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Prophylactic Vertebroplasty May Reduce the Risk of Adjacent Intact Vertebra From Fatigue Injury

An *Ex Vivo* Biomechanical Study

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Study Design. *In vitro* biomechanical study using human spine specimens.

Objective. To find the biomechanical consequences of prophylactic vertebroplasty post fatigue loading.

Summary of Background Data. Percutaneous vertebroplasty man be an effective treatment for osteoporotic vertebral compression fracture. One frequently observed complication post surgery is the adjacent vertebral failure (AVF). The prophylactic vertebroplasty was proposed to prevent the AVF. The vertebroplasty is, nevertheless, an invasive intervention. More scientific proves are needed for the application of this surgery on a still intact vertebra.

Methods. Fourteen 5-level fresh human cadaveric thoracic motion segments were divided into standard and prophylactic group. Both ends of the specimen were mounted, leaving the center 3 vertebrae free. The lower level of free vertebrae was artificially injured and cement augmented. The center level vertebra of standard group remained intact and nonaugmented. The center level vertebra of prophylactic group also remained intact, but augmented with bone cement. The specimen was applied with a 2-hour, 5-Hz, 630-N (mean) compressive fatigue loading. Impulse test and CT scanning were conducted both before and after fatigue loading to find the variance of strain compliance of cortical shell and height of vertebral body.

Results. The strain compliance of cortical shell is generally not statistically significantly affected by the fatigue loading, cement augmentation and vertebral level (All $P > 0.05$). The only exception is that the cortical strain compliance of augmented vertebrae tentatively decreased post fatigue loading ($P = 0.012$ for tensile strain compliance, and $P = 0.049$ for compressive strain compliance). The height loss of intact vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra is significantly lower than the one adjacent to a 1-level augmented (or injury-augmented) vertebra ($P = 0.014$). For an osteoporotic vertebra, neither cortical strain compliance nor vertebral height loss is connected with bone mineral density (all $P > 0.05$).

Conclusion. The strain compliance of cortical shell is generally not a sensitive indicator to predict risk of fatigue injury if the fatigue loading is mild. The prophylactic augmentation strengthens the osteoporotic vertebrae, decreases the progression of vertebral height loss, reduces the anterior body shift, and hence protects the adjacent intact vertebra from elevated flexion bending. It can be cautiously suggested that if the vertebra is osteoporotic and adjacent level is located at pivot or lordotic level of spinal column, the prophylactic augmentation may be an option to prevent the AVF.

Key words: intact-augmented vertebrae, bone cement, strain compliance, vertebral height, bone mineral density.
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Adjacent vertebral failure (AVF) is a frequently observed post surgery complication of percutaneous vertebroplasty.[1,2] Clinical studies showed 12% to 24% of patients suffered subsequent fractures post vertebroplasty within 1 year,[2,3] and 41% to 67% of the subsequent fractures occurred in the level adjacent to the augmented vertebra.[3–5] The fracture rate of adjacent vertebra is 3 times higher than the one of nonadjacent one.[6] The causes of AVF post vertebroplasty were suspected to be the increased stiffness of vertebral body,[7] load transferring in adjacent vertebra,[8,9] shifted body weight[10] and the ongoing progression of osteoporosis.[4,11]

The low bone mineral density (BMD) was reported to be a critical risk factor for osteoporotic vertebral fracture.[11] The prophylactic augmentation of the adjacent vertebra was suggested to reduce the high incident rate of AVF. The rationale of prophylactic augmentation is to improve the bone quality, to reduce spinal deformity,[12] and thus to stabilize osteoporotic vertebra.[13,14] An *in vitro* biomechanical evaluation reported prophylactic vertebroplasty enhanced failure strength and maintain stiffness of nonfractured vertebrae.[15] A finite element analysis further suggested a successful prophylactic reinforcement at high-risk levels required the filling ratio to be 20% of vertebra volume.[2] It is obvious that the cement augmentation will increase the strength of either fractured or intact vertebrae, but how about the biomechanical performance of the new adjacent intact vertebra? Is the failure risk of new adjacent vertebra similar to the one of the prophylactic augmented vertebra before augmentation? Does the prophylactic augmentation just move the same failure risk 1 level above? Since the cement augmentation is still an invasive intervention, the

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practice of cement augmentation on a still intact vertebra needs more scientific proves.

The purpose of this study was to find the biomechanical consequences, *i.e.*, the strain compliance of cortical shell and the height of vertebral body, of prophylactic vertebroplasty on the osteoporotic vertebrae post fatigue loading. The following questions are to be answered. What is the effect of fatigue loading, cement augmentation, vertebral level and BMD on the biomechanical properties of vertebrae? Is the mechanical performance of the vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra similar to the one adjacent to a 1-level augmented (or injured-augmented) vertebra? Does the prophylactic cement augmentation also prevent the adjacent intact nonaugmented vertebra from fatigue injury? Can the BMD be an index for the prophylactic augmentation? The answer of these questions helps to determine the necessity or criteria of prophylactic vertebroplasty.

