Adjacent vertebral failure after vertebroplasty

A BIOMECHANICAL INVESTIGATION

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Vertebroplasty, which is the percutaneous injection of bone cement into vertebral bodies has recently been used to treat painful osteoporotic compression fractures. Early clinical results have been encouraging, but very little is known about the consequences of augmentation with cement for the adjacent, non-augmented level.

We therefore measured the overall failure, strength and structural stiffness of paired osteoporotic two-vertebra functional spine units (FSUs). One FSU of each pair was augmented with polymethylmethacrylate bone cement in the caudal vertebra, while the other served as an untreated control.

Compared with the controls, the ultimate failure load for FSUs treated by injection of cement was lower. The geometric mean treated/untreated ratio of failure load was 0.81, with 95% confidence limits from 0.70 to 0.92, (p < 0.01). There was no significant difference in overall FSU stiffness. For treated FSUs, there was a trend towards lower failure loads with increased filling with cement (r² = 0.262, p = 0.13).

The current practice of maximum filling with cement to restore the stiffness and strength of a vertebral body may provoke fractures in adjacent, non-augmented vertebrae. Further investigation is required to determine an optimal protocol for augmentation.

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The technique of percutaneous vertebroplasty with polymethylmethacrylate (PMMA) was initially presented as a treatment for vertebral angiomas. Cannulas are passed through the pedicles into the vertebral body and cement is injected to provide strength and support. More recently, this technique has also been used to treat painful osteoporotic compression fractures. Early clinical results were very encouraging since relief from pain was immediate and reliable, and the complication rate was low. Biomechanical studies have shown significant increases in parameters of stability after augmentation of vertebral bodies with PMMA and also with calcium phosphate cements. Vertebroplasty is therefore gaining popularity for the treatment of patients with osteoporosis of the spine. Very little is known, however, about the consequences of augmentation on the adjacent, non-augmented level. In single cases, fresh compression fractures have been reported in vertebrae caudal or cranial to that augmented with PMMA. This could be an expression of the natural course of the disease since existing fractures are also a strong, independent predictor of the risk of future vertebral fracture. Additional fractures, however, may also be provoked by an adjacent rigid reinforcement.

To test this hypothesis, we performed a biomechanical study comparing the overall structural stiffness and failure strength of paired osteoporotic spinal segments, one of which was augmented with PMMA in the caudal vertebra.

Materials and Methods

We obtained ten human cadaver thoracic spines within 24 hours of death, sealed in plastic and frozen at -20°C. Before testing they were thawed and the bone mineral density (BMD) determined in the lateral projection on each vertebra using the dual-energy x-ray absorptiometry technique (DXA; Hologic QDR 2000, Hologic Inc, Waltham, Massachusetts). Osteoporosis was defined according to the World Health Organisation (WHO) as a BMD of more than 2.5 standard deviations below the mean of a young healthy reference population of the same gender (‘T-score’). Using our local reference database, osteoporosis corresponded to a BMD of <0.75 g/cm² and the cut-off for osteopenia (i.e. T-score <-1.0 sd according to the WHO criteria) was at
0.95 g/cm². Radiographs in two planes were used to exclude specimens with lytic lesions or other bony abnormality apart from osteopenia.

The four lower thoracic specimens were separated into two functional spine units (FSUs, T9-T10 and T11-T12). All soft tissues, except for the intervertebral ligaments and facet joint capsules, were removed from the specimens. To minimise the effects of variability in BMD and the level of treatment, specimen pairs were sorted according to BMD from lowest to highest, and then assigned in an alternating sequence to two groups for level of vertebroplasty treatment (T10 or T12). From each pair of FSUs, one was selected for treatment by injection with cement while the other served as an untreated control.

**Cementing technique.** We placed 2 mm Kirschner (K-) wires transpedicularly into the centre of the caudal vertebral body of the FSU. Insertion was defined by anatomical landmarks and inspection and controlled by lateral fluoroscopy. Bone-marrow biopsy needles (8 gauge; Somatex, Reitzenneudorf, Germany) were guided over the K-wires. The final positioning was again fluoroscopically documented.

For augmentation, a low-viscosity bone cement (Palacos E-flow; Essex Chemie AG, Lucerne, Switzerland) was used, enhanced with non-ionic liquid contrast dye (Iopamiro 300; Bracco SA, Switzerland) in the ratio of 10 ml of contrast dye to one portion of PMMA. The material was filled into 2 ml standard syringes. Two minutes after mixing the cement, injection of the vertebral bodies was begun using fluoroscopic control in the lateral view. This procedure is equivalent to the in vivo intraoperative technique. Cement was injected bipedicularly, with the degree of cement filling limited by the increasing viscosity as the cement polymerised or by a major leakage of cement.

