Finite Element Analysis of Osteoporotic Vertebrae with First Lumbar (L1) Vertebral Compression Fracture

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***Abstract*—The aim of this work is to assess the biomechanical response or load transfer response between osteoporotic (with first lumbar (L1) vertebral compression fracture) and healthy vertebrae in five vertebral physiological motions namely as compression, flexion, extension, lateral bending and axial rotation. For this purpose, an image-basedheterogeneous three-dimensional patient-specific of lumbar and thoracic spinal unit (T12-L2) finite element models for healthy and osteoporotic subjects were created.**

The finite element analysis have shown that one of the most significant effects of osteoporosis is the tendency to produce higher stress and strain in the cancellous region of the vertebral body. The maximum stress and strain was 4.53 fold (compression) and 5.43 fold (axial rotation) higher for the osteoporotic than the healthy subject, respectively, under the similar loading activity. Uneven stress distribution patterns also have been detected in the osteoporotic vertebrae rather than the healthy vertebrae. All of these characteristicsare reflected bya reduced structural strength and bone mass which might lead to an increased risk of fracture. These results strengthen the paradigm of a strong relationship between osteoporosis and its high susceptibility to fracture.

***Index Terms*—Biomechanics, finite element analysis, osteoporosis, vertebrae.**

I. Introduction

Osteoporosis is the most common disease affecting both men and women [1], and it is becoming increasingly prevalent in aging society [2]. Itsclinical significance lies in the high vulnerability and susceptibility to bone fracture [3]. It is characterized by low bone mass and micro-architectural deterioration of bone tissue [4]. Even though osteoporotic fractures can occur anywhere in the human body [5], the most prevalent fracture site is the spine [6], particularly in the elderly population [3]. In Japan, there are more than 10 million osteoporosis patients [7]. It is believed that this number will significantly increase in relation to Japan's life expectancy continues to rise. In the United States, about 1.5 million fractures due to osteoporosis are reported annually including over 700,000 vertebral fractures with high mortality rates. It was reported that, the survival rate was 72% after one year the symptom was first detected and this figure was then drastically reduced to only 28% after five years. Therefore, early detection of osteoporotic disease play a significant role in order to improve the health quality of the community and to organize early treatment as preventive and precautionary measures.

Human spine is consisted of 24 spinal bones which are known as vertebrae. The vertebral bone is the main compressive load-bearing structure in the spine.The vertebral bone is composed of a porous internal trabecular bone coresurrounded by a thin shell of cortex. In osteoporosis, the cortex layer of the osteoporotic patient is thinner than the normal people due to the effect of reduced bone mineral density even in the outer layer of bone. The vertebral bones are stacked on top of one another in a flexible curved structure to create the spinal column. The spinal is divided into three main segments namely as cervical (7 bones of the neck), thoracic (12 bones of the torso) and lumbar (5 bones of the lower back). The primary function of the spinal column is to provide body's main upright support [1]. It was reported that, the most favorite site of vertebral fracture is the first lumbar vertebra (L1) due its position near the inflection of the vertebrae.

In the previous study, it had shown that the cause of osteoporotic bone fractures were mainly contributed by normal daily load rather than traumatic events [6]. They are few possible justifications for such non-traumatic fracture. Firstly, since the osteoporotic vertebrae characterized by a reduced bone mass, therefore there is a less bone for load distribution. Hence, generating higher local tissue stresses and finally increased the risk of fracture [8], [9]. Secondly, uneven load distribution within the vertebrae has been recognized as one of the main contributors for such type of fracture [2]. This typical phenomenon is related to an altered trabecular micro-architecture with less transverse trabeculae exhibits by osteoporotic vertebral trabecular bone. Thirdly, the reduced strength of osteoporotic bone tissue may potentially increase the risk of fracture even for normal tissue load. Based on these findings, it can be concluded that it is adequate enough to accurately evaluate the load transfer response on the vertebral body only by considering the basic physiological motions of the spine namely as compression, flexion, extension, lateral bending and axial rotation.

Finite element analysis (FEA) has been introduced in the field of biomechanics over the last few decades. This methodology has been utilized in many clinical applications and gaining popularity especially in the prediction of vertebral strength due its subtle relationships that exist between structure and functionality under a variety of conditions.Moreover, due to the complexity and difficulty of in-vitro and in-vivo experiments, FEA seems to give more promising results. Furthermore, this computational approach reduces the cost and danger of other testing procedures, allowing one to achieve certain individualization when organ geometry and specific loading condition can be

Manuscript received May 14, 2014; revised July 14, 2014.

