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A review of plasma treatment and the clinical application of polyethylene fibres to reinforcement of acrylic resins

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The available literature on the plasma treatment of polyethylene fibres and their clinical applications has been critically reviewed. It is suggested that some misconceptions could inhibit the full exploitation of these materials.

1. Introduction

There is now a substantial body of literature dealing with the plasma treatment of polyethylene fibres and their various applications. These include the reinforcement of acrylic resins for orthopaedic surgery use [1]. In view of the apparent contradictions in some of the research there appears to be merit in attempting to review the situation critically.

There are two principal types of highly oriented polyethylene fibres, produced by (a) melt spinning of comparatively low molecular weight polymer [2] (e.g. Tenfor and Certran), and (b) gel spinning of high molecular weight polymer [3] (e.g. Spectra and Dyneema). In both cases a very high draw ratio drawing process is employed to produce a high modulus/high strength fibre. Both types of fibres have similar chemical and morphological characteristics, resulting in outstanding tensile modulus, strength and fracture toughness along the fibre axis, low density, chemical inertness, low moisture absorption and good biocompatibility. On the negative side, both types of material share poor compression properties, low surface energy (poor wettability) and an inadequate interfacial bond between the fibres and various matrix resins.

The gel spun fibres, with their high molecular weight, have a higher strength than the melt spun fibres, but the latter have a lower manufacturing cost. In many respects the two types of fibres are qualitatively identical and studies made with one type can be safely extended to the other. For simplicity, where reference is made to research embracing both types of fibres, the material will be referred to as high performance polyethylene fibres.

2. Survey of the early literature on the surface modification of high performance polyethylene fibres

2.1. Adhesion to industrial resins

Ladizesky and Ward studied the pull-out adhesion of untreated and plasma-treated melt spun polyethylene fibres to epoxy and polyester resins [4, 5]. They found that plasma treatment with oxygen gas produces a cellular structure on the surface of the filaments (subsequently referred to as surface roughness by some authors) into which the resin penetrates to give mechanical keying between the resin and the filament. The combined effect of this and other factors increased the pull out adhesion by a factor of about $\times 10$. Nardin and Ward [6] made a detailed examination of the influence of surface treatments (chemical and oxygen gas plasma etching) on the adhesion of melt spun fibres to epoxy resin. They concluded that the resultant adhesion depends on three factors: (i) wettability, (ii) the extent of the surface cellular structure (including the depth and aspect ratio of the pits); and (iii) the number of chemical bonds per unit area between the fibre and the resin. These three factors are broadly additive and optimum pull out strength is obtained when their respective contributions reach maximum values.

In a further study Ladizesky and Ward [7] examined the effect of various chemical and plasma treatments of melt spun polyethylene fibres on the pull-out adhesion to epoxy resin. They found that plasma treatment with oxygen gas results in maximum adhesion, although using other gases such as argon and helium increased the adhesion by a factor of $\times 4$. Argon and helium produced only a very minor cellular surface topography and it was shown that the mechanism of failure during pull-out, as well as the increase of adhesion, were both clearly related to the level of the surface roughness.

2.2. Adhesion to clinical resins

The known biocompatibility of polyethylene has also focused attention on the application of high performance polyethylene fibres as reinforcement for acrylic resins in clinical use, namely bone cement and denture base polymers. These materials are virtually identical [8, 9], consisting of poly(methyl methacrylate) or PMMA. Studies on these materials are therefore complementary, while bearing in mind the different technological requirements of each application. The first

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TABLE I Tensile strength and pull-out adhesion of high performance polyethylene fibre

Fibre	Plasma treatment	Surface cellular structure	Resin	Fibre tensile strength (GPa)	Pull-out adhesion (MPa)	Authors and year of publication
Melt spun and drawn	None	None	Epoxy	0.98[a]	0.5	Ladizesky and Ward (1983) [4]
	O_2	High		0.58[a]	4.9	
	He	Minor		–	1.9	Ladizesky and Ward (1989 [7]
	Ar	Minor		–	2.3	
	CF_4	None		–	0.9	
	None	None	PMMA	–	0.4	Ladizesky (1990) [11]
	O_2	High		–	2.7	
	None	None	RM-3	–	0.5	
	O_2	High		–	3.3	
	None	None	Bis-GMA	–	0.4	
	O_2	High		–	3.5	
Spectra 900	None	None	Epoxy	3.53	0.6	Li and Netravali (1992) [13]
	Allylamine	None		3.44	1.8	
Spectra 900	None	None	PMMA	3.40	0.6	Hild and Schwartz (1992) [12]
	Ar	None		3.44	1.6	
	CO_2	None		3.31	1.7	
	N	None		3.47	1.7	
Spectra 1000	None	None	PMMA	3.58	0.8	
	Ar	None		3.56	2.1	
	CO_2	None		3.73	1.9	
	N	None		4.14	2.2	