After the procedure, and before mechanical testing, anteroposterior and lateral radiographs were taken. The volume of the vertebral body was calculated, assuming an elliptical shape for the endplates with compensation for concavity of the body, using dimensions measured from the plain radiographs.

**Mechanical testing.** Impressions of the cranial and caudal ends of each FSU were made in semicured moulding material (Beracryl, Fuhlenbach, Switzerland), ensuring parallel orientation of the outer surfaces of each moulded block as well as a perpendicular orientation of the FSU with respect to the loading axis. Mounting blocks formed in this way maintained the FSU in position during mechanical testing, without providing any structural reinforcement to the vertebral bodies. The FSU and mounting blocks were centred between the platens of a servohydraulic testing machine (Bionix; MTS Systems Corporation, Eden Prairie, Minnesota). The outer surface of the FSU was kept moist throughout testing.

In order to characterise the structural stiffness of the FSU, the specimens were subjected to cyclic sinusoidal dynamic compression (50-450 N compression, 1 Hz frequency, 600 cycles total), before and after treatment by vertebroplasty. Load and displacement data were recorded at a frequency of 40 Hz, and the stiffness determined from a linear regression of the loading portions of the load-versus displacement curves from the final ten loading cycles. Immediately after the second dynamic load series, the FSU was compressed at a constant displacement rate of 0.5 mm/s to a total compression of 10 mm. Load and displacement data were recorded at 100 Hz, and the stiffness and failure load of the FSU were measured. The failure load of the FSU was defined as the peak load measured, and the stiffness as the slope of the near-linear portion of the load-versus displacement curve. After testing, anteroposterior and lateral radiographs were taken.

Differences in the overall stiffness and failure load of the FSU between cement-augmented specimens and unaug-
Fig. 2
Load-displacement curves in compression for one matched pair of spinal units. The ultimate failure load of the segments which were augmented with cement is substantially lower than that of the non-augmented segment. The overall stiffness of the spinal segments (the slope of the load-displacement curve) was not affected by augmentation with cement.

Fig. 3
Comparison of the failure strength for all paired segments. The mean failure load for segments treated by cement augmentation was 19.0% lower than that of untreated segments.

Fig. 4a
Radiographs showing a typical pattern of failure observed after destructive mechanical testing. Reduction in vertebral height (a) from the FSU shown in Figure 1, and wedge-shaped deformity (b) are apparent in the cranial vertebra.

Fig. 4b

Results
In the treated vertebrae, the total volume of cement injected was 8.8 ± 1.8 ml (SD) per vertebral body, which was approximately 22.9 ± 7.4% of the total volume of the vertebral body. There was a negative correlation between the possible degree of filling and the BMD, with a lower degree of filling in vertebrae with denser bone ($r^2 = 0.42$, $p = 0.04$). The pattern of filling was consistent for all vertebrae. Anteroposterior and lateral radiographs showed an even distribution of cement to both sides of the vertebral body (Fig. 1). Extrusion of cement from the vertebral body through nutritional vessels was observed in four cases.

Comparison of the dynamic load versus displacement curves before and after vertebroplasty for treated FSUs showed no significant differences in overall stiffness for loads up to 1000 N. The geometric mean postvertebroplasty/prevertebroplasty ratio of low-load stiffness was 0.94, with 95% CI from 0.76 to 1.12 and a p value of 0.45. No significant differences were found in low-load stiffness between treated FSUs and the untreated controls from the same spine. The geometric mean treated/untreated ratio of low-load stiffness was 0.90, with 95% CI from 0.69 to 1.11 ($p = 0.18$).

At higher loads approaching the failure load of the
vertebra, there was no significant difference in stiffness between the augmented FSUs and the untreated controls. The geometric mean treated/untreated ratio of stiffness was 0.89, with 95% CI from 0.70 to 1.06 (p = 0.06). The ultimate failure load, however, for FSUs treated by injection of cement was, on average, 19.0% lower than that of the adjacent, untreated FSU (3648 ± 1269 N and 4435 ± 1441 N, respectively (Figs 2 and 3)). The geometric mean treated/untreated ratio of failure load was 0.81, with 95% CI from 0.70 to 0.92 (p < 0.01). In the treated FSUs, failure always occurred in the non-augmented, cranial vertebral body. Radiologically, the fracture was either not visible or the vertebral body showed a wedge-shaped crush fracture (Fig. 4). In the untreated control FSUs, failure occurred in both cranial and caudal vertebral bodies, with a similar pattern of fracture.

