Published in final edited form as:

Comput Methods Biomech Biomed Engin. 2013 July ; 16(7): 736–746. doi:
10.1080/10255842.2011.636741.

Simulation of the behaviour of the L1 vertebra for different material properties and loading conditions

Ibrahim Erdem, Ph.D, P.E.¹, Eeric Truumees, MD², and Marjolein C.H. van der Meulen, Ph.D³

¹Exponent Inc., 420 Lexington Ave. Suite 1740, New York, NY

²William Beaumont Hospital and the Beaumont Comprehensive Spine Center, Royal Oak, MI

³Sibley School of Mechanical and Aerospace Engineering, Cornell University, Ithaca, NY

Abstract

Three-dimensional finite element models of the thoracolumbar junction (T12-L2) and isolated L1 vertebra were developed to investigate the role of material properties and loading conditions on vertebral stresses and strains to predict fracture risk. The geometry of the vertebrae was obtained from CT images. The isolated vertebra model included an L1 vertebra loaded through polymethylmethacrylate (PMMA) plates located at the top and bottom of the vertebra, and the segment model included T12 to L2 vertebrae and seven ligaments, fibrous intervertebral discs and facet joints. Each model was examined with both homogeneous and spatially varying bone tissue properties. Stresses and strains were compared for uniform compression and for flexion. Including material heterogeneity remarkably reduced the stiffness of the isolated L1 vertebra, and increased the magnitudes of the minimum principal strains and stresses in the mid-transverse section. The stress and strain distributions further changed when physiological loading was applied to the L1 vertebra. In the segment models, including heterogeneous material properties increased the magnitude of the minimum principal strain by 158% in the centre of the mid-transverse section. Overall, the inclusion of heterogeneity and physiological loading increased magnitude of the strains up to 346% in flexion and 273% in compression.

Keywords

thoracolumbar; spine; vertebra; fracture risk; finite element model

Introduction

Osteoporosis is characterized by low bone mass and altered tissue properties that reduce bone strength and make the skeleton, especially the spine, hip, and wrist, susceptible to fracture under physiological loads. Osteoporosis is responsible for 1.5 million fractures^[1] and associated with \$18 billion costs annually in the U.S.^[1, 2] A better understanding of fracture risk is critical to reducing the considerable morbidity and mortality with which osteoporotic fractures are associated. Currently, fracture risk assessment is based on bone mineral density (BMD) measured by dual energy x-ray absorptiometry (DXA).^[3] However,

Ibrahim Erdem, PhD, PE, Engineer Exponent Inc., 420 Lexington Avenue Suite 1740, New York, NY 10170, Tel: 212-895-8112, Fax: 212-895-8199, ierdem@exponent.com

Eeric Truumees, MD, Orthopaedic Surgeon, William Beaumont Hospital and the Beaumont Comprehensive Spine Center, Royal Oak, MI 48073, Tel: 248.663.1909, Fax: 248.663.1901, truumees@wowway.com

Marjolein C. H. van der Meulen, PhD, Professor, 219 Upson Hall, Cornell University, Ithaca, NY 14853, Tel: 607-255-1445, Fax: 607-255-1222, mcv3@cornell.edu

substantial overlap exists in the BMD values of individuals who do and do not sustain skeletal fractures^[4-7], suggesting a need for alternate approaches of fracture prediction.

Failure of the load-bearing ability of a bone results if the stresses due to an applied load are greater than the strength of the bone tissue at any location. In the vertebrae of the spine, for example, loading is applied through the intervertebral disc (IVD), facet joints and ligaments. The structural stiffness and strength of a whole vertebra are determined by the external geometry, cancellous microstructure, cortical shell structure and tissue material properties. Therefore, the contribution of these factors to loading and bone strength should be taken into account when predicting fracture risk. Finite element (FE) models have the potential to improve clinical predictions of bone fracture by taking the aforementioned factors into account.^[7-11] Moreover, FE models facilitate obtaining the distribution of tissue stresses and strains that cannot be measured experimentally. Hence, FE methods have been used to identify vertebrae at high risk for fracture as well as effects of fracture treatment techniques such as vertebroplasty and Kyphoplasty.^[11-23]

The accuracy and reliability of FE models depend on the specific geometry, material and boundary conditions selected. Simplified geometry, loading or material properties are often implemented to expedite model solution. In this study, the effects of loading conditions and material properties on FE-based predictions of L1 vertebral stresses and strains under different applied loads were investigated. An isolated L1 vertebra loaded through the vertebral body was compared with a motion segment containing T12-L2 vertebrae with associated IVDs, facet joints and spinal ligaments. To understand the role of material properties on the FE results, each model was examined with homogeneous tissue properties and with spatially varying material properties scaled directly from the bone apparent density. The solutions for the four models were compared under uniform compression and flexion since the fracture of the vertebrae frequently occurs under these applied loads.