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DOI: 10.7763/IJAPM.2014.V4.297

fullycustomized by means of medical image treatment and biomechanics simulation technology.The reliability of FEA is subsequently strengthened by the recent finding which demonstrated its better correlations to vertebral strength then Dual X-Ray Absorptiometry (DXA) approach [10]. Recent studies, have also indicated that bone density measurements by DXA might not be an accurate diagnosis approach because they can be an overlap between bone densities measurements in people with and without osteoporotic bone fractures [11]. In pharmaceutical field of study, the FE methods have been adopted in order to evaluate the clinical effects of drug therapy and delivery for osteoporosis [12], [13]. FE methods have also been used to address complex clinical issues, such as the use of intervertebral implants [14], the assessment vertebral implants fixation stability [15], vertebral failure risk evaluation, or to investigate the controversial effects of vertebroplasty [16]-[18]. Concrete evidence of the applicability and accuracy of this numerical method had also been proven when this method had been used by several research group to study the mechanics of the normal and injured human spine. The results of these studies showed in a good agreement with measured data from in vitro studies [19]-[31].

In the present study, FEA was performed to quantitatively analyze and differentiate the stress and strain distributions on vertebrae of healthy and osteoporotic subjects under five basic vertebral physiological motions (compression, flexion, extension, lateral bending and axial rotation) as shown in the Fig. 1. The development of the osteoporotic vertebral FE model with first lumbar (L1) vertebral compression fracture (VCF) has opened new perspectives in understanding the behavior of the load transfer patterns in osteoporotic patients with severe vertebral deformity. Therefore, early treatments and preventive measures can be taken as soon as possible to avoid any untoward incidents such as bone failurein the future.

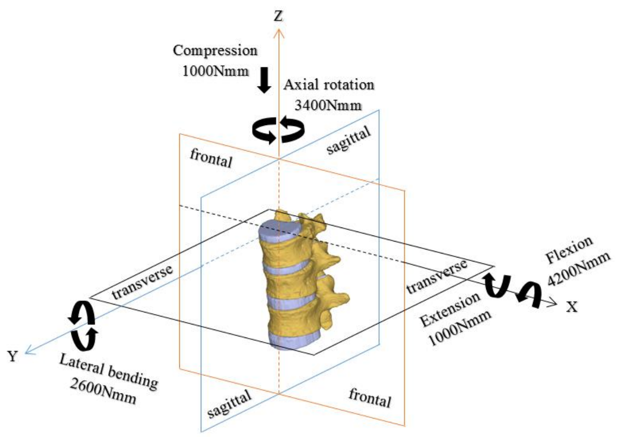

Fig. 1. Five basic vertebral physiological motions [32].

II. Materials and Methods

A. Patient Specific of FE Modeling

Bone geometrical features were extracted from CT images using MECHANICAL FINDER^{TM}software (Research Center of Computational Mechanics Co. Ltd. Japan) [33]. Individual complex bone shapes and heterogeneous bone density distributions were considered in this bone modeling procedure. Heterogeneous bone density distributions are related to the Young's modulus of bone, and vary among cancellous bone and around the regions between cortical and cancellous bone. To reflect this heterogeneity in the FEAs, the MF software program calculates the apparent bone density, and determines the Young's modulus of each element separately [34]-[36].

Two sets of spinal models of healthy and osteoporotic subjects were developed. Written informed consent, permission and cooperation of 29-year-old Japanese male healthy subject (78kg weight and 176cm height) and 72-year-old Japanese female osteoporotic patient (60kg weight and 163cm height) were obtained.

To create the FE models, the CT scan images of the patient'svertebrae from twelfth thoracic vertebra (T12) to the second lumbar vertebra (L2) of the subjects were transferred to a PC. The 3D image-based FE models were then constructed based on the extracted bone edges of the region of interests (ROI) around the outer region of the cortical bone on the CT scan imagesto obtain the anatomical structure of the spinal bone. Because of the structural complexity of the vertebrae, we adopted tetrahedral elements instead of cubic elements to represent the smooth surface of the spinal bone [37]. The trabecular and the inner portion of the cortical bone were modelled using 3mm linear tetrahedral elements. Triangular shell elements with the thickness 0.4mm were also adopted on the outer surface of the cortex to represent the thin cortical shell. On average, there were 804,467 and 790,408 tetrahedral solid elements and 105,252 and 103,844 triangular shell elements for healthy spinal model and osteoporotic model, respectively.

