ELSEVIER

Biomaterials 22 (2001) 2749–2757

Biomaterials

www.elsevier.com/locate/biomaterials

Polypropylene fumarate/phloroglucinol triglycidyl methacrylate blend for use as partially biodegradable orthopaedic cement

Muthu Jayabalan[a,*], Vinoy Thomas[a], P.N. Rajesh[b]

[a]*Polymer Division, Biomedical Technology Wing, Sree Chitra Tirunal Institute for Medical Sciences and Technology, Thiruvananthapuram 695 012, India*
[b]*School of Technology and Applied Sciences, Mahatma Gandhi University, Kottayam, India*

Received 19 April 2000; accepted 8 January 2001

Abstract

Polypropylene fumarate/phloroglucinol triglycidyl methacrylate oligomeric blend-based bone cement was studied. Higher the percentage of phloroglucinol triglycidyl methacrylate, lesser the setting time. An optimum setting time could be arrived with 50 : 50 blend composition of the two oligomers. Composite cement of 50 : 50 blend prepared with hydroxyapatite granules of particle size 125 μm binds bovine rib bones. The tensile strength of this adhesive bond was found to be 1.11 kPa. The thermal studies suggest the onset of cross-linking reaction in the cured blend if the blend is heated. The absence of softening endotherm in the cured blend shows the thermosetting-like amorphous nature of blend system, which may restrict the changes in creep properties. The in vitro biodegradation studies reveal possible association of calcium ions with negatively charged units of degrading polymer chain resulting in slow down of degradation. Relatively slow degradation was observed in Ringer's solution. The study reveals the potential use of polypropylene fumarate/phloroglucinol triglycidyl methacrylate as partially degradable polymeric cement for orthopaedic applications.

Keywords: Polypropylene fumarate; Phloroglucinol triglycidyl methacrylate; Bone cement; Orthopaedic applications

1. Introduction

Biodegradable bone cements that would provide immediate structural support and accelerate normal bone healing and remodelling process have potential orthopaedic and maxillofacial applications such as temporary stabilization for bone in-growth prostheses and augmentation of fracture fixation especially when the underlying bone stock is deficient or osteoporotic [1]. Tri calcium phosphate (TCP) and hydroxyapatite (HA) have been used as alternatives to bone grafts in the filling of skeletal defect [2]. These materials are considered as osteoconductive and non-osteoinductive [3]. Attempts have been made to alleviate this problem using composites of these particulate and compounds such as plaster of Paris, fresh blood and polylactic acid [4]. The idea of using polymers as binders for particulate ceramics to produce composites with improved handling properties and to overcome the problem of brittleness associated with ceramic bone implants has been studied by Bonfield et al. [5]. Poly(α-hydroxy) acids, mainly lactide and glycolide homo- and copolymer are being the first choice among other biodegradable materials in the preparation of biomaterials for hard tissue repair [6–10]. Such composites are in effect biodegradable bone cements though their utility depends on the physical, mechanical and biological properties. Sandner et al. [11] have studied the polymerization shrinkage of composite resins and evaluated the working time of composite resins.

For orthopaedic and maxillofacial applications, it is suggested that biodegradable bone cement should meet certain criteria [12]. It must behave like cement that can be molded, shaped or injected into certain complex internal cavities in bone and harden in situ and develop mechanical properties sufficient to permit functional loading of the region. Bone cement must be biodegradable so that it does not act as a barrier to bone remodelling or fracture healing, but rather is replaced by host bone over a period of time. For orthopaedic applications, a synthetic, biodegradable and injectable oligomer, which can crosslink into hard material in vivo would be

* Corresponding author. Fax: +91-471-314814.
E-mail address: jaybalan@sctimst.ker.nic.in (M. Jayabalan).

0142-9612/01/$ - see front matter
PII: S0142-9612(01)00018-7

more useful for correcting osseous defects and inducing bone regeneration. Bioresorbable poly(propylene fumarate) (PPF), its copolymers and composites have been studied as promising biomaterials for bone tissue engineering [13–15].

PPF was synthesized through different methods by various research groups. Sanderson [16] has prepared PPF as powder by a trans-esterification reaction between diethyl fumarate and propylene glycol in the presence of an acid catalyst, *p*-toluene sulphonic acid by heating to 250°C over a period of 5 h and vacuum drying at 220°C for 4 h. Gerhard and Hayes [17] have synthesized low molecular weight PPF (M_n = 500–1200, polymerization index (PI) = 3–4) by the condensation reaction between propylene glycol and fumaric acid by heating to 145°C for 5 h and then to 180°C. Gresser et al. [18] produced PPF (M_n-2600, PI-2.6) by the reaction of fumaric acid and propylene glycol using *p*-toluene sulphonic acid catalyst and *t*-butyl hydroquinone. Domb et al. [19] has prepared PPF by the reaction between propylene glycol and fumaric acid at 130°C for 10 h and then at 180°C for 2 h to get a viscous liquid with molecular weight (M_n) of 300–2000 and PI of 1.5–1.7. Domb et al. [20] have also prepared PPF by another two steps process. Initially, trimeric bis(2-hydroxy propyl fumarate) (HPF) was prepared by the reaction of propylene glycol and mixture of fumaric acid and pyridine in acetone. Then, propylene bis (hydrogen maleate) (PHM) was prepared by the reaction of propylene glycol and maleic anhydride in toluene at 100°C for 24 h. The intermediate product PPF (M_n-750, PI-2.4) was then prepared by melt condensation of HPF and PHM at 140°C for 24 h. Domb et al. [20] have introduced various terminal groups (acrylate and epoxide) in PPF to study the degree of crosslinking and mechanical properties of the composite without crosslinking monomer. Yaszemski et al. [21] and Peter et al. [22] also have produced PPF (M_n-850, PI-2.0) by two step process. Initially, bis(2-hydroxy propyl fumarate) was prepared by the reaction of fumaryl chloride and propylene glycol at room temperature. The trans-esterification was carried out at 160°C for 24 h under vacuum with antimony trioxide catalyst.

