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Preoperative planning and post-operative estimation of vertebroplasty using CT/CAD/CAE systems

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The aim of the work was to determine the stiffness of vertebral bodies after vertebroplasty by means of radiological examination (CT), computer-aided design (CAD) and computer-aided engineering (CAE) systems. Twenty six patients with angiomata in vertebrae have been examined. A vertebra with pathological changes has been modelled twice, i.e. before operation and after the process of vertebral body filling with cement. The processing of CT images for the purposes of generation of 3D vertebral body models using Mimics software is also shown. In the analysis, non-homogeneous material properties of bone in the analysed areas are taken into consideration. Some problems related to the determination of non-homogeneous areas of particular material properties are discussed. FEM analyses described in the paper yielded the distributions of the stresses, strains and displacements in vertebral bodies. The stiffnesses of healthy vertebral bodies, bodies with pathological changes and bodies with bone cement injected were compared. The usefulness of the results obtained from the analyses of vertebral body stiffness for medical application in vertebroplasty was emphasised.

The method presented above allows us to put forward a different approach to the problem consisting in individual examination of each patient and planning the surgery according to the case by case conditions.

The computer-aided approach, using CT/CAD/CAE system, proposed above allows both improving the surgery performance and post-operative control of the patient condition.

Key words: *angioma, vertebroplasty, vertebral bodies, CT/CAD/CAE systems*

1. Introduction

The vertebroplasty is a low-invasive method for treatment that has been applied in the case of spine pathological changes. The method considerably reduces the feeling of pain and increases mechanical strength of the pathologically changed vertebral bodies [17].

The vertebroplasty consists in injection of cement into a pathologically changed vertebral body by means of a puncturing needle inserted through the vertebral pedicle. Since only analgesics are needed (in approx. 80–90% of cases [1], [7], [15]) and the operation is relatively simple, patients recover quickly. The first surgery of that type was performed by Herve Deramond in 1984 in the Clinic of Radiology and Neurosurgery of the University Hospital, Amiens, France. The surgery was successful since a considerable reduction of pain in the cervical part of the spine was achieved. Deramond applied the cement injection in the cases of compression fractures resulting from primary and metastatic sarcoma and also in the case of angiomata in vertebral bodies. Later on he applied the method also to compression fractures due to osteporosis. The first information on the vertebroplasty in literature appeared in 1987 when Galibert described and analysed over 20 cases of angiomata in vertebral bodies [3]. In the late 90s and at the beginning of the current century, the vertebroplasty has been more and more frequently applied to treatment of pathological

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changes of vertebral bodies. Recently the method has become very common in medical centers all over the world [4], [9], [8], [14].

HIGGINS et al. [5] showed a way how a geometrical model of vertebral body can be utilised. They used three-dimensional model of vertebral body extracted from cadaver in order to compare its volume with the amount of injected cement. The study was done to calculate the amount of cement that has to be applied without the risk of its efflux from the vertebral body. LIEBSCHNER et al. [1] studied 19 healthy vertebral bodies extracted from cadavers. They created three-dimensional models of vertebral bodies by means of finite element method. The models were used in biomechanical and strength studies that had experimental and cognitive character.

Until now the efficiency of vertebroplasty has been estimated only qualitatively. The scales of life quality (like VAS or SF-36) used in such cases gave only a subjective assessment of vertebroplasty efficiency made by the patient. There were no quantitave methods at surgeons' disposal. Hence, there is a strong need for a research method that could allow the quality of surgeon's work to be determined based on a mathematical approach.

Therefore, an estimation of the efficiency of a surgery, like through-skin verebroplasty, should also involve mechanical strength of vertebra after cement injection as well as the possibilities of complications, such as progressing deformation of vertebra pathologically changed and the fractures of adjoining vertebrae, very common in this surgical method. The papers on an estimation of the mechanical strength of vertebral body published so far are limited to experimental studies on the basis of autopsy. The studies yielded favourable results in regard to mechanical strength estimation, however, they concerned only vertebrae filled with cement in vitro.

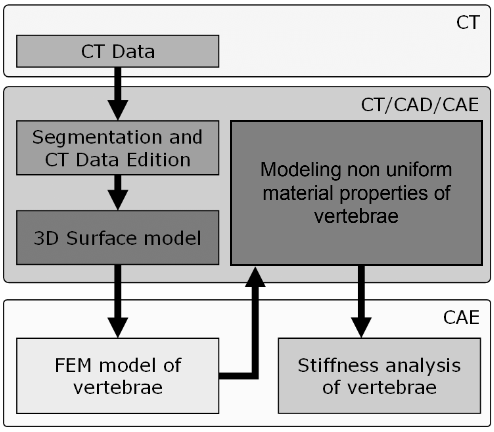


Fig. 1. Methodology

The aim of the work was to determine the stiffness of vertebral bodies after vertebroplasty by means of radiological examination (CT), computer-aided design (CAD) and computer-aided engineering (CAE) systems (figure 1).

