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Original Article

In vitro comparative assessment of the mechanical properties of PMMA cement and a GPC cement for vertebroplasty

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ABSTRACT

Aims: To develop a Glass Polyalkenoate Cement that is suitable for vertebroplasty.

Methods: Testing was carried out to assess the effect of gamma irradiation used for sterilisation, on the glass transition temperature as well as its mechanical properties, including compressive strength and biflexural strength in vivo as well as testing GPC and PMMA cements post injection in cadaveric human vertebral bone.

Results: There was a trend to a higher failure load required for the GPC cement group compared to the current standard PMMA injected group but this was not statistically significant with this small sample size.

Conclusion: The results are encouraging for future research to continue on GPC cements for use in vertebroplasty.

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1. Introduction

This is a study into a Glass Polyalkenoate Cement (GPC) as an alternative cement to PMMA for vertebroplasty and to our knowledge, it is unique in that it assesses this in human cadaveric osteoporotic vertebrae. There are approximately 700,000 diagnosed vertebral compression fractures annually in the USA; however, it is estimated that only one-third of vertebral fractures come to clinical attention.¹ More than 50 per cent of the thoraco-lumbar spine vertebral compression fractures occur in the four vertebrae about the thoraco-lumbar

junction from T11 to L2 inclusive, where there is a change in the spinal curvature from kyphotic to lordotic.

Vertebroplasty has now been used in the treatment of painful osteoporotic fractures since 1993, having first successfully been performed by Galibert in France in 1984 for the treatment of a painful cervical vertebral haemangioma.² Pain relief following vertebroplasty has been shown to be immediate (with the majority occurring within 24 hours of the procedure), long lasting (up to 12 months) and statistically significantly greater than that achieved by conservative measures, as evidenced by the largest randomised control study comparing vertebroplasty to conservative management VERTOS II.³

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It should be noted that two double blinded multi-centre randomised control studies, both of which were published in 2009, looked at vertebroplasty versus sham vertebroplasty procedures and both found no significant difference between vertebroplasty and sham local anaesthetic procedures in terms of pain relief at one and three months.^{4,5} These studies were criticised for having high exclusion rates and fractures that were up to 12 months old and may have been at varying stages of healing. Two further randomised controlled trials in Holland, VERTOS IV ongoing since 2011 and VERTOS V ongoing since 2013 have been underway and to date remain unreported in the literature.^{6,7}

PMMA cements are currently the most commonly used cements for vertebroplasty and are considered the current standard though they are not without their problems. Vertebroplasty may work by either mechanically stabilising the vertebral segment or by simply burning the nerve endings, since protein damage occurs at approximately 45 °C while PMMA cement polymerisation is known to reach temperatures of 80 °C. It then cools as the polymerisation process slows down and this can lead to shrinkage in the cement volume. Furthermore, cement leakage is the most common complication of this procedure and can occur to varying degrees in up to 43% of patients.⁸ Cement leakage at the posterior cortex thus has a high chance of causing thermal damage to surrounding soft tissue including nerve roots and the spinal cord. In addition, this high polymerisation temperature may result in bone necrosis and may further weaken the vertebra.

Augmenting a vertebral fracture level with PMMA cement changes the biomechanics of the spine resulting in increased strain at both the superior and inferior neighbouring vertebrae but the majority is redistributed to the inferior vertebra which is more likely to fail.⁹ Studies have shown that there is a 19% incidence of sustaining a neighbouring vertebral fracture in the first year following the index vertebral fracture even without surgical intervention.¹⁰ In one study, eighty per cent of the patients who sustained an adjacent fracture had these occur within two months of the vertebroplasty procedure.¹⁰ Of course, given the already likely osteoporotic state of neighbouring vertebra, some of these fractures would have occurred even if the fractured vertebra had not been cemented.

PMMA has a much higher Young's modulus of elasticity than either cortical or cancellous bone leading to a modulus mismatch and a stress riser effect. Vertebroplasty causes a stress riser due to the augmented vertebral body becoming stiffer with the injected cement, which in turn increases the intradiscal pressure in the adjacent discs. This increased intradiscal pressure is then transferred to the adjacent vertebral bodies resulting in their subsequent fracture.¹¹

The type of cement used determines the stiffness of the augmented vertebra and the magnitude of the stress riser. The volume of cement used also affects the strength and stiffness of the augmented vertebra. Studies have shown that as little as 2 ml is required to restore vertebral strength to prefracture levels.¹² Larger cement volumes result in a higher required injection pressure and cause higher rates of cement leakage. A thirty per cent volume fraction was found to lead to an increased stiffness fifty per cent greater than the original prefracture vertebral stiffness.¹³

