Experimental Study

Effect of Augmentation Material Stiffness on Adjacent Vertebrae after Osteoporotic Vertebroplasty Using Finite Element Analysis with Different Loading Methods

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Background: Vertebroplasty is an effective treatment for osteoporotic vertebral fractures, which are one of the most common fractures associated with osteoporosis. However, clinical observation has shown that the risk of adjacent vertebral body fractures may increase after vertebroplasty. The mechanism underlying adjacent vertebral body fracture after vertebroplasty is not clear; excessive stiffness resulting from polymethyl methacrylate has been suspected as an important mechanism.

Objectives: The aim of our study was to compare the effects of bone cement stiffness on adjacent vertebrae after osteoporotic vertebroplasty under load-controlled versus displacement-controlled conditions.

Study Design: An experimental computer study using a finite element analysis.

Setting: Medical research institute, university hospital, Korea.

Methods: A three-dimensional digital anatomic model of L1/2 bone structure was reconstructed from human computed tomographic images. The reconstructed three-dimensional geometry was processed for finite element analysis such as meshing elements and applying material properties. Two boundary conditions, load-controlled and displacement-controlled methods, were applied to each of 5 deformation modes: compression, flexion, extension, lateral bending, and torsion.

Results: The adjacent L1 vertebra, irrespective of augmentation, revealed nearly similar maximum von Mises stresses under the load-controlled condition. However, for the displacement-controlled condition, the maximum von Mises stresses in the cortical bone and inferior endplate of the adjacent L1 vertebra increased significantly after cement augmentation. This increase was more significant than that with stiffer bone cement under all modes, except the torsion mode.

Limitations: The finite element model was simplified, excluding muscular forces and incorporating a large volume of bone cement, to more clearly demonstrate effects of bone cement stiffness on adjacent vertebrae after vertebroplasty.

Conclusion: Excessive stiffness of augmented bone cement increases the risk of adjacent vertebral fractures after vertebroplasty in an osteoporotic finite element model. This result was most prominently observed using the displacement-controlled method.

Key words: Bone cements, displacement-controlled method, finite element analysis, load-controlled method, osteoporosis, osteoporotic fracture, polymethyl methacrylate, vertebroplasty

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Osteoporosis is characterized by reduced bone mass and disruption of bone architecture, resulting in increased bone fragility and fracture risk (1). Vertebroplasty is an effective treatment for osteoporotic vertebral fractures, which are common fractures associated with osteoporosis (2). Polymethyl methacrylate (PMMA) is the most widely used bone cement for vertebroplasty; it strengthens augmented vertebrae and can result in significant and rapid pain relief in 80 – 90% of patients (3). However, clinical observation has shown that the risk of adjacent vertebral body fractures may increase after vertebroplasty (4-6). The mechanism underlying adjacent vertebral body fracture after vertebroplasty is not clear; excessive stiffness resulting from PMMA use has been suspected as an important mechanism.

However, efforts to identify whether the stiffness resulting from PMMA increases the risk of adjacent vertebral body fracture have failed to show consistent results. Baroud et al (3) adapted a lumbar L4/5 segment as a three-dimensional (3-D) finite element (FE) model. They compared the pre- and post-augmentation stiffness and load shift with the same quasi-static axial compression of 2.8 mm. The result revealed a 17% greater compressive stress at the L4 trabecular bone in the post-augmented motion segment. In contrast, Rohlmann et al (7) suggested that vertebral body fractures in the adjacent vertebrae after vertebroplasty are not influenced by the elastic modulus of bone cement; instead, they suggested that the dominant factor in these fractures is the anterior shift of the upper vertebral body that is induced by fracture shape. In addition, studies to show the effects of filler materials with lower stiffness, such as low-modulus PMMA (8) and cancellous bone granules (9), on adjacent vertebrae after vertebroplasty have demonstrated conflicting results.

Those inconsistent results might be due to the different loading methods applied. The load-controlled method (LCM) is a traditional loading method that applies the same pure moment loading method to the spinal constructs. In contrast, the displacement-controlled method (DCM) applies different moment loads, so that the same overall ranges of motion (ROM) are achieved. Previous studies have demonstrated that adjacent vertebrae were more obviously affected under the DCM rather than the LCM in spine surgical models (10,11). Hence, we hypothesize that the risk of adjacent vertebral body fracture increases after vertebroplasty under the DCM. We also discussed and explained why vertebroplasty augmented with a low-modulus PMMA instead of the regular PMMA reduces the fracture risk of the adjacent vertebral body by using 3-D FE analyses.

