Biomechanical Alteration of Stress and Strain Distribution Associated with Vertebral Fracture

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ABSTRACT

The phenomenon of recurrent fractures at the adjacent level of a fractured vertebra is becoming a major concern amongst medical practitioners. To date, the underlying cause of this phenomenon is still elusive; therefore, a further investigation is in dire need in order to achieve satisfactory clinical outcomes in the future. In the present study, an image based finite element analysis (FEA) was used to investigate the biomechanical alterations of spine that have been diagnosed with first lumbar (L1) vertebral compression fracture as compared to a healthy spine. The FEA assessment was made based on the model's stress and strain distributions. A complimentary examination of bone density distribution and kyphotic deformity angle of the model would give further details on the underlying cause of this phenomenon. The results showed that the vertebral fracture model tends to produce higher stresses and strains generation in comparison to the healthy vertebral model, especially at the adjacent level of the fractured vertebra. These conditions were highly correlated to the bad quality of the bone strength due to osteoporosis, and the kyphotic structural of the fractured vertebral model. The combination of these two elements has put the structural integrity of the vertebrae at the stake of bone fracturing even under the influence of daily living activity.

Keywords: *Recurrent Fracture, Finite Element Analysis, Vertebral Compression Fractures, Osteoporosis, Spine.*

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Introduction

Finite element analysis (FEA) has been widely implemented in the field of biomechanics over the last few decades. This methodology has been utilized in many clinical applications and is gaining popularity due to the complexity of in-vitro and in-vivo experiments, which makes the FEA approach more practical and able to give more promising results. Besides, this method can also reduce costs and danger of other testing procedures and allow one to achieve certain individualization. A recent study showed that the correlation of the bone strength could be better achieved by adopting FEA than Dual X-Ray Absorptiometry (DXA) approach [1]. Another study revealed that the bone density measurements by DXA could give inaccurate diagnosis due to the possibility of an overlap between the bone density measurement in people with and without osteoporotic bone fractures [2]. Last but not least, this numerical approach has also shown a good agreement with experimental data obtained from in vitro studies of the mechanical behavior of normal and injured spine [3][4].

In general, osteoporosis is the most common disease affecting both men and women, and it is becoming increasingly prevalent in aging society. In Japan, there are more than 10 million osteoporosis patients [5], and in the future this number is expected to increase in tandem to the continuously increasing pattern of the total life expectancy in Japan. 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% and only 28% after one and five year(s) of initial bone fracture was detected, respectively [6]. Moreover, the probability of secondary vertebral bone fracture was reported to be 500% following primary bone fracture, which increased the chances of multiple levels of vertebral bone fracturing with the accompanying effects of high morbidity and mortality [7]. In this context, we believed that FEA is the most reliable tool to diagnose and overcome those problems based on its ability to address complex clinical and mechanical osteoporosis-related issues such as drug therapy and delivery [8] [9], vertebral implant assessment [10] [11], fracture risk evaluation [12-14], etc.

The objective of this study is to quantitatively analyze the stress and strain profiles of vertebral healthy and fracture models. In addition, the phenomenon of subsequent fractures at the adjacent level of the fractured vertebra is also investigated by evaluating the bone density distribution and kyphotic deformity angle of the fractured vertebra. By doing so, this might give an insight as to what extent a vertebral compression fracture (VCF) could induce a formation of secondary fractures at the adjacent level of the fractured vertebra.

Materials and Methods

Finite element modelling

Three-dimensional (3D) finite element (FE) models of T12-L2 were constructed in MECHANICAL FINDERTM software (Research Center of Computational Mechanics Co. Ltd. Japan). The image data of T12-L2 were obtained from CT scan images of thoracolumbar spines of healthy and osteoporotic patients (Department of Orthopedic Surgery, Juntendo University, Japan). The healthy subject was 29-year-old male subject without any previous medical history. The osteoporotic was 86-year-old female subject that has been diagnosed with fist lumbar VCFs. The FE models were developed from the extracted bone edges of the region of interests (ROIs) to obtain the anatomical structures of the spine. They were modelled with 1 mm linear solid tetrahedral and shell triangular elements.

The bone material properties were assigned to be non-linear, inhomogeneous and isotropic. The heterogeneity of the bone material properties (the Young's modulus and yield strength) was obtained through the relationship as reported by Keyak et al. [15] based on the Hounsfield Unit (HU) values obtained from the CT scan images. The Poisson's ratio for each of the element was set to a constant value of 0.4 [15]. The material properties for each of the shell triangular element were set to be equivalent to that of the adjacent tetrahedral element located underneath the shell element. The intervertebral disc and facet joint were assumed to be linear, homogeneous and isotropic. The Young's modulus and Poisson's ratio for the intervertebral disc were set at 8.4 MPa and 0.45 respectively, while the facet joints were set at 11 MPa and 0.2, respectively [11, 16].

