Structural and mechanical properties of new block copolymers designed for biomedical use

P. Ferruti*, E. Martuscelli, L. Nicolais, M. Palma and F. Riva
Laboratorio di Ricerche su Tecnologia dei Polimeri e Reologia del CNR, Arco Felice (Napoli), Italy

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Copolymers containing blocks of conventional polymers and polyamidoamine blocks have been prepared in order to couple good mechanical properties with the ability to form stable complexes with heparin which will induce non-thrombogenic characteristics to the polymer. The tensile properties of these new non-thrombogenic materials (polyamidoamine/polystyrene block copolymers with different polystyrene contents) have been studied and compared with structural data obtained by X-ray diffraction technique. The results show that two distinct phases are present in the copolymers. These materials have potential for applications in biomedical fields.

INTRODUCTION

The aim of this paper is to report a study on the mechanical properties, and some structural characteristics of polystyrene/polyamidoamine block copolymers (PS/PAA). The results are analysed by means of current theories of composite materials.

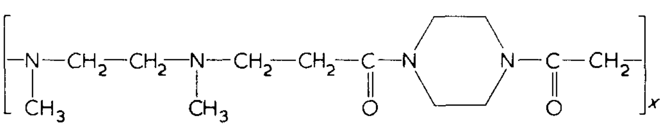
The polystyrene/polyamidoamine block copolymers, whose preparation is reported elsewhere in detail[1], constitute very promising materials for biomedical use, owing to their ability to become, after heparinization, permanently non-thrombogenic[1]. It is well known that one of the main problems encountered in using polymers for artificial organs designed for long-term implantation (e.g. artificial heart) is the thrombogenic activity of these materials when in contact with the blood[2].

An interesting approach to the preparation of non-thrombogenic materials is the adsorbtion of heparin on the surface of synthetic polymers[3,4] by using quaternary ammonium salts. However this heparinization method has several disadvantages which make it unsuitable for long term artificial prosthesis. The quaternary ammonium salts are only weakly bonded to the polymeric surface and a slow release (of these salts) into the blood stream usually cannot be avoided. Furthermore, chemical compounds containing quaternary ammonium groups usually have haemolytic properties, and may exert unfavourable side effects on platelets[5]. In previous work it was found that a group of synthetic polymers of polyamidoamine structure were able, in a crosslinked form, to adsorb heparin from plasma or blood in a very selective way, giving stable complexes without exerting any unfavourable side effect, either on the plasma proteins or on the blood cells, including platelets[6–9]. Because the mechanical properties of these polymers were unsatisfactory, copolymers containing blocks of conventional polymers besides the polyamidoamine groups, have been prepared so as to couple good mechanical performance with the ability to form stable complexes with heparin.

The PS/PAA block copolymers constitute a first example of this type of materials.

EXPERIMENTAL

The block copolymers PS/PAA, have a central block derived from the polyaddition of *N,N'*-dimethylethylenediamine to 1,4-bis-acryloyl-piperazine:



and lateral blocks of polystyrene. The overall structure of these copolymers, is probably that shown in *Figure 1*. Samples were prepared by varying the relative amounts of the polyamidoamine and polystyrene blocks as described previously[9]. The volume concentrations of PAA in the polystyrene matrix were: $V_f = 0.09, 0.16, 0.25, 0.33, 0.50$.

The PS/PAA yarns were obtained by extrusion of bulk materials under controlled conditions. The extrusion temperature decreases, with the amount of PAA in the copolymers as reported in *Table 1*. The extrusion of samples with a very high percentage of PAA (≃80%) was not possible for the high elasticity of the material.

Stress–strain data have been obtained using an Instron universal testing instrument. Tests were made at room temperature and at a constant strain rate of 0.5 min^{-1}.