Szmeresanyi et al. [23] and Andreis [24] have prepared PPF through the condensation reaction between maleic anhydride and propylene glycol and studied the isomerization of maleate to fumarate groups. The synthesis of PPF by all these methods require careful implementation of polymerization conditions since the condensation reaction is an equilibrium reaction and the liberated by-product should be removed. The molecular weight of the PPF is limited by the reversible polyesterification and trans-esterification reactions proceeding in temperatures above 180–200°C for several hours. Moreover, the condensation reaction at 190°C also involves cyclization side reaction involving the maleic anhydride and glycols which reduces the unsaturation (about 10–20%). However, the condensation reaction carried out at lower temperature enables only partial isomerization of maleate to fumarate groups. In order to get low molecular weight PPF with highly reactive and more planar trans-fumarate configuration, a meticulous synthetic procedure is needed.

Polypropylene fumarate based unsaturated prepolymers could be further polymerized in situ through crosslinking of double bonds with unsaturated monomers. The mechanical strength and biodegradation of the bone cement in vivo dependent on the main chain groups, end groups, co-reactants and crosslink density. As part of the development of biodegradable and injectable bone cement for orthopaedic applications, diethylene glycol acrylate–*n*-vinyl pyrrolidone copolymer resins [25] and poly(propylene fumarate-*co*-ethylene glycol) based bone cements were prepared and reported earlier [26]. Combination of epoxy group and acrylic and fumarate groups can offer a convenient route to a combination of thermosetting and thermoplastic properties. By crosslinking an epoxy acrylate and polypropylene fumarate oligomers a quite complex cured product that may have improved creep properties in vivo could be obtained. The cement should not undergo creep changes due to the effect of body temperature and or mechanical stress before the onset of degradation and bone growth. The present work also aims at possible development of bone cement which can set into a hard mass without the use of generally used fast activating amine accelerator, *N*, *N* dimethyl aniline and *N*, *N* dimethyl *p*-toluidine, etc. It has been found that these tertiary amines have been found to be toxic and mutagenic in host [27]. Therefore, the objective of the present paper is to develop and evaluate blends of polypropylene fumarate/phloroglucinol triglycidyl methacrylate as partially biodegradable and injectable bone cement for orthopaedic applications.

2. Materials and methods

2.1. Preparation of resins

Polypropylene fumarate (PF_1) was prepared using maleic anhydride and 1,2-propylene glycol in presence of *p*-toluene sulphonic acid catalyst as reported elsewhere [28]. The second component phloroglucinol triglycidyl methacrylate (PA_2) was also prepared and reported [29]. In short, phloroglucinol triglycidyl ether was prepared initially using phloroglucinol, epichlorohydrin, benzyl triethyl ammonium chloride and alcoholic sodium hydroxide. Then it was reacted with methacrylic acid in the presence of triethylamine catalyst. The residual triethylamine was not removed from the final oligomeric resin PA_2. The residual tertiary amine acts as a moderate activating amine for the crosslinking of PA_2 and PF_1.

2.2. characterization of bone cement components

2.2.1. Determination of acid number

The acid constituent in the resins PF_1 and PA_2 was determined by finding the acid number as per ASTM standard D 2849. One gram of the resin was dissolved in benzene/ethyl alcohol mixture (1:1 v/v) and titrated against 0.1 N alcoholic KOH using, phenolphthalein indicator. A blank titration was also carried out. The quantity of KOH required to neutralize acidic constituents in 1 g of resin was calculated as the acid number. Acid number $= [(A - B)N \times 56.1]/w$, where '$A$' and '$B$' are the volume of KOH solution required for titration of the blank and test sample, respectively.

2.2.2. Infrared spectral studies of PF_1 and PA_2

The IR spectrum of PF_1 and PA_2 was recorded using a resin smear on a NaCl window. A nicolet (Impact 410) FT-IR instrument was used.

2.3. Preparation of blends of polypropylene fumarate/phloroglucinol triglycidyl methacrylate acrylate (PF_1/PA_2)

A series of blends of varying compositions of PF_1 and PA_2 were prepared with and without HA. Blends of $PF_1:PA_2$ compositions 25:75, 50:50 and 75:25 were coded as PF_1/PA_2–A, PF_1/PA_2–B and PF_1/PA_2–C, respectively. The initiator benzoyl peroxide (3.75%) was added to initiate crosslinking of PF_1 and PA_2. Blends of $PF_1:PA_2$ having composition 50:50 were prepared with hydroxyapatite [$Ca_{10}(PO_4)_6(OH)_2$] as filler. Blends with filler HA content 2.5%, 5.0% and 10% were coded as PF_1/PA_2–D, PF_1/PA_2–E and PF_1/PA_2–F, respectively (Table 1).

2.4. Characterization of bone cement

2.4.1. Determination of setting time

The setting time of adhesive blend composites with and without HA was determined as per ISO 5833/1-1999 E Standard. The components were mixed thoroughly in presence of initiator benzoyl peroxide at 37°C and the time taken for curing the blend to solid mass was determined. The average temperature was determined by taking half the sum of ambient temperature and exothermic setting temperature. The time taken to reach the average temperature was found as setting time.