Methods

Computed tomography (CT) images from a 55-year-old female cadaver, who was in slightly flexed position (5 degrees) and had normal bone density and no evident tissue deterioration or excessive bone tissue loss due to osteoporosis, were used to generate the bony geometry of the T12, L1 and L2 vertebrae. The scans were taken on a clinical scanner (Philips Mx8000 IDT 16) with 1 mm axial slices and 0.453 mm in-plane voxel dimension. The height of all three vertebrae at the center of the body was 26 mm. The height of the IVDs varied through the sagittal section. The cranial IVD (T12-L1) and caudal IVD (L1-L2) had heights of 5.9 mm and 8.5 mm at the center, respectively.

To develop the FE models of the vertebrae, first, the surface geometry of each vertebra was extracted from the CT scan (Mimics, Materialise, Belgium). The resulting geometry consisted of a triangular surface mesh in STL format. Next these surfaces were converted to surface splines in NURBS format (Studio, Raindrop Geomagic, NC). Finally, the surface models were converted to hexahedral (brick) FE meshes (Truegrid, XYZ Scientific Applications, CA). The macro scale FE meshes modeled the overall geometry of the vertebrae but are not able to capture the physical microstructure of the vertebrae. A single mesh size was used for all models, based on a mesh refinement analysis. The mesh fineness was determined when the average stresses and strains within the regions of interest differed less than 5% between the coarse mesh and fine mesh. Although the coarse mesh was satisfactory, the fine mesh was used to better capture the geometry of the vertebrae.

Two FE models were created to represent simplified and anatomic boundary conditions (Fig. 1). The first model, referred to as isolated L1, included only the L1 vertebra with loads

applied to the body through flat polymethylmethacrylate (PMMA) plates interfacing with the irregular surface of the superior and inferior endplates of the L1. A perfect bond between the endplates and the PMMA was assumed. The isolated L1 model consisted of 26,452 eight-node hexahedral (brick) elements. The second FE model, referred to as T12-L2 spinal motion segment, included three vertebrae (T12, L1 and L2) and two IVDs. The vertebrae and IVDs were modeled with 67,156 eight-noded hexahedral elements. The L1 vertebra was identical to the isolated L1 model. The motion segment included seven structurally relevant ligaments modeled by linear elastic cable elements: anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), intertransverse ligament (ITL), supraspinous ligament (SL), interspinous ligament (IL), ligamentum flavum (LF) and capsular ligament (CL) (Table 1). Ligament locations, cross-sectional areas and stiffnesses were selected in the range obtained experimentally^[24, 25] and used previously in the literature^[9, 17] since they could not be determined from the CT data. Facet joints between the vertebrae were modeled as frictionless contact surfaces braced with capsular ligaments. In total, 2,752 two-node axial elements were used to model the ligaments and disc fibers in the spinal motion segment.

To model the IVDs, the nucleus pulposus and annulus fibrosis were included. The annulus fibrosis was represented by four layers of stiff fibers embedded in a soft ground material ($E=4.2$ MPa).^[9, 16, 17, 19] The disc fibers, comprising 19% of the volume of the annulus,^[16, 26] were oriented in a crisscross pattern making an angle, on average, of 30 degrees with the endplates of the vertebrae^[16, 17, 19, 26] and were also modeled with cable elements. Moving from the inner to the outer annulus layer the spacing between the concentrically placed fibers was reduced by 81% at each layer whereas the cross-sectional area and elastic modulus were increased by 11% to 25%^[17] to represent the anisotropy and heterogeneity of the annulus fibrosis (Table 1). The nucleus, comprising about 40% of total IVD volume,^[26] was modeled as a nearly incompressible solid with stiffness of 0.2 MPa^[9, 27] to represent its gelatinous behavior.

