Russian Journal of Biomechanics, Vol. 5, № 1: 12-23, 2001

Russian Journal of Biomechanics
www.biomech.ac.ru

THE BONE TISSUE REMODELING IN THE PROXIMAL FEMUR UNDER THE VARIATIONS OF HIP JOINT DAILY LOADING HISTORY

Yu.V. Akulich, Yu.I. Nyashin, R.M. Podgayets, I.A. Dmitrenok

Department of Theoretical Mechanics, Perm State Technical University, 29a, Komsomolsky Prospect, 614600, Perm, Russia, e-mail: auv@theormech.pstu.ac.ru

Abstract. The change of elastic modulus distribution in the sponge bone tissue of the proximal femur under the variations of the hip joint daily loading history has been investigated. The computer simulation of the bone remodeling process has been performed on the basis of two-dimensional finite element model of the proximal femur and kinetic equation of the bone tissue remodeling with a strain stimulus of adaptation. The physiological loading history has been considered as a superposition of several distinct loading conditions with different magnitudes and directions of the forces acting on the proximal femur, and with different number of loading cycles. As a variation of the character of hip joint loading, some hypothetical modification of human activity that causes changes in relative contribution of each load case, has been considered. It is shown that the elastic modulus distribution in the hip head and femur neck varies continuously under the load character modification, but after the customary loading history is restored it gradually returns to its original pattern. The problem on the determination of local safety factor of a bone tissue under the multiply-direction loading has been discussed.

Key words: internal remodeling, elastic modulus, bone tissue, strain remodeling stimulus, hip head, multiply-direction loading, load case, number of loading cycles

Introduction

The present paper continues the authors' investigations on the problem of bone tissue remodeling. In the paper by Akulich and Podgayets [1] the brief review of known equations of sponge bone tissue remodeling was presented, their numerical analysis was carried out and the following kinetic equation of adaptive remodeling of the bone tissue elastic modulus with local strain stimulus was suggested:

$$dE(\mathbf{x},t)/dt = C\,[\varepsilon_i(\mathbf{x},t) - \varepsilon_i^{hom}(\mathbf{x})]\,. \qquad (1)$$

Here $E(\mathbf{x},t)$ and $\varepsilon_i(\mathbf{x},t)$ are the elastic modulus and the strain intensity at the point $\mathbf{x}$ at the instant of time t, respectively; C is a remodeling rate factor; $\varepsilon_i^{hom}(\mathbf{x})$ is the strain intensity at the point $\mathbf{x}$ under conditions of homeostasis, which took place in the bone tissue under physiological load. In papers by Akulich et al. [2, 3] this kinetic equation was applied in a numerical experiment for prediction the changes in mechanical characteristics of sponge bone tissue in the hip head and femur neck during the post-operational rehabilitation period. The main feature of this period is a gradual recovering of the bone tissue elastic modulus after the long period of low load in the conditions of bed regimen that takes place due to the special system of therapeutic motions.

In these works the physiological loading history was considered as a superposition of several distinct cyclic load cases with different magnitudes and directions of the forces acting on the hip head and greater trochanter, and with different cycle numbers of each specific load case. The assumption was made that the time variations of therapeutic loading consisted only in proportional change of force magnitudes in each load case, while their inclinations to the vertical and a number of cycles remained unchanged.

In the present paper a variation of the hip joint loading history is considered as some hypothetical modification of human activity that causes changes in relative contribution of each load case, with unchanged magnitudes and directions of the forces in these load cases. Similar to paper [3], it is assumed that the remodeling rate factor at the negative magnitude of remodeling stimulus $[\varepsilon_i(\mathbf{x},t)-\varepsilon_i^{hom}(\mathbf{x})]$, i.e. under weakening of bone tissue, is 1.33 times larger than the same factor under bone strengthening. That is in a good agreement with medical expertise, saying that the resorption of a bone tissue is a faster process than its reposition.

Methods

Under different kinds of human physical activity different loads act on the hip joint and proximal femur, and they are dependent on the external forces and the forces generated by the muscles attached to the femur. The determination of these loads is a very difficult problem, which is a subject of numerous investigations. The loads are cyclic, and during the study of stresses and strains in the femur these forces are taken as their maximal magnitudes (or amplitudes in symmetric cycles). In present paper we employed simplified two-dimensional model of adult proximal femur with only two forces: the joint reaction force and the hip adductor force. In our model the joint reaction force was presented as a pressure distributed on an arc about a quarter of the hip head circle by the cosine law, with the resultant force $\mathbf{F}_1$ directed at angle α with the vertical. The hip adductor force was considered as a uniformly distributed pressure acting on the greater trochanter surface, with the resultant force $\mathbf{F}_2$ inclined at angle β to the vertical. Depending on the character of human physical activities the magnitudes and directions of these forces for the twenty-four hours are different, and the bone tissue architecture and mechanical characteristics are dependent on their overall effect. According to Carter et al. [4], Weinans et al. [5], the following three load cases were considered (see Fig. 1):

1st load case	F_1 = 2317 N	α = 24°	F_2 = 702 N	β = 28°
2nd load case	F_1 = 1158 N	α = -15°	F_2 = 351 N	β = -8°
3rd load case	F_1 = 1548 N	α = 56°	F_2 = 459 N	β = 35°

The first load case with maximal magnitudes of forces acting on the proximal femur corresponds to one-limb-stance phase of gait, and two other load cases are selected to represent extreme ranges of human motion with reduced force levels. As a normal physiological load per day we assumed 6000 cycles of load applications for the first load case and 2000 loading cycles each for the other two cases.

Following Svesnsson et al. [6], the investigation of stresses and strains in the proximal femur was carried out by quasi-two-dimensional finite element model with side cortical plates (Fig. 1).

