Acta of Bioengineering and Biomechanics
Vol. 15, No. 2, 2013

Original paper
DOI: 10.5277/abb130202

Analysis of the influence of a metha-type metaphysical stem on biomechanical parameters

ANDRZEJ POZOWSKI[1]*, KRZYSZTOF ŚCIGAŁA[2], ANDRZEJ KIERZEK[1],
MAŁGORZATA PAPROCKA-BOROWICZ[1], JADWIGA KUCIEL-LEWANDOWSKA[1]

[1] The Department of Physiotherapy at the Faculty of Health Sciences at Wrocław University of Medicine, Poland.
[2] The Department of Biomedical Engineering and Experimental Mechanics, Wrocław University of Technology, Poland.

The full postoperative loading of the limb is possible if patients are properly selected and qualified for hip arthroplasty and the requirements as to the proper position of the metaphysial stem are met. The lack of precision, and patient qualification which does not satisfy the fixed criteria may result in stem setting inconsistent with the assumptions. An analysis based on the finite element method (FEM) will enable one to find out how to plan the magnitude of operated joint loading on the basis of the position of the stem in the postoperative radiograph. By analyzing the distribution of bone tissue deformations one can identify the zones where the spongy bone is overloaded and determine the strain level in comparison with the one determined for a model of the bone with the stem in proper position. On the basis of the results obtained one can estimate the range of loads for the operated limb, which will not result in the loss of the stem's primary stability prior to obtaining secondary stability through osteointegration. Moreover, an analysis of the formation of bone structures around the stem showed that the incorrect setting of a Metha-type stem may lead to the initiation of loosening.

Key words: *metaphysial stem, osteointegration, primary stability, secondary stability, stem loosening, finite element method*

1. Introduction

Even though hip arthroplasty as the method of choice in the treatment of advanced arthrosis has been used for several decades there is still no certainty that when performed, the operation will be the first and the last one. One of the reasons is the difference in elasticity and strength between the bone and the implant. As a result, the loads and stresses transferred between the bone tissue and the implant significantly differ from the normal parameters. This is one of the main causes of implant destabilization, as reported by numerous authors [3], [4], [6], [11], [14], [17]. Thanks to physical model studies and computer simulations ever new solutions are introduced [2], [6], [7], [10], [12], [13], [15], [16]. The distribution of stress and the transfer of load are analyzed not only in the femoral bone's proximal segment, but also in its distal segment [3], [5], [18]. Thanks to the material and structural changes introduced as a result of interdisciplinary research the percentage of aseptic loosenings has decreased and there is no statistically significant difference in the survival period between the uncemented implant and the cemented one. The long-term results are comparable. The above also applies to the porous titanium coating, the hydroxyapatite coating or the mixed porous titanium/hydroxyapatite coating [1], [4], [8], [9].

As the shape of the fixing and the methods of primary fixing of the stem are improved and optimized, the number of such complications consistently decreases. This does not mean, however, that the problem has been ultimately solved. Multidirectional analyses carried out using various methods have shown that the restoration of the proper function of the hip joint through treatment involving endoprostheses to a high

* Corresponding author: Andrzej Pozowski, The Department of Physiotherapy at the Faculty of Health Sciences at Wrocław University of Medicine, Poland.

Received: July 16th, 2012
Accepted for publication: January 3rd, 2013

degree depends on how well the implant fits into the surrounding tissue environment and how close the transfer of loads and stresses resembles the natural process. The fit applies to both geometrical and biomechanical parameters. The search for a compromise has led to solutions which, sparing the bone structures, concentrate the released static and dynamic forces in places where they are similar to the normal ones.

In opposition to long standard stems, there is resurfacing. However, the indications for its use are limited. An intermediate solution are metaphysial stems. Their use requires precision operative technique. In the case of short stems of the Metha type, the geometrical fit, i.e., the correct position of the implant within the proximal epiphysis of the thigh bone, is critical. Only when patients are properly selected and qualified for this type of procedure and the requirements concerning the correct position of the metaphysial stem are met, the full postoperative loading of the limb is possible.

a) 
b) 

Fig. 1. Possible adjustment of angular setting of short Metha stem

It should be noted here that owing to the modular design of Metha stems the angular settings of their components can be adjusted in a wide range (Figs. 1 and 2). In principle, it is possible to fully restore the proper geometrical interrelations, including the neck-shaft angle and the angle antiversion, within the hip joint.