For each model two bone material distributions were simulated: an isotropic homogeneous elastic modulus of 1000 MPa, or an isotropic heterogeneous modulus assigned based on an exponential density-elasticity relationship.^[28] A linear relationship between the Hounsfield Units (HU), obtained from CT scans, and bone apparent density (ρ , g/cm³) was utilized:

$$\rho=0.18+0.001244 \text{ (HU} - 100\text{)}.$$

The elastic modulus (E , MPa) of the bone tissue was then assigned using the density-elasticity relationship of Morgan et al.:^[28]

$$E=4730 \rho^{1.56}$$

The moduli ranged from 52 to 7970 MPa and the average elastic modulus in the centrum of the vertebrae varied from 200 to 300 MPa, consistent with the range given for the cancellous bone (Fig. 2).^[29–31,45]

Two loading cases were investigated for each model: a uniformly distributed compressive force of 400 N or a pure flexion moment with a magnitude of 7.5 Nm applied to the superior endplate to determine the effects of each loading separately. A bending moment case was also investigated for validation of the segment model. The moments were created by applying concentric forces on the superior end plate of PMMA (isolated L1 model) or T12 (segment model) while creating zero axial force. Both models were restrained in all

directions at the inferior-most endplate. PMMA was modeled as homogenous and isotropic with an elastic modulus of 2,500 MPa and Poisson's ratio of 0.3.

Solutions were obtained using a commercial finite element package (ABAQUS, Hibbitt, Karlsson & Sorensen, Inc, RI). For the isolated vertebra models, we calculated axial and flexion stiffness to characterize the global behavior of the vertebra and to validate our results with previous experimental^[10, 32, 33] and analytical studies.^[10, 21] For the segment model, we calculated axial displacement in compression, and rotations in flexion and bending to obtain the global behavior and to validate our results with previous experimental^[34–39] and analytical^[17] studies. The principal stresses and strains, and von Mises stresses in the mid-transverse and mid-sagittal sections of the L1 vertebra were compared across boundary conditions and material properties. Since the most common fracture mode was compressive, we mostly illustrated the minimum principal strains. Von Mises stresses were also calculated because they have been used as a fracture risk assessment method.

Results

We validated the assigned heterogeneous bone properties by calculating the axial stiffness of the isolated L1 vertebra model and comparing the value with the axial stiffness reported by other researchers. The calculated axial stiffness of 14 kN/mm was comparable to several previous studies.^[21, 32] However, our stiffness value was larger than those values reported in two other studies, probably due to elder (85±8 years old)^[10] or osteoporotic/osteopenic specimens used for previous investigations.^[33]

To validate the segment model, an axial displacement of 0.52 mm was calculated in compression, comparable with the values in the literature.^[34] The average bulging was 0.21 mm and 0.31 mm for the cranial and caudal IVDs, respectively, in the range reported previously.^[39] In flexion and bending 3.2 and 4.3 degree rotation was obtained, respectively, in the range obtained from previous studies^[17, 34, 36–38] that covers a wide range because of the differences in the test protocols and specimens. We also determined the ligament stresses for compression and flexion, which were small percentage of their strengths obtained experimentally.^[24, 25]

When the heterogeneity of the bone tissue was not included, the distributions of the stresses and strains of the isolated vertebra were affected only by the loading and external geometry. The asymmetric shape of the vertebral body and the irregular curvature of the superior and inferior endplates contributed to variations in the stress and strain distributions in the cross sections under applied compression (Figs. 3,4). Although the compressive load was uniformly distributed, the anterior side of L1 was subjected to higher stresses and strains than the posterior aspect. The largest magnitude minimum principal strain and the maximum von Mises stress in the mid-transverse section occurred at the left lateral cortex of the vertebral body (Table 2). When the isolated homogeneous vertebra was loaded in flexion, stress and strain distributions were similar to those of a homogenous beam under flexure (Fig. 5), i.e., the magnitudes of the stresses and strains were highest at the anterior and posterior sides of the vertebral body and lowest at the center (Table 3).

