Comparison of the material properties of PMMA and glass-ionomer based cements for use in orthopaedic surgery

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The intrisic benefits of low exotherm and bioactivity have generated interest in utilizing glass-ionomer cements (GIC) as a bone cement replacement in orthopaedic surgery. This paper is concerned with evaluating the mechanical properties of compressive strength, flexural strength, and fracture toughness for two traditional GICs, one resin-modified GIC (an experimental bone cement) and two polymethylmethacrylate (PMMA) cement systems. To determine the suitability of a GIC system for use in the clinical orthopaedic setting, the additional characteristics of setting exotherm and setting time have also been evaluated. The characterization of these two vastly different cement systems has raised some concern as to the applicability of using the current orthopaedic standards for the testing of GIC systems. In particular, issues relating to the strain rate dependence of PMMA cement and the exothermic basis for determining setting time are not applicable as these factors are not characteristic of GIC systems. Whilst the intrinsic benfits of current GIC systems are well understood and generally accepted, this study has shown their intrinsic mechanical properties to be inferior to current PMMA cements. Improvement in the mechanical properties of traditional GICs have been achieved with the addition of a resin component (HEMA).

1. Introduction

Diarthrodal joints rendered useless by disease or injury are frequently treated by replacement (prosthetic) arthroplasty. Polymethylmethacrylate (PMMA) bone cement is currently used as a fixation material for anchorage of the prosthetic device to the skeleton. The primary function of PMMA is to fix and to secure the prosthesis to the bone during arthroplasty of the hip, knee or shoulder [1–3]. PMMA bone cement provides this fixation as a grouting agent [4]. There is no adhesion by chemical means and fixation is achieved only by mechanical interlock at the cement/bone interface [5].

In recent years, there has been an increased interest in better methods of fixation for orthopedic implants [6, 7]. A direct chemical bond between the bone and the implant has attracted investigation and may be desirable [8]. A material that satisfies this requirement, glass-ionomer cement (GIC), has also attracted interest as a suitable fixation material [9–13].

GICs in their original, self-hardening form became available to the dental profession in the mid-1970s. They were invented by Wilson and Kent at the United Kingdom Laboratory of the Government Chemist [14] and became known as acid-base cements [15]. Their setting involves neutralization of acid groups on a water-soluble polymer, typically poly(acrylic acid) and the powder is formed from fluoro-aluminosilicate glass which acts as the base in the sense that it accepts protons from the acid, even though it is not soluble in water.

Using animal models, GICs have been shown to be stable to bone contact, and to promote bone growth [9, 16], an effect attributed to ion release (fluoride) from the material [16–19]. Aluminum ionic release, on the other hand, has recently been demonstrated [20–22] to have a negative effect on bone mineralization [23] and to accelerate calcium mobilization from bone [24]. Whilst this is undesirable it has been shown that low concentrations of aluminum stimulate the proliferation of osteoblasts and new bone formation [25]. Despite these reports, favorable biological outcomes have occurred outside the field of dentistry. They included ear, nose and throat surgery, where GICs have been used to cement cochlea implants [26], to seal imperfections in the skull through which cerebrospinal fluid would otherwise leak [27] and to create prefabricated artificial ossicies [28]. The latter procedure has been particularly successful, with over 2000 operations worldwide [29].

Load bearing applications demand greater strength and toughness and in this respect GICs have been used with limited success. Human trials of GICs in total hip arthroplasty demonstrated a success rate of 65% over a follow-up period of 2.5–5 years [9]. This high failure rate was largely accounted for by the fact that only high-risk cases with poor bone stock were performed [9]. GIC has also been used to reinforce osteoporotic femoral heads to improve the primary stability of dynamic hip screws [30].

GICs have mechanical properties which are more than satisfactory for dental applications such as bonding all ceramic or porcellain fused to metal crown restorations [31]. However, they have relatively low tensile strength and fracture toughness. There have been many attempts to improve the mechanical properties of this cement [32, 33] such as the incorporation of experimental glass fibers [34].

