

Mechanical Behavior of Lumbar Spine Functional Units in Response to Cement Augmentation of Vertebra Body

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Introduction: Cement augmentation in vertebrae is used to promote mechanical strength after spinal fracture and recently vertebroplasty gaining popularity as a treatment for patients. The numerical simulation could be helpful to enhance the quality of treatments such as vertebroplasty via exact modeling of the lumbar spine. **Materials and Methods:** In this study, a three-dimensional finite element model created from CT images of L1-L3. According to clinical observation and recent studies, we consider that L2 augmented with two different volumes in 10 different distributions. Loadings were assumed to be pure momentum which applied in three anatomical directions (axial rotation, flexion, and lateral bending). **Results:** Our results were validated with experimental data which shows segments range of motion, ligaments forces, and intradiscal pressure had good agreement with our results. Cement augmentation increases max Von Misses stress in L2 cancellous bone and Increment in Cement volume has the same result. Cement augmentation increases L1-L2 intradiscal pressure. Cement augmentation decreases segments range of motion. Finally, Cement augmentation increases total stiffness of model. **Conclusion:** Taken together, vertebroplasty as a well-known method to treat the fractured vertebra, could be optimized to enhance patients' range of motion and decrease the complication of treatment.

Keywords: Vertebroplasty; Cement Augmentation; PMMA Cement

Introduction

Bone augmentation of the vertebra body i.e., vertebroplasty with polymethyl methacrylate (PMMA) cement was presented as a treatment for filling a tumor cavity in 1987 (1, 2). Different groups of patients treated by the method included those who encounter with osteoporosis, trauma, osteolytic metastasis, or hemangioma (3). It seems more detailed investigation needs to propose safe and optimized method of treatment of vertebral compression fracture.

In addition to stabilizing fractured vertebra, vertebroplasty significantly decreases the pain in patients (5, 6). It has been shown that the physical properties of PMMA cement and its distribution have a high impact on the mechanical response of the spine (3, 7). Also, clinical observations reveal that in a long-

term period patients treated by vertebroplasty suffered from a fracture of adjacent vertebra body (3, 8). Moreover, the physiological condition of the intervertebral disc (IVD) plays a pivotal role in multi-functional spinal units (FSU). It has been revealed that degeneration of IVD affects the mechanical behavior of FSU which could affect other treatment symptoms (9).

A variety of experimental and numerical biomechanical studies have examined the effects of cement augmentation on adjacent vertebra fracture but the results are controversial. Most of the studies have reported that augmented cement increases the stiffness of vertebra (7,10,11). In contrast, a minority of researchers found augmented cement did not affect the vertebra stiffness (12). The others found alternation in adjacent segment stresses distribution with vertebroplasty (13). It has been reported that the incidence of none osteoporotic patients was

36% and it seems that compression loading in studies is not sufficient to investigate spinal column response (3). Hence multi-directional loadings should consider in all experimental and numerical studies to investigate cement augmentation effects.

In this study, FSU model is investigated to show how the mechanical response is altered by cement augmentation. Geometry, soft tissue nonlinear stress-strain, and complex contact cause nonlinearity in the nature of segments range of motion. Intervertebral disc consists of the nucleus and annulus. The nucleus acts like an incompressible fluid and has mobility during loading. In contrast annulus ground substance occupied by organized oriented collagenous fiber layers. The average inclination angle of layers is ± 30 . Due to the collagenous structure of ligaments, they act as nonlinear spring to stabilized large motion of segments. Numerous interrelated parameters exist which are unknown during *in vivo* tests like intradiscal and abdominal pressure, therefore most of the data are obtained from in-vitro experiments and loading conditions in numerical models are applied in accordance with the in vitro studies.

In accordance with the importance of the function of lumbar segments after vertebroplasty, a three-dimensional nonlinear model of L1-L3 developed to investigate ROM, ligaments forces, intradiscal pressure and maximum Von misses stress while pure momentum applied in three anatomical directions i.e., flexion, axial rotation, lateral bending.

