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Anatomical and biomechanical justification and role of the residual angle of deviation of the lower leg in unicompartmental knee replacement

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Abstract. ***Aim:*** to calculate the optimal values of the residual angle of the deviation of the lower leg during unicompartmental knee replacement by mathematical modeling. ***Object and methods of research.*** For mathematical modeling, a calculated load scheme of the knee joint with a single-pole endoprosthesis was chosen, in which the femur is represented by a structure with two points of articulated fixation of the support. The value of the force F was assumed to be 700 N. The angle of the inclination g of the force F was changed from 0 to 20° in the direction of the varus and valgus deformations in increments of 2°. The simulation used a method for calculating statically indeterminate systems. Calculations were performed in a package of application programs for solving technical computing problems MathLab R2012a. ***Results.*** The residual angle of the deviation of the lower leg during unicompartmental endoprosthetics leads to a significant redistribution of loads between the femoral condyles in direct proportion to the value of the angle of inclination for a stable joint and non-linearly in case of lack of stability. It is proved that it is relatively safe to deviate the angle in the range of ±2–3°, and the exceeding of this indicator leads to the appearance of horizontal components of the reaction forces of the support with a significant value. ***Conclusion.*** The calculation of the optimal values of the residual angle of deviation of the lower leg during unicompartmental knee replacement allows improving the biomechanical conditions for the functioning of the endoprosthesis.

Key words: residual angle, unicompartmental knee replacement, knee joint.

Introduction

The achievement of excellent clinical results of unicompartmental knee replacement is possible only if the lower limb axis is optimally corrected. Many methods for planning and calculating the optimal correction angle have been developed, but there is no single standard algorithm [3, 7, 8].

According to a number of researchers, one of the prerequisites for successful knee replacement is to restore its neutral position with simultaneous external rotation of the femur as well as the symmetry of the flexor and extensor joint spaces [7]. To do this, the cut of the articular surface of the distal femur must be perpendicular to the mechanical axis of the femur, and the cut of the articular surface of the proximal tibial pigtail must be perpendicular to the mechanical axis of the lower leg.

Long-term dynamic monitoring of patients who were operated in our clinic using the unicompartmental endoprosthetics method for 15 years (224 patients, 237 joints) showed that the best survival of endoprosthetics elements took place among those patients who in the postoperative period had an angle of deviation of the lower leg towards the endoprosthetics within 1–5^{0}. In the same group of patients, there were the least signs of progression of degenerative changes in the contralateral part of the joint.

Traditionally, the mechanical axis of the lower limb runs from the center of the femoral head to the center of the knee joint and to the center of the articular line of the ankle joint. It is believed that the restoration of the mechanical axis of the lower limb contributes to an even distribution of the load between the condyles of the femur, which leads to a decrease in the «wear» of implants and the frequency of instability formation. According to the literature, the restoration of the mechanical axis is the main factor in achieving optimal functional ability of the knee joint [5, 6].

According to other researchers, the «gold standard» of knee replacement is the deviation of the position of the endoprosthesis components from the mechanical axis within ±3°. The issue of preserving the residual varus deviation remains debatable. It is known that the deviation of the axis of the lower limb relatively to the mechanical up to 3° varus is considered «natural». The so-called «natural» deviation is formed due to the varus deviation of the tibial axis by an average of 3° and the distal femur by an average of 3° valgus in relation to the mechanical axis of the lower limb. The «constitutional varus», characterized by a varus deviation of the lower limb axis by 3° or more relatively to the mechanical axis of the lower limb, is observed in 17–32% of healthy individuals who have reached skeletal maturity [7, 8].

Taken into consideration the wide individual variability of the «natural» axis of the lower limb, the «traditional» restoration of the mechanical axis during endoprosthetics may have unfavorable kinematic and clinical treatment results. The restoration of the mechanical axis in individuals with the «constitutional varus» of the knee is associated with the excessive excision of soft tissues and the significant resection of the tibia, which is associated with low functional capacity of patients in the long-term period.

Thus, the problem of choosing the optimal value of the residual angle of the deviation of the lower leg during unicompartmental knee replacement is an urgent task of the modern medicine and requires theoretical and practical justification.

Objective: to calculate the optimal values of the residual angle of deviation of the lower leg during unicompartmental knee replacement by mathematical modeling.

Object and methods of research

To conduct the mathematical modeling of the load conditions of the knee joint after the single-pole endoprosthetics, a calculated load scheme of the knee joint with a single-pole endoprosthesis was chosen, in which the femur is a structure with two points of support (fig. 1).

