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Synthesis, formulation and evaluation of novel zinc-calcium phosphate-based adhesive resin composite cement

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Abstract

Three novel adhesive oligomers having carboxylic acid and methacrylate groups were synthesized, characterized and used to formulate composite bone cements with newly synthesized zinc-calcium-silicate phosphate. The optimal formulation was determined based on types of oligomer, oligomer/diluent ratio, initiator concentration, and filler level using compressive strength (CS) and curing time (CT) as screening tools. Shrinkage, exotherm and aging of the formed composite cements were also evaluated. Results show that the experimental cement was 186% higher in CS, 16% higher in diametral tensile strength, similar in flexural strength, 56% less in exotherm and 64% less in shrinkage, as compared to conventional polymethylmethacrylate cement. The optimal concentrations for initiators were found to be 1.5% (by weight) for both benzoyl peroxide and *N*,*N*′-dimethyl-*p*-toluidine. With increasing initiator concentration, diluent content and zinc oxide content in the cement formulation, CS of the cement increased but curing time decreased. Shrinkage and exotherm of the cement decreased with increasing filler level. CS was not proportional to an increase of filler level and CT increased with an increase of filler level. During aging, the cement showed an increase of strength over 24 h and then no change for over nine months. It appears that this novel cement may be a potential candidate for orthopedic restoration if its biological performance is good and the formulation is optimized.

Keywords: Zinc-calcium-silica phosphate; Composite bone cement; Adhesive oligomer; Synthesis; Formulation; Evaluation

1. Introduction

Traditional polymethylmethacrylate (PMMA) cements have been used for cementing prosthetic hips for over 40 years, since 1960s [1]. Although the use of PMMA has enabled the successful rehabilitation of many elderly patients with a relatively short life expectancy, its inherently poor biocompatibility, lack of adhesion to bone, high exotherm and large shrinkage are major factors in the loosening and subsequent failure of hip prostheses [1,2]. Therefore, there is a strong need to develop new bone cements as an alternative or a substitute for PMMA. Considerable efforts have been made in developing new bone cements since 1990s [3–9]. Major research has focused on development of resin composite, glass-ionomer and calcium phosphate bone cements, such as hydroxyapatite–BisGMA/TEGDMA composites [3,4], bioactive glass–ceramics–BisGMA/TEGDMA

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composites [5,6], calcium phosphate/water cements [7], and glass-ionomers [8,9].

Resin composites have been successfully used in dentistry as esthetic filling restoratives for decades. They are very well known for their high mechanical strengths. However, lack of direct chemical adhesion to tooth and slow degradation make them not very useful without applying extra bonding agents [10]. BisGMA/TEGDMA resin based systems are major components being studied in current bioactive composite bone cements [2–6]. These bioactive bone cements exhibit many advantages such as high bioactivity to bone tissue and high mechanical strengths. However, degradation and low durability are still major concerns for these materials for applications in high stress-bearing sites [11–14]. Few reports on long-term mechanical behavior of these cements are available.

Glass-ionomer cements (GICs) have shown better adhesion, higher compressive strength (CS), less shrinkage and lower exotherm, as compared to PMMA [10]. However, their tensile and flexural strengths are relatively low compared to composite resins and one of the key elements in these cements, the aluminum ion, has been reported to be not suitable to orthopedics due to its potential cytotoxicity to neurotissues [15].

Compomers, a composite resin/glass-ionomer hybrid restorative, combine the desirable properties of GICs and composite resins. They have only been used in dentistry for about 10 years and are still being developed and evaluated [16–18]. They behave more like composite resins than GICs in mechanical strengths but their aging behavior is closer to GICs. Compomers are a one-paste and light-curable composite composed of an ion-leachable glass, usually a calcium–aluminum-fluorosilicate glass, embedded in a matrix of polymerizable monomers and bifunctional oligomers carrying both carboxylic acid functionality and carbon–carbon double bonds [16–18]. The primary setting reaction involved is free-radical polymerization initiated by light, followed by a slow post acid–base reaction initiated by water invasion. Therefore, compomers are able to release fluoride, have less degradation and potentially bond to tooth structure. However, compomers still contain aluminum which is a toxin to neurotissues [15].

The zinc cation (Zn^{++}) has been found to be very osteoconductive and to stimulate osteogenesis [19,20]. Since zinc oxide is able to form salt-bridges between zinc cations and carboxyl anions, a simply blended mixture of zinc oxide (ZnO)/hydroxyapatite (HA) was used to form bone cement with polyacrylic acid aqueous solution [21]. However, the low CS (less than 100 MPa) and short working time (less than 1 min) made the cement not very attractive to orthopedic applications [22]. Zinc polycarboxylate cements have also been used in dentistry for many years but their strengths (CS = 57–99 MPa and DTS = 3.6–6.3 MPa) still cannot compete with those of GICs (CS = 93–226 MPa and DTS = 5.3–24 MPa) [10].

