Acta of Bioengineering and Biomechanics
Vol. 8, No. 2, 2006

Influence of remodelling, stimulating factor selection on bone density distribution in pelvic bone model

ANNA DĄBROWSKA-TKACZYK, MAREK PAWLIKOWSKI

Institute of Mechanics and Design, Warsaw University of Technology

The pelvic bone was considered to be the part of hip joint in the case of loading corresponding to the conditions of stance phase during gait cycle. For that load the model of bone adaptation processes was analyzed. Computer simulations were performed for different stimulating factors in bone remodelling equation. The following stimulators were considered: Huber von Misses stress, principal stresses and strain energy density. Bone density distributions were compared and analyzed. MIMICS and ABAQUS software were applied in order to determine non-homogeneous bone material properties and characteristic, to calculate stresses and strains and to simulate bone adaptation procedure.

1. Introduction

Computer-aided calculations are indispensable in the "custom design" process of total hip joint arthroplasty (THA). The most important task is to investigate the reasons of failures leading to revision surgery. Bone density is the physical factor that determines changes in the bone structure. The tool commonly used is numerical modelling based on FEM method, but the results of calculations are greatly dependent on modelling assumptions, i.e., geometrical and material structure and boundary conditions.

The main goal of this work was to investigate the results of the simulation of remodelling processes, depending on various stimulating factors. The object of simulation was pelvic bone model, obtained by reconstruction bone structure *in situ*. The procedure of processing data from CT images to non-homogeneous 3D numerical model was performed using CT data of young man's pelvic bone and software: MIMICS and MIMICS-FEA and ABAQUS. Numerical calculations, applying realistic bones properties and boundary conditions, allowed the stress and strain energy density to be obtained. Principal stress components, Huber von Misses stress and strain energy density were applied as the stimulating factors in remodelling model.

Bone density distributions for all the cases of remodelling model were calculated and analyzed, showing the presumed areas of structure strengthening or weakening.

2. Numerical analysis of adaptation processes in non-homogeneous model of pelvic bone

2.1. Transformation of CT data into 3D model of non-homogeneous material structure

The CT images were processed [3] in the MIMICS software (figure 1). The contour lines and then the surface patches of the pelvic bone were obtained. In the second step, using the MIMICS FEA to process the Hounsfield units into apparent density (ρ_a) and then the Young's modulus distribution, eventually the non-homogeneous model was obtained.

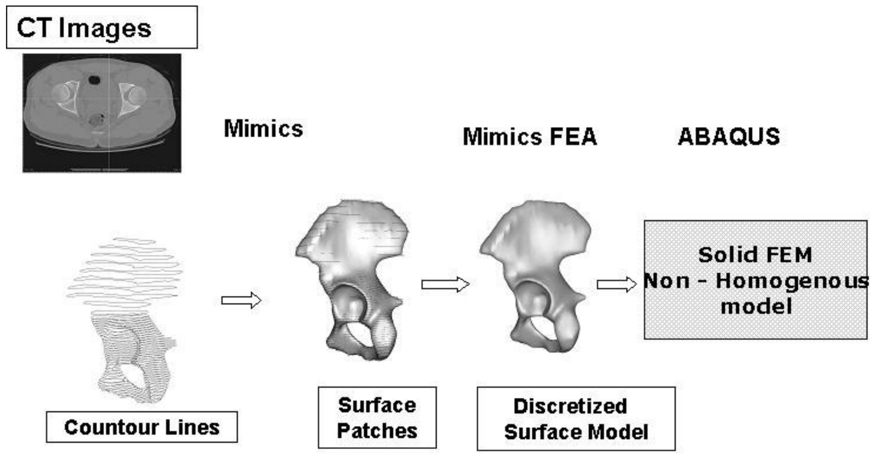

Fig. 1. The procedure of solid FEM non-homogeneous model creation

The values of HU are different for tissues of various densities, hence the bone is shown in MIMICS as non-homogeneous structure. In MIMICS, it is possible to calculate the physical density of bone on the basis of HU values by using an empirical equation [4] relating the two values. In our analysis, we used the following equation:

$$\rho_a = 0.001067 \cdot HU + 0.131, \tag{1}$$

where ρ_a is the bone apparent density.

