Acta of Bioengineering and Biomechanics
Vol. 21, No. 3, 2019

Original paper
DOI: 10.5277/ABB-01387-2019-03

Analysis of fatigue loads of the knee joint during gait

ANDRZEJ A. STĘPNIEWSKI*

University of Life Sciences in Lublin, Poland.

Purpose: The aim of the work was to show that the fatigue load of bone tissue causes permanent structural changes in it. *Methods*: On the basis of the movie recording of gait, the time courses of angular changes in the joints of the lower limb were determined. Using the method of transforming Denavit–Hartenberg coordinate systems, the course of force loading the hip joint and, after that, the course of normal contact reaction of the femoral head of the knee joint during gait for the support phase were determined. On the basis of the Hertz formula, the course of contact stresses in the femoral joint head and the damage coefficient were determined according to the Palmgren–Miner damage accumulation hypothesis. *Results*: A calculation example was made using own software. The course of the obtained damage factor was compared to the image fixed in the X-ray image after its appropriate processing. The thesis of the work has been confirmed to a satisfactory degree. *Conclusions*: The nature of the lesions is similar to the image of structural changes in the head of the joint. It should be assumed that the image fixed in the bone is the result of the stored history of loads. Analysis of the obtained image can be used to determine the state of bone strength.

Key words: *knee joint, support phase, gait, accumulation of lesions, structural changes in the bone*

1. Introduction

Walking is the most frequently performed human activity. The walking cycle begins with the heel touching the ground (the beginning of the support phase), rolling the foot, fully touching the foot with the ground, rolling the foot and lifting the toes from the ground. At this point, the support phase ends and the swinging phase begins. The average number of steps per minute for slow, normal and fast walking is 70, 90, 130, respectively, and the support phase is 0.5, 0.4, 0.3 seconds, respectively, for an average step length of 0.76 m [1].

The knee joint takes over almost all loads related to statics and motion dynamics. It is also subjected to much higher loads, e.g., during running, jumps [2]. It is the most vulnerable joint for various types of injuries.

When walking, muscles and bones undergo fatigue stress. Muscle fatigue does not cause permanent and adverse changes. Properly trained muscle increases volume and increases the contraction force. The effects of these changes can only be determined by experimental methods. There are many studies on muscle fatigue. For example, on the group of 18 people, the influence of unilateral fatigue of the ankle flexor on the lower limb biomechanics was studied [3]. Another study on the group of 20 young women identified changes in kinematics of the knee, kinetics and stiffness that occur during gait due to nerve-muscle fatigue of the lower limb [4]. In many works, however, no specific conclusions from the conducted research have been formulated.

During the support phase, the joints of the lower limb are periodically loaded, causing permanent fatigue changes in the bones. The connection of the femur with the tibia, on a defined section of the joint with a length depending on the range of angular changes in the joint, is subjected to cyclic strains causing stresses in the bone tissue.

Many scientific papers concern the determination of fatigue strength of bones. Laboratory tests were conducted on human bones [5] and on animal bones [6]. The authors post the results of measurements and simu-

* Corresponding author: Andrzej A. Stępniewski, University of Life Sciences in Lublin, Akademicka 13, 20-950 Lublin, Poland. Phone: 607 414 593, e-mail: andrzej.stepniewski@up.lublin.pl

Received: May 21st, 2019
Accepted for publication: September 12th, 2019

lations using computer software [7], [8]. Research is carried out on femurs, tibias, bone fragments cooperating with neighboring bone, fragments cut from half-length, e.g., [9], cortical or spongy structure. The authors formulate various and often contradictory conclusions, mainly concerning the influence of microstructure and anisotropy on fatigue strength.

A large group of works concerns the determination of strength for various types of loads (stretching, compression, shear, bending), Young's modulus and Poisson's ratio.

The value of the Young's modulus of the trabecular bone (spongy tissue) from the femoral head is many times smaller than the Young's modulus of the cortical bone [1]. The authors of many papers give similar data on bone strength for bending [10], also for various types of loads together with the value of Young's modulus.

Joint loads consist of the body's gravity and inertia forces caused by the movement of the body mass. The inertia forces depend on the acceleration of the movement of the center of mass of the body, which depends on the gait speed, individual gait specifity and anatomic dimensions. Kinematic gait parameters can be determined by simultaneous use of theoretical and experimental methods. To model the kinematics of walking, it is necessary to experimentally determine the course of time changes in angles in individual joints of the lower limb. The results of such measurements can be found in many works, e.g., [11], [12]. A wide variation in the results obtained can be observed, which indicates the individual character of gait.

Static, dynamic and fatigue bone loads are sure to cause deformations and stresses in the bone tissue. One can risk the hypothesis that changes in the bone structure are largely permanent and each subsequent load causes the accumulation of these changes. In other words, the history of loads is recorded in bone tissue. The analysis of these changes by medical specialists could be useful for assessing bone wear, existing diseases and movement abnormalities.

The paper attempts to authenticate the hypothesis that can be accepted or rejected.

