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HIP JOINT IMPLANTS – SURVEY OF NUMERICAL MODELING

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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 have learned from clinic observation and rather tests and trial methods than systematic research. Numerical simulations have been limited to simple stress analysis. All the phenomena responsible for the damage in the treated joint are not fully explained. The complexity of the problem is 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 by $10 \div 20\%$.

Key words: biomechanics, hip joint, numerical modeling, FEM

1. Introduction

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

The 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 inferior and anterior cartilage surfaces. Another main part of the hip joint is the round-shaped femoral head, half-way seated in the acetabulum. Stabilization of the joint is ensured by its anatomical shape, and,additionally by a 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* structure of the hip joint together with numerous and powerful muscles provide a broad motion range in each plane. According to Pauwels, the force acting in the hip joint during gait is four times bigger than the body weight (Poitoud et al., 1991). The mean force acting on the femoral head surface is $16\,\mathrm{kg/cm^2}$ (Poitoud et al., 1991). In the affected hip joint, due to muscle tension, forces on the femoral head may increase even 10 times.

In a normal hip joint, the motion is as follows: flexion 130°, extension (hyperextension) 10°, external rotation 45°, internal rotation 40°, abduction 80°, adduction 40°. A pathology affecting soft tissues and joint surfaces leads to limits gradually the motion range, which is 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 etiology of which is unknown. The second, bigger group comprises secondary arthroses, caused frequently by congenital deformities, specific and non-specific inflammations, injuries, static disorders, etc. Rheumatic diseases cause destructive changes in joints, and. therefore, are considered as a separate group of arthroses. Clinically, both degenerative and destructive hip diseases are similar, starting with recurrently, and – in more advanced cases – continuous pain in groin, buttock, and, commonly, in the knee on the affected side. Usually, the pain is accompanied by limitations of motion of different intensities, which deteriorate patient's mobility. If a conservative treatment, consisting of pharmacotherapy, physio- and kinesitherapy is unsuccessful, in the case of advanced destructive and degenerative changes of the hip joint total arthroplasty is choisen of mobility. The purpose of hip prosthesis is to provide painless walking and a suitable range in the operated joint. The joint replacement could be considered as one of the most efficient methods of treatment, because during just one operative procedure painless function of the hip can be restored.

The attempts at replacement of diseased body tissues with artificial elements have had 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 a golden plate (Śmiłowicz, 1978). Surgeons Campbell, Mac Ausland, Murphy, Bauer attempted to use different materials to cover destroyed femoral heads; like, fascia lata, the wall of a pig's urinary bladder, celluloid foil. Unfortunately, the results of such operations were only temporary satisfying, and from the clinical point of view totally disappointing. Covering the femoral head performed by M.N. Smith-Petersen, with 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 (CoCrMo) (Amstutz et al., 1992; Eftekhar, 1993). In 1938 Philip Wiles undertook an unsuccessful attempt at replacing acetabulum and femoral neck with artificial elements made of stainless steel (Śmiłowicz, 1978). In the 1940s Austin T. Moore produced his hemiprosthesis, which was meant to replace only the femoral head and neck, and was fixed with its stem in the 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 the femoral neck (Garlicki and Kreczko, 1974). Also Adam Gruca, a pioneer in Polish orthopaedics, produced and implanted his own total hip prosthesis in 1949; unfortunately, poor results of this clinical trials discouraged him from further experiments.

In Europe, in the 1950s, intensive efforts were directed towards production of a total hip prosthesis. Pioneers in this field were J. Charnley, Watson, G.K. McKee and Farrar. McKee and Farrar created a total prosthesis, both components of which; i.e., acetabulum and femoral stem with its neck and head, were made from metal fixed with bone cement. The bone cement was first used by Kieera and Jansen in 1951 (Śmiłowicz, 1978). In 1956, a Russian surgeon, Sivash, constructed his own model of cementless hip prosthesis, head of which was permanently fastened with the acetabulum, acting as the articulated link (Garlicki and Kreczko, 1974, Kreon, 1992). Incontrovertibly, greatest contribution towards development of total hip arthroplasty made Sir John Charnley. In 1958 he first used *low friction arthroplasty*. After unsuccessful results with teflon acetabular implants, he started using high-density polyethylene. In order to minimize friction between the components of prosthesis, Charnley used a small, 22 mm in diameter prosthesis head, what consequently caused increase in pressure, exerted by the prosthesis head on the artificial acetabulum. Moreover, Charnley introduced the bone cement to orthopaedic practice. Basing on his original model, numerous modifications of his prosthesis were created, of different stem shapes and lengths, head diameters and acetabular shapes.