Fiber incorporation has been an interesting means of achieving better mechanical performance and these GICs are generally termed reinforced GICs. Poolthong *et al*. [32] in a comparative evaluation of the biaxial flexural strength of GICs showed that the incorporation of treated glass fibers resulted in ductile like failure due to fiber pull-out and crack tip bridging. Inclusion of alumina fibers has been reported to improve flexural properties [35] and incorporation of carbon and safil fibers showed increases in strength and modulus [36]. If brittle safil fibers are incorporated, brittle fracture of the reinforced GIC will take place with no fiber pull-out [36]. Brittle behavior of GICs can be changed to a more ductile or predictable one by carbon fiber reinforcement whereby flexibility of the fiber permits fiber pull-out. Generally, fiber reinforcement will prevent catastophic fracture caused by surface and internal flaws which is common to brittle materials. However, such resistance to catastrophic fracture or fracture stability of fiber reinforced cements has not been demonstrated [37].

In addition to fiber toughening mechanisms, improvements in strength have been sought by such means as the inclusion of finely divided silver alloy [38] or of a silver cermet formed from the glass plus silver in the fusion process [39]. The addition of either metals and/or fibers to reinforce traditional GICs has led to significant problems with mixing, and only marginal increases in strength [34, 40].

A further type of GICs incorporate a resin component and are usually hybrids that involve the incorporation of polymerizable components into the acid-base glass-ionomer cement. In their simplest form, the polymerizable substance is hydroxyethylmethacrylate (HEMA), together with an appropriate initiation system. The original resin-modified glass-ionomers were designed to be cured by the application of visible light and employed a camphorquinone-amine initiation system [41]. More recently, resin-modified GICs are capable of curing without light initiation on mixing e.g. benzoyl peroxide with amine accelerator. There is a longer working time because HEMA slows the traditional acid-base reaction [42].

As resin-modified GICs are polymer based, they can deform prior to fracture [43]. Their modulus of elasticity is substantially lower than conventional GICs [44] and the failure mode is changed from brittle to tough [45]. Differences in test methodology preclude direct comparison of individual GIC systems, but in general, resin-modified GICs outperform conventional GICs in terms of compressive and tensile strength [46, 47]. There is evidence of slight swelling of the cured cement in aqueous media [48] in contrast to the shrinkage of PMMA systems. Additionally, clinical indications for their use have been promising, demonstrating good adhesive characteristics [10].

The bioactive benefits of GICs are well established but the evaluation of their mechanical properties have received limited attention. If these materials are to present themselves as a bone cement replacement in orthopedic surgery then as a preliminary stage their material properties must be determined. To achieve this, this paper is concerned with the determination of compressive and flexural strength, modulus, fracture toughness, reaction exotherm and setting time for current PMMA and GIC systems.

2. Materials and methods

2.1. Materials

Simplex bone cement (Howmedica International Limited, London, United Kingdom) and CMW3 bone cement (DePuy International Limited, CMW Laboratories, Blackpool, United Kingdom) have been assessed in this study. These two materials are representative of current PMMA bone cements used in orthopaedic surgery. The ''traditional'' GICs that have been investigated are Fuji IX and Fuji II (GC Corporation, Tokyo, Japan). Both are current dental cements. One resin-modified (experimental GIC bone cement), incorporating a 10% (by weight) hydroxyethylmethacylate (HEMA) component has been included in this study. This formulation has been included in the study with the expectation that its mechanical behavior will be characteristic of both the brittle GIC systems and the viscoelastic PMMA systems. GICs are supplied by the manufacturer in either an encapsulated or powder/liquid form. Both have been tested. All PMMA cement systems and the resin-modified GIC bone cement have been mixed using the open bowl method (i.e. hand mixed). All cement systems have been prepared according to manufacturer's instructions.

Additional cement systems have been included in this study to support the results of the above mentioned cement systems for each test. The additional GIC cement is Fuji I (GC Corporation, Tokyo, Japan) and the additional PMMA cements are Osteobond (Zimmer, Warsaw, USA) and Palacos-R (Schering-Plough International). The mechanical testing was performed using a Shimadzu AG mechanical testing machine (Shimadzu, Japan).