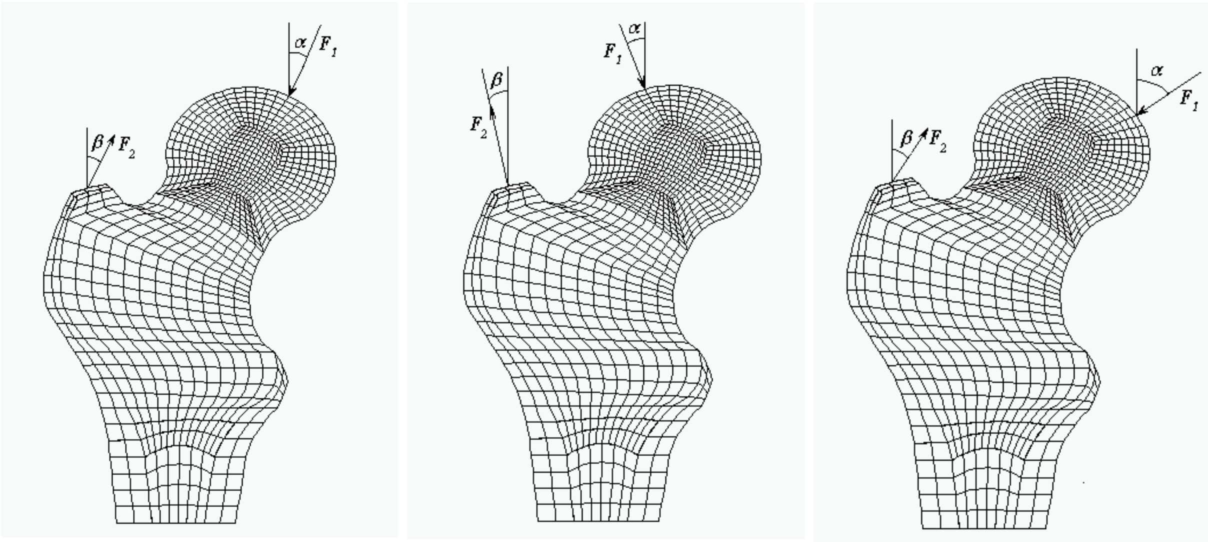


Fig. 1. The central layer of the proximal femur finite element model under three load cases.
a) the first load case: F_1 = 2317 N; α = 24°; F_2 = 702 N; β = 28°;
b) the second load case: F_1 = 1158 N; α = -15°; F_2 = 351 N; β = -8°;
c) the third load case: F_1 = 1548 N; α = 56°; F_2 = 459 N; β = 35°.

In order to study the bone tissue remodeling process by the kinetic equation (1), we have to know:
i) the initial pattern of bone tissue elastic modulus and pattern of homeostatic strain intensity in the proximal femur under the above mentioned daily physiological loading history,
ii) the current distribution of the strain intensity $\varepsilon_i(\mathbf{x},t)$ for the multiple loading by forces with different magnitudes and directions.

The first problem was solved in our paper [2] regarding the bone tissue adaptation during the morphogenesis, with the energy remodeling stimulus proposed by Fyhrie and Carter [7]. The obtained pattern of elastic modulus (Fig. 2) corresponds well enough with the experimentally determined elastic modulus field in the proximal femur (Brown et al. [8, 9]). The pattern of homeostatic strain intensity in the hip head is shown in Fig. 3. To answer the second question, three single-loading-direction solutions for each load case are calculated for each time instant, and in local kinetic equation (1) for each finite element the current strain intensity $\varepsilon_i(\mathbf{x},t)$ is meant as an equivalent strain intensity $\varepsilon_i^{eq}(\mathbf{x},t)$ calculated by averaging the strain intensities referred to all the load cases, with taking into account their numbers of loading cycles:

$$\varepsilon_i^{eq}(\mathbf{x},t) = \left[\sum_{k=1}^{N_{LC}} (n_k/n_T) \left[\varepsilon_i^{(k)}(\mathbf{x},t) \right]^M \right]^{1/M}, \qquad (2)$$

where N_{LC} is the number of load cases and subscript k designates a specific load case; n_k is the number of loading cycles of k^{th} load case per day; $n_T = \sum_{k=1}^{N_{LC}} n_k$ is the total number of loading cycles per day; the strain exponent M is a constant. In our calculations we used $N_{LC} = 3$, $M = 2$.

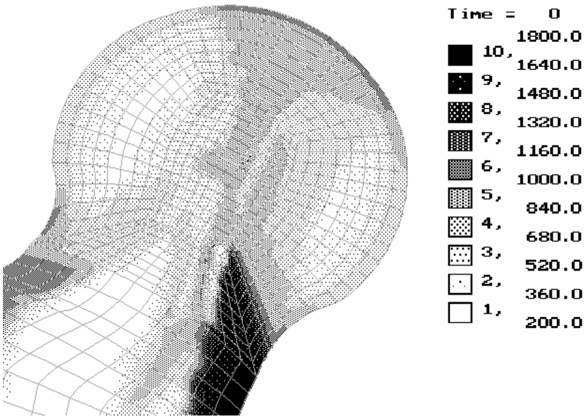


Fig. 2. The elastic modulus pattern in the hip head and femur neck under normal physiological loading condition (in MPa).

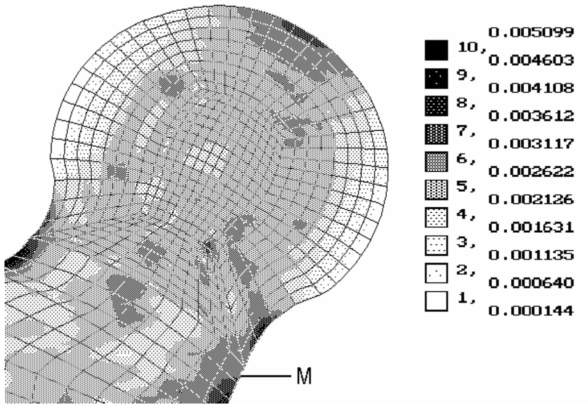


Fig. 3. The homeostatic strain intensity pattern in the hip head and femur neck (the letter M indicates the position of the finite element considered below at the medial surface of the femur neck).

