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INVESTIGATION OF BONE REMODELING AFTER IMPLANTATION OF PEDICLE SCREWS

A b s t r a c t

Mechanical load is related to morphology of bone tissue, in a way that bone tissue structure is adapted to load of which influence is exposed to. Process that regulates this relationship is named bone remodeling. Bone remodeling can be mathematically described by bone remodeling equation that can be integrated with finite element method in order to simulate this adaptive process. Being familiar with mechanisms of bone remodeling is of great importance for implants design because it enables monitoring of bone tissue behavior during post-operative implanted conditions. Post-operative implanted conditions change mechanical load of bone tissue that could be cause of bone remodeling initialization. Therefore, this paper deals with investigation of bone remodeling caused by implantation of pediclescrews.

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ISTRAŽIVANJE ADAPTACIJE KOŠTANOG TKIVA NAKON UGRADNJE KIČMENIH ZAVRTNJEVA

Izvod

Mehaničko opterećenje je povezano sa strukturom koštanog tkiva na način da se struktura koštanog tkiva prilagođava opterećenju kojem je izložena. Proces koji reguliše ovu vezu se naziva adaptacija koštanog tkiva. Adaptacija koštanog tkiva može biti matematički opisana jednačinom adaptacije koja se može integrisati sa metodom konačnih elemenata kako bi se ovaj adaptivni proces simulirao. Poznavanje mehanizama procesa adaptacije koštanog tkiva je od velikog značaja za projektovanje implantata jer omogućava praćenje ponašanja koštanog tkiva u postoperativnim uslovima nakon ugradnje implantata. Postoperativni uslovi nastali ugradnjom implantata izazivaju promjenu mehaničkog opterećenja koštanog tkiva koja može biti uzrok pokretanja adaptivnog procesa. Stoga se u ovom radu istražuje proces adaptacije koštanog tkiva izazvan ugradnjom kičmenih zavrtnjeva.

1. INTRODUCTION

Research regarding the relationship between mechanical load and morphology of bone tissue can be traced back to work of Germanan atomist Wollf from 1892 year. Based on his research Wollf suggested a hypothesis, known as Wollf's law, after which structure of bone tissue can adapt under influence of mechanical load to which is exposed. Wollf also suggested that change of bone tissue structure caused by change of mechanical load is optimal, what means that tissue tends to minimal mass needed to bear load. Process, which regulates relationship between mechanical load and structure of bone tissue is usually called bone remodeling. First hypothesis about mechanisms which cause bone remodeling was suggested by German surgeon Roux in 1895, he assumed that cells of bone tissue can "sense" mechanical load and respond to that stimulus at the cell level by initiating bone remodeling.

Starting point of all contemporary theories of bone remodeling is Wollf's and Roux's work. Therefore, it is believed that bone tissue structure adapts to mechanical load to which is exposed. It is also believed that bone tissue contains specialized cells (osteoclasts and osteoblasts), which at the same time act as sensors which "sense" load and as regulators of bone remodeling which increase or decrease bone mass. Change of bone mass during bone remodeling can be achieved by changing density or geometry.

First mathematical model, based on general principles of continuum mechanics, which establishes functional relationship between mechanical load and bone remodeling was suggested by Cowin and Hegedus in 1976 [1]. Model, suggested in 1986 by Fyhrie and Carter [2], considers bone remodeling as optimization process during which bone tissue adapts its structure and density to stress-strain state. Recently simulation of bone remodeling based on integration of finite element method, as an effective tool for accurate determination of stresses and strains, and mathematical model of bone remodeling was proposed by many authors (Weinans et al.,1992,[4]; Mullender et al.,1994,[5]; Xinghua et al.,2002,[7]).

Being familiar with mechanisms of bone remodeling is of great importance for implants design because it enables monitoring of bone tissue behavior during post-operative implanted conditions. Because post-operative implanted conditions change mechanical load of bone tissue that could be cause of initialization of bone remodeling. Application of bone remodeling theory in implants design is object of research conducted by Huiskes et al. during 1987 [3]. Researches, conducted during 1998 by Martinez et al. [6] and during 2003 by Waide et al. [9], were investigating behavior of bone tissue during post-operative implanted conditions, which initiate bone remodeling.