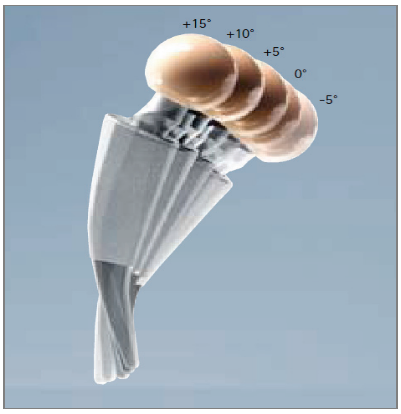


Fig. 2. Possible adjustment of stem angular setting in femoral canal

Thanks to the modular design of the implant one can also change the linear dimensions, including the endoprosthesis offset, whereby the limb shortening parameters can be controlled to a certain extent (Fig. 3). Unfortunately, the modular design and the adjustability of the angular dimensions may, in some cases, lead to such a prosthesis position for which it will be rather difficult, because of the disadvantageous distribution of the biomechanical parameters, to achieve and maintain stem stability.

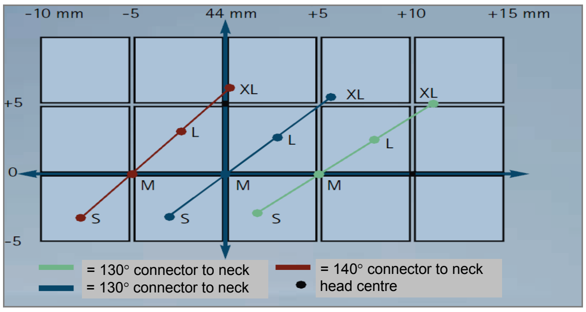


Fig. 3. Possible changes of limb length and endoprosthesis offset

Particularly dangerous seems to be an improper positioning of the stem within the proximal metaphysis of the femoral bone, i.e., such a change in the angular setting of the stem in the frontal plane which will disturb the planned transfer of loads from the stem to the femoral bone tissues.

For this reason, this paper is devoted to the analysis of the influence of a metaphysial Metha stem setting on the distribution of displacements, stresses and strains in the compact and spongy tissue of the thigh bone and the effect of this setting on the forming and remodelling of the spongy tissue structures.

2. Materials and methods

As part of clinical observations an analysis of the results of treatment involving the use of metaphysial stems in 47 patients of the Area Health Service Orthopaedic-Rehabilitation Unit in Wrocław was carried out. In the group of patients under study, stems of all sizes (Fig. 4) were used, but stems with a neck–shaft angle of 135° and an antiversion–retroversion angle predominated.

The biomechanical analysis included numerical computations of stress, deformation and displacement distributions in the implant/bone system for both correct and incorrect stem settings. FE models of the femoral bone, based on CAT measurements of a physical bone model of the SAWBONE type, were created. Then a model of the metaphysial Metha stem, based on the measurements of the real implant, was incorporated into the femoral bone model. In the first model developed the stem was positioned correctly, whereas in the second model the stem was so positioned that its distal end did not come to rest on the side wall of the bone's cortical layer while its proximal end was shifted sideways and slightly upwards relative to the normal position.

The geometrical models developed were subjected to discretization using a Tetra solid element with 10 nodes and 3 DOFs in each node (Fig. 5).

The discrete models of the femoral bone with the metaphysial stem were subjected to load typical of the phase of standing on the lower limb, consistently with a load model developed in the Biomedical Engineering Department of Wrocław University of Technology (Fig. 6).

a) b) c)