2. Method

In order to carry out a correct analysis of vertebra, it is necessary to reconstruct its geometry, in particular the surface topology. Three-dimensional model creation is a complex process that is carried out in several steps. In the first step, human internal tissues are visualised using non-invasive method, i.e. computer tomography (CT). The most effective data acquisition is made by means of spiral tomography. The CT data is saved in the DICOM (Digital Imaging Communication in Medicine) format on a CD. The data can be then easily transmitted from the computer connected to CT machine (UNIX system) to a computer which allows geometrical modelling and strength analyses (Windows system).

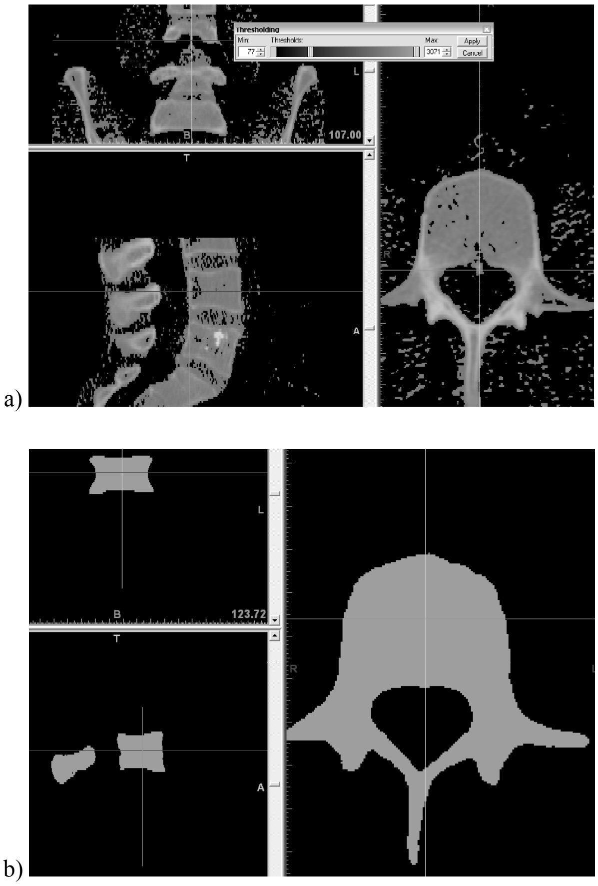


Fig. 2. Setting threshold values of *HU* (a), segmented and "manually" edited mask for L4 (b)

The CT data comprises information on scanned layers. The number of layers in one CT study depends on the gap between the layers and the range of CT scanning. The gap between the layers in spine CT study is usually 3 mm.

Every layer comprises information on radiological density value in *HU* (Hounsfield Units). In order to create a three-dimensional geometrical model, it is indispensable to discern regions containing bone tissue. Thus, image segmentation has to be done by setting a threshold values of *HU* for bone (figure 2). The region of bone is saved in the so-called mask, i.e. one-colour region.

Setting the threshold values of *HU* is insufficient to generate a geometrical model of bone structure, i.e. body of lumbar vertebra, because the mask generated automatically contains noises that affect the real shape of bone structure. It is, therefore, necessary to “manually” modify the mask to erase the noises and fill empty regions of bone (figure 2b).

In order to be able to perform further analyses, three-dimensional models of vertebral bodies have to be generated. The next step is then to separate the regions of the same material properties. This is done by creating masks that comprise one region of the same properties each and by erasing superfluous information coming from adjacent bodies. Mask edition must be performed on all the CT projection layers which is time-consuming; however, a correct geometrical model of the bone structure should be generated. The model of every vertebral body is related to a mask that defines its volume without selection of bone structures, i.e. cortical and trabecular tissues. In order to select the two bone tissues and cement/angioma, additional masks have to be created for these regions.

In the next step of modelling, three-dimensional surface mesh is created on the basis of the first mask (whole body). The mesh represents the bone structure of the body (figure 3). The parameters used during the mesh generation can be controlled, hence it is also easy to control the quality of the object.

The MIMICS system, which was used to generate the models, makes it possible to define the material properties (ρ – the density, E – the Young modulus, ν – the Poisson ratio) of bone structures in a few ways. In every case the properties depend on radiological density (in *HU*). Although bone tissue has extremely anisotropic properties, for the sake of simplicity it was modelled as an isotropic material. It is, therefore, possible to generate the model of non-homogeneous material properties by the determination of corresponding material groups.

