Mechanical efficacy of vertebroplasty: Influence of cement type, BMD, fracture severity, and disc degeneration

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Abstract

Introduction: Osteoporotic vertebral fractures can be treated by injecting bone cement into the damaged vertebral body. “Vertebroplasty” is becoming popular but the procedure has yet to be optimised. This study compared the ability of two different types of cement to restore the spine’s mechanical properties following fracture, and it examined how the mechanical efficacy of vertebroplasty depends on bone mineral density (BMD), fracture severity, and disc degeneration.

Methods: A pair of thoracolumbar “motion-segments” (two adjacent vertebrae with intervening soft tissue) was obtained from each of 15 cadavers, aged 51–91 years. Specimens were loaded to induce vertebral fracture; then one of each pair underwent vertebroplasty with polymethylmethacrylate (PMMA) cement, the other with another composite material (Cortoss). Specimens were creep loaded for 2 h to allow consolidation. At each stage of the experiment, motion segment stiffness in bending and compression was measured, and the distribution of compressive loading on the vertebrae was investigated by pulling a miniature pressure transducer through the intervertebral disc. Pressure measurements, repeated in flexed and extended postures, indicated the intradiscal pressure (IDP) and neural arch compressive load-bearing (F₇). BMD was measured using DXA. Fracture severity was quantified from height loss.

Results: Vertebral fracture reduced motion segment stiffness in bending and compression, by 31% and 43% respectively (p<0.001). IDP fell by 43–62%, depending on posture (p<0.001), whereas F₇ increased from 14% to 37% of the applied load in flexion, and from 39% to 61% in extension (p<0.001). Vertebroplasty partially reversed all these effects, and the restoration of load-sharing was usually sustained after creep-consolidation. No differences were observed between PMMA and Cortoss. Pooled results from 30 specimens showed that low BMD was associated with increased fracture severity (in terms of height loss) and with greater changes in stiffness and load-sharing following fracture. Specimens with low BMD and more severe fractures also showed the greatest mechanical changes following vertebroplasty.

Conclusions: Low vertebral BMD leads to greater changes in stiffness and spinal load-sharing following fracture. Restoration of mechanical function following vertebroplasty is little influenced by cement type but may be greater in people with low BMD who suffer more severe fractures.

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Introduction

Osteoporotic vertebral fractures are a serious health problem that can lead to disabling pain, kyphotic deformity and decreased quality of life [1]. Until recently, fractures were managed conservatively, but during the last decade an increasing number have been treated using vertebroplasty. Vertebroplasty is a minimally-invasive technique that involves percutaneous injection of bone cement, usually polymethylmethacrylate (PMMA), into a weakened vertebra in order to strengthen and stabilise it. It was introduced in 1986 by Galibert et al. as a treatment for vertebral haemangiomas [2] and the same group subsequently pioneered its use in the treatment of osteoporotic vertebral
fracture [3]. Randomised trials to assess the long-term effectiveness of the procedure have not been reported, but non-randomised prospective studies report pain relief in 80–95% of patients who are treated for vertebral fracture [4,5].

Despite the success of vertebroplasty in alleviating pain, the procedure has several potential drawbacks including cement leakage, thermal damage to adjacent tissues, and increased fracture risk at adjacent levels. Clinical studies have shown that the risk of leakage is greater as the volume of injected PMMA increases, [6] and biomechanical studies suggest that factors such as the placement and viscosity of the injected cement may also be important [7–9]. The properties of the cement also influence heat damage: in the case of PMMA, the polymerisation process can lead to temperature rises sufficient to cause protein denaturation, cell necrosis and nerve ablation [10–12]. The latter effect may possibly explain pain relief following vertebroplasty.

New types of injectable bone cement have been developed in order to reduce some of these problems and to improve biocompatibility and absorbability [13]. One such cement is Cortoss (Orthovita, Malvern, USA) which is already in clinical use in Europe, and is currently undergoing clinical trials in the United States [14,15]. Cortoss is a bioactive composite material consisting of an acrylic resin reinforced with glass ceramic particles [16,17]. It is biocompatible and has better radiopacity, [16] a lower polymerization temperature, [12] and greater strength and stiffness [17,18] compared to PMMA. Clinically, it is as effective as PMMA in alleviating the pain of vertebral fracture [19].