GPC cements are formed by an acid base reaction between a glass powder and an aqueous polyacrylic acid. GPC cements are currently used in dental applications and it is the aim of this research to contribute to their modification for orthopaedic application. They are quite biocompatible with few adverse reactions in over twenty years of dental use. GPC cements also have exothermic reactions that reach 10 °C above ambient temperature.¹⁴ Unlike PMMA, GPC cements are not prone to volumetric shrinkage. They reach most of their compressive strength within 24 hours but as cations continue to crosslink, their compressive strength continues to improve even up to a year.^{15,16} It is these properties that have led to their development for orthopaedic use.

In order to develop the GPC cement for vertebroplasty, tests were devised to assess its practical application at point of orthopaedic use. Thus, its mechanical properties including its compressive strength and biaxial flexural strength were tested over time in water and in blood, pre- and post-gamma irradiation, which is the method of choice in the sterilisation of medical products. Previous experience has shown that irradiation in air has compromised the mechanical properties of some materials such as the polyethylene used in arthroplasty due to excessive oxidation, which resulted in unacceptable failure rates within a short period of time. The effect of the gamma irradiation on the GPC cement's mechanical properties was assessed.

We also compared the incidence of neighbouring vertebral fractures and load to failure post injection of PMMA and GPC cements into osteoporotic harvested human cadaveric lower thoracic and upper lumbar spinal segments. This is to assess if the GPC cement would be better than PMMA cement in vertebroplasty with regard to its compressive strength being closer to that of bone and thus leading to a lower stress riser effect and thereby a lower likelihood of neighbouring secondary vertebral compression fractures.

2. Materials and methods

2.1. GPC cement synthesis

The GPC glass (0.48 SiO₂-0.36 ZnO-0.12 CaO-0.04 SrO) was synthesised of silica, zinc, calcium and strontium oxides. The GPC glass powder was prepared by weighing out the required amounts of analytical grade reagents (*Sigma-Aldrich, Dublin, Ireland*) and then subjected to ball milling for one hour. This mix was then dried in an oven at 100 °C for one hour. This was then fired at 1480 °C for one hour in a crucible and then shock quenched in water. The resultant frit was then dried, ground and then sieved producing a GPC glass powder of maximum particle size of 45 µm. The polyacrylic acid (PAA) that was used in these experiments was purchased from *Advanced Healthcare, Kent, U.K.* The PAA had a molecular weight (Mw) of 80,800. The PAA was ground and sieved to retrieve particles sized <90 µm. The GPC cement was prepared by thoroughly mixing the glass powder with the PAA and distilled water.

2.2. PMMA cement

A commercially available PMMA bone cement Simplex P® (*Stryker Howmedica, Limerick, Ireland*) was used. Simplex P®

contains 75% Methylmethacrylate-styrene copolymer, 15% Polymethylmethacrylate (PMMA), and 10% Barium Sulphate. Benzoyl peroxide is encapsulated within each methyl methacrylate-styrene-copolymer bead. The PMMA cement was prepared and mixed in a Stryker Mixevac III (Stryker Howmedica, Limerick, Ireland) cement mixing bowl with a vacuum attachment in order to decrease the porosity of the cement. The PMMA cement was then loaded into a 10 ml syringe and injected into the moulds or into the vertebrae.

2.3. Gamma irradiation

The GPC cement was gamma-irradiated according to the following ISO standards: ISO 11137:2006 sterilisation of Healthcare Products, MD76165; ISO 9001:2000 Quality Management System, FM76164 and ISO 13485:2003 Quality System – Medical Devices MD76165. The dose of gamma irradiation used was between a minimum and maximum dose of 31.5 kGy and 31.8 kGy. The commercially available PMMA Simplex P[®] cement which has a molecular weight of 198,000 came pre-sterilised by gamma radiation in its packaging.

2.4. Compressive strength testing

The compressive strength testing of the PMMA and GPC cements was carried out in accordance with the International Organisation for Standardisation ISO 9917 standards.¹⁷ Split ring moulds were made with sets of five cylindrical samples, each sample with a diameter of 4 mm and a height of 6 mm. These were filled to excess with freshly mixed PMMA and GPC cements and then covered with acetate sheets. The moulds were then sandwiched between two stainless steel plates, clamped and then incubated at 37 °C for at least 1 h.

