Hip joint implants — survey of numerical modeling*

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Abstract

In the paper the discussion of the modeling of hip joint implants is presented. The historical attempts to implant construction finally led to complex solutions. To date almost all experiences were gained by clinic observation and rather test and trial method than systematic research. Numerical simulations were limited to simple stress analysis. All the phenomena responsible for the damage in the treated joint are not totally explained. The complexity of the problem was presented in the paper. The stress concentration in the region of rigid inclusion is pointed as one of the mechanical destructive factors. The change of the stem form and material rigidity could reduce extremal stresses 10–20%.

Streszczenie

W pracy omówiono modelowanie implantów stawu biodrowego. Historyczne próby ich konstruowania doprowadziły do bardzo złożonych rozwiązań. Dotąd niemal wszystkie doświadczenia zdobywano w obserwacjach klinicznych i próbach, a nie w wyniku systematycznych badań. Numeryczne symulacje ograniczano do prostej analizy stany naprężenia. Większość zjawisk odpowiedzialnych za uszkodzenia badanych stawów nie jest w pełni wyjaśniona. W pracy zwrócono uwagę na złożoność zagadnienia. Za jeden z czynników niszczących uznano koncentracje naprężeń w strefach sztywnych wtrąceń. Zmiana kształtu trzpienia i sztywności materiału pozwala zmniejszyć ekstremalne naprężenia o 10-20%.

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1 Introduction

Nowadays, there are over 100 million people aged over 50 in the USA and Europe. One of the consequences of aging is deterioration of bone quality; also, gradual loss of bone mass is observed, starting from 30 – 40 years. One of the frequent consequences of aging of the skeletal system are degenerative changes often occurring in the hip and the knee, which may lead to even serious handicap.

Hip joint, classified as "ball and socket" type, is the second biggest joint in the human organism. Acetabulum, placed in the pelvic bone, is one of the hip joint components, it is round – shaped and deepened by surrounding labrum. Inside the acetabulum there are lunar - shaped cartilage faces, oriented inferiorly and anteriorly. Another main part of the hip joint is the round – shaped femoral head, half – way seated in the acetabulum. Stabilization of the joint comes from its anatomical shape, and, moreover, is provided by strong capsular and ligamentar system. The main role is played by three ligaments: iliofemoral, pubofemoral and ischiofemoral, which together stabilize the joint and prevent excessive motion. The "ball and socket" construction of the hip joint together with numerous and powerful muscles provide a broad motion range in every plane. According to Pauwels, the force acting in the hip joint during gait is four times bigger than the body weight [39]. The mean force acting on femoral head surface is 16 kg/cm^2 [39]. In affected the hip joint, because of stress muscle tension, pressure on femoral head may increase even 10 times.

In a normal hip, the joint motion is as follows: flexion 130° extension (hyperextension) 10° external rotation 45° internal rotation 40° abduction 80° adduction 40°. Every pathology affecting soft tissues and joint surfaces leads to gradual loss of motion range, frequently not noticed by the patient, even for a long time.

Arthritic changes of the hip joint can be classified into two big groups. The first group consists of essential arthroses, where etiology is unknown. The second, bigger group, are secondary arthroses, caused frequently by congenital deformities, specific and non – specific inflammations, previous injuries, static disorders. Rheumatic diseases cause destructive changes in joints, and, therefore, are considered as a separate group of arthroses. Clinically, both degenerative and destructive hip disease manifest similarity, starting with recurrent, and – in more advanced cases – continuous pain of groin, buttock, and, commonly, of knee on the affected side. Usually, pain is accompanied by – differently intensified – restriction of motion, deteriorating the patient's mobility. If conservative treatment, consisted of pharmacotherapy, physio – and kinesitherapy, is unsuccessful, in the case of advanced destructive and degenerative changes of hip joint total arthroplasty is choice therapy. The purpose of hip prosthesis is providing painless walking and useful range of the operated joint. Joint replacement could be considered as one of the most efficient methods of treatment, because during just one operating procedure restoration of painless function of the hip can be obtained.

