

A Finite Element Approach to Predict the Post-treatment Evolution of an Orthopedically Treated Vertebral Body Fracture

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Abstract—Our research employs FEM to simulate the evolution of an orthopedically treated vertebral fracture and compares the result against the real healing process observed through CT and X-ray investigations. Applying FEM to the simulation of vertebral fracture healing is an original endeavor in studying the evolution of the fracture stability and represents a theoretical pretest for applying the FEM in the prediction of fracture healing progress.

Index Terms—lumbar fracture, finite element method (fem), computer tomography (CT).

I. INTRODUCTION

The Finite Element Method (FEM) represents a widely used research tool in the simulation of the spinal mechanical behaviour under given specific stresses, loads and movements. [1]-[3] To this end, various elements and processes have been simulated, such as the vertebral bodies' reaction under compression forces; the resistance of vertebroplasty to compression loads; the mechanical behaviour of various spinal segments in flexion, extension, rotation; the pressure on the intervertebral disc, etc. [4]-[6].

The process of virtual modeling / model development generally involves several stages:

- The reconstruction of the spinal element geometry, which, along the last decade, has been most often achieved employing CT images
- Adjoining the soft tissues (discs, ligaments)
- Setting the mechanical properties of each anatomical structure involved
- Model testing and validation, allowing for comparative analysis between the simulation product and the results observed *in vivo* or *in vitro*.

This research intends to simulate, by the means of MEF, the evolution of a vertebral fracture orthopedically treated, whose actual progress is known. The simulation regards only the first 4 weeks of the healing process.

II. MATERIALS AND METHOD

A. Case Presentation

The patient is a 32 year old male, of 1.6 m tall and of 52 kg weight who was diagnosed with fractures at L1 vertebral level after plunging from about 4 m height.

Following the ER protocol, the spinal fracture was classified based on radiologic and CT investigations. According to Load Scherring Scale (LSS) [7.] and Magerl AO [8] classification the fracture was considered stable, and the patient was treated conservatively and was immobilized for a period of 8-12 weeks with a 3 point hyperextension orthosis. The orthostatic posture was allowed to the limit of pain three days after trauma. Anti-inflammatory, anticoagulant and pain medication was prescribed to complete recovery. The X-ray image one day before discharge displayed no modification as compared to the admission image. The next follow-up investigation occurred one month after the trauma, the patient being one week late for the required medical control.

B. 3D Model Description

Based on the CT scan sections, the spinal segment T12-L2 was reconstructed, employing MIMICS, a software with medical applicability, allowing the 3D reconstruction using CT.

The model was subsequently transferred in a CAD software where the contours of the vertebral body were manually adjusted, and the volumes of the intervertebral discs and of the artefacts were eliminated.

In a first stage, the bone structures are constructed, closely following, with maximum possible accuracy, the actual alignment and anatomical proportions. (Fig. 1).

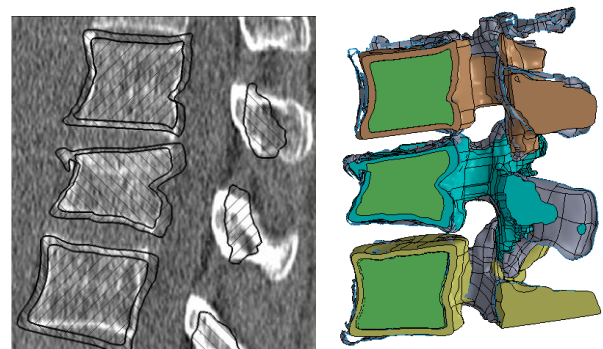


Fig. 1. Overlapping CAD model sections and CT scans.

Manuscript received October 1, 2017; revised October 30, 2017.

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In our study, the alignment accuracy was verified by super-posing the obtained CAD model section (obtained with SolidWorks software) with the CT section. Both sections were realized in the same plan. After correctly aligning the vertebrae, the spongy matter of the vertebral body was separated from the compact bone tissue. As the L1 vertebra is our main research object, it was modeled with maximum accuracy. The wall of the vertebral body was 2 mm thick, the superior vertebral plate was 0.6 mm thick and the inferior one was of about 1 mm. The fracture lines at the superior vertebral plate level and walls were also reconstructed corresponding to the CT scan.