Accounting for material heterogeneity changed the mechanical behavior of the L1 vertebra. The axial and flexion stiffness of the L1 vertebra were reduced by 55% and 39%, respectively. Heterogeneity of the bone tissue also altered the distributions of stresses and strains in the mid-transverse and mid-sagittal sections of L1 for compression. The magnitude of the minimum principal strain in the mid-transverse section exceeded the homogeneous maximum value at the left and right lateral and anterior cortex by at least 125%. Moreover, the largest von Mises stress in the mid-transverse section increased by 45% at the left cortex

(Table 2). Although the heterogeneous and homogeneous L1 models had similar stress profiles on the superior endplates, the regions of high stresses and strains were altered. Heterogeneity of the bone tissue caused both stress relief and increased stresses and resulted in a slight shift of the stresses towards the pedicles (Fig. 3). The inclusion of heterogeneity caused relatively more dramatic changes in the distributions of the stresses and strains for flexion (Figs. 5–7). Although the highest stresses and strains occurred at the anterior and posterior sides of the vertebral body, the change in the minimum principal strain was greatest at the center of the vertebral body (134%) (Table 3). The changes in the minimum principal strains were almost always higher than the changes in the von Mises stresses (Table 3). In the mid-sagittal section, the highest magnitude minimum principal strains shifted from the anterior superior endplate to just above and below the mid-anterior cortex of the vertebra (Fig. 7).

The stress and strain distributions changed further when physiological loading was applied to L1 through T12 and L2, the intervertebral discs, ligaments and facets. In compression, although the facets and capsular ligaments contributed to the load transfer, the disc mainly transferred the load to the neighboring vertebra. Including the capsular ligaments and facet joints shifted the stresses towards the posterior side of the vertebra (Fig. 3). The magnitude of the minimum principal strain at the anterior lamina in the mid-transverse section of the homogenous model increased 11.8-fold with the more physiological loading (Fig. 4). The maximum change in both von Mises stresses and magnitude of the minimum principal strains occurred in the posterior cortex of the mid-transverse section with more severe changes in the strains (Table 2). The magnitude of the stresses and strains at the left cortex in the mid-transverse section were at least 29% larger than those at the right cortex (Table 2). Although the inclusion of heterogeneous bone properties did not change the overall displacements or rotations of the vertebrae (less than 5%), the strains were affected more than the stresses (Figs. 3,4). The von Mises stresses at the center and anterior of the mid-transverse section increased by only 9% (Table 2). However, the magnitude of the minimum principal strains increased by 99% to 166% in all five regions with heterogeneous bone properties (Table 2).

The physiological loading had remarkable effects on the response of L1 when the segment was loaded in flexion. Facets and all ligaments, except the anterior longitudinal ligament, played a significant role in the load transfer mechanism. In the mid-transverse section, stresses and strains shifted from the posterior vertebra towards the laminae (Figs. 5,6). When compared to the isolated L1 model, von Mises stress and the magnitude of the minimum principal strain in the anterior cortex in the mid-transverse section of the L1 in the segment model were reduced by 61% (Table 3), implying that the load was partially transferred to the posterior elements, facets and ligaments. In the mid-sagittal section of the segment, the high magnitude stresses and strains observed in the anterior cortex of the isolated vertebra body reduced and the minimum principal strains at the superior posterior wedge diminished (Fig. 7). Unlike the isolated vertebra in flexion, the magnitude of the stresses and strains in the mid-transverse section were at least 31% higher in the left cortex than right cortex (Table 3). As in compression, the heterogeneity of the bone tissue did not substantially change the overall displacements or rotations of the vertebra (less than 5%) but did change the stresses and strains. The change in the minimum principal strains was more pronounced than that of the von Mises stresses in the mid-transverse section (Table 3). In the segment models, including heterogeneous bone properties increased the magnitude of the minimum principal strain by 61% and 50% at the center and anterior of mid-transverse section compared to the homogenous case (Table 3, Fig. 6). In addition, the inclusion of heterogeneity in the segment model increased the magnitude of the minimum principal strains in the inferior endplate in the mid-sagittal section under flexion (Fig. 7).

Discussion

In this study, we investigated the effects of heterogeneity of bone tissue and physiological loading on stress and strain predictions in the L1 vertebra using macro scale FE models. Not surprisingly, different material properties and loading conditions changed the stresses and strains up to 346% in the transverse and sagittal sections examined. In compression, the changes in both stresses and strains were most pronounced in the posterior of the vertebral body, whereas in flexion the most-affected region varied depending on the outcome parameter, and material or loading conditions. In general stresses were affected more by the loading conditions, whereas changing the material properties substantially altered the strains. The overall changes across the models were most pronounced in flexion. In the laboratory, bone tissue failure depends on the applied strain levels,^[23, 32] underscoring the need for accurate local strains to predict failure.