In the present paper this kinetic equation is employed to investigate a remodeling process of bone tissue elastic modulus at some hypothetical variation of human physical activity which causes changes in numbers of loading cycles for the considered load cases. Under normal physiological multiple-direction loading, when the first load case is prevailing as in force magnitudes as in number of loading cycles, the current distribution of strain intensities coincides with the homeostatic one, and the elastic modulus pattern remains unchanged. If we suppose that in some conditions the second or third load case may become prevailing in number of loading cycles, it will cause the deviations of the current strain intensities in finite elements from the homeostatic pattern, and due to this strain stimulus, the remodeling process of the bone tissue elastic modulus will be initiated.

Results and discussion

With the aim of simulation of the process of bone tissue remodeling we considered a situation when a person changes abruptly his or her physical activities over a period of three months, and then returns to customary loads. As a changed human physical activity we assumed daily loading history consisted of 500 loading cycles each for the first and second loads cases and 9,000 cycles of load applications for the third load case. In our computations of the bone remodeling process we have considered two schemes of returning back to normal physiological loading regime (Tab. 1). According to the scheme A after 90 days of changed physical activity the original numbers of loading cycles of the considered load cases are restored at once, but by the scheme B this process follows a multi-step course within three months.

Table 1. The schemes of the load time variations.

Stage	Number of cycles per day			Stage duration, days	
	Load case 1	Load case 2	Load case 3	Scheme A	Scheme B
1	500	500	9,000	90	90
2	2,000	1,000	7,000	–	30
3	3,000	1,500	5,500	–	30
4	4,000	2,000	4,000	–	30
5	6,000	2,000	2,000	180	90

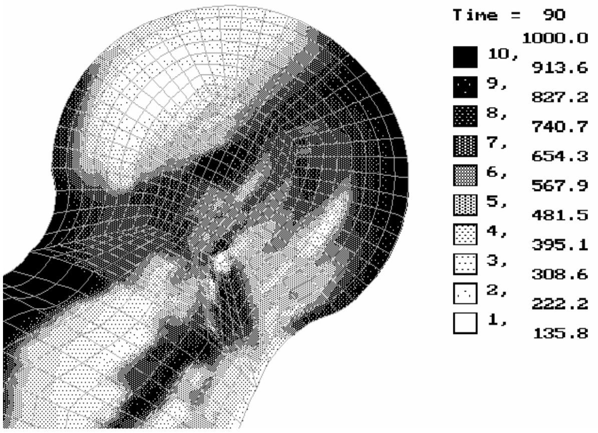


Fig. 4. The elastic modulus pattern in the hip head and femur neck in 90 days after the daily loading history was changed (in MPa).

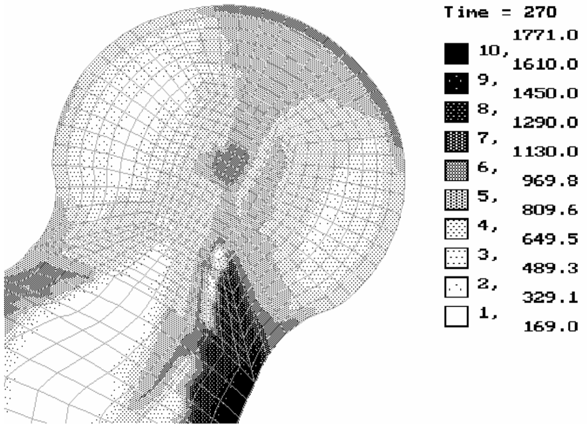


Fig. 5. The elastic modulus pattern in the hip head and femur neck in 6 months after the normal physiological loading condition was restored in line with the scheme A (in MPa).