As the conventional GIC materials can be severely stressed by hydration or dehydration shrinkage [15], it is recommended that glass ionomer samples be protected by a suitable agent. At this stage it is not clear how susceptable these materials are to hydration or dehydration therefore, in this study, we have tested specimens with the prescribed coating FujiCoat (GC Corporation, Tokyo, Japan) to prevent desiccation.

2.2. Mechanical testing

2.2.1. Three-point bend test

The three-point bend test is used frequently for the determination of flexural strength and modulus of elasticity. Traditional uniaxial tensile testing poses some technical problems when applied to brittle materials such as GICs. This generally relates to load misalignment, which induces a bending moment on the specimen and leads to premature failure. For these reasons, it was decided that the measurement of flexural strength offered the best practical and reliable estimate of tensile strength. The flexural strength of cements were measured using beam shaped specimens 20 mm × 5 mm × 1.6 mm. They were prepared by directly casting the prepared cement into a mould. The mould was then clamped for 10 min and stored at 37 °C for 1 h. The cured cement was removed from the mould and stored in distilled water at 37 °C for 24 h before testing.

Three-point loading was used with the specimen supporting rollers set at 15.5 mm apart. The load was applied by the central loader, to the mid-point of the specimen at a rate of 1.0 mm/min. The failure stress, σ_b is determined from the following equation:

$$\sigma_b = \frac{3PL}{2bd^2}$$

where P is the force at fracture, L is the distance between outer rollers, b is the specimen breadth and d is the specimen height. The modulus of elasticity is calculated from the gradient of the stress strain curve obtained. The flexural strength is reported to be the average of a minimum of five specimens of a given group, reported to the nearest Megapascal (MPa). The elastic modulus is be reported in Gigapascals (GPa).

2.2.2. Compression test

Orthopedic bone cement specimens were prepared in accordance with International Standard 5833 [49] and International Standard 9917 [50]. International Standard 9917 [50] was specifically drafted for the dental use of GICs whereas International Standard 9917 [50] was specifically drafted for the orthopedic use of PMMA cements. The fundamental differences between the two standards are the required specimen dimension and the rate of mechanical testing. ISO9917 requires cements to be prepared as cylinders of dimension 6.0 ± 0.1 mm high and 4.0 ± 0.1 mm in diameter whilst ISO 5833 requires 12.0 ± 0.1 mm high and 6.0 ± 0.1 mm in diameter. Specimens prepared for ISO5833 and ISO9917 were tested at crosshead speeds of 20 and 1 mm/min respectively, i.e. four strain rates. The specimens were stored in water at 23 °C for 24 h prior to compressive strength testing. The compressive strength was calculated as the failure load divided by the measured cross-sectional area taken at the 2% strain offset for PMMA cements and as absolute maximum stress for GIC systems. The compressive strength was the average of a minimum of five specimens of a given group, reported to the nearest Megapascal.

2.2.3. Fracture toughness

The fracture toughness for GIC and acrylic cements have been determined according to ASTM E399-83 [51]. The fracture toughness, K_{Ic}, is considered to be a good indicator of the integrity of cement systems because of the porosity and less than optimal filling of bone cavities that exist in clinical practise. The importance of this parameter is attributed to the fact that fracture toughness characterizes the resistance of a material to fracture in the presence of a sharp crack under severe tensile constraint. A K_{Ic} value is believed to represent the lower limiting value of fracture toughness. This value may be used to estimate the relation between failure stress and defect size for a material in service. Using this method [51], the crack length, a is nominally equal to the thickness, B, and is between 0.45 and 0.55 times the width, W. The ratio W/B is nominally equal to two, to satisfy plain strain conditions. The specimens dimensions that have been tested have a width of 5 mm and a thickness of 2.5 mm.