In the 1970s, the double-cup shaped Wagner prosthesis was invented. The advantage of this design was that only a small resection of the bone ends was needed; unfortunately the results of using the Wagner prosthesis were not satisfying (cf Wagner, 1978; Salamon et al., 1983). Since the bone cement seemed to be the weakest point in the artificial hip, numerous efforts were made to improve the prosthesis – bone fixing. In 1976 Asmutz introduced blocking of the femoral shaft with a bone stopper; achieving 30% increase in pressure during cementing of the prosthesis (Salamon and Ratomski, 1988). In the late 1970 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 cement strength and fatiguelike.

The 1980s brought recapitulation of arthroplasty, and many critical opinions. The rate of loosening was 15-21% (Halley and Wroblewski, 1986). 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 1970s. Mittelmeier's model had a ceramic, threaded acetabulum, and a ceramic head, placed on a steel stem. In 1975 Lord and Boncel used their own cementless prosthesis, which consisted of metal acetabulum with polyethylene insert and a porous stem. In early 1980s Zweymuller and Parhofer-Monch team used *press-fit* cementless prostheses. According to this method, the properly chosen stem of prosthesis precisely fills the femoral shaft. Nowadays, cementless prostheses are widely used, stabilized by the bone ingrowth into implant micropores. It is proved that for best results the micropores on prosthesis surface should be 150-200 μm (Bobyn et al., 1980; Gallante, 1971; Homsy, 1972). In order to obtain a porous surface the implant is in the production process vacuum-covered with pure titanium molecules using 20 000°C gas flame ([1]; Barun and Papp, 1993). The porous surface should cover 20-40% of total surface of the prosthesis stem. The process of bone ingrowth into prosthesis pores resembles bone healing. During this process, it is necessary to achieve *mechanical silence* into implant vicinity; otherwise instead of bone the 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 achieve the initial stability the surgeon must precisely 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 $60 \div 100\,\mu$m. This method was first used by Geesink (1990).

It is observed, that in cementless prostheses the stem loosening occurs more frequently than that 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 (Callaghan, 1992; Davey and Harris, 1989).

Basing on the above presented data, there are three types of the prostheses used nowadays. The first group consists of cemented prostheses, being in use for the longest time and, thus, best known. In this type, stabilization of the implant is achieved by cementing it with acrylate cement. Transmission of stress occurs in three main contacting surfaces; i.e., head-acetabulum, bone-cement, cement-implant (the weakest point). In this method the friction wear and superficial fatigue are dealt with (Będziński, 1997). The friction wear occurs if loose fragments of prosthesis material get between joint surfaces, causing their gradual destruction. Those fragments could origin from 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 the femoral neck (Lieberman et al., 1994). In order to minimize contact stresses, the surgeon should choose a prosthesis which fits bone beds best; also the best implant-bone contact must be ensured and the thickest polyethylene acetabulum should be used. The contact between different materials, with substantially, different rigidity, brings more mechanical problems. For example, the rigidity of titanium alloys is 114000 MPa, cancellous bone 500 MPa, cortical bone 15000 MPa,while for bone cement 2200 MPa. Among the prosthesis components, cement is the most compliant to superficial fatigue, revealing, unlike metal, high fatiguelife, while bone is able to autorepair. Another disadvantage of cement seems to be its high temperature of polimerization ($45 \div 70°C$) causing marginal necrosis of bone; moreover, both the patient and the operating team are subject of toxic influence of cement.

The prosthesis, depending on the material used, can be grouped as follows:

- Polyethylene acatabulum, metal or ceramic head
- Metal acetabulum and head
- Metal acetabulum, polyethylene head (not used nowadays).