B. Calculation of the Bone Materials Properties of Spine FE Models

The bonedensity of an element was determined from the average CTvalue (HU: Hounsfield Unit) of 17 points (Fig. 2), which werecomposed of the centre point and four points distributed onfour lines connecting the centre point to each apex of thetetrahedral element [38]. The bone density of each FEwas computed by the relationship as follows [33]:

$$\rho = \begin{cases} 0.0 & (HU \le 1) \\ (0.945 \times HU + 1.347) \times 1010^{-3} & (HU > 1) \end{cases} \quad (1)$$

where $\rho(g/cm^3)$ denotes the bone density as described in the Fig. 2. Poisson's ratio was set to constant value of 0.4 by reference to Keyak *et al*. [34], Reilly and Burstein [39] and Van Buskirk and Ashman [40]. The elastic modulus of each finite element was determined based on the relationship between Young's modulus, $E(MPa)$ and the bone density provided by Keyak *et al*. [20] as follows:

$$E = \begin{cases} 0.01\ (\rho = 0.0) \\ 33900\rho^2\ (0.0 < \rho \le 0.27 \\ 5307\rho + 469\ (0.27 < \rho < 0.6) \\ 10200\rho^{2.01} (0.6 \le \rho) \end{cases} \quad (2)$$

Since Young's modulus is defined by individual elements one by one as described above, the heterogeneity of the Young's modulus in the femoral bone can be directly reflected in the FE models.

Meanwhile the yield stress $\sigma(MPa)$ of the models was calculated from the bone density as proposed by Keyak *et al.* [34] as follows:

$$\sigma = \begin{cases} 1.0 \times 10^{20} (\rho \leq 0.2) \\ 137\rho^{1.88} (0.2 < \rho \leq 0.317) \\ 114\rho^{1.72} (0.317 \leq \rho) \end{cases} \quad (3)$$

The final FE models consisted of T12-L2 vertebrae, intervertebral discs and facet joints (Fig. 3). The material properties of the intervertebral disc and facet joint are listed in Table I.

TABLE I: MATERIALS PROPERTIES OF FE MODELS

Material	Young's Modulus (MPa)	Poisson's Ratio
Intervertebral Disc	8.4	0.45
Facet Joint	11	0.2

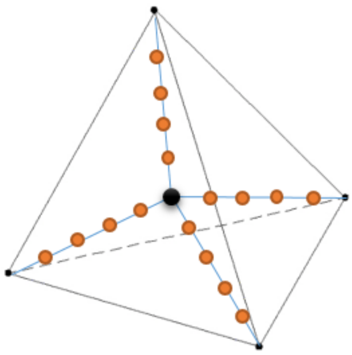

Fig. 2. Determination of bone density in tetrahedral elements. The bone density of an element was determined from the average number of Hounsfield units obtained for a total 17 points that were composed of the center point (solid circle) and 4 points (open circles) evenly distributed on 4 lines connecting the center point to each of the apexes of the tetrahedral element [7].

C. Analysis

The FE models were loaded with compressive force of 1000N and four rotational/moment loadings on the superior surface of T12 intervertebral disc to stimulate the four physiological motions/functions of the spine which are representingthe movement of flexion, extension, lateral bending and axial rotation.The inferior side of the L2 intervertebral disc was rigidly fixed. The loading details are listed and depicted in Table II and Fig. 1, respectively.

TABLE II: LOADING CONDITIONS [41]

Motion	Flexion	Extension	Lateral Bending	Axial Rotation
Loading(N.m)	4.2	1.0	2.6	3.4

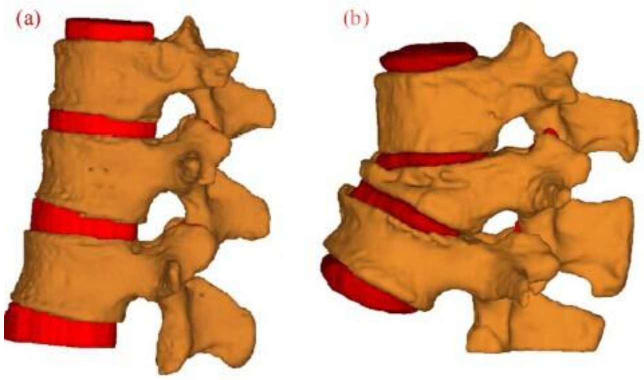

Fig. 3. Three-dimensional FE model: (a) healthy and (b) osteoporotic subject.