In order to define non-homogeneous material properties of the bone models we made use [6] of a very well known equation (2) expressing the Young's modulus E as a function of apparent density:

$$E = 4249 \cdot \rho_a^{\ 3} . \tag{2}$$

The non-uniform model of pelvis created in the MIMICS software was calculated in ABAQUS finite element system. The non-uniformity of Young's modulus results from realistic bone density in the pelvis.

2.2. Modelling assumptions

The numerical model of pelvic bone, whose construction is described above, consisted of 5546 nodes and 22790 tetrahedral elements. In the numerical calculations, the forces of muscle actuators, corresponding to stance phase of gait cycle [4], [5], were taken into account (figure 2 B, C). Interaction between acetabulum cup and femur head has been modelled by means of the reaction force in hip joint, distributed along the upper part of the acetabulum (figure 2C). The X, Y and Z components of the reaction force fin the considered period of walking cycle (30–40%) were equalled to 532 N, 273 N and 1715 N, respectively [2]. Interaction of sacral bone was simulated by application of the upper body weight in the sacroiliac joint area (figure 2A). In the same area, the degrees of freedom in x and y directions were fixed. In the pubic symphasis area (figure 2A), the displacements in x and z directions were fixed.

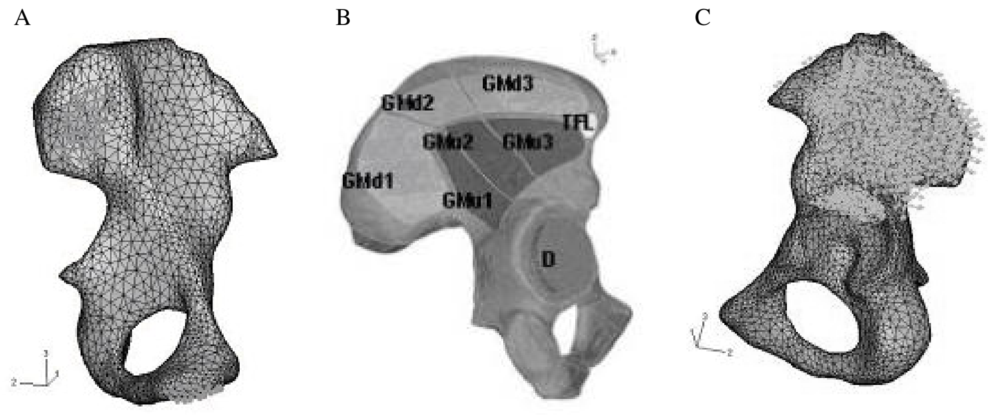

Fig. 2. Displacement and loading boundary conditions

The numerical analyses were performed by means of finite element method (FEM) using ABAQUS software. Thus, the analyses consisted mainly in simulation of bone remodelling phenomenon.

2.3. Functional adaptation of the bone procedure

Mathematically the phenomenon is described [2], [3] as bone density change with time:

$$\frac{d\rho_a}{dt} = \begin{cases} B(S-(1+s)S_0), & S>(1+s)S_0, \\ 0, & (1-s)S_0 \le S \le (1+s)S_0, \\ B(S-(1-s)S_0), & S<(1-s)S_0, \end{cases} \tag{3}$$

where: ρ_a – the bone apparent density, B – the remodelling rate constant, S – the remodelling stimulating signal, s – the half-width of the so-called "dead zone" interval, S_0 – the reference stimulating signal.