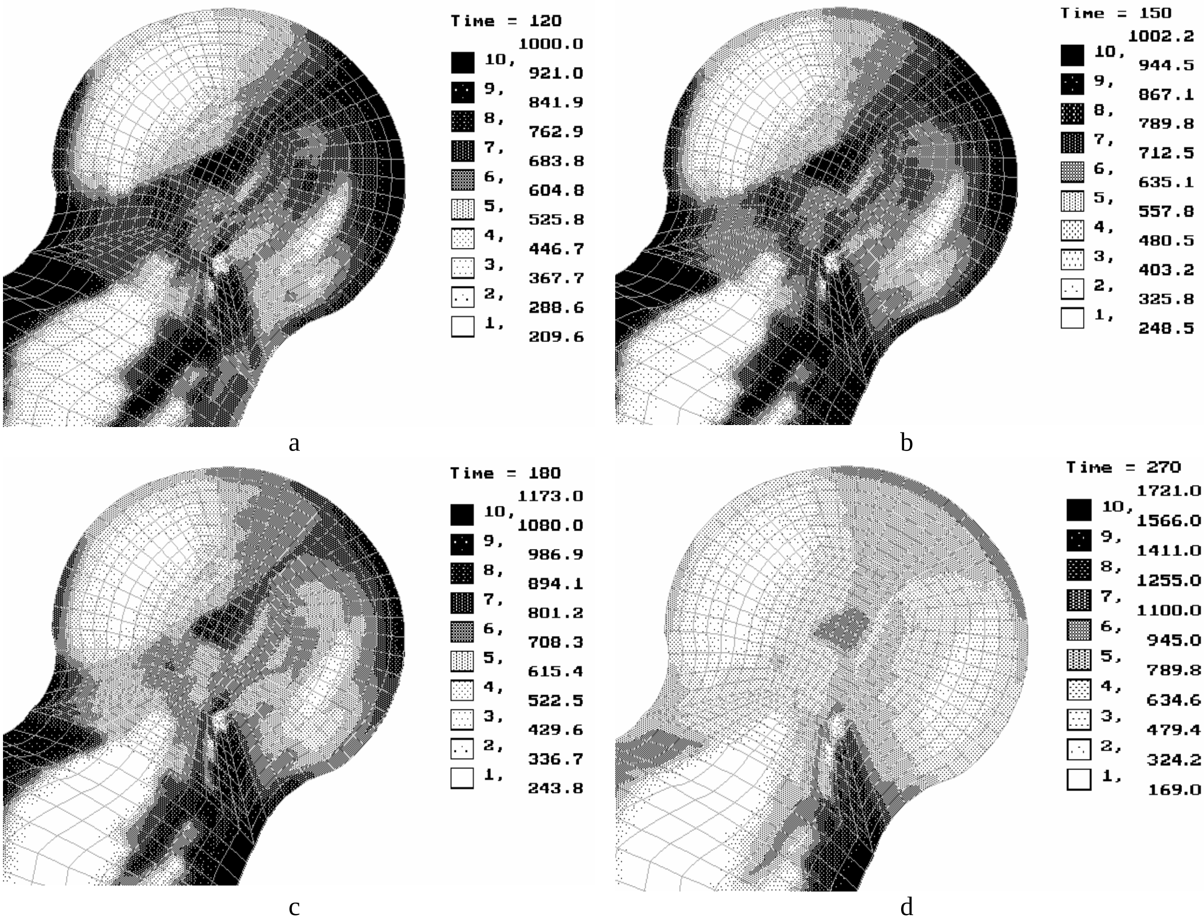
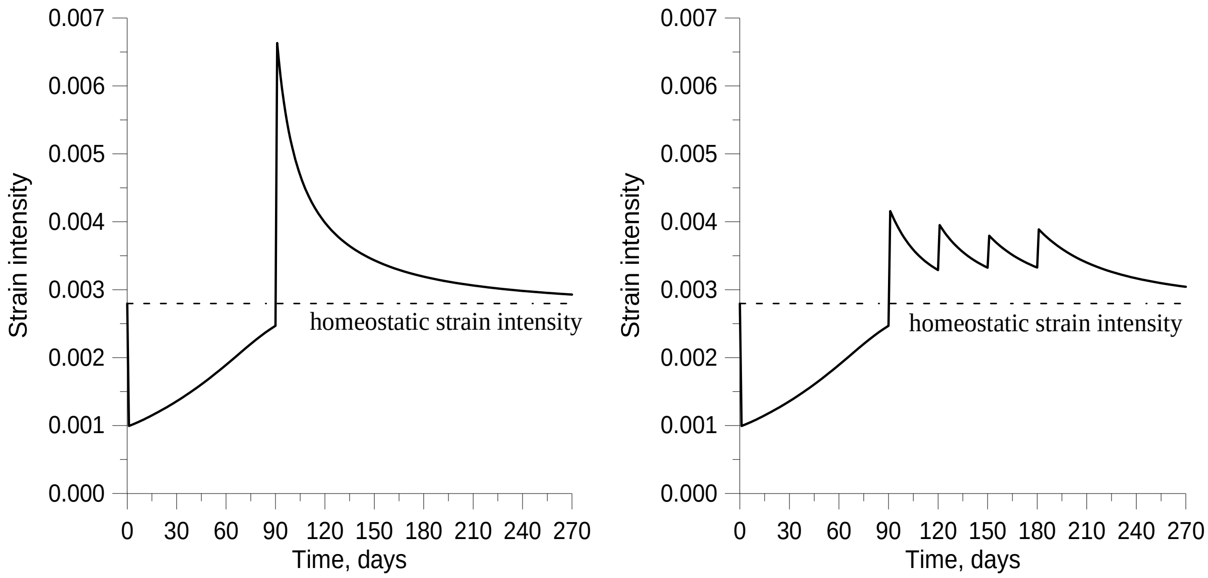


Fig. 6. The bone tissue elastic modulus patterns (in MPa) in the hip head and femur neck at different stages of the stepwise restoration of normal physiological load according to the scheme B (time in days, see Table 1); a) the end of the second stage; b) the end of the third stage; c) the end of the forth stage; d) the end of the fifth stage.

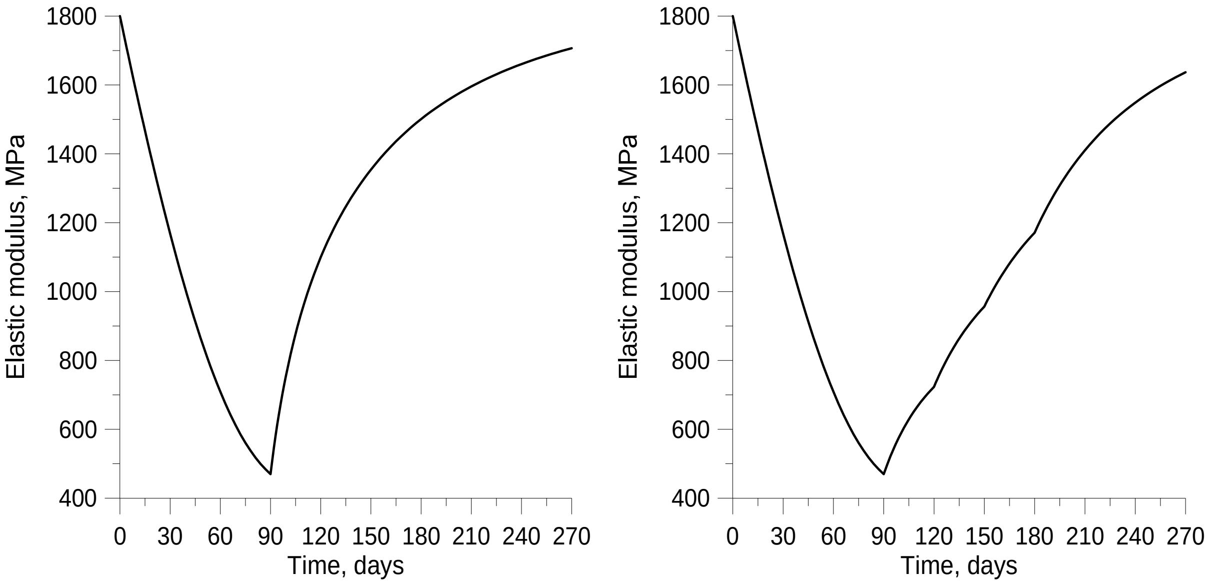
As the initial state we used elastic modulus pattern under the prescribed above normal physiological loading history (Fig. 2), and Fig. 4 shows the elastic modulus distribution after 90 days of changed loading condition. It can be seen that change of the prevailing direction of forces acting on the hip head causes appreciable modification of the bone tissue architecture, and specifically the bone tissue weakening at the medial surface of femur neck. After the customary physiological load was restored, the elastic modulus pattern was recovering step by step, and in six months it was nearly the same as the original one (see Figs. 5, 6).