The biomechanical effects of osteoporotic bone model was analyzed and compared to the healthy bone model. Drucker-Prager stress and minimum principle stress distributions on the vertebrae were evaluated.

III. RESULTS AND DISCUSSIONS

The load transfer properties (stress and strain) were significantly different in healthy vertebrae as compared to osteoporotic vertebrae in five different vertebrae physiological motions (Fig. 4 and Fig. 5). In general, the osteoporotic subject tended to produce higher stress and strain in comparison to the healthy subject in all physiological movements.

The maximum Drucker-Prager stresses (Fig. 6) for the healthy subject were 1.48MPa, 4.00MPa, 1.00MPa, 2.28MPa and 1.47MPa for compression, flexion, lateral bending and axial rotation, respectively. Meanwhile, the Drucker-Prager stresses for the osteoporotic subjects were 8.19MPa, 6.60MPa, 1.50MPa, 3.56MPa and 6.15MPa for compression, flexion, lateral bending and axial rotation, respectively. The largest relative difference (Fig. 8) was found in compression activity (453%). It was then followed by axial rotation (334%), flexion (65%), lateral bending (56%) and extension (50%) activities.

The minimum principle strains (Fig. 7) for the healthy subject were -4000 µstrain, -2000 µstrain, -400 µstrain, -1000 µstrain and -700 µstrain for compression, flexion, lateral bending and axial rotation, respectively. Applying the same distribution pattern as the drucker-prager stress distributions, the osteoporotic subject owning relatively higher minimum principle strain than the healthy subject. Topping the list (Fig. 8) was axial rotation (543%), followed by compression (350%), flexion (200%), extension (150%), and end up the list with lateral bending vertebral motion (100%).

In order to evaluate the stress distribution within and between the vertebral bodies, fifteen points (five points for each vertebra) were selected to extract the average Drucker-Prager stress. This point represented a square plate that could measure the average stress distribution distributed uniformly throughout its square volume. The plate was placed perpendicular to the vertebral endplates. The distance between each of the plate was set to 5mm. The result of the average Drucker-Prager stress distributions are shown in Fig. 9.

The results showed that the greatest drucker-prager stress for both subjects was found during compression. For the osteoporotic subject this stress was substantially higher under relatively similar level of compressive loading, approximately 4.53 times higher for the osteoporotic than the healthy subject. It is also important to note that, the least relative stress difference was 50% under similar extensive loading with the osteoporotic subject exhibits higher stress than the healthy subject. The high degree of these stresses were then correlated with high degree of principle strain or in other words the drucker-prager stress values were directly proportional to the principle strain values and most of the time the strains were concentrated in the middle of the trabecular region for each of the vertebrae.

Based on this result, firstly, we could rank the osteoporotic vertebral physiological motions with high vulnerability to bone failure according to the stress and strain patterns, with compression activity topping the lists, followed by extension, axial rotation, lateral bending and extension. Secondly, for the osteoporotic patient, the most affected part is the trabecular region rather than the cortical region by considering the highest strain and strain generated at that region. These discoveriessuggest that patients with osteoporosis should be cautious of fracture risk even during daily activities [1], [3]. These phenomena could be associated with the structural change and low bone mass of the osteoporotic vertebrae which highly affected the strength and stiffness of the bone [6]. The higher stresses generated also could be relate to an agefactor which is in agreement with the previous study [1]. In that particular study, it had shown that the elderly population tended to produce higher von-Mises stress than the middle age and young age population by 50% and 120%, respectively.The loss of bone strength and structural change due to aging effect is mainly characterized by a decrease in bone volume fraction, a shift from plate-like trabeculaeto more rod-like microstructures, as well as a decrease in connectivity density combined with anincrease in orientation of trabeculae along the main axis of loading[42] - [44], which can be partiallyexplained by adaptive remodeling [45].

