Tensile Properties of Denture Base Resin Reinforced with Various Esthetic Fibers

Hakan Demir,[1] Orhan Murat Dogan,[1] Arife Dogan[2]

[1]*Department of Prosthodontics, Faculty of Dentistry, Cumhuriyet University, Sivas, Turkey*
[2]*Department of Prosthodontics, Faculty of Dentistry, Gazi University, Ankara, Turkey*

Received 26 March 2010; accepted 26 November 2010
DOI 10.1002/app.33831
Published online 15 September 2011 in Wiley Online Library (wileyonlinelibrary.com).

ABSTRACT: This study was conducted to determine the reinforcement effect of five types of esthetic fibers on the tensile properties of a conventional denture base resin. E-glass, polyester, rayon, nylon 6, and nylon 6/6 fibers were cut into 2, 4, and 6 mm lengths and added into resin randomly at a concentration of 3% by weight. For each formulation, five tensile specimens, as well as control specimens without fibers, were prepared in a dumbbell shape using a stainless steel mold, constructed according to ASTM Standard D638*M*-91a. Tensile properties were evaluated by using a universal testing machine. Surfaces of the tensile sections were also observed under the scanning electron microscope (SEM). Tensile strength of the specimens reinforced with fibers in varying lengths was found to be lower than that of the unreinforced control group. Among the trial groups, the specimens reinforced with 6 mm long polyester fibers showed the highest tensile strength. All the SEM fractographs indicated both weak adhesion and pull out of fibers from the matrix. None of the incorporated esthetic fibers appeared to improve tensile strength of the resin. © 2011 Wiley Periodicals, Inc. J Appl Polym Sci 123: 3354–3362, 2012

Key words: resins; fibers; mechanical properties; electron microscopy; adhesion

INTRODUCTION

Since early 1940s, poly(methylmethacrylate) (PMMA) has remained to be the principal material of choice for the denture base construction. Its popularity has mainly been due to its working characteristics, ease of processing and repairability, accuracy of fit, stability in oral environment, esthetic appearance, and low cost.[1–4] However, insufficient strength and stiffness of this material is still a cause of concern.

Denture fractures could result from weak mechanical properties of the denture base resin or because of the multiple factors leading to the failure of denture base material, such as increased ridge resorption, deep incisal notching at the labial frena, sharp changes at the contours of denture base or deep scratches and induced processing stress.[5,6] Dentures are subjected to various types of stress including compressive, tensile or shear stress during function,[6] and the stress intensification seems to be responsible for most of the denture fractures occurred inside the mouth. Waters[7] has described how stressed polymer chains could be distorted around stress concentrations.

Correspondence to: O. M. Dogan (mdogan1955@yahoo.com or mdogan@cumhuriyet.edu.tr).

Journal of Applied Polymer Science, Vol. 123, 3354–3362 (2012)

Over the years, numerous methods have been tried to improve the strength and performance of PMMA denture base. These have included some chemical approaches such as copolymerization and crosslinking of acrylic resin, reinforcement of the resin with different types of fibers[8–11] or by the addition of metal strengtheners.[12] Recently, much attention has been directed toward the fiber-reinforcement. It has been shown that carbon,[13–15] aramid,[1,6] glass,[1,4,6,16–18] or ultra high-molecular weight polyethylene fibers[19–23] had a significant strengthening effect when the amount of fibers was high enough and the technique was properly applied. In such a system, the fibers are embedded in a polymer matrix, which binds the fibers and forms a continuous phase surrounding the fibers. The polymer matrix transfers loads to the fibers, which are the strongest component of the composite.[24,25]

Reinforcing capacity of the fibers depends on their orientation and adhesion to the polymer matrix resin, and their impregnation with the matrix resin.[26,27] The glass fibers appear to be favored in dental applications because of good adhesion of silanated glass fibers to polymer matrix and their good esthetic qualities. It has been reported that glass fibers improved mechanical properties of the denture base polymers, especially fatigue resistance. Denture base reinforced with glass fiber has also

become clinically more successful.[4,15–29] Nevertheless, efforts still continue to search for alternative materials with better mechanical properties than the commonly used PMMA. In this context, materials such as nylon 6, nylon 6/6, viscous rayon and polyester have been tested as reinforcing agents, and they have been shown to have improved some of the mechanical properties and esthetic appearance of PMMA.[1–3,6,30,31]

In our previous studies, we have tested the reinforcement effects of these fibers on impact strength and flexural properties of a conventional heat polymerized PMMA denture base material. Random incorporation of nylon 6, nylon 6/6, polyester, viscous rayon or glass fibers into the resin at 3% concentration by weight, and in 2, 4, or 6 mm lengths, was found to have no significant effect on flexural properties,[30] whereas impact energy tended to increase with fiber length, giving the highest value for rayon reinforced specimens of 6 mm length.[31] Because the tensile strength is a measure of the maximum nominal stress to be induced in a body under rupture,[8] the tensile test can also be expected to provide useful information on the ultimate strength properties of the polymers. Therefore, the main objective of the present investigation was to examine tensile strength properties of the same denture base resin, with or without reinforcement, using the same experimental conditions, with the hope to combine the promising mechanical properties of these fibers.