Methods

Modeling Procedure and Analysis Conditions

Three-dimensional image data of the L1/2 functional spinal unit (FSU) were gained from computed tomographic scans of the lumbar spine of a 63-year-old woman who had no abnormal findings on radiographs taken at 0.2 mm intervals. The specialist software Mimics (Mimics 16.0, Materialise, Leuven, Belgium) was used to process the digital medical images and derive the geometry. A 3-D digital anatomic model of L1/2 bone structure was established, containing cortical bone, cancellous bone, bony endplate, and posterior elements. Based on the L1/2 bony surface, the other structure volumes, such as annulus fibrosus and nucleus pulposus, were secondarily produced in Mimics 16.0 (Fig. 1a). The reconstructed 3-D geometry was imported to the computer-aided design (CAD)/FE software (ANSYS 12.1, ANSYS Inc., Canonsburg, PA), and the preprocessing for FE analysis, such as meshing element and applying material properties and boundary conditions, was performed. Fig. 1b shows a meshed model. The material properties for FE analysis are shown in Table 1. The material properties for the osteoporotic bony structures were reduced, compared with normal bony structures, by 66% of the elastic modulus for cancellous bone and by 33% for cortical bone, bony endplate, and posterior elements (9). Because osteoporosis is clinically related to an age-related degeneration process, the process for intervertebral disc modeling was taken from Kurutz and Oroszvary's study (12).

The FE static analysis was performed under 5 circumstances: normal, osteoporotic, and augmented with bone cement at 1,800, 500, and 150 MPa in an osteoporosis model. The bone cement with an elastic modulus of 1,800 MPa represented regular PMMA, and 500 MPa represented low-modulus PMMA. Bone cements 3, 5, or 10 mL were injected into the center of L2 trabecular bone and augmented the supporting endplate similar to an upright pillar. FE static analyses were also performed with ANSYS 12.1.

Deformation Mode and Boundary Condition

Two boundary conditions, LCM and DCM, were applied to each of 5 deformation modes: compression, flexion, extension, lateral bending, and torsion. When load or force moment was fixed, it was said to be LCM.

Fig. 1. *Reconstructed geometry (a) and meshed model (b) of the L1/2 functional spinal unit.*

Table 1. Material properties for finite element analysis.

Materials	Young's modulus (MPa)	Poisson's ratio	Status	References
Bony structures				
Cortical bone	12000	0.3	Normal	3, 12
	8040	0.3	Osteoporosis	9
Trabecular bone	150	0.3	Normal	12
	34	0.3	Osteoporosis	9
Bony endplate	1000	0.3	Normal	9
	670	0.3	Osteoporosis	9
Posterior structure	3500	0.3	Normal	12
	2345	0.25	Osteoporosis	9
Intervertebral disc				
Nucleus pulposus	1	0.499	Normal	9
	9	0.4	Osteoporosis	12
Annlus matrix	4.2	0.45	Normal	9
	5	0.45	Osteoporosis	12
Annulus fibers	455	0.3	Normal	9
	455	0.3	Osteoporosis	9
Filler materials				
Bone cement 1	1800	0.3		8
Bone cement 2	500	0.3		8
Bone cement 3	150	0.3		

When displacement or rotation angle was fixed, it was said to be DCM.

The distortion energy theory was used for FE performance. The distortion energy theory, which is also known as the von Mises theory, is a famous theory of failure criterion for ductile materials. The von Mises stress is a failure criterion stating that yielding occurs when the von Mises stress reaches the yielding stress (13). The L2 inferior endplate's bottom surface centroid was fixed, and the load and moments were added at the centroid of the L1 superior endplate's top surface. An axial load of 500 N and 0.5 mm displacement was applied to the z direction in the compression mode under LCM and DCM boundary conditions, respectively (Fig. 2a). Bending moments of 5,000 Nmm in the flexion mode and 2,500 Nmm in the extension, lateral bending, and torsion modes were applied under LCM, whereas a rotation angle of 2° in the flexion mode and 1° in the extension, lateral bending, and torsion modes were applied under DCM (Fig. 2b). The L2 inferior endplate and the L1 superior endplate were connected centrally with rigid elements and multi-point constraints. Only flexion mode was applied to FE models with 3 and 5 mL of bone cement to compare the effects of cement volume. The contact of the facet joints was ignored.

Results

The maximum von Mises data with 5 deformation modes under LCM and DCM boundary conditions showed similar results (Fig. 3–9). The maximum von Mises stresses in the cortical bones and endplates of the osteoporotic FSU were larger than those of the normal FSU. The cement augmentation decreased the maximum von Mises stresses in the augmented L2 vertebra under both LCM and DCM. The L1 vertebra of the FSU, irrespective of augmentation, revealed nearly similar maximum von Mises stresses under the LCM (Fig. 3–9a). However, for the DCM boundary condition, as shown in Fig. 3–8b, the maximum von Mises stresses in the cortical bone and inferior endplate of the adjacent L1 vertebra increased significantly except in the torsion mode (Fig. 9b).