The samples were then removed from the moulds and placed in either distilled water or human blood and labelled and then incubated at 37 °C for 1, 7 and 30 days before compression testing was carried out. On completion of each of the above time periods, the samples for compression testing

were then loaded on an Instron 4082 universal testing machine (Instron Ltd., High Wycombe, Bucks, U.K.) using a 5 kN load cell at a crosshead speed of 1 mm/min. There were five samples of each of the cement types, for each of the incubation fluids and each of the incubation periods.

The compression strength for each sample was then calculated according to the following equation: $C = \frac{4\rho}{\pi d^2}$ where C is compressive strength (Mega Pascals), ρ is maximum applied load (Newtons) as measured by the Instron and d is diameter of the sample (millimetres).

2.5. Biaxial flexural strength

Biaxial flexural strength testing of the cements was carried out in a similar method to that used by Williams et al.¹⁸ using a testing jig with three support bearings on which rests a disc of the cement to be tested and onto which descends a bearing centred on the cement disc (Fig. 1).

After mixing the cements, samples were placed in pre-made circular rubber moulds of 12 mm diameter and 2 mm thickness and filled to excess. These were then placed between two acetate films and the moulds were then sandwiched between two stainless steel plates, clamped and incubated at 37 °C for at least one hour. Samples were then removed from the moulds and then placed in either distilled water or human blood and labelled, then incubated at 37 °C for 1, 7 and 30 days before biaxial flexural strength testing was carried out. Cement disc sample thickness was measured using a Vernier calliper for each sample immediately prior to testing. The test jig described above was fixed into an Instron 4082 universal testing machine (Instron Ltd, High Wycombe, Bucks, U.K.) using a 1 kN load cell at a crosshead speed of 1 mm/min. There were five samples of each of the cement types, for each of the incubation fluids and each of the incubation periods.

The biaxial flexural strength for each sample was then calculated according to the following equation:

$$BFS = \frac{\rho(N)}{t^2} \left\{ 0.63 \ln \left(\frac{P}{t} \right) + 1.156 \right\}$$

Fig. 1 – Cement disc samples being tested for biaxial flexural strength.

where BFS is biaxial flexural strength (Mega Pascals), ρ is fracture load (Newtons), t is sample thickness (millimetres) and r is the radius of the support diameter (millimetres).

2.6. Preparation of cadaveric spines

Three human cadaveric spines were obtained including lower thoracic and upper lumbar spinal segments. The spines were initially removed *en bloc* and then the soft tissues and ribs (except for intervertebral ligaments and facet joint capsules) were dissected from the specimens. Radiographs in two planes – antero-posterior and lateral – were used to exclude specimens with lytic lesions or other bony abnormalities not related to osteoporosis, which might affect testing results. Vertebrae that had been damaged or were incomplete due to the initial *en bloc* removal were also excluded. Bone mineral density was determined on each vertebral level using Dual-Energy X-ray Absorptiometry with a DEXA scanner (Lunar IDXA model, GE Healthcare).

Spines were then divided sequentially into sets of two vertebrae creating functional spinal units (FSUs), leaving intact the intervertebral disc and ligaments between the two vertebrae. The various FSUs were then labelled with tissue markers. To minimise the effects of variability in bone density and the spinal level of treatment: specimen pairs were then sorted according to BMD and then assigned in an alternating sequence to three groups for vertebroplasty treatment. The first acting as a control group with no cement injected, the second would be injected with the GPC cement and the third would be injected with the PMMA cement.

In order to be consistent and to allow for comparison between the two cement groups, only the superior vertebrae were injected. X-rays were taken to confirm position of the 13 gauge trocar in the vertebral bodies prior to and following injection of cement to assess the location of cement injection (Fig. 2). A separate cement mix was used for each vertebral level injected using either PMMA or GPC cement so as to minimise the time from completion of cement mixing to injection and maximise the working time of the cement in order to optimise its injection capacity. The spines that were injected were left at room temperature to allow the cements to set for at least a period of twenty-four hours prior to further testing.

2.7. Compression testing of cadaveric vertebrae

The FSUs were then mounted into a servohydraulic testing machine ensuring parallel orientation of the outer surfaces of each FSU and perpendicular orientation of FSU with respect to

Fig. 2 – Injected GPC cement in superior vertebral body of FSU.

Fig. 3 – Servohydraulic testing machine with mounted FSU for testing.

