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MODELLING A SEGMENT OF THE LUMBAR OF THE HUMAN VERTEBRAL COLUMN BY MEANS OF FINITE ELEMENT METHOD

Jan Awrejcewicz and Tomasz Antosik

Technical University, Division of Control and Biomechanics (I-10),
1/15 Stefanowskiego St, 90-924 Łódź, POLAND
e-mail:awrejcew@ck-sg.p.lodz.pl

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Abstract. *A biomechanical model of the human spinal segment (lumbar part) is presented and a new method of stabilization of an injured segment using an implant is outlined. The model includes 3D shapes of the vertebrae, discs, spinal joints, implants and other spine elements. The static investigations are focused on an optimal selection of the shape, mechanical properties and a fixation place of the proposed implant.*

1 INTRODUCTION

A spine (*columna vertebralis*) is the most complicated system of a human skeleton (see figure 1). Its stability is significant because the spine is related to the central nervous system (spinal cord) for which it is the only protection against potential mechanical damages[6]. The strength and resistance of the spine increase downwards due to the stronger construction and a bigger size of the vertebrae. This is why the lumbar vertebrae are bigger and more massive than the thoracic[1].

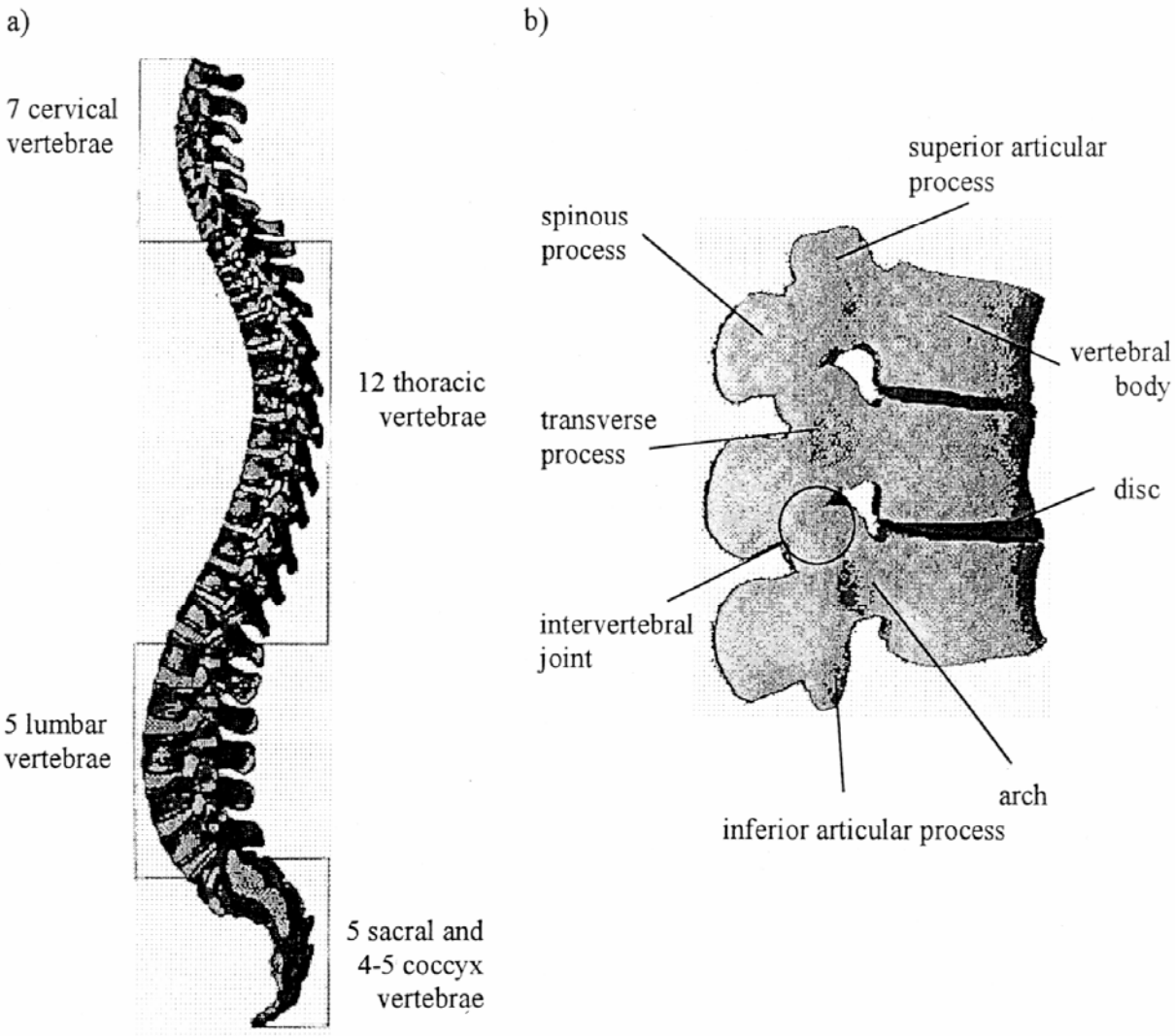

Figure 1: Human spine: a) cervical, thoracic, lumbar, sacral and coccyx vertebrae b) lumbar part of the spine taken from a dead body (anatomical specimen)

The most important segment of the human spine is the lumbar part[15] (see figure 1b). A human usually has five lumbar vertebrae *(vertebrae lumbalec)*. This part is the most frequently subject to different kinds of injury or pathologies.