The cementless prosthesis design has two weak points. The first one concerns material resistance to wear, what may be connected with prosthesis geometry. The second problem consists in biological reactions caused by metal, polyethylene, ceramics also the process of bone remodelling around the implant, possible allergic reactions, metal corrosion could lead to further problems. The prosthesis wear causes release of small material fragments causing reaction against a foreign body. Growing granulation is mainly responsible for the implant loosening. Corrosion occurs regardless of 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 the alergy to nickel, chromium and cobalt occurs in approximately 10% of population (Szulc and Ratomski, 1988).

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

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

Currently, intensive efforts are made to improve the hip prosthesis design, 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 introduced into orthopaedia in 1972 (Brekelmans, 1972). 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 the two dimensional ones. 3D 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 (Huiskes, 1986).

Realistic 3D models yielded more precise geometrical description. However, the analysis in such a case is complex and separate phenomena occuring; e.g., decohesion propagation, cracks, friction etc., cannot be investigated straightforward.

It was mentioned in the first chapter that the forces which act on the joint considerably exceed the body weight. The non-axial load and a couple involved by muscles must be carried by the bone. Walking or jumping increases the force amplitudes and makes the hip joint one of the most heavily 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 (Bergman et al., 1993, 1995; Guo et al., 1994; Alexander, 1995).

2.1. Interface modeling

Vroemen et al. (1986) developed a 2D FEM 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 under tension. The variable hip load was applied. Sensitivity analysis was carried out to evaluate the effects of bone and material properties, cup stiffness and the friction coefficient in the case of loose cup. Finally, basing 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.

A simple gap can be introduced (Brown and DiGioia, 1984) between the head and acetabulum. Two separate FEM systems with a small gap between them were coupled by the 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 acetabulum took a more horizontal position, 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 cementless 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 done by the authors prove it. In Fig.2 dimensionless σ_x and σ_y are depicted.

Simple shape optimization shows the form of the implant stem for which higher and lower stress concentrations are obtained (Fig.3). As the stress concentration indicator the difference r between 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 the system is subject to changes the result.

A rigid inclusion increases the stress concentration. In Table 1 the stress concentration value for uniform and mixed material is presented. Both thick and thin prosthesis stems were considered.

Table 1. Stress concentration indicators in uniform and mixed materials (dimensionless units)