[a] Experiments carried out with 300 μm diameter monofilaments and batch plasma treatment. For 15 μm diameter filaments and continuous plasma treatment, as used for composites, the tensile strength values are 1.08 GPa (untreated) and 0.93 GPa (plasma treated) [15].

pull-out adhesion results for melt spun polyethylene fibres (untreated and plasma treated with oxygen gas) to PMMA were reported by Braden *et al.* [10]. Ladizesky [11] extended this work in a study which included PMMA, as well as two other acrylic type resins in clinical use, namely a urethane dimethacrylate (RM-3) and an acrylic/epoxy composition usually known as Bis-GMA. This work also investigated the effect of water immersion at 37 °C (up to 9 months) on the pull out adhesion (a matter of particular importance in clinical applications). The adhesion levels obtained with all these resins were dramatically increased by plasma treatment of the fibres with oxygen gas. Compared with these increases, the effect of water immersion was relatively minor.

3. Survey of recent work on the surface modification of high performance polyethylene fibres

3.1. Pull-out adhesion and fibre tensile strength

Pull-out of gel spun (Spectra) fibres was measured by Hild and Schwartz [12] (PMMA) and Li and Netravali [13] (epoxy resin), both works published in 1992. In view of the then proven capability of oxygen gas plasma treatment to increase the adhesion of melt spun polyethylene fibres to resins [4–7, 11], it is perhaps surprising that none of the above authors reported results with this gas, concentrating instead on argon, nitrogen and carbon dioxide [12] and allylamine plasma deposition [13]. Two reasons were given for this approach: (i) safety problems related with the use of pure oxygen gas, and (ii) oxygen plasma treatment increases interfacial bond strength at the expense of fibre strength (as reported by Ladizesky and Ward [4]).

Regarding the safety aspects of plasma treatment with oxygen gas, Kaplan and Rose [14] state "oxygen is one of the more commonly employed (plasma) process gases". Furthermore, it hardly needs to be emphasized that oxygen gas is in widespread use in many branches of industry, as well as in clinical procedures, and including surgical operating theatres.

Table I summarizes the pull-out adhesion and fibre tensile strength results presented by Ladizesky *et al.* [4, 7, 11], Hild and Schwartz [12] and Li and Netravali [13]. The data corresponds to the plasma treatment parameters giving maximum adhesion enhancement. Gases other than oxygen give a significant increase in the pull-out adhesion, but well below the values obtained with oxygen plasma treatment. From Table I, it is seen that only oxygen gas gives rise to a high cellular structure on the surface of the fibres, thus adding an extra mechanism to the factors contributing to the resultant adhesion, according to the analysis made by Nardin and Ward [6].

Hild and Schwartz [12] and Li and Netravali [13] are correct in pointing out the significant decrease of the fibre tensile strength associated with oxygen gas plasma treatment (≈40%), as seen in Table I, and their results confirm the earlier results of Ladizesky and Ward [4, 7]. The main aim of the early work was to establish the inherent suitability of the novel technique, with little attention paid to the optimization of the plasma parameters involved and the resultant

properties of the material. In particular, the oxygen gas plasma treatment referred to in Reference 4 was a batch process applied to thick monofilaments of 300 μm diameter. Further work, published as early as 1986 [15] investigated the effect of continuous oxygen gas plasma treatment (bobbin to bobbin, the whole system at low pressure) applied to thin filaments of 15 μm diameter which were made of a different parent linear polyethylene, as used for composites. In this case the plasma parameters for optimum adhesion reduced the tensile strength of the filaments from 1.08 GPa (untreated) down to 0.93 GPa (oxygen gas plasma treated). This treated material also showed a significant cellular surface topography, but the individual pits were smaller and of different shape compared to the pits seen in treated monofilaments.