Materials and Methods

Model geometry

A three-dimensional model of two functional segment unit was used in this investigation. The geometry of the model was taken from CT images of 20-year women. The model consists of intervertebral discs and bony parts supported by ligaments. CT image was obtained by the cooperation of Jam institute in Esfahan in DICOM format. Mimics10 interactively read CT/MRI data in the DICOM format. Segmentation and editing tools enable the user to manipulate the data to select a bone and soft tissue. It provides an interface to Rapid Prototyping systems via sliced files with patented support structure generation. The 3D model was finalized in Rapidform XOR2 and components imported to ABAQUS6.10 to analyze the procedure.

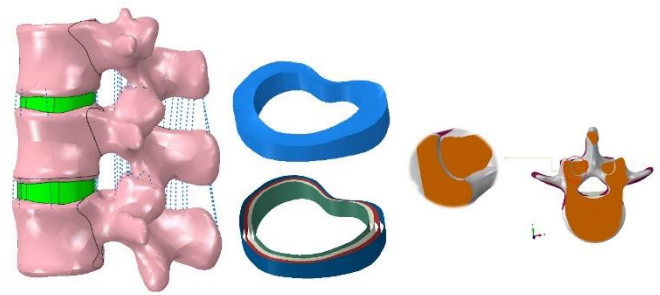


Figure 1. 3D component of L1-L3 model including annulus layers, annulus ground substance and facet joints

Intervertebral disc

Since Soft tissues are not visible in CT images, intervertebral disc created according to extrusion of bony endplates. Firstly, two endplates of adjacent vertebrae were selected based on the CT images, and then we assumed that the total volume between endplates was occupied with intervertebral disc. The portions of the total volume dedicated to the nucleus and annulus were selected based on the previous reports. The nucleus occupies 44% of IVD volume and acts as an incompressible fluid. To simulate real conditions, hydrostatic FLUID elements were used. Fluid property in healthy and degenerated condition were reported on literature (14). The bulk module of fluid was assumed to be 2300 for a healthy nucleus.

The annulus was reinforced by four membranes that were embedded in annulus ground substance and in each membrane, two layers were placed in opposite orientation angle (± 30 to the transverse plane of the layer) (15, 16). The point that the outer membranes are stiffer than the inner one was applied in the model (16). Also; 19% of annulus content is collagenous fibers therefore collagenous fiber thickness calculated and normal distance between fibers assume 1mm (15-17). Mooney-Rivlin hyperplastic material model used for annulus ground substance and material coefficients are $C01=0.54$, $C10=14$ (17).

Bony parts

The geometry of cortical bone constructed from CT image and its thickness is assumed 0.6 mm (18). About 12% of applied pressure distributed on the posterior element and it increases the stiffness of FSU. Isotropic elastic material used for the posterior element. On the other hand, cortical and cancellous bone



Table 1. Mechanical Properties of Ligaments

	ALL	PLL	LF	CL	ISL	SSL	Reference
Young module for small strain(Mpa)	7.8	10	15	7.5	8	10	[19]
Transition strain(%)	12	11	6.2	25	20	14	[19]
Young module for larg strain(MPa)	20	5	19	33	15	12	[19]
Max. failure force(N)	510	384	340	284	130	200	[20]
Spring number	5	5	8	6	6	3	<i>This study</i>

Anterior longitudinal ligament, ALL; Posterior longitudinal ligament, PLL; Ligamentum flavum, LF; Transverse ligament, TL; Capsular ligament, CL; Supraspinous ligament, SSL.

Table 2. Mechanical Properties of Annulus Layer

layer	Elastic module	Poassion ratio	refrence
1,2	550	0.3	[10]
3,4	485	0.3	[10]
5,6	420	0.3	[10]
7,8	360	0.3	[10]

Material is assumed anisotropic, reported in the literature. Superior surface of L1 was assumed to be rigid.

Ligaments

All Ligaments (ALL, PLL, SSL, ISL, FV, CL) were modeled as nonlinear spring using force-displacement curves. Springs were assumed to be a tension-only element. Anatomical origin and insertion of ligaments have a crucial role in model prediction but CT images are useless. Ligaments cross-sections, length, origin, and insertion obtain from literature.