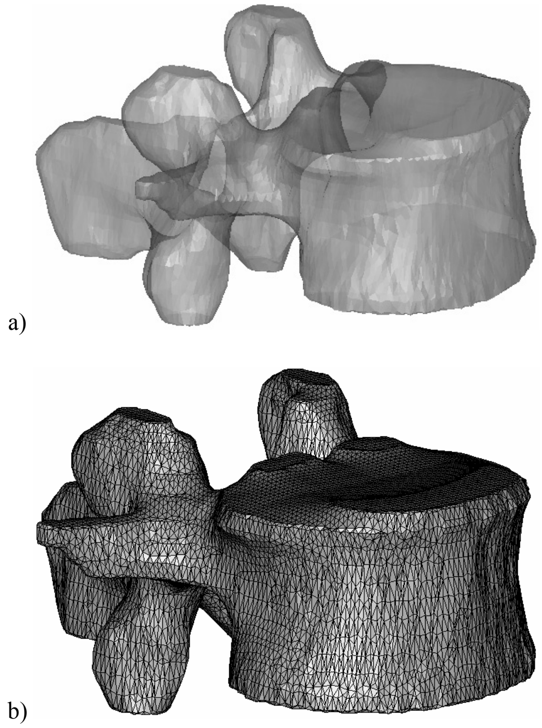
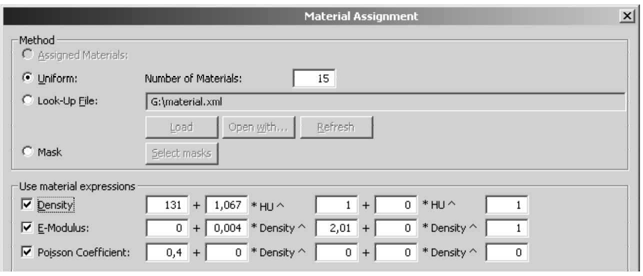


Fig. 3. Three-dimensional surface model: generated from the mask (a), meshed by means of surface elements (b)

After three-dimensional surface model generation it is necessary to mesh it by means of volume finite elements. This is done in the system which allows a finite element analysis to be carried out. In order to automatically generate volume finite element mesh, we made use of the Abaqus system. The model meshed with volume finite elements is transmitted to the software in which CT data are processed. In this software, the definition of material properties is completed.

After calculating *HU* value, in every element the material properties are defined (figure 4).



Method
○ Assigned Materials:
● Uniform: Number of Materials: 15
○ Look-Up File: G:\material.xml
○ Mask

Use material expressions
☑ Density 131 + 1,067 * HU ^ 1 + 0 * HU ^ 1
☑ E-Modulus: 0 + 0,004 * Density ^ 2,01 + 0 * Density ^ 1
☑ Poisson Coefficient: 0,4 + 0 * Density ^ 0 + 0 * Density ^ 0

Fig. 4. Definition of material properties

2.1. Automatic definition of material properties

In the first way of defining material properties, the Young modulus is determined automatically based on the equation relating the Young modulus to the radiological density *HU*. In a histogram, the software shows the number of elements that were attributed to a given radiological density. Every element has its own *HU* level based on CT data. To convert *HU* into material properties two approaches can be adopted. In the first one, discretisation follows in which the *HU* range is re-divided. The division depends on the approach accepted.

In the approach of uniform discretisation, *HU* range is divided into sections of equal intervals. Each section represents different material property. Average value of *HU* in each section is selected as the representative value for a given range (figure 5).

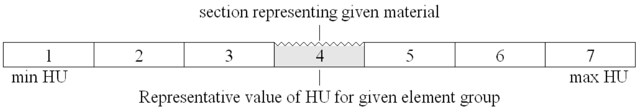


Fig. 5. Discretisation of *HU* range uniform approach

In the next step, the material properties of bone tissue are defined by empirical equations:

- for the material density ρ [kg/m^3]

$$\rho(HU) = a_1 + a_2(HU)^\wedge a_3 + a_4(HU)^\wedge a_5, \tag{1}$$

- for the Young modulus E [MPa]

$$E(\rho) = b_1 + b_2\rho^\wedge b_3 + b_4\rho^\wedge b_5, \tag{2}$$

- for the Poisson ratio ν [–]),

$$\nu(\rho) = c_1 + c_2\rho^\wedge c_3 + c_4\rho^\wedge c_5, \tag{3}$$

where a_i, b_i, c_i ($i = 1, 2, ..., 5$) are the coefficients dependent on bone tissue (cortical or trabecular) [6], [12], [13].

The main disadvantage of the uniform approach is that determining the Young modulus for elements is inaccurate. The empirical equations given in literature [6], [12], [13] relate to either cortical bone or trabecular bone (figure 6). Whereas in the software one can use only one equation which determines the Young modulus for both bone tissues.