Therefore, this paper deals with investigation of bone remodeling after implantation of pedicle screws. Pedicle screws are specially designed screws that are meant to be implanted into the pedicles of the spinal vertebrae. They have traditionally been used in the lumbar spine, and with recent advances in technology and technique, surgeons are now using them in the thoracic spine too. Screws provide strong "anchorage" points to which rods can be attached. Rods can then be contoured to correct deformities of spine.

2. METHODS

2.1 Model of bone remodeling

Simulation of bone remodeling, used in this paper, is based on remodeling model of Mullender et al. (5). Regarding this model continuum, which is filled by bone tissue is boiled down to finite elements. Each element contains a sensor cell placed into its centroid. According to model of Mullender sensor cells, after "sensing" mechanical load which initiates bone remodeling, send signals which cause initialization of adaptive process within range extending out of boundaries of elements inside which cells are placed. Effect of this signal to bone remodeling decreases as remoteness from sensor cell location increases. This assumption includes influence of all sensor cells into bone remodeling with respect to its remoteness from location at which bone remodeling takes place.

According to this model equation of bone remodeling can be expressed:

$$\begin{aligned} &\frac{d\rho(x,t)}{dt} = B \cdot \sum_{i=1}^{n} f_i(x) \cdot \left[\frac{U_i}{\rho_i} - k\right] \\ &0 < \rho \leq \rho_{cb}, \end{aligned} \tag{2.1}$$

where n is number of finite elements, U_i density of strain energy at centroid of finite element, (i density of bone tissue of finite element, k reference stimulus value and B constant of bone remodeling. Remodeling model suggested by Mullender involves function of spatial influence which physically simulates influence of sensor cells on neighboring bone tissue:

$$f_i(x) = e^{-\frac{d_i(x)}{D}}, \tag{2.2}$$

where $d_i(x)$ is remoteness from sensor cell to location x and D range of sensor cell influence. Young's modulus of finite element can be determined by the following equation:

$$E = C \cdot \rho^{\gamma}, \tag{2.3}$$

where C and γ are constants.

Mullender converted diferential equation of bone remodeling into an explicit time integration scheme using constant time step Δt, after which the following expression is obtained:

$$\begin{aligned} &\Delta\rho(x,t) = \Delta t \cdot B \cdot \sum_{i=1}^{n} f_i(x) \cdot \left[\frac{U_i}{\rho_i} - k\right] \\ &0 < \rho \leq \rho_{cb}. \end{aligned} \tag{2.4}$$

New value for density of finite element can be determined from the following expresion:

$$\rho(x, t + \Delta t) = \rho(x,t) + \Delta\rho(x,t). \tag{2.5}$$

Bone remodeling in each finite element is considered to be converged if one of the following three conditions is satisfied:

(i) reached preset reference stimulus value of ratio between density of strain energy and density of bone tissue k,

(ii) reached density of cortical bone tissue $\rho = \rho_{cb}$,

(iii) complete resorbtion of bone tissue from finite element ρ=0.01 g/cm^3. Reaching one of the bone remodeling equilibrium conditions can be achieved by changing density of bone tissue in finite element.

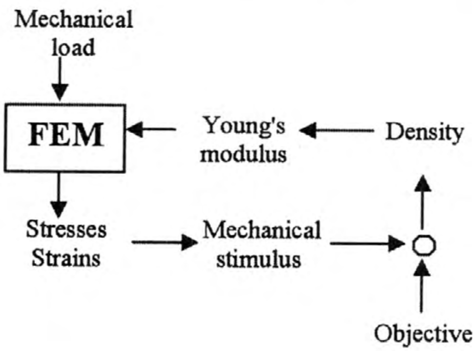

Figure 1. The iterative feed-back mechanisms of simulation of bone remodeling integrated with finite element method

Simulation of bone remodeling integrated with finite element method, Figure 1, keep on going iterativly until process would be finished in all finite elements.