2.4.2. Studies of adhesive bonding of bones using $PF_1/PA_2/HA$

Bovine rib bones were procured from the slaughter house, cleaned well with isotonic saline solution, removed adhering muscles and fixed in 10% formaldehyde solution for 3 days. The bone samples were washed well with distilled water, and dried in oven at 60°C. The bone cement composite of PF_1/PA_2/HA (125 μm) were applied to bone joints as soon as the initiator was added and blended well with the mixture. The bonded bone joints were subjected to tensile tests as per ASTM standard as reported earlier [26]. Tensile strength was found by using an Instron Corporation, series IX automated materials testing system 7.43.00 at a crosshead speed of 5 mm/min.

Table 1
Formulation of bone cement blends

Cement	PF_1 (g)	PA_2 (g)	Initiator (g)	HA (g)
PF_1/PA_2–A	0.6	0.2	0.03	—
PF_1/PA_2–B	0.2	0.6	0.03	—
PF_1/PA_2–C	0.4	0.4	0.03	—
PF_1/PA_2–D	0.8	0.8	0.06	0.04
PF_1/PA_2–E	0.8	0.8	0.06	0.08
PF_1/PA_2–F	0.8	0.8	0.06	0.16

2.5. Studies on cured bone cement blends

2.5.1. Thermal studies of cured bone cement blends

The thermal behaviour of the cured blends was determined by thermogravimetric (TG) and differential thermal analysis (DTA). The samples were heated at 10°C to a maximum temperature of 500°C in nitrogen atmosphere using a universal VI-12 ETA thermal analyzer.

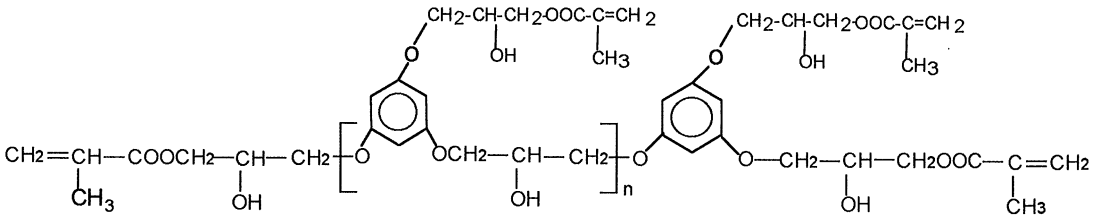
2.5.2. Studies on degradation of cured bone cement

The accelerated hydrolytic degradation of the polymers was studied by in vitro aging in various media such as phosphate buffered saline (PBS) and Ringer's solution [30]. Phosphate buffered saline was prepared as per standard procedure [26,28]. Disodium hydrogen phosphate 0.2 M (28.4 g of anhydrous Na_2HPO_4 per 1000 ml) and 0.2 M sodium dihydrogen phosphate (27.6 g of $NaH_2PO_4 \cdot H_2O$ per 1000 ml) in 0.9% saline solutions were prepared separately and mixed in the ratio 77:23 (v/v) such that pH would be 7.4. Ringer's solution (pH = 8.4) was prepared by dissolving sodium chloride (9 g), sodium hydrogen carbonate (0.42 g), calcium chloride (0.24 g) and glucose (1 g) in 1000 ml distilled water. Uniform pieces of cured bone cement blend composites were accurately weighed and immersed in different media. The accelerated aging was carried out at 70°C for 30 days. Every 5 days interval, the samples were taken out, dried and weight loss was determined. The weight loss (%) was plotted against duration of exposure and the relative aging stability was determined.

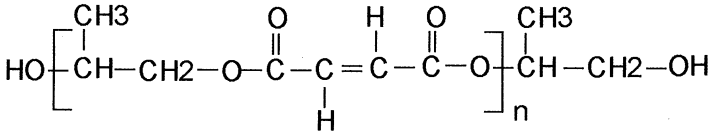
3. Results and discussion

3.1. Materials

Acrylic bone cements have been used for orthopaedic applications as two-part component [31]. The major



PA_2



PF_1

Scheme 1. Structure of blend components — phloroglucinol triglycidyl methacrylate (PA_2) and polypropylene fumarate (PF_1).

disadvantages associated with acrylic cements are high exothermic heat, shrinkage, and allergic and toxic reactions of unreacted monomers, residual initiator and accelerator, and loosening of bone–implant interface [32]. Modifications of acrylic bone cements were attempted with hydroxy ethyl methacrylate, ethylene glycol dimethacrylate and ultrahigh molecular weight polyethylene though limited in success [33]. Bisphenol A glycidyl methacrylate-bonding agent is in routine use for dental cement composite [34]. Poly(propylene fumarate) has been studied as biodegradable material for tissue engineering and orthopaedic applications. However, polypropylene fumarate–phloroglucinol triglycidyl methacrylate blend system could be a better combination giving fast setting, adhesive bonding and controlled biodegradability. Both polypropylene fumarate (PF_1) and phloroglucinol triglycidyl methacrylate (PA_2) are oligomeric viscous liquids [28,29]. The molecular structures of blend components are given in Scheme 1. Hydroxyapatite was prepared [35] and supplied as 125 μm size granules from Bioceramics Division, SCTIMST.