For all FSUs, there was a significant correlation between increasing BMD and increasing failure load (r² = 0.402, p < 0.01). For treated FSUs, there was a trend towards a lower failure load with increased degree of filling with cement (r² = 0.262, p = 0.13). There was no correlation between BMD and overall stiffness of the FSU, with or without augmentation, or between overall stiffness and failure load.

Discussion

Percutaneous vertebroplasty with PMMA offers an efficient means of augmenting vertebral bodies. Clinical experience in the treatment of lytic vertebral lesions and, more recently, osteoporotic deficiencies have been very encouraging in the short term.²,⁷,¹⁰ Biomechanical studies have clearly demonstrated the restoration of lost strength and stiffness of fractured vertebral bodies, or the reinforcement of intact but weak vertebral bodies. Very little is known, however, of the effect of vertebroplasty on the overall alignment and function of the spine, particularly the adjacent vertebrae. We therefore investigated entire FSUs instead of single vertebrae.

Our findings show that the failure strength in compression of FSUs treated by augmentation of the caudal vertebra with PMMA is lower than that of untreated FSUs. Patterns of failure in the vertebral body observed in these tests were similar to those which are seen clinically. The mechanism for such failure is not clear, but it is possible that the increased stiffness of the augmented vertebrae alters the biomechanics of load transfer to the adjacent vertebrae. The early failure of the adjacent, non-augmented level may be caused by a ‘stress-riser’ effect and a significant disparity in biomechanical properties between the two involved vertebral bodies. It is important to note that all augmented vertebrae were filled bipedicularly with the maximum possible amount of PMMA, resulting in a mean filling of 22.9% of the vertebral volume. Previous studies have shown that this results in significant increases for both strength and stiffness of the individual osteoporotic vertebral body. The results of our study, however, suggest that rigid augmentation may also provoke failure of the adjacent, non-augmented level. It is not known whether this finding could be prevented by either less filling or different augmentation materials. It therefore seems difficult to define the optimal amount of filling as well as the ideal reinforcement. Belkoff et al¹¹ suggested that merely restoring the initial strength of osteoporotic vertebrae after fracture is not an adequate goal for treatment by vertebroplasty, but rather that attempts should be made to restore the strength of the vertebral body to healthy, normal values. It has been shown by Tohmeh et al¹² that unipedicular or bipedicular augmentation with cement increases the strength of the vertebral body significantly above that of intact, osteoporotic vertebrae. While Belkoff et al¹¹ and Tohmeh et al¹² found no differences in stiffness of the vertebral body before and after augmentation, with applied compressive loads of up to 1100 N, Berlemann et al reported that augmentation with cement increased the maximum stiffness of individual osteoporotic vertebral bodies by a mean of 174%. This may also have been due to the different types of cement which were used in these studies. Consequently, in view of the observed trend for lower failure strength of adjacent vertebrae with increased filling with cement, caution should be exercised in recommending the current practice of maximum possible filling to obtain a satisfactory clinical outcome. Indeed, clinical studies have not demonstrated a correlation between the volume of filling and the amount of relief from pain,¹³ indicating possible mechanisms for relief from pain other than those which are simply mechanical.

There were no significant differences in the overall stiffness of treated FSUs before and after augmentation, and there was no significant difference in stiffness between untreated and treated FSUs. This is not surprising, since the overall compliance of mult vertebral spinal segments is likely to be influenced by factors other than stiffness of the vertebral body alone as no correlation was found between the properties of the intact vertebral body and the overall stiffness of the FSU. While it has been shown that augmentation of fractured vertebrae with cement can increase the overall stiffness of damaged segments,¹⁴ it is likely that the properties of the intervertebral disc determine the stiffness of the segment once individual vertebral bodies have been restored to their original level of stiffness. Since there was no correlation between the stiffness of the FSU and the failure load, stiffness itself was not a contributory factor for failure of the segment.

Our study has shown that augmentation of vertebral bodies with cement has an influence on the ultimate failure strength of spinal segments. The failure strength of FSUs treated by augmentation with cement in one vertebral body was lower than that of untreated controls. In order to understand better the subtle changes in spinal biomechanics after vertebroplasty, further studies are planned to measure directly local bone strains in augmented and adjacent vertebral bodies.
brae, before and after treatment, under a variety of physiological loading conditions.

No benefits in any form have been received or will be received from a commercial party related directly or indirectly to the subject of this article.

References