In order to formulate non-aluminum-containing cement with improved mechanical strengths and extensive working time, we proposed to prepare sintered zinc-calcium-silicate phosphate glass and formulate it with the bifunctional oligomers having carboxylic acid and methacrylate groups, which may provide a route to form a cement with better bioactivity, less cytotoxicity and higher mechanical strengths.

The objective of this study was to synthesize and characterize novel bifunctional oligomers, to use them to formulate a novel bone cement incorporating a newly synthesized biocompatible and reactive glass containing zinc-calcium-silicate phosphate, and to evaluate the properties of the formed cements.

2. Experimental

2.1. Materials

Calcium phosphate tribasic [$Ca_{10}(OH)_2(PO_4)_6$ or HA] and silicon (IV) oxide (SiO_2) were used as received from Alfa Aesar Co. Pyridine, butylated hydroxy toluene (BHT), zinc oxide (ZnO), hydrochloric acid (HCl, 37%), γ-(trimethoxysilyl)propyl methacrylate (TMSPMA), triethylene glycol dimethacrylate (TEGDMA), tetrahydrofuran (THF), methanol, petroleum ether, and diethyl ether were used as received from Acros/Fisher Scientific Inc. Glycerol dimethacrylate (GDMA), 1,2,4,5-benzenetetracarboxylic dianhydride (BTCDA), 3,3′,4,4′-benzophenonetetracarboxylic dianhydride (BPTCDA), 4,4′-(4,4′-isopropylidenediphenoxy)-bis(phthalic anhydride) (IPDPBisPA), benzoyl peroxide (BPO), *N*,*N*′-dimethyl-*p*-toluidine (DMT), strontium fluoride (SrF_2), and anhydrous magnesium sulfate ($MgSO_4$) were used as received from Aldrich Chemical Co. without further purifications. Orthoset® 3 bone cement (PMMA) was supplied by Wright Medical Technology, Inc. EB-40 (tetraacrylate monomer) and trimethylolpropane trimethacrylate (TMPTMA) were supplied by UCB Chemical Corp.

2.2. Synthesis and preparation

2.2.1. Synthesis and characterization

To a three-neck flask containing 0.1 mole of dianhydride-containing monomer, 0.8 g of pyridine and 0.4 g of BHT in 180 ml dry THF, a solution of 0.22 mole of GDMA in 80 ml dry THF were added. The mixture was stirred under a nitrogen blanket at 50–60 °C for two days. Fourier transform-infrared (FT-IR) spectroscopy was used to monitor the reaction. After the reaction was completed, the mixture was precipitated with diethyl ether/petroleum ether. The obtained oil was dissolved in diethyl ether, filtered, and washed with 4% HCl aqueous solution followed by washing with distilled water three times. The solution was dried with anhydrous $MgSO_4$ and the solvent was removed using a rotary evaporator. The oily product was stored at room temperature prior to use.

The synthesized oligomers were identified using FT-IR and proton nuclear magnetic resonance (^{1}HNMR) spectroscopy. The FT-IR spectra were obtained with a FT-IR spectrophotometer (Mattson Research Series FT/IR 1000) and ^{1}HNMR spectra were obtained on a FT-300 MHz Bruker ARX-300 spectrophotometer using deuterated methyl sulfoxide as solvent.

2.2.2. Synthesis of glass fillers

A glass mixture comprising HA, SrF_2, ZnO and SiO_2 in a ratio of 35/6/47/12 or 47/6/35/12 was thoroughly mixed using a vortex vibrator, sintered at 1400 °C for 3 h, and then quenched in the air [23]. The sintered filler was ground again in the ball mill and sieved to obtain fine particles having a size equal to or less than 63 μm. The sieved fillers were then treated with silane (TMSPMA)–methanol solution (10% by weight), followed by heat-treatment at 110 °C.

2.2.3. Preparation of composite cements

A two-paste system (A and B) was used to formulate the cements. Paste A and Paste B were prepared by mixing oligomer, diluent and glass filler with BPO (0.5–2%, by weight) and with DMT (0.5–2%), respectively. A filler level at 75% (by weight) and a resin composition at 50/50 oligomer/diluent (by weight) were used throughout the study unless specified. The specimens were made by mixing an equal weight of Paste A and Paste B thoroughly before being placed into the molds.