More recent work [16] was carried out with a modified plasma equipment, so that the feeding and take-up bobbins remain at room conditions. Also, further changes were made in the parent linear polyethylene. In this case oxygen gas plasma treatment for optimum adhesion reduced the tensile strength of melt spun fibres from 1.3 GPa (untreated) to 1.1 GPa (plasma treated). Under similar conditions the tensile strength of gel spun (Spectra) fibres was reduced from 2.6 GPa (untreated) to 2.3 GPa (plasma treated). In all cases plasma treatment resulted in an extensive cellular fibre topology.

The results discussed above suggest that Hild and Schwartz [1, 12, 17] and Li and Netravali [13] have been overpessimistic in their assessment of the effect of plasma treatment with oxygen gas on the tensile strength of high performance polyethylene fibres. The work of these authors, using gases other than oxygen has added substantial understanding to the earlier studies of Ladizesky and Ward [4, 7] and Nardin and Ward [6]. However, to assess the technological merits of the new treatments it should be noted that (a) plasma treatment with oxygen gas gives rise to the highest fibre/resin adhesion, with only a minor effect on the tensile strength of the fibres (Table I and related text), and (b) as will be seen in the following sections, the only reported results of mechanical properties of composites reinforced with fibres plasma treated with gases other than oxygen gas correspond to acrylic resins with less than 1 wt % chopped fibre loading (Hild and Schwartz [1]). Such low loadings do not permit a meaningful comparison of the effects of the new and old treatments on the mechanical properties of the respective composites.

3.2. Mechanical properties of composites reinforced with unidirectional fibres

Table II summarizes the most important results available for the mechanical properties of composites reinforced with unidirectional high performance polyethylene fibres. All treatments shown in Table II produced a minor decrease in the tensile strength of the fibres, significant changes in their surface topology and a consistent increase of interlaminar shear strength (ILSS). The properties of reinforced acrylic resins appear to benefit from the increase in adhesion associated with the plasma treatment of the fibres, but no significant effect is seen in the properties of epoxy resin reinforced with the melt spun fibres. On the other hand, the flexural strength of epoxy resin reinforced with gel spun (Spectra) fibres increases with increased adhesion, probably because of the very low ILSS of this system when using untreated fibres.[1]

Table II reveals some inconsistencies among the reported flexural modulus results for the reinforced epoxy resin. For the gel spun (Spectra) fibre composites, Tissington *et al.* [18] quote significantly higher values than those reported by Kaplan *et al.* [20, 21], the latter being similar to the flexural modulus of the composites reinforced with melt spun fibres. Since gel spun (Spectra) fibres have substantially higher tensile modulus than the melt spun variety, it is likely that the values reported by Tissington *et al.* [18] are the most reliable.

As shown in the previous discussion, the 5 year period prior to 1988 saw substantial research effort on the effect of plasma treatment on the adhesion and mechanical properties of melt spun polyethylene fibre composites. The results have been reported in the first publication, Reference 4 (1983), as well as Reference 15 (1986), 23 (1986), 24 (1986), 6 (1987) and 10 (early 1988). In view of this considerable body of available know-how, it is somewhat surprising to read the following statement by Kaplan *et al.* [20] published in 1988 "The facile manner by which plasma surface treatment modifies the interface adhesion to allow the improvement of composite properties is also noteworthy. These improvements were not the result of a long tedious program, but rather the results of preliminary programs". It should be noted that these authors do not state the gas used in their experiments.

3.3. Application of chopped high performance polyethylene fibres to the reinforcement of clinical acrylic resins

The clinical demands on PMMA as a denture material and as a bone cement lead to different mechanical properties requirements. Mandibular and maxillary dentures fail predominantly by impact and fatigue, respectively [25, 26]. Therefore, improvements of the impact toughness and fatigue resistance of the material are desirable, including a reduction or elimination of the high notch sensitivity of PMMA [19, 27, 28]. Furthermore, an increase in stiffness is also advantageous for dental purposes to minimize deformation of the appliance during function.

[1] These low ILSS values may be due to the relatively large diameter of gel spun (Spectra) fibres when compared with melt spun drawn fibres. A recent paper by Woods and Ward [22] has shown that, for a given fibre content, the ILSS value decreases as the reinforcement diameter increases. This applies for both untreated and plasma treated fibres.