Cement

PMMA cement injected in a fractured area which usually located in the anterior part of the vertebra. Cement not only stabilizes the fractured area but also helps height restoration. In this study, the cement cavity is located in 1/3 anterior and 2/3 anterior. In addition, symmetry of cement is taken into account. Cement elastic moduli were simulated as either 0.5 or 8.0 GPa. Although

Table 3. Mechanical Properties of Cement and Bony

part	Mechanical properties(MPa)	Refrence
Cortical bone	$E_{xx}=11300,$ $G_{xy}=3800,$ $\nu_{xy}=0.484$	[21]
	$E_{yy}=11300,$ $G_{yz}=5400,$ $\nu_{yz}=0.203$	
	$E_{zz}=22000,$ $G_{xz}=5400,$ $\nu_{xy}=0.203$	
Cancellous bine	$E_{xx}=140,$ $G_{xy}=48.3,$ $\nu_{xy}=0.45$	[22]
	$E_{yy}=140,$ $G_{yz}=48.3,$ $\nu_{yz}=0.315$	
	$E_{zz}=200,$ $G_{xz}=48.3,$ $\nu_{xy}=0.315$	
Posterior element	$E=3500, \nu=0.3$	[21]
PMMA cement	$E=3000, \nu=0.4$	[10]



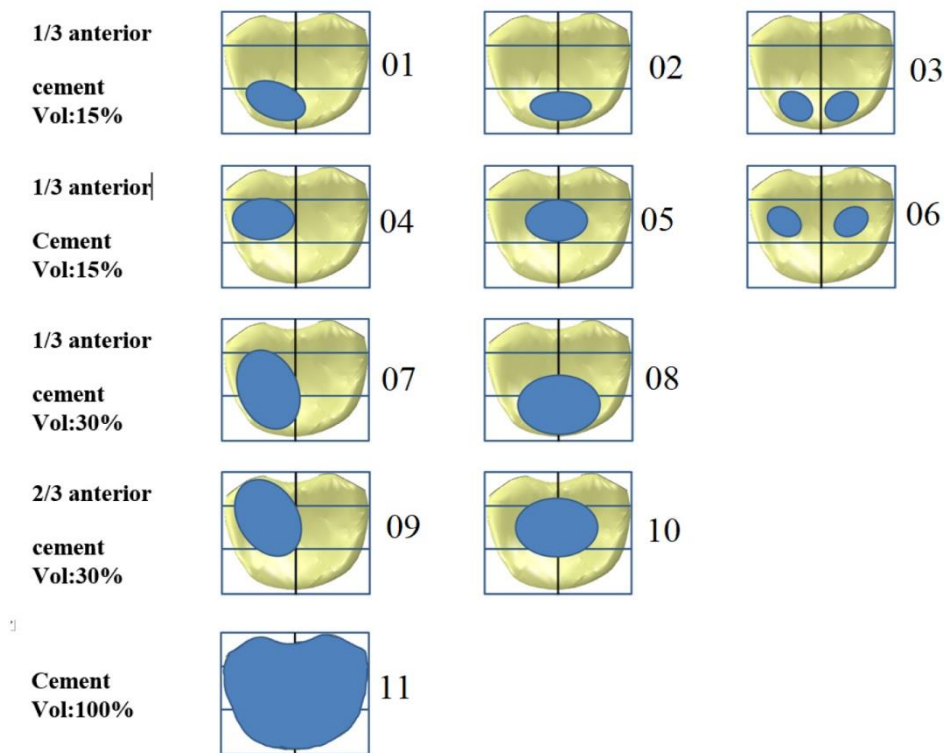


Figure 2. Cement distribution in L2 cancellous in 11 cement localization

cement modulus is typically about 2.0 GPa. Figure 2 displays a cross-sectional view of augmented models, showing the cement within the bone.

Contacts

Facet joint contact pressure-clearance has studied by some researchers but it gives higher contact pressures than in reality in the presence of cartilage. A soft-pressure overclosure would be more appropriate; however, this required knowledge of the relationship between contact pressure and clearance which has not been widely reported in the literature. Tied contact was assigned to the vertebral body-IVD contact surfaces and bone-cement surfaces.