2.3. Evaluation

2.3.1. Strength measurement

Specimens were mixed and fabricated at room temperature, according to the published protocol [24]. Briefly, the cylindrical specimens were prepared in glass tubing with dimensions of 4 mm in diameter by 8 mm in length and 4 mm in diameter and 2 mm in thickness for compressive (CS) and diametral tensile strength (DTS) tests, respectively. The specimens for the flexural strength (FS) test were prepared using a rectangular Teflon mold with dimensions of 3 mm in width by 3 mm in depth by 25 mm in length. The specimens were removed from molds after 15–30 min, and conditioned in distilled water at 37 ± 2 °C for 24 h, prior to testing.

Testing of specimens was performed on a hydraulic mechanical testing machine (Model 858 Mini Bionix, MTS Systems Corp., Eden Prairie, MN), with a cross-head speed of 1 mm/min for CS, DTS and FS measurements. The FS test was performed in three-point bending, with a span of 20 mm between supports. The sample sizes were $n = 6$–8 for each test.

CS was determined using an equation of $\mathrm{CS} = P/\pi r^2$, where P is the load at fracture and r the radius of the cylinder, and DTS was determined from the relationship $\mathrm{DTS} = 2P/\pi dt$, where P is the load at fracture, d the diameter of the cylinder and t the thickness of the cylinder. The flexural strength in three-point bending was obtained using the expression $\mathrm{FS} = 3Pl/2bd^2$, where P the load at fracture, l the distance between the two supports, b the breadth of the specimen, and d the depth of the specimen.

2.3.2. Estimates of curing time

A metal rod was used to evaluate the curing time (or working time) [25]. The rod was inserted into the center of mixture of the cement, which was mixed and packed into a-two-end opened glass tubing with diameter of 4 mm. Curing time was taken as the period from which the mixing process was initiated to the moment at which the metal rod could not be moved by hand.

2.3.3. Exotherm measurement

The heat generated from the setting reaction of the cement was measured with a slightly modified ASTM F-451 procedure. Briefly, the cement paste was placed in a cylindrical Teflon mold with dimensions of 30 mm in diameter by 6 mm in height and covered with a Teflon plunger having holes for allowing the excess cement to escape. A digital thermocouple (Fisher Scientific, Springfield, NJ) was inserted in the center of the cement and used to record the temperature change. The peak temperature was defined as an exotherm.

2.3.4. Determination of polymerization shrinkage

The polymerization shrinkage was determined using an equation of $\%\ \mathrm{Shrinkage} = (1 - d_{\mathrm{uncured}}/d_{\mathrm{cured}}) \times 100$, where d_{cured} = density of cured cement and d_{uncured} = density of uncured cement [26]. The densities of the uncured and cured cements were determined by weighing the cement paste injected from a calibrated syringe and weighing the cured cylindrical specimens whose volumes were measured in a calibrated buret, respectively. The volumes of both PMMA and experimental cement were measured in water and hexane, respectively. The mean values were averaged from three readings.

2.3.5. Viscosity determination

The viscosity of the liquid formulated with oligomer and diluent was determined at 23 °C using a programmable cone/plate viscometer (RVDV-II + CP, Brookfield Eng. Lab. Inc., MA, USA), as described elsewhere [28].

2.3.6. Statistical analysis

One-way analysis of variance (ANOVA) with the *post hoc* Tukey–Kramer multiple range test was used to determine significant differences of strengths among the materials in each group. A level of $\alpha = 0.05$ was used for statistical significance.

3. Results and discussion

3.1. Synthesis and characterization

Three oligomers having both carboxylic acid and methacrylate groups were synthesized. Their structures and schematic diagram for the synthesis are shown in Fig. 1. After purification, the yields for BTTEMA (an adduct of BTCDA and GDMA), BPTEMA (an adduct of BPTCDA and GDMA), and IPTEMA (an adduct of IPDPBisPA and GDMA) were determined with values of 68%, 80% and 78%, respectively.

The FT-IR spectra for BTTEMA, BPTEMA and IPTEMA are shown in Fig. 2. All spectra show broad peaks between 2400 and 3700 cm^{-1} for the hydroxyl group on carboxylic acid and medium strong peak at 3000 cm^{-1} (CH stretch) for methylene and methyl groups, a strong and sharp peak at around 1723 cm^{-1} for carbonyl (ester and carboxyl) groups and a characteristic peak at 1637 cm^{-1} for carbon–carbon double bond on methacrylate. Furthermore, all spectra show medium-weak peaks at 1400–1600 cm^{-1} for aromatic C=C stretch, medium-weak peaks at 1100–1250 cm^{-1} (C–O stretch) for ester groups and weak peaks at around 900 cm^{-1} (O–H oop) for ester groups.