3.2. Characterization of cement components PF_1 and PA_2

The infrared spectrum of the oligomer phloroglucinol triglycidyl methacrylate (PA_2) is given in Fig. 1. The appearance of a strong peak at 1725.15 cm^{-1} indicates the ester group of acrylate. The presence of –OH group in the resin is also indicated with the peaks at 1409.79 and 3431.41 cm^{-1}. The aromatic ring of phloroglucinol is seen with sharp peak at 1599.74 cm^{-1}. The residual epoxide group of resin is observed with the peak at 1295.33 cm^{-1}. The IR spectral responses of polypropylene fumarate (PF_1) reveal presence of unsaturation (C=C) in fumarate unit at 1643.32 cm^{-1} and ester (C=O) unit at 1741.13 cm^{-1} (Fig. 1). The peaks around 2983 and 2944 cm^{-1} are due to aliphatic (C–H) groups in the two spectra. The more intense peak at 3499.13 cm^{-1} indicates PF_1 has more hydroxyl end groups suggesting the liquid resin in the oligomeric form. The acid value of oligomers PF_1 and PA_2 are 156.519 and 51.612, respectively. This indicates that PF_1 resin is end capped with acid group whereas PA_2 is end capped with both acid and hydroxyl units.

3.3. Preparation and characterization of blends of cement

Determination of setting time revealed the extent of the fast crosslinking reaction between the unsaturated double bonds of PF_1 and PA_2. The data of setting time for blends of PF_1 and PA_2 are given in Table 2 and Fig. 2. The curve A represents the setting PF_1 and PA_2 without HA. Lesser the percentage of PF_1 and the higher the percentage of PA_2, the lower is the setting time. The crosslinking reaction of PF_1 and PA_2 resulted in a crosslinked product comprising of PF_1 and PA_2 in a random fashion (Scheme 2). The crosslinking took place through C=C bond of PF_1 and PA_2 as the IR spectrum of crosslinked polypropylene fumarate showed absence of peak around 1644 cm^{-1} and around 982 cm^{-1} (due to –C–H bending of –CH=CH– of units) as reported earlier [26]. The slow crosslinking reaction in blends having higher percentage of PF_1 is attributed to the presence of maleate ester in PF_1 in addition to fumarate units. The maleate units were less reactive than vinyl, acrylic, and fumarate units [34]. The setting characteristics of blends of PF_1 and PA_2 with HA are given Table 2 and curve

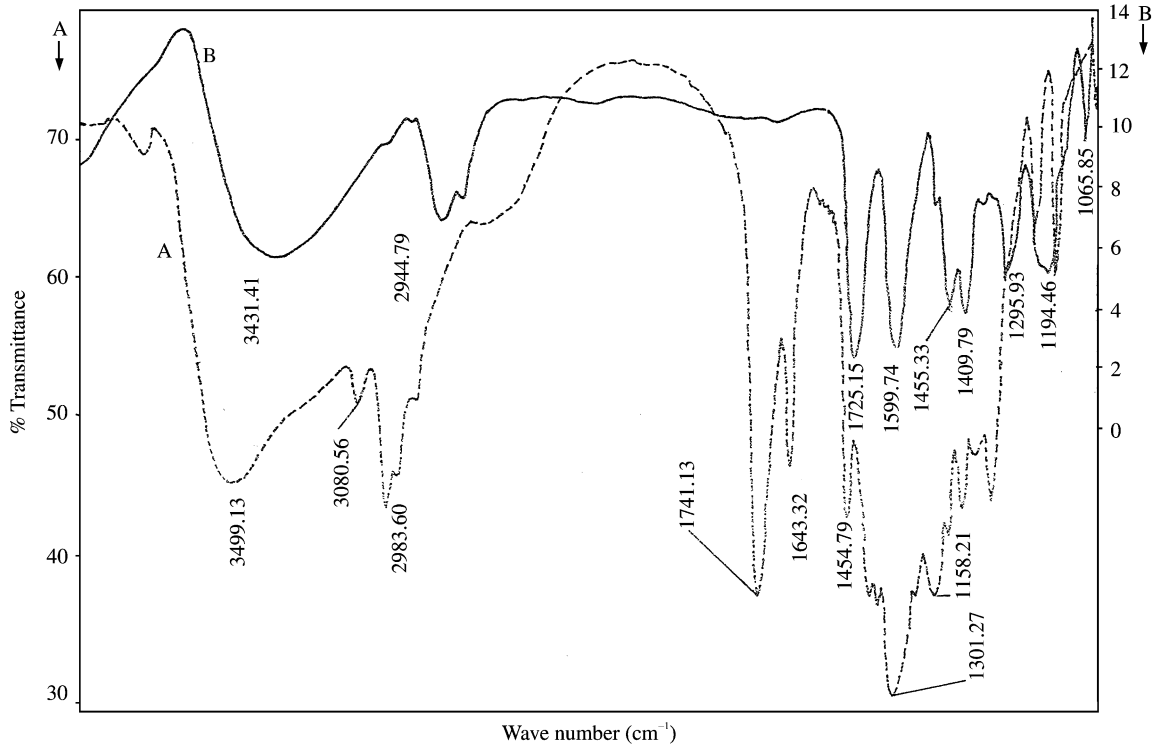


Fig. 1. FT-IR Spectrum of bone cement components: A — PF_1, B — PA_2.

B of Fig. 2. The HA content has an important role in the setting reaction. An optimum amount (5%) of HA gave lower setting in comparison with the composition having the lowest and highest amount of HA. The basicity of HA influenced the setting reaction. Blend having 2.5% HA may be more acidic and that with 10% HA may be more basic, which hinders crosslinking. Therefore, an optimum 5% HA enabled the polymer composite to have comparatively lower setting time.