■ Materials and Methods

Specimen Preparation

Fourteen fresh 5-level osteoporotic thoracic motion segments (T1–T5, N = 8; T6–T10, N = 3; T3–T7, N = 1; T8–T12, N = 2) from 10 human spines were used. All spines were first examined with computer tomography (CT) (LightSpeed VCT, GE) and dual energy radiograph absorptiometry (DEXA) (QDR 4500A, Hologic). The CT examination of the specimens showed normal degenerative changes and no deformity or anatomic defects. The DEXA examination found the BMD of vertebrae. All specimens were carefully cleaned off muscle tissue. The centered 3 vertebrae were labeled as VB1, VB2, and VB3 from top to bottom (Figure 1).

An artificial fracture was created in VB3 using a 25-mm-diameter disc saw. The cleft was horizontally penetrated into the vertebral body by 10 mm through the center of anterior cortical shell. The size of the cleft was approximately 10 mm in depth, 30 mm in width, and 2 mm in height. To prevent the cement leakage from the fissure site, the surface of the cortical bone was sealed with bone cement. Then, 3.5 mL polymethylmethacrylate bone cement (DePuy CMW 1, Johnson & Johnson, UK) was injected into the vertebra with 11-gauge bone biopsy needle through left pedicle. The monomer-to-powder ratio of the bone cement was 0.75 mL/g.[16] The curing time was 70 seconds at room temperature. In the prophylactic group, 3 mL bone cement was further injected into the intact center vertebra (VB2) (Figure 1). Axial compressive force or extension traction, which simulates the internal muscle/ligament force or clinical manipulation during surgery, was not applied during the bone cement injection to minimize the experimental factors and variations.

The specimen was then provided with quick setting gypsum mount at both ends with the middle vertebra horizontally aligned. Four metallic nails were screwed in the anterior, posterior, and lateral site of the mount as reference points for the alignment of CT graphic measurement. All specimens were placed in plastic bags, and frozen at −20°C until testing.

Impulse Testing Apparatus and Protocol

A "drop-tower type" testing apparatus was used for impulse testing (Figure 2).[17] The impulse energy was generated by the free fall of impactor and was transmitted to the specimen through the impounder. The shock absorber was placed on top of the impounder to control the impulse contact period. The contact period is the time interval that impactor compresses the impounder before it bounces back. The shock absorber was designed to give the contact period at 40 milliseconds.[18] The specimen was mounted vertically below a uni-axial load cell (STC-500 kg, Vishay Intertechnology, Inc., Malvern, PA). The measured force was the resultant force because of impulse energy. A high-speed data acquisition card (PCI-6071E, National Instrument, Inc., Austin, TX) was used to record the signals from force load cell and strain gauge rosettes (the application of rosettes will be described in experimental procedure). Half seconds data were recorded at 10 kHz sampling rate. The signals were low-pass filtered at 500 Hz frequency using Butterworth filtering algorithm. The impulse weight was 12 kg and the impulse height was 1 cm, which resulted in a 1.2-J impulse energy. The averaged peak loading during impulse was 580 N, which was smaller than the average of fatigue loading (630 N, see following paragraph). The damage of motion segment due to impulse loading was minimal.

Fatigue Loading Apparatus and Protocol

A vibrator composed of 2 eccentric rotors and 1 motor was mounted into the impactor of the impulse testing apparatus to create the fatigue loading. The fatigue loading was produced by the rotation of 2 counter-rotated eccentric rotors driven by a motor. The vibration was transmitted to the specimen through the impounder (Figure 2).[17] The root mean squared averaged fatigue loading was 630 N (peak to peak; 550 N–750 N). The magnitude of fatigue loading was kept constant through out

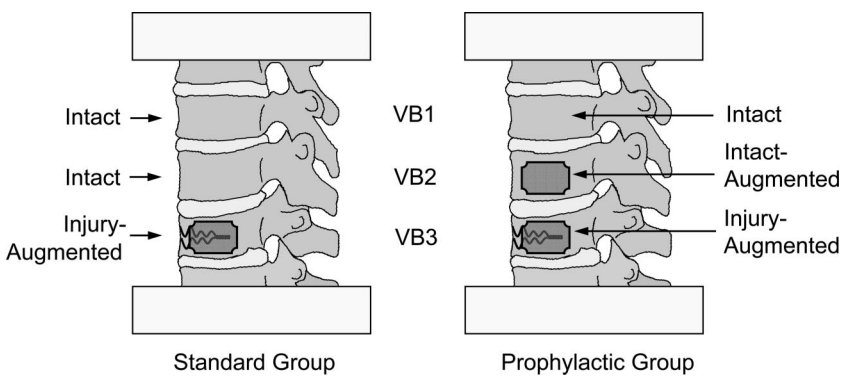

Figure 1. Schematics of specimen preparation. The lower level vertebra (VB3) was artificially injured and cement augmented. The center level vertebra (VB2) of standard group remained intact and nonaugmented. The VB2 of prophylactic group also remained intact, but augmented with bone cement.