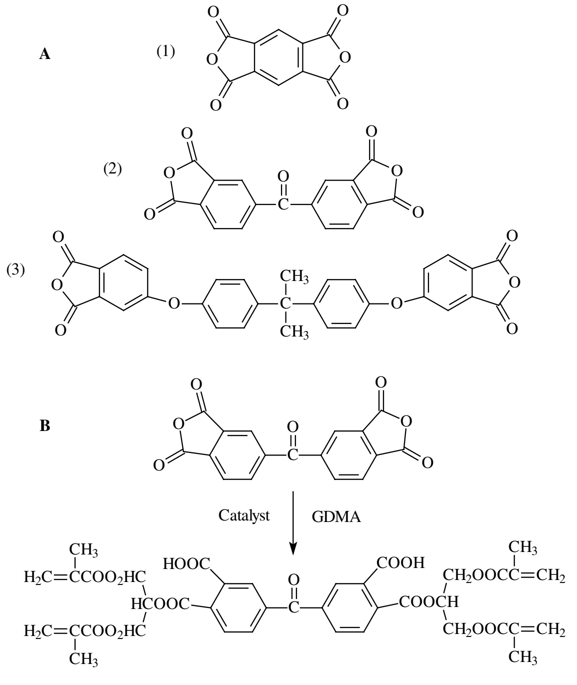

Fig. 1. Structure and synthesis scheme: A. Structures of three dianhydrides: 1,2,4,5-benzenetetracarboxylic dianhydride (BTCDA); 3,3′,4,4′-benzophenonetetracarboxylic dianhydride (BPTCDA); 4′-(4,4′-isopropylidenediphenoxy)-bis(phthalic anhydride) (IPDPBisPA); B. Synthesis scheme for BPTEMA (an adduct of BPTCDA and GDMA).

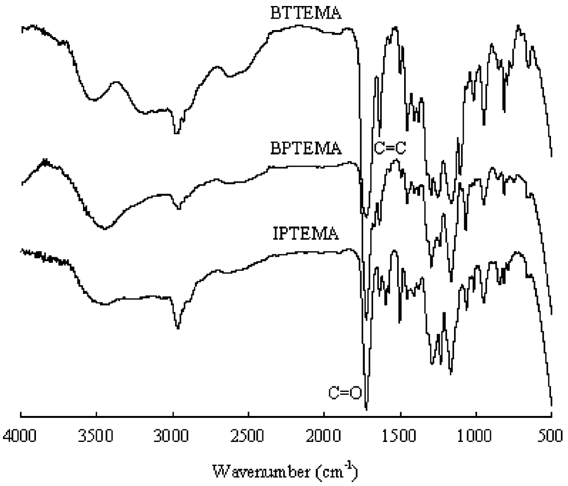

Fig. 2. FT-IR spectra for BTTEMA, BPTEMA and IPTEMA.

Fig. 3 shows the ^{1}HNMR chemical shifts for BTCDA, GDMA and BTTEMA. The chemical shifts for the phenyl ring on BTCDA were found at 7.9, 8.3 and 8.7 ppm on the upper curve. The chemical shifts of GDMA were found at 6.0 and 5.7 (CH_2=), 5.4 and 5.1 (OH), 4.1 (CH_2), 3.6 and 3.4 (CH), and 1.9 ppm (CH_3) on the middle curve. The chemical shifts for the product BTTEMA were found at 8.2–7.9 (ArH), 6.1 and 5.7 (CH_2=), 4.5 (CH_2), 3.6 (CH), and 1.9 ppm (CH_3) on the

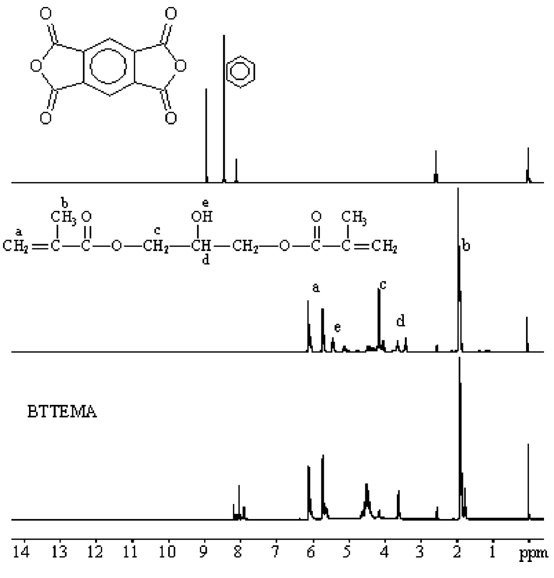

Fig. 3. ^{1}HNMR spectra for GDMA, BTCDA and BTTEMA.

lower curve. Although the characteristic peak for carboxylic acid was seen in FT-IR (Fig. 2), the proton for carboxylic acid on BTTEMA was not observed in NMR, which may be attributed to relatively weak peak of the carboxyl acid compared to the others.