The convergence behaviour of bone remodeling can be investigated by objective function F which is defined by the following expresion:

$$F = \frac{1}{m} \cdot \sum_{i=1}^{m} \Big| \sum_{i=1}^{m} f_i(x) \cdot \Big(\frac{U_i}{\rho_i} - k \Big) \Big|, \tag{2.6}$$

where m is number of finite elements inside which bone remodeling keep on going.

Values of constant parameters of model used for simulation of bone remodeling (5): B=1 $(\mathrm{g/cm^3})^2/(\mathrm{MPa} \cdot \text{time unit})$, ρ_{cb}=1.74 g/cm^3, C=100 MPa/(g/cm^3)^2, γ=2, ν=0.3. Poisson's ratio, ρ_0=0.8 g/cm^3 density of bone tissue in all finite elements at the beging of bone remodeling simulation.

Remaining model parameters were chosen so that stability of bone remodeling would be provided, as well as that final structure of bone tissue would be similar to the real one: D=0.5 mm, k=0.006 J/g, Δt=2 time unit.

2.2 Physical and geometric model of lumbar vertebra L_5

Lumbar vertebra L_5, shown in Figure 2, was used as basis to generate geometric model.

Figure 2. Physical model of vertebra L_5

3D geometric model of vertebra, shown in Figure 2, was generated in environment of comercial geometric modeler Mechanical Desktop 6. Modeling was realized by an automatic generator of geometric models of lumbar vertebrae, which is presented in work of J. Jovanović and M. Jovanović (10) in 2002 year.

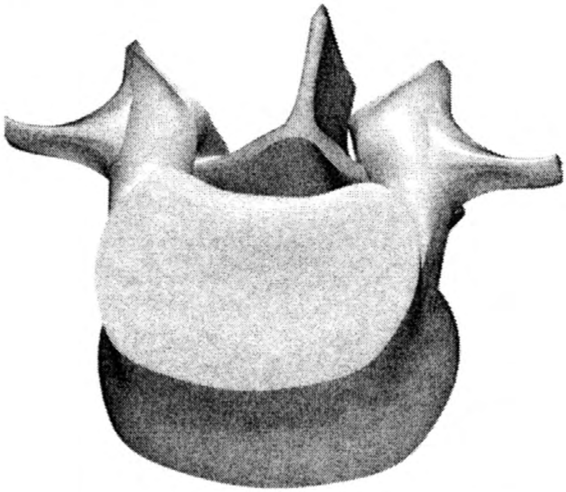

Figure 3. 3D geometric model of vertebra L_5

3D geometric model of vertebra, shown in Figure 3, was used as basis to generate 2D geometric model of vertical section of vertebral body, which is shown in Figure 4.

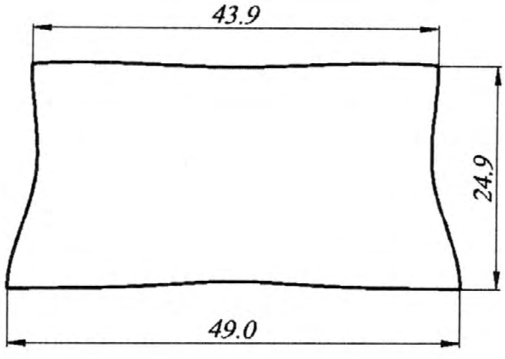

Figure 4. 2D geometric model of vertebra L_5

2.3 Physical model of pedicle screw

Physical model of used pedicle screw with diameter of 6.35 mm is shown in Figure 5. It is a product of factory Corin Spinal System from USA. This implant is made of titanium alloy Ti6Al-4V with specific density of ρ=4.43 g/cm^3.

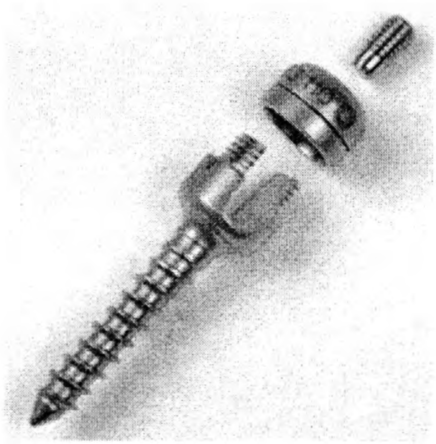

Figure 5. Physical model of pedicle screw

2.4 FEM model of lumbar vertebra L_5

2D geometric model of vertebra, shown in Figure 4, is used as basis to generate mesh of plane linear isoparametric elements with nodes at

