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Influence of the fixation region of a press–fit hip endoprosthesis on the stress–strain state of the “bone–implant” system

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ABSTRACT

Although significant progress has been made in the development of total hip replacement, behaviour of the femoral component of an endoprosthesis in relation to the type of its fixation in the bone is still not fully understood. In this paper, behaviour of the femoral bone and the stem prosthesis is studied taking into account different types of prosthesis fixation in the medullary canal of the femur under the action of functional loads. For an analysis, a three-dimensional model of a femur has been developed based on the results of a computed tomography. The stress–strain state governing behaviour of the femoral bone and the stem prosthesis has been estimated with the use of the finite element method (FEM).

The FEM analysis has shown that for the diaphyseal fixation, the area of contact between the surface of the endoprosthesis and the bone is insufficient and leads to large stresses in the implant accompanied by stress concentration in the distal femur. An increase in the area of contact between the implant and the bone raises the stiffness of the "bone–implant" system, which, in turn, reduces stresses in the implant. The applied metaphyseal-type fixation yielded an improvement of results regarding behaviour of the femoral bone and the stem prosthesis. Namely, the study yielded the distribution of stress in the bone similar to the physiological stress state.

1. Introduction

Numerous causes of the hip joint impairment can be listed. The most common disorder of the hip joint is a pathological destruction of a joint cartilage, called osteoarthritis. Congenital deformation of the hip joint, accompanied by damage, contributes to the destruction of both the bone and the cartilage. This leads to pain and immobilisation of the hip joint. Nowadays, hip replacement is one of the most effective methods of treatment of hip joint diseases, which enables to both save the patient from pain and, simultaneously, restore full static–dynamic functions of the joint [1–3]. According to the USA analytical services, an increase of 180% in the number of total hip replacements will be observed by 2030 [4].

However, hip replacement is also accompanied with negative consequences which may occur about 10 years [5] after installation of the hip joint endoprosthesis (Fig. 1). In the case of excessive concentration of stresses on small areas of the femur, an increase in the density and volume of the bone tissue is observed, i.e. the cortical hypertrophy arises [6,7]. Factors which increase the risk of femur fractures are patient's reduced bone density, defects and cracks of the femur, and also the peculiarities of installing the endoprosthesis in the bone [8,9]. In the case of exclusion of any volumes of bone structures from the process of transferring the loads, atrophy and lysis of the bone tissue is possible [10]. Moreover, in patients who have osteolysis of the bone surrounding the implant, osteoporosis, or allergic reactions to metal, loosening and instability of endoprosthesis components may occur [11]. If the limit of elasticity or durability of materials used to manufacture implants is exceeded, plastic deformation or destruction of the prosthesis is likely to happen [12].

In the process of designing and constructing implants, efforts should be aimed at solving the afore-mentioned problems and developing fundamentally new designs of implants. These new designs of structures must be based on the data of biomechanical research, application of new materials and modern manufacturing methods [13–15]. Taking these elements into consideration ensures creation of high-quality implants being able to maintain their functional properties for a long time. One of the most important stages in the development of endoprosthesis is biomechanical rationale for performance and reliability of implants [16–19].

The efficiency of the “bone–implant” biomechanical system is

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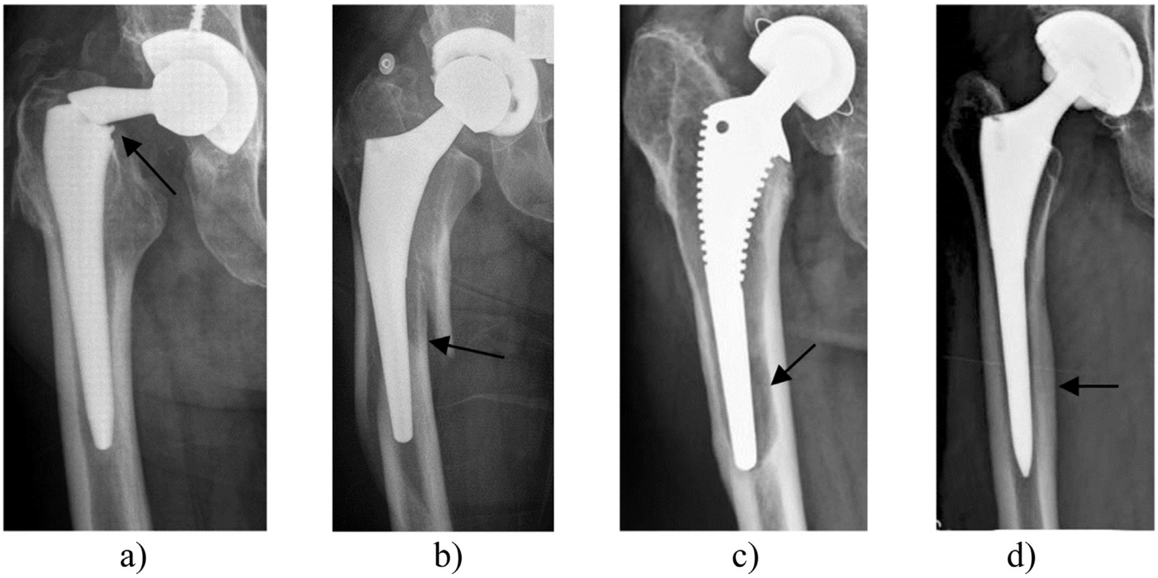