2. Methods

To obtain data for calculations, a digital camera KENOX S860/Samsung with the possibility of filming was used. In the calculations, the method of transforming Denavit–Hartenberg coordinate systems and the Palmgren–Miner fatigue cumulation hypothesis were used. The obtained results of the calculations were compared with the record obtained after appropriate processing of the image of the knee fixed on X-ray plate. The following order of procedure was used:

1. Motion of the lower limbs was filmed during normal gait.
2. On the basis of the film record, the coordinates of the extreme points of angular positions in the hip, knee and ankle joints were determined.
3. Using a polynomial approximation, time courses of angular displacements in the joints were determined, followed by time velocity and acceleration waveforms.
4. Using the method of transforming Denavit–Hartenberg coordinate systems, the acceleration of vertical body mass motion was determined in order to determine the value of the inertia force of the body mass acting on the hip joint.
5. Using the method of transforming Denavit–Hartenberg coordinate systems, the course of normal contact reaction of the femoral head of the knee joint during gait for the support phase was determined.
6. On the basis of the Hertz formula, the course of contact stresses in the femoral head of the femoral joint was determined.
7. Using of the Palmgren–Miner damage cumulation hypothesis, the course of the damage factor along the femoral head was determined.
8. The course of the damage coefficient was compared with the image obtained after the X-ray image was processed.

In this work, due to the high complexity of the biological structure, the joint was treated as a simplified flat mechanism. The same geometry and the same load on both condyles were assumed. The influence of the joint support phase and pelvic movements was omitted. Elements of the joint were treated as perfectly rigid. Many factors that affect the proper functioning of the joint, such as spatial movement or the presence of synovial liquid, have not been taken into account, therefore the results obtained should be regarded as approximate. The theoretical analysis of the knee joint described in the work was used [13]. The results obtained were analyzed and conclusions were formulated.

2.1. Displacements, velocities and angular accelerations in joints

The movement of the lower limb in the sagittal plane was filmed during normal gait (about 90 steps per minute). Markers were affixed to the extremity in the joints axes and in the heel and toe area of the foot. Based on 30 film records, the values of five extreme angular positions in the hip, knee and ankle joints were determined.

The determined coordinates of the extreme points $P_{k,i}(\beta_{k,i},\ t_{k,i})$ were matched to the polynomial. The coefficients $a_{k,n}$ were determined from the equations of matching the polynomial of the degree $n = 6$ to the set of five extreme points with the coordinates $P_{k,i}(\beta_{k,i},\ t_{k,i})$ and the condition of zeroing of derivatives.

$$\begin{bmatrix} \beta_{k,1}-\beta_{k,0} \\ \ldots \\ \beta_{k,4}-\beta_{k,0} \\ 0 \\ \ldots \\ 0 \end{bmatrix} = \begin{bmatrix} t_{k,1}^2 & t_{k,1}^3 & t_{k,1}^4 & t_{k,1}^5 & t_{k,1}^6 \\ \ldots & \ldots & \ldots & \ldots & \ldots \\ t_{k,4}^2 & t_{k,4}^3 & t_{k,4}^4 & t_{k,4}^5 & t_{k,4}^6 \\ 2t_{k,1} & 3t_{k,1}^2 & 4t_{k,1}^3 & 5t_{k,1}^4 & 6t_{k,1}^5 \\ \ldots & \ldots & \ldots & \ldots & \ldots \\ 2t_{k,4} & 3t_{k,4}^2 & 4t_{k,4}^3 & 5t_{k,4}^4 & 6t_{k,4}^5 \end{bmatrix} \begin{bmatrix} a_{k,2} \\ a_{k,3} \\ a_{k,4} \\ a_{k,5} \\ a_{k,6} \end{bmatrix}, \tag{1}$$

where:

$a_{k,n}$ – coefficients of adjustment of the sixth degree polynomial to changes in the angle of the joints, k takes values: 1 – hip joint, 2 – knee joint, 3 – ankle joint,

$t_{k,i}$ – time of reaching the extreme point i in the joint k,

$\beta_{k,i}$ – angle value in joint k at time $t_{k,i}$,

$\beta_{k,0}$ – angle value in joint k at the moment of beginning of the support phase (in time $t_{k,i} = 0$, $i = 0$).

The equations were solved by the Gaussian elimination method. Having calculated values of coefficients $a_{k,n}$, displacements, velocities and angular accelerations in joints were determined:

$$\begin{aligned} \beta_k(t) &= \beta_{k,0} + [t^2 \quad t^3 \quad t^4 \quad t^5 \quad t^6] \\ &\quad \cdot [a_{k,2} \quad a_{k,3} \quad a_{k,4} \quad a_{k,5} \quad a_{k,6}]^T, \\ \dot{\beta}_k(t) &= [t \quad t^2 \quad t^3 \quad t^4 \quad t^5] \\ &\quad \cdot [2a_{k,2} \quad 3a_{k,3} \quad 4a_{k,4} \quad 5a_{k,5} \quad 6a_{k,6}]^T, \\ \ddot{\beta}_k(t) &= [1 \quad t \quad t^2 \quad t^3 \quad t^4] \\ &\quad \cdot [2a_{k,2} \quad 6a_{k,3} \quad 12a_{k,4} \quad 20a_{k,5} \quad 30a_{k,6}]^T. \end{aligned} \tag{2}$$

2.2. Time course of the force loading the hip joint

The main load on the hip joints during gait is the sum of the torso weight force and the inertia force of the torso mass center. The kinematics of the center of mass depends on the time course of changes in the positions in the kinematic pairs of the three joints – hip, knee and ankle. It also depends on the geometric dimensions of the limbs, speed and type of gait. The time course of force can be determined empirically by measuring the foot pressure on the ground using force sensors placed in the footwear. Another possible way to obtain data to determine the inertia force is to film the motion of the center of mass and perform a double differentiation of the course of vertical displacement versus time. Because of the small displacement of the center of mass in the vertical axis, this method is not very precise. The vertical movement of the center of mass can be determined more accurately (due to larger displacements) based on temporary displacements in the three joints of the lower limb. The method of homogenous transformations of Denavit–Hartenberg coordinate systems was used to determine the vertical displacements of the body during gait. The system $\{x_0, y_0\}$ associated with the hip joint was assumed as the reference system (Fig. 1).