The strain intensity and elastic modulus in one of the finite elements at the medial side of the femur neck for both schemes of daily loading history variations are plotted in Figs. 7 and 8 as functions of time (the position of the finite element is indicated in Fig. 3 by the letter M). Right after the daily loading history is changed, when the third load case became the prevailing one in number of loading cycles per day, the equivalent strain intensity in this element that originally was equal to the homeostatic strain intensity $\varepsilon_i^{hom} = 0.00280$, abruptly falls down to 0.00099. Within 90 days of the first stage, the strain intensity in this element increases to 0.00247 but the elastic modulus decreases from $1800\ \text{MPa}$ to $469.6\ \text{MPa}$. When the normal physiological load is restored all at once (scheme A), the strain intensity immediately increases nearly three times, and its following diminishing is accompanied by the rise of elastic modulus, in 6 months it reaches the magnitude of $1706.9\ \text{MPa}$. Under the stepwise changing numbers of daily cycles of the load cases (scheme B) the jumps of strain intensity at the beginning of each stage are far less in size, and the elastic modulus increases over the same period of time up to $1636.9\ \text{MPa}$. The presented plots of elastic modulus versus time show that in both schemes of loading change the elastic modulus is nearly restored to its original value. However, if under the loading regime changing abruptly (scheme A) the restoration of elastic modulus runs faster, one may suppose that the multi-step change of daily loading history (scheme B) is more safe, as it does not involve large jumps of strain intensity.



Scheme A Scheme B

Fig. 7. The dependence of strain intensity on time at the medial side of the femur neck for two schemes of load change.



Scheme A

Scheme B

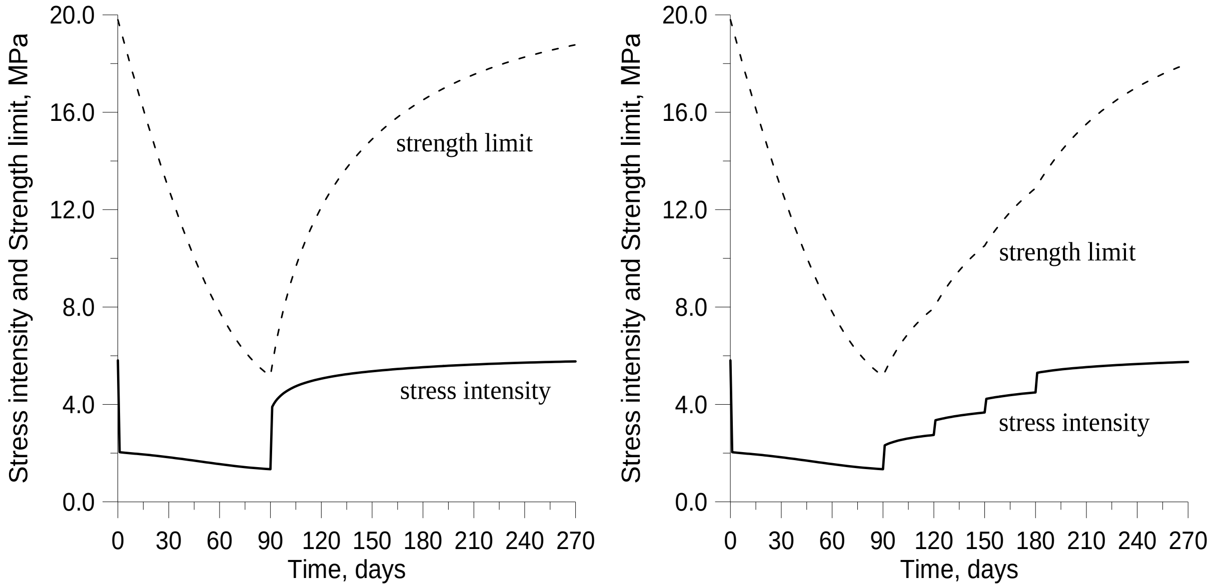
Fig. 8. Time dependence of elastic modulus of the bone tissue at the lateral side of the femur neck for two schemes of the hip joint loading variations.

To answer the question on the safety of any scheme of changing the daily loading history of the hip joint, the bone tissue fracture criterion has to be considered. In the paper by Brown et al. [9] it is presumed that such a criterion is stress-to-strength ratio or safety factor (reciprocal of stress-to-strength ratio). At approach of stress-to-strength ratio to unity, probability of bone tissue collapse critically increases. We accepted the stress intensity as a measure of a stress state and used a hypothesis that bone tissue collapse occurred in a location corresponding to some finite element, when the stress intensity in this element reached the strength limit.