Compression

Under the displacement-controlled boundary condition, as shown in Fig. 3b, the maximum von Mises stresses in the cortical bone of the adjacent L1 vertebra increased by 32% for bone cement of 1,800 MPa, 27% for 500 MPa, and 18% for 150 MPa. The maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra increased by 34% for bone cement of 1,800 MPa, 28% for 500 MPa, and 19% for 150 MPa.

Flexion

As shown in Fig. 4b, with the rotation angle-controlled boundary condition, the maximum von Mises stresses in the cortical bone of the adjacent L1 vertebra increased by 19% for bone cement of 1,800 MPa, 17% for 500 MPa, and 11% for 150 MPa. The maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra increased by 22% for bone cement of 1,800 MPa, 19% for 500 MPa, and 12% for 150 MPa. In flexion mode, different cement volumes, 3 and 5 mL, were additionally used. The tendency seems similar, however, the less cement volume used, the smaller the increase in the maximum von Mises stresses. In the case of cement volume of 5 mL, under the rotation angle-controlled boundary condition (Fig. 5b), the maximum von Mises

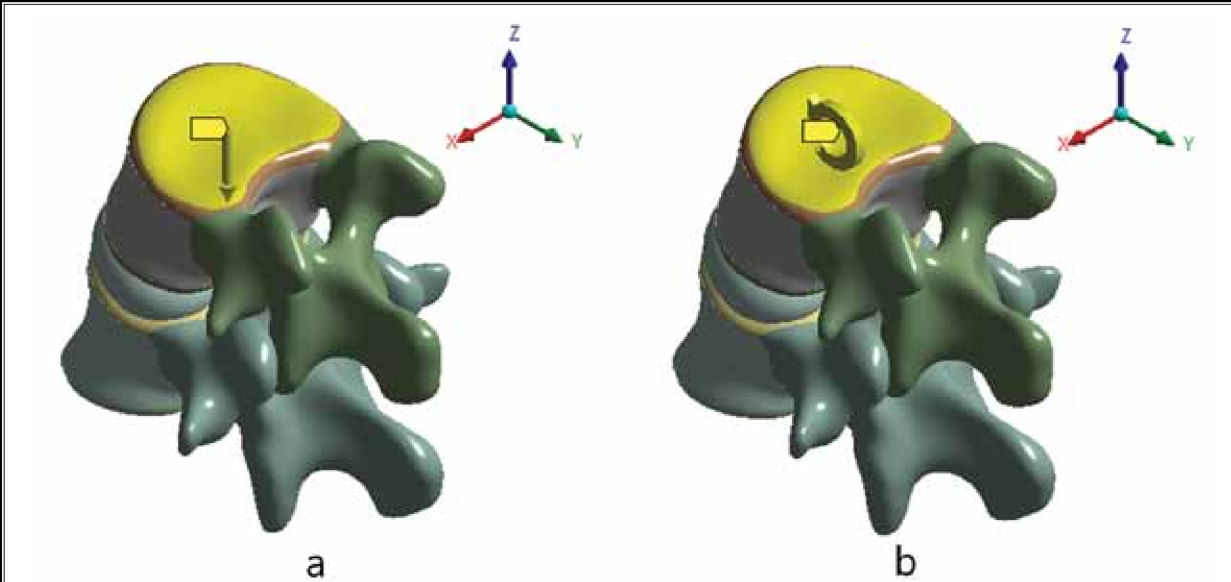

Fig. 2. *Load or displacement-controlled compression (a) and force moment or angle-controlled flexion (b).*