TABLE II Mechanical properties of polymeric resins reinforced with 48–56 vol % of uniaxially oriented high performance polyethylene fibre

Fibre	Plasma treatment	Surface cellular structure	Resin	Fibre tensile strength (GPa)	Mechanical properties of composites						Authors and year of publication
					ILSS (MPa)	FS (MPa)	FM (GPa)	TM (GPa)	TS (GPa)	CS (MPa)	
Melt spun and drawn	Unreinforced		Epoxy	–	–	95	3.0	3.0	0.82	120	Ladizesky and Ward (1986) [15]
	None	None		1.08	20	156	21	22	0.34	74	
	O_2 (continuous)	High		0.93	25	144	19	20	0.34	76	
	None	None		1.26	14	–	–	–	–	–	Woods and Ward (1993) [16]
	O_2 (batch)	High		1.23	20		–	–	–	–	
	O_2 (continuous)	High		1.23	28	–	–	–	–	–	
	None	None		–	16	172	27	31	0.47	88	Tissington *et al.* (1991) [18]
	O_2 (batch)	High		–	28	169	23	28	0.47	91	
	Unreinforced		PMMA		–	105	3.2	–	–	–	Ladizesky and Chow (1992) [19]
	None	None		–	16	158	17	–	–	–	
	O_2 (batch)	High		–	22	188	22	–	–	–	
	None	None	RM-3	–	26	195	21	–	–	–	
	O_2 (batch)	High		–	28	209	25	–	–	–	
	None	None	Bis-GMA	–	20	213	20	–	–	–	
	O_2 (batch)	High		–	36	238	26				
Spectra	None	None	Epoxy	2.63	7.2	–	–	–	–	–	Woods and Ward (1993) [16]
	O_2 (Not stated)	High		2.26	22	–	–	–	–	–	
	None	None		–	8.0	142	68	41	0.62	79	Tissington *et al.* (1991) [18]
	O_2 (batch)	High		–	14	184	62	43	0.65	81	
	None	None		2.59	8	146	20	–	–	–	Kaplan *et al.* (1988) [20]
	Unstate gas (continuous)	Yes (not shown)		2.33	31	234	28				
	None	None		–	8	146	20				Kolluri *et al.* (1988) [21]
	Mixture O_2 + tetrafluor-methane	High		–	31	234	28	–	–	–	

The requirements for bone cement are somewhat more complex, with added emphasis of physicochemical and biological characteristics. For load-bearing structures such as hip joint replacement, the combination of poor adhesive properties of the cement and the mismatch of the mechanical properties of the bone–cement–metal implant components give rise to frequent implant loosening. It is often stated [1, 29, 30] that, with Young's moduli of the structure in ratios of 10:1:100, the weak link in the prostheses is the poor mechanical properties of PMMA. Knoell *et al.* [31] found that PMMA reinforced with carbon fibre resulted "in a significant increase in stiffness (modulus of elasticity) without compromising the flexural strength of the material .... The exotherm also decreased significantly". They concluded "on the basis of this brief investigation, it appears feasible to use graphite fibre additive to improve certain of the mechanical and thermal properties of surgical bone cement". Thus, these authors see an increase in stiffness as a worthwhile improvement of the material.

Conversely, it has been argued [32–34] that an increase in the stiffness of the bone cement may have a detrimental effect on the load transmission from the prostheses to the surrounding bone. Owing to the complexity of the problem, the arguments at present are on a qualitative level.

Other requirements for bone cement are better understood. For example, the observed failure of the structure within the bulk of the cement may be minimized by increasing the ductility and fatigue resistance of the material. Other, non-mechanical properties of importance include dimensional changes taking place during polymerization and during sorption of body fluids, as well as the exothermic nature of the polymerization. Dimensional changes are detrimental to the integrity of the interfaces in the structure and give rise to stresses in the bulk of the polymer, whereas the heat produced during curing are associated with traumatization of the body tissue.

Various approaches have been followed in order to improve the properties of PMMA for dental and orthopaedic surgery applications. These include rubber-toughening of the material [33, 35] and porous PMMA to allow bone ingrowth within the cement and improve the interface strength [36]. Also, other polymers have been explored with the aim of replacing the conventional material [37]. These new types of cements can also be filled with bioactive ceramic powders [34].

Notwithstanding the various approaches explored, there has been a concentration of effort on the reinforcement of PMMA with high performance fibres, namely glass fibres [38], carbon fibres [31, 32, 39–41], Kevlar [8, 14, 42] and, during the last 10 years, high performance polyethylene fibres. A full review of the various lines of current and past research lies outside the scope of this publication. Instead, emphasis will be given to the assessment of PMMA with high performance polyethylene fibres for dental applications [10, 11, 19, 28, 43–53] and for surgical applications [1, 12, 54, 55]. Owing to the similarity of the materials used, the known-how from these two lines of research is largely complementary [8, 9].