Attempts of replacement of diseased body tissues with artificial elements have a long history. In 1565 Petronius used a golden plate to cover palatoschisis. In 1902 surgeon Mr. Robert Jones covered a stump of femoral neck with golden plate [43]. Surgeons Campbell, Mac Ausland, Murphy, Bauer attempted to use different materials to cover destroyed femoral head, like fascia lata, the wall of a pig's urinary bladder, celluloid foil. Unfortunately, the results after such operations were only temporary satisfying, and – from the clinical point of view – totally disappointing. Experiments with covering femoral head performed by M. N. Smith–Petersen, started in 1923, seemed much more effective. Initially, he made his "cup" from glass, then from viscaloid, pyrex, bakelite, and, finally, in 1938, from vitallium alloy (Co–Cr–Mo) [4, 19]. In 1938 Philip Wiles undertook an unsuccessful attempt of replacing acetabulum and femoral neck with artificial elements made of stainless steel [43]. In the forties Austin T. Moore produced his hemiprosthesis, which was meant to replace only the femoral head and neck, and was fixed with its stem in femoral marrow cavity. At the same time, the Judet brothers made their acrylate prosthesis, replacing only the femoral head, to be fixed with its metal stem in femoral neck [21]. Also Adam Gruca, a pioneer in Polish orthopaedics, produced and implanted own total hip prosthesis in 1949; unfortunately, poor results of this clinical trial discouraged him from further experiments. In Europe, in the fifties, intensive efforts were undertaken in order to produce total hip prosthesis. Pioneers in these works were J.Charnley, Watson, G.K. McKee and Farrar. McKee and Farrar created total prosthesis, which both components – acetabulum and femoral stem with its neck and head - were made from metal fixed with bone cement. Bone cement was first used by Kieera and Jansen in 1951 [43]. In 1956, a Russian surgeon, Sivash, constructed his own model of cementless hip prosthesis, which head was permanently connected with acetabulum, acting as the articulated link [21, 32]. Incontrovertibly, greatest contribution in developing of total hip arthroplasty had Sir John Charnley. In 1958 he first used "low friction arthroplasty". After unsuccessful experience with teflon acetabular implants, he started using high–density polyethylene. In order to minimize friction between components of prosthesis, Charnley used a small, 22 mm diameter prosthesis head, what consequently caused increase of pressure, exerted by prosthesis head on artificial acetabulum. Moreover, Charnley introduced the usage of bone cement to orthopaedic practice. Based on his original model, numerous modifications of his prosthesis were created, differing in stem's shapes and lengths, head diameters and acetabular shapes. In the seventies, the double–cup shaped Wagner prosthesis was invented. The advantage of this model was that only a small resection of bone ends was needed; unfortunately late results of using the Wagner prosthesis were not satisfying [46, 42]. Because bone cement seemed to be the weakest point in the artificial hip, numerous efforts were undertaken to improve prosthesis – bone connection. In 1976 Asmutz introduced blocking of femoral shaft with a bone stopper; achieving 30% increase of pressure during cementing of prosthesis [41]. In the late seventies Harris started to use “cement gun”, obtaining more regular distribution of cement around the implant. Further step in cementing was “vacuum mixing system”, what significantly increased the resistance of cement against strains and fatigue. The eighties brought recapitulation of arthroplasty, and many critical opinions. The rate of loosening was 15–21% [25]. Such high rate of poor results pushed numerous surgeons to search for new solutions in cementless arthroplasty. P.A. Ring presented his cementless prosthesis in 1964; P. Boutin and H. Mittelmeier created their prostheses in the seventies. Mittelmeier’s model had a ceramic, threaded acetabulum, and a ceramic head, placed on steel stem. In 1975 Lord and Boncel used their own cementless prosthesis, which consisted of metal acetabulum with polyethylene insert and porous stem. In early eighties Zweymuller and Parhofer–Monch team used “press – fit” cementless prostheses. According to this method, properly chosen stem of prosthesis precisely fills the femoral shaft. Nowadays, cementless prostheses are widely used, stabilized by the ingrowing of the bone into implant micropores. It is proved that for best results micropores on prosthesis surface should be 150–200 μm [11, 20, 27]. During the production of a implant, in order to obtain porous surface, the implant is vacuum–covered with pure titanium molecules, with use of 20 000°C gas flame [1, 6]. Also, porous surface should cover 20–40% of total surface of prosthesis stem. The process of bone ingrowing into prosthesis pores resembles bone healing. During this process, it is necessary to obtain “mechanical silence” in implant’s vicinity; otherwise instead of bone, fibrous tissue will be produced, which does not provide proper stabilization of the implant. The bone–implant gap should not exceed 100 μm. Thus, in order to obtain initial stability, the surgeon must ideally fit the implant into its bed.

Another method of obtaining the best bone – prosthesis contact is covering the implant with hydroxyapatite, a basic unorganic bone substance. Hydroxyapatite will be gradually built into the bone; the optimal thickness of its layer is 60–100 μm. This method was first used by Geesink [22].

It is observed, that in cementless prostheses loosening of stem occurs more frequently than of acetabulum. This pushed numerous surgeons, especially in the USA, to introduce the use of the so-called “hybrid” prostheses, consisting of cementless acetabulum and cemented stem [16, 18].