However, new cements may introduce new problems. In cadaver experiments, fractured vertebral bodies injected with Cortoss were stronger and stiffer than those injected with PMMA [20]. High stiffness may not be entirely beneficial, because strengthening and stiffening of one vertebra may potentiate fracture at other levels [21] possibly by increasing pressure in the adjacent intervertebral disc [22–25]. The effects of vertebroplasty on adjacent non-augmented vertebrae are widely debated and must be considered in the context of epidemiological findings which suggest that up to 20% of people with untreated vertebral fractures suffer a subsequent vertebral fracture in the following year [26]. Nevertheless, several clinical studies report an increased risk of fracture following vertebroplasty [27–29], especially in adjacent vertebrae [30–32]. The mechanical effects of vertebroplasty on injured and adjacent vertebrae may also be influenced by characteristics of the treated spine, such as bone mineral density (BMD), the type and severity of fracture, and disc degeneration, but these possibilities are largely unexplored.

The present authors have developed techniques which allow both the local and more widespread mechanical effects of vertebroplasty to be assessed on cadaveric spines. A miniature pressure transducer is pulled along the mid-sagittal diameter of an intervertebral disc in order to measure the distribution of compressive loading on the adjacent vertebral bodies and neural arches [33]. The clinical relevance of these measurements has been demonstrated by the finding that abnormal load-sharing is associated with abnormal distribution of BMD, and with susceptibility to compression fracture [34]. When applied to vertebroplasty, these techniques have shown that injection of PMMA into fractured vertebral bodies partially restores intradiscal pressure and neural arch load-bearing towards pre-fracture values [35].

The objectives of the present study were to compare the ability of Cortoss and PMMA to restore normal load-sharing in cadaveric spinal motion segments with fractured vertebrae, and to assess the influence of individual factors (such as BMD and fracture severity) on the efficacy of both cements. Results should help to optimise this important new treatment for vertebral osteoporotic fracture, and assist in the selection of patients who would benefit most from it.

Materials and methods

Cadaveric material

Fifteen thoracolumbar spines were obtained from cadavers donated for medical research. There were 6 male and 9 female spines, aged 51–91 years (mean 75 years). They were stored at –20 °C in sealed bags until required for testing. Spines were subsequently thawed at 3 °C, and each was dissected to provide two “motion segments” consisting of two adjacent vertebrae with the intervening intervertebral disc and ligaments. Motion segments were obtained from spinal levels between T10 and L5, with the choice of level being determined by the need to avoid large osteophytes (which interfere with disc stress measurements) and the need to maximise use of scarce human tissue. This resulted in our rejecting one whole spine and four motion segments at different levels from three other spines. At the end of the experiment, the discs were sectioned in the transverse plane and the grade of disc degeneration determined by visual inspection, using points 1 (non-degenerated) to 4 (severely degenerated) on the scale defined by Adams et al. [36]. Specimen details are shown in Table 1.

Overview of experiments

One of each pair of motion segments from the same spine was randomly assigned to undergo vertebroplasty with a PMMA bone cement (Spineplex®, Stryker Instruments, Howmedica International, Limerick, Ireland), and the other was treated with Cortoss® (Orthovita, Malvern, PA, USA). Each motion segment was radiographed in the sagittal, frontal and transverse planes and set in plaster for loading on a materials testing machine. An initial “creep” test (1.0 kN compression for 2 h) was performed to simulate the diurnal change in intervertebral disc water content and height that would occur in life [37]. The motion segment was then fractured and a second set of radiographs was taken. The force at which fracture was initiated was determined from the load-deformation curves. Vertebroplasty was performed on the fractured vertebra, after which a final set of radiographs was obtained to demonstrate the area of cement filling. The motion segment was then compressed for a further 2 h at 1.0 kN to allow cement consolidation. Compressive and bending stiffness of the motion segment, and the distribution of compressive “stress” inside the intervertebral disc, were measured at each stage of the experiment: after the preconditioning creep test, after fracture, after vertebroplasty, and finally after consolidation. Fracture characteristics and vertebral deformity were determined from the radiographs.