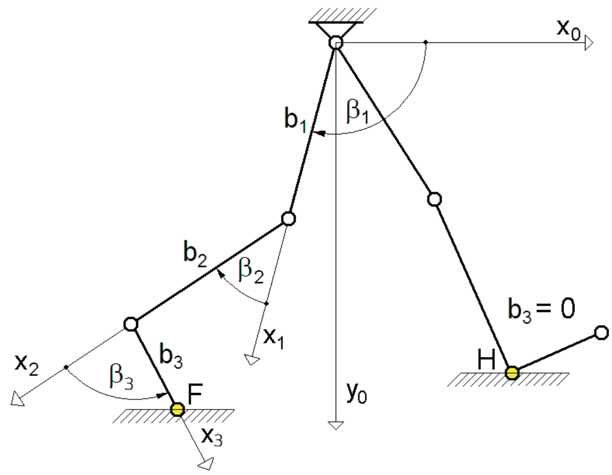

Fig. 1. Lower limb in Denavit–Hartenberg coordinate systems

The matrices of transformations of coordinate systems will have the form

$$\mathbf{B}_k = \begin{bmatrix} c_k & -s_k & b_k c_k \\ s_k & c_k & b_k s_k \\ 0 & 0 & 1 \end{bmatrix}, \quad \text{where } k = 1, 2, 3, \tag{3}$$

wherein $s_k = \sin\beta_k$, $c_k = \cos\beta_k$, b_k – length of the element k.

The vector of the start position of the coordinate system associated with the foot (point F of the end of the large finger) relative to the system assumed to be immobile $\{x_0, y_0\}$ determines the relationship

$$\mathbf{r}_{F,0} = \left(\prod_{i=1}^{i=n} \mathbf{B}_i \right) \mathbf{r}_{F,F} , \tag{4}$$

where $\mathbf{r}_{F,F} = [0 \quad 0 \quad 1]^T$.

After performing the actions defined by the dependence (4), you get

$$\mathbf{r}_{F,0} = \begin{bmatrix} b_1c_1 + b_2c_{12} + b_3c_{13} \\ b_1s_1 + b_2s_{12} + b_3s_{13} \\ 1 \end{bmatrix} , \tag{5}$$

wheras

$$c_{12} = \cos(\beta_1 + \beta_2),\ \ c_{13} = \cos(\beta_1 + \beta_2 + \beta_3),$$
$$s_{12} = \sin(\beta_1 + \beta_2),\ \ s_{13} = \sin(\beta_1 + \beta_2 + \beta_3) .$$

Acceleration of the center of mass of the torso in the vertical axis is determined by double differentiation of the second row of the matrix (5)

$$a_{F,0}(t) = [b_1 \quad b_2 \quad b_3] \begin{bmatrix} \ddot{\beta}_1 c_1 - \dot{\beta}_2^2 s_1 \\ \ddot{\beta}_{12} c_{12} - \dot{\beta}_{12}^2 s_{12} \\ \ddot{\beta}_{13} c_{13} - \dot{\beta}_{13}^2 s_{13} \end{bmatrix} , \tag{6}$$

wherein

$$\dot{\beta}_{12} = \dot{\beta}_1 + \dot{\beta}_2,\ \ \dot{\beta}_{13} = \dot{\beta}_1 + \dot{\beta}_2 + \dot{\beta}_3,$$
$$\ddot{\beta}_{12} = \ddot{\beta}_1 + \ddot{\beta}_2,\ \ \ddot{\beta}_{13} = \ddot{\beta}_1 + \ddot{\beta}_2 + \ddot{\beta}_3.$$

The acceleration determined by the dependence (6) is right after the heel is detached from the ground (point H). Until the heel is detached from the ground, the calculation is supplemented with a condition, if

$$|b_1s_1 + b_2s_{12} + b_3s_{13}| \le |b_1s_1 + b_2s_{12}| \Rightarrow b_3 = 0 . \tag{7}$$

The course of the force loading the hip joint during the walk in the support phase can be defined as

$$F_G(t) = m_t[g + a_{F,0}(t)] , \tag{8}$$

where:
g – gravitational acceleration,
m_t – torso mass.

2.3. The course of the contact reaction of the femoral head

The analysis was carried out for a simplified, flat model of the knee joint mechanism. In this case, for the permanent system $\{x_0, y_0\}$, a system was adopted whose x_0 axis passes through the attachment points of the cruciate ligaments to the tibia bone (Fig. 2). To transform vector of $\mathbf{r}_{oz,n}$ (points with the designation "oz"), defined in the systems "n" in one of the vectors $\mathbf{r}_{oz,0}$ relative to the fixed system $\{x_0, y_0\}$, the method of transforming Denavit–Hartenberg coordinate systems was also used

$$\mathbf{r}_{oz,0} = \left(\prod_{i=1}^{i=n} \mathbf{A}_i \right) \mathbf{r}_{oz,0} , \tag{9}$$

wherein the matrix $\mathbf{A}_i$ defined with respect to the angles Θ_i and lengths of the elements l_i has the form analogous to the matrix (3), $\mathbf{r}_{oz,0} = [x_{oz,0} \quad y_{oz,0} \quad z_{oz,0}]^T$ – the vector of the position of the point with the designation "oz" with respect to the coordinate system "n".