It was shown in the experiments by Brown et al. [7] that the strength limits of bone tissue samples from different parts of proximal femur were proportional to their local elastic moduli. In paper by Akulich et al. [3] it was supposed that this proportion remained valid also under variations of local elastic modulus in consequence of bone tissue remodeling. The proportionality factor has been found from the condition that average sponge bone elastic modulus of 500 MPa corresponds to strength limit of 5.5 MPa (Ueo, Tsutsumi et al [10]).

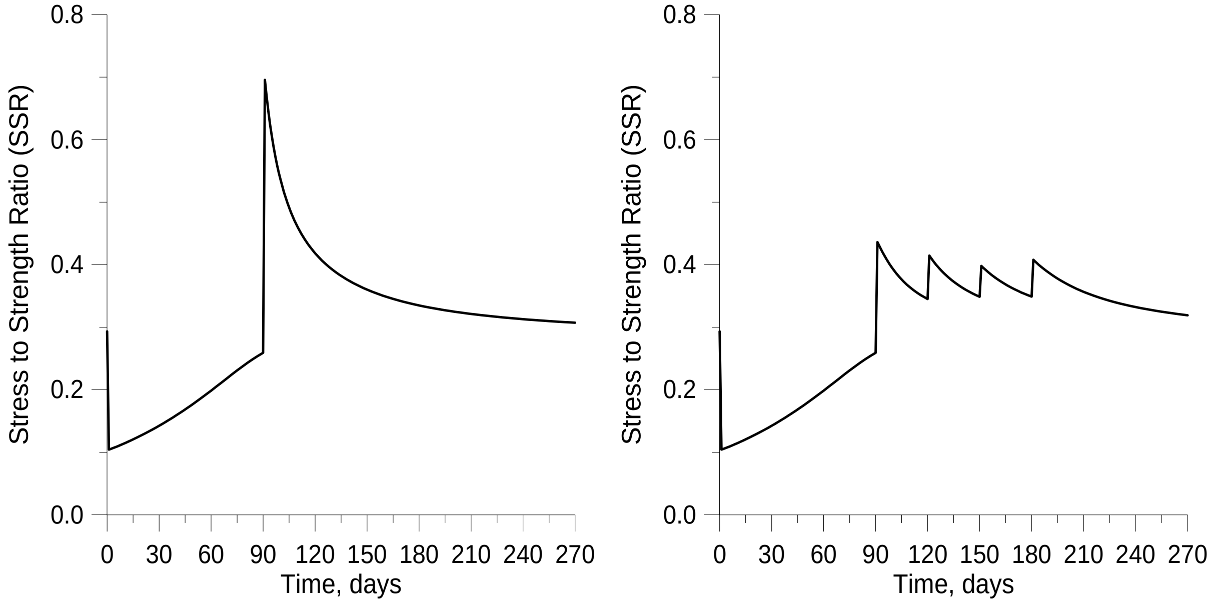
In Fig. 9 the development in time of stress intensity (solid line) and strength limit (dotted line) in the same location at the medial side of the femur neck is shown for both schemes of load change. It can be seen that at abruptly restored customary load condition (scheme A) the stress intensity only once comes nearer to the strength limit, but this does not occur under the stepwise change of load regime (scheme B).

Fig. 10 shows time development of stress-to-strength ratio in the same element. When the normal physiological load is restored abruptly (scheme A), the stress-to-strength ratio reaches the value of 0.696 that corresponds to safety factor of 1.437. But under stepwise restoring of usual daily loading history (scheme B) stress-to-strength ratio displays moderate jumps at the beginning of each stage and nowhere exceeds 0.45, that means more than twofold safety factor. Thus, the load change conforming to the scheme B really is much safer.



Scheme A Scheme B

Fig. 9. Time dependence of stress intensity (solid line) and strength limit (dotted line) at the medial side of the femur neck for two schemes of the hip joint load change.



Scheme A Scheme B

Fig. 10. . Time dependence of the stress-to-strength ratio at the medial side of the femur neck for two schemes of daily loading history change.

It has to be noted that the consoling forecast of local safety factor that comes from Fig. 10 has an essential limitation. The matter is that the stress intensity presented in Fig. 9 is an equivalent stress intensity calculated similarly to equivalent strain intensity [see Eq. (2)] by averaging the stress intensities referred to all the load cases, with taking into account their numbers of loading cycles, and the magnitude of stress-to-strength ratio was calculated by this averaged stress intensity. Meanwhile any time in each finite element an averaged stress intensity does not exist but there exists the specific stress intensity generated by one of the three load cases realized at the given time instant. This stress intensity may be greater or less than the averaged one. Therefore the local safety factor in each finite element should be defined by the very stress intensity in this element which is maximal for considered load cases. If some load case has large forces and a few loading cycles per day (might be just one cycle), this loading condition will hardly affect the bone tissue remodeling but it most likely cause the bone damage.

In Table 2 the main characteristics in the finite element considered above are presented for several time instants: time = 0 is an initial state with customary physiological load; time = 90 days is the end of the first stage with changed daily loading history; time = 91 days is the beginning of the second stage in terms of the scheme B (or the fifth stage by the scheme A), and time = 270 days is the end of the considered bone remodeling process. The magnitudes of stress-to-strength ratio and local safety factor were calculated by the averaged stress intensity, and by the stress intensities related to each load case.

The Table 2 shows that the stress intensity generated by the third load case is always the least one, and the corresponding safety factor is the greatest. On the contrary, the first load case is the most dangerous one because at any time instant the local stress-to-strength ratio referred to this load case is maximal among all the load cases and more than averaged

Table 2. The main characteristics of the stressed state and the local safety factor in the finite element at the medial side of femur neck.