Fig. 9 shows the average stress distributions within and between the vertebral bodies spanning from the superior endplate of the T12 to the inferior endplate of the L2. Based on this figure we can see that the stress distribution pattern for the healthy patient was distributed more evenly than the osteoporotic subject in all vertebral physiological motions. In general, when the magnitude of the stress is omitted, the average stresses exhibited more or less similar stress pattern. Therefore, based on this observation we can conclude that, the stress acting on the vertebral body for the osteoporotic subject is almost similar to the healthy subject with relatively higher stress level. Conversely, when the magnitude is considered, the stress difference in the two adjacent points in the vertebral body become more significant and this correlation has significantly contribute to uneven stress distribution within the vertebral body.

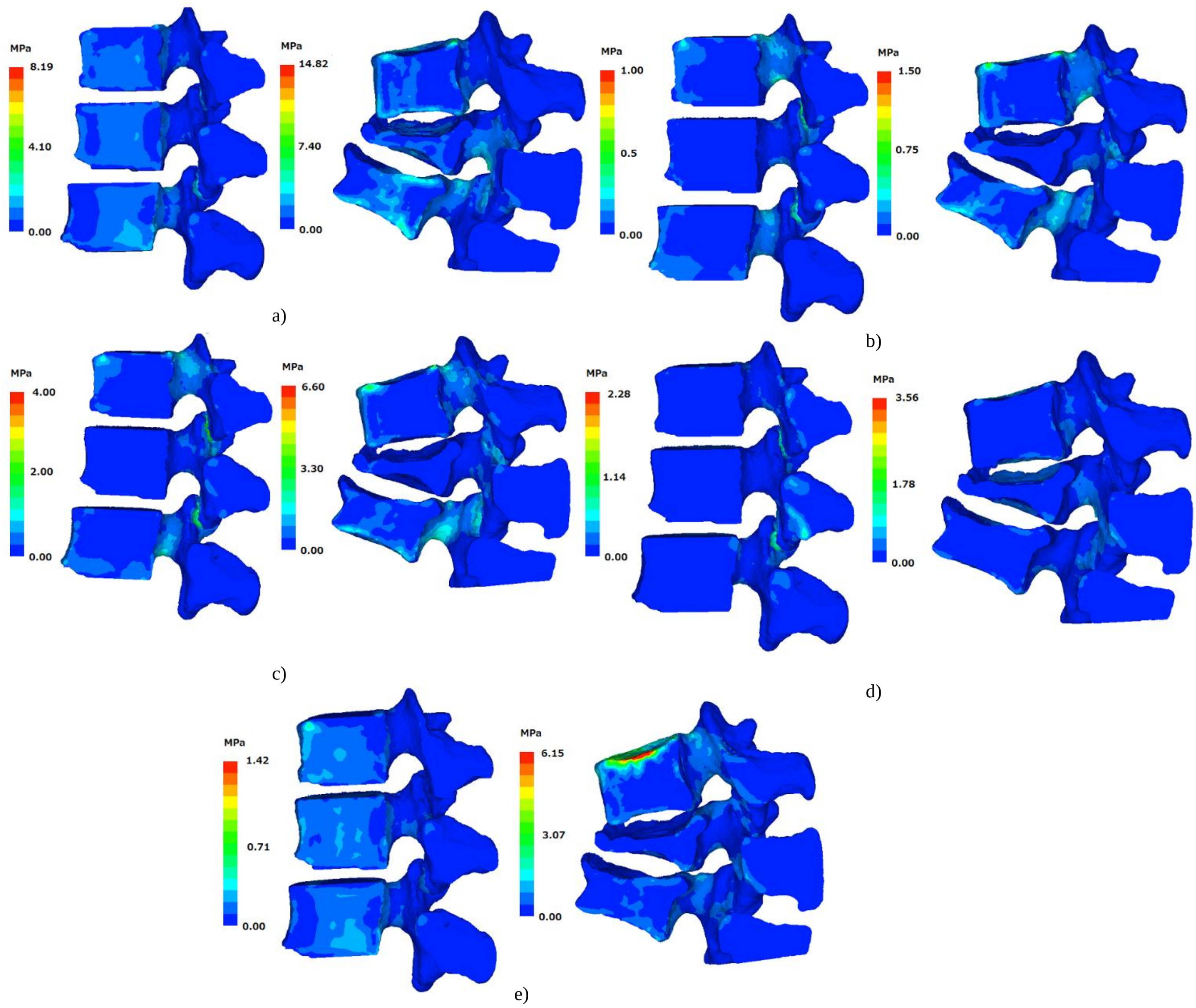

Fig. 4. The cross-sectional view of drucker-prager stress distribution on the vertebral body in five different vertebral physiological motions: (a) Compression (b) Extension (c) Flexion (d) Lateral bending and (e) Axial rotation.