Fig. 4 – Vertebral wedge compression fracture of inferior non-injected vertebra.

the loading axis (Fig. 3). Each FSU was then compressed at a constant displacement rate of 0.5 mm/s with a load of 0.2 kN/mm²/min. The failure load of the FSU was defined as the peak compressive load measured. FSU specimen loading stopped two seconds after peak compressive load was reached and the specimens removed. After testing, antero-posterior and lateral radiographs were taken to determine the presence and location of fractures and/or reduction in vertebral height (Figs. 4 and 5).

2.8. Statistical analysis

All the data obtained from the experiments were entered into SPSS 22 for later statistical analysis and review. A repeated measures analysis of variance (ANOVA) with time as the repeated measure and medium as the between samples factor was carried out to explore significant differences in outcome variables over time and by medium. A level of significance of $p < 0.05$ was used for statistical tests with the level of significance adjusted for multiple pairwise comparisons. Independent samples t tests were used to compare means across the two groups (GPC, PMMA).

Fig. 5 – Vertebral coronal plane fracture of inferior non-injected vertebra.

3. Results and discussion

3.1. Transition temperature post-gamma irradiation of GPC cement

Gamma irradiation of the GPC cement resulted in a colour change from white to a grey colour. A differential thermal analysis was carried out on the samples pre- and post-gamma irradiation to ensure that there was no change in the glass transition temperature and were found to be very similar, 675.06 °C and 672.64 °C respectively, indicating no significant disruption to the glass network by the gamma irradiation at a dose of approximately 31.5 kGy.

The Simplex P glass transition temperature is 110 °C pre-radiation and was found to decrease by a maximum of 12 °C post-gamma irradiation at a dose of 23.5 kGy in previous experiments also indicating very little structural change by gamma irradiation.

3.2. Pre- and post-gamma irradiation GPC cement mechanical properties in water and blood

The compressive and biflexural strengths of the GPC cement were compared pre- and post-gamma irradiation (while incubated in water for 1, 7 and 31 days at 37 °C). The results of the means of the five samples in each group on each of the testing days, for biflexural strength (BFS) testing pre- and post-irradiation are illustrated in Fig. 6. Similarly, those for compressive strength (CS) testing are illustrated in Fig. 7.

These figures show a large decrease in mean compressive strength and to a lesser degree in mean biflexural strength when comparing post-irradiation to pre-radiation values. This demonstrates a decrease in the mechanical properties post-gamma irradiation of the GPC cement, which improves with time.

Fig. 6 also illustrates that the biflexural strength of the GPC cement in blood post-irradiation improves to close to that prior to irradiation, likely due to the initial small molecular weight glass washout secondary to irradiation being partially

Fig. 6 – Mean BFS pre- and post- (γ) irradiation in water and post-irradiation in blood.

Fig. 7 – GPC mean compression strength pre- and post- (γ) irradiation in water and post-irradiation in blood.

compensated for by the ions in the blood and increasingly so with time.

There is no statistically significant difference in biflexural strength over time ($p = 0.25$) or between media ($p = 0.11$) in a repeated measures ANOVA with time as the repeated measure (Day 1, Day 7 and Day 31) and media (blood or saline) as the between samples factor. The low p -values would suggest, however, that this may be as a result of the small sample size rather than no real differences.

Fig. 7 similarly illustrates that the compression strength post-irradiation in blood improves with time towards that of its original strength pre-gamma irradiation for the same reasons mentioned before. In repeated measures ANOVA, there is a statistically significant difference in compression strength over time ($p < 0.001$) with the measurements taken at day 1 significantly different from Day 31 ($p < 0.001$). There is also a statistically significant difference between the measurements taken at Day 7 and Day 31 ($p < 0.001$) but no significant difference between Day 1 and Day 7 ($p = 0.74$). Medium (blood, saline) is not a statistically significant factor in this analysis ($p = 0.12$) but given the low p -value, this may again be due to the small sample size rather than no real differences.

In both biflexural and compressive strength testing, there was an increase in strength with time and more so in blood than in water. This is a desirable effect as it shows that the GPC

Table 1 – PMMA mechanical properties (units of measurement MPa).

	Day 1	SD	Day 7	SD	Day 30	SD
CS	95.6	5.9	98.4	2.1	116.6	1.7
BFS	147.7	10.7	131.2	9.8	138.7	21.7

Table 2 – Bone mineral density means.