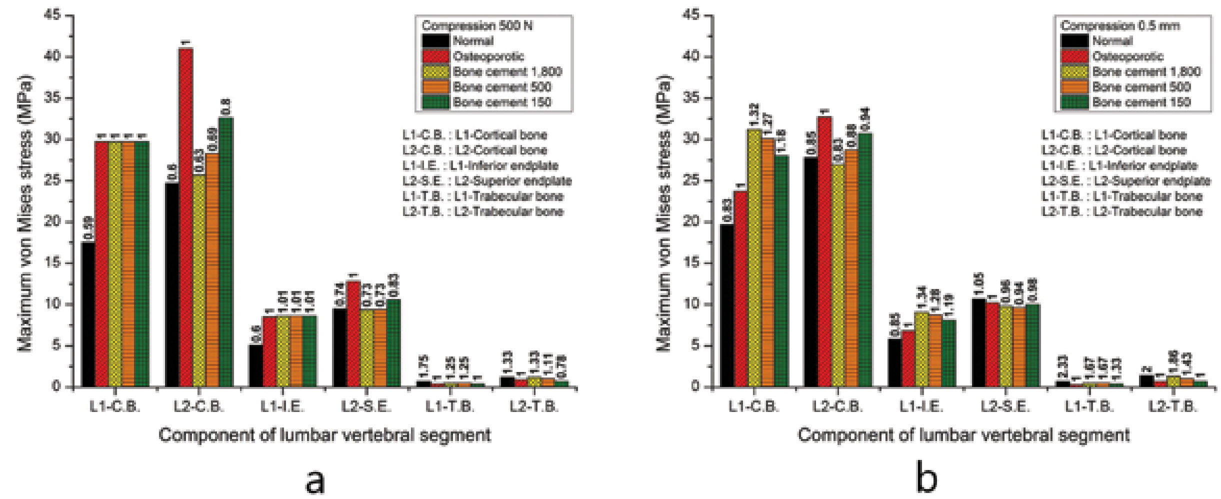

Fig. 3. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 10 mL under load-controlled compression, 500 N (a), and displacement-controlled compression, 0.5 mm (b).*

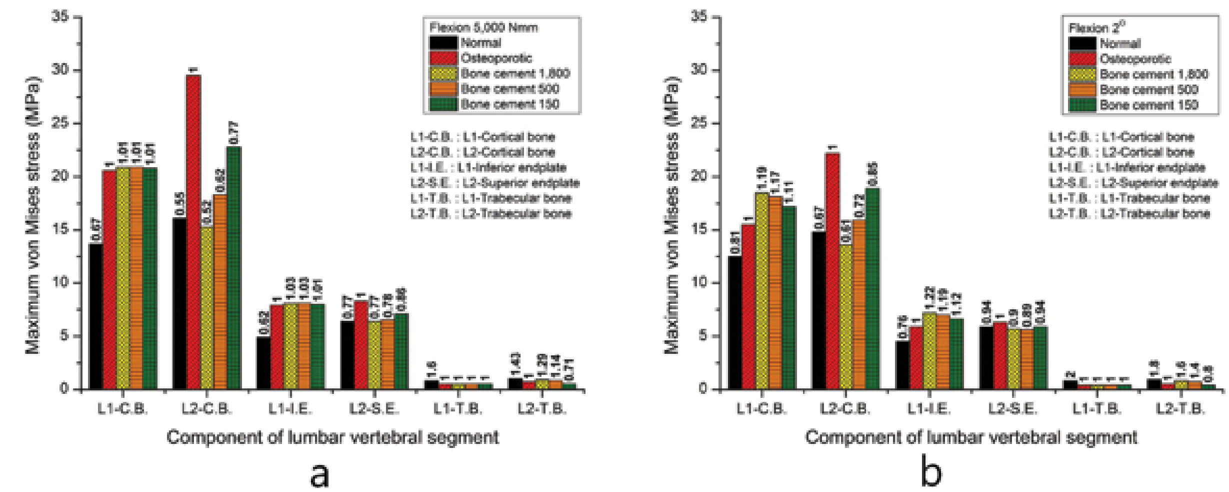

Fig. 4. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 10 mL under bending moment-controlled flexion, 5,000 Nmm (a), and rotation angle-controlled flexion, 2° (b).*

stresses in the cortical bone of the adjacent L1 vertebra increase by 12% for bone cement of 1,800 MPa, 10% for 500 MPa, and 6% for 150 MPa. The maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra increased by 14% for bone cement of 1,800 MPa, 12% for 500 MPa, and 7% for 150 MPa. In the case of cement volume of 3 mL, under the rotation angle-controlled boundary condition (Fig. 6b), the maximum von Mises stresses in the cortical bone of the adjacent L1 vertebra increase by 7% for bone cement of 1,800 MPa, 6% for 500 MPa, and 3% for 150 MPa. The maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra increased by 8% for bone cement of 1,800 MPa, 7% for 500 MPa, and 3% for 150 MPa.

Extension

Under the rotation angle-controlled boundary condition, the maximum von Mises stresses in the corti-