Mechanical testing apparatus

Each motion segment was secured in two cups of dental plaster (Ultrahard Die Stone Iso-Type IV, Kerr S.p.A., Italy) and loaded on a computer-controlled, hydraulic materials testing machine (Dartec-Zwick-Roell, Leominster, UK). The testing rig allowed complex loading to be applied in bending and compression by means of one or two low-friction rollers (Fig. 1) [38]. The height of the rollers
could be adjusted to apply either pure compression (rollers of equal height) or compression combined with either moderate flexion (posterior roller lower) or extension (posterior roller higher). In the tests of bending stiffness, the posterior roller was completely removed to enable the specimen to flex forwards about its own natural centre of rotation, without prohibiting shearing movements or coupled rotations [39].

**Stiffness in compression and bending**

To determine compressive stiffness, each motion segment was compressed at 600 N/s while positioned in 2° of flexion. The maximum compressive load applied was either 1.2 kN or 1.5 kN, depending on specimen size and BMD. Deformation of the specimen was measured from the output of the linear variable displacement transducer (LVDT) of the materials testing machine, and compressive stiffness was determined from the maximum slope of the load-deformation curve. Validation tests showed that compressive deformation of the plaster and apparatus is less than 10% of specimen deformation, and remains constant in repeated tests.

Methods for the measurement of bending stiffness have been described in detail previously [39] so are explained only briefly here. Spinal flexion was induced by applying an off-centre compressive force by means of the front roller only (Fig. 1). Vertebral rotation was measured by attaching 5 mm diameter reflective markers to the metal cups, the roller axis, and to 10 mm long pins inserted into the anterior and posterior margins of each vertebral body. Movements of the markers in the sagittal plane were tracked at 50 Hz by a 2-D MacReflex infrared motion analysis system (Qualisys Ltd, Goteborg, Sweden). Custom software was used to calculate the rotation angle between adjacent vertebral bodies using simple trigonometry based on four markers, two on each side of the disc.

### Table 1

<table>
<thead>
<tr>
<th>Spine level</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Disc degeneration</th>
<th>BMD (g/cm³)*</th>
<th>Yield strength (kN)</th>
<th>Height loss (mm)†</th>
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<tbody>
<tr>
<td>C</td>
<td>S</td>
<td>C</td>
<td>S</td>
<td>C</td>
<td>S</td>
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<tr>
<td>L1–L2</td>
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<td>2</td>
<td>0.086</td>
<td>0.136</td>
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<td>0.300</td>
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<td>0.335</td>
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<td>0.099</td>
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<td>0.139</td>
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<td>1.5</td>
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<td>0.241</td>
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<td>0.113</td>
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<td>0.154</td>
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<td>0.115</td>
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<tr>
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<td>3</td>
<td>0.129</td>
<td>0.116</td>
<td>2.3</td>
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</table>

Mean (S.D.) 75 (13) 0.169 (0.081) 0.184 (0.107) 2.5 (1.2) 2.9 (1.3) 2.34 (0.57) 2.16 (0.49)

C=Cortoss; S=Spineplex; *BMD refers to fractured vertebra; †height loss refers to whole motion segment.

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Fig. 1. Apparatus used to load cadaver spine motion segments. The height of the rollers was adjusted so that the specimen was compressed at the desired flexion/extension angle. The posterior roller was removed altogether for tests of bending stiffness. Stress profilometry was performed by pulling a pressure transducer along the mid-sagittal diameter of the loaded disc.
of the upper and lower cups. Comparison of rotation angles obtained from markers on the vertebral bodies and those on the cups indicated if there was any slipping of the specimen in the cups [39]. Bending moment acting on the specimen was calculated by multiplying the applied compressive force (measured by the Dartec load cell) by the lever arm of that force relative to the geometric centre of the disc. This lever arm increased as the flexion angle increased and was calculated from the dimensions of the apparatus and the position of the roller as measured by the MacReflex. Bending moment-rotation angle graphs were then plotted as shown in Fig. 2 to determine the bending stiffness (in forwards flexion). Bending stiffness was defined as the gradient of the tangent to the linear region of the graph at 10 Nm.