	Control	SD	GPC	SD	PMMA	SD
BMD superior	0.777	0.055	0.767	0.089	0.796	0.072
BMD inferior	0.806	0.059	0.740	0.064	0.798	0.056
BMD average	0.791	0.054	0.751	0.071	0.797	0.039

cement does not weaken over time and in fact gains more strength in the vertebrae's bone marrow blood environment.

Table 1 shows the compression strength and biflexural strength values for PMMA cement, which are significantly higher than those for the GPC cement and also of bone. Cancellous trabecular bone has a compressive strength of 4 to 12 MPa while normal non-osteoporotic cortical bone has a compressive strength of 131 MPa.

3.3. Vertebral testing of cements

The analysis of the vertebral testing is based on several factors. This includes the bone mineral density of the tested vertebrae, the cement penetration of the injected vertebrae, the ultimate load failure on compression and the various severity and locations of the vertebral height loss and vertebral compression.

3.4. Bone mineral density of tested vertebrae

Analysis of the randomised functional spinal unit vertebral pair's bone mineral density of the superior injected vertebrae and the inferior non-injected vertebrae as well as the non-injected superior and inferior control group vertebrae was undertaken and shown in Table 2.

There was no statistically significant difference in the mean BMD across all the groups for either the superior ($p = 0.85$) or inferior ($p = 0.24$) vertebrae, or the average of the two ($p = 0.45$). The lack of a statistically significant difference in the BMD between groups indicates that vertebral FSU randomisation

Table 3 – Mean load to failure in each cement group.

	Mean (Newtons)	Standard deviation (SD)
Control	6850	645.5
GPC	4900	1278.7
PMMA	4650	191.5

was effective and thus on average the cement groups were tested on similar vertebral FSU pairings in terms of the BMD.

3.5. Load to failure results

The failure load was also measured to assess if the cement augmented vertebral functional spinal units required a higher load prior to failure.

Table 3 shows that surprisingly the uncemented control group had higher failure loads than the cemented groups. There was also a much greater spread in the failure loads for the GPC cement (with a much larger standard deviation value) compared to the PMMA cement, which had the narrowest failure load spread. Overall, there was a statistically significant difference in mean failure load across groups ($p = 0.01$) with the control group significantly different from both the GPC ($p = 0.03$) and PMMA group ($p = 0.02$), but there is no significant difference between the GPC cement and PMMA cement ($p = 0.92$). The GPC cement had a mean failure load, which was only 5 per cent higher than the PMMA cement.

The higher failure loads may well be due to a lack of a modulus mismatch between the two vertebrae in the control sample set of functional spinal units, especially since the mean BMD for both superior and inferior vertebrae were almost the same. In contrast, the cemented vertebrae would have a modulus mismatch possibly leading to earlier failure of the cemented functional spinal units. Unfortunately, it was not possible to assess if the failure occurred firstly in the cemented superior vertebrae or in the uncemented inferior vertebrae or simultaneously in both due to the speed at which the vertebral failure occurred.

3.6. Vertebral height loss analysis

By looking at whether there was more height loss in the superior or inferior vertebrae, it is possible to extrapolate which was the weaker of the two. Tables 4 and 5 show the

Table 4 – Superior vertebral height loss post compression compared to post cement injection height.

		Anterior	Middle	Posterior
Control	Mean (SD)	15.49% (12.02%)	19.64% (15.31%)	22.41% (16.41%)
GPC	Mean (SD)	15.56% (21.57%)	23.77% (11.14%)	21.71% (20.81%)
PMMA	Mean (SD)	14.08% (7.91%)	16.87% (15.36%)	18.77% (13.34%)

Table 5 – Inferior vertebral height loss post compression.

		Anterior	Middle	Posterior
Control	Mean (SD)	24.31% (15.86%)	24.08% (17.06%)	18.71% (9.02%)
GPC	Mean (SD)	19.11% (17.86%)	23.09% (15.27%)	16.64% (20.70%)
PMMA	Mean (SD)	7.27% (9.51%)	21.42% (15.21%)	14.26% (17.88%)

Fig. 8 – Cement thickness on lateral X-rays in different zones of vertebral body by cement type.

mean percentage height loss sustained in the lateral radiograph in the superior and inferior vertebrae.

In the superior vertebrae, there is a trend to posterior and central vertebral body height loss more so than anterior vertebral body across all three groups even in the cemented groups. Interestingly, when it came to the inferior vertebral height loss, in the control group, it was more so on the anterior and middle section of the vertebra than the posterior. In contrast, in the cemented groups, the height loss was more at the central segments likely directly underneath the more cemented and thus stiffer areas of the superior vertebrae. In the PMMA group, there was less anterior height loss and thus less anterior compression of the inferior vertebrae compared to the GPC cement group. The compression pattern of the GPC cement was more similar to that of the control than the PMMA cement group.