3.4. Determination of adhesive bonding strength in bone

Bone cement comprising PF_1 (47.06 ppw), PA_2 (47.06 ppw), benzoyl peroxide (3.53 ppw) and HA (2.35 ppw) were blended and used for bonding bovine rib bones. The tensile test data revealed that the adhesive bonding has strength 1.11 kPa and low percentage strain. The toughness was found to be 0.0688 kPa, which indicated that the bonded bone cement is brittle in nature.

3.5. Thermal studies of cured bone cement

The thermal studies revealed the degree of curing in a cured product as well as the thermal stability of the cured product. The data on differential thermal analyses of the cured bone cement with and without HA filler are given in Table 3. The DTA curves of bone cement without HA showed two mild transitions at 55°C and 102°C. These peaks were somewhat well defined in the case of PF_1/PA_2–A, which could be due to the presence of homopolymers by the polymerization of PA_2 and PF_1 in the blend. The high setting time (230 min) also supported this possibility. The DTA curve of PF_1/PA_2–A showed exothermic peak at 124.65°C which was due to the crosslinking reaction between the remaining PF_1 and PA_2 components (Table 2). PF_1/PA_2–B exhibited the exothermic peak at 118.91°C, which could be due to individual crosslinking of PA_2 due to the presence of higher quantity of fast reactive phloroglucinol component. The lesser area under this peak for PF_1/PA_2–B

Table 2
Setting and curing characteristics of bone cements

Cement	Setting time at 37°C (min)	Crosslinking temperature[a] (°C)
PF_1/PA_2–A	230	124.65
PF_1/PA_2–B	21	118.91
PF_1/PA_2–C	40	138.26
PF_1/PA_2–D	133	130.36
PF_1/PA_2–E	101	129.95
PF_1/PA_2–F	240	122.03

[a]From DTA analyses.

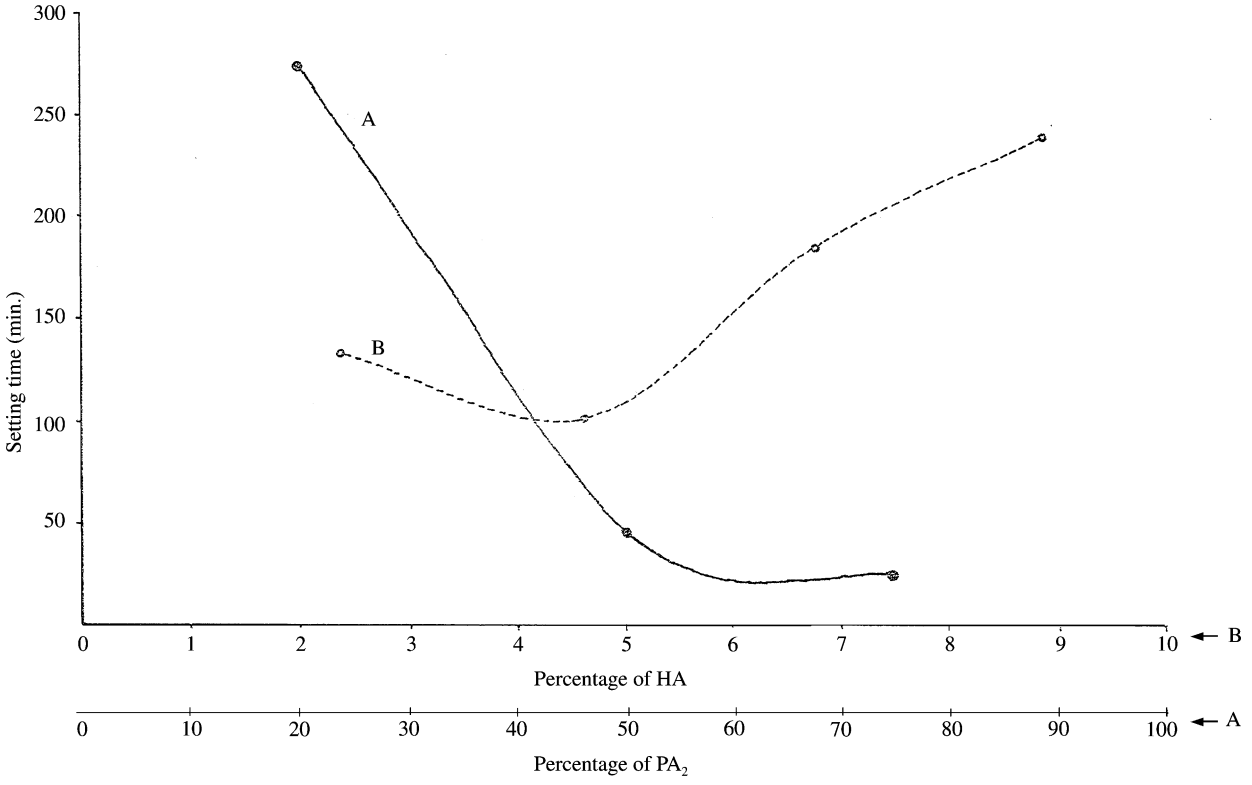
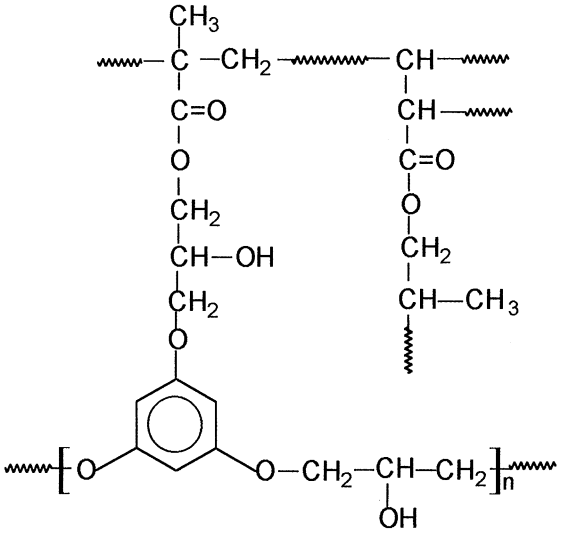


Fig. 2. Setting characteristics of bone cement blends.