In the preliminary analyses, equations (4) and (5), proposed in [13], were used. A constant Poisson ratio was assumed:

$$\rho = 1.122 \cdot HU + 47, \tag{4}$$

$$E = 1.92 \cdot \rho - 170, \tag{5}$$

$$\nu = 0.3. \tag{6}$$

In the second approach, discretisation is done by "manual" definition of *HU* range sections (figure 8). Here the data, which determine *HU* ranges for different tissues, taken from [6], [18] can be utilised. Thus, one can separate bone structures, i.e. trabecular tissue and cortical tissue, as well as pathological change of bone and injected cement. Constant material properties are determined for each structure. This is completed by means of a file .xml read in the software, which contains description of *HU* ranges and the corresponding mechanical properties.

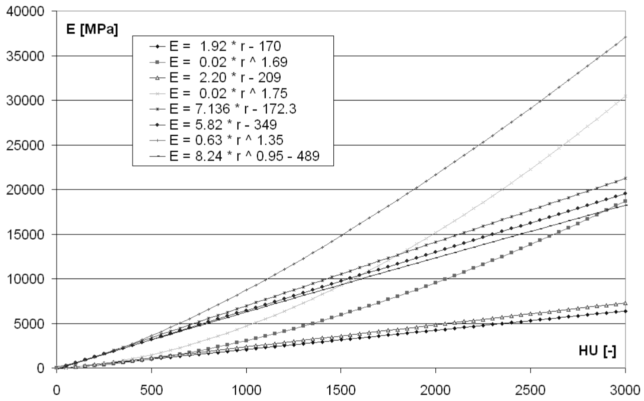


Fig. 6. Relations between material properties and *HU* [6], [13]

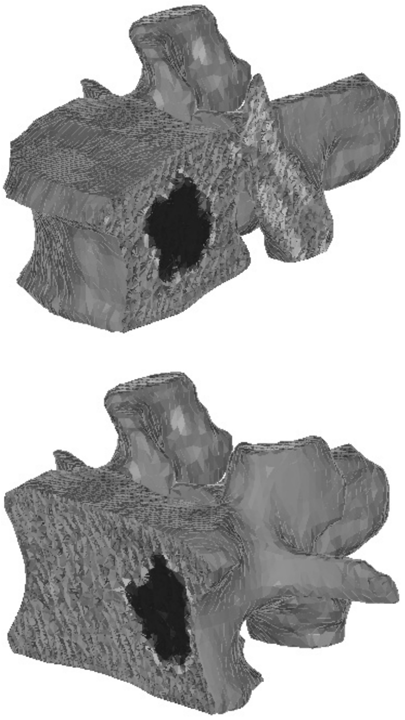


Fig. 7. Material properties automatically assigned to the vertebrae model (cross-section)

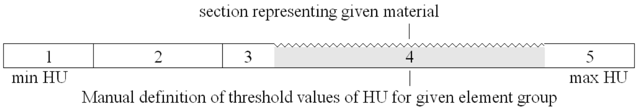
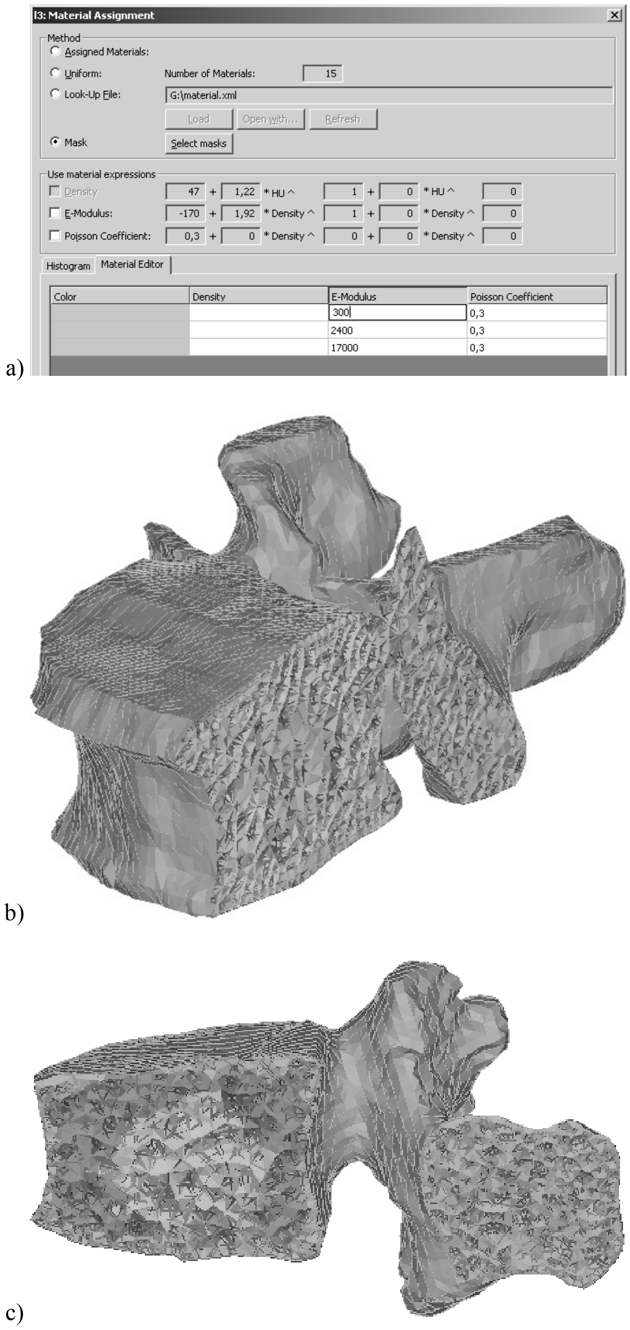