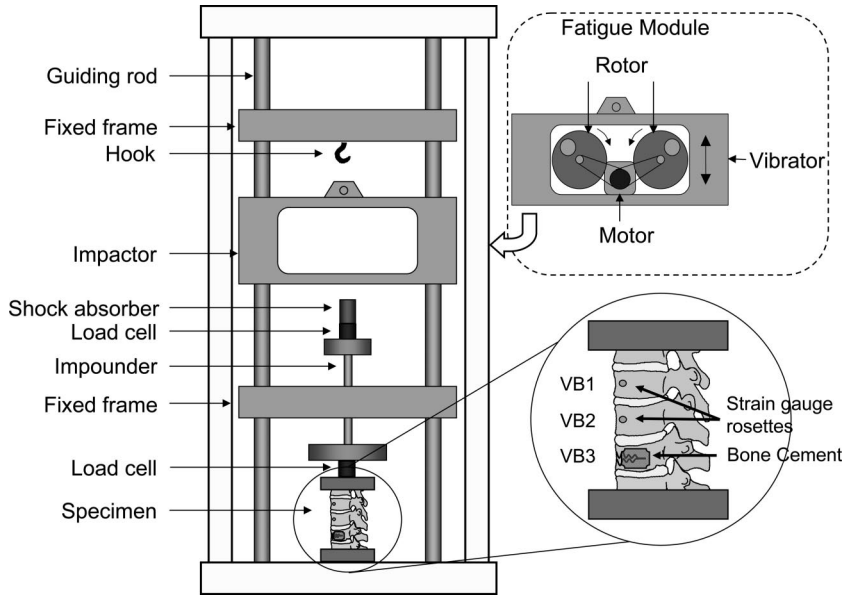

Figure 2. Schematic of fatigue and impulse loading apparatus and application of strain gauge rosettes. The impactor was guided by 2 rods to give a vertical motion. The fatigue module created the cyclic loading by a pair of counter rotated eccentric rotor. The specimen was mounted vertically below the impounder. The strain gauge rosettes were applied on the anterior cortical shells of upper and center level vertebrae.

the experiment. The loading frequency was 5 Hz and the loading time was 2 hours.

Experimental Procedures

The intact specimens were thawed at room temperature for at least 6 hours before testing. After the specimen was fully thawed, two 3-axis strain gauge rosettes (FR-1A12L30W05MS, Minebea Co., Ltd., Japan) were applied on the anterior cortical surface of upper (VB1) and center vertebrae (VB2). The specimens were CT scanned again to find the vertebral height as a base line for later comparison. To get the better spatial resolution in sagittal plane, the specimens were stand vertically on the CT bed with the anterior wall facing the gantry. The scanning thickness was 0.625 mm. The field of view was 96 mm, hence the spatial resolution was 0.1875 mm/pixel. An impluse testing was also applied to find the bone strains of cortical shell. The specimens were then applied with fatigue loading. Impulse testing and CT scanning were applied again after fatigue loading to find the variations of bone strain and vertebral height (Figure 3).

Data Analysis

The independent variables of this study include: fatigue loading, cement augmentation, level of vertebrae, and BMD. The dependent variables include the strain compliance of cortical shell and height of vertebral body.

The principal cortical strains of VB1 and VB2 were calculated from the 2 strain gauge rosettes. Both the first and third principal strains were calculated. The first principal strain is the maximum tensile strain, while the third principal strain is the maximum compressive strain. The strain compliance, *i.e.*, the ratio of maximal principal strain over the maximal force, during the loading history was calculated. The higher magnitude of strain compliance means lower strain stiffness.

The vertebral height was measured from the CT images. The metallic screws in the mount were used to make sure the compared frames before and after fatigue loading were in the same plane. Since the deformation of the vertebral body was generally less than 0.5 mm, multiple locations of the vertebral height at anterior, posterior, middle, left and right lateral site were measured, and then averaged to get the mean vertebral height.

The cortical strain compliance and vertebral height before and after fatigue loading were first compared using dependent paired Student *t* test. If the variation before and after the fatigue loading was not significant, the effect of cement augmentation and levels of vertebra both before and after fatigue loading on the variables will be examined using independent Student *t* test. The variable will also be linear regressed with BMD. If the variation between before and after fatigue loading was significant, the variation would be calculated. Then, the effect of cement augmentation and vertebral level on the variation will be examined using independent Student *t* test. The variation will also be linear regressed with BMD. All tests were considered to be significant if $P < 0.05$. The linearity of regression

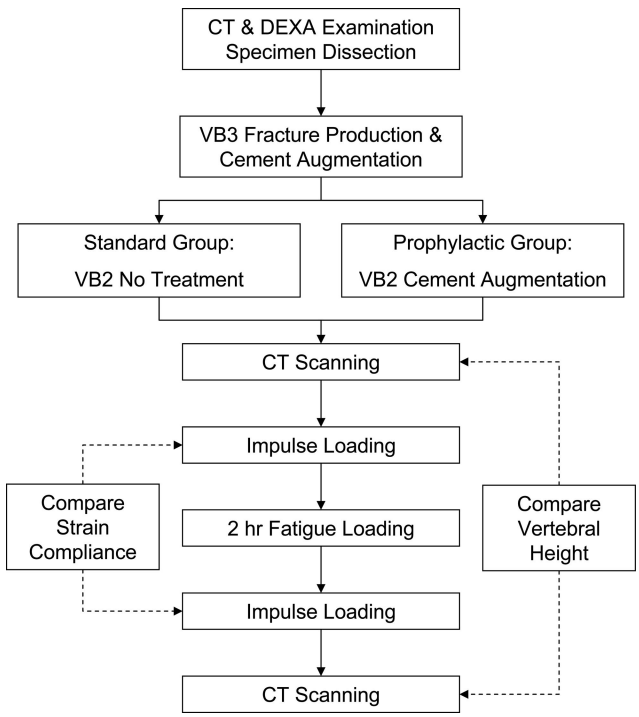

Figure 3. Experimental protocol.