Scheme 2. Crosslinked structure of PF_1/PA_2 blend.

unlike PF_1/PA_2–A also indicated the lesser degree of individual crosslinking reaction. The lower temperature transitions were not clear in bone cements with filler. This may be due to the absence of higher quantities of individual homopolymer. The exothermic peak for crosslinking reactions also appeared as mild in the HA filled bone cement. Comparing the DTA curves of PF_1/PA_2–C (without HA) and PF_1/PA_2–D (with HA), a small exothermic peak appeared at 130.36°C for the latter. However, PF_1/PA_2–F exhibited a well-defined exothermic peak at 122.03°C for crosslinking which indicated slow speed of crosslinking reaction in the presence of higher amount of HA. There was no clear melting endothermic peak in DTA curves, owing to the thermosetting-like amorphous nature of blend components. The decomposition peak for the highest decomposition temperature (T_{d_4} or T_{d_3}) of unfilled and filled cured bone

Table 3
Thermal decomposition characteristics of cured bone cement blends

Blend system	Temperature[a] (°C) at weight loss of			Decomposition temperatures[b] (°C)			
	10%	50%	70%	T_{d_1}	T_{d_2}	T_{d_3}	T_{d_4}
PF_1/PA_2–A	210	370	400	195.0	294.8	341.0	374.9
PF_1/PA_2–B	240	350	455	192.0	—	342.2	—
PF_1/PA_2–C	230	360	400	196.0	292.5	340.5	381.5
PF_1/PA_2–D	250	375	440	200.0	300.2	371.1	—
PF_1/PA_2–E	230	380	460	199.5	301.1	376.6	—
PF_1/PA_2–F	225	390	450	200.0	298.0	376.4	—

[a] From TGA curves.
[b] From DTA curves.

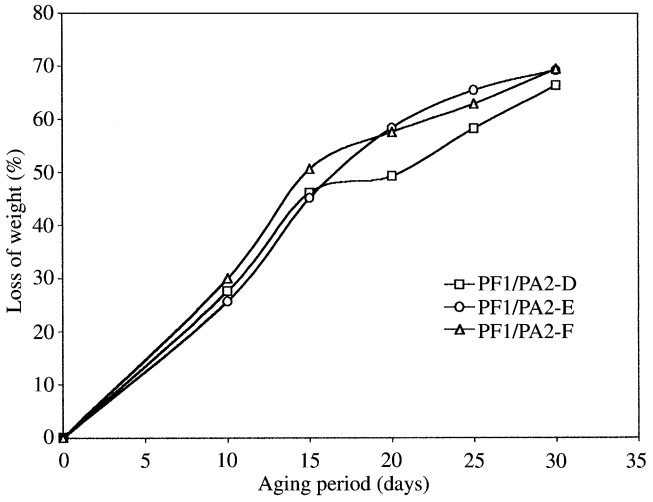


Fig. 3. Degradation of bone cement blend in phosphate buffered saline.

cements were clear at 373.94°C for PF_1/PA_2–A with higher percentage of PF_1 and at 342.17°C for PF_1/PA_2–B with higher percentage of PA_2 (Table 3). For PF_1/PA_2–C two equally intense decomposition peaks appeared 340.54°C and 381.46°C. For cured bone cements with HA, the peak appeared at 340.54°C for PF_1/PA_2–C disappeared. Moreover, the decomposition peak at 381.46°C for PF_1/PA_2–C shifted to lower temperature with the addition of HA (Table 3). Weight loss (70%) occured at higher temperature in filled bone cements when compared to the weight loss of unfilled PF_1/PA_2–C.

3.6. Studies on in vitro biodegradation

A biodegradable bone-filling material should be degraded about the same rate of the new bone tissue formation. The profile of accelerated hydrolytic degradation in vitro media is shown in Figs. 3 and 4. The weight loss during aging clearly indicated the hydrolytic degradation of cured bone cements. The pH (7.4) of the solution decreased appreciably especially in PBS medium and reached a constant value (Table 4). This revealed degradation and association of calcium ions with negatively charged units of PF_1 and PA_2 backbone. This acted as a buffer and arrested further lowering of pH. This ionic association slowed down the hydrolytic degradation after an initial period of 17 days. The higher decrease of pH ($\approx 40\%$) observed in PBS media than in Ringer's solution ($\approx 20\%$) suggested higher degree of ionic association in Ringer's solution than in PBS. Therefore, more degradation was noticed in PBS medium (Figs. 3 and 4). After 30 days of aging more than 50% of degradation was observed. HA granules can neutralize the acidic degradation products of polyester and further reduce the auto-catalytic degradation of the polymer phase. The studies revealed that the present bone cement was partially biodegradable having proper degradation rate that would degrade the polymer matrix within 2–3 months.