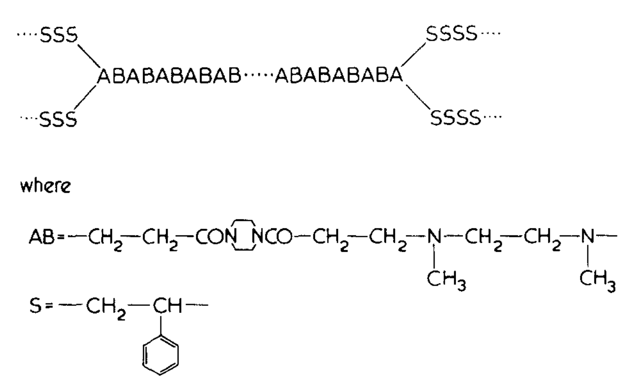


Figure 1 Most probable structure of the PS/PAA block copolymers

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* Istituto di Chimica del Politecnico di Milano, Milano, Italy.

Table 1 Extrusion temperatures of PS and PS/PAA copolymers

T_e(°C)	V_f(% PAA)
165	0
163	9
152	16
148	25
144	33
140	50

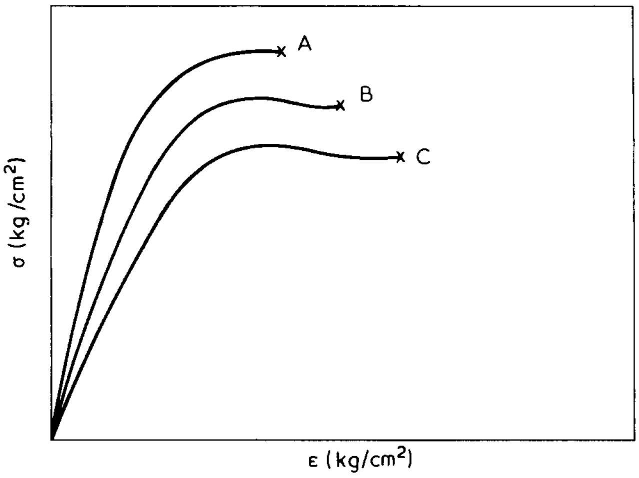

Figure 2 Typical stress–strain curve of (A) PS and PS/PAA copolymers at (B) 9% and (C) 25% of PAA

The tensile strength, σ_b, and the yield stress, σ_y, were related to the initial cross-section area of the unstretched sample; strain-at-break, ϵ_b, was determined as the ratio of maximum length of the sample to the initial distance between the clamps that was constant and equal to 2 cm.

The yield stress is defined as the first point at which the tangent of the stress–strain curve becomes zero, the corresponding value of the strain is referred to as yield strain ϵ_y.

The work to break has been calculated as the area under the stress–strain curve. The reported values are always the average of at least 40 experiments.

X-ray diffraction analysis of PS/PAA copolymers both in bulk or in yarn was carried out by using for the bulk material a Philips powder diffractometer, X-ray generator PW 1130, goniometer 1050/25 and normal electronic panel PW 1360 (CuKα radiation) and for yarns a cylindrical camera (ϕ = 57.3 mm, CuKα radiation).

RESULTS AND DISCUSSION

Typical stress–strain curves of PS/PAA copolymers at different PAA content are shown in *Figure 2*. The modulus and the strength decrease with PAA content, while the elongation at break increases. This behaviour is similar to that of rubber-modified thermoplastics[10]. The initial elastic modulus of the copolymers E_c decreases with the PAA content and can be expressed as a function of its volume fraction V_f by the Kerner equation[11] valid in the case of composites with spherical inclusions:

$$E_c = E_p \frac{1 + ACV_f}{1 - CV_f} \qquad (1)$$

where

$$A = \frac{7 - 5\mu_{PS}}{8 - 10\mu_{PS}} = 1.17$$

and

$$C = \frac{E_f/E_p - 1}{E_f/E_p + A} \simeq -\frac{1}{A} = -0.85$$

μ_{PS} is the Poisson's ratio of PS, E_f and E_p are the Young's moduli of the PAA and PS, respectively, and E_f is assumed to be much lower than E_p.

In *Figure 3* the elastic moduli calculated from equation (1) are compared with experimental results. From this Figure it can be seen that the agreement between the experimental data and the theoretical values is good enough to imply that the PAA acts as a second rubbery phase in a glassy polystyrene matrix.