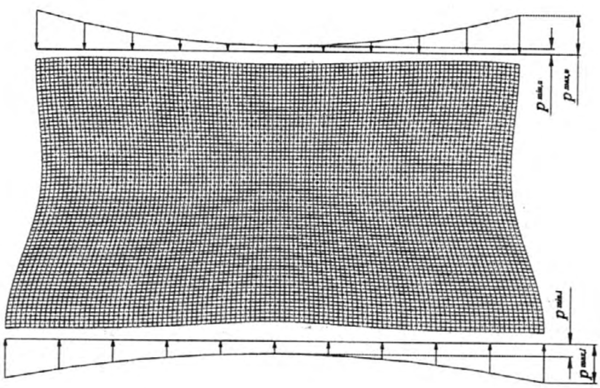

Figure 6. 2D FEM model of vertebra L_5 before implantation

each corner (11). Generated mesh is consisted of 110x70 finite elements, which makes structure with 15762 degrees of freedom (dof).

Load of vertebral body section, which is caused by every days activities, of young person is shown in Figure 6. According to research that is conducted by Xinghua et al. (7) shape of load distribution is symmetrically concave parabola, and total magnitude of vertical load is 117.3 N.

Parameters of load distribution of upper and lower vertebral body endplate were chosen so that shape and magnitude of vertebral load from mentioned research would be satisfied. Values of load distributions parameters are: $p_{max,u}=4.8$ N/mm^2, $p_{min,u}=1.6$ N/mm^2, $p_{max,l}=4.525$ N/mm^2 and $p_{max,u}=1.325$ N/mm^2.

3. RESULTS

Obtained distributions of bone tissue density and Von Mises stress which is result of simulated bone remodeling of lumbar vertebra L_5 before implantation is shown in Figures 7 and 8. Process was monitored until bone remodeling was finished in each finite element.