The spongy tissue of the L1 vertebra was delimited in two volumes by a horizontal plan situated at the lower limit of the compacted tissue visible both in the CT scan and MIMICS sections. These volumes correspond to the damaged and intact parts of the vertebra (Fig. 2). The next step involved modeling the soft anatomical structures (Fig. 2): the anterior longitudinal ligament, posterior longitudinal ligament, the fibrous ring with its four layers and the pulposus nucleus as an amorphous mass inside the ring. For image simplification purposes the posterior elements - ligamentum flavum, articular capsule, and the inter and supra spinous ligaments were modelled as connection lines, in accordance with the trajectories of the physiological forces.

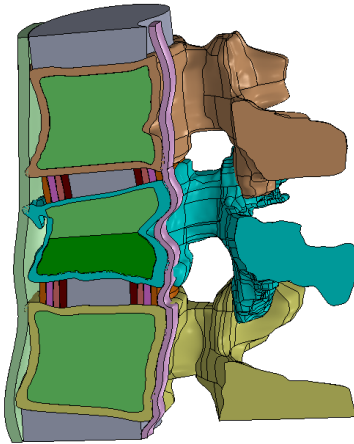


Fig. 2. Final model for analysis.

C. FEM Model Description and analysis

The model is composed of 569675 nodes and 287278 tetrahedral elements with 10 nodes each. The L2 vertebra mesh density is lower than the density of the other vertebrae but this has no negative influence over the final result. On one side, this vertebra has only a support role and on the other side, the forces implied in the studied mechanical processes do not pose any challenges over its resistance and integrity. Each anatomical element (vertebrae, ligaments and intervertebral discs) was separately meshed.

Regarding the settings of the mechanical properties of the elements in the model, due to the large dispersion of values available in the literature. [9,10,]. Ezquerro et al [11] proposed a method of data calibration consisting of the selection of *in vitro* gathered values that are the most similar to the results obtained through MEF modeling. We employed these values, as they are presented in Tables 1, 2 and 3. For best accuracy we considered the materials as being

anisotropic.

TABLE I. BONE PROPERTIES [11]

	Young Modulus [MPa]	Poisson ratio
Cortical bone	$E_{xx}=E_{yy}=11300$	$\nu_{xy}=0.484$
	$E_{zz}=22000$,	$\nu_{yz}=\nu_{xz}=0.203$
	$G_{xz}=G_{yz}=5400$	
	$G_{xy}=3800$	
Spongy bone	$E_{xx}=E_{yy}=140$	$\nu_{xy}=0.45$
	$E_{zz}=200$	$\nu_{yz}=\nu_{xz}=0.315$
	$G_{xz}=G_{yz}=G_{xy}=48.3$	
Rear elements	$E=3500$	$\nu=0.25$

TABLE II. SOFT MATERIAL PROPERTIES [11]

Ligament	Section [mm ²]	λ_{0j} [MPa]	λ_{1j} [MPa]	ε_{1j}
LLA	50	23.06	84.87	0.26
LLP	17	0	73.60	0.37
Joint capsule	70	0.15	89.37	0.37
Flavum ligament	67	0.10	79.87	0.53
Interspinous ligament	35	0.88	4.69	0.41
Supraspinal ligament	30	0.02	4.54	0.21

TABLE III: INTERVERTEBRAL DISC MATERIAL PROPERTIES

	Young Modulus [MPa]	Poisson ratio
Kernel	1	0.499
Fibrous rings	45	0.3

The linkages between the longitudinal ligaments and bone structure were defined as being very tight. Anatomical particularities were respected though, so that, at the middle of the vertebral body the link between the LLA and the cortical is loose. Also, the contact between the LLA and the intervertebral disc has been defined as very tight. The same contact was defined for the fibrous ring and superior and inferior plateaus. The facet joints were defined as contact surfaces.

The model analysis was performed using ANSYS software and consists in imitating the movements that may occur in the spinal column when the patient, though being immobilized in the hyperextension orthosis, is allowed to stand. In that case, we consider that, at the level of the fractured segment, only axial loads may occur while the flexion - extension loads are considered impossible. As the probability that the patient hadn't strictly observed the movement restrictions is considerable, was also tested to flexion and extension movements.

The normal axial load was set at $F = 350$ N and the flex torque was set to $M = 15$ Nm. The compression angle on a sagittal CT section of the fractured vertebral body measured after the accident was of 12.8° and the angle measured one month later was of 17.56° (Fig. 6).