The yield stress of the PS/PAA copolymers decreases as the volume content of PAA increases. The decrease could be simply a reflection of the reduced cross-sectional area of the polymer bearing the load. A simple model for the dependency of the yield stress on the volume fraction of filler for glass bead filled thermoplastics has been presented[12]. According to this model the load is carried only by the continuous phase i.e. the rigid polymer. Further, yielding is assumed to occur in the minimum cross-section of the continuous phase which is perpendicular to the applied load; i.e. in the cross-section when the stress is at its maximum. For spherical inclusions this model yields the following equation:

$$\sigma_c = \sigma_p(1 - 1.21\, V_f^{2/3}) \qquad (2)$$

where σ_c and σ_p are the strength of the PS/PAA and that of PS, respectively.

In *Figure 4* the experimental values of the strength are in fair agreement with those calculated, once more supporting the hypothesis of two phases present in the copolymer.

In *Table 2* all the data relative to the tensile tests are reported. While the strength σ_b of these materials is re-

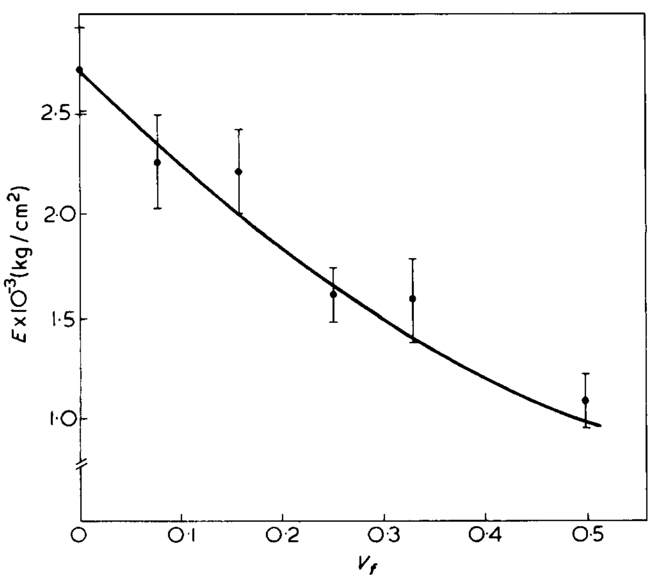

Figure 3 Young's modulus *E versus* PAA content (V_f) for PS/PAA copolymers. The line plotted refers to the Kerner equation[11]

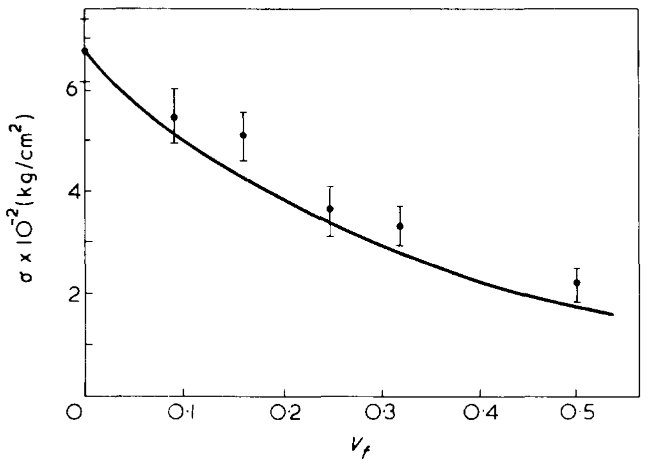

Figure 4 Yield stress of PS/PAA copolymers *versus* PAA content. The line is a theoretical prediction[12]

Table 2 Tensile properties of PS and PS/PAA copolymers

V_f (% PAA)	E (kg/cm^2)	σ_y (kg/cm^2)	ϵ_y	σ_b (kg/cm^2)	ϵ_b	W (kg cm/cm^3)
0	26 900	685	0.038	674	0.045	18.3
9	22 500	548	0.039	551	0.055	21.9
16	22 000	513	0.031	514	0.058	18.0
25	16 000	362	0.031	372	0.076	25.8
33	15 800	336	0.035	391	0.086	23.9
50	11 000	249	0.037	252	0.057	13.0

duced by increasing PAA content, the elongation at break ϵ_b increases up to 33% PAA and then decreases. The work to break W, finally, is not much effected by the presence of PAA up to 33%.