Boundary and loading condition

Loadings condition assign to model according to *in vivo* testing protocol. The inferior surface of L3 assumed to be fixed and multidirectional loading applied on the superior surface of L1. In axial rotation superior surface of L1 restricted in coronal and sagittal surfaces. In flexion, to omit any asymmetry effects, L1

superior surface exclusively is free in sagittal surface and in lateral bending L1 superior surface exclusively is free in the coronal surface. Moments were applied on the superior surface of L1.

Results

Validation of model

To ascertain the validity of the FE model used in this study, validation was carried out by comparing predicted results with experimental results under similar loading and boundary conditions. Segments Range of Motion (ROM), the tension in the annulus, force in ligaments, and intradiscal pressure are compared with literature. ROM of segments in multidirectional loadings have a good agreement with experimental results. Although in flexion our model has less ROM than experimental results but in axial rotation and lateral bending results show good agreement. Figure 3 depicts the ROM of segments in a multidirectional moment.



In comparison to other FEM studies, results are in accordance with Wang *et al.*, (23) which assigned a viscoelastic model to investigate lumbar ROM. Figure 3 shows that L2-L3 ROM in multidirectional loading has good agreements with other FEM model. Ligaments force in flexion compare with Rohlmann *et al.*, (24) in Figure 5 and the Order of forces in ligaments shows good agreement, although max force occurred in different ligaments. Maximum loads in ligaments in multidirectional moments showed that ligament function in flexion significantly important. CI and ALL have maximum force in axial rotation and lateral bending, respectively.

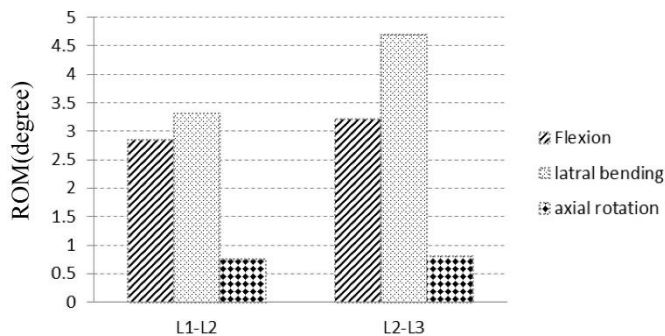


Figure 3. Segments range of motion in multidirectional loadings

The nucleus is assumed to act as a fluid cavity with hydrostatic fluid. This characteristic helps to obtain intradiscal pressure in multidirectional loadings. Figure 6 shows that maximum intradiscal pressure occurred in lateral bending. Wilke *et al.*, measured intradiscal pressure by a pressure transducer. They reported that the highest pressure occurred in flexion and Rohlman *et al.*, reported the same pattern.

According to results, the model is validated and capable to investigate cement augmentation effects on the mechanical response of the lumbar spine.