In mechanical point of view, uneven stress distribution is recognized as one of the most significant contributor to the vertebral failure in osteoporosis. Based on the previous research, this typical phenomenon is related to an altered trabecular micro-architecture with less transverse trabeculae exhibits by osteoporotic vertebral [8], [9]. This altered structural architecture resulting in uneven stress distribution, with some trabeculae virtually overloaded and others unloaded [2]. In addition, this characteristic also related to the less bone for load distribution which is likely to increase trabeculae local stress and finally increasing the risk of bone fracture. Moreover, unloaded bone region will become less dense and weaker and finally fractures will be initiated on that region.Less bone for load distribution depicts the deterioration level of reduced bone mass as characterized by osteoporosis [42] and this phenomenon is generally known as osteopenia.

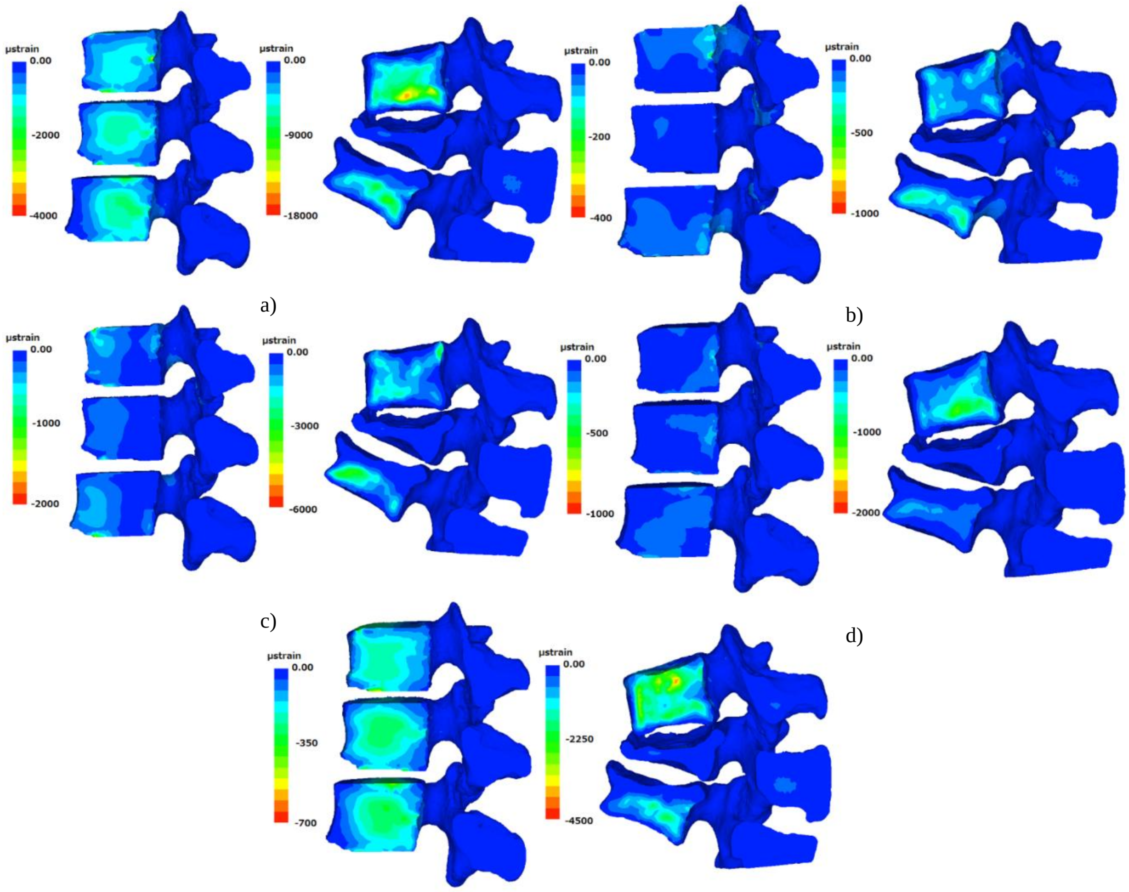

Fig. 5. The cross-sectional view of minimum principle strain distribution on the vertebral body in five different vertebral physiological motions: a) Compression, b) Extension, c) Flexion, d) Lateral bending, e) Axial rotation.