These results show that the addition of PAA in a PS matrix leads to composites with lower modulus and strength, higher elongation at break without any significant variation of the work to break. Furthermore, these materials seem very attractive for use in the biomedical field, owing to their high elasticity coupled with high toughness.

To confirm the presence of two distinct phases in the PS/PAA copolymers further structural analysis on these materials has been carried out.

In *Figure 5* wide-angle X-ray (CuKα radiation) diffraction patterns of pure atactic PS, pure PAA and some PS/PAA copolymers with different PAA content are shown. While PS/PAA samples having content up to 50% were completely amorphous, the samples with 80% PAA gave crystalline peaks. The spectrum is very similar to that of pure PAA, but crystallinity is lower.

Atactic PS shows two halos with different intensities at 2θ values of 9.5° ($d = 9.3$ Å) and of 19.5° ($d = 4.65$ Å). Pure PAA shows, on the contrary, only one amorphous halo at about the 2θ value of the more intense halo of atactic PS. No diffraction is given by pure PAA at $2\theta = 9.5°$.

These considerations lead one to assume that in the PS/PAA copolymers the ratio R between the area of the amorphous halo at $2\theta = 9.5°$ and the total area is proportional to the weight fraction of PS in the samples.

We can derive the following simplified equation:

$$R = \frac{(1-p)2A}{(1-p)(2A+B)+pC} \qquad (3)$$

In the above equation $2A$ and B are the areas of two amorphous halos of pure atactic PS, whose maxima are at $2\theta = 9.5°$ and $2\theta = 19.5°$, respectively; C is the contribution to the diffusion of the amorphous halo of PAA with the maximum at about 19.5° and p is the PAA weight fraction in the copolymers.

The method followed for the separation of the areas in the X-ray patterns of PS/PAA copolymers, in order to calculate R, is shown in *Figure 6*.

The constants A, B and C were experimentally measured from the spectra of pure PS and pure PAA. It should be noticed that in the calculation of R we have assumed that the contribution of PAA and PS blocks are additives.

This assumption is also supported by the mechanical behaviour of the PS/PAA copolymers previously discussed.

The experimental values of R and those calculated according to the equation (3) are reported in *Figure 7*. The theoretical curve interpolates fairly well the experimental points.

The wide-angle X-ray diffraction patterns of yarns obtained by extrusion of PS/PAA copolymers having a PAA content up to about 50% show that these yarns are amorphous, but an equatorial orientation in the halo at lower

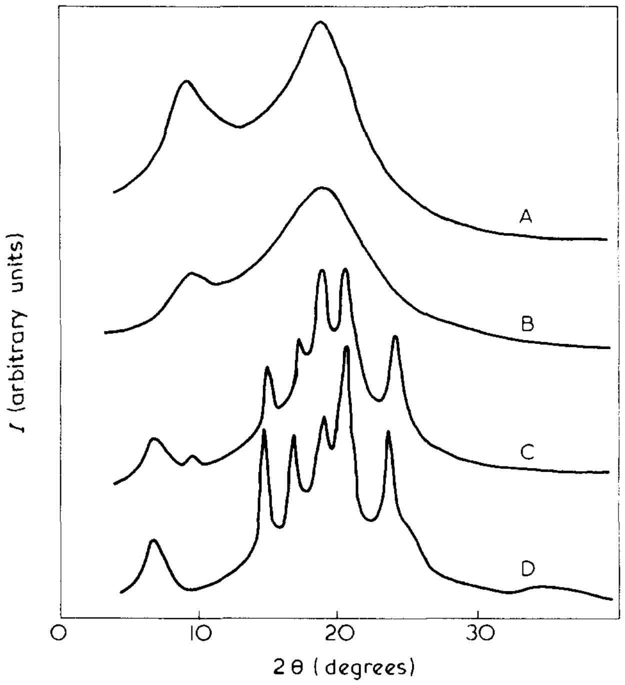

Figure 5 Wide-angle X-ray CuKα diffraction traces of PS/PAA samples of copolymers as a function of composition. The percentage of PAA in the copolymer: A, 0; B, 50; C, 80; D, 100