The standard specimen is a single edge notched beam loaded in three-point bending with a support span, S, nominally equal to four times the width. The testing apparatus in our study had a span of 20 mm. Calculation of, K_{Ic} for each specimen was performed using the equation below. The units of fracture toughness are MPa · $m^{1/2}$.

$$K_{Ic} = \left[\frac{P_Q S}{BW^{3/2}}\right] f\left(\frac{a}{W}\right)$$

$$f\left(\frac{a}{W}\right) = \frac{3(a/W)^{1/2}[1.99-(a/W)(1-(a/W))(2.15-3.93(a/W)+2.7(a^2/W^2))]}{2(1+(2a/W))(1-(a/W))^{3/2}}$$

where P_Q is the load (kN); B, the specimen thickness (cm); S, the span (cm); W, the specimen depth (width) (cm); and a the crack length as determined (cm).

2.2.4. Exotherm and setting time

A Datataker datalogger (Data Electronics USA Inc, Irvine, USA) connected to two thermocouples (Industrial Pyrometers Aust, Sydney, Australia) was used to record the temperature for a fixed volume [49] of GIC and PMMA cement. One thermocouple was used to record ambient temperature and the second was placed within the cement volume. A continuous temperature was logged from the onset of mixing to the eventual setting of the cement. The temperature acquisition system has an accuracy of $\pm$ 0.5 °C.

For each unit of cement, plots of cure temperature versus time and highest temperatures (exotherm) were recorded. The setting time [49] is defined as the time taken to reach the temperature midway between the ambient and maximum recorded temperatures. For each unit of cement, this setting time, T, is measured from the beginning of mixing until the temperature of the polymerizing mass reaches;

$$\frac{T_{\max} + T_{\text{amb}}}{2} = T_{\text{amb}} + \left(\frac{T_{\max} - T_{\text{amb}}}{2}\right)$$

where T_{amb} is the ambient temperature and $T_{\max}$ is the maximum temperature.

3. Results and discussion

The compressive strength, flexural strength and fracture toughness are shown in Tables I–III. Comparing the

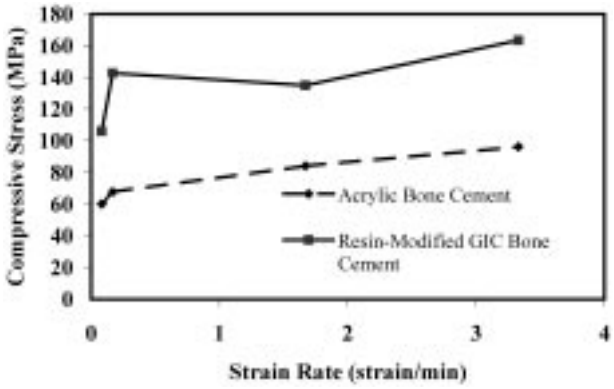

Figure 1 PMMA and resin-modified GIC bone cements show an almost similar viscoelastic response. Traditional GIC systems exhibit a purely brittle failure.

results for PMMA (Simplex and CMW3) and traditional GIC systems (Fuji IX and Fuji II) the data indicates that there are marked differences in the mechanical properties between these two cement systems. The test results indicate that traditional GICs (Fuji II and Fuji IX) display on average higher compressive strength, lower bending strengths, lower fracture toughness and higher modulus when compared to the PMMA bone cements tested. Results for the resin-modified GIC (experimental bone cement) show that the addition of the resin component to the traditional GIC will increase the flexural strength, reduce the compressive strength and increase the fracture toughness compared to the traditional GICs tested.

Compressive strength tests were conducted in accordance with the orthopaedic standard for acrylic cement systems, International Standard 5833 [49], and the dental standard for water-based systems, International Standard 9917 [50]. Due to the specimen size and loading rates specified in each standard, the compressive tests were conducted at 4 strain rates (0.083, 0.167, 1.67 and 3.33 per min). The compressive strength results of the PMMA cement systems demonstrated an increased strengthening trend with strain rate – increasing in a linear manner (Fig. 1). Similar strain rate responses in other modes of loading, flexure [52] and shear [53, 54] have also been reported. For the resin-modified GIC, at very low strain rates (0.083 and 0.167 per min) the increase in strength with strain rate also existed, but resulted in a greater increase in strength compared to the PMMA cements. At even higher strain rates, the increase in strength is the same for both the resin-modified GIC and PMMA bone cements (Fig. 1). The compressive strengths of traditional GICs were, on average, over twice those of PMMA cement systems, however these systems could not be tested at higher strain rates due to limitations in specimen size. The average compressive strength for the traditional GICs tested (Fuji II and Fuji IX) was 168.7 MPa compared to 72.1 MPa for PMMA cements at a strain rate of 0.167 per min (Tables I and II). The compressive strength of resin-modified GIC was 106.1 MPa (0.167 per min).