Mechanical loads of the spine are the main reason for its illnesses, especially of the lumbar and cervical parts. Frequent changes and damage of the spinal structures are caused by exceeding of the resistance boundary values of the particular parts of the spine, for instance during street accidents, high altitude falls or carrying heavy loads[5,7]. These injuries can result also from normal life activities undertaken in non-anatomic conditions. Even a continuous impact of cyclic loads within the load efficiency of the vertebrae can lead to serious spinal injuries. A disturbance of the spinal structure due to an injury, fatigue or any other illness changes in the spinal area is a subject of interest of orthopaedists and biomechanical engineers[5].

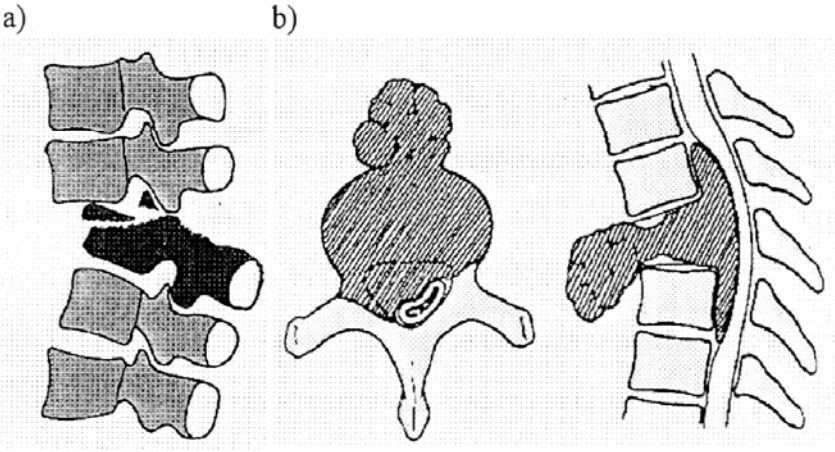

Figure 2: Spinal illnesses: a) multiple fracture of a vertebral body with a dislocation, b) spinal cancer

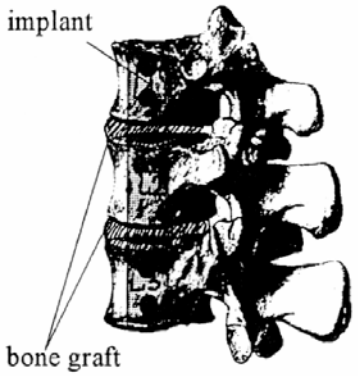

Figure 3: Stabilisation of a sick spine by an implant

The statistical investigation of the patients in Poland show that more than a half of the patients of orthopaedic and rheumatologic polyclinics are those with spine diseases[8]. Considering spinal pathologies, such as fractures or neoplasms, it is necessary to obtain a synostosis of the unstable (sick) spinal segment (see figure 2a,b). To achieve this, the spinal discs are removed and substituted by the autogenic (taken from the patients ilium) bone grafts, and then the sick spinal segment is fixed (stabilised) using an implant (see figure 3).