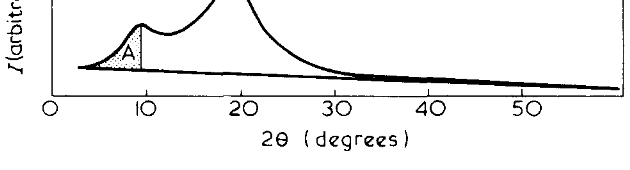

Figure 6 Method of separation of areas in the X-ray traces of PS/PAA copolymers in order to calculate R. $R = 2A$/total area

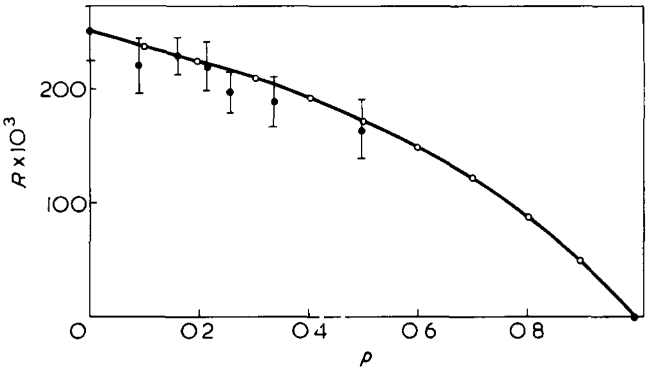

Figure 7 Dependence of R upon the percentage of PAA: ●, experimental values calculated from the X-ray traces; ○, theoretical values calculated according to relation (3). $R = (1-p)2A/(1-p)(2A+B)+pC$

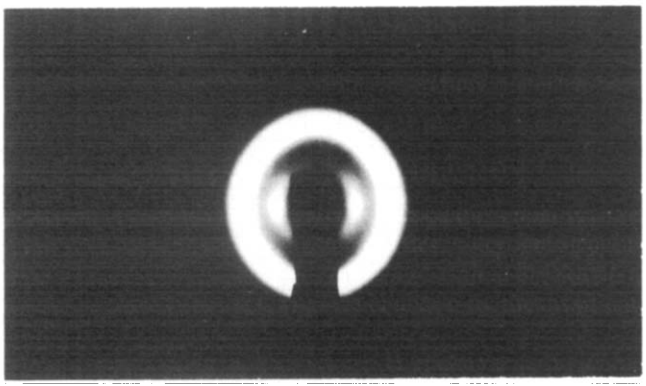

Figure 8 Wide-angle X-ray cylindrical (CuKα) camera photograph of yarn of PS/PAA copolymer showing equatorial orientation of the amorphous halo at lower angle

angle is observed in some cases (see *Figure 8*). This result together with the optical birefringence observed in all fibres, indicates that the molecules are slightly oriented. In the more stretched yarns (i.e. the ones mechanically oriented at room temperature by means of the Instron machine) this orientation effect is more pronounced.

In conclusion, from the structural analysis it can be inferred that in the PS/PAA copolymers there is a second rubbery PAA phase in the glassy PS matrix. Similar conclusions were drawn from the mechanical experiments. The resulting properties of these new materials show potential for use in biomedical applications. In particular their high toughness and good elastic properties together with the possibility of achieving, by heparinization, permanently non-thrombogenic characteristics, present the possibility of preparing artificial prostheses for human implantation.

Work in this respect has started and probes of PS/PAA copolymers have been inserted in the inferior vena cava of test animals (dogs) with very encouraging results[1].

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