Stem profile	Mixed material	Uniform material
thin	2.66	2.56
thick	2.55	2.42

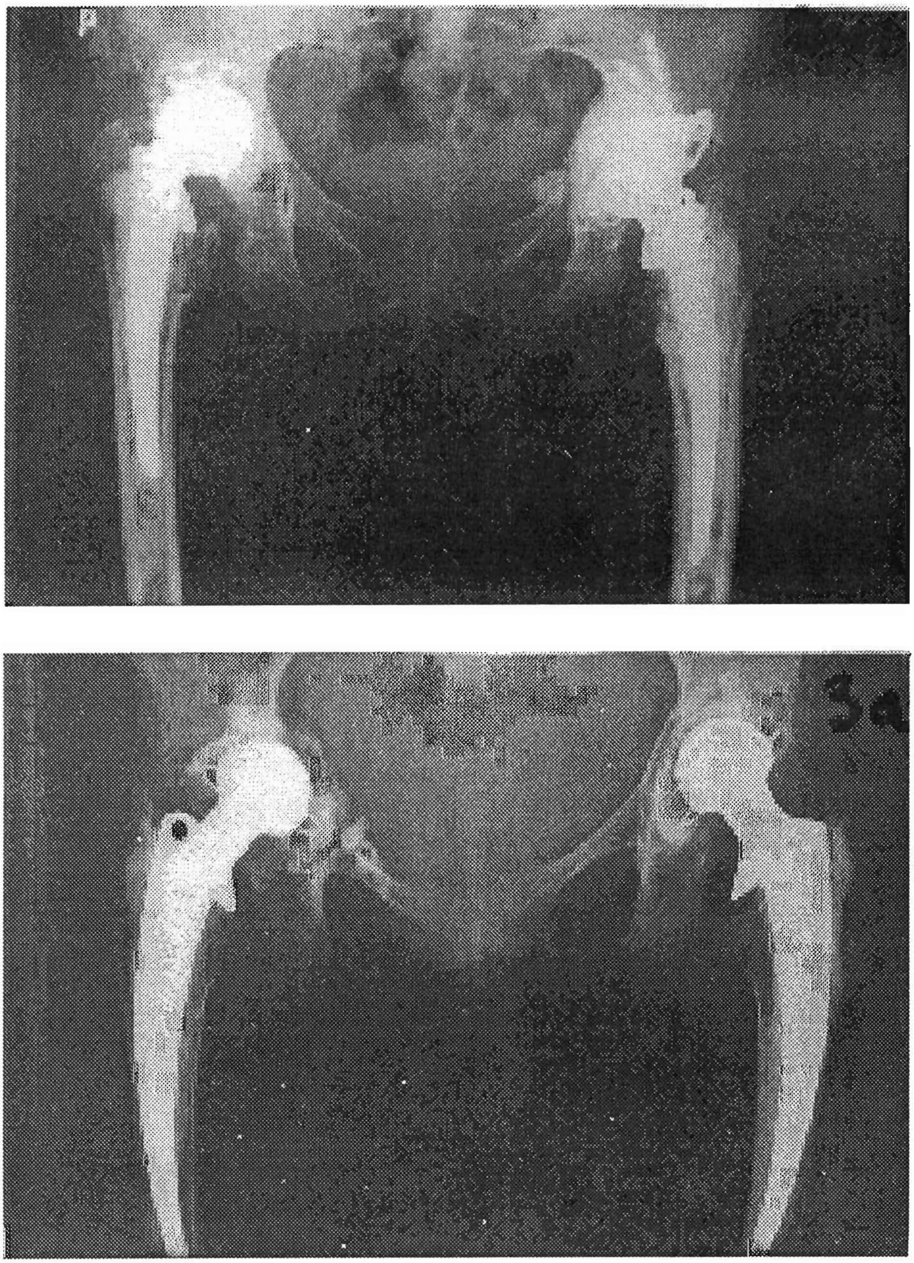

Fig. 1. Loosening of the acetabulum in the implanted hip joint