Fig. 1. Radiographs of the main complications following the hip replacement surgery (in the late post-operative period): (a) fracture of the stem prosthesis; (b) fracture of the femur; (c) loosening and instability of the femoral component; (d) cortical hypertrophy.

defined by the condition of the stress–strain state and mechanical behaviour of each element of the system under the action of functional loads. If in certain areas of the bone or the implant the functional load causes stress which exceeds some value (e.g., tensile strength, fatigue limit, etc.), then a destruction or plastic deformation of one or more components occurs, which leads to partial or complete loss of functionality of the whole system. It usually happens among patients who have an overly active lifestyle or suffer from osteoporosis and/or overweight. It may also happen in the case of subsurface voids or inclusions (or both), which could increase the stress two or even three times. Moreover, such condition may be caused by local weakening, incorrect attachment of the stem during surgery, or incorrect choice of the size of the prosthesis, etc. [20].

Currently, one of the most effective and informative methods of research of problems related to biomechanics is the method of numerical simulation, i.e. the finite element method (FEM). Thanks to the FEM, it is possible to avoid difficulties associated with the use of analytical methods for calculation of the stress–strain state of biomechanical systems and, most importantly, to get results with high accuracy [21–23]. In general, numerous approaches simulating bone behaviour and the process of adaptation to the applied loads have been already developed and some authors proposed techniques to reduce stress shielding effect [24–27]. However, in the literature, there have been no studies combining the FEA, used to analyze the influence of the region of fixation of a press–fit hip endoprosthesis on the state of the “bone–implant” system, with a bone remodelling model. The following research provides a numerical analysis of the stress-strain state of the femur and a tapered stem, considering different types of stem fixation in the medullary canal of the femur under the action of functional loads. The analysis is used to determine the best conditions for long-term functioning of the "bone–implant" system, what will lead to the improvement of results of the surgery.

2. Methods

For the analysis, a three-dimensional model of the femur has been developed based on the results of a computed tomography (CT) (Fig. 2).

For this study, frozen cadaver bones were used. Bones were scanned at the Dnipropetrovsk State Medical Academy. Scanning was conducted using an AQUILION RXL 16 (Toshiba Medical Systems) 16-slice CT scanner. DICOM images were obtained with a 0.5–mm slice thickness. In the first step, CT images of the femur have been acquired for subsequent segmentation of the object. The CT data were processed using MIMICS (Materialise, N.V. – Belgium) software.

The mechanical characteristics of the femur have been found by calculating the analytical dependences between the Hounsfield units (HU) obtained from the analysis of computed tomograms (Fig. 2). The Hounsfield units determine the dependence between the radiographic density of the femur tissue, presented in arbitrary units [28,29], the actual bone density ρ (g/cm^3), and the elastic modulus E (MPa) [30,31]. In our study, the relationship for density ρ =1+7.185 E−4· HU [29] and for the dependence between the elastic modulus and the density E =−388.8+5925·ρ have been used. Poisson's ratio has been assumed to be equal to 0.3 for the whole analyzed bone [32].

For the relationship between bone adaptation to mechanical loading and different types of endoprosthesis fixation, the strain energy density (SED) model of bone remodelling has been used [33]. The model employs a ratio of the strain energy density (U) and the local apparent density (ρ) as a mechanical signal that advances and controls the bone remodelling process. It defines the remodelling stimulus as:

$$S = \frac{U}{\rho}. \tag{1}$$

And U can be expressed in terms of stresses and strains as follows:

$$U = \frac{1}{2}\sigma_{ij}\varepsilon_{ij} \tag{2}$$

where σ_{ij} and ε_{ij} are the local stress and strain tensors, respectively.

The signal responsible for bone resorption or formation is therefore assumed to be the difference between the actual stimulus level S and some reference stimulus k. It is also assumed that there exists a form of a "lazy zone" (or “dead zone”) [34] where no remodelling takes place. The limits of bone remodelling are *(1+s)k* and *(1-s)k*. Thus, the remodelling rule can be expressed as the following set of conditional equations:

$$\frac{d\rho}{\mathrm{dt}}\begin{cases} \mathrm{B}\left[\frac{\mathrm{U}}{\rho}-(1-\mathrm{s})\mathrm{k}\right], \mathrm{if}\frac{\mathrm{U}}{\rho} < (1-\mathrm{s})\mathrm{k}, \\ 0,\ \mathrm{if}(1-\mathrm{s})\mathrm{k} \le \quad \frac{\mathrm{U}}{\rho} \le (1+\mathrm{s})\mathrm{k}, \\ \mathrm{B}\left[\frac{\mathrm{U}}{\rho}-(1+\mathrm{s})\mathrm{k}\right], \mathrm{if}\frac{\mathrm{U}}{\rho} > (1+\mathrm{s})\mathrm{k}, \end{cases} \tag{3}$$

where ρ is the local apparent density, B is a constant which char-