Ladizesky *et al.* [19, 45, 56] have shown substantial improvements in the mechanical properties of PMMA reinforced with increasing volume fractions of uniaxially oriented or woven high performance polyethylene fibres. They also showed that dental prostheses with these two forms of reinforcement and 30–40 vol % fibre content can be successfully produced with standard dental techniques [49, 56]. Nevertheless, *a priori* it appears that chopped fibres (in lengths of between 1 and 20 mm) are the most convenient form for PMMA reinforcement, particularly for bone cement, which has to be prepared "*in situ*".

In view of the proven increase in performance of PMMA with increasing fibre fraction [19, 45, 56], it is a matter of some concern that all the research on the reinforcement of bone cement has concentrated on small loadings of chopped fibre. This situation is not restricted to high performance polyethylene fibres [1, 54, 55], but also applies to the use of other fibres such as Kevlar [8, 29] and carbon fibres [32, 39, 40]. In fact, the highest chopped fibre loading quoted in the publications on bone cement is 7 wt % prior to mixing with the liquid monomer [8, 29, 32, 55]. Since polymerization incorporates the monomer into the resin, the final maximum weight fraction of fibre would be below 5 wt %. (Knoel *et al.* [31] incorporated 10 wt % of carbon fibre, reporting considerable mixing difficulties and a compressive strength of only a fifth of the unreinforced cement, owing to the presence of voids.) The generally indifferent results obtained with these maximum polyethylene fibre loadings lead Hild and Schwartz [1] to limit their studies of reinforced PMMA to a maximum fibre content of 1 wt % prior to mixing with the liquid monomer. Wagner and Cohn [54], in an earlier work using similar materials and fibre loadings concluded "since the fibre content is necessarily very low due to constraints inherent to the use of such low-density fibres, and unless a change in conception is implemented, this approach seems to be of limited value".

In 1991, Ladizesky *et al.* [28, 57] reported a newly developed, simple incremental mixing technique which achieves over 30 vol % of chopped melt spun polyethylene fibres into PMMA. The fibres can be either untreated or plasma treated and the quoted loadings are those present in the resin after polymerisation. Table III shows the results obtained with such high fibre loadings, as well as results reported by other authors using low fibre loadings. The table includes only the tests common to the literature being discussed, namely flexural strength and flexural modulus.

It is seen that the flexural strength of the acrylic resin is broadly insensitive to the reinforcement, whatever the level of fibre loading. Hild and Schwartz [1] and Wagner and Cohn [54] interpret this result by speculating on the effect of fibre/resin interface strength, low fibre volume fraction and the geometrical design of the test. The measurement of flexural strength of acrylic resin reinforced with high chopped fibre content has shown that this insensitivity is an intrinsic characteristic of this type of composite. Such

TABLE III Mechanical properties of PMMA resin reinforced with chopped high performance polyethylene fibre

Fibre	Fibre loading (vol %)	Plasma treatment	Surface cellular structure	Fibre tensile strength (GPa)	Mechanical properties of composites		Authors and year of publication
					FS (MPa)	FM (GPa)	
Melt spun	0	–	–	–	93	2.9	Gutteridge
and drawn	2	None	None	1.08	95	3.6	(1992) [44]
		O_2	High	0.93	89	3.2	
	0	–	–	–	105	3.2	Ladizesky
	2.4	None	None	1.26	103	3.3	*et al.* (1993) [48]
	0	–	–	–	104	3.2	Cheng *et al.*
	37	None	None	1.26	83	7.6	(1993) [28]
		O_2	High	1.23	98	9.7	
Spectra	0	–	–	–	66	2.8	Wagner and
	1	None	None	–	63	2.7	Cohn (1989) [54]
		O_2	Not shown	–	60	2.7	
	0	–	–	–	68	1.3	Pourdeyhimi and
	1	None	None	–	78	0.7	Wagner (1989) [55]
	0	–	–	–	58	1.7	Hild and
	1	None	None	3.4	58	2.6	Schwartz
		Ar	None	3.4	63	3.0	(1993/2) [1, 12]
		CO_2	None	3.3	62	2.9	
		N	None	3.2	59	2.8	

behaviour can be qualitatively understood in terms of the mechanical anisotropy of the fibres, their distribution within the composite and the tensile and compressive strength of the fibre and the resin [28].