When performing calculations, the force value F was assumed to be equal to 700 N, which corresponds to the average human body weight. The angle of deviation g of the force F was changed from 0 to 20° in the direction of varus and valgus deformations with increments of 2°.

In addition, the fact of the presence of an endoprosthesis on one of the condyles was taken into account by giving to the support 2 the mechanical properties of titanium. According to the literature data [2], the ratio of the elastic modulus of the cortical bone and titanium is approximately 1:6, so we assumed: EK=1 ET=6, where EK

Figure 1 The calculation scheme of the load of the knee joint in the frontal plane, where *F* is the force of action of the human body weight; *Fx*, *Fy* are the components of the force *F* in the Cartesian coordinate system; *g* is the angle of action of the force *F*; N_1, N_2 are the reaction forces of supports acting on the femoral condyles

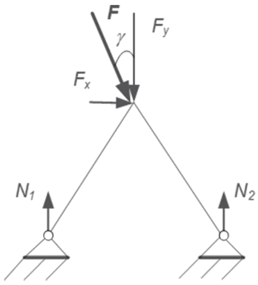
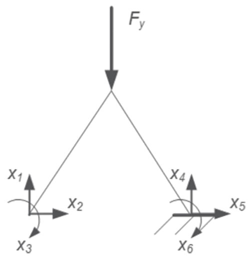


Figure 2 Scheme for calculating the reaction of the system to the influence of the force 1 H, where X_1 is the value of the vertical component of the force N_1, X_2 is the value of the horizontal component of the force N_1, X_3 is the value of the torque formed by the force N_1; X_4 is the value of the vertical component of the force N_2, X_4 is the value of the horizontal component of the force N_2, X_6 is the value of the torque formed by the force N_2



is the elastic modulus of the cortical bone; ET is the elastic modulus of titanium.

We investigated the support option, i.e. the hinge type of the fixing. The simulation used a method for calculating statically indeterminate systems. Calculations were performed in a package of application programs for solving technical computing problems MathLab R2012a (License № 536828 HostID: 001A92CC27A1) [1].

Results

Let's consider how a change in the angle of action g of the body weight of the force F affects the distribution of loads on the femoral condyles after the single-pole endoprosthetics. The first step in solving this problem, according to the method [2], is to release the support 1 and apply to it the force of 1 H, i.e. N_1=1 H.

Let's calculate the forces acting in the system under such a load (fig. 2).

Let's make a system of equations describing the equilibrium of the system under the action of the force N_1=1 H:

$$\begin{cases} \delta_{11}x_1 + \delta_{12}x_2 + \delta_{13}x_3 + \Delta_{1F} = 0 \\ \delta_{21}x_1 + \delta_{22}x_2 + \delta_{23}x_3 + \Delta_{2F} = 0 \\ \delta_{31}x_1 + \delta_{32}x_2 + \delta_{33}x_3 + \Delta_{1F} = 0 \end{cases} \quad (1),$$

where:

x_1 is the value of the vertical component of the force N_1;

x_2 is the value of the horizontal component of the force N_1;

x_3 is the value of the torque formed by the force N_1;

δ_{11}–δ_{33} are values of displacements caused by the corresponding forces;

Δ_{1F}–Δ_{3F} are values of displacements caused by the corresponding moments.

The values of displacements in the system elements are calculated according to the reference data given in the technical literature [2]:

$$\delta_{11} = \frac{1}{3EJ}\left(a^2(l_1 + 3l_2) + l_2 b(b + 3a)\right) \quad (2),$$

where *E* is the elastic modulus of the material; *J* is the moment of inertia.

Similarly:

$$\delta_{12} = \delta_{21} = -\frac{a}{EJ}\left(\frac{1}{3}cl_1 + \left(\frac{1}{2}a + \frac{1}{3}b\right)l_2\right) \quad (3)$$

$$\delta_{22} = \frac{1}{EJ}\left(\frac{1}{3}l_1c^2 + \frac{1}{3}l_2c^2\right) = \frac{c^2}{3EJ}(l_1 + l_2) \quad (4)$$

$$\delta_{33} = \frac{1}{EJ}(l_1 + l_2) \quad (5).$$

Let's define the movements which occur under the influence of the torque of the force F:

$$\Delta_{1F} = -\frac{F_Y}{EJ}\left(\frac{1}{6}bl_2\left(a + 2(a+b)\right)\right) = -\frac{F_Y bl_2(3a+2b)}{6EJ} \quad (6)$$

$$\Delta_{2F} = \frac{F_Y}{EJ}\left(\frac{1}{6}bcl_2\right) = \frac{F_Y bcl_2}{6EJ} \quad (7)$$

$$\Delta_{3F} = -\frac{F_Y}{EJ}\left(\frac{1}{2}bl_2\right) = -\frac{F_Y bl_2}{2EJ} \quad (8).$$