EXPERIMENTAL

Materials

Five types of fibers, supplied as threads [(E-glass (SMC3) (Cam Elyaf Sanayi, Kocaeli, Turkey); polyester (PE), rayon (RY), nylon 6 (N6), and nylon 6/6 (SM6) fibers (Kordsa, Kocaeli, Turkey)] were chosen for the reinforcement of a conventional heat polymerized acrylic resin (Meliodent, Heraeus Kulzer, Germany). The fibers were cut to lengths of 2, 4, and 6 mm with special scissors (Tipo, Spezial, Stahl, Solingen, Germany) and used without any surface treatment. The fiber content was determined as 3% by weight for all the fiber-reinforced groups. The fiber amounts to be added were determined relative to the premixed, measured resin powder weight (Sartorious AG, Gottingen, Germany) and not to combined powder and liquid or mixed resin weight.

Preparation of specimens

Five specimens were prepared for each length of the fibers tested and for the control PMMA resin. Tensile specimens that conformed to ASTM Standard D638*M*-91a[32] were prepared in dumbbell shape using a stainless steel mold in 115 × 30 × 2 mm^3 dimensions. Sufficient number of wax patterns were obtained using this mold, then flasked and eliminated in the conventional manner.

Acrylic resin with or without the fibers was mixed thoroughly at a powder/liquid ratio of 2.34 g/mL in an agate mortar manually. A dough-like consistence was attained; the resins were then packed into the spaces created by wax elimination. Dental flasks were pressed with a force of 100 kg (Emmevi SpA, Parma, Italy) and left for 15 min to remove voids. Excess flash was trimmed away on trial packing. The flasks were fixed with clamps and cured in a 70°C water bath for 1 h, then in boiling water for 30 min. Upon completion of polymerization, the flask was left to cool at ambient temperature before being opened. Deflasked specimens were manually polished with a 600-grit water-proof silicon carbide paper under the tap water.

All of the specimens were stored in 37°C distilled water for 24 h before the mechanical test. The test was performed in the standard laboratory atmosphere of 23°C ± 2°C and 50% ± 5% relative humidity.

Mechanical test

Each of the groups was subjected to the tensile test by using a universal testing device (Lloyd NK5, Lloyd Instruments Ltd., Fareham Hampshire, UK) at a cross head speed of 5 ± 1 mm/min until rupture. The tensile strength was calculated by the formula below:

$$T = F/A$$

where T is the tensile strength (MPa); F, the force at failure (N); and A, the original cross-sectional area (mm^2).

Statistical analysis

After data were obtained, mean values and standard deviations were calculated by using an SPSS statistical software program (13.0 version, SPSS Inc., Chicago, USA). Differences between the control and fiber groups at a given length and also those between different lengths for the same fiber group were evaluated by Kruskal-Wallis analysis of variance and Friedman tests, respectively. For pairwise comparison of the same length of fibers and for the different length of fibers, Mann-Whitney U test and Wilcoxon test were used, respectively. The groups, which had statistically significant differences, were indicated using the same superscripted letters.

Scanning electrone microscopic analysis

Surfaces of the tensile sections of test specimens reinforced with 6 mm fibers were observed under a

TABLE I
The Young's Modulus Mean Values (MPa)

Groups	Length			Statistical test	
	2 mm ($\bar{x} \pm Sd$)	4 mm ($\bar{x} \pm Sd$)	6 mm ($\bar{x} \pm Sd$) ($\bar{x} \pm Sd$)	Friedman	Wilcoxon
N6	2041.45 ± 167.78	2037.04 ± 160.12	2110.57 ± 358.12	$P = 0.549$	$P > 0.05$
RY	1901.78 ± 256.78	1823.25 ± 259.46	1928.97 ± 241.46	$P = 0.549$	$P > 0.05$
SMC3	1908.79 ± 244.00	2200.28 ± 250.39	2052.33 ± 429.24	$P = 0.247$	$P > 0.05$
SM6	2138.23 ± 225.54	1986.72 ± 112.22	2097.37 ± 103.39	$P = 0.165$	$P > 0.05$
PE	1887.86 ± 244.93	2080.42 ± 222.99	2098.29 ± 204.01	$P = 0.449$	$P > 0.05$
Control	2174.97 ± 223.48	2174.97 ± 223.48	2174.97 ± 223.48		
	KW = 7.78	KW = 8.56	KW = 5.29		
	$P = 0.169$	$P = 0.128$	$P = 0.381$		
	$P > 0.05$	$P > 0.05$	$P > 0.05$		

$n = 5$.

scanning electrone microscope (SEM) (QUANTA 400, FEI Company, Eindhoven, Netherlands) to obtain some insight into the interface between fibers and polymer matrix during the loading process. Tensile sections of the test specimens were prepared by using an Isomed low-speed saw (Buehler Ltd., Lake Bluff, IL) after the tensile test.

RESULTS AND DISCUSSION

Many of the investigations have shown that the principal strength deficiency of PMMA lies in its low tensile strength, particularly under impact or fatigue conditions.[8] To obtain an alternative material with relatively higher strength so as to be unbreakable under clinical conditions, to date research employing fibers such as carbon, aramid, ultra high molecular weight polyethylene, or glass fibers for the reinforcement of PMMA has continued. Some esthetic fibers such as rayon, polyester, nylon 6 or nylon 6/6 fibers, have also been tested,[1–3,6,30,31] and these have provided encouraging results in mechanical properties—especially on impact values—of PMMA. Because of the lack of conclusive evidence concerning fiber reinforcement and tensile strength, this *in vitro* study was performed to determine tensile properties of PMMA after the addition of these esthetic fibers.