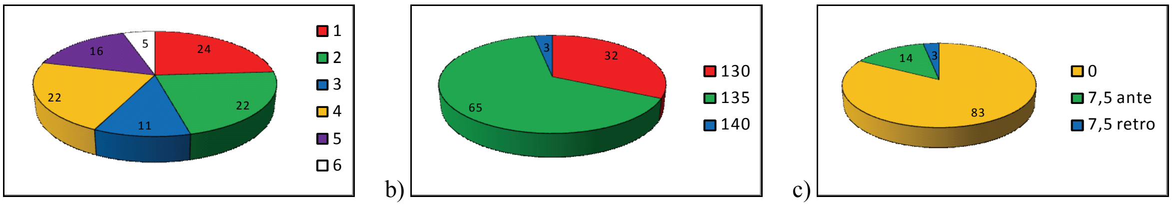
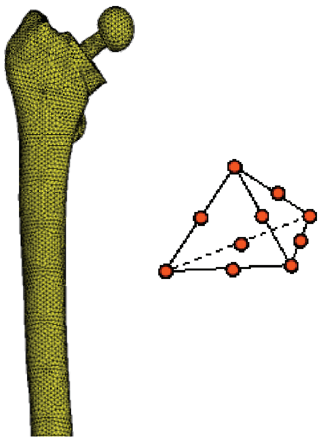
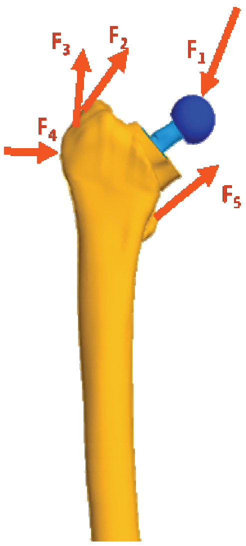


Fig. 4. Percentage of prostheses with parameters: (a) stem size, (b) neck–shaft angle, (c) antiversion–retroversion angle, in all operations performed



Volume	Number of finite elements	Number of nodes
Compact bone tissue	68498	109669
Spongy bone tissue	57356	87667
Stem of endoprosthesis	19702	31007
Head of endoprosthesis	3421	5345

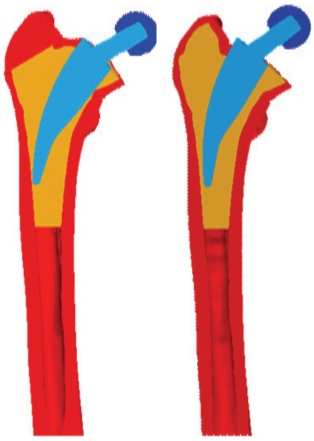
Fig. 5. FE model of femoral bone with short Metha stem, and grid parameters



Force	F_x [N]	F_y [N]	F_z [N]
F_1 – reaction to head	–490	–1302	–70
F_2 – gluteus minimus	88	66	–23
F_3 – gluteus medius	303	218	–202
F_4 – tractus iliatibialis	41	98	–15
F_5 – iliopsoas	42	330	–357

Fig. 6. Load model used in analysis

a) b)



Material	Modulus of elasticity [MPa]	Poisson's ratio [–]
Compact bone tissue	18600	0.3
Spongy bone tissue	478	0.42
Stem of endoprosthesis	105 000	0.3
Head of endoprosthesis	300 000	0.3

Fig. 7. Stem setting in femoral bone canal (a), properties of tissues and materials (b)

The models took into account the division into compact bone tissue and spongy bone tissue, which were modelled as isotropic materials with a linear stress–strain characteristic. The mechanical properties were selected on the basis of the authors' own research and literature reports (Fig. 7).

A separate part of the analysis consisted in numerical simulations of the formation and remodelling of the osseous structures of the spongy tissue surrounding the metaphysial stem.

The discrete models used in this simulations reflected the breakdown into compact tissue and spongy tissue, with the former modelled as a solid material and the latter modelled as porous material with an assumed initial uniform density, retaining structural isotropy. The model with the assumed initial structure of spongy tissue is shown in Fig. 8.

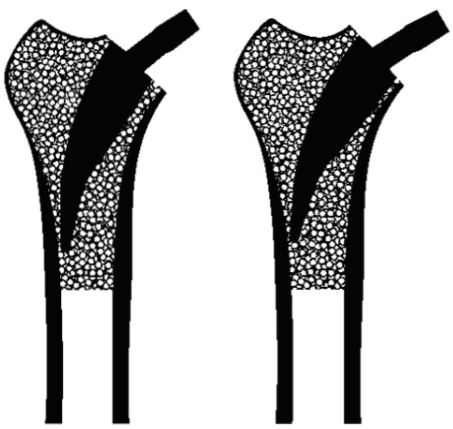


Fig. 8. FE models used to analyze bone structure formation

The simulations of the formation and remodelling of the trabecular structures were based on the Tsubota model [19], which assumes that the stimulus for the formation and resorption of trabecular structures is the nonuniformity of stress distribution on the surface of the bone trabeculae.