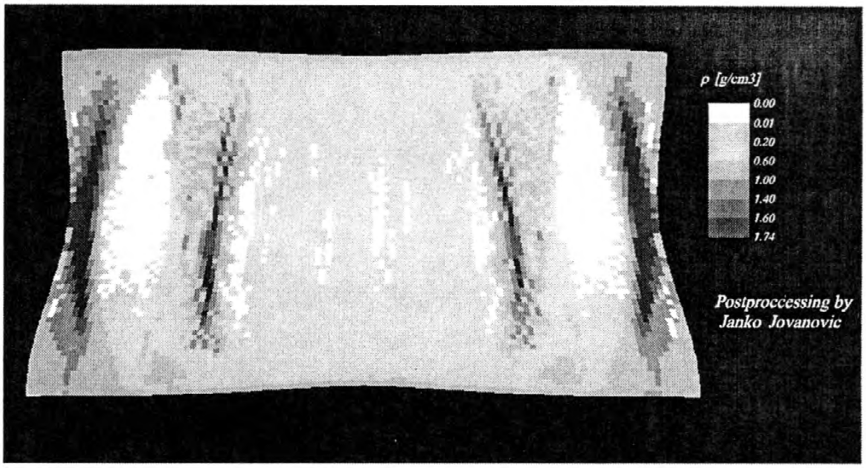

Figure 7. Distribution of bone tissue density before implantation

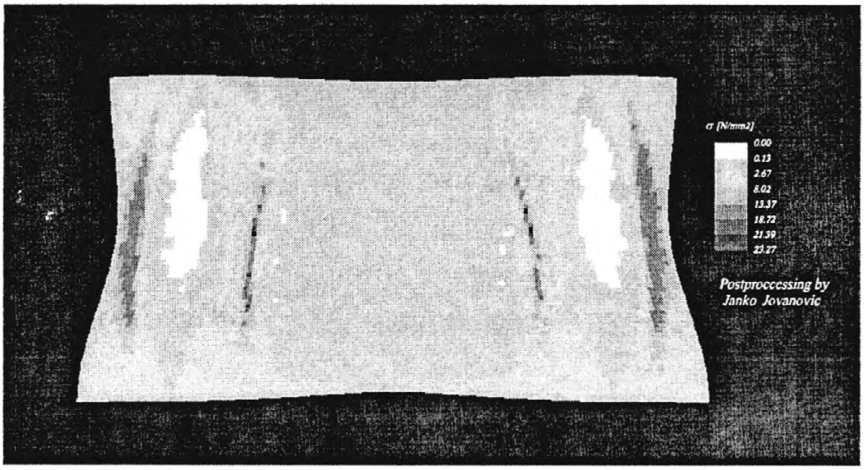

Figure 8. Distribution of Von Mises stress in bone tissue before implantation

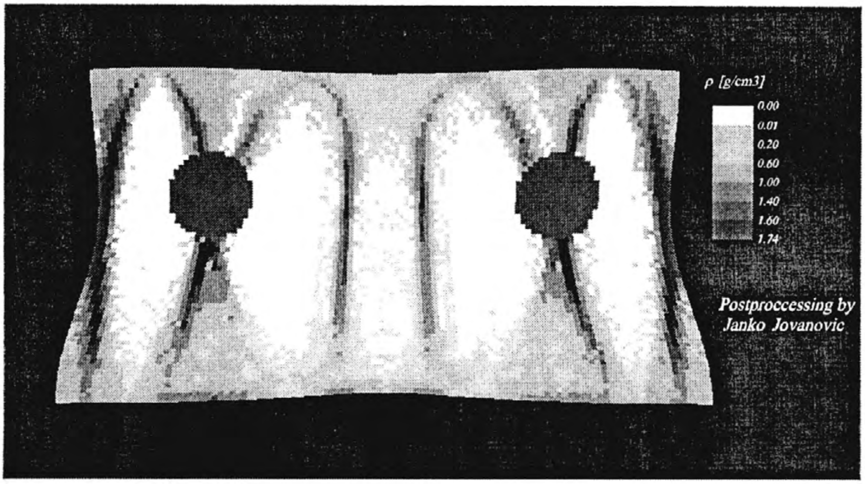

Figure 9. Distribution of bone tissue density after implantation

Post-operative implanted conditions caused by pedicle screws implantation initiate bone remodeling causing bone tissue density redistribution. 2D FEM model, shown in Figure 6, involving cross section of implanted pedicle screws is used for simulation of bone remodeling initiated by implantation. Obtained distribution of bone tissue density which is result of simulated bone remodeling of lumbar vertebra L_5 after implantation is shown in Figure 9.

Adaptive process before and after implantation ended after 170 and 139 iterations respectively, during which change of vertebral mass was investigated, as well as objective function as indicator of convergence of bone remodeling. Obtained dependencies of ratio between current m and initial mass of vertebra m_0, as well as objective function F during bone remodeling are shown in Figures 10 ÷13.