It has been stated that the length and orientation of fibers in the resin matrix play a major role in determining mechanical properties of the reinforced PMMA. The placement of fibers for reinforcement purposes is optimized when they are positioned in continuous unidirection orientation parallel to the denture surface and perpendicular to the expected loading force.[13] However, incorporation of unidirectional fibers during packing has been found to be difficult, because fiber protrusion outside the mold compartment presented technical difficulties.[33] Some fibers may also be lost during finishing and polishing.[15,34] In addition, lateral spreading of the fiber rovings during pressing may result in a nonhomogenous distribution of the fibers in the matrix[17,35] Therefore, ease of manipulation of the reinforcing agent is a critical factor for practical use, and this has become an important selection criterion.

The use of short fiber system tends to be less expensive and more flexible when used in conventional methods of fabricating dentures.[13,20,34] They can be simply added and oriented randomly in the polymer matrix. As for polyethylene fibers which were over 4% (w/w) in concentration, Gutteridge[20] has pointed out the difficulty in manipulating them. Chen et al.[1] have tested aramid, glass, and polyester reinforcing fibers by incorporating fibers into the polymer matrix in 2, 4, or 6 mm lengths and at concentrations of 1, 2, and 3% (w/w). Their results have shown that impact strength tended to increase with fiber length and concentration, particularly with polyester fibers at 3% concentration and 6 mm in length. In this study, therefore, the short fibers of different lengths were used at 3% (w/w) concentration.

We chose the fibers not only for their strengths[36] and esthetic properties, but also in the lights of our previous experience. All of the fibers were cut to the length of 2, 4, or 6 mm, and used without any surface treatment so that the results obtained could be compared with those of our previous studies.[30,31] Weight percentages of the fibers added were based only on the premixed, measured resin powder weight, and not on the combined powder and liquid or mixed resin weight. Thus, the fibers were added to the recommended P/L ratio instead of replacing some of powder to avoid changing the polymerization reaction.

The results of the tensile test measurements, given in Tables I–III, showed Young's modulus, tensile strength, and maximum extension at break values for each combinations of the fibers. As can be seen, the unreinforced control specimens had the highest values for tensile strength (55.28 ± 5.44 MPa) and Young's modulus (2174.97 ± 223.48 MPa) among the groups. The tensile strength and the modulus of

TABLE II
The Tensile Strength Mean Values (MPa)

Groups	Length			Statistical test	
	2 mm ($\bar{x} \pm$ Sd)	4 mm ($\bar{x} \pm$ Sd)	6 mm ($\bar{x} \pm$ Sd)	Friedman	Wilcoxon
N6	$53.43 \pm 11.19^{a,b}$	52.86 ± 8.39	51.62 ± 13.18	$P = 0.819$	$P > 0.05$
RY	$45.50 \pm 7.94^{a,c,d}$	43.41 ± 6.37	49.46 ± 4.89	$P = 0.449$	$P > 0.05$
SMC3	$51.47 \pm 14.08^{b,e,f}$	48.80 ± 5.48	47.76 ± 12.01	$P = 1.00$	$P > 0.05$
SM6	$52.51 \pm 6.36^{c,e}$	52.07 ± 2.26	52.03 ± 4.49	$P = 0.819$	$P > 0.05$
PE	46.76 ± 4.88	50.50 ± 7.17	53.84 ± 12.39	$P = 0.247$	$P > 0.05$
Control	$55.28 \pm 5.44^{d,f}$	55.28 ± 5.44	55.28 ± 5.44		
	$KW = 13.70$	$KW = 8.46$	$KW = 7.84$		
	$P = 0.018$	$P = 0.132$	$P = 0.165$		
	$^*P < 0.05$	$P > 0.05$	$P > 0.05$		

$n = 5$.
* Groups with the same superscript letter are significantly different by Mann-Whitney U test at 5 % level ($P < 0.05$).

elasticity values of the conventional heat-cured denture base resins have been determined to be 48.3–60.1 MPa[37] and 2.5×10^3 MPa,[38] respectively. In this study, although the strength values of most of fiber-reinforced specimens remained within this range, it seemed that the addition of different types of fibers into the resin matrix did not improve tensile strength properties. Furthermore, the use of different lengths of fibers had no noticeable effects on tensile properties of the trial groups, except for SMC3 fiber reinforced specimens which yielded some statistical differences for extension at break values ($P < 0.05$). For this group, the highest extension at break value was recorded for the 2 mm long fiber (3.28 ± 0.68 mm), followed by 6 mm long fiber (2.48 ± 0.30 mm) (Table III).

The Kruskall-Wallis analysis of variance revealed that there were differences between control and trial test groups with 2 mm long fibers, in terms of the tensile strength values (Table II), and the maximum extension at break of the specimens ($P < 0.05$, Table III). Young's moduli values, however, did not show any difference ($P > 0.05$, Table I). When the 2 mm long fibers were added to PMMA matrix, N6 fiber-reinforced specimens produced the highest tensile strength value (53.43 ± 11.19 MPa), whereas RY fiber-reinforced specimens had the lowest (45.50 ± 7.94 MPa). With Mann-Whitney *U* test, control specimens, N6 fiber-, and SM6 fiber-reinforced specimens were found to be statistically different from RY fiber- and SMC3 fiber-reinforced specimens ($P < 0.05$). In terms of extension at break values, PE fiber-reinforced specimens showed the maximum value among the groups tested, and this group was statistically different from SMC3 fiber-, SM6 fiber-reinforced specimens, and also from control group ($P < 0.05$). Specimens reinforced with N6 and SM6 fibers also had differing extension values ($P < 0.05$).