III. RESULTS

In standing position the highest compression takes place at the posterior edge of the fibrous T12-L1 ring and at the anterior side of the fractured vertebral body L1. Red colored areas suggest compressions of about 1.5mm. Smaller compression values, of 0.5 - 1 mm may be observed both at the level of the intervertebral disc T12-L1 and at the fracture level. Bone areas with the highest forces occurring during axial loads can be found at the level of the postero-superior vertebral body area, that is the intact area of the posterior wall

and the posterior apophyseal ring that remained unaffected after the trauma. The maximum tension in this area is 136 MPa. Areas with high tension can also be found on the superior and inferior edges of the T12 vertebra pedicles as well as of T12-L1 facet joints. These areas are visible in figure no.4 a where red colored areas represent tensions between 100 and 170 MPa.

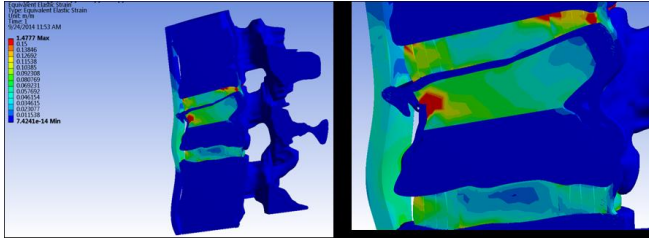
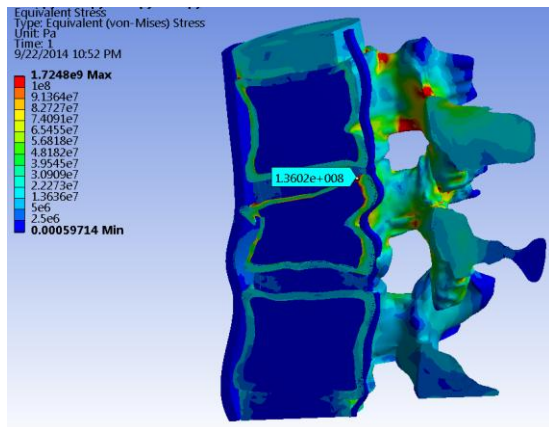
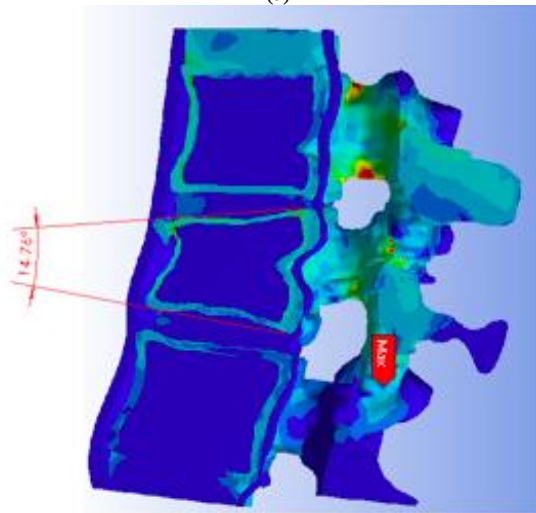


Fig. 3. Deformation distribution during axial load.



(a)



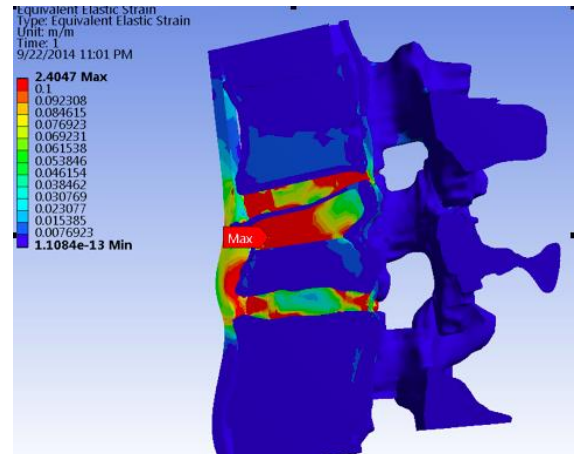
(b)

Fig. 4. Compression angle of the vertebral body after the accident (a) and after axial load (b).

