimensional accuracy of styrene was superior to that of nylon and conventionally processed PMMA, and not significantly better than injection-molded PMMA, there are other considerations that do not make styrene the material of choice for routine complete denture construction. First, it is difficult to adjust and polish. Second, denture teeth do not chemically adhere to the material, so mechanical retention such as diatorics is required. Finally, it is not possible to reline styrene because the denture has to be heated to 550°F, which not only distorts the denture but also damages the denture teeth.

Therefore, if there is a clinical indication to reline a styrene denture that treatment is not an option, and the denture will have to be remade.

The difficulty of the study is relating the degree of misfit observed in the laboratory to a clinical situation. Woelfel[15] examined warpage of complete dentures, and found that the shrinkage of 0.5 mm across the posterior region did not cause a serious misfit or discomfort clinically. When it was increased to 0.9 mm, the dentures did not fit properly.

Denture tooth movement subsequent to processing shrinkage is clinically important. If there is minimal processing error, the changes in the vertical dimension of occlusion are clinically insignificant before decasting; therefore, a laboratory remount and clinical remount procedures to correct the occlusion may be minimized.

Conclusions

1. In general, for all groups the greatest distortion occurred in the posterior palate across the arch (B–C dimension).
2. With conventionally processed PMMA, injection-molded PMMA, and nylon, the greatest distortion occurred when the processed denture was removed from the master cast, whereas for styrene, the greatest distortion occurred during processing.
3. When removed from the master cast, nylon had the greatest anteroposterior and cross-arch distortions, and styrene had the least.
4. The greatest overall distortion occurred with nylon, and the least with styrene.

Acknowledgments

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