For each point lying on the surface of the bone trabeculae, a certain neighbourhood (Fig. 9) is determined for which a stress analysis is carried out.

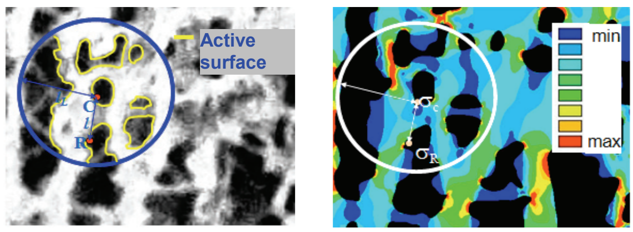


Fig. 9. Determination of local stress distribution heterogeneity

The stimulus itself is determined from relation (1) describing the ratio of the stress at the analyzed point to the stress at the other points of its neighbourhood.

$$\Gamma = \ln\left(\frac{\sigma_C}{\sigma_D}\right), \tag{1}$$

where: σ_c – stress at currently analyzed point C, σ_d – a value determined on the basis of the stress values in a precisely defined neighbourhood of point C (in accordance with Fig. 9).

The stress values in the neighbourhood of the analyzed point are calculated from relation (2).

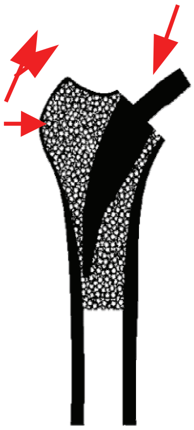
$$\sigma_D = \frac{\int_S w(L)\sigma_R dS}{\int_S w(L) dS}, \tag{2}$$

$$w(L) = \begin{cases} 1 - \dfrac{L}{L_L} & (0 < L < L_L), \\ 0 & (L_L \le L), \end{cases} \tag{3}$$

where: S – the active surface, L – the distance between points C and R, σ_R – the stress value at point R, L_L – the radius of the circular area analyzed.

It should be noted that the effect of the stresses on the remodelling stimulus value depends on their location in the neighbourhood of this point and it is described by relation (3).

The models were subjected to the load typical of the walk support phase. The location of the loading forces and their values are shown in Fig. 10.



Force	F_x [N]	F_y [N]
Reaction to head	–490	–1302
Gluteus minimus	88	66
gluteus medius	303	218
Tractus iliatibialis	41	

Fig. 10. Load model used to analyze remodelling of trabecular structures of spongy tissue around stem

3. Results

Selected results of treatment involving the use of metaphysial stems are shown in Fig. 11.

The numerical analysis of the state of displacement showed that under the load the proximal epiphysis of the femoral bone displaces towards the medial side, forwards and downwards. From the three components the highest and lowest values assume respectively the downward displacements and the forward displacements. A comparative analysis showed that for the incorrect setting the displacements towards the medial side and the forward displacements decrease significantly relative to the values obtained for the model with the correctly positioned stem. It should be noted, however, that at the same time the downward displacements for the incorrectly positioned stem are slightly larger (Fig. 12), which indicates that in the case of incorrect setting it may be much more difficult to achieve the full primary axial stability. The stem will tend to migrate axially in the proximal epiphysis of the thigh bone. Taking into consideration possible negative remodelling caused by the overloading of the bone structures and possible mechanical degradation of the overloaded tissue structures, this process may intensify in the distant postoperative period.