3.2. Formulation and evaluation of cements

As we know, commercially available resin composites are made of BisGMA and TEGDMA (50/50, by weight) and 70–80% aluminosilicate fillers, which exhibit fairly strong mechanical strengths, especially compressive (CS) and flexural strengths (FS) [3–6,10,14]. However, they lack of durability in strength due to relatively weak interfacial bonding between glass fillers and resin matrix [11–14,27]. The beauty shown by dental glass-ionomer cements and compomers are their increased or unchanged mechanical strengths [16–18,28–30]. The cements we developed in this study also demonstrated similar behavior (see effect of aging below). The key is attributed to the fact that we have incorporated reactive zinc-containing fillers and carboxyl-containing oligomers into the cements. In this study, we have utilized the simple principle of experimental design to formulate the cements. The optimal formulation based on the newly synthesized glass fillers and oligomers was determined using the following procedures: selection of the optimal oligomer and filler; determination of the optimal initiator concentration; selection of the optimal ratio of oligomer/diluent and combinations of diluent; and determining the optimal glass filler level. Basically, we used CS and curing time (CT) as screening tools to formulate the cements.

3.2.1. Selection of optimal oligomer and filler

As shown in Table 1, the cement A formulated with BTTEMA/TEGDMA showed the highest CS, followed by C (IPTEMA/TEGDMA) and B (BPTEMA/TEGDMA), where B and C were not significantly different from each other. The relatively low molecular weight of BTTEMA may contribute to a higher CS of A. A and D are cements with different glass compositions where the former contained more zinc oxide whereas the latter contained more hydroxyapatite. The zinc cation is more basic and reactive than the calcium cation when it contacts with the oligomer with carboxylic acid so that A exhibited higher CS.

3.2.2. Determination of optimal concentration of initiator

It is important to use suitable amount of initiators in the cement system. Neither too low nor too high amounts of initiators are favorable to the polymerization [29,31]. Insufficient initiator will cause insufficient initiation and excessive initiator will lead to a lower MW of polymer network formation and act as a plasticizer. The results in the study showed that both BPO and DMT at 1.5% demonstrated higher CS (235.5 MPa) and reasonable CT (5.5 min), as shown in Fig. 4. The cements with both 1.0% and 2.0% initiator concentrations showed CT either too long (8.5 min) or too short (3.3 min), although their CS were very similar to each other

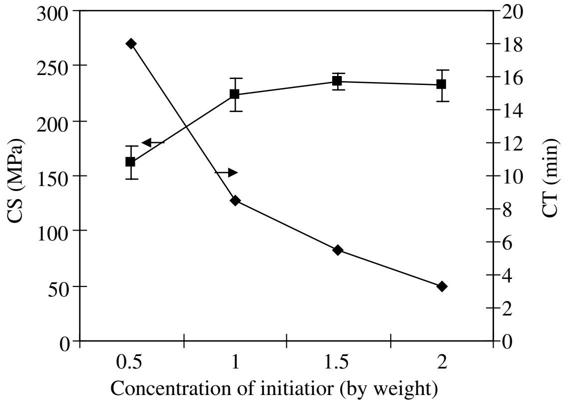

Fig. 4. Selection of optimal initiator concentration: Filler composition = A (described in Table 1); Filler level = 75%; BTTEMA/TEGDMA = 50:50.

Table 1
Effects of oligomer and glass composition on CS and CT

Code	CS* [MPa] (SD)	Viscosity (cp)	Composition (by weight)
Different synthesized oligomers			
A	235.5 ± 7.3	172	BTTEMA/TEGDMA
B	184.8 ± 15.5^a	151	BPTEMA/TEGDMA
C	199.4 ± 8.0^a	84	IPTEMA/TEGDMA
Different glass compositions			
A	235.5 ± 7.3	172	35/6/47/12 (HA/SrF_2/ZnO/SiO_2)
D	213.7 ± 10.1	172	47/6/35/12 (HA/SrF_2/ZnO/SiO_2)

B and C had the same filler composition as A did; D had the same resin composition as A did; Initiator concentration = 1.5%; Oligomer/diluent = 50:50; Filler level = 75%.

[*] Entries are mean values with standard deviations in parentheses; the mean values with the same superscript letter were not significantly different ($p > 0.05$).

and no statistically significant difference was found between them.