Basing on above presented data, contemporary prostheses can be classified into one of three types. The first group are cemented prostheses, being longest in use and, thus, best known. In this type, stabilization of the implant is obtained by cementing it with acrylate cement. Transmission of stress occurs in three main contacting surfaces; first on head – acetabulum, second on bone–cement; third, the weakest point, on cement-implant. In this method friction wear and superficial fatigue are dealt with [7]. Friction wear occurs if loose fragments of prosthesis material get between joint surfaces, causing their gradual destruction. Those fragments could origin in all substances used in the production of prosthesis; i.e. bone cement, polyethylene and metal. It is proved that loose polyethylene fragments play an important role in the enhancement of bone resorption and osteolysis of femoral neck [33]. In order to minimize contact stresses, the surgeon should choose a prosthesis which fits bone beds best; also best implant – bone contact must be achieved and the thickest polyethylene acetabulum should be used. Contacting of different materials, with substantially different rigidity brings more mechanical problems. For example, titanium alloys rigidity is 114000 MPa, cancellous bone is 500 MPa, cortical bone is 15000 MPa, bone cement is 2200 MPa. Among prosthesis components, cement is the most compliant to superficial fatigue, unlike metal, representing high resistance against fatigue, and bone, being able to autorepair. Another disadvantage of cement seems to be its high temperature of polimerization (45–70°C), causing marginal necrosis of bone; moreover, both the patient and the operating team are subject of toxic influence of cement.

Depending on material used in the construction of prosthesis, they can be classified into three types:

- polyethylene acatabulum, metal or ceramic head;
- metal acetabulum and head;
- metal acetabulum, polyethylene head (contemporary not in use).

Construction of cementless prosthesis has two weak points. The first one concerns material resistance against wear, what may be connected with prosthesis geometry. The second problem are biological reactions caused by metal, polyethylene, ceramics; also process of bone remodelling around the implant, possible allergic reactions, metal corrosion could lead to further problems. Wear of prosthesis causes release of small material fragments, causing reaction against foreign body. Growing granulation is mainly responsible for loosening of implant. Corrosion occurs regardless the type of metal used, causing several reactions known as metallosis; metal can also cause metabolic, bacteriological, immunological and oncogenic effects. It is proved that alergy against nickel, chromium and cobalt occurs in approx. 10% of population [44].

It seems that hydroxyapatite porous prostheses, recently introduced into clinical practice, meet the best criteria to proper and long–lasting implant osteointegration.

In contrast to the number of prostheses types, the choice of prosthesis is still depending on own surgeon's experience, patient's age and status of bone tissue.

Currently, intensive efforts are undertaken on improvement of hip prosthesis, and the number of questions, still awaiting answers, indicates that the ideal prosthesis will not be developed soon.

2 Modeling of hip joint implants

The finite element method was introduces in the orthopeadic literature in 1972 [14]. The early applications concerned stress analysis of entire bones, following traditional interest of orthopaedic and anatomical science. For the first time irregularity of structure, material properties and load could be considered. A greater part of the FEM analyses was focussed on joint replacement, in particular the total hip prosthesis. Early models were two dimensional ones. 3–D effects were taken into account for example by using of two finite element layers, one of each simulated the implant and the other the bone surrounding implant, or by using axi–symmetric forms [29].

Realistic 3–D models were more precise in their geometrical description. However, complexity if the analysis in such a case disabled the simple investigation of the pure phenomenon chosen, as for example decohesion propagation, cracks, friction etc.

It was mentioned in the first chapter that forces which act on the joint considerably exceed the body weight. The nonaxial load and a couple involved by muscles must be carried by the bone. Walking or jumping increase the force amplitudes and make the hip joint one of the most loaded part of the human skeleton.

The loading system is different in each function of the active life and may vary in the individual case. Several papers were devoted to the time function of loading forces during walking, running and climbing the stairs [8, 9, 10, 24, 2].

2.1 Interface modeling

Vroemen et al. [45] developed 2–D FE model of the Wagner femoral cup. The cup/bone connection is either assumed fixed or governed by non–linear stick–slip conditions with dry friction and loosening on tension. The variable hip load was applied. Sensitivity analysis were carried out to evaluate the effects of bone and material properties, cup stiffness and the friction coefficient in the case of loose cup. Finally, based on the histological findings, bone resorption and soft–tissue interposition was simulated by reducing the stiffness of the elements in the appropriate areas in subsequent steps.

Simple gap can be introduced [15] between the head and acetabulum. Two separate FE systems with small gap between them were coupled by non–penetration condition. Nowadays the stem loosening is the most frequent type of the implant damage. Both the stem of the implant and the acetabulum can be loosened. In Fig. 1 the implanted right

Table 1: Stress concentration in uniform and mixed material (dimensionless units).

stem profile	mixed material	uniform material
thin	2.66	2.56
thick	2.55	2.42

acetabulum took the more horizontal position (upper and lower photography) which is the result of loosening.

There is no doubt that the stress concentrations that occur in the implant–bone interface (even in the case of implant fixed by the cement coat) accelerate the fatigue and destructive processes. The simple numerical analysis of the bone with the cement–less implant loaded with the force distributed on the implant head shows the increased stresses, much more higher than both in the bone and steel implant. Simple calculations carried on by authors prove it. In the Fig. 2 σ_x and σ_y in dimensionless units are depicted. Simple shape optimization shows the form of the implant stem for which higher and lower stress concentration is obtained (Fig. 3). As the stress concentration indicator the difference r of maximum value of σ_{max} and minimum value of σ_{min} was taken. For the left figure the r=2.65 while for the right case r=2.47. Unfortunately, the full optimization is complex. The most important data is the load characteristics. The change of force that subject the system changes the result. Rigid inclusion increases stress concentration. In Table 1 the stress concentration value for uniform and mixed material is presented. Both thick and thin prosthesis stem was considered.