Fig. 8. Discretisation of *HU* range of non-uniform approach (look up file)

2.2. Manual definition of materials properties

The second way of material properties definition consists in the separation of several regions, i.e. cortical bone, trabecular bone, cement/angioma, as masks and the determination of material properties of those masks (figure 9). This way makes it possible to precisely manually define volumetric regions of similar bone structures.



I3: Material Assignment

Method
○ Assigned Materials:
○ Uniform: Number of Materials: 15
○ Look-Up File: G:\material.xml
● Mask

Use material expressions
☐ Density 47 + 1,22 * HU ^ 1 + 0 * HU ^ 0
☐ E-Modulus: -170 + 1,92 * Density ^ 1 + 0 * Density ^ 0
☐ Poisson Coefficient: 0,3 + 0 * Density ^ 0 + 0 * Density ^ 0

Color	Density	E-Modulus	Poisson Coefficient
		300	0,3
		2400	0,3
		17000	0,3

Fig. 9. Manual assigning of non-homogeneous material properties (a) to the vertebrae using mask (b, c)

3. Analyses

Twenty six patients with angiomata in their vertebrae have been examined. A vertebra with pathological changes has been modelled twice, i.e. before operation and after the vertebral body filling with cement.

Three geometrical models of vertebrae of three different patients were generated in order to analyse numerically the vertebra stiffness by means of finite element method. The material properties of the models were non-homogeneous. Each model consisted of tetrahedral elements of linear characteristics. The model of vertebral body after vertebroplasty consists of the following material regions:

a) trabecular bone,
b) cortical bone,
c) bone cement.

In the second model, the region of bone cement was interpreted as pathological change, i.e. angioma.

In the third model, the element group corresponding to cement had mechanical properties of trabecular bone. Mechanical properties of each region that constitutes the vertebra models are shown in the table.

Material regions and their mechanical properties in numerical analyses

	E [MPa]	*ν* [–]	Literature
Cortical bone	17000	0.3	[16]
Trabecular bone	300	0.3	[10]
Cement 1	2 400	0.3	[7]
Cement 2	1 900	0.3	[–]
Cement 3	1 400	0.3	[–]
Angioma	0.00001	0.45	[–]

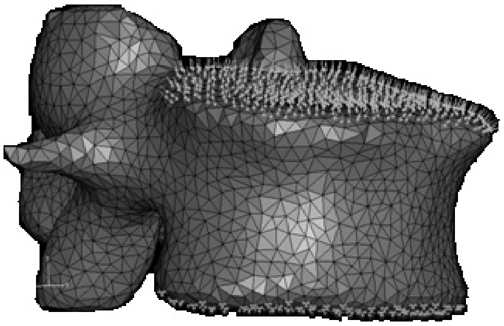


Fig. 10. Boundary conditions defined in numerical analysis

Numerical analyses by means of finite element method with displacement and stress boundary conditions were carried out for each model. In the analysis with displacement boundary conditions, the following boundary conditions were defined (figure 10):

- on the lower face of the body all displacement degrees of freedom were fixed,
- 1 mm displacement of axial direction and negative sense was applied to 302 nodes on the upper face.

In the stress analysis, the following boundary conditions were defined:

- on the lower face of the body all displacement degrees of freedom were fixed,
- on the upper face a 4 MPa pressure was applied; the pressure corresponds to the total force F distributed uniformly on the upper face ($A \approx 1300$ mm^2, $F \approx 5200$ N).