3.2.3. Determination of optimal liquid formulation

In both dental composite resin [10–14] and composite resin bone cements [3–6], TEGDMA is the most commonly used resin diluent because of its low viscosity. Most oligomers such as Bisphenol A glycidyl dimethacrylate (BisGMA) or other urethane dimethacrylate are quite viscous due to their higher MW and molecular interactions between molecules such as hydrogen bonding. Without a diluent these viscous oligomers can lead to low conversion of polymerization [10,32,33]. Based on this principle, we first studied the ratio of oligomer/diluent and then compared three different diluents including TEGDMA (a dimethacrylate), TMPTMA (a trimethacrylate), and EB-40 (a tetramethacrylate) and their effects on CS and CT of the formed cements. As shown in Fig. 5, with decreasing BTTEMA/TEGDMA ratio from 70/30 to 40/60, the CS increased from 188 to 249 MPa and the CT was decreased from 15 to 2.3 min. More diluent in the system actually increased the degree of conversion and speeded up the reaction due to lower viscosity and ease of molecular movement, thus leading to a higher CS and a shorter CT. As proposed in the introduction, the purpose of this study was to use the novel oligomers to formulate the cements for better interfacial bonding between glass particles and polymer resins. Therefore, we tried to keep the amount of the oligomer as high as possible in order to have better durability. In doing so, we chose 40/60 (BTTEMA/TEGDMA) for further formulations. As shown in Table 2, TEGDMA still was the best among di-, tri-, and tetra-methacrylates, based on CS and viscosity values. A lower viscosity means ease of manipulation which is of great clinical importance. When we compared the combinations of three component systems, we found that BTTEMA/TEGDMA/TMPTMA were higher in CS than the BTTEMA/TEGDMA/EB-40. Among them the cement with a ratio of 40/40/20 was considered the best, because this formulation not only demonstrated the highest CS (261.5 MPa) but also exhibited a lower viscosity (99 cp). Both TMPTMA and EB-40 belong to a multifunctional crosslinker, which usually makes the material more brittle [34]. However, some studies showed that using tetra- or pentamethacrylates could increase the mechanical strength [25]. Further, the results in this study showed that a combination of both TEGDMA (dimethacrylate) and TMPTMA (trimethacrylate) with more TEGDMA and less TMPTMA demonstrated a higher CS compared to either TEGDMA alone or a combination of TEGDMA and EB-40.

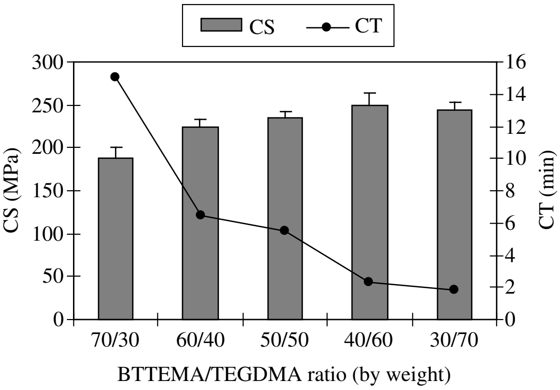

Fig. 5. Selection of optimal ratio of BTTEMA/TEGDMA: Filler composition = A; Filler level = 75%; Initiator conc. = 1.5%.

3.2.4. Determination of optimal filler level

Filler level can significantly affect the properties of the cements [28–31] or composite resins [35,36]. From Fig. 6, it is clear that with the increase of filler level, the CT increased, which may be attributed to an increased distance between resin molecules and thus a slower

Table 2
Effect of diluent on CS and CT

Materials in resin liquid	CS* [MPa] (SD)	Viscosity (cp)	Ratio in liquid (by weight)
BTTEMA/TMPTMA	216.0 ± 16.0^{c}	610	40/60
BTTEMA/EB-40	183.0 ± 12.5^{d}	799	40/60
BTTEMA/TEGDMA	249.0 ± 14.0^{a}	51	40/60
BTTEMA/TEGDMA/TMPTMA	261.5 ± 6.3^{a}	99	40/40/20
BTTEMA/TEGDMA/TMPTMA	260.8 ± 9.6^{a}	159	40/30/30
BTTEMA/TEGDMA/TMPTMA	$231.7 \pm 11.2^{b,c}$	244	40/20/40
BTTEMA/TEGDMA/EB-40	$241.1 \pm 10.9^{a,b}$	175	40/40/20
BTTEMA/TEGDMA/EB-40	$227.2 \pm 14.9^{b,c}$	207	40/30/30
BTTEMA/TEGDMA/EB-40	189.6 ± 10.6^{d}	322	40/20/40

Filler composition = A; Initiator concentration = 1.5%; Filler level = 75%.

[*] Entries are mean values with standard deviations in parentheses; the mean values with the same superscript letter were not significantly different ($p > 0.05$).