Stress profilometry and compressive load-sharing

A miniature pressure transducer (Gaeltec, Dunvegan, Scotland), side-mounted in a 1.3-mm diameter needle, was pulled through the intervertebral disc, along the mid-sagittal diameter, while the motion segment was subjected to a compressive force of 1.0 kN or 1.5 kN [35,38]. Profiles of vertical and horizontal compressive stress were obtained in successive tests with the transducer membrane facing upwards then horizontally. Profiles were obtained with the specimen positioned in 2° of extension, to simulate the erect standing posture, [40,41] and in 2–8° of flexion (depending on specimen mobility) to simulate a slightly stooped posture. Stress profiles were analysed to determine the intradiscal pressure (IDP) and the maximum stress in the anterior (S\textsubscript{A}) and posterior (S\textsubscript{P}) annulus (Fig. 4). IDP was calculated as the average pressure in the “functional nucleus” of the disc where vertical and horizontal stresses were approximately equal. Anterior (S\textsubscript{A}) and posterior (S\textsubscript{P}) stress peaks were calculated by subtracting IDP from S\textsubscript{A} and S\textsubscript{P} respectively.

Stress “integration” was used to quantify the compressive force acting on the anterior (F\textsubscript{A}) and posterior halves (F\textsubscript{P}) of the disc (and vertebral body) as described previously [33]. The compressive force acting on the neural arch (F\textsubscript{N}) was calculated by subtracting F\textsubscript{A} and F\textsubscript{P} from the total compressive force applied to the specimen (1.5 kN). F\textsubscript{A}, F\textsubscript{P} and F\textsubscript{N} were each expressed as a percentage of the applied compressive force.

Bone mineral content and density

Bone mineral content (BMC) of the vertebrae was measured prior to mechanical testing using a PIXImus dual photon X-ray absorptiometer (Lunar Corporation, Madison, WI, USA). BMD of the fractured vertebral body was calculated as BMC/volume, where the volume of the vertebral body was determined, after mechanical testing, by a water displacement method. Validation tests showed that BMC correlated highly with ash weight (\(r^2 = 0.987\)) [34].

Vertebral fracture and deformity

Each motion segment was positioned in 10° of flexion to simulate a forwards stooped posture, and compressed at 3 mm/s until the load-deformation curve indicated the yield point by a reduction in gradient (stiffness). Fracture was confirmed from radiographs taken before and after overload. The severity of injury was quantified as the specimen height loss (in mm) measured by the LVDT mounted on the actuator of the materials testing machine (Fig. 1) while the specimen was subjected to a reference compressive force of 1.0 kN. Vertebral deformity was assessed from measurements of the wedge angle, defined as the angle between the superior endplate line and the horizontal line bisecting the vertebral body on lateral radiographs [42]. The superior endplate line was determined by connecting the most posterior and anterior margins of the superior endplate [43].

Vertebroplasty

In the PMMA group, vertebroplasty was performed on the fractured vertebra using two 10 G vertebroplasty needles. These were gently tapped into the vertebral body by the transpedicular route, one needle being introduced through each pedicle. A radiograph confirmed that the tips of the needles were located in the anterior/inferior quadrant of the vertebral body. PMMA cement was then prepared by mixing 20 g of powder with 10 ml of monomer liquid. The cement was drawn into a 5 ml syringe while it was still at a low viscosity. The stylets of the vertebroplasty needles were removed and 3.5 ml of cement were injected through each needle into the vertebral body so that 7.0 ml in total was injected. This volume should be sufficient to restore vertebral body strength and stiffness in thoracic and lumbar vertebrae [44]. After injection, stylets were reinserted into the needle barrel and the needles removed from the vertebra.