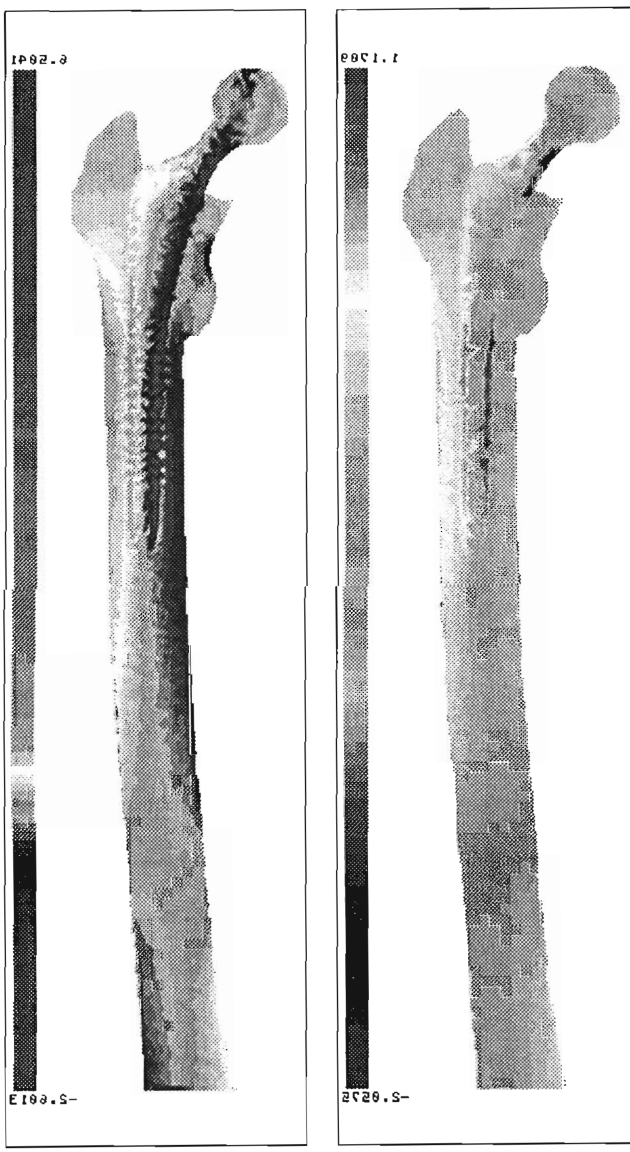

Fig. 2. Stress concentration around the stem

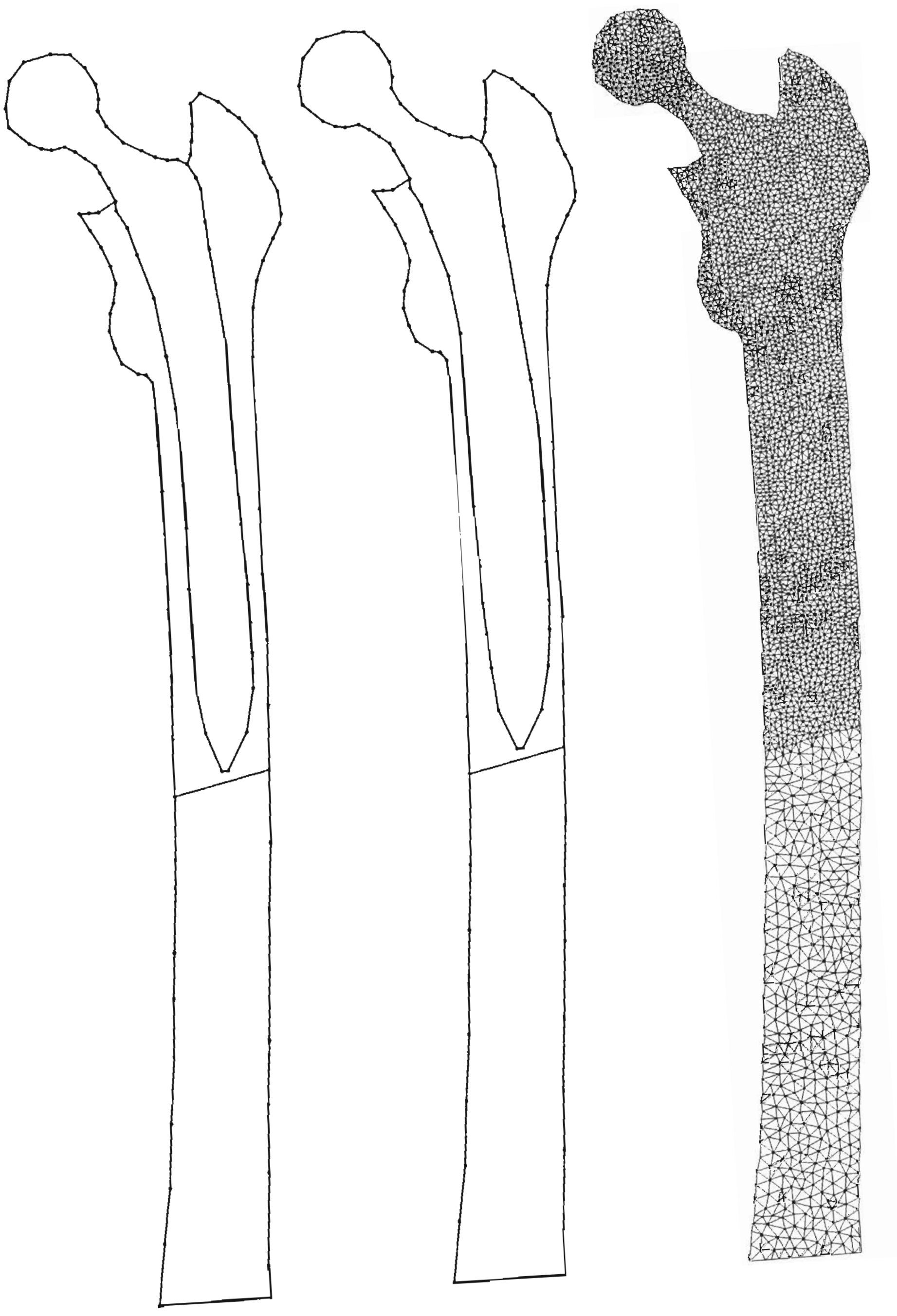

Fig. 3. Stem shape with higher and lower stress concentrations and the finite element mesh