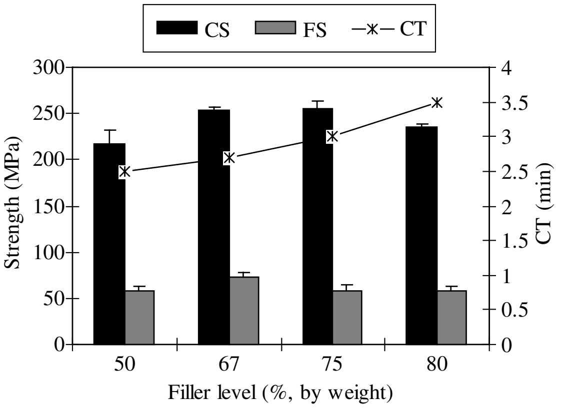

Fig. 6. Selection of optimal filler level: Filler composition = A; Initiator conc. = 1.5%; BTTEMA/TEGDMA/TMPTMA = 40:40:20.

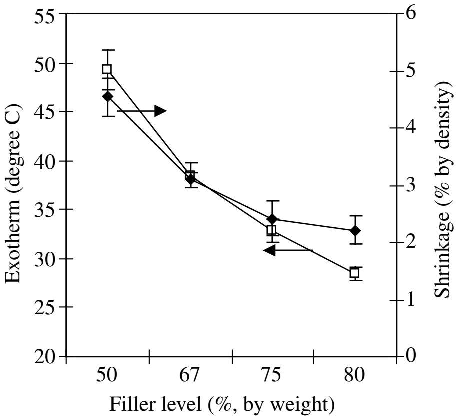

Fig. 7. Effect of filler level on exotherm and shrinkage: Filler composition = A; Initiator conc. = 1.5%; BTTEMA/TEGDMA/TMPTMA = 40:40:20.

polymerization propagation. However, the increase in CS was not directly proportional to an increase of filler level. In fact, the filler level at both 67% and 75% gave the top values in CS (253 and 255 MPa). Flexural strength (FS) was used to distinguish the two cements and the result showed that the one with 67% showed the highest FS (73.2 MPa), as compared to the others (57.9, 58.5 and 58.8 MPa).

3.2.5. Exotherm and shrinkage

Exotherm (peak temperature) was measured using a slightly modified ASTM F-451 standard procedure and shrinkage was determined using a simple published method with a minor modification [26]. Due to the hygroscopic character of the synthesized oligomers, hexane was used as the liquid for shrinkage measurements for the experimental cements instead of water. The results seem very reasonable. The effects of filler level and oligomer/diluent ratio were determined. As shown in Fig. 7 and Table 3, both exotherm and shrinkage decreased from 49.2 to 28.4 °C and from 4.54 to 2.21%, respectively, with increasing filler level (50–80%).

The effects of oligomer/diluent ratio are also shown in Table 3. With decreasing BTTEMA/TEGDMA ratio, i.e., 70/30 to 30/70, both exotherm and shrinkage increased, which can be correlated to more carbon–carbon double bonds in TEGDMA due to its lower MW, as compared to the BTTEMA oligomer. By comparing TEGDMA (dimethacrylate), TMPTMA (trimethacrylate) and EB-40 (tetraacrylate) at the same ratio (40/60), EB-40 exhibited the highest exotherm and lowest shrinkage, followed by TMPTMA and TEGDMA. The exotherm can also be correlated to different quantities of double bonds that exist in different diluents. However,

Table 3
Exotherm and shrinkage

Variable	Peak temperature (°C)	Curing time (min)	Shrinkage (%)
Filler level (%, by weight)[a]			
50	49.2 ± 2.0	5.3 ± 0.8	4.54 ± 0.33
67	38.5 ± 1.2	3.5 ± 0.5	3.10 ± 0.13
75	32.8 ± 1.2	3.3 ± 0.3	2.42 ± 0.30
80	28.4 ± 0.7	7.5 ± 0.4	2.21 ± 0.25
Oligomer/diluent ratio (by weight)[b]			
70/30 (BTTEMAT/TEGDMA)	26.6 ± 0.5	8.3 ± 0.8	2.06 ± 0.33
40/60 (BTTEMAT/TEGDMA)	29.7 ± 1.2	4.3 ± 0.4	3.12 ± 0.49
30/70 (BTTEMAT/TEGDMA)	31.1 ± 0.4	3.8 ± 0.3	4.00 ± 0.29
40/60 (BTTEMAT/TMPTMA)	32.7 ± 0.6	4.3 ± 0.3	2.55 ± 0.36
40/60 (BTTEMAT/EB-40)	37.0 ± 0.7	3.8 ± 0.5	0.57 ± 0.21
40/40/20 (BTTEMAT/TEGDMA/TMPTMA)	32.8 ± 1.2	3.3 ± 0.3	2.42 ± 0.30

Filler composition = A; Initiator concentration = 1.5%.