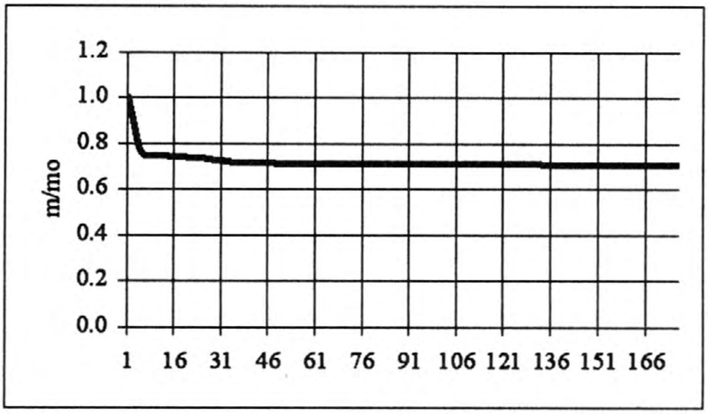

Figure 10. Mass change during bone remodeling before implantation

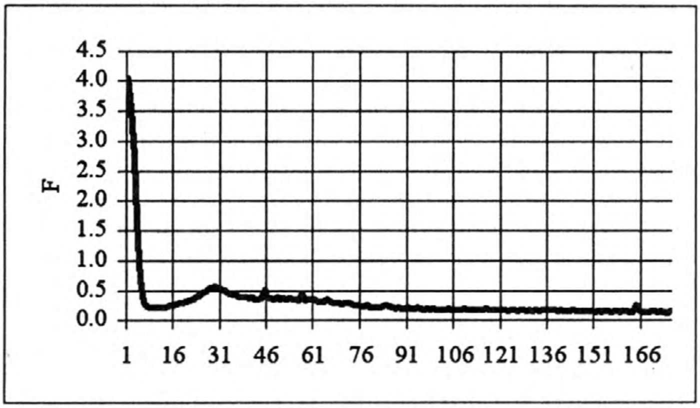

Figure 11. Objective function change during bone remodeling before implantation

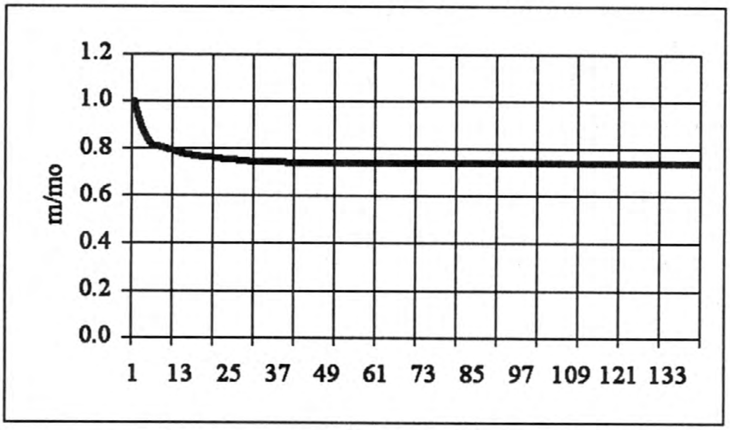

Figure 12. Mass change during bone remodeling after implantation

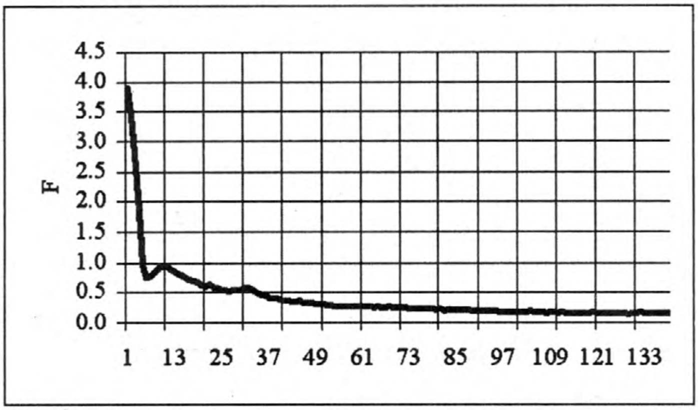

Figure 13. Objective function change during bone remodeling after implantation

4. CONCLUSION

Regarding the fact that used FEM model was based on number of assumptions and simplifications, distribution of bone tissue density obtained according to bone remodeling model is surprisingly similar to reality. Model of vertebra is two-dimensional, relationship between Young's modulus and density is expresed by crude approximation. Choice of density of strain energy, as mechanical stimulus of bone remodeling, was also more or less based on the fact that it is an easily interpretable physical scalar which is related to stress and strain.

Determinantion of bone tissue distribution based on bone remodeling in pre and post-implanted conditions enables analysis of bone tissue stress distributions before and after implantation as well as after implantation of different types of implants. Being familiar with bone tissue stress distribution in post-implanted conditions is of crucial importance for implant designe. It makes possible chosing optimal shapes, dimensions and materials for implants in order to avoid post-implanted troubles caused by disadvantegous stress distribution that could lead to bone fracture during post-operative period.

Analysis of influence of different types of hip-implants on change of mechanical load of proximal femur and bone remodeling caused by that change was object of research conducted by Wiade et al. at 2003 (9). This research experimentaly and theoreticaly proved that different adaptive process caused by post-operative implanted conditions of different types of hip-implants was cause of less clinical succes of Muller-Curved implant compared to Lubinus SPII implant.

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