Compressive failure of the traditional GICs tested was observed to be classically brittle. The compressive failure of the PMMA cements, on the other hand, showed a general viscoplastic behavior with a marked plastic deformation prior to final fracture (Fig. 2). As the strain rate increased, brittle failure occurred more markedly in the traditional GIC systems, giving a sharper narrower peak whilst the failure became much more defined (extended roll-over) for the PMMA cements. Interestingly, the resin-modified GIC failed in a manner with characteristics of both cement systems – consisting of a small viscous component and a sharp final fracture.

The compressive results for the traditional GICs showed greater dispersion in compressive strength and modulus than did the PMMA cements. One could argue that this may be attributed to their intrinsic brittle nature, voids and/or specimen misalignment. The addition of the resin component to the traditional GIC changed the nature of failure from brittle to ductile and reduced the compressive strength by 37%, but this is still greater than the compressive strength of PMMA.

Flexural testing results show an opposite trend to the compressive results. Due to the short setting characteristics of GIC systems, a limitation was placed on the maximum physical specimen size that could be utilized

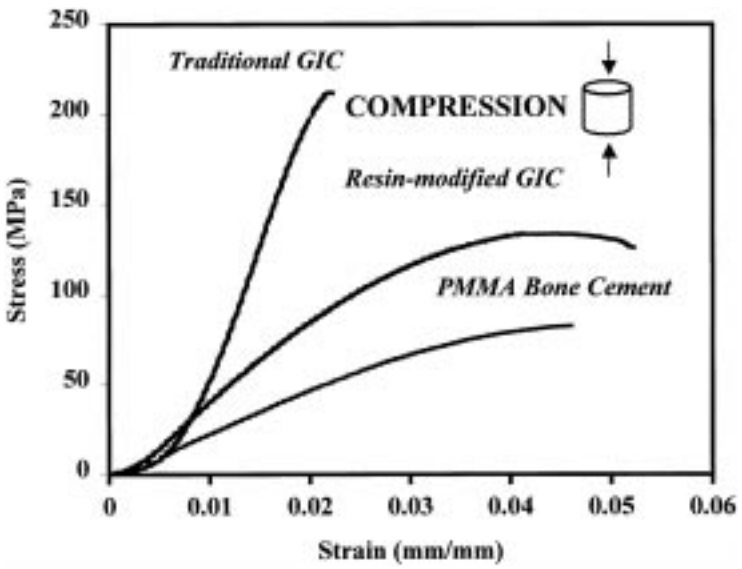

Figure 2 Compressive stress–strain response of PMMA bone cement, traditional GIC and resin-modified GIC. The traditional GIC shows maximum strength and minimum yield (visco-plastic response), whereas the PMMA bone cement and experimental GIC indicate substantial plastic deformation prior to failure.

TABLE I PMMA Cements – compression and flexural properties

		Compression				Flexure
		ISO 9917:1991(*E*)		ISO 5833:1992(*E*)		Ref. 26
		1 mm/min	20 mm/min	1 mm/min	20 mm/min	1 mm/min
Simple *x*	σ	67.69 ± 3.57	96.12 ± 1.90	60.10 ± 1.91	84.09 ± 2.33	72.56 ± 3.95
	E	2.41 ± 0.12	2.53 ± 0.05	2.15 ± 0.05	2.34 ± 0.07	2.63 ± 0.07
CMW3	σ	75.04 ± 1.04	100.71 ± 2.30	68.34 ± 1.29	92.68 ± 2.48	64.77 ± 3.70
	E	2.57 ± 0.081	2.60 ± 0.056	2.27 ± 0.035	2.45 ± 0.10	2.30 ± 0.061
Palacos	σ	73.61 ± 3.3	100.31 ± 2.90	71.94 ± 0.69	86.12 ± 2.08	ND
	E	2.25 ± 0.12	2.79 ± 0.03	2.27 ± 0.03	2.38 ± 0.08	ND