Nevertheless, the insensitivity of the flexural strength with fibre loading is of little clinical importance in the dental field because dentures are not subjected to deformations approaching their failure deflection. In the orthopaedic field, however, an increase in the bone cement strength would have been an advantage. It is to be noted that the failure strain (ductility) of the reinforced resin is not compromised by the addition of the fibres, even at high volume fractions [28].

Stiffness relates to the resistance to deformation. The flexural modulus data presented in Table III for the melt spun fibres/PMMA (dental) resin is conclusive, namely, the modulus is not affected by low fibre loading, whereas high fibre loading more than doubles its value, whether the fibres are untreated or plasma treated. On the other hand, the available data for gel spun (Spectra) fibre/PMMA (bone cement) resin, all for low volume fraction of fibres, is somewhat confusing. Some authors report no change [54], others report a significant decrease [55], while a further group of researchers [1] finds a $\times 1.8$ increase in the modulus of the resin after incorporation of the fibres, whether untreated or plasma treated. In the two latter cases [55, 1] the modulus quoted for the unreinforced resin is particularly low.

It should be noted that the incorporation of a large volume fraction of chopped fibres into PMMA has other benefits not included in Table III [28]. For example, the impact toughness of the resin increases by a factor of $\times 4$ while, at the same time, the material loses its well-documented notch sensitivity to become notch insensitive. These advantages are probably more important for dental applications than in a surgical context, although Pilliar *et al.* [39] comment on the deleterious effect of notches on the impact resistance of PMMA bone cement, both unreinforced and reinforced with carbon fibres.

It has also been shown [51–53] that the water sorption decreases linearly and directly proportional to fibre content, whereas dimensional changes during polymerization and during immersion also decrease linearly but proportional to twice the fibre content (these effects are broadly independent of fibre form and treatment). It follows that an acrylic resin with 40 vol % chopped high performance polyethylene fibres has negligible dimensional changes. As remarked above, bone cement serves as an interfacial phase between the metallic implant and the bone, and the elimination of dimensional changes should result in a substantial reduction of internal stresses within the system and more stable interfaces, leading to longer lasting implants.

The experiments with high volume fractions of chopped fibres [28] have been carried out with heat curing, dough-type denture base resin, whereas the bone cement is similar to the cold curing dough-type denture base resin [8, 9]. In all cases the presentation is similar, namely a powder (polymerized PMMA plus initiator) and the liquid MMA (monomer plus, for cold curing resins, an activator). In the dental laboratory or the operating theatre, the powder and the liquid are mixed and used a few minutes later, when the dough stage is reached. It was found [28] that the dough stage of the resin with high chopped fibre volume content can be delayed for several months by placing the mix in a refrigerator or freezer unit. When returned to room temperature the mix can then be used in the standard manner. These findings could be of particular importance for the technology of bone cement because the mix (powder, fibre and liquid, the latter without activator) could be prepared beforehand, leaving for the operating theatre the incorporation of an activator.

Summarizing, reinforcement of PMMA with high polyethylene chopped fibre content produces a material with significantly enhanced modulus and impact strength, notch insensitive and dimensionally stable. Ductility remains unaffected. It should also be noted that fatigue studies of these systems, presently in progress, are encouraging. The resin reinforced with at least 30 vol% polyethylene fibres is likely to have a significantly reduced exotherm during curing, owing to the removal of reactants. The material can be prepared beforehand, stored under refrigeration and used when required.

The literature on bone cement reinforced with low content of gel spun (Spectra) and Kevlar fibres contains some excellent mathematical analysis of the flexural deformation tests [8, 29, 54, 55]. Hild and Schwartz [1] further expanded this analysis by examining the energy needed for fibre slippage (toughness index) after crack propagation, providing new understanding on the failure processes taking place in low fibre content reinforced resins. It would be of considerable interest to see such analysis applied to the latest technological developments, namely the reinforcement of acrylic resins with high fibre content [28].

4. Conclusions

In this review of the surface modification and application of high performance polyethylene fibres we have attempted to review the considerable body of knowledge already available in these subjects. It appears that there is unnecessary scepticism of the practical value of the incorporation of the fibres into clinical resins and that considerable and valuable effort has been expended on lines of research which do not represent an optimum technological approach.

It is hoped that this review will serve to focus future work on the most promising areas of research and lead to the successful application of these important new materials which have already been demonstrated in very extensive clinical trials and research [10, 11, 19, 28, 45–53, 56].

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