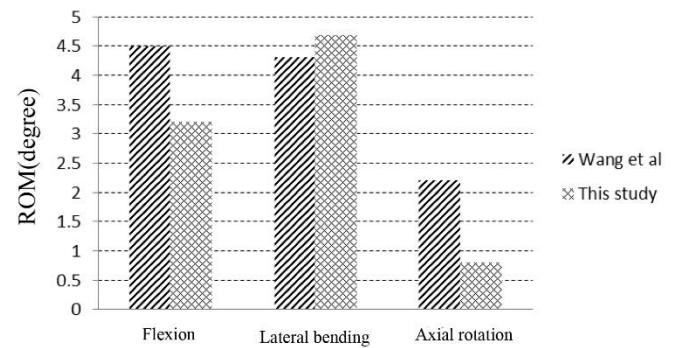


Figure 4. L2-L3 Rang of motion comparing with Wang *et al.*,

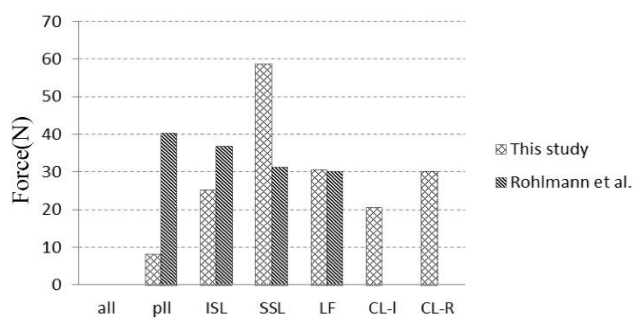


Figure 5. Ligaments force in flexion comparing with Rohlman *et al.*,

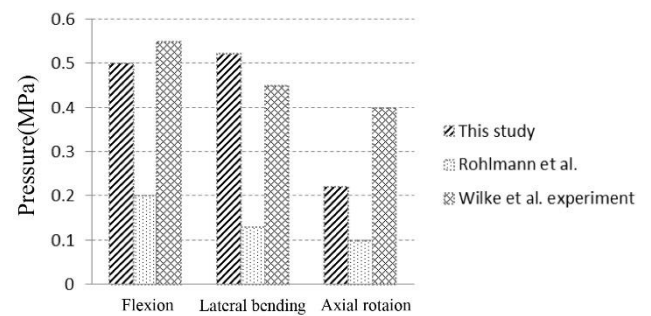


Figure 6. Intradiscal press in IVD1



Vertebroplasty

Cement distributed in 11 different cases in L2 cancellous bone. Loadings and boundary conditions were applied on L1 and L3. Cement presence alters stress distribution in L2 cancellous, intradiscal pressure in L1-L2 IVD, and total stiffness of model.

Maximum Von misses stress in annulus affects by cement augmentation. Figure 7 shows that in axial rotation max stress in IVD12 increases but stress in IVD23 decreases after cement augmentation. Max stress in flexion and lateral bending are not sensitive to cement augmentation. Cement augmentation increases total segment stiffness in all cases (Table 4). In axial rotation, stiffness increases more than others.

Figure 8 shows that increasing cement volume increase intradiscal pressure more than 5% in flexion. In lateral bending pressure increase about 4% but in axial rotation pressure remains constant.

Figure 9 shows that axial rotation max Von misses stress increases in L2 by cement augmentation. Average max Von misses stress in 1/3 anterior is lower than 2/3 anterior. When cement volume is lower than 15% max Von misses stress does not change. When cement injected in bipedicular method, max Von misses stress increase. Flexion and lateral bending results show the same pattern.

Table 4. Total stiffness of L1-L3 in multidirectional moment

	00	1	2	3	4	5	6	7	8	9	10	11
L1-L2 stiffness in Axi. Rotation (NM/Radian)	0.1127	0.1128	0.1128	0.1128	0.1129	0.1128	0.1130	0.1130	0.1132	0.1132	0.1133	0.1345
L1-L2 stiffness in flexion (NM/Radian)	0.02774	0.02777	0.02777	0.02776	0.02776	0.02776	0.02777	0.02780	0.02781	0.02779	0.02779	0.02798
L1-L3 stiffness in lat. bending (NM/Radian)	0.02774	0.02777	0.02777	0.02776	0.02776	0.02776	0.02777	0.02780	0.02781	0.02779	0.02779	0.02798