Units: Modulus (GPa) and strength (MPa).

for flexural testing. This maximum specimen size was 20 mm × 5 mm × 1.6 mm and therefore International Standard (ISO) 5833 [49] could not be used to determine flexure strength for both material systems because it specifies a much larger specimen size. As a direct result of the limitation in specimen size and due to the compressive tests having already established the rate dependency of the cement systems, it was decided to conduct the flexural tests for both material types at the same rate and same specimen size.

All materials tested, both traditional GIC, resin-modified GIC and PMMA failed in a traditionally brittle manner. There was no evidence of any viscous component in the PMMA cements. This was in contrast to the earlier finding for the compressive test. The average traditional GIC flexural strength was on average less than half the average PMMA cement strength, being 33.6 and 68.6 MPa, respectively. The addition of the resin component led to an increase in the flexural strength to a value of 59.3 MPa – approaching the strength of the acrylic cements tested (68.6 MPa).

A further difference between traditional GICs, resin-modified GICs and PMMA cement systems is the difference in modulus. The flexural modulus of the traditional GICs tested (av. 12.8 GPa) was found to approximately double the flexural modulus of the resin-modified GIC (av. 6.5 GPa). Additionally, the flexural modulus of the resin-modified GIC (Tables I and II) was more than double that of the PMMA systems (av. 2.5 GPa). The average compressive modulus of the traditional GICs was 11.9 GPa compared to 2.4 GPa for the PMMA cements and 4.7 GPa for the resin-modified GIC bone cement. There were small but not significant differences in the compressive and flexural moduli for each material. The difference can be attributed to specimen condition, the nature of each test, and the coatings applied to the GICs to prevent dehydration. The coatings are required to prevent early desiccation of the GIC materials and will have different effects in flexure and compression.

Cracking of cement mantles around the prosthesis has been indicated as a major source of failure [11, 13]. The size and distribution of voids, coupled with an alternating stress regime place emphasis on the importance of a material having a high fracture toughness. Whilst the fracture toughness measurements determined from this study indicate that PMMA cements have fracture toughness's of the order 1.6–1.7 MPa · $m^{1/2}$ (Table III). Given the existing frequency of fracture of cement mantles due to inherent voids, it would indicate that for a material to eventually replace PMMA as an orthopedic ''grout'', this material would require an equivalent or greater fracture toughness. The fracture toughness of the traditional GICs indicate a shortcoming in this regard.

TABLE II Glass-ionomer compression and flexure properties

		Compression ISO 9917:1991 (*E*) 1 mm/min	Flexure Ref. 26 1 mm/min
Fuji I	σ	175.21 ± 12.07	12.82 ± 4.78
	E	7.36 ± 0.67	8.86 ± 0.86
Fuji II	σ	153.21 ± 9.41	36.23 ± 6.99
	E	10.18 ± 2.06	12.60 ± 2.31
Fuji IX	σ	211.16 ± 27.92	29.17 ± 7.56
	E	14.34 ± 2.65	13.09 ± 2.93
Experimental ISO 5533 – 1 mm/min	σ	113.72 ± 5.58	
	E	4.45 ± 0.65	
Experimental ISO 5533 – 20 mm/min	σ	134.99 ± 6.38	
	E	4.99 ± 0.74	58.62 ± 6.73
Experimental ISO 9917 – 1 mm/min	σ	145.07 ± 13.98	6.47 ± 0.21
	E	4.67 ± 0.73	
Experimental ISO 9917 – 20 mm/min*	σ	163.37 ± 13.99	
	E	3.76 ± 0.61	

Units: Modulus (GPa) and strength (MPa).
*Specimen size – $\phi 6x12$ mm.