The coordinates of the position vector of the kinematic pair C (O_2) in the stationary system are expressed as follows

$$\begin{bmatrix} x_{O2,0} \\ y_{O2,0} \\ z_{O2,0} \end{bmatrix} = \begin{bmatrix} c_{12}l_2 + l_1c_1 \\ s_{12}l_2 + l_1s_1 \\ 0 \end{bmatrix} , \tag{10}$$

where:
$s_{12} = \sin(\Theta_1 + \Theta_2)$,
$c_{12} = \cos(\Theta_1 + \Theta_2)$,
$l_1 = l_{AB}$,
$l_2 = l_{BC}$.

Based on Fig. 2, dependency is possible

$$\Theta_1 + \Theta_2 = \beta_2 + \Psi + 0{,}5\pi , \tag{11}$$

where: Ψ – fixed angle, contained between the line u parallel to the femur and axis x_2.

Based on Fig. 2, it can be written as

$$l_3^2 = (l_4 - x_{O2,0})^2 + {y_{O2,0}}^2 , \tag{12}$$

where:
$l_3 = l_{CD}$,
$l_4 = l_{AD}$.

After taking into account dependence (10) and (12), the following equation was obtained

$$D_1 = A_1c_2 + B_1s_2 , \tag{13}$$

hence the values of angles are determined Θ_1 and Θ_2

$$\Theta_1 = 2\text{atan2}(B_1 + \sqrt{A_1^2 + B_1^2 - D_1^2}, A_1 + D_1) , \tag{14}$$

$$\Theta_2 = 2\text{atan2}(B_2 + \sqrt{A_2^2 + B_2^2 - D_2^2}, A_2 + D_2) , \tag{15}$$

where:
$A_1 = 2l_1(l_2c_{12} - l_4)$,

$B_1 = 2l_1l_2s_{12}$,

$D_1 = -l_1^2 - l_2^2 + l_3^2 - l_4^2 + 2l_2l_4c_{12}$,

$A_2 = 2l_2(l_1 - l_4c_1)$,

$B_2 = 2l_2l_4s_1$,

$D_2 = -l_1^2 - l_2^2 + l_3^2 - l_4^2 + 2l_1l_4c_1$.

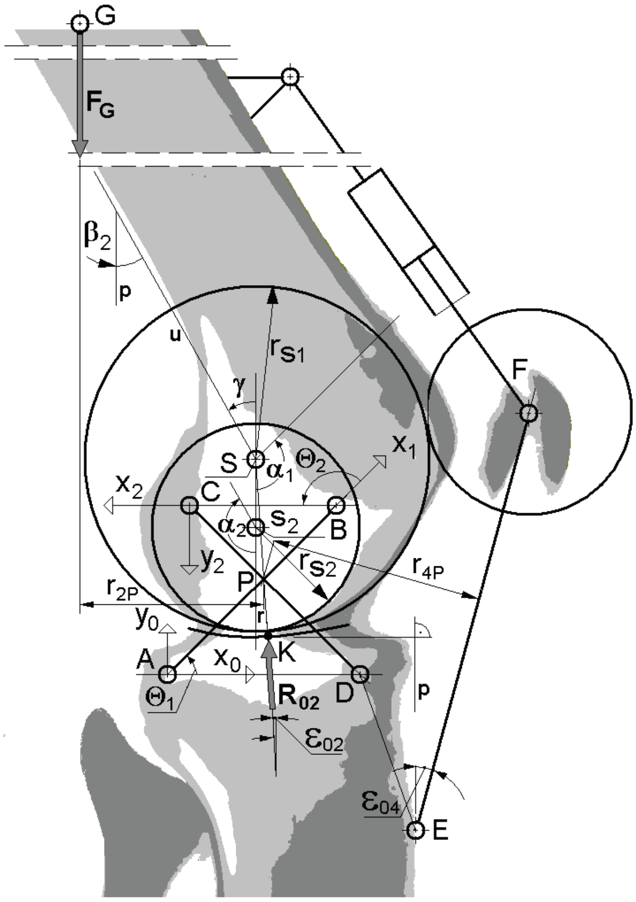

Fig. 2. Knee joint; geometry, dimensions, reaction of normal

The reaction of the bone contact $R_{02}(\beta_2)$, was determined from the equation of the equilibrium of forces in the direction determined by this reaction and the equation of the sum of moments of forces

$$R_{02}(\beta_2, t) = F_G(t)\left[\cos\varepsilon_{02} + \frac{r_{2P}}{r_{4P}}\cos(\varepsilon_{04} - \varepsilon_{02})\right], \quad (16)$$

where:

$r_{2P} = x_{P,0} - x_{G,0}$,

$$r_{4P} = \frac{\left|a_{E,F}(x_{P,0} - x_{E,0}) - y_{P,0} + y_{E,0}\right|}{\sqrt{1 + a_{E,F}^{\;2}}},$$

$\varepsilon_{02} = 0.5\pi - \operatorname{atan} a_{K,S}$, $\varepsilon_{04} = 0.5\pi - \operatorname{atan} a_{E,F}$,

wherein $a_{E,F}$ i $a_{K,S}$ are directional coefficients of straight lines passing through the points, respectively, $E(x_{E,0}, y_{E,0})$, $F(x_{F,0}, y_{F,0})$ i $K(x_{K,0}, y_{K,0})$, $S(x_{S,0}, y_{S,0})$.