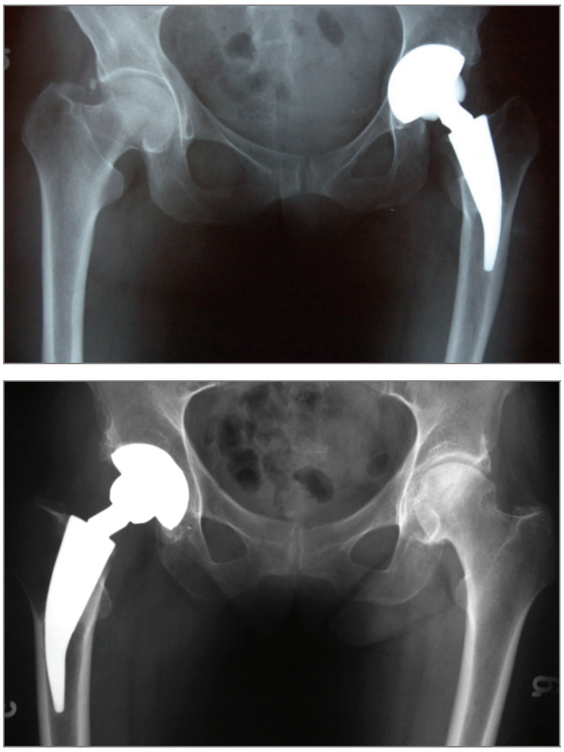


Fig. 11. Exemplary results of clinical treatment

An analysis of the stress in the cortical tissue of the proximal part of the femoral bone showed that because of the dominant bending of the bone in the frontal plane and the critical bending in the sagittal plane, distinct stress concentrations occurred in the distal part of the model. The maximum tensile stresses occurred on the posterior-medial side of the bone shaft. A comparative analysis of the stress values for the correct setting and the incorrect setting of the stem showed that the changes in stress in the bone shaft are negligibly small. In the case of both compressive and tensile stress, the stress values obtained for the incorrect stem setting were higher than for the correct stem setting, but in both cases the difference did not exceed 5%.

But significant differences were found when analyzing the distribution of stress and strain in the spongy tissue of the bone. The distribution of strain in the tissue surrounding the bone, presented in Fig. 14, reveals differences on the lateral side (the greater trochanter) as well as on the medial side, relative to the shaft. The strain values in the region of the

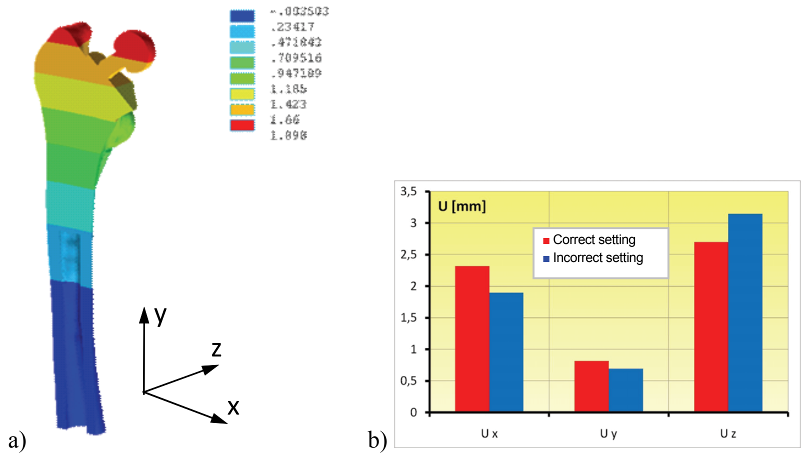


Fig. 12. Typical distribution of displacement component U_x (a), comparison of maximum displacement components for two stem settings (b)

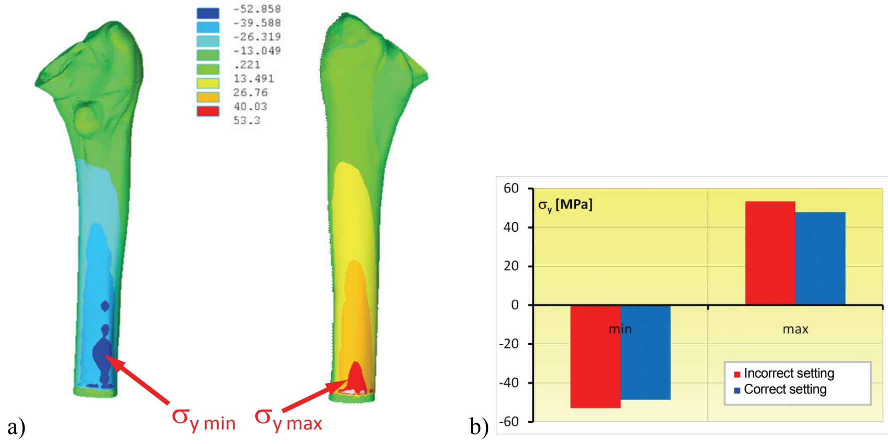


Fig. 13. Exemplary distribution of stress σ_y (a), comparison of stresses for two stem settings (b)