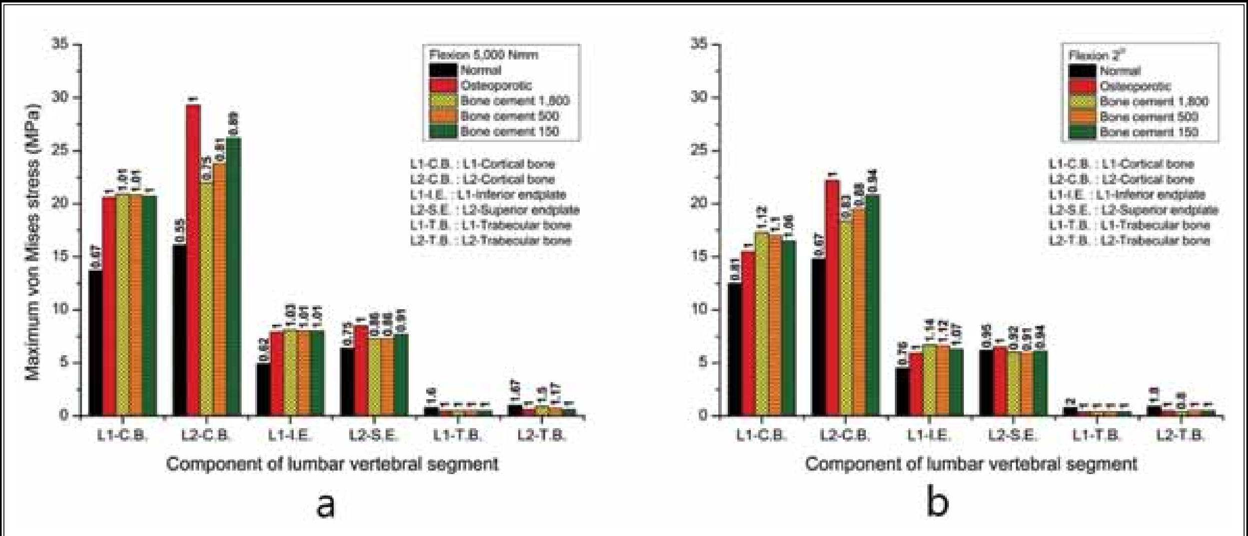

Fig. 5. Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 5 mL under bending moment-controlled flexion, 5,000 Nmm (a), and rotation angle-controlled flexion, 2° (b).

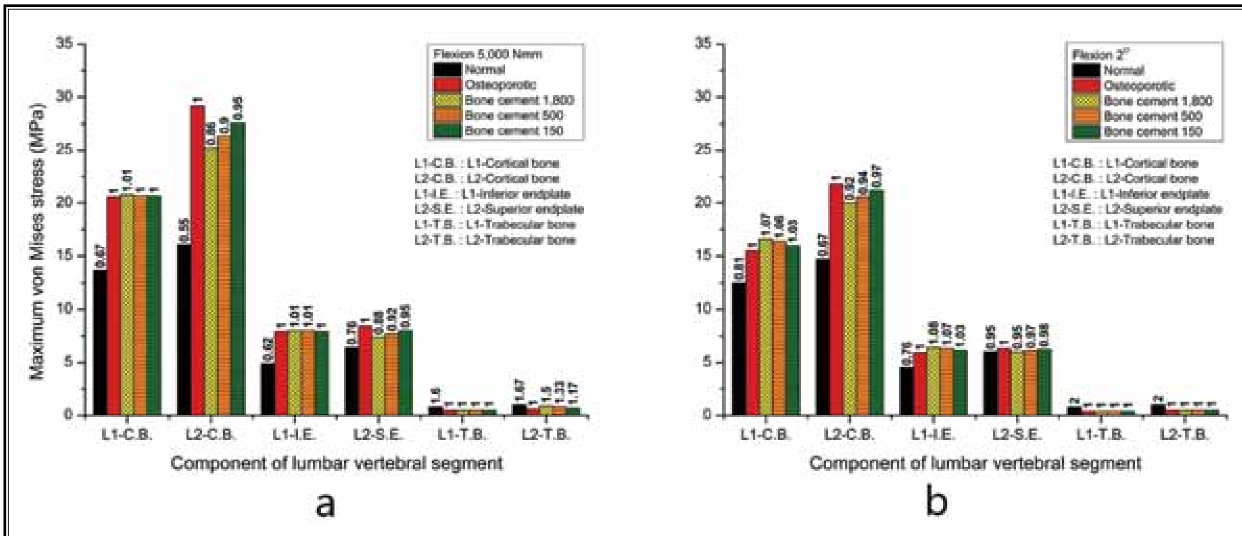

Fig. 6. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 3 mL under bending moment-controlled flexion, 5,000 Nmm (a), and rotation angle-controlled flexion, 2° (b).*

cal bone of the adjacent L1 vertebra were increased by 19% for bone cement of 1,800 MPa, 15% for 500 MPa, and 10% for 150 MPa and maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra were increased by 20% for bone cement of 1,800 MPa, 17% for 500 MPa, and 10% for 150 MPa (Fig. 7b).