Huiskes (1984) described several numerical tests compared with the stress experimental analysis. The steel bar fixed in the bone with the cement coat was loaded transversally, acting as a cantilever. Diagrams of radial σ_r and shear σ_{rt} stresses exhibit the concentrations at both ends of the bar, which exceed more 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 conducted through the cement mass, implant and bone. The concentration of heat flows can increase locally the temperature of the bone and damage the bone irreversibly (Mazullo et al., 1991). For example Huiskes et al. (1982) took 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 used the experimentally established relation (Moritz and Henriques, 1947), which determined the bone cell necrosis

$$\Delta T^{7.143} t_e > 3.55 \cdot 10^{10} \tag{3.1}$$

where $\Delta t = 37°\mathrm{C}$ is the bone temperature and t_e is the exposure time in seconds.

The temperature level can be lowered by thickening of the cement layer, cooling the operation region prior and during the cement insertion, reducing the polymerization rate, increasing the ratio between polymer powder and monomer liquid. The cement penetration into the trabecular voids increases significantly probability of bone necrosis.

More advanced study was presented by Mazullo et al. (1991). The kinetics of polymerization was investigated. The exposure time t necessary to reach thermal bone necrosis at a given temperature u (bone necrosis time) is given in the following form

$$t(u) = M \exp\Big(\frac{\mu}{R(u-310)}\Big) \qquad u > 310 \tag{3.2}$$

The factor M and the activation energy μ are identified by the linear regression of the data presented by Moritz and Henriques (1947). If η denotes the thermal damage measure $(0 \leqslant \eta \leqslant 1)$ under non-isothermal conditions its value can be estimated as an integral of the fraction of exposure time at assigned temperatures

$$\eta(r,z) = \int_0^T \frac{dt}{t\Big(u(r,z,t)\Big)} \qquad (r,z) \in \Omega \tag{3.3}$$

The authors verified several hypotheses 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 of the border and the rubber membrane separating cement and bone. The last case shows positive effects.

Numerous authors have dealt with the numerical analysis of stresses in selected medical cases. For example Lotz et al. (1991) analyzed 3D models of proximal femur. Nonlinear material properties of the cortical and trabecular bones were assumed to verify the coincidence of the model and *in vitro* strain gage data and failure loads. While there was poor correspondence between the strain gage data and model predictions, an excellent agreement was readed between the *in vitro* failure data and the linear model, especially when using the von Mises effective strain failure criterion.

Recent papers give advanced analysis of the contact region. However, they give only quantitative results for known or expected phenomena. For example Merz et al. (1997) the humero-ulnar joint was analysed numerically. Incorrect materials between the joint contact surfaces involves bicentric distribution of the contact stresses. Tensile stresses occur during the pressing of the humerus to the subchondral bone. Since the curvature of the humerus is lower then the curvature of the subchondral bone, all the result could be simply predicted. Himeno et al. (1990) assumed a rigid-body spring model for the joint contact pressure distribution. The method presented by Kawai and Takeuchi (1981) was adopted. Two kinds of springs were considered: normal and transversal. The elbow joint was investigated for the joint pressure and dislocation conditions.

4. Friction and wear

A group of papers present the friction problem in the human joint. The experiments show extremely low friction coefficient (see data in Fig.4).

However, there is no agreement on what mechanism works in heavily loaded joints. The experimental investigation was described by Bogacz and Ryczek (1997). The interesting discussion was presented by Higginson (1977). In the

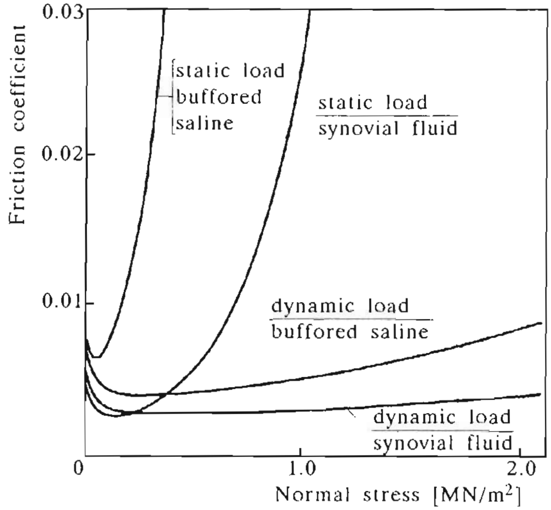

Fig. 4. Friction coefficients in synovial joints

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 probability to occur.