A sequence of biomechanical tests are needed before the direct application of implants in the clinical practice.

2 BIOMECHANICAL INVESTIGATION

A considerable progress in biomechanical engineering, connected with surgery and orthopaedy, has been achieved especially over the last forty years [5]. Over the past decade an

intensive development has been reported in the area of new implants used for stabilisation of all spinal segments[20,25]. The biomechanical tests define the implant efficiency during the stabilization of an arbitrarily chosen spinal segment[1,9] and the quality of the junction. They also allow one to compare the efficiency of different implants depending on a type of the unstable spinal segment and on the geometry of the fixation points of the implant to the spine.
The biomechanical investigations can be divided into direct and model ones, and numerical modelling.

In the direct investigations, a biological material is used as a model spine, which is often a spinal segment taken from a dead body. This kind of investigation is, however, rarely available because it is difficult to obtain an appropriate segment free from any pathological changes.

In the model investigations, an artificial spinal segment is used, for instance, made of silicon caoutchouc. However, there are a lot of drawbacks in this investigation. First of all, there is little resemblance between the model and the object since it is impossible to obtain the real biomechanical properties of the spine using an artificial model. In addition, the investigation is burdensome and expensive[5].

This is why the numerical modelling is the most promising as it also decreases the costs of research. The results of the investigation are obtained in figures or graphics, and it is not necessary to process them any further. Additionally, in numerical modelling it is relatively easy to repeat experiments[24].

3 FINITE ELEMENT METHOD (FEM) AND A SPINE MODEL

First FEM spinal models were developed in the 2D space[13,16]. However, their biomechanical accuracy was low and they did not take into account real three dimensional displacements of the spine and its segments. More appropriate 3D models have been built by Natarajan, Ke, Andersson[17]; Ueno and Liu[21]; Goel[10]; Zagrajek[24]. However, these models described only a structure, properties and the fundamental static principles of the human spine. Only a few authors used those models to test the efficiency of the stabilisation of a spine with an implant[3,4].

The presented numerical model was created in FEM in 3D space, and additionally, it was verified with the literature data concerning the in-vitro experiments made on human spines. In this case, the model was used to select an implant and estimate the efficiency of the applied stabiliser.

4 HUMAN SPINAL SEGMENT MODEL USING ANSYS 5.2 FEM APPLICATION

A human spinal segment is modelled using the numerical package ANSYS (version 5.2). The program allows for applying non-linear geometry and non-linear modelling of mechanical systems. Large changes in the geometry of the structure while transformation has to be considered in mechanical analysis of a human spine. For the purpose of numerical simulation two kinds of elements (A, B) have been used.

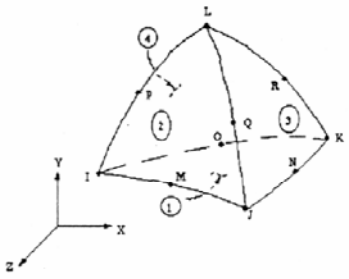

Figure 4: Element (SOLID 92)

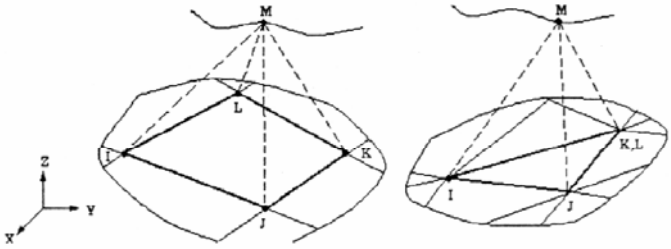

Figure 5: Element (CONTACT 49)