In [28] Huiskes describes several numerical test compared with experimental stress analysis. The steel bar fixed in the bone with the cement coat was subjected transversally, as a cantilever. Diagrams of radial σ_r and shear σ_{rt} stresses exhibit concentrations at both ends of the bar that exceeds more the 10 times the stresses in the internal part of the bar.

3 Heat generation

Numerous papers deal with the temperature problems in the case of prosthesis–cement–bone system. The heat of polymerization is conduced through the cement mass, the implant and the bone. The concentration of heat streams can increase locally the temperature of the bone and damage the bone irreversibly [35]. For example Huiskes et al. [30] take into account the plane or axisymmetric structure. The analysis of the acetabular cup fixation in hip joint replacement determined the osteocyte necrosis and vascular damage zones. He uses the experimentally established relation [37], which determines the

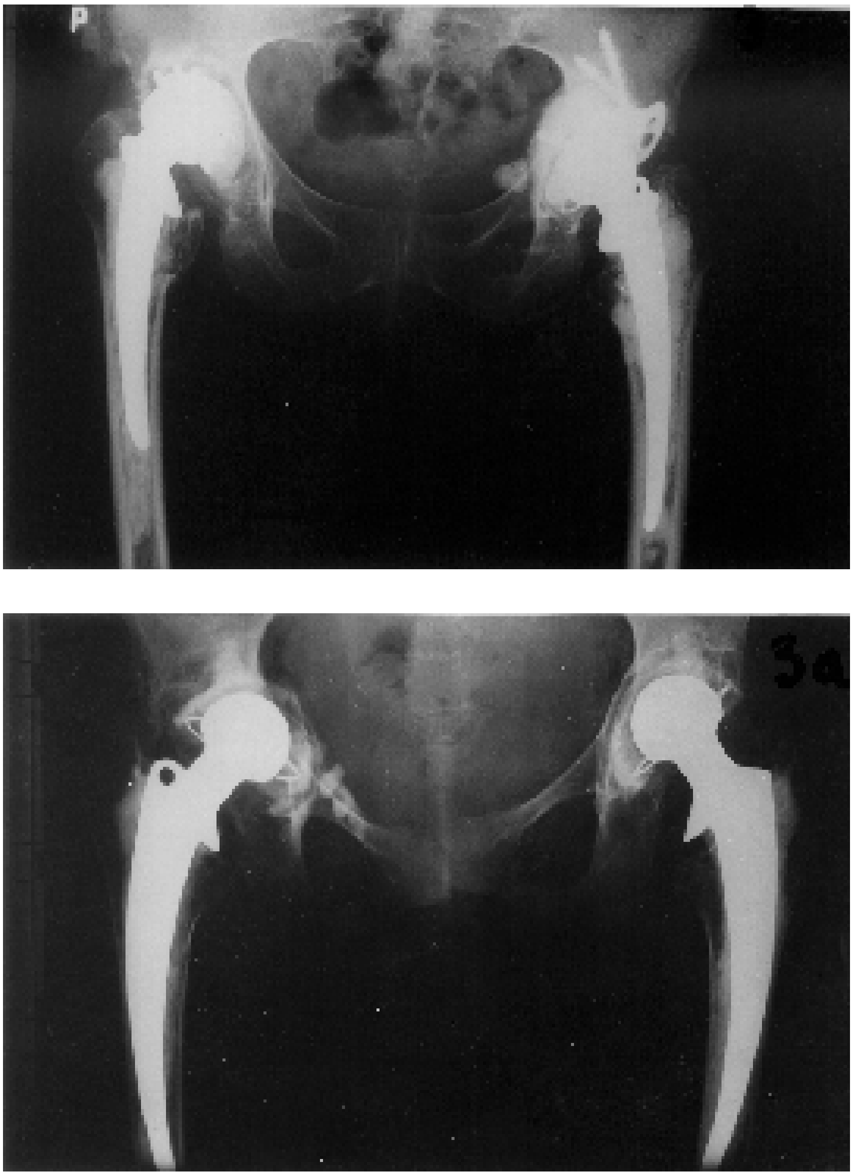

Figure 1: Loosening of the acetabulum in implanted hip joint.