4. Results

CT data studies of twenty six vertebral bodies after vertebroplasty by means of the method described above make it possible to estimate the degree of filling angioma with bone cement. It was observed that full filling of angioma was obtained in 10 (38.5%) vertebral bodies. Partial filling was observed in 16 (61.5%) vertebral bodies. The observations were essential for engineering analyses, because identification of the left-over angioma has the considerable importance for defining the material properties of the regions. In the cases of partial filling, angioma occupied on average 21.79% of body volume (in the range from 9.8% to 42%). An average volume of angioma totally filled in respect of the volume of the whole body was 9.29% (from 4.6% to 11.4%). On the basis of stiffness studies it was observed that in the case of filled vertebrae the average stiffness of vertebra before surgery was 0.96086 (± 0.01272), from 0.9419 to 0.9725; after surgery 1.17348 (± 0.07008), from 1.0818 to 1.3053. In the case of partially filled bodies before surgery, average body stiffness was 0.92186 (± 0.03413), from 0.8404 to 0.9861; after surgery 1.36118 (± 0.26468), from 1.0779 to 1.8240 [2].

The aim of stress analyses was to observe the phenomena that take place in vertebral bodies under load of such value that it may cause body damage. It cannot, however, be stated unequivocally whether or not the calculated stresses correspond to the values that may damage the vertebra, because the maximum stresses the particular bone structures are able to withstand are not known. It is possible to estimate the calculated stresses in respect of the allowable stress of bone structures, provided that this stress is known. The aim of the analyses was to calculate and compare reduced stresses, strains and resultant displacements in the three vertebral bodies caused by the particular load. The distributions of stresses, strains and displacements are shown in figures 12–14.

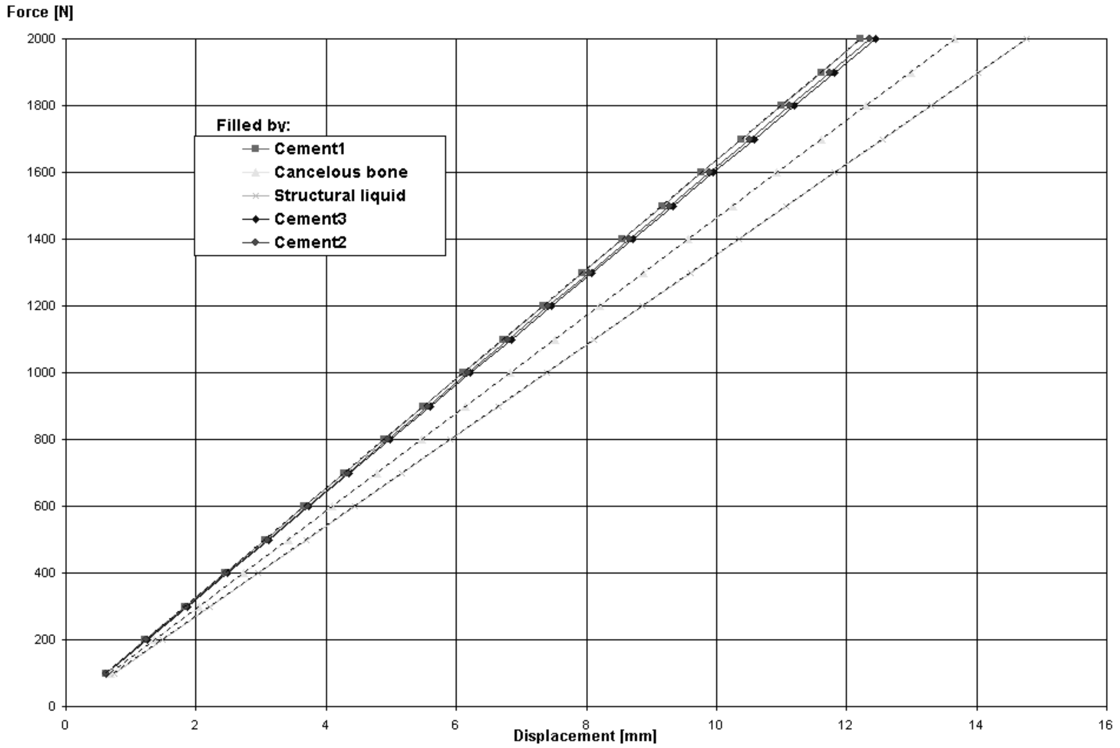


Fig. 11. Stiffness characteristics of healthy vertebra before and after vertebroplasty (three different cement types used)