A. Tetrahedral element with 10 nodes (SOLID 92) - Fig. 4.
The element is defined by 10 nodes with three-degree-of-freedom. As well as the non-linear material orthotropic material can be used in analysis, especially while big displacements.
B. Surface type contact element (CONTACT 49) - Fig. 5.
It allows one to model two contact surfaces with slip possibilities (with or without friction). This element is defined by a node belonging to the contact surface and the target surface. The latter can by defined by 3 or 4 nodes. The contact neighbourhood is defined by a special sphere.

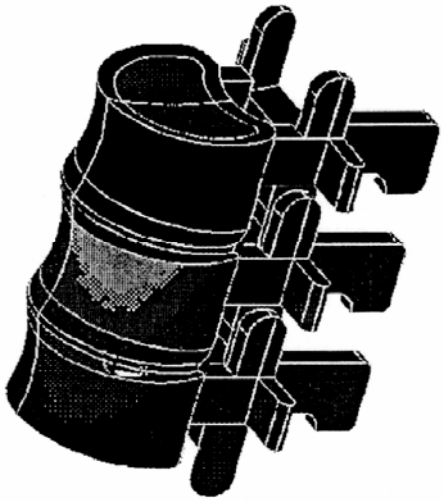

Figure 6: Human spine model, the lumbar part (solids)

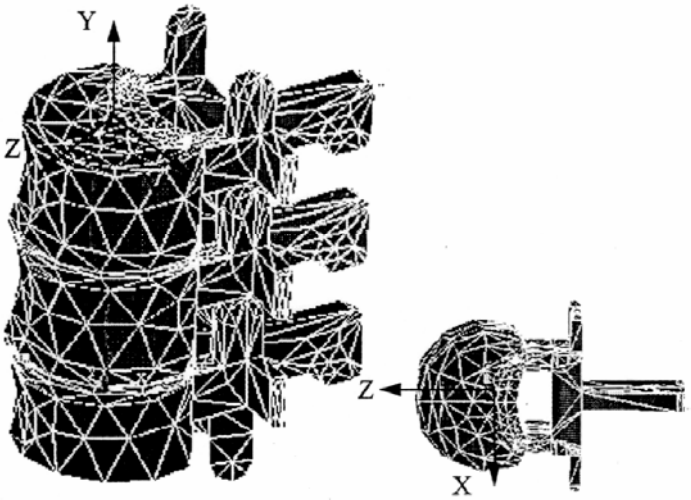

Figure 7: Model of the lumbar part divided into finite elements

Three lumbar vertebrae (see figure 6) have been modelled in the 3D space on the basis of anatomic atlases[22] and of a plastic spinal model of an optimum size. The disc (nucleus, annulus fiber) and bone structure (cortical, cancellous hard and soft bone) were taken into account. The existing solids were divided into the finite elements (see figure 7) on the basis the elements SOLID 92 and CONTACT 49.