2.4. Cumulation of stresses

Assuming simplifications resulting from Hertz theory, the contact stresses in the case of mutual pressure of two pairs of spherical surfaces (convex with concave) determine the relationship [14]. Assuming that both joints of the joint are loaded evenly, the relationship will take the form

$$\sigma_H(\beta_2, t) = \frac{1}{\pi}\sqrt[3]{0.75R_{02}\left[\frac{E(r_{S1/S2} - r_P)}{r_{S1/S2}r_P(1-\nu^2)}\right]^2}, \quad (17)$$

where:

E – Young's modulus of bones,

$r_{S1/S2}$ – the radius of the femoral knee joint r_{S1} or r_{S2} depending on the angular position in the knee joint β_2,

r_P – the radius defined by the contact point K of the femoral head with the tibia bone,

ν – Poisson's coefficient of bone,

$\sigma_H(\beta_2, t)$ – contact stress in the bones of the knee joint according to Hertz as a function of time t and angle β_2 in the joint.

The relationship between the amplitude of stress σ_H and the number of N_f cycles causing damage is represented by the Wöhler curve. In double logarithmic coordinates, they can be approximated with a straight line and apply the Basquin equation [15]

$$\sigma_H = R_c(2N_f)^b, \quad (18)$$

b – exponent of fatigue strength ranging from –0.15 to –0.05,

N_f – number of load cycles to fatigue crack for tension with σ_H amplitude,

R_c – limit breaking stress after one cycle in static compression.

From the equation, the number of cycles N_f load σ_H causing fatigue fracture can be determined

$$N_f = 0.5\left(\frac{\sigma_H}{R_c'}\right)^{\frac{1}{b}}. \quad (19)$$

The stress $\sigma_H(\beta_2, t)$ is calculated with the assumed step of angle $\beta_2(t)$. The whole range of changes of $\beta_2(t)$ angle is divided into p ranges from p = 1 to p = k. In a given interval p angle $\beta_{2,p}(t)$ occur from i = 1 to n_p load cycles with different stress amplitude $\sigma_{H,p}(\beta_2)$. According to the Palmgren–Miner of stress summation hypothesis for a given interval p

$$D_p = \sum_{i=1}^{n} \frac{n_p}{N_{f,p}}, \quad (20)$$

D_p – damages after n_p stress cycles in the interval p with the amplitude $\sigma_{H,p}(\beta_2)$,

n – the number of phases of the limb support,

n_p – number of repetitions of the same range of changes of β_2 angle in the interval p during the phase of supports with different values of $\sigma_{H,p}(\beta_2)$,

$N_{f,p}$ – number of cycles in the interval p to the fatigue crack for tension with amplitude $\sigma_{H,p}(\beta_2)$.

Finally, based on the dependence (17), (19), (20), damage D_p after n_p stress cycles in successive intervals p as a function of the angle $\beta_{2,p}(t)$ determines the dependence

$$D_p(\beta_{2,p},t) = \sum_{i=1}^{n} 2n_p \left(\frac{1}{\pi R_c} \sqrt[3]{0.75 R_{02} \left[\frac{E(r_{S1/S2} - r_P)}{r_{S1/S2} r_P (1-\nu^2)} \right]^2} \right)^{-\frac{1}{b}} . \quad (21)$$

Assuming the value of the fatigue strength exponent $b = -0.1$ is obtained

$$D_p(\beta_{2,p},t) = 8.17 \cdot 10^{-6} \sum_{i=1}^{n} \left(\frac{1}{R_c} \right)^{10} \left(\frac{E}{1-\nu^2} \right)^{\frac{20}{3}} \cdot n_p R_{02}^{\frac{10}{3}} \left(\frac{r_{S1/S2} - r_P}{r_{S1/S2} r_P} \right)^{\frac{20}{3}} . \quad (22)$$

After substituting the values of endurance parameters for the cortical bone: Young's modulus E = 20 GPa, limit breaking stress R_c = 800 MPa, Poisson's ratio $\nu = 0.39$, dependence (22) takes the form of

$$D_p(\beta_{2,p},t) = 10^{-5} \sum_{i=1}^{n} n_p R_{02}^{\frac{10}{3}} \left(\frac{r_{S1/S2} - r_P}{r_{S1/S2} r_P} \right)^{\frac{20}{3}} . \quad (23)$$

2.5. The possibility of verification of calculation results

Loads cause structural changes in the bones. The intensity of changes depends on the value and frequency of the loads. Certainly walking is the most frequently performed activity, so it should cause noticeable structural changes in the bones. These changes can be read after appropriate X-ray image processing according to the method described in patent application PL413247 [16]. The method is very simple. X-ray image made on film should be highlighted with polarized light. Then the photo should be photographed with a digital camera. As a result, a color photo with clear boundaries separating the individual colors is obtained. Color saturation can be increased by any program for digital photo processing, e.g., the popular *Picture Manager*. Each of the colors represents a different bone structure and allows you to assess the changes and limits of changes in bone structure.

This work uses the described method to verify the obtained results in terms of the nature and geometry of structural changes.

2.6. Numerical example

Based on 30 gait films, arithmetic mean values of the extreme angular positions in the joints of the lower limb were determined. The results of measurements are given in Table 1.