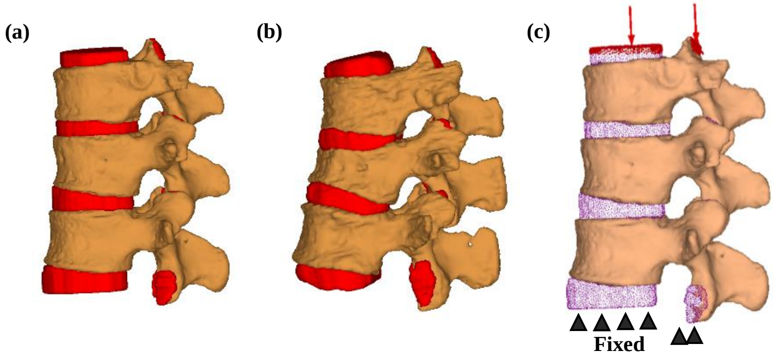


Figure 1: FE model of (a) vertebral healthy and (b) fracture spine, and (c) their corresponding load and boundary conditions.

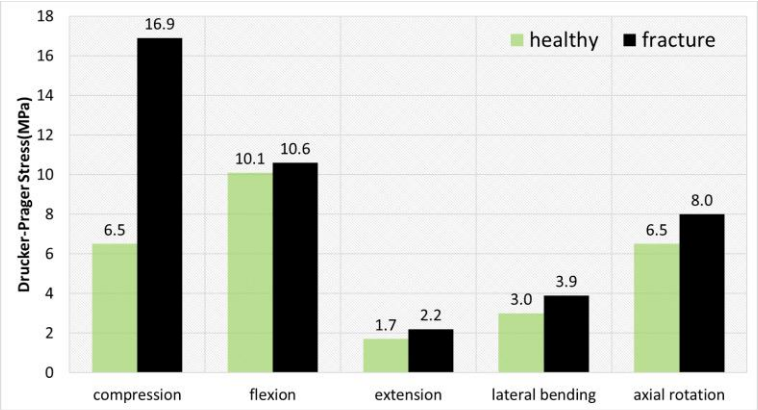
The models (Figure 1(a) and (b)) were loaded with compression (1 kN), flexion (4.2 Nm), extension (1 Nm), lateral bending (2.6 Nm), and axial rotation (3.4 Nm), based on the spinal three column load bearing concept with the proportion of 85% and 15% of the load being applied to the anterior and posterior column of the vertebra, respectively [17,18]. The loads were applied to the superior surface of the intervertebral disc and facet joints located immediately on the top of T12, while the inferior surface of the intervertebral disc and facet joints located immediately below L1 was fixed in all directions (Figure 1(c)). In order to inhibit any relative movement during simulation, the facet joint and intervertebral discs were securely attached to the vertebral body through perfectly bonded interface conditions.