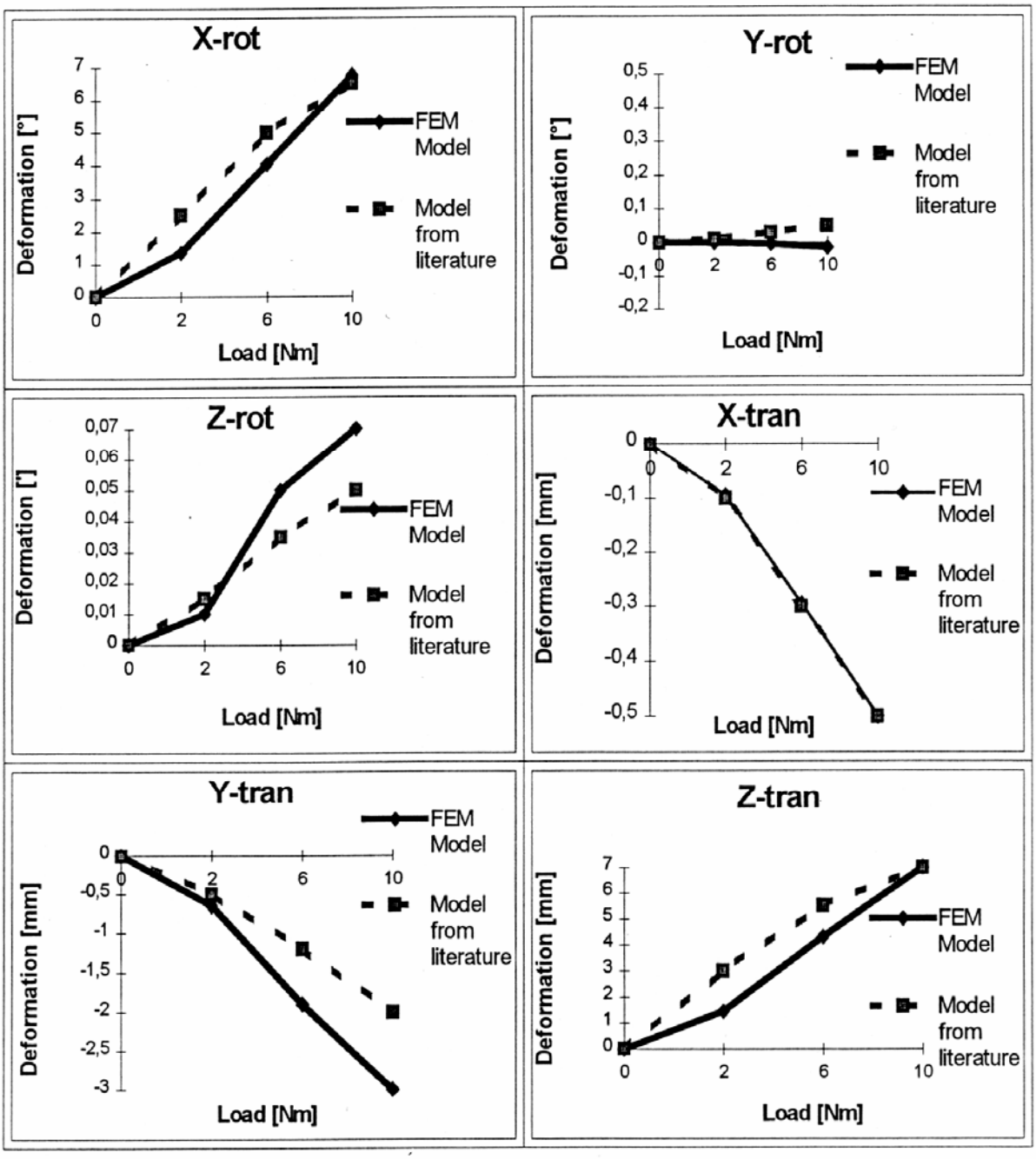

Figure 8: Deformation of the lumbar part of a spine without an implant

Table 1. Mechanical properties of a bone tissue adopted in the FEM model.

	E (MPa)	ν
Cortical bone[23]	11032	0.25
Cancellous bone[23]	345	0.2
Bony posterior elements[11]	3500	0.25
Annulus (fiber)[14,23]	20	0.45
Nucleus pulposus[23]	10	0.49

The adopted mechanical properties of the spine are presented in Table 1. This made it possible to check if the numerical spine model was created correctly. The results of the dislocations under the static loads were compared with the results obtained on the in-vitro biological specimens achieved by different authors. The Figure 8 presents the analysis of the