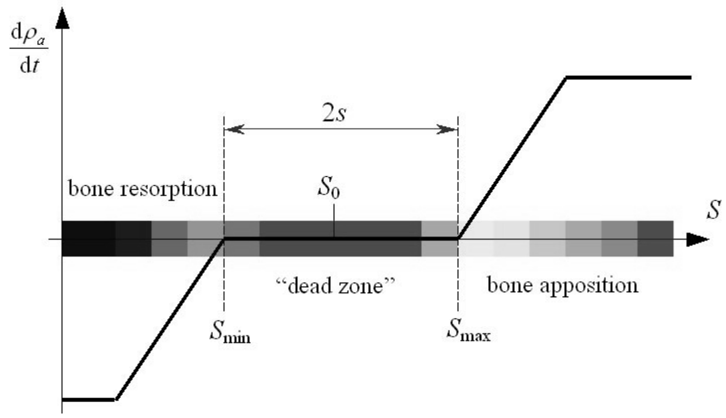

Fig. 3. Graphical representation of remodelling model

The graphical representation of equation (3) is shown in figure 3. Depending on the actual value of the stimulating signal S, the bone density can decrease (bone resorption), increase (bone apposition) or remain the same ("dead zone"). Equation (3) was applied to simulate bone adaptation.

Strain energy density (SED) and Huber von Misses stress were selected as the stimulating signal. Also principal stresses were applied as the stimulating signal. In such a way, we could determine the bone density distribution in the three principal directions. This consequently allows creation of the orthotropic model of bone and orthotropic model of the remodelling phenomenon.

2.4. Results of calculations

At the beginning, the geometrical model of pelvic bone with density distribution on the surfaces and cross-section of bone is presented in figure 4. These are the results of the transformation of CT images. The areas of maximum values of density concentration at pelvic bone surfaces correspond to their anatomical structure, but in the section across the bone, the external layer of cortical bone can be seen.

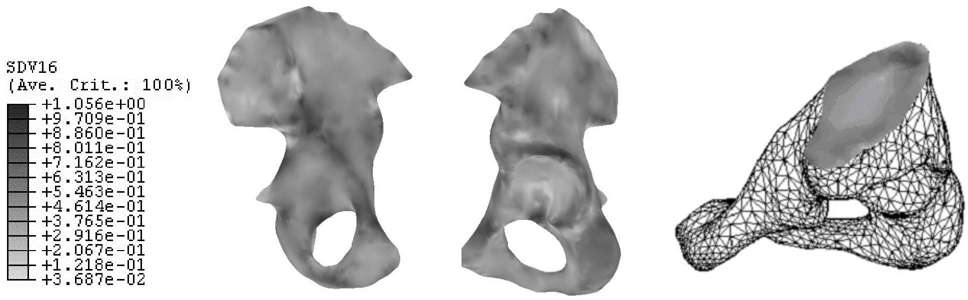

Fig. 4. Normal pelvic bone geometrical and material structure

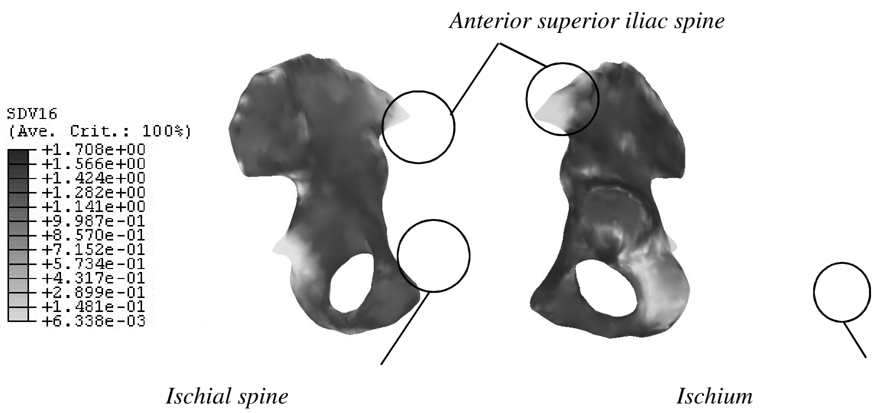

Fig. 5. Bone density distribution after remodelling procedure with SED stimulating factor

Bone density distributions in pelvic bone model after adaptation procedure, obtained for the constant load assumed, are presented in figures 5–7. Results in figure 5 represent the phenomenon, where the strain energy density (SED) is the stimulating signal. Huber von Mises stress as the stimulating signal yields different remodelling results (figure 6). However, in both density distributions, certain similarities can be observed, e.g., bone resorption in the regions where in the real pelvis the bone tissue exists. These regions are the ischial spine, the anterior superior iliac spine and the ischium (figures 5–7). This discrepancy between the numerical bone distributions and the real pelvic bone structure results from the fact that some of the muscles that attach to the regions were not taken into account in the numerical model of pelvis.