In 4 mm long fiber added specimens, the tested groups did not exhibit any difference in terms of Young's moduli and in also tensile strength values ($P > 0.05$). SMC3 fiber reinforced specimens yielded the lowest extension at break value (2.10 ± 0.20 mm). Moreover, pairwise comparisons showed that the latter parameter was statistically different from those of other test groups ($P < 0.05$).

TABLE III
The Maximum Extension at Break Mean Values (mm)

Groups	Length			Statistical test	
	2 mm ($\bar{x} \pm$ Sd)	4 mm ($\bar{x} \pm$ Sd)	6 mm ($\bar{x} \pm$ Sd)	Friedman	Wilcoxon
N6	5.25 ± 3.49^{a}	2.75 ± 0.32^{e}	2.68 ± 0.42	$P = 0.449$	$P > 0.05$
RY	2.57 ± 0.56	2.65 ± 0.63^{f}	3.76 ± 2.08	$P = 0.449$	$P > 0.05$
SMC3	$3.28 \pm 0.68^{b,A,B}$	$2.10 \pm 0.20^{e,f,\ g,h.i,\ A,C}$	$2.48 \pm 0.30^{B,C}$	$P = 0.015$	$^*P < 0.05$
SM6	$2.47 \pm 0.25^{a,c}$	2.52 ± 0.10^{g}	2.42 ± 0.17	$P = 0.819$	$P > 0.05$
PE	$5.55 \pm 0.63^{b,c,d}$	2.84 ± 0.38^{h}	3.43 ± 1.54	$P = 0.247$	$P > 0.05$
Control	2.56 ± 0.30^{d}	2.56 ± 0.30^{i}	2.56 ± 0.30		
	$KW = 13.85$	$KW = 13.29$	$KW = 8.35$		
	$P = 0.017$	$P = 0.021$	$P = 0.138$		
	$^*P < 0.05$	$^*P < 0.05$	$P > 0.05$		

$n = 5$.
* Values labeled with lowercase letters indicate statistical differences between the groups; uppercase letters indicate statistical differences within each of the groups.

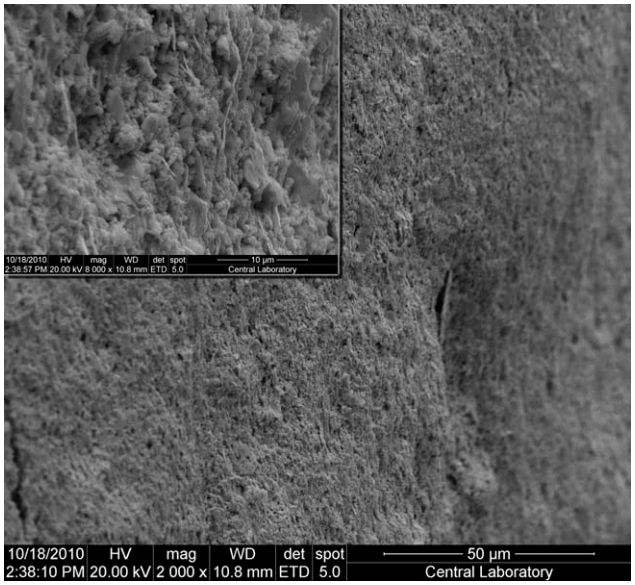

Figure 1 Tensile fractured surface of a control specimen under SEM (×2000 magnification; inner box at ×8000 magnification).

Addition of 6 mm long fibers resulted in no significant differences in any of the groups for the parameters tested ($P > 0.05$). However, longer fibers caused some increase in the tensile strengths and Young moduli for RY and PE fiber- reinforced specimens. After the control group, the specimens reinforced with 6 mm of PE fibers produced the highest tensile strength value (53.84 ± 12.39 MPa) in all conditions tested.

Data showed that all the fiber-reinforcements used in this study did not cause any substantial improvement in the tensile strength. Although tensile elongation of denture base polymers does not reach the limit of failure strain during service,[23] loss of tensile strength of the fiber-filled PMMA still remains to be an important issue. It has been generally known that in order for the fibers to be effective against craze initation, or to act as energy absorbers, a good fiber/matrix interfacial adhesion is essential.[25,39] The effect of each of the fiber reinforcements was obvious in the SEM fractographs at different magnifications. The micrograph of PMMA without reinforcement showed a homogeneous surface appearance (Fig. 1), whereas those of SMC3 fiber-reinforced specimens showed that the fibers were not well distributed in the polymer matrix, some fibers formed clusters in some areas (Fig. 2). All of the remaining fibers had almost similar SEM images, on which they appeared to be pulled out or slipped away from the resin matrix, leaving many void spaces and/or leading to small cracks propagating from circular pores under tensile loading. The spatial distribution of fibers was observed to be relatively uniform in the matrix with some fibers kinked out. However, at higher magnifications, SEM fractographs showed no evidence of PMMA particulates adhering onto the surface of these fibers (Figs. 3–6). The differences between the SEM appearances of the specimens reinforced with SMC3 fiber and other polymeric fibers may be attributed to their densities. While the density of glass fibers was 2.54 g/cm^3, those of the polymeric fibers tested ranged from 1.14 to 1.52 g/cm^3.[40,41] In other words, the volume fraction of glass fibers was about twofold less than that of other polymeric fibers,