Table 1. Age, Gender, Body Weight, Vertebral Level, Bone Mineral Density, and T-Score of the Specimens Used in the Standard Group and Prophylactic Group

Standard Group Specimen	Age	Gender	Weight (kg)	Level	BMD (g/cm^2)
1	68	F	45.4	T1–T5	0.453
2	68	F	45.4	T6–T10	0.627
3	83	M	59.0	T1–T5	0.669
4	83	M	59.0	T6–T10	0.637
5	87	F	84.0	T6–T10	0.534
6	62	F	36.3	T1–T5	0.388
7	86	M	54.5	T3–T7	0.528
				Average	0.55 (0.10)
Prophylactic Group Specimen	**Age**	**Gender**	**Weight (Kg)**	**Level**	**BMD (g/cm^2)**
1	71	M	81.7	T1–T5	0.777
2	87	F	84.0	T1–T5	0.419
3	86	M	54.5	T8–T12	0.539
4	67	F	56.8	T8–T12	0.651
5	97	F	54.5	T1–T5	0.341
6	52	M	120.3	T1–T5	0.686
7	83	M	45.4	T1–T5	0.649
				Average	0.58 (0.14)

was evaluated with *f* test. The quality of linear fitting was evaluated by the correlation coefficient (r^2).

The cement filling ratio, *i.e.*, the cement volume over the vertebral body volume, of augmented vertebra was also calculated from the CT stack images. The cement volume was the summation of gray area covered by cement, while the vertebral volume was the summation of area inside cortical shell. It should be noted, the calculated cement volume may be higher then the physically injected cement volume. It is because the contoured gray area of CT images covered both cement and cancellous bone, so the calculated cement volume is the summation of physically injected cement volume and the cancellous bone that the cement resides in. The cement filling ratio between the injury-augmented and intact augmented vertebra was compared using nonpaired independent *t* test.

■ Results

The averaged BMD of specimens was 0.56 (0.13) g/cm^2. The specimens were evenly distributed into 2 groups by the quality of BMD [standard group, BMD = 0.55 (0.10) g/cm^2, N = 7; prophylactic group, BMD = 0.58 (0.14) g/cm^2, N = 7]. The BMD between these 2 groups were not statistically different (P = 0.655) (Table 1). The filling ratio of intact-augmented vertebrae [30 (12) %] is higher than the one of injury-augmented vertebrae [23 (6) %], but not statistically significant (P = 0.206).

The mean tensile cortical strain compliance of intact vertebrae (VB1 and VB2 in standard group, and VB1 in prophylactic group) ranged from 0.79 $\mu\varepsilon$/N to 0.90 $\mu\varepsilon$/N before fatigue loading (Table 2). The tensile strain compliance decreased after fatigue loading, but not statistically significant (All $P > 0.05$). The tensile cortical strain compliance of intact-augmented (VB2 in prophylactic group) vertebra was 0.64 $\mu\varepsilon$/N, and significantly decreased to 0.42 $\mu\varepsilon$/N post fatigue loading (P = 0.012, Table 3). The mean compressive cortical strain compliance of intact vertebrae ranged from 1.06 to 1.50 $\mu\varepsilon$/N before fatigue loading (Table 2). The compressive strain compliance decreased after fatigue loading, but not statistically significantly (All $P > 0.05$). The compressive cortical strain compliance of intact-augmented vertebra

Table 2. Mean (SD) of Principal Tensile and Compressive Strain Compliance ($\mu\epsilon$/N) of Vertebral Cortical Shell Before and After Fatigue Loading

	VB1 (intact VB)		VB2 (intact VB)	
Standard Group	Before Fatigue	After Fatigue	Before Fatigue	After Fatigue
Tensile	0.80 (0.44)	0.71 (0.46)	0.79 (0.26)	0.70 (0.23)
Compressive	−1.06 (0.54)	−0.98 (0.50)	−1.50 (0.69)	−1.34 (0.60)
Ratio of tensile/comp	78 (17) %	69 (23) %	59 (24) %	58 (20) %
	VB1 (intact VB)		VB2 (intact-augmented VB)	
Prophylactic Group	Before Fatigue	After Fatigue	Before Fatigue	After Fatigue
Tensile	0.90 (0.59)	0.71 (0.47)	0.64 (0.45)	0.42 (0.32)
Compressive	−1.30 (0.77)	−1.21 (0.77)	−1.40 (0.67)	−1.06 (0.47)
Ratio of tensile/comp	80 (30) %	62 (23) %	44 (21) %	37 (21) %