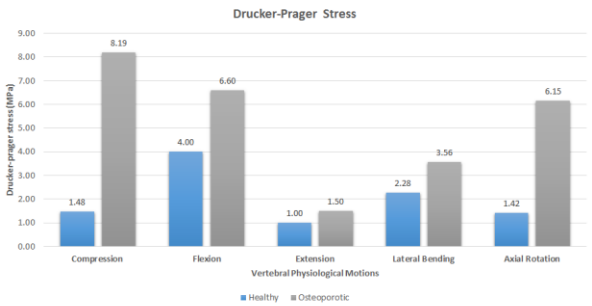

Fig. 6. Drucker-Prager stress on the vertebral body for healthy and osteoporotic subject.

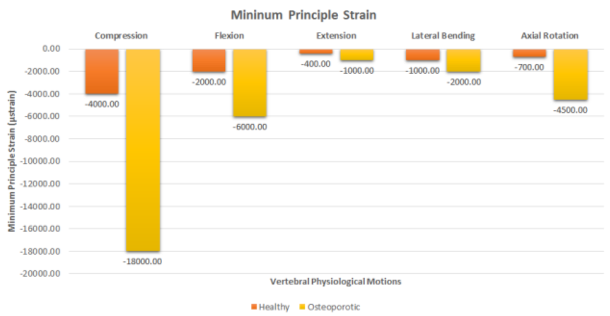

Fig. 7. Minimum principle strain on the vertebral body for healthy and osteoporotic subject.

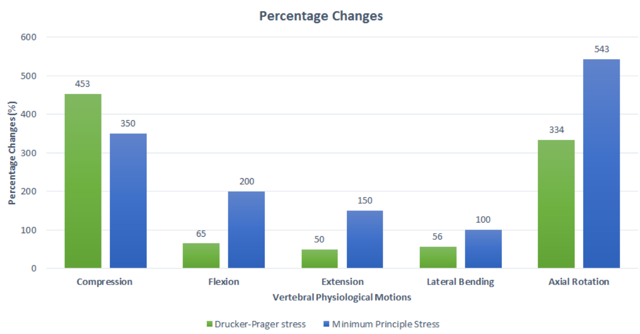

Fig. 8. Drucker-prager stress and minimum principle strain percentage changes between healthy and osteoporotic subject.

Based on this findings, the load transfer pattern in osteoporotic patient with L1 VCF could be recognized by its tendency to generate higher stress and strain associated with uneven stress distribution within and between the vertebral bodies. These characteristics can be used as an indicator to an existence of VCF. Early detection and prevention of VCF is very crucial, the occurrence of VCF increases the chances of additional fractures by 500% [46]. It was also reported that only one third of the osteoporotic vertebral fractures were actually detected, as a large portion were thought to be asymptomatic [47], [48]. Therefore, the significance of this study lies in the importance of characterizing the load transfer pattern in osteoporotic patient to be able to detect abnormal stress and strain pattern.