The following data was used for the calculations:

Torso weight m_t = 60 kg.

Lower limb: $b_1 = 45 \cdot 10^{-2}$ m, $b_2 = 43 \cdot 10^{-2}$ m, $b_3 = 22 \cdot 10^{-2}$ m.

Geometry of the knee joint: $l_{AB} = 4{,}0 \cdot 10^{-2}$ m, $l_{BC} = 2{,}3 \cdot 10^{-2}$ m, $l_{CD} = 4{,}5 \cdot 10^{-2}$ m, $l_{AD} = 3{,}8 \cdot 10^{-2}$ m, $l_{EF} = 8{,}3 \cdot 10^{-2}$ m, $r_{S1} = 4{,}4 \cdot 10^{-2}$ m, $r_R = 2{,}4 \cdot 10^{-2}$ m, $r_P = 5 \cdot 10^{-2}$ m, $\alpha_1 = 110°$, $\alpha_2 = 160°$, $\gamma = 28°$, $\Psi = 52°$.

Coordinates of the fastening point of the patellar tendon in the $\{x_0, y_0\}$ system:

$x_{E,0} = 4{,}5 \cdot 10^{-2}$ m, $y_{E,0} = -3{,}0 \cdot 10^{-2}$ m.

The coordinates of the points relative to the $\{x_2, y_2\}$ system related to the femur:

$x_{S1,2} = -1{,}4 \cdot 10^{-2}$ m, $y_{S1,2} = -2{,}2 \cdot 10^{-2}$ m,
$x_{S2,2} = -2{,}0 \cdot 10^{-2}$ m, $y_{S2,2} = -0{,}2 \cdot 10^{-2}$ m,
$x_{G,2} = 25 \cdot 10^{-2}$ m, $y_{G,2} = -32 \cdot 10^{-2}$ m,
$x_{H,2} = -4{,}5 \cdot 10^{-2}$ m, $y_{H,2} = -10{,}5 \cdot 10^{-2}$ m.

Calculations have been made for 5 million cycles n.

Table 1. Values of extreme angular positions in joints (in degrees)

Time $t_{k,i}$ [s]	0	0.030	0.055	0.165	0.200	0.250	0.300	0.330	0.400
Hip joint $\beta_{1,i}$ [°]	55*	58	–	–	96	–	–	103	105
Knee joint $\beta_{2,i}$ [°]	8	–	20	4	–	–	60	–	0
Ankle joint $\beta_{3,i}$ [°]	–90	–	–	–102	–	–75	–	–96	–90

* Value in accordance with the dimensioning method adopted by Denavit–Hartenberg.

3. Results

The results of the calculations are presented in Figs. 3 and 4. The temporal courses of angular displacements in the joints were obtained on the basis of the data in Table 1 by fitting to the 6th degree polynomials according to the relation (1). Figure 3a shows the temporal courses: of force loading the hip joint F_G, of the angle at the knee joint β_2 obtained on the basis of the relationship (8), the reaction R_{02} obtained on the basis of the relationship (16). Figure 3b shows the course of reaction R_{02} as a function of the angle β_2 and the a course of the damage accumulation coefficient $D_p(\beta_{2,p})$ (points) for the next 29 intervals of the angle β_2 calculated on the basis of the relationship (24). Figure 4 shows the course of $D_p(\beta_{2,p})$ along the curvature of the bone head and comparison with the image recorded in the X-ray. The amplitude was adjusted by the use of a graphic scale.

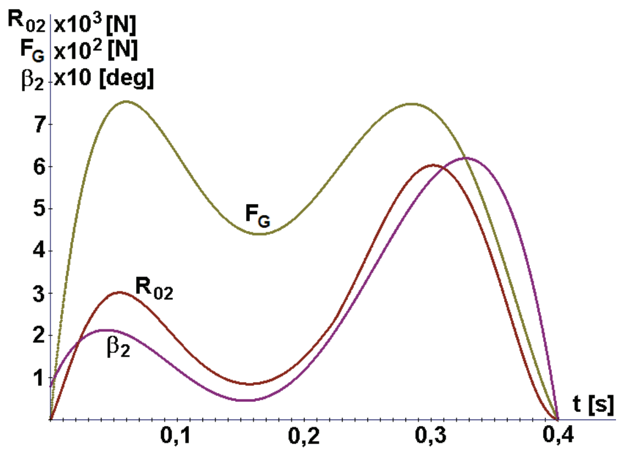

Fig. 3a. Time courses of the joint angle $\beta_2(t)$, of the force loading hip joint $F_G(t)$ and reaction $R_{02}(t)$

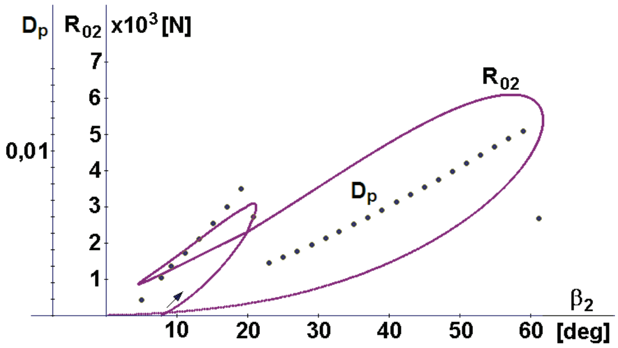

Fig. 3b. The courses of $R_{02}(\beta_2)$ and $D_p(\beta_2)$ in the subsequent intervals of p