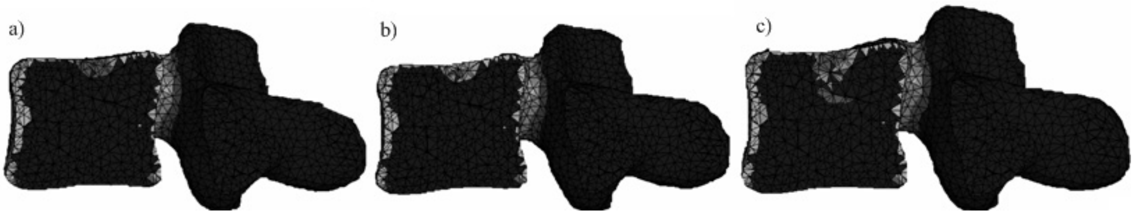


Fig. 12. Stress in cross section: healthy vertebral body (a), before (b) and after (c) vertebroplasty

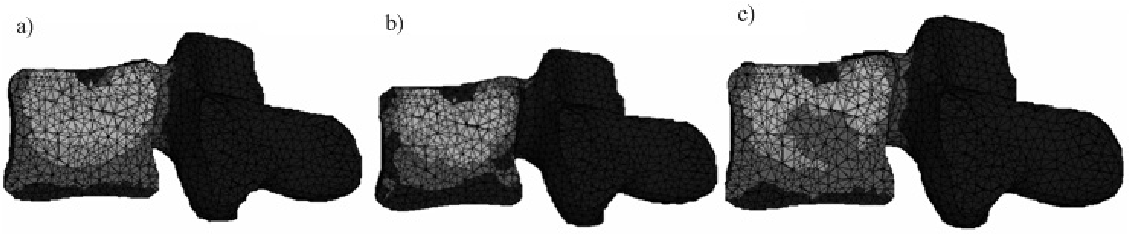


Fig. 13. Strain in cross section: healthy vertebral body (a), before (b) and after (c) vertebroplasty

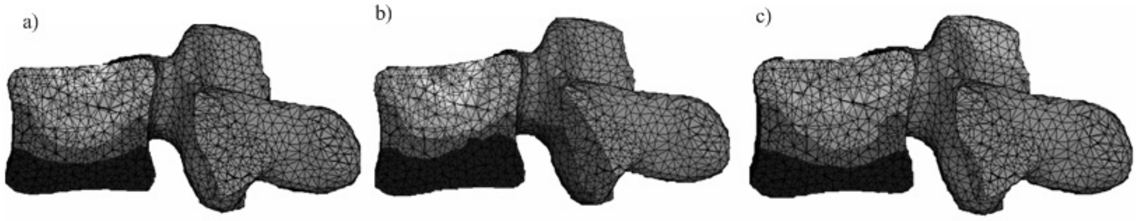


Fig. 14. Displacements in cross section: healthy vertebral body (a), before (b) and after (c) vertebroplasty

It is worth mentioning that an evaluation of vertebra functionality in the spine segment should also include the stress redistribution in the neighbouring vertebrae. Because of the complexity of the problem and a limited range of the work, our considerations are limited to the analysis of the stiffness of one vertebra.

5. Conclusions

In the paper, the methodology of the geometrical and material modelling of vertebral bodies was shown. The method makes it possible to quantitatively estimate the body stiffness after vertebroplasty.

CT images used in building 3D geometrical models of vertebral bodies in a CAD system by means of finite elements allow a correct diagnosis and plan of a surgery. Then based on 3D models built after the surgery, the degree of cement penetration into the vertebral body as well as possible post-operative primary complications could also be estimated.

The existing methods of treatment effectiveness estimation were based on subjective opinion on pain complaint. Thus, the developed method of objective estimation of surgery effectiveness together with CT or X-ray data acquires an essential importance. The importance of the stiffness bone cement, which is used in vertebroplasty in the case of angioma, has not been analysed yet. The studies carried out by the authors indicate that there is a possibility of damage of pathologically changed vertebral body treated with cement injection due to significant bone cement stiffness.

The results of the research presented above have proved that the issue of the mechanical strength the vertebral body reveals after the cement injection should also be considered when assessing the efficiency of the low-invasive through-skin vertebroplasty.

Based on the on figures presented the possibilities of post-operative complications, e.g. further development of vertebral body deformation, fractures of the adjoining vertebrae, etc., should be analysed as well.

The results of research in this field available in the literature were obtained from autopsy-based experiments or in vitro investigation. The method presented above allows us to put forward a different approach to the problem consisting in individual examination of each patient and planning the surgery according to the case by case conditions. Post-operative assessment of the surgery result is the second step of the method.

The method presented makes it possible to quantitatively estimate the body stiffness after vertebroplasty. On the basis of FEM analyses the following conclusion can be drawn:

- in the vertebroplasty planning the importance of the bone cement applied has not been analysed,
- the method of material property definition makes it possible to relate the bone radiological density to material properties,
- filling the whole angioma with cement leads to an increase of vertebral body stiffness (this may be observed in figure 11),
- the results of numerical calculations with displacement boundary conditions show a slight difference between the stiffness of healthy vertebral body and that of the body with angioma, if the volume of angioma is small in respect of the whole body volume,
- stress analysis indicates that stress concentration appears around the injected cement,
- the most time-consuming work in the model preparation for numerical calculations is the detection of regions with the same material properties.