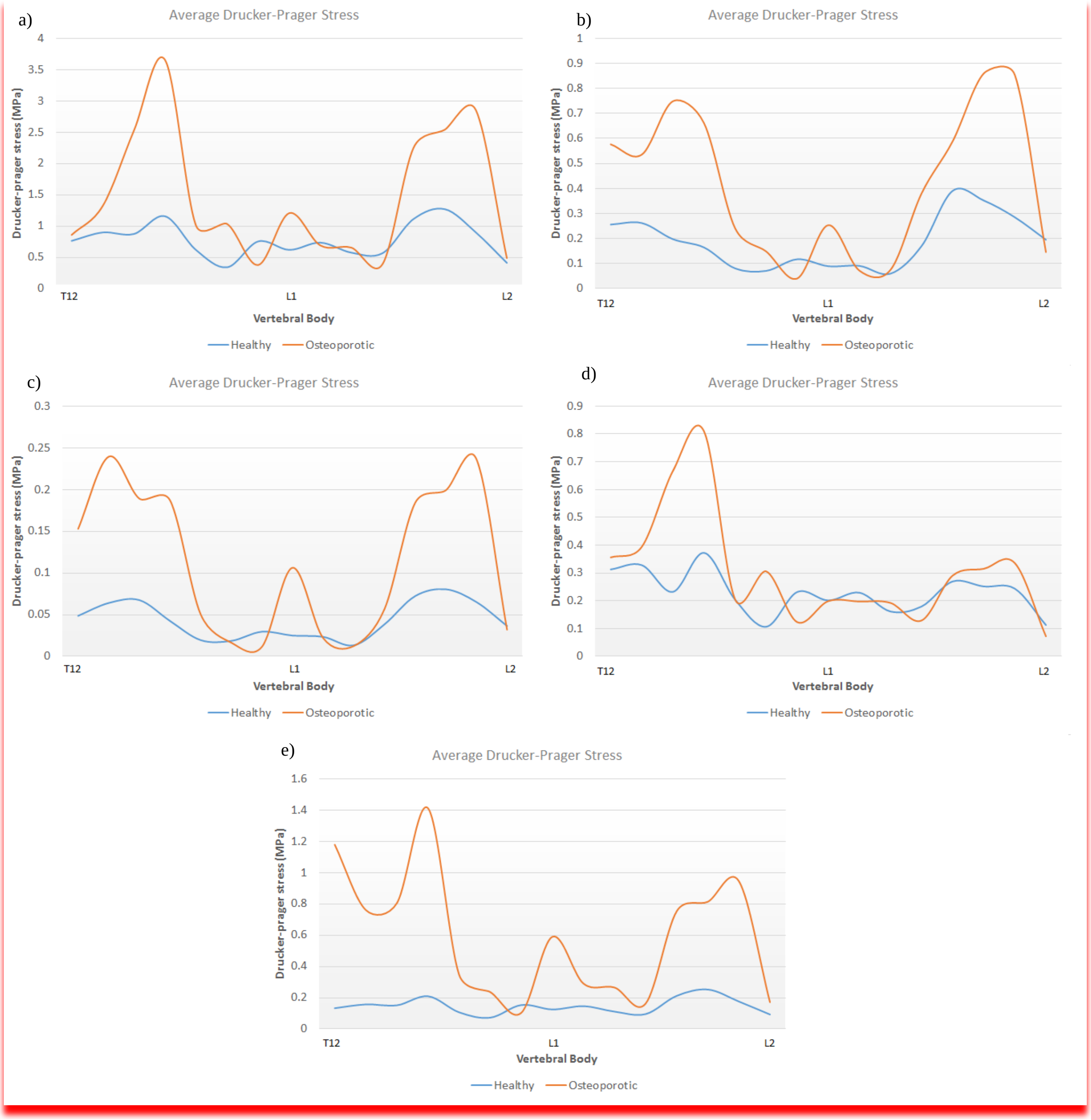

Fig. 9. Average drucker-prager stress distributions in the vertebral body for healthy and osteoporotic subject measured from superior endplate of T12 to inferior endplate of L2: (a) Compression (b) Flexion (c) Extension (d) Lateral bending and (e) Axial rotation.

IV. Conclusions

Osteoporotic vertebrae model with L1 vertebral compression fracture (VCF) has significantly affecting the load transfer pattern (stress and strain) distributions within the vertebral body. By utilizing the stress and strain distributions of the healthy subject as a comparison tools, it seems that the osteoporotic subject exhibits extremely higher stresses and strains than the healthy subject under five basic vertebral physiological motions. To make this situation even worse, this condition was accompanied by uneven stress distribution within and between the vertebral bodies. Therefore, it is strongly suggest that, for the osteoporotic subject the risk of vertebral fracture can occur at any time even with daily living activities. It was also has been proven that this phenomena are well correlated with the deterioration of bone structural strength and reduced bone mass as characterized by osteoporosis.

In the future, it is believed that the present work could be expended to evaluate the vertebral fracture risks evaluation by deploying nonlinear fracture analysis with additional loading conditions such as the combination of axial and bending loads as typically occurred in vertebral compression fractures. Therefore, the region of predicted fracture sites and its behavior can be further investigated and analyzed.

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