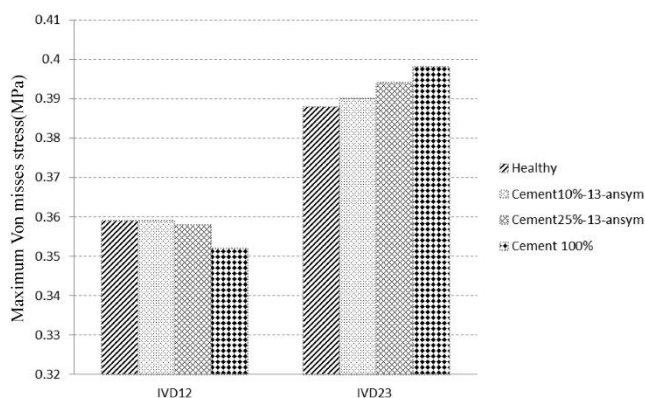


Figure 7. Max Von Misses stress in annulus during axial rotation

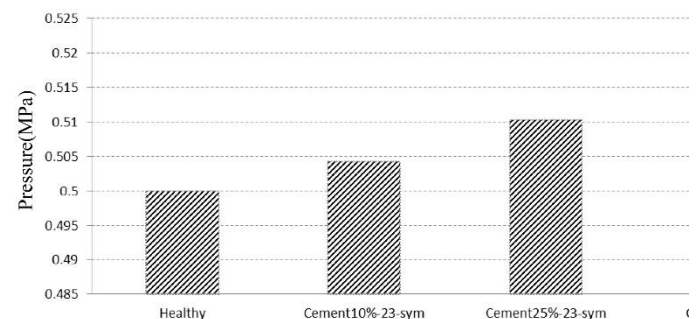


Figure 8. Interdiscal pressure in IVD L1-L2 during flexion



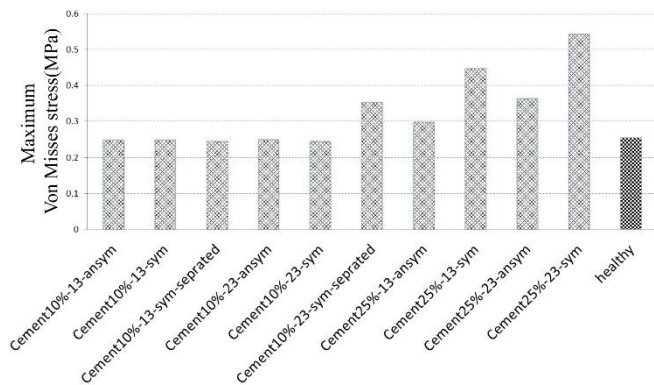


Figure 9. L2 cancellous max Von misses stress in different cement distribution

Discussion

The underlying motivation for this study was to develop a FEM of the lumbar spine that could ultimately be used to identify the effect of cement augmentation in Vertebroplasty on the mechanical response of two functional spine unit.

Three dimensional detailed model of L1-L3 is presented in this paper. Pure moments apply to simulate in vitro test conditions. Nucleus model by fluid elements. Model validated by experimental and FEM results in the literature .

Segments ROM in axial rotation and lateral bending have good agreements with literature, in flexion model prediction is lower than the reported data by other researchers. In addition, ligaments force in flexion is higher than the other cases. Ligaments properties might be the origin of extra flexion stiffness. Ligaments forces show that in axial rotation capsular ligaments and facet joints play an important function to stabilize segments motion and SSL is significantly important in flexion. Also in an in vitro study, Mimura *et al.*, (1994) found that the range of motion is reduced for degenerated discs in flexion and lateral bending while it is increased for axial rotation .

Cement augmentation effects investigate by some numerical and experimental studies. Micheal *et al.*, performed a comprehensive finite-element analysis to provide a theoretical framework for understanding and optimizing the biomechanics of the Vertebroplasty procedure. They found that only a small amount of bone cement (~15% volume fraction of fill) is necessary to restore the compressive stiffness of the damaged vertebral body to its value before the damage. Furthermore,

modest increases in the volume of the cement can substantially increase vertebral stiffness beyond its intact value.it should be mention that they just apply compressive load in their model.

Berlmann *et al.*, 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. Our model predicts that cement augmentation increases Max Von misses stress in L2 cancellous bone. This change could be the reason of lower strength.

Baroud *et al.*, found that cement augmentation increase intradiscal pressure under compressive load. Our results predict the same results and they claim that this extra pressure could be a risk factor for adjacent vertebra fracture. Cement volume effects were investigated by some researchers. Most of them found that increasing cement volume, increase FSUs stiffness and our results approved their funding. Yuan *et al.*, (25) reported that augmentation of a lower amount of cement volume not only has less complication compare to higher volume but also restore patient function. By considering our analysis, we proposed to the clinicians to augment a lower amount of cement volume for fractured vertebra, which is in accordance with clinical observation.

Conclusion

Taken together, by increasing cement volume, max Von misses stress was increased in L2/L3 annulus. Hence it is obvious that force distribution in lumbar structure is disrupted by cement augmentation. In addition, cement presence in lumbar structure increased max Von misses stress in cancellous bone and intradiscal pressure in the nucleus. We postulate that these alternations have synergically increased the chance of fracture of adjacent vertebrae.

Conflict of Interest: 'None declared'.

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