4. Conclusion

Studies on polypropylene fumarate/phloroglucinol triglycidyl methacrylate oligomeric blend have revealed

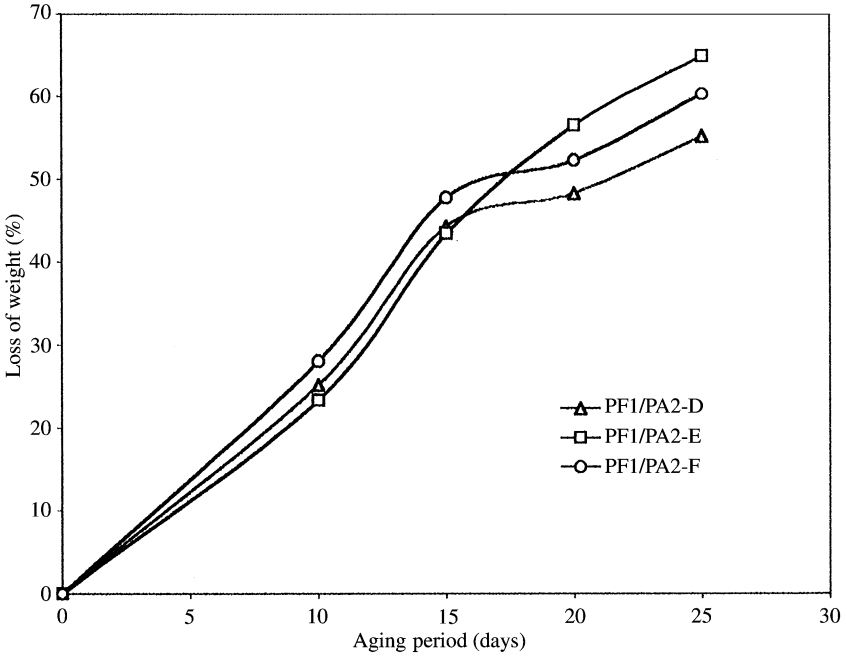


Fig. 4. Degradation of bone cement blend in Ringer's solution.

Table 4
In vitro biodegradation of cements in hydrolytic media at 70°C for 30 days

Cement	Ringer's solution (pH 8.4)		Phosphate buffered saline (pH 7.4)	
	Maximum weight loss (%)	pH of medium after aging	Maximum weight loss (%)	pH of medium after aging
PF_1/PA_2–D	59.56	6.92	66.08	4.64
PF_1/PA_2–E	66.44	6.67	69.0	5.42
PF_1/PA_2–F	63.79	6.79	69.20	4.40

that the blend composition influences the setting polymerization reaction in the presence of free radical catalyst (benzoyl peroxide) at 37°C. The higher the percentage of phloroglucinol triglycidyl methacrylate the lesser the setting time. An optimum setting time could be arrived with 50 : 50 blend composition of the two oligomers. Addition of HA increases the setting time of the blend. Though aromatic amine activator could be used to reduce the setting time further, it is wise to avoid and use safe formulation. Composite cement of 50 : 50 blends prepared with HA granules of particle size 125 μm bind bovine rib bones. The tensile strength of this adhesive bond was found to be 1.11 kPa. The toughness and elongation data suggested that the cured cement is brittle in nature due to insufficient macromolecular chain growth. The thermal studies suggest the onset of crosslinking reaction in the cured blends if the blend is heated. The absence of melting endotherm shows the thermosetting-like amorphous nature of blend system, which can resist changes in creep properties. The in vitro biodegradation studies of HA added blends reveal possible association of calcium ions with negatively charged units of degrading polymer chain resulting slow down of degradation. Relatively slow degradation was observed in Ringer's solution. The study revealed the potential use of polypropylene fumarate/phloroglucinol triglycidyl methacrylate as an in situ polymerizable, injectable and partially biodegradable orthopaedic cement.

Acknowledgements

The authors gratefully acknowledge Dr. P.R. Hari Krishna Varma for providing hydroxyapatite granules and Mr. P.R. Hari and Ms. C. Radhakumari for their help rendered during the evaluation of materials. We are also grateful to Head, Biomedical Technology Wing, and Director, Sree Chitra Tirunal Institute for Medical Sciences and Technology for providing the facilities to carry out the work.