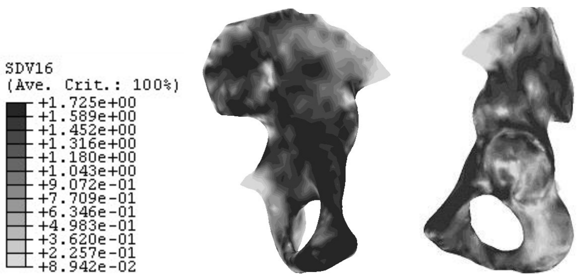

Fig. 6. Bone density distributions after remodelling procedure with Huber von Misses stress stimulating factor

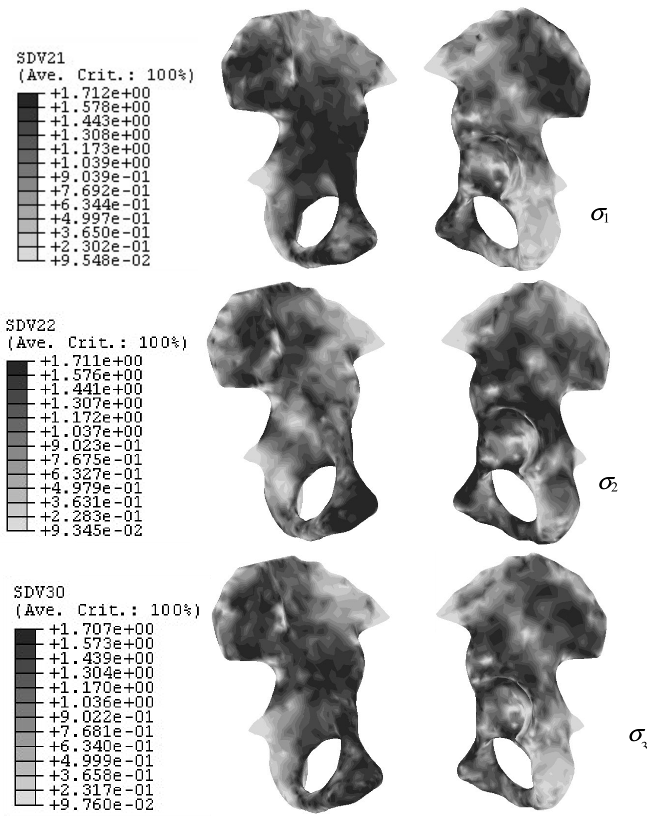

Fig. 7. Bone density distributions after remodelling procedure stimulated by principal stresses: σ_1, σ_2, σ_3

The same can be observed in figure 7, where bone density distributions in pelvis model obtained for maximal, medium and minimal principal stresses as stimulating signals, respectively, are shown.

3. Discussion and conclusions

Based on the results shown above, several important factors can be distinguished. The first one is non-homogeneous model of bone structure *in situ* with density distribution changing in the bone sections, depending on the physical properties of the bone. The second one consists in the changes of bone density on the surfaces of the bone influenced by remodelling simulation procedure. The results, shown in figures 5–7, allow selection of the areas of increased and decreased density corresponding to presumable bone strengthening or atrophy. The areas of strengthening correspond roughly to physical structure of pelvic bone, but the areas of density decrease greatly depend on model assumptions, especially on force boundary conditions. Only the main muscles were considered, whose activity was measured by EMG analysis. The forces of passive soft tissue structures (ligaments) were neglected. Density distribution in an initial model of pelvic bone *in situ* depends on the accuracy of evaluating the borders of layers in CT images. The soft tissue covering the bone *in situ* makes the procedure difficult. The principal stresses applied as stimulating factors in remodelling procedure allow us to observe the density changes in principal directions and to calculate Young's modulus in those directions. The results will be used to create a model of material orthotropic properties.

The results of investigation in this stage do answer exactly the question about the values of changes, but they determine the directions of parameter changes and the directions of further investigations.

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