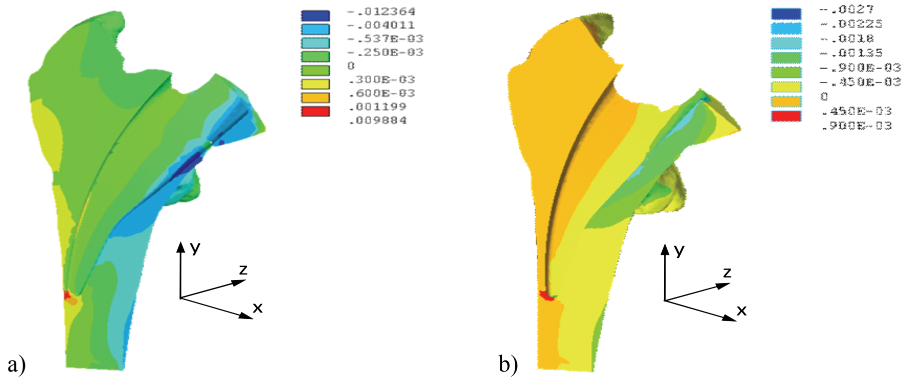
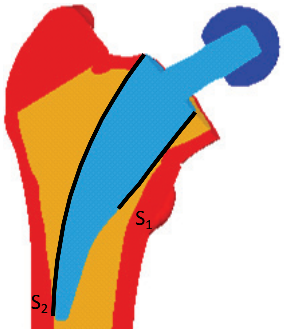


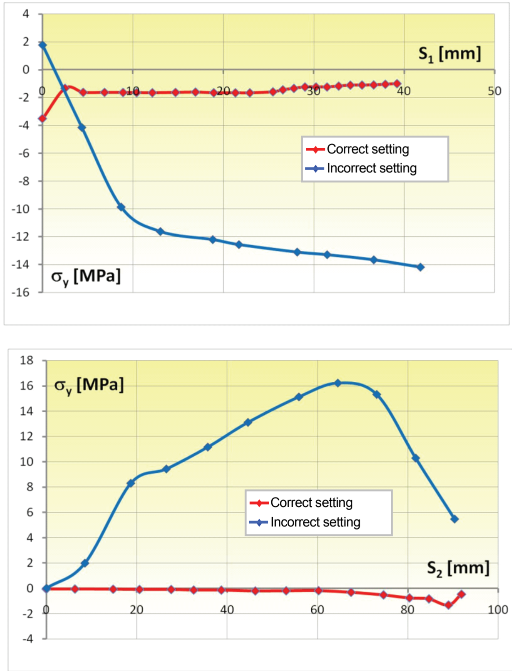
Fig. 14. Exemplary distribution of strain σ_y: (a) for correct stem setting, (b) for incorrect stem setting

proximal part of the stem are markedly lower for the incorrect setting of the stem. Because of the weaker mechanical stimulation of the spongy tissue until growth and permanent union with the stem take place, one should not consider this result as positive. A distinct stress concentration can be observed in the region of the stem end.

However, for the incorrect stem setting the stress concentration is clearly shifted into the bone. Also the area in which the maximum strains occur is distinctly larger. Hence one can conclude that in the case of incorrect stem setting, insufficient stability of the proximal part and overload of the bone tissue in the region of the stem end will simultane-



a)



b)

Fig. 15. Lines along which stress distributions were analyzed (a) and results of analysis (b)

ously occur, which may lead to the loss of axial stability and the sinking of the stem into the osseous base.

The state of stress was analyzed along two paths: S_1 and S_2, as shown in Fig. 15. Along path S_1 for the correct stem setting the stress is uniformly distributed. In the case of the incorrect stem setting it was found that stress sharply increases in the proximal part of the path and the slope of the stress distribution curve changes at about one fourth of the path length, but the stress continues to increase, reaching at the distal path end nearly seven times higher values than for the correct stem setting.