References

[1] Brauer GM, Hass SS, Dickson GA. Analysis of the ingredients and determination of the residual components of acrylic bone cements. J Biomed Mater Res 1977;11:577.
[2] Kury RP, Hartz MJ. Effect of molecular weight on the fracture surface energy of PMMA in cleavage. J Mater Sci 1976;11:1475.
[3] Saha S, Pal S. Mechanical properties of bone cement — a review. J Biomed Mater Res 1984;18:435.
[4] Lee ACJ. Some properties of PMMA with reference to its use in orthopeaedic surgery. Clin Orthop Relat Res 1973;95:287.
[5] Bonefield W, Grynapas MD, Tully AE, Bowman AJ, Abram J. Hydroxyapatite reinforced polyethylene — a mechanically compatible implant material for bone replacement. Biomaterials 1981;2:185.
[6] Hollinger JO. Preliminary report on the osteogenic potential of a biodegradable copolymer of polylactide (PLA) and polyglycolide(PGA). J Biomed Mater Res 1983;17:71.
[7] Verheyen CCPM, De Wijin JR, Van Blitterswijk CA, De Groot K. Evaluation of hydroxyapatite/poly(L-lactide) composites: mechanical behaviour. J Biomed Mater Res 1992;26:1277.
[8] Boeree NR, Dove J, Cooper JJ, Knowles J, Hasting GW. Development of a degradable composite for orthopaedic use: mechanical evaluation of a hydroxyapatite–polyhydroxybutyrate composite material. Biomaterials 1993;14:793.
[9] Thomson RC, Yaszemski MJ, Powers JM, Mikos AG. Poly(α-hydroxy ester) foams for bone regeneration. Biomaterials 1998;19:1935–43.
[10] Ural E, Kesenci K, Fambri C, Migliaresi C, Piskin E. Poly(D,L-lactide/ε-caprolactone)/hydroxyapatite composites. Biomaterials 2000;20:2147–54.
[11] Matschinske U, Tappe A, Sandner B. Zahn Mund Kieferheilkd Zentralbl 1989;77:783–9.
[12] Gerhart TN, Roux RD. Antibiotic release from an experimental biodegradable bone cement. J Orthop Res 1988;6:585–92.
[13] Suggs LJ, Krishnan RS, Garcia CA, Peter SJ, Anderson JM, Mikos AG. In vitro and in vivo degradation of poly(propylene fumarate-*co*-ethylene glycol) hydrogels. J Biomed Mater Res 1998;42:312–20.
[14] Suggs LJ, West JL, Mikos AG. Platelet adhesion on a bioresorbable poly(propylene fumarate-*co*-ethylene glycol) copolymer. Biomaterials 1999;20:683–90.
[15] Lewandrowski K, Cattaneo MV, Gresser JD, Wise DL, White RL, Bonassar L, Trantolo DJ. Effect of a poly(propylene fumarate) foarming cement on the healing of bone defects. Tissue Eng 1999;4:305–16.
[16] Sanderson JE, Bone replacement and repair putty material from unsaturated polyester resin and vinyl pyrrolidone. US Patent 4,722,948, 1988. p. 1–14.
[17] Gerhart TN, Hayes WC. Bioerodible implant composition. US Patent 4,843,112, 1989. p. 1–16.

[18] Gresser JD, Hsu SH, Nagaoka H, Lyons CM, Nirratko DP, Wise DL, Barabino GA, Trantolo DJ. Analysis of a vinyl pyrrolidone/PPF resorbable bone cement. J Biomed Mater Res 1995;29:1241–7.
[19] Domb AJ. Poly(propylene glycol fumarate) composition for biomedical application. US Patent 4,888,413, 1989. p. 1–32.
[20] Domb AJ, Laurencin CT, Israll O, Gerhart TN, Langer R. The formation of propylene fumarate oligomers for use in bioerodible bone cement composites. J Polym Sci Polym Chem A 1990;28:973–85.
[21] Yaszemki MJ, Mikos AJ, Payne RG, Hayes WC. Biodegradable polymer composites for temporary replacement of trabecular bone. In: Mikos AG, Murphy R, Bernstein H, Peppas NA, editors. The effect of polymer molecular weight on composite strength and modulus in biomaterials for drug and cell delivery, vol. 331. Pittsburgh: Materials Research Society, 1994. p. 251–6.
[22] Peter SJ, Yaszemski MJ, Suggs LJ, Payne RG, Langer R, Hayes WC, Unroe MR, Alemany LB, Engel PS, Mikos AG. Characterization of partially saturated poly(propylene fumarate) for orthopaedic applications. J Biomater Sci Polym Ed 1997;8(11):893–904.
[23] Szmerczanyi IV, Marcos LK, Greger K, Makaybodi E. Investigation of maleate–fumarate isomerisation during esterification with various glycols. J Polym Sci 1961;53:240–8.
[24] Andreis M, Veksli Z, Meic Z. Polymer 1986;26:1099.
[25] Celin D, Vinoy Thomas, Jayabalan M. Diethylene glycol acrylate–*n*-vinyl pyrrolidone copolymer resins for bone cement applications. Indian J Eng Mater Sci 2000;71:160–6.
[26] Jayabalan M, Vinoy Thomas, Sreelatha PK. Studies poly(propylene fumarate-*co*-ethylene glycol) based bone cements. Biomed Mater Eng 2000;10:57–71.
[27] Davindenko N, Cohen ME, Diaz JM, Santre R. Development of new photopolymerisable dental sealants. J Biomater Sci Polymer Ed 1998;9:891–903.
[28] Anitha NS, Vinoy Thomas, Jayabalan M. Studies on polypropylene fumarate–*n*-vinyl pyrrolidone copolymer based bone cement: setting and in vitro biodegradation. J Indian Inst Sci 1999; 79:431–42.
[29] Jose Jacob, Jayabalan M. Studies on biodegradable polymers of phloroglucinol/triglycidyl acrylates. Proceedings of National Seminar on Polymers for New Millenium, Madras, India, 1999.
[30] Jayabalan M, Lizymol PP, Vinoy Thomas. Synthesis of hydrolytically stable low elastic modulus polyurethane–urea for biomedical applications. Polym Int 2000;49:88–92.
[31] Charnley J. Acrylic bone cements in orthopaedic surgery. Edinburgh, London, 1970. p. 107.
[32] Park JB, Lakes RS. Biomaterials — an introduction. New York, 1992. p. 346.
[33] Yang JM. Study of polymerisation of acrylic bone cement: effect of HEMA and EGDMA. J Biomed Mater Res 1998;43:54–61.
[34] Melvire HM, Burnett GM. J Polym Sci 1966;13:417.
[35] Varma HK, Sivakumar R. Preparation and characterization of free flowing hydroxyapatite powders. Phosphorous Res Bull 1996;6:35–8.