			Average value	Load case 1	Load case 2	Load case 3
Time = 0	E = 1800.0 MPa, σ_s = 19.80 MPa	Stress intensity, MPa	5.807	6.887	5.073	0.768
		Stress-to-strength ratio	0.293	0.348	0.256	0.039
		Safety factor	3.413	2.874	3.906	25.64
Time = 90	E = 469.6 MPa, σ_s = 5.165 MPa	Stress intensity, MPa	1.338	4.462	3.189	0.576
		Stress-to-strength ratio	0.259	0.867	0.617	0.112
		Safety factor	3.861	**1.153**	1.620	8.928
Time = 91 days	Scheme A, E = 509.4 MPa, σ_s = 5.603 MPa	Stress intensity, MPa	3.898	4.646	3.336	0.584
		Stress-to-strength ratio	0.696	0.829	0.595	0.104
		Safety factor	1.437	**1.206**	1.681	9.615
	Scheme B, E = 483.4 MPa, σ_s = 5.317 MPa	Stress intensity, MPa	2.318	4.534	3.246	0.581
		Stress-to-strength ratio	0.436	0.853	0.610	0.109
		Safety factor	2.294	**1.172**	1.639	9.174
Time = 270 days	Scheme A, E = 1706.8 MPa, σ_s = 18.775 MPa	Stress intensity, MPa	5.767	6.831	5.083	0.715
		Stress-to-strength ratio	0.307	0.364	0.271	0.038
		Safety factor	3.257	2.747	3.690	26.32
	Scheme B, E = 1636.9 MPa, σ_s = 18.006 MPa	Stress intensity, MPa	5.745	6.800	5.068	0.708
		Stress-to-strength ratio	0.319	0.377	0.281	0.039
		Safety factor	3.135	2.653	3.559	25.44

stress-to-strength ratio plotted in Fig. 10. The local safety factor related to the first load case is the minimal among all the load cases, and less than the safety factor calculated by the averaged stress intensity.

In the initial state, when the daily loading history is normal physiological one and the first load case is prevailing in number of cycles per day, the averaged local safety factor does not differ too much from the safety factor related to the first load case. But within the first stage, when the daily loading history is changed, and the third load case becomes the prevailing one, their difference rapidly increases and reaches its maximum by the end of the stage. At this time instant (90 days) the first load case safety factor falls down to critically low level of 1.153, whereas the calculation by the averaged stress intensity yields nearly fourfold safety factor. After the numbers of different loading case applications restore their customary proportion, the difference between these safety factors decreases again.

As can be seen from Figs. 3 – 9 and Table 2, the considered bone remodeling process ends with nearly complete restoration of elastic modulus, stresses and strains in the proximal femur bone tissue. The histories of stresses and strains at the medial side of the femur neck for both schemes of this process are visualized in Fig. 11 as graphs of stress intensity versus strain intensity. The digits on the graphs mean time in days and denote the stages of the bone remodeling process.

The initial stress and strain state of the bone tissue is designated in Fig. 11 by 0. The straight-line segment from 0 to 1 corresponds to stresses and strains decrease at the beginning of the first stage, when the third load case becomes the prevailing in number of cycles (see Table 1). The decrease proceeds at the originally high elastic modulus, it is corroborated by a high angle of this segment with an abscissa. The portion of a curve from 1 to 90 represents weakening of the bone tissue at the first stage of the process that lasts 90 days. Increase of the strain intensity (Fig. 7) and the stress intensity decrease (see Fig. 9) accompany the reduction of the elastic modulus at this stage (see Fig. 8). Further, for the scheme A the straight-line segment from 90 to 91 corresponds to stresses and strains increase just after restoration of usual regime of the hip joint loading (the fifth stage in Tab. 1), and this increase proceeds at minimal value of the elastic modulus. The portion of a curve from 91 to 270 represents

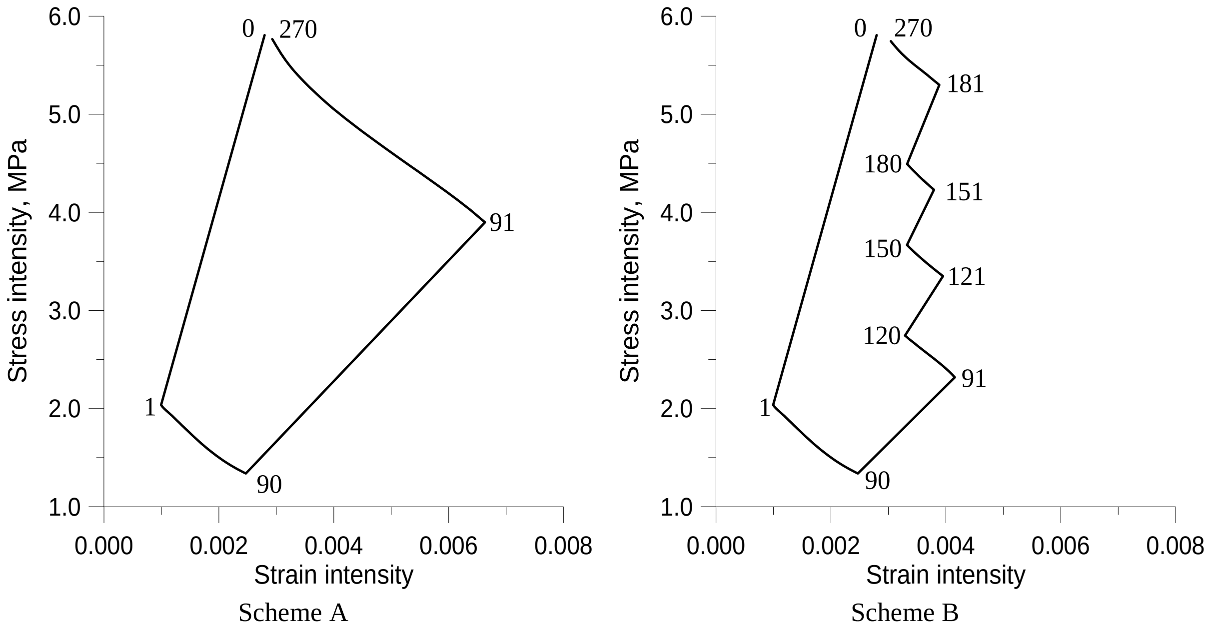


Fig. 11. The image of the process of bone tissue remodeling for two schemes of load change plotted as a graph of stress intensity versus strain intensity of deformations (the numbers on the graphs denote time in days, see explanation in the text).

strengthening of the bone tissue at the six-month final stage of the process, when the rise of the elastic modulus causes decrease of the strain intensity and increase of the stress intensity. For the scheme B, with a stepwise transition to normal physiological loading, the bone remodeling process runs similarly but it is divided in four stages, and during each subsequent stage the rise of stresses and strains proceeds at a greater value of the elastic modulus. In the both plots the point 270 that corresponds to the final state, nearly coincides with the initial point 0 (for the scheme A these points are situated some closer each to other), and this is a corroboration of practically complete recovering of the bone tissue stress and strain state.