Result and Discussion

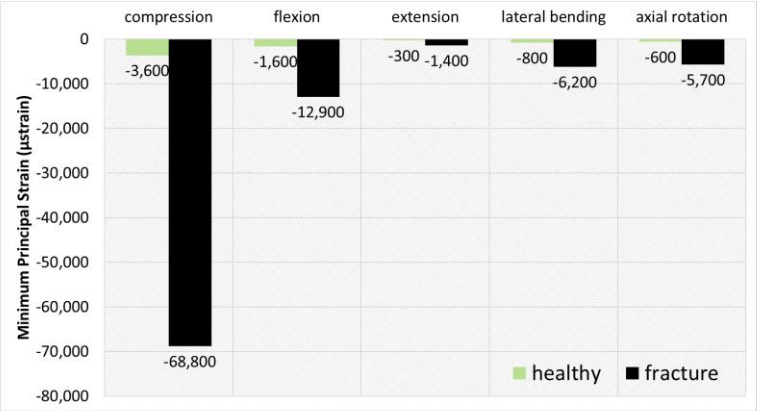
Vertebral bone strength

Figure 2 shows the Drucker-Prager Stress, Minimum Principal Strain and Maximum Principal Stress of the vertebral healthy and fracture models under the application of spine physiological motions. These values were recorded for the most maximal and minimal of the stresses and strains produced by each of the vertebral model. Obviously, for the vertebral fracture model, those values were detected at the adjacent level of the fractured vertebra. The maximal Drucker-Prager stress for the healthy vertebral model in compression, flexion, lateral bending and axial rotation was 6.5, 10.1, 1.7, 3.0, and 6.5 MPa, while the fractured vertebral model was 16.9, 10.6, 2.2, 3.9 and 8.0 MPa, respectively. The minimal Minimum Principal Strain for the healthy vertebral model of the dedicated spine motions was 3,600, 1,600, 300, 800, and 600 microstrains, while the fractured vertebral model was 68,800, 12,900, 1,400, 6,200 and 5,700 microstrains, respectively. The maximal Maximum Principal Stress for the healthy vertebral model of the assigned spine motions was 4.1, 12.5, 1.8, 3.2 and 7.8 MPa, while the fractured vertebral model was 20.0, 12.9, 2.8, 5.2 and 11.1 MPa, respectively. Noticeably, in all spine motions, the vertebral fracture model tends to produce greater stress and strain generation than the healthy vertebral model.

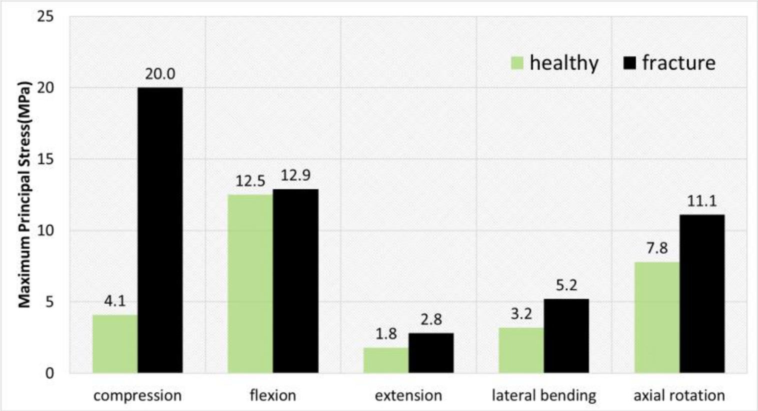
Theoretically, the values indicated by the maximal Drucker-Prager stress, minimal Minimum Principal Strain and maximal Maximum Principal Stress were reflected the model's yield strength, crushing strength and critical strength, respectively. Moreover, by taking into consideration the dynamic factor that can influence a steady state analysis of up to a five-fold increased impact [19], the safety factor for each of the vertebral models can be further evaluated. Based on the previous report, the average normal people's bone's yield strength, crushing strength and critical strength is 83 MPa [20], -27,900 microstrains [21] and 66 MPa [22], respectively. Therefore, in order to avoid any untoward incidents, the stress and strain should not exceed one-fifth of the average normal people's bone strength criteria as mentioned earlier. Apparently, in almost all of the spine motions, the structural integrity of the vertebral fracture model fell within the tolerable range of the required safety factor. However, in compression activity, the minimal Minimum Principal Strain was produced beyond the permitted level which was 8.5 times more negative than the normal people's bone's crushing strength. According to the failure criteria of human bone as described by Bessho et al. [22], this result did not necessarily mean that the vertebral fracture model was at the stake of structural fault since its maximal Drucker-Prager stress did not exceed the critical requirement of the normal people's bone's yield strength.



(a)



(b)



(c)

Figure 2: Vertebral bone strength based on the (a) Drucker-Prager, (b) Minimum Principal Strain and (c) Maximum Principal Stress evaluation.

Bone mineral density

Figure 3 shows the average bone mineral density taken from 30 regions of interest, which were represented by square plates placed in perpendicular to the inferior and superior endplate of a vertebral body. From this distribution we found that the average bone mineral density of the vertebral healthy and fracture model was 232 mg/cm3 and 188 mg/cm3, respectively. According to Kurtz et al. [23], the average apparent bone mineral density for a healthy individual is 100 mg/cm3. Therefore, the region that exhibited the average bone mineral density lower than this value is expected to have a greater risk of vertebral bone failure. Thus, based on this consideration, we concluded that the vertebral fracture model was considered to have been afflicted with severe osteoporosis especially at T12 and L2 vertebrae. Moreover, the average bone density of the vertebral healthy model was far evenly distributed than the vertebral fracture model. However, the event of higher bone density distributions at the vertebral fracture model was elusive and intriguing which need to be further corroborated in the future.