Lateral Bending

As shown in Fig 8b, the maximum von Mises stresses in the cortical bone of the adjacent L1 vertebra increased by 17% for bone cement of 1,800 MPa, 14% for 500 MPa, and 9% for 150 MPa under the rotation angle-controlled boundary condition. Maximum von Mises stresses in the inferior endplate of the adjacent

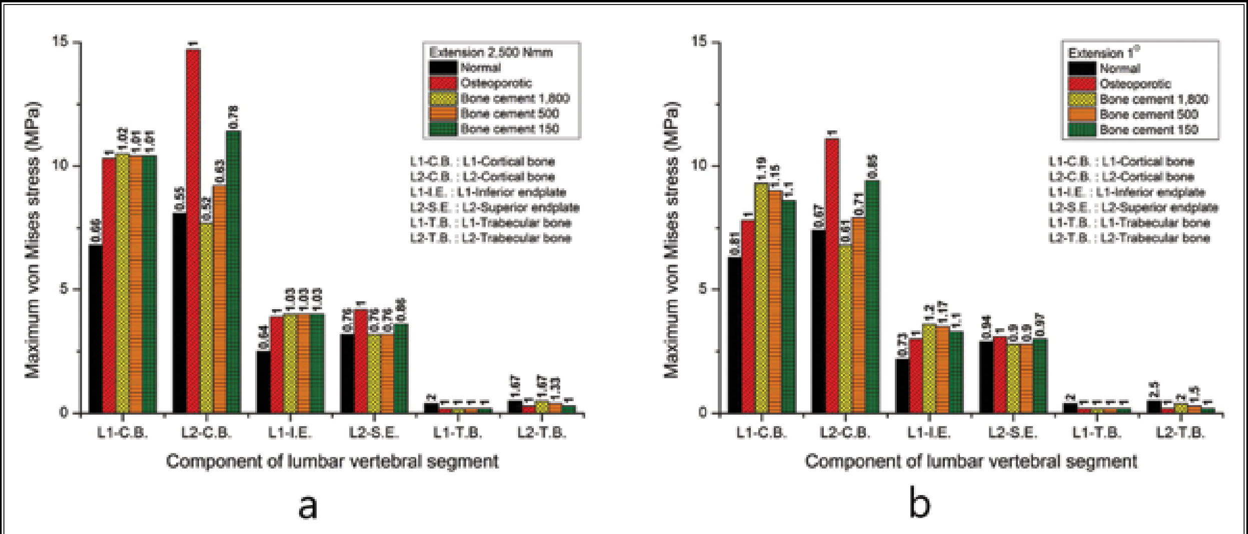

Fig. 7. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 10 mL under bending moment-controlled extension, 2,500 Nmm (a), and rotation angle-controlled extension, 1° (b).*

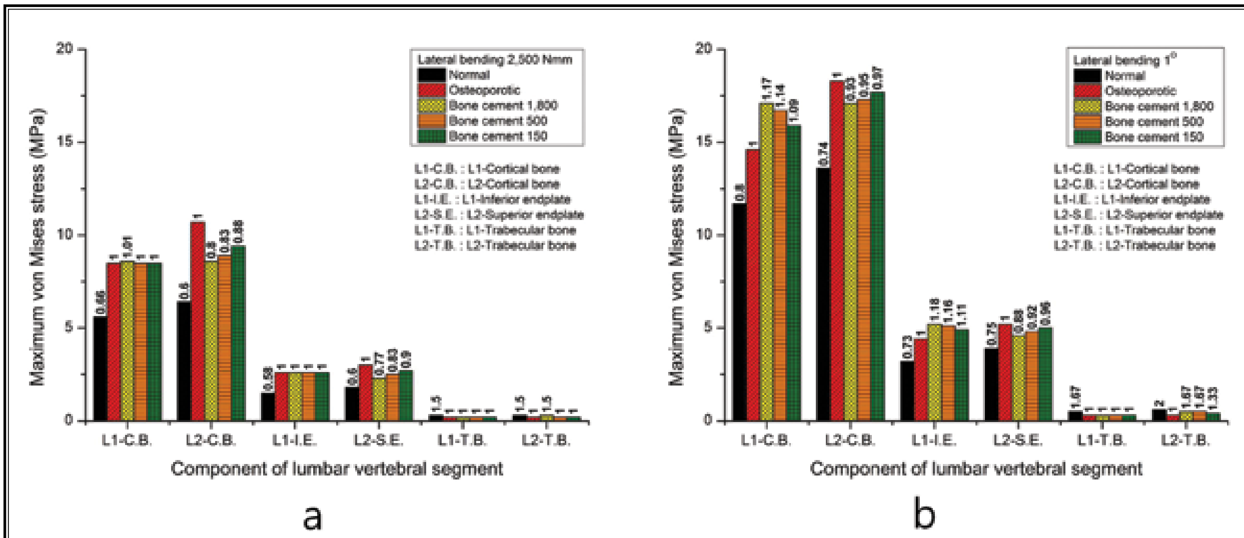

Fig. 8. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 10 mL under bending moment-controlled lateral bending, 2,500 Nmm (a), and rotation angle-controlled lateral bending, 1° (b).*

L1 vertebra were also increased by 18% for bone cement of 1,800 MPa, 16% for 500 MPa, and 11% for 150 MPa.