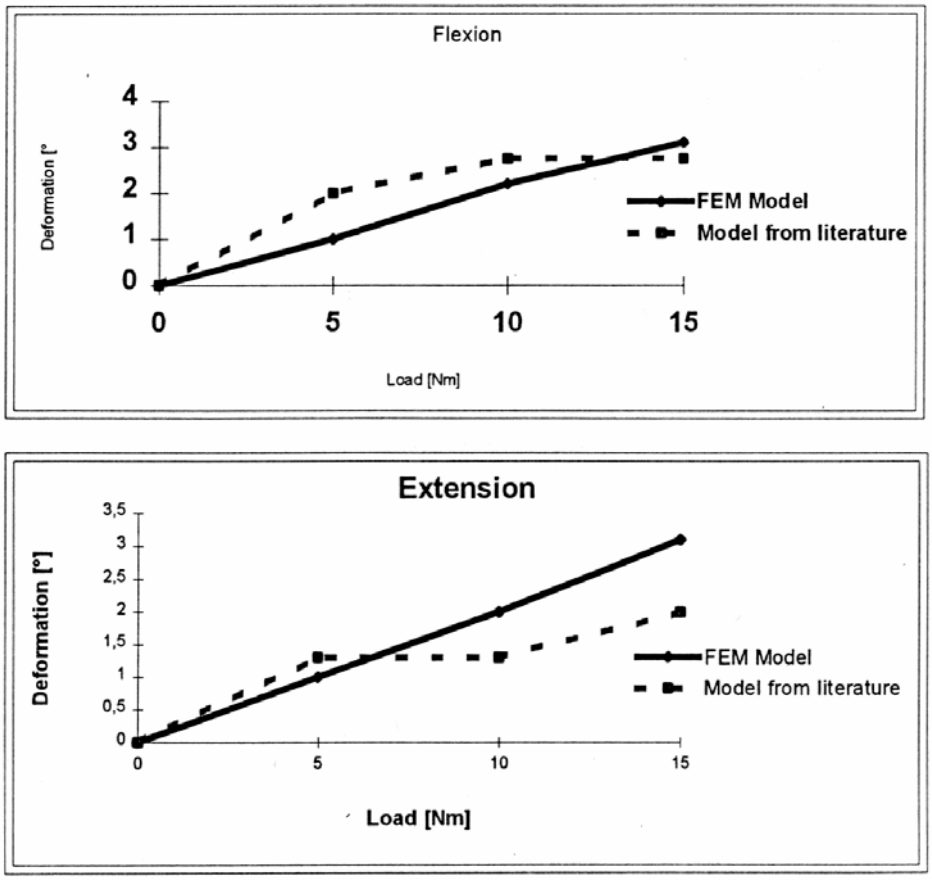

Fig. 9. Deformation of the lumbar part of a spine without an implant

Table 2. Mechanical properties of a bone tissue adopted in the FEM model.

	E (MPa)	ν
Cortical bone[23]	11032	0.25
Cancellous bone[23]	345	0.2
Bony posterior elements[11]	3500	0.25
Annulus (fiber)[23]	92	0.45
Nucleus[23]	10	0.49

dislocations of the lumbar segment (three vertebrae) of the six-degree-of-freedom (three rotations about the XYZ axes, and three dislocations along these axes). Kenneth's[12] results achieved in the in-vitro investigation were taken into account here. He loaded the investigated spine segment with 2,6,10 Nm causing a flexion. Our model was loaded with the same loads, and the obtained results agree with the biological model taken from a dead body. The numerical model was compared with other spine models investigated on the in-vitro stands. In this case, the mechanical properties of the bone tissue and of the intervertebra disc presented in Table 2 were adopted. The model was loaded with the moments: 5, 10, 15 Nm causing the flexion-extension. The dislocations of the numerical model caused by those loads are presented in the figure 9. These deformations were compared with the results of the investigation conducted by Panjabi.