For our investigation, we chose the thoracolumbar junction (T12 to L2) as osteoporotic fractures occur most frequently here and in the mid-thoracic spine.^[4, 40] One prior T12-L2 junction model did not include posterior elements or ligaments^[15], which we believe affected their results at the L1 vertebra. Another thoracolumbar junction model included only T12 and L1,^[19] excluding the disc under the L1, which also probably affected the stresses and strains at the L1 vertebra. On the other hand, our model had an L1 vertebra that was loaded through IVDs, and had the ability to capture the load shared by the posterior elements with the incorporation of the ligaments and facet joints. Including the posterior structures better simulated the kinetics of the L1 vertebra; hence, our model better captured the physical loading conditions of the L1. Many studies have analyzed isolated vertebral bodies in the thoracic and lumbar spine without simulating physiological loading.^[7, 11, 21–23, 43, 44] Our results show that both the values and distribution of stresses and strains are very different when physiological loading is applied, affecting both failure location and timing.

Studies of spinal segment models^[8, 9, 14–20, 45] focused mainly on the behavior of the IVD, ligaments and facets, and modeled these structures in detail, or the overall mechanics of the motion segment. As our results show, the overall behavior of the spinal segment with and without heterogeneous bone properties did not change. Among the spinal segment models, stress or strain values in the vertebrae, can be directly compared to our results.^[9, 14, 15, 17] These models were created using different levels in lumbar spine and reported stresses or strains under different load magnitudes. When scaled appropriately, our results are in agreement with several models^[15, 17] but our strain values are substantially lower than those reported previously, probably due to the use of a low elastic modulus (34 MPa) for the entire volume of cancellous tissue in the earlier study^[9]. Our stress values for the mid transverse section are lower than those reported for a different vertebral section (cranial).^[14]

In light of the above discussion the most significant strengths of our study are the incorporation of heterogeneous bone properties and physical boundary conditions for the stress analysis of the L1 vertebra. While homogeneous bone properties isolate the effects of geometry, loading and boundary conditions on vertebral stresses and strains, the material properties determine the load sharing (or stress distribution) within the vertebral tissue. Our strain distributions changed substantially due to the spatial variation of the elastic modulus of the bone tissue which will affect fracture risk prediction. The stress distribution in L1 when loaded through a PMMA plate was remarkably different than when loaded through IVDs, ligaments and facets due to several factors. In the segment model the vertebra was constrained by flexible IVDs, unlike the loading through less flexible plates in the isolated vertebra model. The material property differences between the homogeneous, isotropic PMMA and the heterogeneous, anisotropic IVD also contribute to the differences we

observed. Finally, the posterior processes play an important role in load sharing during flexion of the segment and are completely omitted in isolated vertebra models.

The incorporation of patient specific vertebral geometry and bone material properties is another strength of our approach. Modeling the vertebra with symmetry, average geometry or flat endplates will underestimate stresses and strains and change the load paths. As our results show, the stress and strain path is not symmetrical due to the asymmetrical vertebral geometry. Therefore, for better fracture risk assessment, simplifications in the geometry of the vertebra should be avoided.

Our models have some limitations. We did not conduct mechanical testing on the specimen to verify our model. Although the behavior of the models is similar to the data in literature, the exact behavior of the individual specimen may be different from what we obtained from the FE model due to the properties based on the literature for the IVD and ligaments. The height of CT slices might not have captured the endplate geometry accurately, which would alter the stress distribution on the endplates. We did not include viscosity, porosity and nonlinearity of the materials. Therefore, impact or time dependent loading or full mechanical behavior may not be simulated. Bone tissue is not isotropic^[41, 42] and we excluded the orthotropic (or transversely isotropic) behavior of the bone tissue in our model, which may underestimate the transverse stresses and strains. We analyzed the vertebrae under either compression or flexion to distinguish between the effects of the changes in boundary conditions or in material properties to the stresses and strains obtained from each individual load. However, in vivo the vertebrae are loaded in combined compression and flexion, so superposition of the two loads may be more appropriate for clinical investigations. Finally, we used a single model to illustrate the significance of material properties and loading conditions on fracture risk assessment due to the time required to develop a smooth patient-specific model.