Table 3. The Significances (*P-value*) of Effect of Fatigue Loading, Vertebral Level, Cement Augmentation, and Prophylactic Augmentation on Principal Tensile Strain Compliance of Vertebral Cortical Shell

	VB1	VB2
Effect of Fatigue (Before *vs.* After Fatigue)		
Standard group	$P = 0.502$	$P = 0.098$
Prophylactic group	$P = 0.099$	$P = 0.012$*
	Before fatigue	After fatigue
Effect of level (VB1 *vs.* VB2)		
Standard group	$P = 0.933$	$P = 0.967$
Prophylactic group	$P = 0.377$	$P = 0.207$
	Before fatigue	After fatigue
Effect of cement augmentation		
Prophylactic VB2 *vs.* standard VB2	$P = 0.472$	$P = 0.094$
	Before fatigue	After fatigue
Effect of prophylactic augmentation on adjacent level		
Prophylactic VB1 *vs.* standard VB2	$P = 0.659$	$P = 0.949$

*Significance: $P < 0.05$.

was 1.40 $\mu\varepsilon$/N, and tentatively decreased to 1.06 $\mu\varepsilon$/N post fatigue loading ($P = 0.049$, Table 4). The ratio of tensile over compressive strain of intact vertebrae ranged from 60% to 80%, whereas the ratio of intact-augmented vertebra was at level of 40% (Table 2).

The effect of vertebra level (VB1 *vs.* VB2 in both standard and prophylactic group) and cement augmentation (VB2 in prophylactic group *vs.* VB2 in standard group) on strain compliance was both not significant (Table 3, 4). The strain compliance of intact vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra was not statistically significantly different from the one adjacent to a 1-level augmented (or injury-augmented) vertebra (VB1 in prophylactic group *vs.* VB2 in standard group) (Table 3, 4). The strain compliance of intact vertebra was not affected by the magnitude of BMD both before ($r^2 = 0.026$, $P = 0.484$) and after fatigue loading ($r^2 = 0.095$, $P = 0.174$).

The vertebral height significantly decreased after 2-hour fatigue loading (All $P < 0.05$) (Table 5). The height loss of augmented vertebrae ranged from 0.11 to 0.16 mm, which corresponding to 0.57% to 0.84% of original vertebral height. The height loss of intact vertebrae ranged from 0.25 to 0.45 mm, which corresponded to 1.36% to 2.35% of original vertebral height. The height loss of augmented vertebrae were all significantly lower than the one of nonaugmented vertebrae (VB2, VB3 in prophylactic group and VB3 in standard group *vs.* VB1 in prophylactic group and VB1, VB2 in standard group, all $P < 0.05$).

Table 4. The Significances (*P*) of Effect of Fatigue Loading, Vertebral Level, Cement Augmentation, and Prophylactic Augmentation on Principal Compressive Strain Compliance of Vertebral Cortical Shell

	VB1	VB2
Effect of Fatigue (Before *vs.* After Fatigue)		
Standard group	$P = 0.496$	$P = 0.188$
Prophylactic group	$P = 0.115$	$P = 0.049$*
	Before fatigue	After fatigue
Effect of level (VB1 *vs.* VB2)		
Standard group	$P = 0.194$	$P = 0.251$
Prophylactic group	$P = 0.797$	$P = 0.667$
	Before fatigue	After fatigue
Effect of cement augmentation		
Prophylactic VB2 *vs.* standard VB2	$P = 0.759$	$P = 0.349$
	Before fatigue	After fatigue
Effect of prophylactic augmentation on adjacent level		
Prophylactic VB1 *vs.* standard VB2	$P = 0.592$	$P = 0.732$

*Significance: $P < 0.05$.

The height loss of intact vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra was significantly lower than the one adjacent to a 1-level (or injury-augmented) vertebra (VB1 in prophylactic group *vs.* VB2 in standard group, $P = 0.014$). In the prophylactic group, the height loss of center vertebra (VB2) was higher than the one of upper level vertebra (VB1), but not statistically significant. The height loss of upper level vertebra (VB1) in standard group was similar to the one in prophylactic group ($P = 0.570$) (Figure 4). The loss of vertebral height was also not affected by BMD ($r^2 = 0.076$, $P = 0.226$).

■ Discussion

Specimen Preparation and Experimental Procedure

The artificial vertebral fractures enable researchers to control the fracture type and location in a reproducible manner. Different sizes of drill bit (6.35–25.4 mm),[19] trephine (5 mm),[20] and inflated balloon and drill bit (9 mm)[21,22] plus mechanical loading were used to create the reproducible compression fracture. The anterior site of wedge fracture vertebra is injured and not continuous before union. We use the 25-mm blade saw to artificially simulate this injury and discontinuity. The failure mode may be attributed to the type A1.2 fracture, *i.e.*, the wedge impaction fracture.[23] We did not apply extra loading to crush the anterior cortical to create the wedge-like vertebra, nor did we simulate the traction/extension manipulation during the vertebroplasty operation to minimize the experimental variation. It is expected that the current procedure of injury creation and cement augmentation can be the best scenario of vertebroplasty operation, after which the injured vertebra restored to its original height.