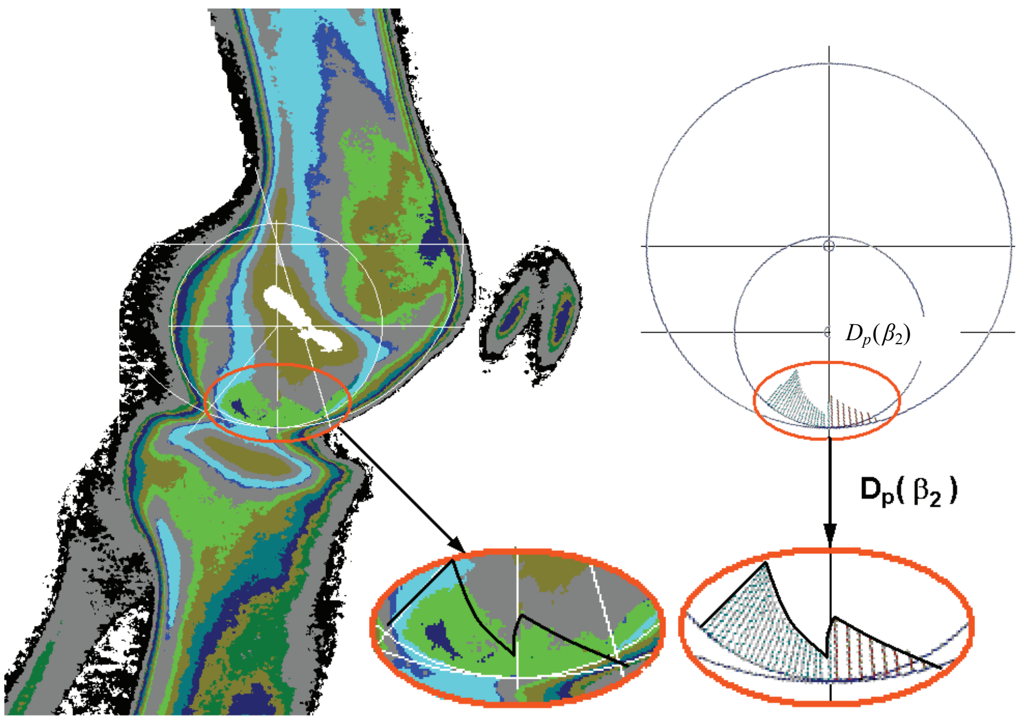

Fig. 4. Comparison of the course of $D_p(\beta_{2,p})$ along the curvature of the head of the bone with the image obtained after processing the X-ray of the knee joint

4. Discussion

On the basis of the published works and own measurements, one can observe a large variation in the results of changes in angles in the joints of the lower limb during gait. The way of walking and the method of loading the knee joint is an individual feature. Therefore, in this work all data and results concern the same person.

In one support cycle, the same point of contact (the same value of angle β_2) can be loaded once, twice times, three times and even four times with forces of different values (Fig. 3b). In order to adjust the waveform to the limits separating the colors (Fig. 4), the scale of the amplitude of the calculated waveform $D_p(\beta_{2,p})$ was selected.

Authors of many works confirm the relationship between load and structural changes inside the bone, e.g., [6], [8]. However, it is difficult to find a study on the course and nature of these changes, so that they can be used to compare with the results obtained in this work.

On the basis of other fatigue strength tests in three main types of loads (stretching, compression and shearing), human bone obtained from cadavers of six individuals of different ages, showed that fatigue strength of cortical bone refers to its internal architecture and certain physical features [5]. In addition, the work of other authors draws attention to the geometry of the bone structure. For example, on the basis of animal bone tests, it was found that the rate of crack propagation is lower when the internal microstructure is parallel to the load [6].

In their works, the authors formulate conclusions that the effect of fatigue loads on the bone adversely affects their strength properties and the greater the loads, the greater the degree of destruction of the "material". However, it should be noted that conducting strength tests is only possible on dead bones. In the case of dead bones, the conclusions can be considered right. In the case of living tissue, the conclusions formulated contradict the generally known principle that training and proper rehabilitation cause an improvement in the endurance status of bones and muscles.

It should also be noted that the applied loads in laboratory tests are not adequate to the most frequently occurring real loads in the human bone system. Fatigue strength was tested, among others on shear [5]. The shear force during compression reaches its maximum value in the plane inclined at an angle of 45 degrees with respect to the direction of the compressive force, while in the plane perpendicular to the direction of the compressive force, the shear stresses are equal to zero.

Using the computer simulation method (finite element method), the "multi-axial load effects" in normal walking conditions were studied, which is not present in the so-called real layout. The conclusion was that fatigue life under the multi-axial load is higher (by 60–80%) compared to uniaxial and does not depend on the degree of anisotropy. At the same time, it was also found that the use of multi-axial loads reduced fatigue life five times [7]. The authors confirmed Galileo's well-known and the oldest endurance hypothesis which indirectly states that the strength of the material is determined by its tensile strength, while compression (triaxial) is incomparably greater. In another work with the help of ANSYS, the fatigue strength of selected areas of the tibia bone was examined. It was found that the highest loads are in the upper part of the bone, in the area of contact with the femur. The authors conclude that fatigue durability has increased due to the reduction of bone burden. Thus, they suggest the patient to rest, reduce body weight in parallel and not lift weights [8]. The applications are obvious and no research is needed to formulate them. There are no conclusions regarding the ability of living tissue to regenerate.