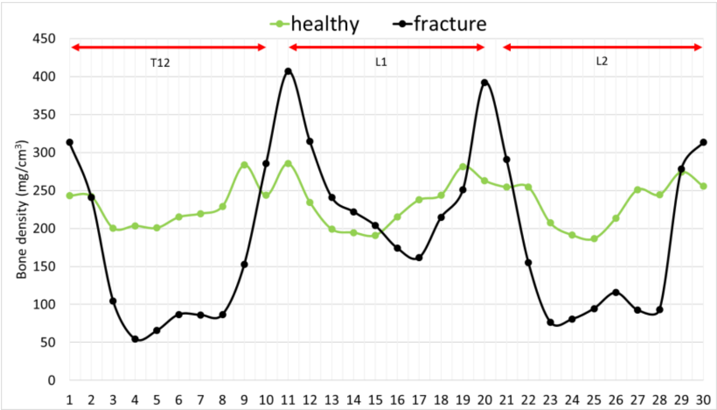


Figure 3: Average bone density distribution of vertebral healthy and fracture models.

Kyphotic angle

The kyphotic angle for each of the vertebral model was computed according to the Cobb Measuring Technique [24]. The kyphotic angle for the vertebral healthy and fracture models was 6° and 16°, respectively. Literally, a higher degree of kyphotic angle will produce a higher generation of moment and force at the anterior portion of the affected and its neighbouring vertebrae. This is because the body's centre of gravity (COG) will be shifted away from the vertebral body and subsequently increase the chances of subsequent fractures at the affected and its adjacent vertebrae.

Osteoporosis and kyphotic deformity

Based on the results, we found that the increased risks of subsequent fractures at the adjacent of the fractured vertebrae were highly related to the osteoporosis severity and kyphotic deformity of the spine. These two complimentary elements have contributed to a greater biomechanical disadvantages to the affected spine. The higher stress and strain generation at the adjacent levels of the fractured vertebra could be associated to the decrease of the vertebral bone stiffness (especially at the trabecular bone regions) and subsequently altered the load transmission mechanisms of the spine. As a result, when the fractured vertebra becomes relatively stiffer than its neighboring vertebrae, the load from the fractured vertebra will be transmitted to the adjacent vertebrae through the centrum rather through the cortex, which in turn put the structural integrity of the adjacent vertebrae at the stake of vertebral bone failures. This phenomenon is in agreement to the studies conducted by Kim et al. [25] and Boroud et al. [26]. In those studies, they found that the existence of osteoporosis has significantly increased the stress generated on the fractured vertebral body from 50% to 120% and subsequently increased the pressure on its adjacent intervertebral discs by up to 19%. In addition, the fractured location, which was situated at the most critical inflection point of the spine (L1), has produced a vast additional downward pressure on the spine especially in bending forward movement [27]. Moreover, a statistical study showed that 58% of the recurrent fractures occurred next to the index fracture (irrespective on the fractured location) [28], and based on this criteria, the probability of a subsequent fracturing at the adjacent levels of the fractured vertebra is expected to be higher if the factor of the fractured location is taken into consideration.

The osteoporosis progression towards the end would cause the vertebral bone to lose its ability to sufficiently absorb the energy afflicted on them and finally make them less capable of sustaining damage in withstanding great magnitudes of any input loads [29]. Generally, the clinical significance of osteoporotic lies in its high vulnerability and susceptibility to bone fractures [30], and the most prevalent fracture site occurs in the spine [27] particularly in elderly people [30]. It is characterized by low bone mass and micro-architectural deterioration of bone tissue [31]. In mechanical viewpoint, low elastic modulus and uneven stress and strain distributions are the main indicators to the existence of osteoporosis [32]. The cause of bone fracturing can either be associated to a traumatic or non-traumatic event (daily living activities) [27]. For the non-traumatic event, this phenomenon can be correlated to a less bone for a load distribution [33-34], uneven load distribution within a vertebral body, and compromised vertebral bone structural integrity [6]. For such conditions, they might increase the local tissue stress and subsequently put the structural integrity of the bone at the stake of bone failures.

Conclusion

The osteoporotic vertebral model with L1 VCFs has significantly altered the load transfer mechanisms of the spine with the accompanying effects of higher vulnerability of subsequent fractures at the fractured and its adjacent vertebrae. The underlying cause of this phenomenon is conclusively attributed to the osteoporosis severity and kyphotic structural of the fractured vertebra. Therefore, for an osteoporotic individual, the risk of vertebral fractures can occur at any times even when performing daily living activities. Thus, early detection is necessary to avoid any untoward incidents to occur in the future.

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