The comparison between the friction coefficients measured and presented in the literature is given in Table 2.

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

Joint tested	Friction coefficient
human ankle	$0.005 \div 0.02$
porcine shoulder	$0.020 \div 0.35$
canine ankle	$0.005 \div 0.01$
human hip	$0.010 \div 0.04$
bovine shoulder	$0.002 \div 0.03$

New materials are introduced to fill the contact surface in the joint. The interesting one is the hydrogel (cf Goldsmith and Clift, 1998). It is a soft, porous-permeable polymer that readily absorb water and maintain its shape. Hydrogels have high water contents $(40 \div 60\%)$, lower Young modulus than polyethylene and lower friction coefficient. They have been proposed as a bearing surface material for cushion joint, e.g. by Bray and Merril (1973). The cushion joint bearing or cushion 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, non-porous 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 a wider range of loading conditions. Moreover in hydrogels under compression the fluid is released, similarly to the behavior of natural articular cartilage. The experimental study of hydrogels concerns friction and lubrication (cf Caravia et al., 1993), wear and biocompatibility (cf Oka et al., 1990).

Wear of sliding surfaces is a phenomenon of boundary degradation involving progressive loss of lubricant from the body as a result of mechanical action. The two conventional types of wear are fatigue wear and interfacial wear. The fatigue wear is independent of the lubrication occuring at the bearing surfaces. 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 at which the cartilage cells still may regenerate the tissue, an accumulation of fatigue micro-damage will occur that may lead to bulk tissue failure.

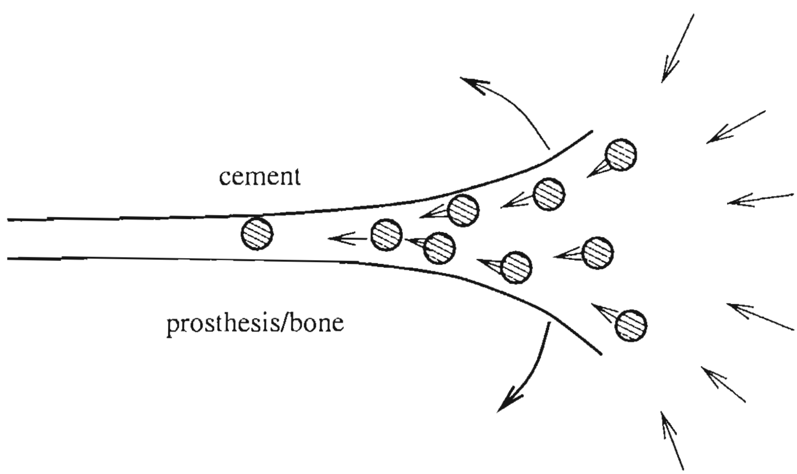

Fig. 5. Debris flow into the cement-bone gap

The gap that opens oscillatory in the metal-cement or cement-bone interface acts as a pump: the wear debris is led into the gap in the opening stage and are hardly removed in the closing stage 5. The crack once produced by the tensile forces is never closed firmly due to the biological reaction. The propagated loosening of the prosthesis in this case is called aseptic loosening.

X-ray pictures show a so-called *lucent line* around the implant. The interesting microanalysis is presented by Revell et al. (1997). After the implantation the metal prosthesis is surrounded by the bone remodeled in its orientation and structure, so that trabeculae are formed parallel to the surface of the implant. There exist two explanations of the nature of the bone-implant interface: some authors describe the direct contact between the bone and biomaterial, others postulate the existence of a fibrous tissue layer. The experiences described by Revell et al. (1997) prove both the mechanisms. Next to an implant, which is aseptically loosened macrophages and fibroblasts are found. These cells form a synovium-like structure at the surface of fibrous tissue. Thorough 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 the process of implant loosening.