During the injection of an injured vertebra, the cement filled the cleft first, then perfused into the cancellous pore. During the injection of an intact vertebra, the pores within the cancellous bone were perfused with bone cement. This is the reason why the filling ratio of injury augmented vertebra is lower than the one of intact augmented vertebra. The filling ratio of bone cement may affect the vertebra strength. An FEM analysis showed higher filling ratio of bone cement reached higher degree of vertebral stiffness and strength.[2] Another *in vitro* study also showed endplate-to-endplate full augmentation is better than the partial augmentation in restoring the vertebral stiffness and strength.[24] In this study, since the level of filling ratio of intact augmented vertebrae (30%) is not statistically significantly different from the one of injury augmented vertebrae (23%), the difference of

Table 5. Mean (SD) of Vertebral Height (mm) Before and After Fatigue Loading

	Standard Group			Prophylactic Group		
	Before Fatigue	After Fatigue	*P*	Before Fatigue	After Fatigue	*P*
VB1	18.1 (1.4)	17.8 (1.4)	0.001	18.2 (2.6)	17.9 (2.5)	0.001
VB2	19.1 (1.2)	18.7 (1.2)	0.000	18.7 (2.6)	18.6 (2.5)	0.000
VB3	19.9 (1.6)	19.8 (1.5)	0.001	19.5 (2.8)	19.4 (2.8)	0.000

The comparison of vertebral height before and after fatigue loading is using the dependent paired Student *t* test.

vertebral stiffness and strength between injured-augmented and intact-augmented vertebra should be minimal.

The magnitude of loading and number of loading cycle determine the degree of fatigue injury. The loading magnitude of current study was 630 N (mean), and the number of cycles was 36,000 cycles. The compressive loading on vertebrae column was 0.2 to 2.5 body weight during walking.[25] In this study, the average body weight of the specimen was 64 kg. The 630 N loading magnitude of the present study was within the range of moderate physiologic load. The spine of normal people subjected to 1 million loading cycles per year,[26] which is equivalent to 85,000 cycles per month. The 36,000 loading cycle could reasonably simulate the loading cycles of few weeks.

Strain Compliance of Cortical Shell

The strain gauge rosette is a sensitive sensor to measure the micro deformation of bony structure. In spine biomechanics research, the strain gauge was used to predict the initiation of burst fracture injury[27] and risk potential of cement augmentation.[21,22,28] The cortical strain at 50 kg external loading, *i.e.*, stain compliance multiplied by 490 N,[27] was compared. The average compressive strain of current study ranged from 480 to 740 $\mu\varepsilon$, which was a little bit higher than the range of previous studies: 200,[29] 300,[27] and 450 $\mu\varepsilon$.[21] The higher measured strain of this study may be due to the poor bone quality of specimen and measurement site. The measurement site of this study was located at the center of anterior cortical shell, which is the site that the largest strain occurred within the vertebra.

The strain compliance of cortical shell tentatively decreased post fatigue loading. This phenomenon is so called the "strain hardening effect" in the study of engineering material science. The "strain hardening effect" was found to be more provoked in the augmented vertebra. This maybe because the strengthening of cement augmented cancellous bone reduced the progression of micro deformation during fatigue loading. This guess may also be indirectly proved by the less magnitude of tensile over compressive strain ratio in the augmented vertebra. The physical appearance for the smaller tensile strain of cortical shell is the cortical shell was "less easy to bulge." The reason of "less easy to bulge" of cortical shell may be due to the infiltrated cement constrained the outward migration of cancellous bone during fatigue loading.

Height Loss of Vertebral Body

The loss of vertebral height is an index for the degree of vertebral compression fracture. As anticipated, the height loss of augmented vertebrae was significantly lower than the one of nonaugmented vertebra. An *in vitro* model showed that the height loss of augmented vertebrae was 10% for vertebroplasty and 22% for kyphoplasty after 10,000 cycles of 45% (mean) of vertebral failure load.[1] An *in vivo* study showed the loss of vertebral height was 4.4% 1 year post kyphoplasty.[30] The loss