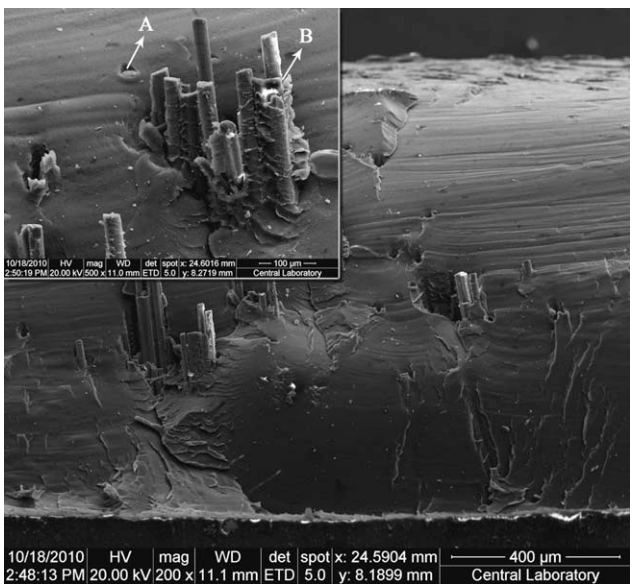

Figure 2 Tensile fractured surface of 6 mm length of glass fiber-containing specimen under SEM (×200 magnification; inner box at ×500 magnification) (A, pull out-; B, interfacial areas).

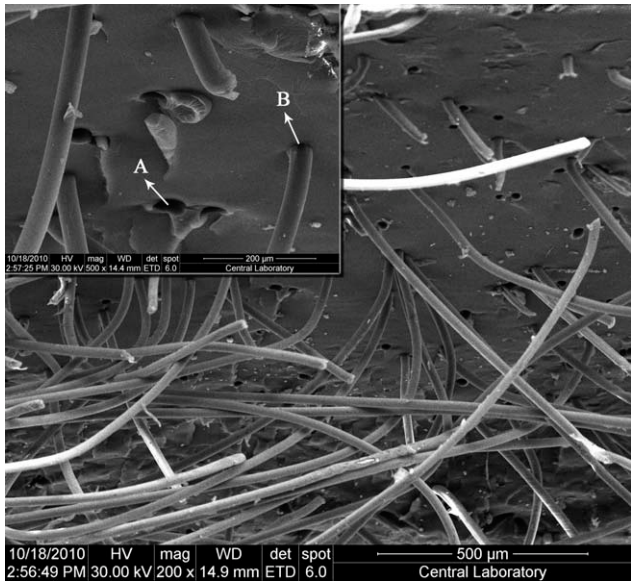

Figure 3 Tensile fractured surface of 6 mm length of nylon 6 fiber-containing specimen under SEM (×200 magnification; inner box at ×500 magnification) (A, pull out-; B, interfacial areas).

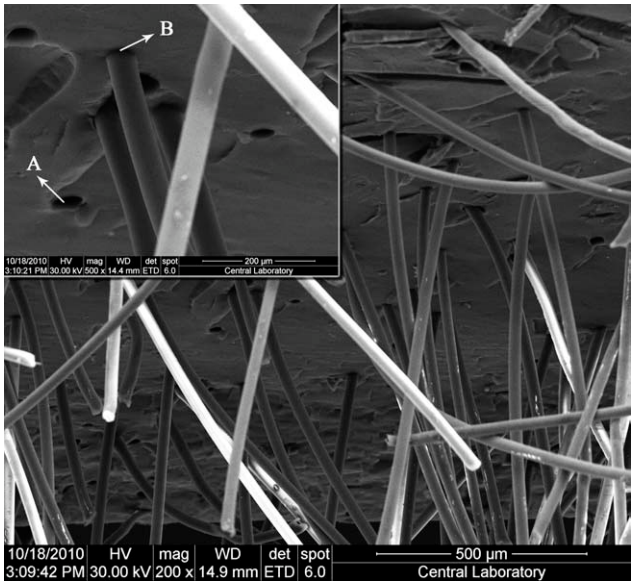

Figure 4 Tensile fractured surface of 6 mm length of nylon 6/6 fiber-containing specimen under SEM (×200 magnification; inner box at ×500 magnification) (A, pull out-; B, interfacial areas).

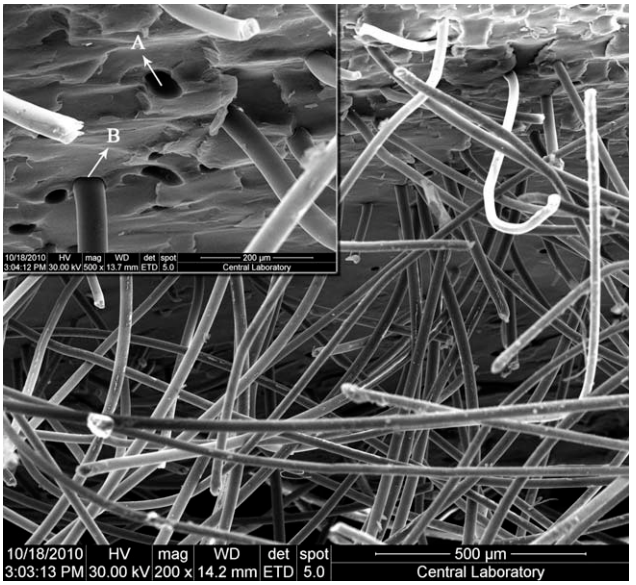

Figure 6 Tensile fractured surface of 6 mm length of polyester fiber-containing specimen under SEM (×200 magnification; inner box at ×500 magnification) (A, pull out-; B, interfacial areas).

implying that the dispersion of glass fibers was more difficult than that of polymeric fibers.