A more detailed understanding of areas of vulnerability under loading will provide insights into failure locations and improve efforts towards reconstruction after fracture and fixation of osteoporotic vertebral bodies. Since current clinical methods cannot predict individuals who will fracture or respond to treatment, patient specific FE models provide an opportunity to mechanistically understand and predict vertebral failure. As our results show, including heterogeneity of the vertebral bone tissue, and physiological boundary conditions and loading are critical to determining the locations and distribution of highly loaded tissue that is at risk for failure. We have demonstrated these effects in a normal spine; the distributions and magnitudes of the stresses and strains will change further when material changes are present because of osteoporosis. In addition, treatments such as vertebroplasty, Kyphoplasty, spinal fusion and disc replacement will change vertebral load transfer and, therefore, fracture risk. To truly understand the effects such treatments have on the vertebra and select appropriate treatments, FE models need to incorporate patient specific material properties in conjunction with realistic loading conditions.

Acknowledgments

We thank Kirk Gunsallus and Jason Long of Cornell University for their assistance with the development of the FE models. We are grateful for the support provided by NIH grant AR053571.

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Figure 1.
Finite element models of isolated segment and T12-L2 segment

a)

b)

Figure 2.
Elastic modulus variation in A) mid-sagittal, B) mid-transverse section of L1

Figure 3.
von Mises stresses (in MPa) in the mid-transverse section due to compression

Figure 4. Minimum principal strains (in microstrains) in the mid-transverse section due to compression

Figure 5.
von Mises stresses (in MPa) in the mid-transverse section due to flexion

Figure 6.
Minimum principal strains (in microstrains) in the mid-transverse section due to flexion

Figure 7.
Minimum principal strains (in microstrains) in the mid-sagittal section due to flexion

Table 1

Ligament and disc fiber properties.

	ALL	PLL	ITL	SL	IL	LF	CL	F1	F2	F3	F4
Area (mm ²)	64	20	10	30	40	40	40	0.35	0.30	0.25	0.20
Modulus (MPa)	20	20	50	15	10	20	20	550	475	400	360

ALL: Anterior longitudinal ligament, PLL: Posterior longitudinal ligament, ITL: Inter-transverse ligament, SL: Supraspinous ligament, IL: Interspinous ligament, LF: Ligamentum flavum, CL: Capsular ligament, F1: Outer most anular fiber layer, F4: Innermost anular fiber layer

Table 2

von Mises stresses and minimum principal strains at the center, posterior, anterior, left and right cortex of the mid-transverse section of L1 under compression.

L1 homo	L1 hetero	T12-L2 homo	T12-L2 hetero	% difference between L1 homo and T12-L2 hetero
Von Mises stresses (MPa)					
center	0.385	0.330	0.307	0.335	-13
posterior	0.263	0.189	0.612	0.383	46
anterior	0.423	0.477	0.236	0.258	-39
left	0.555	0.805	0.492	0.665	20
right	0.407	0.538	0.347	0.467	15
Minimum principal strains (microstrain)					
center	-390	-837	-288	-767	97
posterior	-266	-634	-585	-1187	346
anterior	-421	-1002	-237	-518	23
left	-577	-1296	-513	-1021	77
right	-392	-906	-358	-743	90

L1: isolated vertebral body model; T12-L2: motion segment model; homo: homogeneous bone properties; hetero: heterogeneous bone properties

Table 3

von Mises stresses and minimum principal strains at the center, posterior, anterior, left and right cortex of the mid-transverse section of L1 in flexion.

L1 homo	L1 hetero	T12-L2 homo	T12-L2 hetero	% difference between L1 homo and T12-L2 hetero
Von Mises stresses (MPa)					
center	0.124	0.100	0.174	0.181	46
posterior	1.43	0.758	0.250	0.092	-94
anterior	1.24	1.18	0.484	0.472	-62
left	0.203	0.316	0.309	0.338	67
right	0.284	0.358	0.213	0.178	-37
Minimum principal strains (microstrain)					
center	-115	-269	-166	-429	273
posterior	-446	-623	-78	-58	-87
anterior	-1227	-2428	-479	-953	-22
left	-202	-503	-327	-566	180
right	-273	-597	-214	-313	15

L1: isolated vertebral body model; T12-L2: motion segment model; homo: homogeneous bone properties; hetero: heterogeneous bone properties