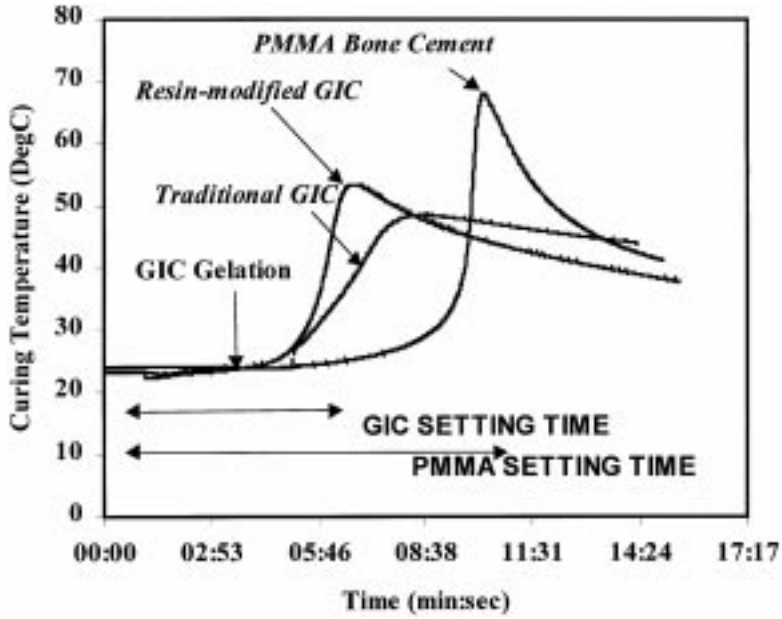

Figure 3 The figure plots the temperature rise following mixing of a PMMA bone cement and two GIC cements and shows the high exothermic temperature and longer setting times characteristic of PMMA cements. The gelation phase for GIC systems occur prior to the onset of temperature and has a viscosity indicative of being considered already set.

The fracture toughness of traditional GIC is low at 0.63 MPa · $m^{1/2}$, however, this is much higher than most posphate cement/bone replacement materials [55]. The addition of the HEMA component has resulted in an increase in fracture toughness to a level of 0.91 MPa · $m^{1/2}$.

Setting time and exotherm results are listed in Table IV. An undesirable characteristic of PMMA cement is they undergo a highly exothermic reaction during setting. This is in stark contrast to the setting characteristics of traditional GICs where the influence of temperature is small, but not minimal as has been reported [56, 57]. The determination of setting time using international standard [49] lends itself to a overestimation for the traditional GIC and the resin-modified GIC bone cement because it is based on the temperature dependence of PMMA systems. The setting of GIC materials cannot be determined accurately from its exothermic profile alone. The second phase of the GIC reaction, termed the gelation phase, is characterized by an initial hardening of the cement with transfer of ions from the glass to the acidic matrix and leading to a significant increase in viscosity. At this point, it was noted that the cement mass is no-longer workable and could be described as being set (Fig. 3). In PMMA cements, the temperature change indicates a rapid transition from the fluid to solid phase and for this material the standard [49] is a good indicator of setting. The setting time of PMMA cements was found to be, on average, 4 min longer than the resin-modified GIC bone cement and 6 min longer than the traditional GICs. The setting time for current PMMA cements is deemed clinically suitable. If the gelation phase of the resin-modified GIC is taken as the setting time, then the setting time has reduced to 2 min, compared with approximately 10 min for PMMA systems. This characteristic has led to the conclusion that the current orthopedic bone cement standard [49] is not appropriate for the determination of setting time for GIC systems. A more realistic test would be to define the setting time and working time as being a function of the viscosity or extrudability of the cement systems. However, the standard is relevant to PMMA cements where there is a sudden onset of exotherm indicating that setting of the cement mass has taken place.

TABLE III Fracture toughness results

	K^{*}_{Ic} (MPa · $m^{1/2}$)
Simplex BC	1.66 ± 0.15
Osteobond BC	1.72 ± 0.15
Fuji IX GIC	0.63 ± 0.07
Experimental GIC	0.91 ± 0.10

*Rate of testing 0.5 mm/min.