Despite far-reaching simplifications of the knee joint model, the waveform "found its reflection" in the obtained color limits. The course obtained is the course of accumulation of defects, however, in the case of biological living tissue, the result can not be treated as an image of permanent damage. In the case of live tissue, any acceptable load results in beneficial changes leading to the strengthening of the fragment being trained. Therefore, the image obtained after processing the X-ray image can also be interpreted as an image of bone tissue strengthening.

The final confirmation of the correctness of these interpretations belongs to specialists in the field of traumatology and orthopedics. The hypotheses and own thoughts were used in the work, therefore it should be emphasized again that the correctness of formulated assumptions can be questioned.

5. Conclusions

Based on the analysis, the following conclusions can be formulated:

1. The image obtained on the basis of the X-ray image is similar to the image obtained by the known

method of elasto-optics and presumably presents the distribution of fixed structural changes in bone tissue.

2. The determined course of changes in the normal contact reaction can be "adjusted" to the line delimiting areas with different ranges of stress values (borders separating colors), which with a high probability confirms the thesis that the history of loads is recorded in of the tissue of bone.

References

[1] HERMAN I.P., *Physics of the Human Body*, Springer, Berlin, 2007.

[2] MRDAKOVIC V., PAZIN N., VULOVIC R., FILIPOVIC N., ILIC D., *Neuromechanical control in submaximal drop jumps: the effects of volitional effort demands and drop height magnitude on soleus muscle activation*, Acta Bioeng. Biomech., 2018, Vol. 20, No. 4, 101–111, DOI: 10.5277/ABB-01210-2018-02.

[3] HUNT M.A., HATFIELD G.L., *Ankle and knee biomechanics during normal walking following ankle plantarflexor fatigue*, Journal of Electromyography and Kinesiology, 2017, Vol. 35, 24–29, https://doi.org/10.1016/j.jelekin

[4] LONGPRÉ H.S., POTVIN J.R., MALY M.R., *Biomechanical changes at the knee after lower limb fatigue in healthy young women*, Clinical Biomechanics, 2013, Vol. 28, Issue 4, 441–447, https://doi.org/10.1016/j.clinbiomech

[5] ZIOUPOS P., GRESLE M., WINWOOD K., *Fatigue strength of human cortical bone: age, physical, and material heterogeneity effects*. J. Biomed. Mater Res. A., 2008, 86 (3), 627–636, DOI: 10.1002/jbm.a.31576.

[6] KIM J.H., NIINOMI M., AKAHORI T., TODA H., *Fatigue properties of bovine compact bones that have different microstructures*, Int. J. of Fatigue, 2007, 29 (6), 1039–1050, DOI: 10.1016/j.ijfatigue.2006.09.018.

[7] FATIHHI S.J., HARUN M.N., KADIR M.R.A., ABDULLAH J., KAMARUL T., CHSNER A.O., SYAHROM A., *Uniaxial and Multiaxial Fatigue Life Prediction of the Trabecular Bone. Based on Physiological Loading*, A Comparative Study Annals of Biomedical Engineering, 2015, DOI: 10.1007/s10439-015-1305-8.

[8] KUMAR SRIVASTAVA R., NIZAMULLA S., JAGADESH KUMAR J., RAVI TEJA G., *Fatigue life prediction of tibia and fibula bones using finite element method*, International Journal of Advance Engineering and Research Development (IJAERD), 2017, Vol. 4, Issue 11, 2348–6406. e-ISSN (O): 2348-4470 p-ISSN (P).

[9] KIM J.H., NIINOMI M., AKAHORI T., TAKEDA J., TODA H., *Effect of Microstructure on Fatgue Strength of Bovine Compact Bones*, JSME International Journal, Series A, 2005, Vol. 48, No. 4, 472–480, DOI: 10.1299/jsmea.48.472.

[10] RABIJ K., KAJZER A., KAJZER W., *Examination of mechanical properties of bovine*, Current Problems of Biomech., Silesian University of Technology, 2015, No. 9, 105–110, ISSN 1898-763X (in Polish).

[11] BŁAŻKIEWICZ M., WIT A., *Artificial neural network simulation of lower limb joint angles in normal and impaired human gait*, Acta Bioeng. Biomech., 2018, Vol. 20, No. 3, 43–49, DOI: 10.5277/ABB-01129-2018-02.

[12] VAN CRIEKINGE T., TRUIJEN S., HALLEMANS A., SAEYS W., *The influence of a thoracolumbosacral orthosis on gait performance in healthy adults during walking*, Acta Bioeng. Biomech., 2018, Vol. 20, No. 4, 15–21, DOI: 10.5277/ABB-01201-2018-02.

[13] STĘPNIEWSKI A.A., *Structural, geometric and static analysis of the mechanism of the human knee joint*, Acta Mechanica et Automatica (AMA), Białystok 2010, Vol. 4, No. 1, OWPB, 80–85, ISSN 1898-4088 (in Polish).

[14] JASTRZĘBSKI P., MUTERMILCH J., ORŁOWSKI W., *Strength of materials*, Part 2, Arkady, Warszawa 1985 (in Polish).

[15] KOCAŃDA S., SZALA J., *Basics of fatigue calculations*, PWN, Warszawa 1985 (in Polish).

[16] STĘPNIEWSKI A.A., *The system for assessing the bone strength status and the way of conducting the assessment*, Patent Application No P.413247, BUP 03/2017, p. 46 (in Polish).