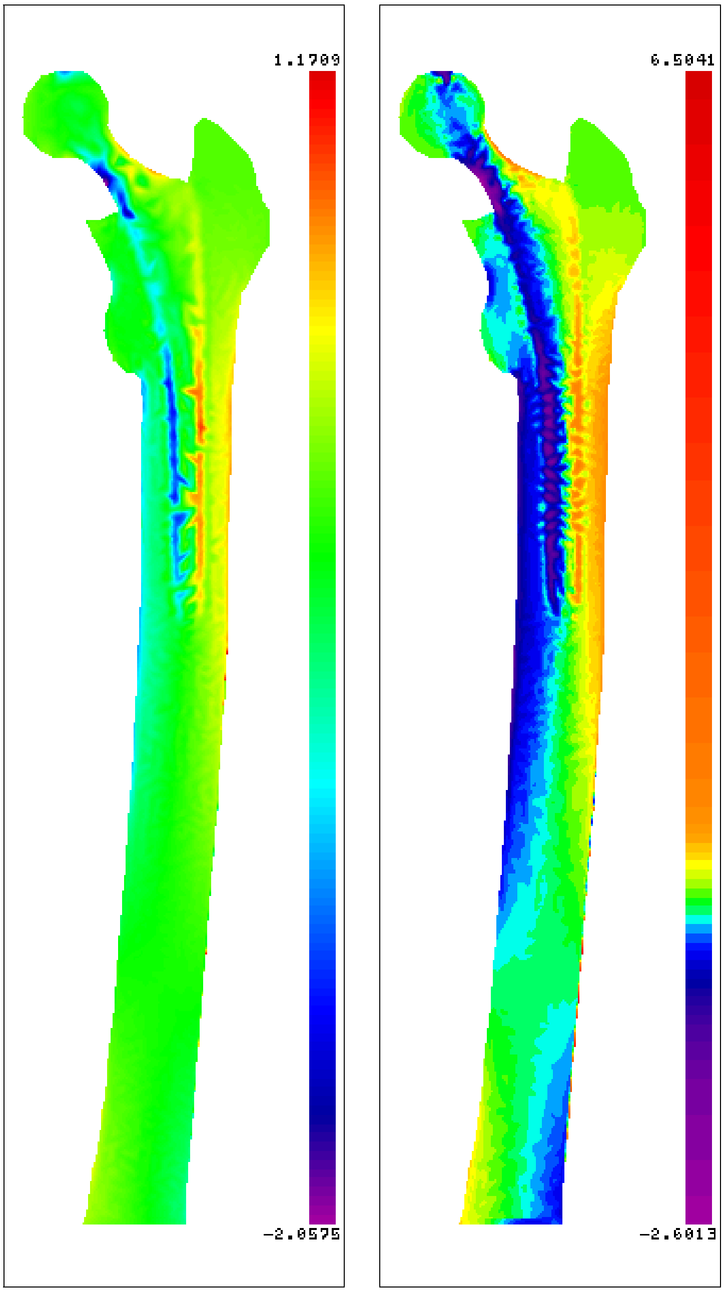

Figure 2: Stress concentration around the stem.