In the Cortoss group, vertebroplasty was performed unilaterally using a single aliquot needle that was gently tapped into the anterior/inferior quadrant of the vertebral body by the transpedicular route. The stylet was then withdrawn through the needle after which a micro-reamer was inserted to create a channel for easier insertion of the catheter. The micro-reamer was removed and the catheter inserted through the needle into the vertebral body. Cortoss was provided in a double-chamber cartridge which attached to a specially designed mix tip and delivery gun which expelled material from two cartridges through the mix tip into a 1 ml syringe where polymerization started. The resulting resin was injected through the catheter into the vertebral body. Injection volume ranged from 3.5 ml to 6.2 ml depending on spinal level, in accordance with the manufacturer’s guidelines. After injection, the catheter and needle were removed from the vertebra.

Statistical analysis

Intra-observer and inter-observer reliability of wedge angle measurements obtained from radiographs were evaluated from the intraclass correlation coefficient. Repeated measures analysis of variance (ANOVA) was used to compare measurements following each intervention, with cement type as a between-subjects factor. Where a significant main effect was found, post-hoc paired comparisons with appropriate Bonferroni adjustment were employed to identify where the differences arose. BMD, degree of disc degeneration, and fracture severity (height loss) were compared in the two cement groups using matched pair \(t\)-tests. Multiple regression analysis was used to determine how BMD, disc degeneration and height loss influenced post-fracture and post-vertebroplasty changes. SPSS 11.5\textsuperscript{®} was used for all statistical analyses.

Results

Vertebral fracture and deformity

Yield strengths recorded during the fracture tests are shown in Table 1 and ranged from 1.1 kN to 5.8 kN. Radiographs showed that 22/30 specimens sustained a fracture of the lower vertebral body and 8 sustained a fracture of the upper vertebral
body. Wedge fractures (indicated by an increase in vertebral wedge angle) occurred in 28/30 vertebral bodies, and two failed by endplate fracture. Damage to the anterior cortex was apparent in 29/30 vertebrae.

**Vertebroplasty**

This was successfully performed in all specimens without any visible leakage (Fig. 3). Baseline results showed no significant difference in BMD, grade of disc degeneration, or severity of fracture between the Cortoss and PMMA groups (Table 1).

**Motion segment stiffness**

Fracture and vertebroplasty both affected compressive and bending stiffness but effects were similar in the two cement groups (Table 2). Fracture reduced compressive stiffness by 42% and 44% in the Cortoss and PMMA groups respectively, and bending stiffness fell by 26% and 35%. Vertebroplasty increased compressive stiffness and bending stiffness by similar amounts in both groups, but after 2 h of creep consolidation, both stiffness parameters remained lower than pre-fracture values.

**Stress profilometry**

Stress profiles obtained before fracture were typical for discs of similar age and grade of degeneration: stress peaks were often observed in the annulus, and these became more marked posteriorly when specimens were compressed in extension to simulate erect standing [38]. Profiles obtained after fracture and after vertebroplasty revealed characteristic changes in the distribution of compressive stress (Fig. 4). However, the extent of these changes was not significantly affected by cement type (Table 2). Fracture reduced IDP by 39% and 46% in flexion, and by 57 and 65% in extension, whereas posterior stress peaks increased by 65% and 196% in flexion, in the PMMA and Cortoss groups respectively. These fracture-induced changes were significantly reversed by vertebroplasty, and except for IDP in extension, this restoration towards pre-fracture values was maintained following creep consolidation.

Stress integration revealed that load-bearing by the anterior and posterior halves of the disc, and by the neural arch, was influenced by fracture and by vertebroplasty but not by the type of cement. Based on average data for the two groups, fracture reduced loading on the anterior half of the vertebral body ($F_{AN}$) from approximately 50% to 28% of applied load in flexion, and from 24% to 11% in extension. Fracture also reduced loading of the posterior half of the vertebral body ($F_{PN}$) in extension, from 38% to 28%. Reduced load-bearing by the disc resulted in increased load bearing by the neural arch ($F_{AN}$) after fracture, from 14% to 37% in flexion, and from 39% to 61% in extension. Vertebroplasty partially reversed these changes in load-sharing, and in flexed posture this restorative effect was maintained or enhanced following creep loading. In extension, creep loading tended to reduce the effects of vertebroplasty (Table 2).