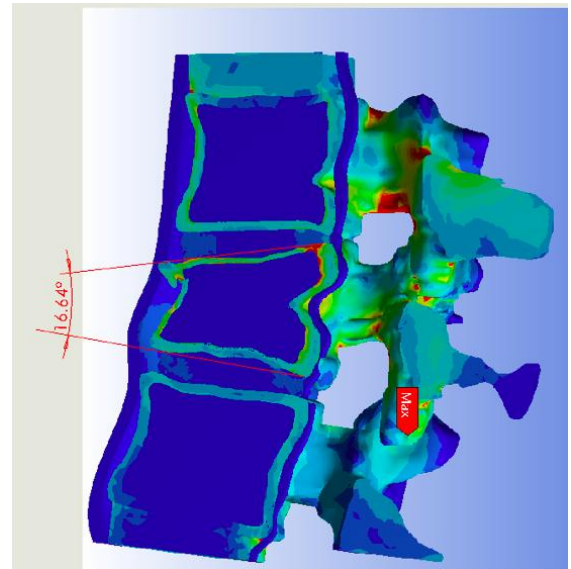
Under axial load the model presents the wedging of the vertebral body of 14.76° which represents an increase of 2.4° over the angle measured after the accident (fig. 4b).

Axial load combined with flexion movements lead to soft tissue compression of a maximum of 2.4 mm in the area of the intervertebral discs and anterior area of the L1 fractured vertebral body (the red coloured areas in figure no. 5a). Lower compression zones are visible to the posterior of the fractured area, of the pulpous nucleus and intervertebral discs, with compression values between 0.01 and 0.1 mm.

High values of tension in the bone tissue accumulate in the anterior and posterior of the apophyseal ring L1 and the zigoapophyseal joints T12-L1 and L1-L2. These values range between 100 and 170 MPa.



a



b

Fig. 5. Distribution of maximum forces during flexing associated with axial load.

The wedging angle that resulted in the simulation was 16.64° ; 0.8° (Fig. 5b) less than the one measured in the control X-ray taken one month after the accident (Fig. 6). Compression angle of the vertebral body is being indicated.

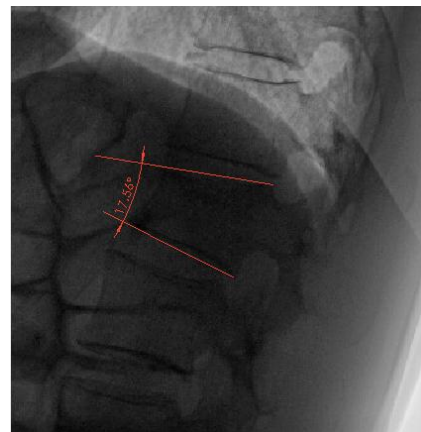


Fig. 6. Control radiography taken one month after the accident.

IV. MODEL VALIDATION DISCUSSIONS.

Two arguments plead for the validation of this model. According to the obtained results, when the fractured vertebra is subjected only to axial loads the vertebral body compresses by 2.4° . When flexion movement is added, the compression angle increases by 4.3° (16.64° in total). This angle is very close to the one measured in the control X-ray (17.5°). (fig 6)

The second validation argument is the behavior of the vertebral disc L1-L2. Biomechanical studies have shown that in physiological conditions, the pulposus nucleus of a normal disc is practically incompressible. Its role is to take over axial loads by transforming them into horizontal forces that act on the fibrous ring which protrudes not more than 1mm.

A secondary aspect that favours the validity of our model represents the stability of the assembly which keeps anatomical relations intact. All these arguments sustain the validity of the model.

Vaccaro *et al.* [12] have argued that the secondary deformation is a consequence of losing the anterior support. Studies of biomechanics show that following a fracture of the anterior and middle column, the capacity to sustain axial loads decreases by 70%. Our model suggests that, at least in the situation in which the posterior cortical is intact, axial loads alone do not lead to a significant deformation of the vertebral body. The flexion movement dramatically changes the situation by doubling the compression angle. This element becomes important also because the flexion movement implies far lower forces than the axial ones. Another element that needs discussion is the efficacy of thoracolumbar hyperextension orthosis. The model shows that there was a flexion movement in the fracture area. This suggests either that the patient did not wear the orthosis permanently or that the corset did not have the capacity to eliminate flex movement altogether. The second alternative is excluded, in our opinion the first one is the most probable.