[a] The liquid formulation was 40/40/20 (BTTEMAT/TEGDMA/TMPTMA).

[b] Filler level = 75%.

less shrinkage exhibited by EB-40 and TMPTMA may be interpreted as the reason that a sudden and quick shrinkage formed by a multiacrylate crosslinker actually blocks further significant apparent shrinkage of the whole network. The exact mechanism involved still needs to be further studied. It seems that there is no specific pattern that can be followed in terms of CT.

3.2.6. *Effect of aging*

Both glass-ionomer cements (GICs) and compomers are known for their increased mechanical strengths with time, especially within the first week [16,28–31]. This was also observed for our new cements. Results in Fig. 8 illustrated the effect of aging on CS. A significant increase in CS was found between 1-h (202.3 MPa) and one-day (253 MPa) tests and after that no significant change was found over nine months (253–262 MPa). The trend for strength increase is reasonably consistent with those results for both GICs and compomers reported elsewhere [28–31]. The reason for this effect can be attributed to the gradual salt-bridge formation between zinc as well as calcium cations from reactive fillers and carboxylic acids from the polymer networks. The initial CS was formed by covalent cross-links between methacrylates on the oligomers, initiated by redox free-radical polymerization. The unchanged CS is attributed to a strong interfacial bonding between glass filler particles and polymer resins due to ionic bond formations, which is the advantage of this type of composite, as compared to conventional dental composite resins. The latter has been reported to show significant reduction of mechanical strength due to degradation caused by poor interfacial bonding between glass fillers and polymer resins [11–14,27].

3.3. *Comparison between optimal cement and PMMA*

CS, diametral tensile strength (DTS), FS, exotherm and shrinkage of our optimal cement were compared with those of Orthoset® 3 bone cement (commercial PMMA cement). As shown in Table 4, the experimental cement exhibited much higher CS (253 MPa), higher DTS (29.6 MPa), similar FS (73.2 MPa), and much less exotherm (38.5 °C) and shrinkage (3.1%), compared to 88.3 MPa, 25.4 MPa, 68.7 MPa, 86.8 °C, and 8.6%, respectively, for PMMA bone cement.

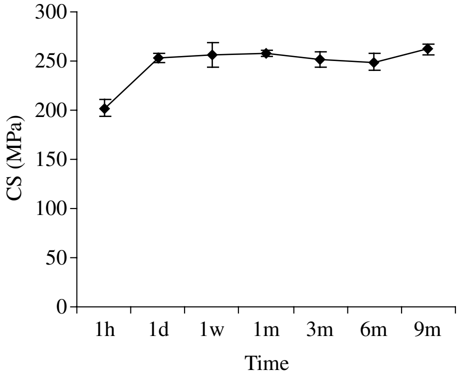

Fig. 8. Effect of aging on CS: Filler composition = A; Initiator conc. = 1.5%; BTTEMA/TEGDMA/TMPTMA = 40:40:20; Filler level = 67%.

Table 4
Property comparisons between experimental and PMMA cements

Property	Optimal experimental cement	PMMA
CS (MPa)	253 ± 4.5	88.3 ± 4.5
DTS (MPa)	29.6 ± 2.6	25.4 ± 2.2
FS (MPa)	73.2 ± 5.5	68.7 ± 4.4
Exotherm (°C)	38.5 ± 1.2	86.8 ± 0.25
Shrinkage (%, by density)	3.10 ± 0.13	8.60 ± 0.19

The cements were conditioned at 37 °C in distilled water for 24 h. For the optimal experimental cement: Filler level = 67%; BTTEMAT/TEGDMA/TMPTMA = 40:40:20; Filler composition = A; Initiator concentration = 1.5%.

4. Conclusions

We have developed a novel non-aluminum-containing bone cement composed of adhesive oligomer as a resin matrix and sintered zinc-calcium-silicate as a reactive filler. The experimental cement exhibited significantly higher compressive strength, higher diametral tensile strength, similar flexural strength, and significantly less exotherm and shrinkage, as compared to conventional PMMA bone cement. The optimal concentrations for initiators were found to be 1.5% (by weight) for both benzoyl peroxide and *N*, *N*′-dimethyl-*p*-toluidine. Generally, with increasing initiator concentration, diluent content and zinc oxide content in the cement formulation compressive strength of the cement increased but curing time decreased. Shrinkage and exotherm of the cement decreased with increasing filler level. Compressive strength was not proportional to an increase of filler level and curing time increased with an increase of filler level. During aging, the cement showed an increase in strength over 24 h and then no change for nine months. Future studies should focus on adhesion to bone, fatigue and biological properties such as bioactivity and biocompatibility.

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