**Vertebral deformity**

Methods used to assess vertebral deformity were reproducible, with ICC’s for inter-observer and intra-observer error in vertebral wedge angle being 0.95 and 0.99 respectively. Vertebral wedge angle changed significantly as a result of the interventions but was not influenced by cement type. Paired comparisons showed that the wedge angle increased after fracture but showed no significant change following vertebroplasty in either group (Table 2).

**Influence of BMD, disc degeneration, and fracture severity**

In the absence of any differences between specimens injected with Cortoss or PMMA, data for the 30 specimens were pooled for this analysis, the results of which are shown in Table 3. Vertebrae with low BMD suffered more severe fractures ($r=0.47; p=0.009$), and stepwise linear regression showed that BMD, fracture severity and disc degeneration all influenced changes in several biomechanical parameters in a posture-dependent manner (Table 3). Following fracture, BMD had a more marked effect in flexion where low BMD was associated with a greater fall in IDP and $F_{PN}$ and a greater increase in $F_{AN}$ (Fig. 5). Severity of fracture tended to have a greater influence in extension where specimens with more severe fractures showed a greater fall in IDP (Fig. 6), a greater increase in neural arch load-bearing, and a greater reduction in $F_{AN}$. Disc degeneration affected disc load-sharing: in flexed postures, more degenerated discs showed a greater fall in $SP_A$ and $F_{AN}$ after fracture. BMD, fracture severity and disc degeneration also influenced the mechanical changes following vertebroplasty (Table 3). Low BMD of the fractured vertebral body was associated with a greater increase in IDP towards pre-fracture values in flexion (Fig. 7) and a greater increase in $F_{AN}$ in extension. Specimens with more severe fractures showed a greater increase in $SP_A$ in extension, and those with more degenerated discs showed a greater reduction in annulus stress peaks ($SP_p$) in flexion following vertebroplasty.
The efficacy of vertebroplasty also depended on the extent of changes induced by fracture. For example, a greater fall in IDP after fracture was associated with a greater increase after vertebroplasty ($r = 0.524$, $p < 0.001$ in flexion; $r = 0.515$, $p = 0.004$ in extension). Similarly, a greater increase in neural arch load-bearing as a result of fracture was associated with a greater reduction as a result of vertebroplasty ($r = 0.511$, $p < 0.001$).

**Discussion**

**Summary of findings**

Vertebral fracture reduced specimen height and stiffness, increased the wedge angle of the vertebral body, reduced pressure in the nucleus of adjacent intervertebral discs, and transferred compressive load-bearing to the neural arch. Vertebroplasty using PMMA or Cortoss was equally effective in partially reversing most of these changes towards pre-fracture values. However, vertebroplasty did not change vertebral wedge angles in the unloaded specimens. Specimens with low BMD, disc degeneration and more severe fractures showed greater mechanical changes after fracture, and benefited most from vertebroplasty.

**Strengths and weaknesses of the study**

The matched pair design (using two motion segments from each spine, randomly allocated to the two cement groups) minimised the influence of confounding factors such as age, gender, spinal level, disc degeneration, vertebral size and vertebral BMD, so that small differences would be detected — if present. In fact no differences were found between the two cement types, but the study design gives confidence in this null result. Other strengths include: validated techniques to characterise the distribution of loading on the whole of the fractured and adjacent vertebrae; the use of elderly human specimens rather than young animal tissues; and the use of physiologically-reasonable complex loading in bending and compression to reproduce anterior wedge fractures and to simulate specific postures adopted by living people. These details are important because anterior wedge fractures are the most common type of vertebral fracture in patients with osteoporosis [45–47] and in life are associated with forward mechanical changes after fracture, and benefited most from vertebroplasty.

The average (S.D.) results obtained at different stages of the experiment are shown in Table 2.