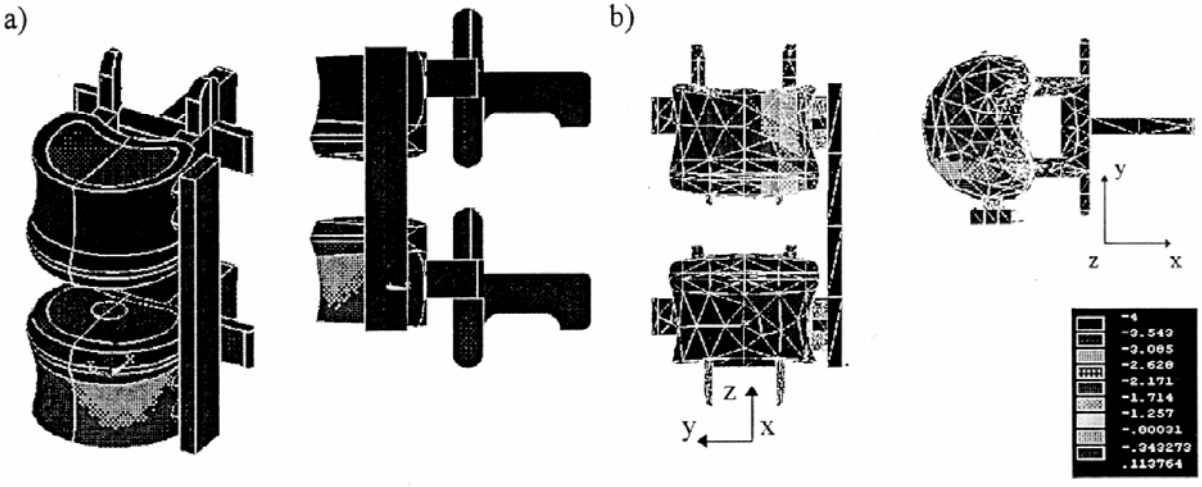

Figure 10: Human spine model with an implant a) solids b) FEM

The results achieved in the numerical model case agree, with the results obtained by Panjabi. This leads to the conclusion that the spine segment was modelled correctly and this model can be used to investigate different kinds of implants.

A middle vertebra was removed from a three-vertebrae spinal segment, and an implant was introduced composed of a metal plate fixed to the neighbouring vertebrae with the metal screws (see figure 10a). This arrangement was divided into finite elements (see figure 10b). The same mechanical properties of a bone were adopted as for the spine without an implant