It is important to emphasize that based on the methodology presented we can suggest that the cement of determined properties, i.e. the stiffness after polymerisation, has to be selected for a concrete clinical case. This should be done by a surgeon who selects a proper cement, which is available at the moment.

The computer-aided approach, using CT/CAD/CAE system, proposed above allows both improving the surgery performance and post-operative control of the patient condition.

Acknowledgements

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References

[1] BELKOFF S.M., MATHIS J.M., JASPER L.E., DERAMOND H., *The biomechanics of vertebroplasty. The effect of cement volume on mechanical behaviour*, Spine, Jul., 2001, 15, 26(14), 1537–41.

[2] CEJMER W., *Ocena skuteczości wertebroplastyki w naczyniakach trzonów kręgów*, PhD thesis, Wojskowy Instytut Medyczny, 2007.

[3] DERAMOND H., DEPRIESTER C., TOUSSAINT P., GALIBERT P., *Percutaneous Vertebroplasty*, Semin. Musculoskelet. Radiol., 1997, 1(2), 285–29.

[4] DIETRICH M., KĘDZIOR K., SKALSKI K., ZAGRAJEK T., KRZESIŃSKI G., SKOWORODKO J., *On concurrent engineering and design of an intervertebral disc of lumbar spine*, Proceedings of the 18-th International Conference on Computer-aided Production Engineering CAPE 2003, Professional Engineering Publishing Limited, Edinburgh, 2003, 209–219.

[5] HIGGINS K.B., HARTEN R.D., LIN D., VIEIRA P., LANGRANA N.A., REITER M.F., *Biomechanics of Patient Specific Vertebroplasty*, 2001, BED-Vol. 50, Bioengineering Conference ASME.

[6] HOBATHO M.-C., RHO J.Y., ASHMAN R.B., *Anatomical variation of human cancellous bone mechanical properties in vitro*, Studies in Health Technology and Informatics, 1997, Vol. 40, 157–173.

[7] JASPER L.E., DERAMOND H., MATHIS J.M., BELKOFF S.M., *Material properties of various cements for use with vertebroplasty*, Journal of Materials Science: Materials in Medicine, 2002, 13, 1–5.

[8] JENSEN M.E., EVANS A.J., MATHIS J.M., KALLMES D.F., CLOFT H.J., DION J.E., *Percutaneous polymethylmethacrylate vertebroplasty in the treatment of osteoporotic vertebral body compression fractures: technical aspects*, AJNR, 1997, 18(10), 1897–904.

[9] KAEMMERLEN P., THIESSE P., JONAS P. et al., *Percutaneous injection of orthopedic cement in metastatic vertebral lesions*, N. Engl. J. Med., 1989, 321, 121.

[10] KEYAK J.H. et al., *Three dimensional finite element modelling of bone: effect of element size*, J. Biomed. Eng., November 1992, Vol. 14.

[11] LIEBSCHNER M.A., KOPPERDAHL D.L., ROSENBERG W.S., KEAVENY T.M., *Finite element modelling of the human thoracolumbar spine*, Spine, 2003, 28(6), 559–65.

[12] Mimisc 9.1 Help

[13] RHO J.Y., HOBATHO M.-C., ASHMAN R.B., *Relations of mechanical properties to density and CT numbers in human bone*, Medical Engineering and Physics, 1995, Vol. 17, No. 5, 347–355.

[14] TAYLOR W.R., ROLAND E., PLOEG H., HERTIG D., KLABUNDE R., WARNER M.D., HOBATHO M.C., RAKOTOMANANA L., CLIFT S.E., *Determination of orthotropic bone elastic constants using FEA and modal analysis*, Journal of Biomechanics, 2002, Vol. 35, 767–773.

[15] TROJANOWSKI T., KAMIŃSKI J., JANCZAREK M., TROJANOWSKA M., *Methyl methacrylate embolisation and stabilisation of vertebral hemangioma*, Zentralbl. Neurochirg., (Suppl.), 1998, 154S.

[16] WEINANS H. et al., *The behaviour for adaptive bone-remodelling simulations models*, J. Biomechanics, 1992, Vol. 25, No. 12.

[17] WHITE A.A., PANJABI M.M., *Clinical Biomechanics of the Spine*, J. B. Lippincott Company, 1990.

[18] WHYNE C.M., HU S., LOTZ C., *Parametric finite element analysis of vertebral bodies affected by tumors*, Journal of Biomechanics, 34, 1317–1324, 2001.