<table>
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<tr>
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<th>Post-fracture</th>
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<td>Cortoss</td>
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<td>10.6</td>
<td>39.1</td>
<td>33.9</td>
</tr>
<tr>
<td>FN-ext (%)</td>
<td>39.9</td>
<td>37.3</td>
<td>61.2</td>
<td>60.3</td>
</tr>
<tr>
<td>Compressive</td>
<td>2951</td>
<td>3086</td>
<td>1717</td>
<td>1716</td>
</tr>
<tr>
<td>stiffness (N/mm)</td>
<td>1029</td>
<td>1252</td>
<td>414</td>
<td>457</td>
</tr>
<tr>
<td>Bending</td>
<td>6.90</td>
<td>8.06</td>
<td>5.10</td>
<td>5.52</td>
</tr>
<tr>
<td>stiffness (Nm/deg)</td>
<td>1.72</td>
<td>2.28</td>
<td>0.50</td>
<td>1.12</td>
</tr>
<tr>
<td>Wedge angle (deg)</td>
<td>5.3</td>
<td>6.2</td>
<td>5.9</td>
<td>8.6</td>
</tr>
</tbody>
</table>

Significance (final column) indicates main effects demonstrated by repeated measures ANOVA. Differences between cement groups were not significant for any of the measured parameters. Post-hoc paired comparisons indicate differences from pre-fracture ($p < 0.05$; $p < 0.01$; $p < 0.001$) and post-fracture ($p < 0.05$; $p < 0.01$; $p < 0.001$) values.
bending and lifting activities [48] when compressive and bending stresses are high [49,50]. Intervertebral discs press on both of their adjacent vertebral bodies, so stress profilometry indicates how fracture and vertebroplasty at one spinal level affects load-sharing at the adjacent level. Specifically, wedge fractures decompress the disc, increasing load-bearing by the neural arches, and reducing loading of the anterior vertebral bodies. This fall in anterior load-bearing may reduce fracture risk in the short-term. However, over time, stress-shielding of the anterior vertebral body will accelerate bone loss in this region and ultimately increase the risk of further fracture [34].

Vertebroplasty was carried out as recommended clinically, and although this resulted in different volumes of Cortoss and PMMA being used, it ensured clinical relevance. It also demonstrated that lower volumes of Cortoss could achieve similar biomechanical results to higher volumes of PMMA. This may have clinical implications in relation to the risk of cement leakage [6].

Weaknesses of the study include using tissue after death and freeze-thawing cycles, both of which have some small influence on the spine’s mechanical properties [51], and using a consolidation time of only 2 h rather than days or weeks. The outer vertebral endplates of motion segments are protected by plaster, so there are arguments for using specimens with three or more vertebrae. However, long spine specimens introduce other problems, including a tendency to buckle under high compressive loading [51]. The generalisability of the study may be limited by the lack of specimens from upper thoracic levels. However, the types of fracture induced in this study are also observed in upper thoracic vertebrae, so similar results might be expected. Future studies should be carried out to substantiate this.

Relationship to other studies

In the present study, most specimens were induced to fail by fracture of the anterior vertebral body, increasing the vertebral wedge angle, but changes observed following fracture and subsequent vertebroplasty were similar to those reported previously following end-plate fracture [35]. Some mechanical changes tend to be greater following wedge fracture, but differences are small, suggesting that the site of fracture is not a major determinant of the changes in stiffness and load-sharing which follow vertebral fracture and subsequent vertebroplasty. From a biomechanical perspective, cement augmentation appears to be suitable for treating either type of vertebral fracture. Vertebroplasty has been reported to have minimal effect on intradiscal pressure [52], but this was in a small group of younger specimens tested at lower load levels.