Conclusions

In the given paper a step in the direction of a more exact quantitative description of the bone remodeling process has been made. The analysis of computer simulations proves that the phenomenological model used in the present study describes the bone tissue remodeling qualitatively correct not only when the force amplitudes of different load cases are changed (as it has been made in our articles [2, 3]), but also under the modification of the character of human physical activity which results in variations of cycle numbers of these load cases with different magnitudes and directions of forces acting on the proximal femur. The obtained data show that the architecture of the bone tissue of the hip head and femur neck is changing under the modification of the daily loading history but after the restoration of usual loading regime it gradually recovers itself.

The problem of the bone tissue safety factor under the multiple-direction loading has been discussed. While the bone tissue remodeling is dependent on the averaged action of all the load cases calculated with taking into account their numbers of loading cycles, the safety factor has to be calculated for each load case separately, and the actual safety factor is the minimal one.

Meanwhile many problems of the hip joint loading and bone tissue remodeling require further theoretical and clinical investigations, e.g. refinement of the forces acting on the hip joint during different kinds of human physical activity, experimental determination of bone remodeling rate factors in different parts of the proximal femur, the bone tissue mechanical properties and strength criteria, etc.

References

1. AKULICH Yu.V., PODGAYETS R.M. The phenomenological model of adaptable adult spongy bone tissue. **Russian Journal of Biomechanics**, 1 (1-2): 53-57, 1998.
2. AKULICH Yu.V., DENISOV A.S., PODGAYETS R.M., AKULICH A.Yu. Computer simulation of the elastic modulus of hip head bone tissue during the hip joint loading variations. **Russian Journal of Biomechanics**, 3 (1): 53-61, 1999.
3. AKULICH Yu.V., DENISOV A.S., NYASHIN Yu.I., PODGAYETS R.M., AKULICH A.Yu. The influence of the scheme of loading variations on the recovering of the bone tissue elastic modulus. **Russian Journal of Biomechanics**, 3 (3): 63-72, 1999.
4. CARTER D.R., ORR T.E., FYHRIE D.P. Relationships between loading history and femoral cancellous bone architecture. **J Biomech**, 22: 231-244, 1989.
5. SVESNSSON N.L., VALLIAPAN S., WOOD R.D. Stress analysis of human femur with implanted Charnley prosthesis. **J Biomech**, 10: 581-588, 1977.
6. FYHRIE D.P., CARTER D.R. A unified principle relating stress to trabecula: bone morphology. **J Orthop Res**, 4: 304-317, 1986.
7. BROWN T.D., WAY M.E., FERGUSON A.B., Jr. Stress transmission anomalies in femoral heads altered by aseptic necrosis. **J Biomech**, 13: 687-699, 1980.
8. BROWN T.D, FERGUSON A.B., Jr. Mechanical property distribution in the cancellous bone of the human proximal femur. **Acta Orthop Scand**, 51 (3): 429-437, 1980.
9. BROWN T.D., BAKER K.J., BRAND R.A. Structural consequences of subhondral bone involvement in segmental osteonecrosis of the femoral head. **J Orthop Res**, 10: 79-87, 1992.

10. UEO T., TSUTSUMI S., YAMAMOTO T., OKIMURA H., SHIMIZI A., NAKAMURA T. Biomechanical aspects of the development of aseptic necrosis of the femoral head. **Arch Orthop Trauma Surg**, 104: 145 – 149, 1985.

ПЕРЕСТРОЙКА КОСТНОЙ ТКАНИ ПРОКСИМАЛЬНОГО ОТДЕЛА БЕДРА ПРИ ИЗМЕНЕНИИ ХАРАКТЕРА НАГРУЗКИ НА ТАЗОБЕДРЕННЫЙ СУСТАВ

Ю.В. Акулич, Ю.И. Няшин, Р.М. Подгаец, И.А. Дмитренок (Пермь, Россия)

С помощью двумерной конечноэлементной модели проксимального отдела бедра, учитывающей адаптацию губчатой костной ткани с использованием деформационного стимула внутренней перестройки, исследуется изменение распределения модуля упругости костной ткани при изменении характера нагрузки на тазобедренный сустав. Физиологическая нагрузка рассматривается как суперпозиция нескольких дискретных вариантов нагрузки с различной величиной и направлением сил, действующих на проксимальный отдел бедра, и с различным относительным числом циклов каждого варианта нагрузки. В качестве изменения характера нагрузки на сустав рассматривается некоторое гипотетическое изменение деятельности человека, приводящее к изменению относительного вклада каждого из вариантов нагрузки. Показано, что распределение модуля упругости в головке и шейке бедра изменяется при модификации характера нагрузки и постепенно восстанавливается при возвращении к обычному режиму нагрузки. Обсуждается вопрос об определении локального запаса прочности костной ткани при многократном нагружении различными силами. Библ. 10.

Ключевые слова: внутренняя перестройка, модуль упругости, костная ткань, деформационный стимул, головка бедра, многовариантное нагружение, вариант нагрузки, число циклов нагружения

Received 19 January2001