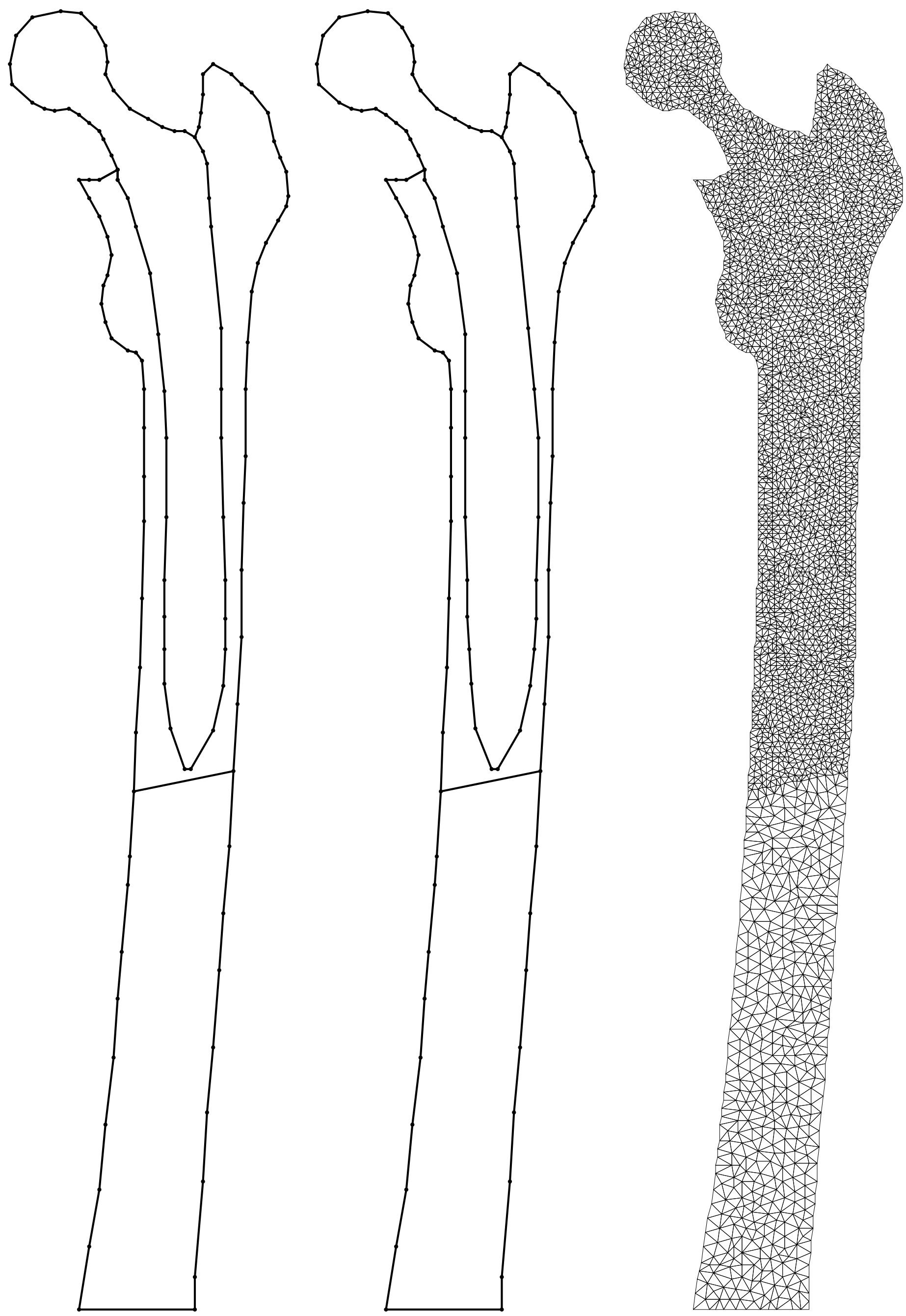

Figure 3: Stem shape with higher and lower stress concentration and finite element mesh.

bone–cell necrosis

$$\Delta T^{7.143} t_e > 3.55 \times 10^{10} \tag{1}$$

where Δt=37^{o}C is the bone temperature and t_e is the exposure time in seconds.

The decrease of the temperature level can be achieved by decreasing the cement layer, cooling the operation region prior and during the cement insertion, decreasing the polymerization rate, increasing the polymer powder to monomer liquid ratio. The penetration of the cement in the trabecular voids increases significantly the chance of bone necrosis.

More advanced study was presented in [3]. The kinetics of the polymerization was investigated. The exposure time t necessary to reach thermal bone necrosis at given temperature u (bone necrosis time) is given in the following form

$$t(u) = M\exp\left(\frac{\mu}{R(u-310)}\right), \quad u > 310 \tag{2}$$

The factor M and the activation energy μ are identified by the linear regression of the data presented in [37]. If η denotes the thermal damage measure ($0 \le \eta \le 1$), under non–isothermal conditions its value can be estimated as an integral of the fraction of exposure time at assigned temperatures

$$\eta(r,z) = \int_0^T \frac{\mathrm{d}t}{t(u(r,z,t))}, \quad (r,z) \in \Omega. \tag{3}$$

The authors test several hypotheses in the aim to verify the real state and to reduce the heat propagation. They assume perfect bonding, perfect contact in some portions of the cement/bone interface and air isolation in the rest if the border and rubber membrane separating cement and bone. The last case shows positive effects.

Numerous authors deal with the numerical analysis of stresses in the selected medical cases. For example in [34] 3–dimensional models of proximal femur are analyzed. The goal is to test the correspondence between the model and *in vitro* strain gage data and failure loads. Nonlinear material properties for the cortical and trabecular bone are included. While there was poor correspondence between strain gage data and model prediction, there was excellent agreement between the *in vitro* failure data and the linear model, especially using the von Mises effective strain failure criterion.

Recent papers give advanced analysis of contact region. However, they give only quantitative results for the known or expected phenomena. For example in [36] the humero–ulnar joint was treated numerically. Incongruity in the joint contact surfaces involves bicentric distribution of contact stresses. Tensile stresses occur during the pressing of the humerus to the subchondral bone. Since the curvature if the humerus is lower then the curvature of the subchondral bone, all the result could be simply predicted. In [5] the authors assume a rigid–body spring model for joint contact pressure distribution. The method presented in [31] was adopted. Two kinds of springs were considered: normal and share. The elbow joint was investigated for the joint pressure and dislocation conditions.