Torsion

Under the rotation angle-controlled boundary condition, the maximum von Mises stresses in the cortical bone of the adjacent L1 vertebra increased by 4% for bone cement of 1,800 MPa, 2% for 500 MPa, and 1% for 150 MPa. Maximum von Mises stresses in the inferior endplate of the adjacent L1 vertebra increased by 4% for bone cement of 1,800 MPa, 0% for 500 MPa, and 150 MPa. The results of stress for both twisting moment and rotation angle-controlled boundary conditions have small differences, as shown in Fig. 9a and 9b.

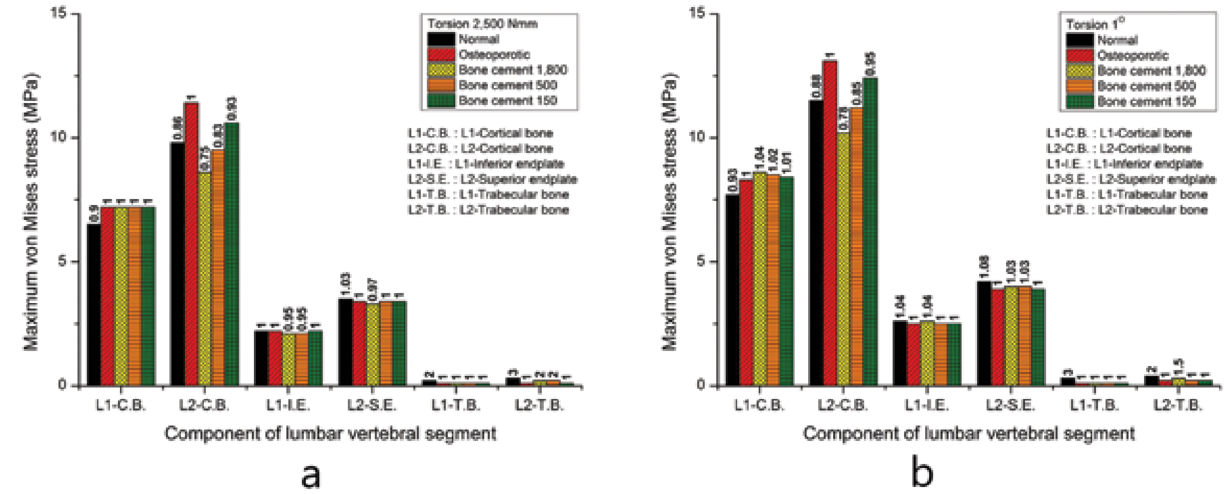

Fig. 9. *Maximum von Mises stresses for components of the L1/2 functional spinal unit with cement volume of 10 mL under twisting moment-controlled torsion, 2,500 Nmm (a), and rotation angle-controlled torsion, 1° (b).*

Discussion

According to this study, excessive stiffness of augmented bone cement increases the risk of adjacent vertebral fracture after vertebroplasty in osteoporotic FE models. The increased risk was more prominent under DCM rather than LCM, which explains the inconsistent results of previous studies.

Biomechanical studies using cadaveric FSU have been conducted previously. Barlemann et al (14) demonstrated that the failure strength of FSUs treated by cement augmentation in one vertebral body was lower than that of untreated controls. In contrast, another study failed to show any difference of the compressive load of failure between augmented and non-augmented vertebrae (15). The former study was conducted with load compression at a constant rate of 0.5 mm/s to a total compression of 10 mm. In the latter study, loading was created with a combination of axial compression and anterior flexion moments at a rate of 200 N/s.

The inconsistency also has appeared with FE analysis. Baroud et al (3) adapted a lumbar L4/5 segment as a 3-D FE model. They compared the pre- and post-augmentation stiffness and load shift with the same quasi-static axial compression of 2.8 mm. The result revealed 17% greater compressive stress at the adjacent trabecular bone in the post-augmented motion segment. In contrast, Rohlmann et al (7) simulated standing by applying a follower load of 500 N; flexion by applying a follower load of 1,175 N and a flexion bending moment of 7.5 Nm; extension, lateral bending, and axial rotation by applying a follower load of 500 N and a corresponding moment of 7.5 Nm; and walking by applying a follower load of 650 N and a torsion moment of 7.5 Nm. They concluded that bone cement's elastic modulus influence on the maximum von Mises stresses in cancellous and cortical bone was negligible.