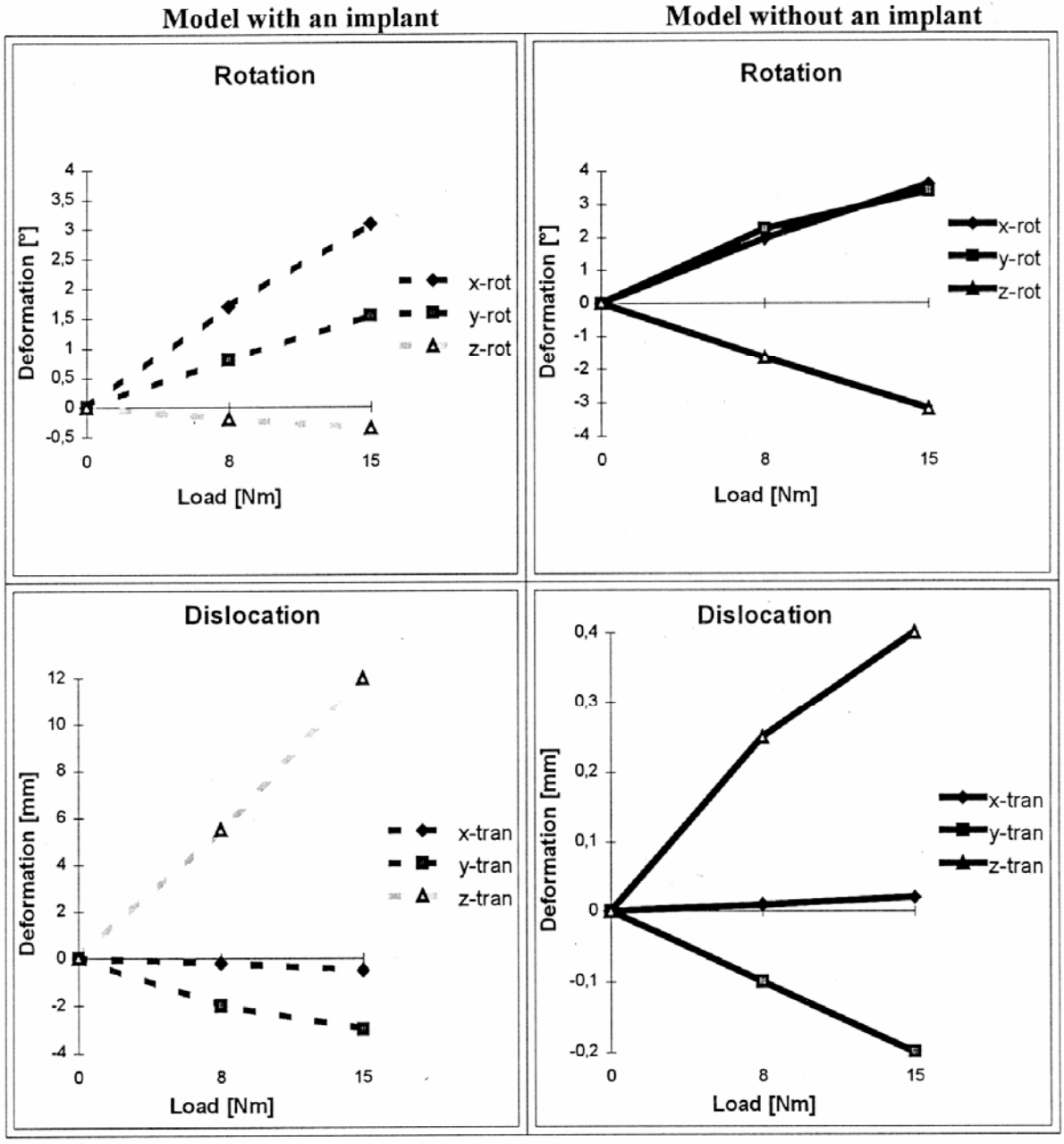

Figure 11: Deformation of the lumbar part with and without an implant

(Table 2) for the plate and the screws Young's modulus $E = 2.1 \times 10^5$ MPa, and the Poisson's ratio $\nu = 0.2$. The spine part has been subjected to the action of moments as it was in the case of a healthy spine (without an implant). In this case the loads were 8 and 15 Nm. The results obtained for the spine with a stabiliser were compared with ones obtained for the same moments acting on the healthy spine without an implant (see figure 11). In the case of the dislocations around the XYZ axes, the values for two analysed spine parts are close to each other, and for the shift along the axis they differ from each other by the order of magnitude. The implant applied by us made the spine more rigid than the healthy spine without an implant.

Like in any other bone, an adhesion appears in the spine when the healing place is stabilised. Generally, we can say that the more stabilised is the place of the expected adhesion, the greater is the chance of its appearance. However, it is unknown how big the rigidity must be in the place of the stabilised bone fractures to make an adhesion possible. On the other hand, it is known that too rigid stabilisation may lead to the appearance of another complication, osteoporosis - decalcification of the stabilised part, and osteoliosis - obliteration of a bone[2,18]. For the last few years, it has been believed that the combining of the bones, either long or vertebral, should be more flexible[19].

5 CONCLUDING REMARKS

The results obtained in the investigation of the created numerical model were compared with the results of the in-vitro investigation conducted by different authors. These results are convergent to a great extent. It can be concluded that the numerical model of the human spine segment was constructed correctly, and it can be used to investigate different kinds of implants. It gives a possibility to choose their optimum shapes, the appropriate mechanical properties of an implant, as well as the place and method of its fixation. The analysed implant led to the translation stabilisation of the spine along the XYZ axes in relation to the healthy spine without an implant. The rotation stabilisation was achieved as in the case of the healthy spine.

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