3.7. Cement penetration into injected vertebrae

Lastly, we analysed the penetration of the cements into the injected superior vertebrae in the lateral and AP radiographs as illustrated by Figs. 8 and 9 respectively.

As Fig. 9 shows, the GPC cement did not penetrate into the anterior segment of the vertebrae at all. This was due to its high viscosity. With the PMMA cement, there was anterior vertebral penetration but to a lesser degree to that of the middle segment and posterior part of the vertebrae illustrating that cement penetration was not uniform. Cement under filling is known to occur in vertebroplasty. This may also be due to the fact that the cadaveric tissues are more rigid and less pliable compared to the more viscoelastic living tissue. The lack of GPC cement penetration into the anterior parts of the vertebrae is concerning as it thus does not stabilise the most weakened and fractured part in the vertebrae.

In contrast, there was a more uniform penetration of both cements in the coronal plane as illustrated by Fig. 9. As the high

Fig. 9 – Cement percentage fill on AP X-rays in different zones of vertebral body by cement type on the right, middle and left sides of the vertebrae.

GPC cement viscosity limited its penetration anteriorly in the sagittal plane, it resulted in a greater penetrance or filling of the segments it did penetrate, averaging over 40 per cent compared to over 20 per cent for PMMA. There was a significant difference in the mean penetration for GPC and PMMA in the right and middle sides (both $p = 0.02$) and a trend towards significance in the left ($p = 0.06$). Overall, the PMMA was able to fill more regions of the vertebrae than the GPC cement due to its longer working time and ease of injection with lower viscosity.

4. Conclusion

In conclusion, this research aimed to further test the GPC cement with regard to developing it further for vertebroplasty. Tests were undertaken that were necessary to bring it closer to its end application in a theatre setting. Gamma irradiation used in its sterilisation did not affect the glass transition temperature (T_g); it did however cause a colour change in the GPC cement from white to grey, which is not of significance. More importantly, it was demonstrated that the mechanical properties of the cement were also altered with a decrease in both compression strength and biflexural strength of the cement post-irradiation on all tested days.

The effect of the GPC cement's target bloody vertebral environment was also investigated and showed that the GPC cement had both increased biflexural and compression strength when setting in blood compared to distilled water on all days tested. This progressively compensated for the decrease sustained in these mechanical properties as a result of the irradiation. This is the first time that this has been shown although there was not a statistical difference between blood and water media shown due to the small sample size. This will need further investigation.

Furthermore, due to its shorter working times and high viscosity, the GPC cement was found to have no penetrance into the anterior vertebrae, the very area that vertebroplasty seeks to support and stabilise since this is the area that is fractured. On vertebral testing post-GPC cement injection of the superior vertebrae in the randomised functional spinal units, there was a trend to a higher failure load required for the GPC cement group compared to the current gold standard PMMA injected group but this was not statistically significant but is encouraging.

4.1. Limitations

Weaknesses of this research include a small vertebral sample size due to limited available human cadaver numbers. The small sample size decreases the statistical power to demonstrate a significant difference between the cements on vertebral testing but it allows for this research to act as a pilot to guide further research on these cements.

4.2. Future work

Future research work will need to include testing this GPC cement with increased working times and lower viscosity and thus allow more anterior vertebral penetrance. This can be achieved with tri-sodium citrate but its effects on the GPC cement's mechanical properties will also need to be assessed. The combined effects of the gamma irradiation and the tri-sodium citrate as well as cement setting in blood that has been demonstrated in this research will need to be further investigated.

In testing the use of radiation and the addition of tri-sodium citrate on the GPC cement, there can be an added advantage of the compression strength of the GPC cement being adjusted to be a lot closer to that of the vertebral bone of different bone mineral densities while simultaneously improving its rheology. Thus, it may be possible to have a patient tailored and vertebra specific GPC cement to more closely match the bone mineral density of the neighbouring vertebrae thereby minimising the modulus mismatch and the stress riser effects, and in doing so, create a more patient-specific cementing solution for the orthopaedic surgeon when carrying out vertebroplasty.

Conflicts of interest

The authors have none to declare.

Consent

Please note that the cadavers used were from the University College Cork Anatomy Department and that the subjects had signed consent forms to the medical university donating their bodies to medicine.

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