Decreases in the strength values with fiber reinforcement may be explained by several factors.[42,43] Because untreated fibers can act as inclusion bodies, and thus weaken the resin system,[18] one of the factors could be the use of fibers without any surface treatment. In our case, the random addition of fibers might also have disturbed the main matrix continuity and interfered with stress transfer between the fiber and polymer or within the same material.

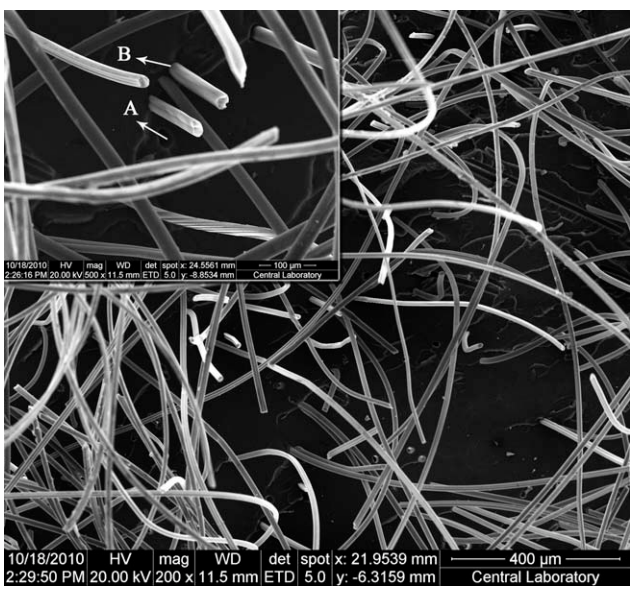

Figure 5 Tensile fractured surface of 6 mm length of rayon fiber-containing specimen under SEM (×200 magnification; inner box at ×500 K magnification) (A, pull out-; B, interfacial areas).

The second factor, which was not taken into consideration in the present study, could be the differences found in the thermal properties of two structure; because of the different coefficients of the thermal expansion between the matrix and the fibers, any bond formed while the resin undergoing polymerization, may later be broken upon cooling.[42,43]

It has also been stated that the impregnation of reinforcing fibers with the resin had important influence on bonding of fibers to polymer matrix, and incomplete wetting of the fiber surfaces by highly viscous PMMA resin may cause a sharp decrease in tensile strength by acting as stress concentration areas.[42,43] This might help explain our findings on SEM fractographs which strongly indicated both weak adhesion and pull out of most of the fibers from the main matrix under tension.

Glass fibers have been widely used in different forms to strengthen dental polymers, and generally treated chemically with silane compounds to produce sufficient bonding with the matrix.[18,28] Solnit[18] has measured transverse strength of PMMA-filled glass fibers, used with or without silane pretreatment and found that untreated fiber reinforced specimens were weaker than unreinforced specimens. This finding could suggest that the silane treated fibers were slightly stronger. On the other hand, Kanie et al.[16] have found no statistical differences on the flexural strength of test specimens reinforced with silanized or unsilanized glass fibers. In that study, they have claimed that bonding between

glass fiber and polymer matrix depended on the mechanical retention by polymerization shrinkage and roughness caused by the longitudinal and transverse threads, that is, on the frictional force between the glass fiber and polymer matrix.

In this study, the tensile strength of the glass fibers was found to be inversely proportional to fiber length that argued against that of Young's modulus. The SEM analysis of specimens reinforced with glass fibers partly supported the suggestion of Kanie et al.[16] Although any substantial improvement in the tensile strength was not achieved by the use of glass fibers, in some areas, the fibers bunched together in a typical bundle and some PMMA particulates were detected on the surface of them (Fig. 2). However, in some areas tension force applied led to some voids left by the fibers, suggesting the existence of poor impregnation areas. The effect of added glass fibers on the strength of PMMA has been demonstrated by several investigators stating that the poor impregnated regions in the fiber composite led to voids between the fibers.[17] If the glass fibers had been completely wetted and a homogenous mixture attained, then they could have strengthened PMMA above its limits.

There is little evidence to justify using a particular type or proportion of esthetic fibers in denture base materials. The choice has often been based on tradition, availability and empiricism. Several materials such as rayon (acetate rayon, viscose rayon), acrylic resin fibers, nylon fibers, PVC, and wool can be used as esthetic fibers. They may be added in different lengths, ranging from 1 to 9 mm, different denier and at percentages between 0.1% and 3% by volume.[3]

Nylon is a generic name for certain types of thermoplastic polymers belonging to the class known as polyamides. This family of condensation polymers results from the reaction of a diacid with a diamine. Amide groups are extremely polar, and hydrogen bonded with each other. Depending on the linking groups between the acid or amine groups, different types of nylon with distinct physical and mechanical properties can be produced.[30,44] The chief advantage of nylon lies in its resistance to shocks and repeated stressing.[6] Moreover, as the backbone of nylon is regular and symmetrical, it forms very good fibers.[30]