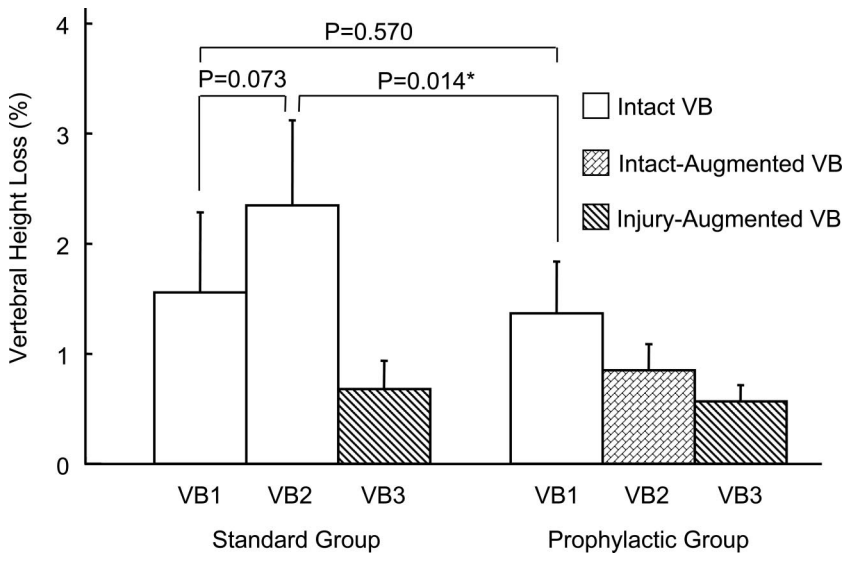

Figure 4. Loss of vertebral height (%) of standard group and prophylactic group during fatigue loading. Error bars represent ±1 SD.

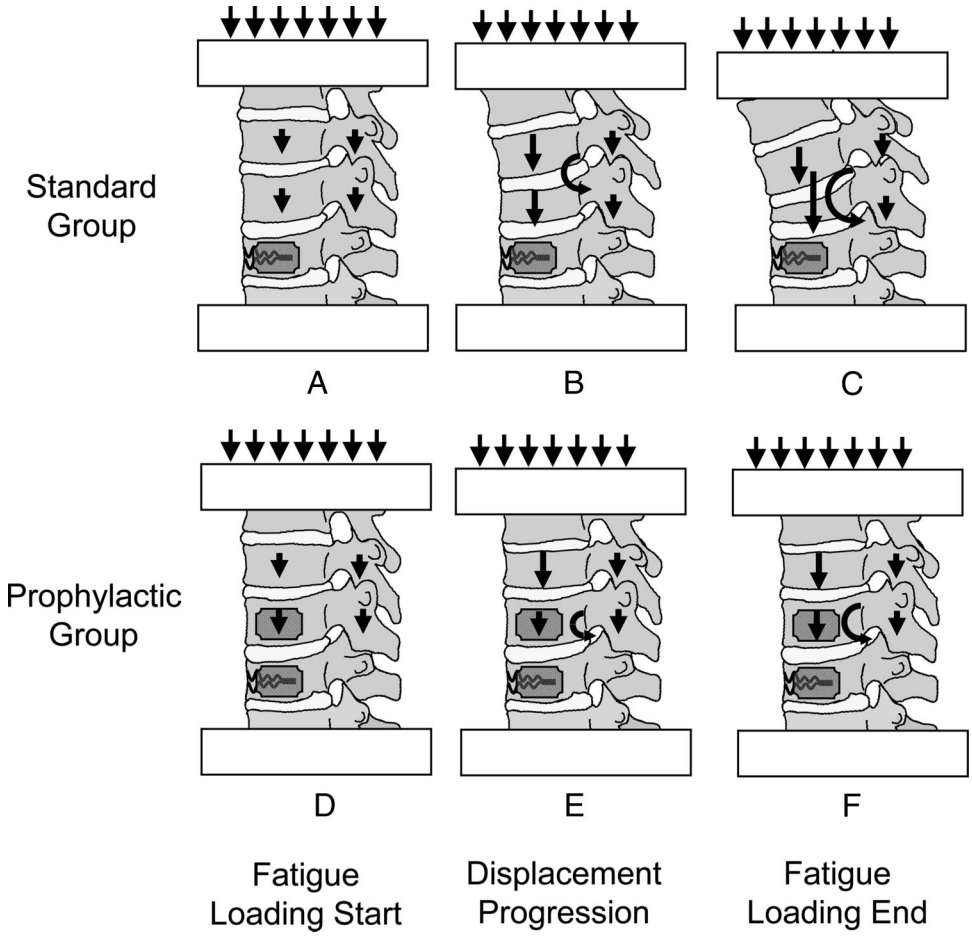

Figure 5. The deformation progression of vertebrae in standard group and prophylactic group during fatigue loading. The deformation of vertebral body and facet joint is similar along the spinal column when fatigue loading start (**A**, **D**). During the progression of fatigue loading, the deformation of osteoporotic vertebral body is higher than the one of facet joint (VB1, VB 2 in **B**, and VB1 in **E**). The pure compression may therefore create a minor forward bending moment, and further compress the center vertebra (**C**). The prophylactic augmentation decreases the deformation of vertebra and the induced flexion moment, hence minimize the deformation of adjacent intact vertebra (**E**, **F**). The length of vertical vector within the vertebra represents magnitude of deformation. The size of circular vector represents the magnitude of flexion moment.

of vertebral height of current study was lower than 3%, which was lower than the results of previous *in vitro* study, but close to the one of *in vivo* study. The low magnitude of height loss of vertebral body may be due to the relatively mild fatigue loading.