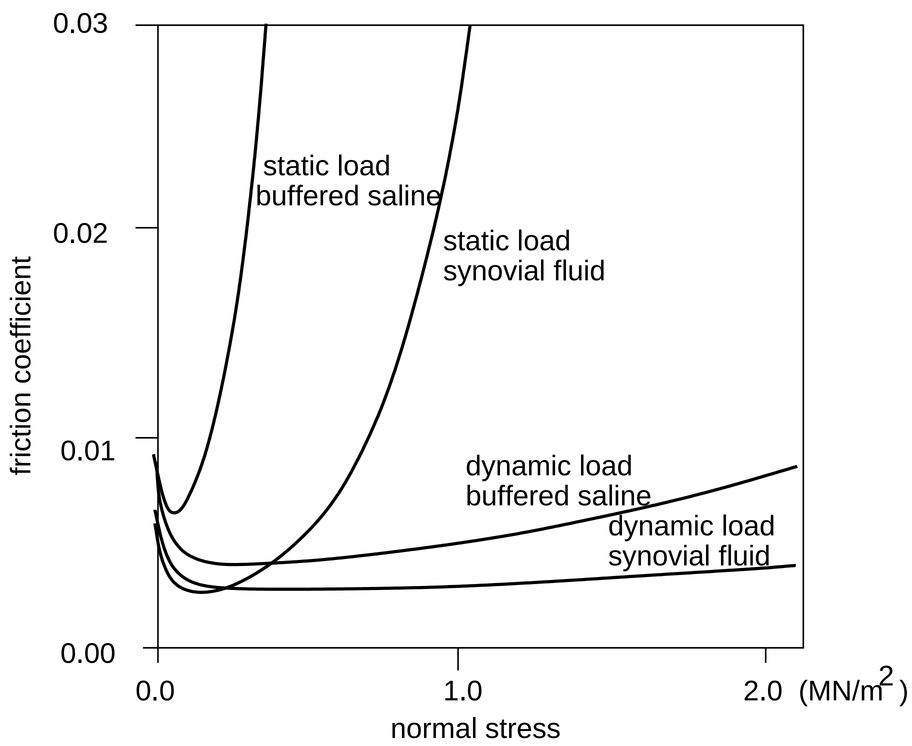

Figure 4: Friction coefficients in synovial joints.

4 Friction and wear

A group of papers consider the friction problem in the human joint. The experiments show extremely low friction coefficient (literature data presented in Fig. 4). However, there is no agreement on what mechanism operates in the heavily loaded joints. The experimental investigation is described in [12]. The interesting discussion is presented in [26]. In the case of the presence of thin layer of lubricant between the solids, when the considerable load must be carried, either high pressure or a big area of the contact is required. In such a case the thickness of a film is below 0.01 mm. The author's calculations show that the squeeze film mechanism dominates in the human joint sliding, especially for loads of short duration. The mechanism in which full film of fluid between the solids is generated by conventional rolling/sliding elastohydrodynamic lubrication has a lower chance to occur.

The comparison of friction coefficient measured and presented in the literature is given in Table 2.

New materials are introduced to fill the contact face in the joint. The interesting one is the hydrogel [23]. It is a soft, porous–permeable polymer that readily absorb water and maintain its shape. Hydrogels have high water contents (40–60 %), lower Young modulus than polyethylene and lower friction coefficient. They have been proposed as a bearing surface material for cushion joint for example in [13]. The cushion joint bearing or cush-

Table 2: Friction coefficients for articular cartilage in synovial joints.

joint tested	friction coefficient
human ankle	0.005–0.02
porcine shoulder	0.02–0.35
canine ankle	0.005–0.01
human hip	0.01–0.04
bovine shoulder	0.002–0.03

ion joint aims to reproduce the lubrication conditions of the natural diartroidal joints to a grater extent than the current generation of replacement joints, which use harder, nonporous bearing surface materials. The aim is to increase the area of contact between two joint surfaces and in the same time to lower the contact stresses and maintain the fluid film lubrication in wider range of loading conditions. What is more, with hydrogels under compression fluid is released, similarly to the behavior of natural articular cartilage. Experimental study of hydrogels concerns friction and lubrication [17], wear and biocompatibility [38].

Wear of bearings is a phenomenon of boundary degradation involving progressive loss of bearing substances from the body as a result of mechanical action. The two conventional types of wear are fatigue wear and interfacial wear. Fatigue wear is independent of the lubrication occuring at the surface of bearings. It is generated by cyclic strain growth. This internal failure within the tissue was observed in a form of collagen fiber buckling and loosening of the normally tight collagen network. If the rate of damage exceeds that by which the cartilage cells may regenerate the tissue, an accumulation of fatigue micro damage will occur that may lead to bulk tissue failure.

The gap that opens oscillatory in the interface metal–cement or cement–bone acts as a pump: the wear debris are led into the gap in opening stage and are hardly removed in closing stage 5. The crack once produced by the tensile forces is never closed firmly for the reason of biological reaction. The propagated loosening of the prosthesis in this case is called aseptic loosening. Radiologically a so–called "lucent line" is observed around the implant. The interesting microanalysis is presented in [40]. After the implantation the metal prosthesis is surrounded by bone remodeled in its orientation and structure so that trebeculae are formed parallel to the surface of the implant. There exist two explanations of the nature of the interface between bone and implant: some authors describe the direct contact between bone and biomaterial, others postulate the existence of a fibrous tissue layer. The experiences described in [40] prove both the mechanisms. Next to an implant which is aseptically loosened macrophages and fibroblasts are found. These cells form a