Nylon fibers are usually added to the denture material to simulate the minute blood vessels underlying the oral mucosa. Recently, John et al.[6] have tested nylon 6 fibers as reinforcing agents and reported that the incorporation of nylon fibers increased the flexural strength of the PMMA denture base specimens, compared to unreinforced specimens. This finding argues against our previous results. Indeed, flexural strength of N6 reinforced specimens was decreased especially with increasing lengths, compared with that of control unreinforced specimens; and among the trial groups the best result was obtained by the addition of SM6 at 6 mm of length.[30] In respect of the impact values, both fibers have showed a significant increase in absorbed energy as their length increased.[31] In this study, N6 and SM6 gave the relatively higher tensile strength values compared to those of other fiber-reinforced groups. Although strengthening performance of N6 showed a small degree of decrease with fiber length, SM6 yielded more stable values for all the lengths used. The specimen shape and size and also the direction of load to break the specimens might lead to difference strength values obtained, for a given fiber by different mechanical test methods.

There is little data in the literature related to the mechanical effect of rayon fibers when used as reinforcing agent for the PMMA.[30,31] It has been reported by Winkler and Vernon[45] that rayon acetate fibers were primarily employed because they were completely insoluble in monomer. However, they absorbed monomer, resulting in changes in their length and diameter, and it was also noticed that the absorbed monomer was slowly released from the fibers. In our previous studies,[30,31] the use of this fiber reinforcement led to a considerable decrease in flexural strength; whereas when used at 6 mm length, it has become the most effective material in terms of impact energy absorbed among the tested groups. By studying the effect of viscose rayon esthetic fibers at different concentrations on the flow properties of an acrylic resin denture base material, Katsikas et al.[3] have demonstrated a rapid increase in viscosity at higher levels, especially at 3%. This result has been explained by stating that larger quantities of fibers consumed the available monomer by absorption. They have also concluded that the increase in viscosity could influence flow properties which might, in turn, adversely affect the molding process. In the present study, rayon fibers seemed to be the weakest reinforcing material in terms of tensile properties. This may be due to the high viscosity of acrylic dough resulting in insufficient impregnation at 3%. Whereas the impact surfaces showed that these fibers tended to remain enveloped in the resin resisting pull out. In the current study, the SEM fractographs indicated no evidence of adhesion between PMMA and the fiber; they were pulled out completely under tension, leaving the voids behind (Fig. 5). These larger voids and cracks might lead to poor tensile properties for the rayon reinforced specimens.

Fiber reinforcement can only be effective if the stress is transferred from the polymer matrix to the fiber. This can be achieved by the fibers having a length equal to or greater than the critical fiber length.[46] In fact, PE fibers showed an increase in strength values with fiber length, and the best result

of such reinforcement was obtained using 6 mm long fibers. Although the test protocols used are different, this finding was in agreement with those of Chen et al.[1] who have concluded that incorporation of 3% (w/w) of 6 mm polyester fiber offered the best formulation for acrylic denture base reinforcement in terms of flexural strength. By evaluating the tensile results, it may be said that fiber reinforcements with this length works effectively. However, differences in extension at break values of specimens could suggest that when 2 mm long fibers were used, the specimens reinforced with PE could withstand better to applied stress without permanent deformation (Table III). Here, it may become obvious that different fiber systems require different lengths to be used.

Strengthening by fiber reinforcement is based on the principle that a relatively soft ductile polymer matrix is fully capable of transferring an applied load to fibers via shear forces at the interface. The fibers will be the main load-bearing constituents, while the matrix forms a continuous phase to surround and hold the fibers in place.[24] The matrix and the fiber, however, have different tensile strains because of their different modulus. In the region of the fiber ends, the strain in the fiber was less than that in the matrix. As a result of this strain difference, shear stresses were induced around the fiber in the direction of the fiber axis and the fiber was stressed in tension. Thus, a fiber length longer than the estimated critical fiber length can withstand a significant load even up to the fracture load and improve the strength of reinforced PMMA in comparison with unfilled PMMA.[33]

In terms of stress transfer, the elastic modulus of fibers should ideally be greater than the elastic modulus of the matrix so that at a given strain the fibers could absorb more stress.[4] Clinicians generally demand that the denture base should be rigid. A high value of modulus of elasticity is therefore advantageous. A high value of elastic limit is also required to ensure that stresses encountered during biting and mastication should not cause permanent deformations.[38] However, under the test conditions of this current study, the elastic modulus of the specimens reinforced with the different fibers appeared to be lower than that of the PMMA unreinforced.

The results of tensile test performed did not offer sufficient information to suggest that these fibers could be used as resin strengtheners. Improving this parameter by the chemical treatment of fibers will be a subject for a future study. Futhermore, their effects on other physical and clinical properties of the denture base resins must also be studied. It thus becomes clear that further work is required to better understand the nature of reinforcement afforded by the herein-evaluated esthetic fibers.

CONCLUSIONS

Within the limits of this study, following conclusions can be drawn:

1. The incorporation of nylon 6, nylon 6/6, rayon, glass, or polyester fibers into the resin led to a decrease in tensile properties of fiber/resin composites.
2. An increase in fiber length had no noticeable effect on the parameters used.
3. Among the trial groups, the highest tensile strength was obtained by the addition of 6 mm long polyester fibers.
4. SEM images indicated a weak adhesion between fiber/resin interface, and the fibers tended to pull out under tension.

The authors thank Professor Ali Fazil Yenidunya, Research Centre for Medical School of Cumhuriyet University, for his invaluable contribution in preparing this manuscript for submission. They also thank Mr. Zafer Artvin, Middle East Technical University, Central Laboratory, for his help in taking of the SEM photographs.