4.1. Stress distribution analysis in healthy and diseased bones

Both 2D and 3D FEM analyses show the distribution of stresses in the bone. Nowadays, even complex form of the bone or implant-bone system does not pose considerable difficulties to analysis. Even simple radiology enables to model the real geometry in a particular case. Tomography allows for almost automatic passage from the human body to the finite element geometric model (Fig.6).

Moreover, the X-ray pictures show the density change in some parts of a bone and at the same time the lower relative load carrying capacity. In most parts the external contour can be determined straightforward. The difficultis appear in the contact region between the acetabulum located in the pelvis and the bone head. The internal contour determines the sites where the density (i.e. X-ray transmission) has a specified low value. Since the bone density and 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 case of trabecular bone, mostly of the lattice structure, rigid enough and 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 of diseased bones both the geometry and material properties may differ considerably from the intact natural bones.

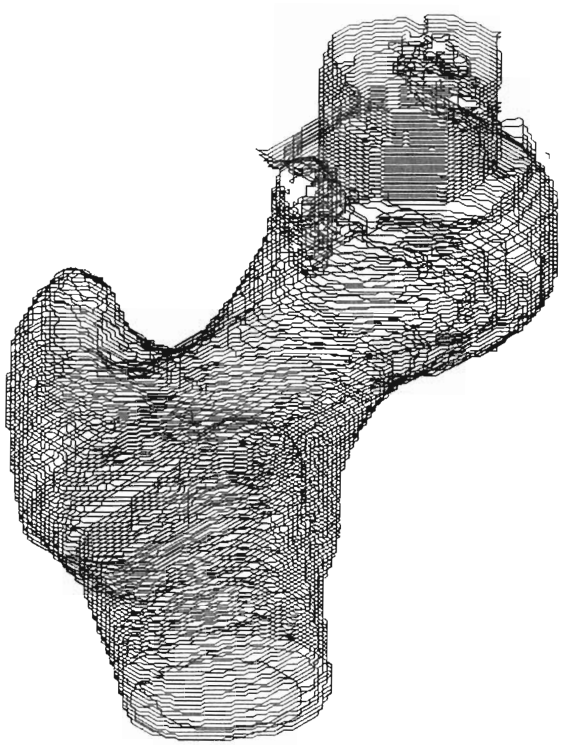

Fig. 6. Contour lines of the hip joint obtained using the tomography

5. Conclusions

There are two main reasons behind the prosthesis loosening:

- Biological, strongly depended on the individual predispositions, which should be investigated on the microscopic level; e.g., wear of the implant, caused both by the biological factors and mechanical factors (for example friction), migration of the wear debris to the bone-cement-steel contact surface, slacken rebuilding of the bone

- Mechanical, when the stress limits are exceeded; it results in the fatigue, especially under cyclic load, crack propagation and material fragmentation.

There exists limited contribution of mechanical research towords solution the problem. The same concerns the microbiological field. A realistic simu-

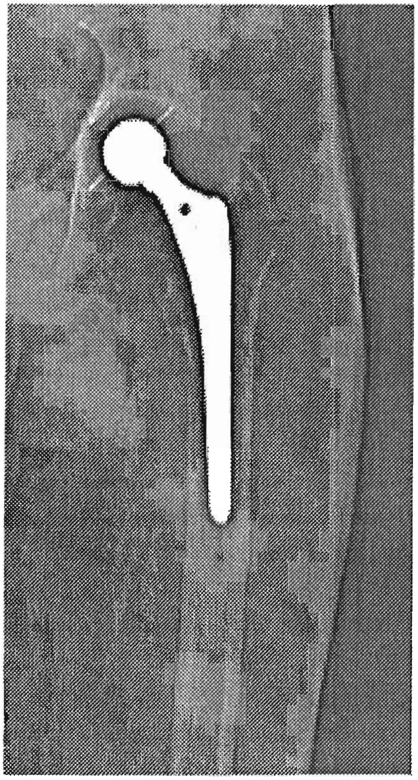

Fig. 7. Hip joint implant

lation of all the phenomena and the authoritative answer to the question of loosening of prosthesis is open.

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Implanty stawu biodrowego – przegląd modelowaia numerycznego

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 stanu 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 $10 \div 20\%$.

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