Similarly in the case of path S_2 for the correct stem setting, the distribution of stress is uniform, only at the distal end of the path a slight local increase in stress can be observed. For the incorrect stem setting initially stress quickly increases and then at about one-fourth of the path length the slope of the curve changes, stress continues to increase and at about three-fourths of the path length the slope of the curve changes again and stress gradually decreases. It should be noted that stress is minimum for the correct stem setting, whereas for the incorrect stem setting, stress is even greater than in the case of path S_1.

An analysis of the formation and remodelling of the bone tissue structures clearly indicated detrimental effects of the incorrect stem setting. For the correct stem setting, well developed trabecular structures are visible on both the lateral side and the medial side of the stem (Fig. 16). On the lateral side one can see bone trabeculae with an average length and a relatively large diameter, forming structures connecting the stem's surface with the bone's cortical layer. Structures running along the medial-lateral direction and structures running nearly parallel to the stem's surface are visible. On the medial side one can see structures supporting the stem, which run diagonally in the superior-medial direction. A reduction in the density of the structures occurs only in the distal part. Distinct support structures have developed in the region of the stem's end.

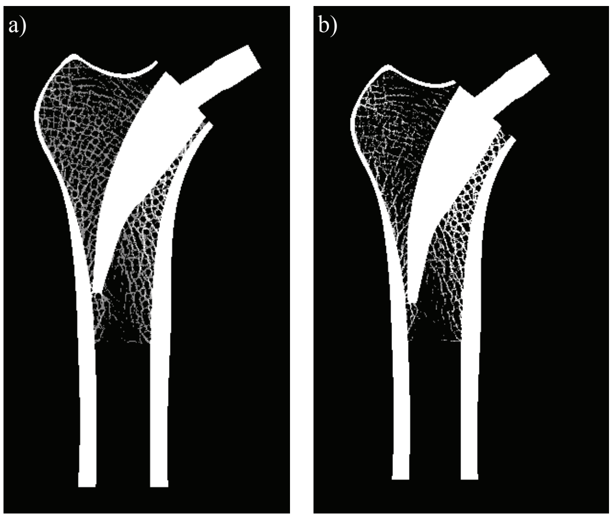


Fig. 16. Simulation results for formation of bone trabecular structures: (a) for correct stem position, (b) for incorrect stem position

If the stem setting is incorrect, the spongy bone structures on the lateral side are undeveloped, actually in decline. The bone trabeculae are clearly longer and smaller in diameter. Also the number of intertrabecular junctions has decreased. On the medial side in the proximal part the trabecular structures are very well developed – large-diameter trabeculae connected into distinctly developed support structure are visible. One should note, however, that in the distal part the structures are practically in decline. The support structures in the region of the stem end are also poorly developed.

4. Discussion

The analysis has indicated several detrimental effects of the incorrect insertion of the endoprosthesis, which may lead to unsuccessful treatment. An incorrect setting of the endoprosthesis stem adversely affects the values and distribution of most of the parameters which describe the biomechanical aspects of the stem/bone interaction. An increase in displacement values indicates that the incorrect stem setting may cause difficulties in achieving the axial stability of the stem. The undesirable changes in the distribution of strains in the region of the stem end combined with considerable changes in the distribution of stresses on both the medial side and the lateral side indicate that the stability loss scenario is quite probable due to the possible mechanical degradation of the tissue structures in the distal part of the stem and to the lack of proper support in the proximal part. These observations are corroborated by the results of the analysis of the formation and remodelling of the trabecular structures. The trabecular structures observed in the region of the stem end do not constitute a proper mechanical support when the stem axially sinks into the osseous base. In the proximal part, well developed structures occur only on the medial side. The lack of stem mechanical support on the medial side may result in the axial migration of the stem.

The decades-long research aimed at reducing or eliminating the causes of the loss of stability of the endoprostheses, has been conducted mainly in two directions. Clinical observations of patients with surgically implanted endoprostheses take into account the intravital causes of their destabilization and involve comparative analyses of the findings for the different types of the endoprostheses used. Such research data together with the results of biomechanical studies of, among other things, the materials used for the manufacture of the endoprosthesis, their tribology, load transfer directions and stress distribution have been inspiring research teams to search for the best solutions [1], [3], [4], [7], [9], [13].