Goel et al (10) analyzed adjacent vertebral effects of artificial discs that used both the LCM and DCM approach, and demonstrated that adjacent vertebral effects were more significant under DCM. This finding is consistent with our result. They suggested that DCM is more clinically relevant because, in real life, people bend their spines within a similar limited ROM, regardless of whether their spine is healthy or has undergone spinal surgery. When people pick up objects from the floor using squatting or bending movements after vertebroplasty, their ROM cannot be changed much; instead, flexion bending moments can be changed owing to augmented bone cements. Hence, we suggest that DCM is a much more clinically relevant approach to evaluate adjacent vertebral effects after vertebroplasty; however, it can be confirmed only after a comparative study of ROM between healthy and post-vertebroplasty lumbar spines is conducted.

Normal functional ROM of the lumbar spine during various activities of daily living was investigated previously (16). The functional ROM required to perform daily activities included in the investigation was 3° to 49° (median, 9°) of flexion/extension, 2° to 11° (median, 6°) of lateral bending, and 2° to 7° (median, 5°) of rotation. Of all the daily activities that were tested, more than 40° of lumbar flexion was required when participants picked up an object from the ground using squatting at the knee or bending at the waist. Although the study included asymptomatic individuals, the functional ROM of patients after vertebroplasty might be larger than the 1° or 2° used in the present study. It means that the maximum von Mises stresses on adjacent vertebrae would be even higher during daily activities, which increases the risk of vertebral fracture after vertebroplasty.

Efforts to overcome the shortcomings of PMMA, including excessive stiffness, have been made. Calcium phosphate cement (CPC) has been suggested as a possible alternative bone cement; it is highly osteoconductive and gradually replaced by new bone, which can provide substantial improvement in the compressive strength of the osteoporotic or fractured bone (17-19). The elastic modulus of PMMA is 2,700 MPa, which is much higher than that of human cancellous bone (168 MPa). In contrast, the elastic modulus of CPC is 180 MPa, which is more effective at avoiding the stress-shielding effect and abnormal load transfer, as well as reducing secondary fracture of adjacent vertebral bodies (20). Low-modulus PMMA also has been investigated, which resulted in lower incidence of upper vertebral endplate fractures compared with regular PMMA. However, despite biomechanical studies showing reduced risk of adjacent vertebral fracture with filler materials with lower stiffness (8,21), only a few clinical investigations have been conducted owing to low-modulus PMMA's inadequate initial stiffness, and they failed to demonstrate a better outcome compared with PMMA (22).

This clinical insignificance may be explained by relatively recent clinical studies that identified the risk factors for new osteoporotic vertebral compression fractures after vertebroplasty (23,24). The most important risk factor is bone marrow density (BMD). An adjacent vertebral fracture would not be affected by stiffness of bone cement as long as BMD is normal. The results of our study also showed vertebroplasty with various stiffnesses of bone cement could not reduce adjacent vertebral body fracture risk to the normal level. A previous study, in which highly osteoporotic bone could not restore the strength of normal vertebra, demonstrated that improvement in stiffness and strength after vertebroplasty depended significantly on BMD (25). Besides BMD, body mass index and cement leakage are also known to increase risk of adjacent vertebral fracture after vertebroplasty, which might have contributed to the negative outcomes in the previous clinical trials.

Some limitations arose from the simplified FE model in this study. One limitation is that trunk and paravertebral muscles were not included in FE analysis. Goel et al (26) compared the effects of muscular forces on a lumbar motion segment under normal muscle condition and totally devoid of muscle condition. The lumbar motion segment was more stable under normal muscular forces. Kong et al (27) also showed that, at higher loads or at more-flexed postures, muscles were found to play a more crucial role in stabilizing the spine. The facet joints help to maintain the stability of the spine and carry between 3% and 25% of the spinal load in axial compression (28). Moreover, the stresses in the facet joints increase significantly when comparing an osteoporotic spine to a healthy spine (29). One of the reasons is that degenerative facets lead to the mechanical instability of the spine motion segment increasing the mechanical stresses supported by the adjacent disc, causing its progressive degeneration (30). To maximize the effect of bone cement, we created 2-level spinal models and injected a higher volume of bone cement than is clinically recommended (31). Although less cement volume produces less von Mises stress on adjacent vertebra, the result still shows a similar trend as larger cement volume; stiffer cement produces greater von Mises stresses. Hence, future study should include the effects of muscular force and facet joints in 5-level lumbar spine models and examine effects with various bone cement volumes and locations.

Conclusion

In this research, the effects of bone-cement stiffness on adjacent vertebrae after osteoporotic vertebroplasty with LCM versus DCM were compared using FE analysis. The results demonstrated that bone-cement stiffness significantly affects adjacent vertebral body fractures only with DCM. Assuming the ROMs between pre-vertebroplasty and post-vertebroplasty are similar (10), DCM is more suitable than LCM to reflect real life. Our investigation suggests that excessive stiffness of bone cement might be a contributing factor to adjacent vertebral fracture after vertebroplasty in osteoporosis. However, the limitation that simplified our FE model and could not fully reflect the complexities of real-life situations should be considered.

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