According to the FEM model, in case of a fractured vertebral body with intact posterior wall that is subjected only to axial loads, the forces will be concentrated only on certain areas, such as the posterior part of the fibrous ring and the underlying bone tissue and also in the area of the T12-L1 zygapophyseal joints. In the long run, these forces could lead to a degenerative process in both the disc and the joints. [13]

It is evident that the deformation in the fractured area is far greater than the one in the tissues of the vertebral disc above (anterior half part of the fibrous ring and nucleus pulposus). This can be explained by the difference in elasticity between the three materials. The fractured area behaves more plastically than the components of the intervertebral disc. When all of them are compressed together, the first one will show the largest deformation.

When the force is no longer applied, the structures of the disk will return to their initial dimensions, except the fractured area. This area is considered a homogenous entity only from a bio-mechanical point of view. Actually, from a histological perspective, it is a complex mass of normal bone spongy tissue with different degrees of resorption and bone marrow, fibrous callus whose structure and biomechanical properties depend on the duration of the post fracture period.

V. CONCLUSIONS

This model may prove valuable for the study of the stability of a vertebral fracture, as an instrument of theoretic pretesting.

The flexion movement combined with axial load, as compared to simple axial load is the one that determines the compression of vertebral bodies.

These elements can induce a degenerative processes at the level of both intervertebral disc adjacent to the fracture as well as of the zygapophyseal joints of the fractured segment.

REFERENCES

- [1] Ruberte LM, Natarajan RN, Andersson GB. Influence of a single-level lumbar degenerative disc disease on the behaviour of the adjacent segments – a finite element model study. *Journal of Biomechanics*, 2009; 42(3) 341-34
- [2] B. Khuyagbaatar, K. Kim, and Y. H. Kim, "Effect of bone fragment impact velocity on biomechanical parameters related to spinal cord injury: A finite element study," *Journal of Biomechanics*, 2014, vol. 47, pp. 2820–2825
- [3] S. K.Chung, Y. E. Kim, and K. C. Wang, "Biomechanical effect of constraint in lumbar total disc replacement a study with finite element analysis," *Spine*, 2009, vol. 34, no. 12, pp. 1281-1286
- [4] A. A. Jaffar, S. J. Abass, and S. S. M. Mukhtar, 2010, "The stress distribution on the zygapophyseal," *Joint of Lumbar Vertebra by Ansys Program Al-Khwarizmi Engineering Journal*, vol. 6, no. 3, pp. 85 – 91, 2010.
- [5] D. D. Jebaseelan, C. Jebaraj, N. Yoganandan, S. Rajasekaran, and J. Yerramshetty, "Biomechanical responses due to discitis infection of a juvenile thoracolumbar spine using finite element modeling," *Medical Engineering & Physics*, 2014, vol. 36, pp. 938–943.
- [6] R. Singh, P. D. Lee, T. C. Lindley, C. Kohlhauser, C. Hellmich, M. Bram, T. Imwinkelried, and R. J. Dashwood, "Characterization of the deformation behavior of intermediate porosity interconnected Ti foams using micro-computed tomography and direct finite element modeling," *Acta Biomaterialia*, 2010, vol. 6, pp. 2342-2351
- [7] T. McCormack, E. Karaikovic, and R. W. Gaines, *The Load Sharing Classification of Spine Fractures*, 1994.
- [8] F. Magerl, M. Aebi, S. D. Gertzbein, H. Harms, and S. Nazarian, *A Comprehensive Classification of Thoracic and Lumbar Injuries*, 1994.
- [9] J. Homminga, R. Aquarius, V. E. Bultink, C. T. J. Jansen, and N. Verdonchot, "Can vertebral density changes be explained by intervertebral disc degeneration?," *Medical Engineering & Physics*, 2012, vol. 34, pp. 453–458
- [10] A. Shirazi-Adl, A. M. Ahmed, and S. C. Shrivastava, "Mechanical response of a lumbar motion segment in axial torque alone and combined with compression," *Spine*, 1986, vol. 11, pp. 914–927.
- [11] F. Ezquerro, F. G. Vacas, S. Postigo, M. Prado, and A. Simon, "Calibration of the finite element model of a lumbar functional spinal unit using an optimization technique based on differential evolution," *Medical Engineering & Physics*, 2011, vol. 33, no. 1, pp. 89-95
- [12] A. R. Vaccaro and J. S. Silber, *Post-traumatic Spinal Deformity*, 2001.
- [13] S. Anghel, "Predictive criteria for the development of thoracolumbar posttraumatic kyphosis based on FEM. 2014," PhD thesis UMF Tg-Mures.



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