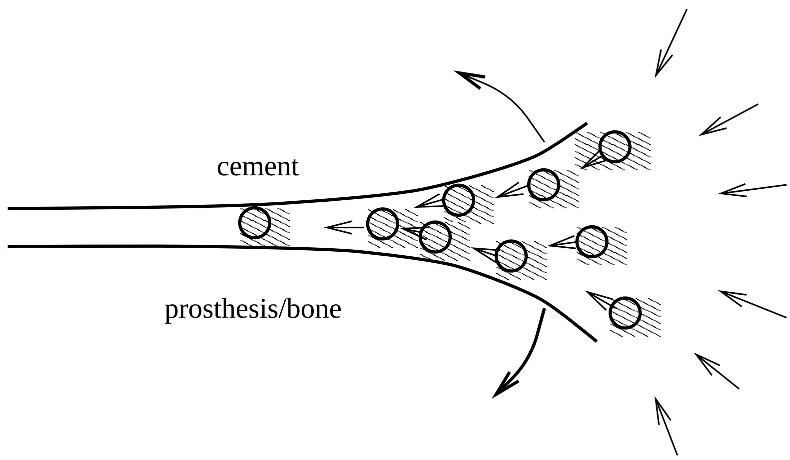

Figure 5: Debris flow into the cement–bone gap.

synovium–like structure at the surface of fibrous tissue. Detailed microscopic observation finally led to the following conclusions:

- polymethylmethacrylate (cement) particulate debris is a main cause of the foreign body reaction and subsequent bone loss,
- polymethylmethacrylate fragmentation is important in implant loosening,

4.1 Stress distribution analysis in healthy and diseased bones

Both 2 and 3 dimensional analysis carried on with the finite element method shows the distribution of stresses in the bone. Nowadays even complex form of the bone or implant–bone couple does not exhibit considerable difficulties. Even simple radiology enables to prepare the real geometry in a particular case. Tomography allows almost automatic passage from the human body to the finite element geometric model (Fig. 6). More, the X–ray pictures show the density change in parts of a bone and in the same way the lower relative load carrying capacity. The external contour can be simply determined in most parts. The difficulty appears in the contact region between the acetabulum located in the pelvis and the head of the bone. The internal contour determines the places where the density (i.e. X–ray transmission) has a predefined low value. Since bone density and bone mineralization change continuously and the bone is laced with voids, the internal contour is much more complex. If the contour line separate voids or liquid–like media, the precision has minor role. However, in the trabecular bone, where we have mostly the lattice structure, rigid enough with low average material density, the proper geometrical and material data are important.

Although for researches the realistic and precise model is not so important, in the case

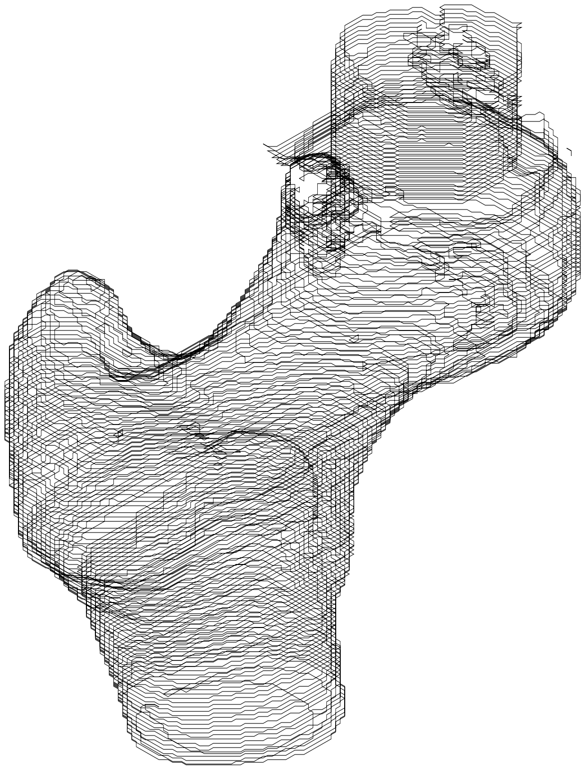

Figure 6: Contour lines of the hip joint obtained by the tomography.

of diseased bones both the geometry and material properties may differ considerably from the typical ones.

5 Conclusion

There are two main reasons of prosthesis loosening:

- biological, strongly depended on the individual predispositions, which should be investigated on the microscopic level; wear of the implant, caused both by the biological factors and mechanical factors (for example friction), migration of wear debris to the contact surface of bone, cement and steel, slacken rebuilding of the bone,
- mechanical, when the stress limits are exceed; it results in the fatigue, especially in the case of cyclic load, crack propagation and material fragmentation.

There exists limited contribution of mechanical research to the problem. The same concerns the microbiological field. The real simulation of all the phenomena and the authoritative answer to the question of loosening of prosthesis is open.

Figure 7: Hip joint implant.

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