References

1. Chen, S. Y.; Liang, W. M.; Yen, P. S. J Biomed Mater Res Part B: Appl Biomater 2001, 58, 203.
2. Chen, S. Y.; Liang, W. M. Mid Taiwan J Med 2004, 9, 203.
3. Katsikas, N. G.; Huggett, R.; Harrison, A.; Vowles, R. W. Dent Mater 1994, 10, 2.
4. Jagger, D. C.; Harrison, A.; Jandt, K. J Oral Rehabil 1999, 26, 185.
5. Jagger, D.; Harrison, A.; Jagger, R.; Milward, P. J Oral Rehabil 2003, 30, 231.
6. John, J.; Gangadhar, S. A.; Shah, H. J Prosthet Dent 2001, 86, 424.
7. Waters, N. E. Dent Pract Dent Rec 1968, 18, 389.
8. Stafford, G. D.; Smith, D. C. Br Dent J 1970, 128, 442.
9. Harrison, A.; Huggett, R.; Jagger, R. C. J Dent 1978, 6, 299.
10. Ruyter, I. E.; Svendson, S. A. J Prosthet Dent 1980, 43, 95.
11. Stafford, G. D.; Smith, D. C. Br Dent J 1968, 125, 337.
12. Sehajpal, S. B.; Sood, V. K. J Prosthet Dent 1989, 61, 746.
13. DeBoer, J.; Vermilyea, S. G.; Brady, R. E. J Prosthet Dent 1984, 51, 119.
14. Schreiber, C. K. Br Dent J 1971, 130, 29.
15. Yazdanie, N.; Mahood, M. J Prosthet Dent 1985, 54, 543.
16. Kanie, T.; Fujii, K.; Arikawa, H.; Inoque, K. Dent Mater 2000, 16, 150.
17. Vallittu, P. K.; Lassila, V. P.; Lappalainen, R. J Prosthet Dent 1994, 71, 607.
18. Solnit, G. S. J Prosthet Dent 1991, 66, 310.
19. Braden, M.; Davy, K. W. M.; Parker, S.; Ladizesky, N. H.; Ward, I. M. Br Dent J 1988, 164, 109.
20. Gutteridge, D. L. Br Dent J 1988, 164, 177.
21. Ladizesky, N. H.; Pang, M. K. M.; Chow, T. W.; Ward, I. M. Aust Dent J 1983, 38, 28.
22. Ladizesky, N. H.; Chow, T. W. Aust Dent J 1992, 37, 277.
23. Taner, B.; Doğan, A.; Tinçer, T.; Akinay, A. E. J Oral Sci 1999, 41, 15.
24. Foo, S. H.; Lindquist, T. J.; Aquilino, S. A.; Schneider, R. L.; Williamson, D. L.; Boyer, D. B. J. Prosthodont 2001, 10, 148.

25. Vallittu, P. K. J Prosthodont 1996, 5, 270.
26. Vallittu, P. K. J Prosthet Dent 1999, 81, 318.
27. Lassila, L. V. J.; Vallittu, P. K. J Contemp Dent Pract 2004, 5, 14.
28. Vallittu, P. K. J Oral Rehabil 1997, 24, 124.
29. Vallittu, P. K.; Ruyter, I. E.; Ekstrand, K. Int J Prosthodont 1998, 11, 340.
30. Dogan, O. M.; Bolayir, G.; Keskin, S.; Dogan, A.; Bek, B. J Mater Sci: Mater Med 2008, 19, 2343.
31. Dogan, O. M.; Bolayir, G.; Keskin, S.; Dogan, A.; Bek, B.; Boztug, A. Dent Mater J 2007, 26, 232.
32. ASTM. "Annual Book of ASTM Standards", ASTM Standard D 638M-91a, Part 08–01; ASTM: Philadelphia, PA, 1992.
33. Marei, M. K. J Prothed 1999, 8, 18.
34. Dixon, D. L.; Breeding, L. C. J Prosthet Dent 1992, 67, 417.
35. Vallittu, P. K.; Lassila, V. P.; Lappalainen, R. Dent Mater 1994, 10, 116.
36. Mark, J. E. Polymer Data Handbook; Oxford University Press, Inc.: United Kingdom, 1999.
37. Craig, R. G.; Restorative Dental Materials; Mosby: St. Louis, 1989.
38. McCabe, J. F.; Walls, A. W. G. Applied Dental Materials; Blackwell Science: Edinburgh, 1998.
39. Cho, K.; Yang, J. H.; Park, C. E. Polymer 1997, 38, 5161.
40. Callister, W. D. Materials Science and Engineering: An Introduction; John Wiley: New York, 2003.
41. Rodriguez, F.; Principles of Polymer Systems; Mc Graw-Hill: New York, 1983.
42. Kilfoil, B. M.; Hesby, R. A.; Pelleu, G. B. J Prosthet Dent 1983, 50, 40.
43. Krause, W. R.; Park, S. H.; Straup, R. A. J Biomed Mater Res 1989, 23, 1195.
44. O'Brien, W. J. Dental Materials and Their Selection; Quintessence Publishing: Chicago, 2008.
45. Winkler, S.; Vernon, M. H. J Prosthet Dent 1978, 40, 4.
46. Nielsen, L. E. Mechanical Properties of Polymer and Composites; Marcel Dekker: New York, 1974.