The influence of individual factors on load-sharing following fracture and vertebroplasty agree with previous studies. Cadaver experiments have shown that BMD is an important predictor of vertebral strength and stiffness [34,53] which explains the link between BMD and fracture severity. It is only to be expected that specimens with low BMD and more severe fractures would show greater changes in mechanical function following fracture. However, these same specimens also showed the greatest restoration of IDP and neural arch load bearing as a result of vertebroplasty. Similar effects have been reported in isolated vertebral bodies where augmentation using the same (relative) volume of cement produced greater increases in strength and stiffness in specimens with lower BMD [54,55]. More recently, a study using multilevel spinal segments indicated that specimens with low BMD showed the greatest increases in strength and stiffness following kyphoplasty [53]. These findings all suggest that spines with low BMD and more severe fractures gain most mechanical benefit from vertebroplasty.

Explanation of results

Changes in spinal load-sharing following vertebral fracture depend on changes within the intervertebral disc. Nucleus pressure (IDP) falls after fracture because the damaged vertebral body deforms more under load, allowing more space for the nucleus which is effectively a pressurised fluid [57]. A decompressed disc bulges radially and loses height, [58] giving slack to intervertebral ligaments, and reducing motion segment stiffness in bending as well as compression [39]. Reduced load-
bearing by the nucleus increases stress concentrations in the annulus, particularly in the posterior annulus [33,38,56]. Disc narrowing increases neural arch load-bearing by bringing adjacent vertebrae closer together, increasing contact stresses in the zygapophyseal joints, particularly in erect or extended postures [59]. (A similar but reduced effect occurs each day as a result of diurnal changes in disc water content and height [60,61].) Zygapophyseal joint impaction explains why greater height loss following fracture is associated with increased neural arch load-bearing and a consequent fall in IDP that is most marked in extension (Fig. 6). Vertebroplasty tends to reverse all of these changes by stiffening the vertebral body and restoring more normal load-bearing to the adjacent disc.

Low BMD leads to greater height loss and to greater changes in load-sharing after fracture because less dense vertebrae have a reduced stiffness, and tend to be more brittle [62], so that more extensive damage can be sustained when the elastic limit is exceeded [63]. These effects are greater in flexed postures because flexion transfers almost all of the spinal compressive force on to the vertebral body, in particular to its anterior half, [34] and this is the region of a vertebra most affected by reduced BMD in osteoporosis [34]. In erect postures, a substantial portion of the compressive load is borne by the neural arch, especially when the intervertebral disc is degenerated and narrowed, [34] and so the neural arch is then able to protect the vertebral body to a certain extent from severe loading and damage.

Unanswered questions and future research

The stress profilometry technique used in the present study is well-suited to quantifying the distribution of compressive load acting on different regions of a vertebra, and to explain the mechanisms of vertebral failure in general terms. However it is unable to characterise the exact distribution of compressive (or other) stresses within the vertebral body. Further studies, perhaps involving finite element modelling, are required to explain the precise mechanisms of fracture, and to analyse systematically the influence of cement volume and placement on the efficacy of vertebroplasty.

Table 3

<table>
<thead>
<tr>
<th>Influence of BMD, disc degeneration and fracture severity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Post-fracture</td>
</tr>
<tr>
<td>BMD</td>
</tr>
<tr>
<td>r</td>
</tr>
<tr>
<td>IDP-flex (MPa)</td>
</tr>
<tr>
<td>IDP-ext (MPa)</td>
</tr>
<tr>
<td>SPa flex (MPa)</td>
</tr>
<tr>
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<td>Fp-flex (%)</td>
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<tr>
<td>Fp-ext (%)</td>
</tr>
<tr>
<td>Fn-flex (%)</td>
</tr>
<tr>
<td>Fn-ext (%)</td>
</tr>
</tbody>
</table>

Results of stepwise multiple regression analyses showing how changes in mechanical parameters following fracture and vertebroplasty depend on BMD, disc degeneration, and fracture severity (measured as height loss). Correlation coefficients (r) and p values are shown for significant relationships only.

Fig. 6. Specimens that suffered a greater loss of height after fracture showed a greater fall in IDP. (Data for specimens loaded in extension.)

Fig. 7. Specimens with low BMD showed a greater increase in IDP after vertebroplasty. (Data for specimens loaded in flexion.) Regression data refer to pooled results from specimens injected with either type of cement.
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References


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