This study has determined that GICs, in larger volumes than specified [49], can generate substantial exotherm reaction temperatures. The average exothermic temperature measured for the traditional GICs was 43.4 °C. Whilst the recorded temperatures are considerably lower than those for the PMMAs tested (average 62 °C) it may indicate that GICs, in larger volumes, could possibly damage bone tissue. Many authors [58–60] have measured the temperature during polymerization for large volumes (10 cm^3) of PMMA cements and obtained peak values ranging from 80 °C to 124 °C. It has been reported that tissue damage occurs within the temperature range 42–47 °C [61]. The exothermic temperatures measured for the GIC materials was 40 °C for Fuji II, 41.2 °C for Fuji IX and 49 °C for Fuji I (Table IV). The resin-modified GIC bone cement has a recorded exotherm of 54 °C and is only marginally lower than the PMMAs at 56 °C to 67 °C. This is a cause for concern, as one of the intrinsic benefits of GICs for orthopedic applications is the reported low exotherm.

There are many problems with the exotherm testing of cement systems with regard to specimen configuration and volume. It has been noted that the setting and exothermic results for GIC materials are subject to test variation since the international standard [49] does not

TABLE IV Exotherm and setting time

	Exotherm*	Setting time*	Time to exotherm
CMW BC	66.69 °C	9:35	10:13
OsteoBond BC	67.38 °C	10:28	11:18
Fuji I GIC	48.96 °C	5:45	7:50
Fuji II GIC	40.03 °C	3:15	5:44
Fuji IX GIC	41.17 °C	3:22	7:40
Experimental GIC	54.33 °C	5:36	6:15

Units min : sec.
*International standard 5833:1992 (*E*).

specify the mass or volume of cement to be used and therefore the thermal energy content and thus exotherm of the material will change. Swenson [62] disclosed increases in the peak temperature from 33 °C to 80 °C as the diameter of the mold containing the cement increased from 0.75 to 2.0 cm. Another report [63] demonstrated that a cement thickness of 1 cm generated a core temperature of 70 °C, whereas a 3 cm thickness of acrylic cement generated a 124 °C core temperature. In the clinical situation, it was determined that the bone-cement interface temperature varied between 48 °C and 58 °C, and duration was only a few seconds.

4. Conclusions

Static mechanical testing is widely used to initially characterize and screen potential bone cements. The type of tests performed in this study are considered as a necessary preliminary mechanical characterization of the material and a precuser to biocompatibility, bonding and fatigue tests. Every orthopedic bone cement must possess certain minimum static strength properties. This study has highlighted some of the problems of testing to a standard designed for a particular material composition. The problem lies in trying to characterize the properties of two very different cement systems for the same clinical application. The chemical composition of both systems make them ultimately suited to the application but the various methods to assess them are distinctly separate. This paper has suggested some modifications to some existing tests and concludes that setting and working time should be determined in a manner that relates to handling characteristics. In addition, it has been shown that GICs are unique materials that show potential to be a successful bone cement.

In contrast to acrylic cements, the setting reaction of traditional GICs do not generate considerable heat so will not cause thermal damage to tissues at the implant site. However, as resin is added to the system to improve mechanical properties this exotherm is increased substantially. The aim, therefore, is to improve the toughness and strength of traditional GICs without a trade-off in the intrisic properties of bioactivity and low-exotherm. Current results demonstrate that the addition of a resin component to a purely traditional acid-base glass can improve the flexural strength and toughness for a small reduction in compressive strength. The introduction of the resin has the ability to modify a purely brittle GIC with a measurable degree of ductility.

Whilst GICs have been used extensively in dentistry for several decades, their use in orthopedic surgery has been somewhat limited. If this type of material does prove to be a successful candidate for this application then a new set of standards will be required for laboratory evaluation and quality control. To this end, the standards must be realistic to the functional requirements of the orthopedic application and specific to the chemistry of the GIC material in much the same way that the existing International Standard [43] is specific to acrylic-based systems.

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