The height loss of intact vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra was significantly lower than the one adjacent to a 1-level augmented (or injury augmented) vertebra. This result may imply that the prophylactic augmentation not only reduce the failure risk of intact-augmented vertebra itself, but also reduce the failure risk of adjacent intact vertebra. Several numerical simulation, *in vitro* and *in vivo* studies hypothesized that the AVF was due to increase of disc pressure, deflection of endplate hence change of load path,[9,31] load shift from augmented vertebra to intact vertebra,[8] and the progression osteoporosis.[4] A lower stiffness of vertebra by reducing the cement filling ratio[32] or injecting low-modulus bone cement[33] was even proposed to reduce the risk of AVF. However, it seems that the phenomenon found in this study cannot be thoroughly explained by these theories.

Two possible properties, at least, between the 1-level augmented (or injury-augmented) vertebra in standard group and the 2-level augmented (or intact-augmented) vertebra in prophylactic group can be suspected to cause the variations in height loss of adjacent level, *i.e.*, the cement filling ratio and the vertebral level.

No decisive conclusion of the optimized cement filling ratio was yet agreed among spine researchers. Previous studies showed that a partially filled cement[32] or low-modulus bone cement[33] may be better to reduce the risk of AVF. However, our study does not support this conclusion. The filling ratio, which reflects the vertebral stiffness and strength, of intact-augmented vertebrae is slightly higher (not statistically significantly though) than the injury-augmented one, but the height loss of vertebrae adjacent to the intact-augmented vertebrae is significantly lower than the injury-augmented one. This phenomenon suggests that the height loss of adjacent vertebra may not be closely connected with filling ratio of augmented vertebrae.

The adjacent intact vertebra in standard group is located at the center level, whereas the one in prophylactic group is located at the upper level. Ideally, the deformation along the vertical line of a spine column should be similar if the loading is in pure compression (Figures 5A, D). However, since the vertebra is osteoporotic, the vertebral body may deform more than the facet joint does during the progression of fatigue loading (VB1, VB2 in Figure 5B, and VB1 in Figure 5E). As the deformation progressed, the facet joint behaves like an anchor, and

the pure compression hence creates a minor forward bending moment. Since the deformation of standard group is larger than the one of prophylactic group, so is the flexion moment (Figures 5B *vs.* 5E). During the flexion compression, the highest deformation locates at the center of concave side, where is the level of center vertebra. As the deformation increases with fatigue loading, so does the flexion moment, and then, a vicious circle is formed (Figure 5C). A prophylactic augmentation will break or reduce the effect this vicious circle by decreasing the deformation of vertebra and the induced flexion moment (Figure 5E, F). This point of view is also similar to the conclusion of a finite element analysis that the AVF may be due to the anterior shift of the upper body, but not the elevated stiffness of augmented vertebra.[10]

Bone Mineral Density

In this study, we scanned the specimen without soft tissue. The most used protocol of *in vitro* DEXA examination was the scanning with immersion of 20 cm water. Our internal study showed that the scanning of specimens without soft tissue decreased the magnitude of BMD by 5% when compared with scanning of 20 cm water immersion. The current study used thoracic specimens, and the averaged BMD was 0.564 g/cm^2. It can be expected that the BMD would be 0.570 g/cm^2 when examined with 20 cm water immersion. The database that translates BMD into T-score is not available for thoracic vertebrae. The 0.570 g/cm^2 BMD magnitude corresponds −2.7 T-score for L1 vertebrae. Our internal study of the data regression of T-score *versus* BMD showed that higher level of vertebrae would reach better T-score at the same magnitude of BMD. The averaged T-score of tested specimens should be at level of −2.7 or higher/better.

The BMD or age[34] is a clinical index for risk prediction of osteoporotic vertebral fracture. A previous study showed that both vertebral strength and stiffness were dependent on the BMD.[35] The current study showed that neither the bone strain compliance nor the loss of vertebral height can be correlated by the magnitude of BMD. This may be due to the already lowed BMD. A similar results was also found that vertebral stiffness was least affected by the BMD when the BMD was below 0.8 g/cm^2.[36]

■ Conclusion

The finding of current study suggests that the AVF may be due to the compression flexion induced by the anchoring effect of facet joint. The prophylactic augmentation strengthens the osteoporotic vertebrae, decreases the progression of vertebral height loss, reduces the anterior body shift, and hence protects the adjacent intact vertebra from elevated flexion bending. It can be cautiously suggested that if the vertebra is osteoporotic and adjacent level is located at pivot or lordotic level of spinal column, where is the site that the anterior body shift effect will be accelarated, the prophylactic augmentation may be an option to prevent the AVF.

■ Key Points

- The strain compliance of cortical shell is generally not a sensitive indicator to predict risk of fatigue injury if the fatigue loading is mild.
- The height loss of intact vertebra adjacent to a 2-level augmented (or intact-augmented) vertebra is significantly lower than the one adjacent to a 1-level augmented (or injury-augmented) vertebra.
- The prophylactic augmentation not only reduces the failure risk of the augmented vertebra but also the adjacent intact nonaugmented vertebra from fatigue injury.
- Neither the strain compliance of cortical shell nor the height loss of vertebral body is connected with bone mineral density if the vertebra is already osteoporotic.
- If the vertebra is osteoporotic and the augmented vertebra is located at pivot or lordotic level of spinal column, the prophylactic augmentation may be helpful to prevent the adjacent vertebral failure.

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