A substantial number of the studies and their results reported in the available literature concern the design of long conventional (simple and modular) stems, taking into account their shape, stiffness and flexibility [6], [7], [10], [11], [13], [17]. It has been shown that load transfer takes place mainly in the metaphysial segment whereby the necessity to implant long stems is avoided. In the last decade many short metaphysial stems have been constructed. Some of them have not proved successful, some continue to be under long-term observation. The fact that Metha-type stems are used by many centres and the reported medium-term results confirm their high suitability. The infrequent cases of premature stem migration and loosening require further analyses of their causes and one of the causes has been explained in this report.

5. Conclusions

1. The statistics of both the short-term and long-term results indicate a need for further research in this direction.

2. Besides the design and material characteristics, also the precision of the operation has an influence on the results of treatment.

References

[1] BARRACK R. et al., *Improved cementing techniques and femoral component loosening in young patients with hip arthroplasty. A 12-year radiographic review*, Journal of Bone and Joint Surgery, British, 1992, Vol. 74B, 385–389.

[2] BEAUPRE G.S. et al., *An approach for time-dependent bone remodelling and remodelling applications: a preliminary simulation*, Journal of Orthop. Res., 1990, Vol. 8, 662–670.

[3] BĘDZIŃSKI R. et al., *Biomechanical aspects of artificial joint implantation in a lower limb*, Journal of Theoretical and Applied Mechanics, 1999, Vol. 37, 455–481.

[4] BUGBEE W.D. et al., *Long-Term Clinical Consequences of Stress-Shielding after Total Hip Arthroplasty without Cement*, Journal of Bone and Joint Surgery, American, 1997, Vol. 79, 1007–1012.

[5] CARTER D.R. et al., *Skeletal Function and Form*, Cambridge University Press, 2001.

[6] CRISTOFOLINI L., *A critical analysis of stress shielding evaluation of hip prostheses*, Crit Rev Biomed Eng., 1997, Vol. 25, 409–83.

[7] CROWNINSHIELD R.D. et al., *An analysis of femoral component stem design in total hip arthroplasty*, The Journal of Bone and Joint Surgery, 1980, Vol. 62, 68–78.

[8] ENGH C., *Porous-coated hip replacement. The factors governing bone ingrowth, stress shielding, and clinical results*, Journal of Bone and Joint Surgery – British, 1987, Vol. 69B, 45–55.

[9] GRUEN T.A. et al., *"Modes of Failure" of Cemented Stem-type Femoral Components: A Radiographic Analysis of Loosening*, Clinical Orthopaedics & Related Research, 1979, Vol. 141, 17–27.

[10] HUISKES R. et al., *Mathematical shape optimization of hip prosthesis design*, Journal of Biomechanics, 1989, Vol. 22, 793–804.

[11] HUISKES R. et al., *The Relationship Between Stress Shielding and Bone Resorption Around Total Hip Stems and the Effects of Flexible Materials*, Clinical Orthopaedics & Related Research, 1992, Vol. 274, 124–134.

[12] HUISKES R. et al., *Osteocytes and Bone Lining Cells: Which are the Best Candidates for Mechano-Sensors in Cancellous Bone?*, Bone, 1997, Vol. 20, 527–532.

[13] KUIPER J.H., *Mathematical Optimization of Elastic Properties: Application to Cementless Hip Stem Design*, Journal of Biomechanical Engineering, 1997, Vol. 119, 166–174.

[14] PAUWELS F., *Biomechanics in Locomotor Apparatus*, Springer-Verlag, 1980.

[15] PRENDERGAST P.J., *Biomechanical Techniques for Pre-clinical Testing of Prostheses and Implants*, Lecture Notes 2, AMAS Polish Academy of Sciences, 2001.

[16] PRENDERGAST P.J., *Finite elements models in tissue mechanics and orthopaedic implant design*, Clinical Biomechanics, 1997, Vol. 12(6), 343–366.

[17] SUMNER D., *Functional adaptation and ingrowth of bone vary as a function of hip implant stiffness*, Journal of Biomechanics, 1998, Vol. 31, 909–917.

[18] ŚCIGAŁA K., *FEM analysis of deformation and surgical correction of tibia bone*, Acta Bioeng. Biomech., 2002, Vol. 4(1), 321–322.

[19] TSUBOTA K., *Spatial and temporal regulation of cancellous bone structure*, Medical Engineering & Physics, 2005, Vol. 27, 305–311.