Let's transfer the free terms in the system of equations (1) to their right-hand side:

$$\begin{cases} \delta_{11}x_1 + \delta_{12}x_2 + \delta_{13}x_3 = -\Delta_{1P} \\ \delta_{21}x_1 + \delta_{22}x_2 + \delta_{23}x_3 = -\Delta_{2P} \\ \delta_{31}x_1 + \delta_{32}x_2 + \delta_{33}x_3 = -\Delta_{3P} \end{cases} \quad (9).$$

Let's represent the coefficients of the equations of the system in the form of a matrix:

$$A = \begin{vmatrix} \delta_{11} & \delta_{12} & \delta_{13} \\ \delta_{21} & \delta_{22} & \delta_{23} \\ \delta_{31} & \delta_{32} & \delta_{33} \end{vmatrix} \quad (10).$$

Let's find the roots of equations using Kramer's formulas:

$$x_1 = \frac{\Delta_1}{\det A};$$

$$x_2 = \frac{\Delta_2}{\det A};$$

$$x_3 = \frac{\Delta_3}{\det A}. \quad (11),$$

where det *A* is the determinant of the Matrix |A| (10) which is determined by the equation:

$$\det A = \delta_{11}\delta_{22}\delta_{33} + \delta_{12}\delta_{23}\delta_{31} + \delta_{21}\delta_{13}\delta_{32} - \delta_{13}\delta_{22}\delta_{31} - \delta_{11}\delta_{23}\delta_{32} - \delta_{12}\delta_{21}\delta_{33} \quad (12).$$

The values Δ_1, Δ_2, and Δ_3 are determined by replacing the corresponding column of the Matrix |A| (10) with a column of free terms from the system of equations (9):

$$\Delta_1 = \begin{vmatrix} -\Delta_{1F} & \delta_{12} & \delta_{13} \\ -\Delta_{2F} & \delta_{22} & \delta_{23} \\ -\Delta_{3F} & \delta_{32} & \delta_{33} \end{vmatrix}$$

$$\Delta_1 = -\Delta_{1F}\delta_{22}\delta_{33} - \Delta_{3F}\delta_{12}\delta_{23} - \Delta_{2F}\delta_{13}\delta_{32} + \Delta_{3F}\delta_{13}\delta_{22} + \Delta_{1F}\delta_{23}\delta_{32} + \Delta_{2F}\delta_{12}\delta_{33} \quad (13)$$

$$\Delta_2 = \begin{vmatrix} \delta_{11} & -\Delta_{1F} & \delta_{13} \\ \delta_{21} & -\Delta_{2F} & \delta_{23} \\ \delta_{31} & -\Delta_{3F} & \delta_{33} \end{vmatrix}$$

$$\Delta_2 = -\Delta_{2F}\delta_{12}\delta_{33} - \Delta_{1F}\delta_{23}\delta_{31} - \Delta_{3F}\delta_{13}\delta_{21} + \Delta_{2F}\delta_{13}\delta_{31} + \Delta_{3F}\delta_{12}\delta_{23} + \Delta_{1F}\delta_{21}\delta_{33} \quad (14)$$

$$\Delta_3 = \begin{vmatrix} \delta_{11} & \delta_{12} & -\Delta_{1F} \\ \delta_{21} & \delta_{22} & -\Delta_{2F} \\ \delta_{31} & \delta_{32} & -\Delta_{3F} \end{vmatrix}$$

$$\Delta_3 = -\Delta_{3F}\delta_{11}\delta_{22} - \Delta_{2F}\delta_{12}\delta_{31} - \Delta_{1F}\delta_{21}\delta_{32} + \Delta_{1F}\delta_{22}\delta_{31} + \Delta_{2F}\delta_{11}\delta_{32} + \Delta_{3F}\delta_{12}\delta_{21} \quad (15).$$

Substitute the values of the coefficients δ from expressions (3–5) and Δ from expressions (6–8) and determine the roots of the equations.

The values of forces and moments arising from the action of the horizontal component of the force of the human body weight are determined in the same way.

The values of the reaction of the support force N2 and the torques acting on this support are determined using equilibrium equations:

$$\begin{cases} x_1 + x_4 - F_Y = 0 \\ -F_X + x_2 + x_5 = 0 \\ x_4(a+b) - F_Y a + F_X c + x_6 = 0 \end{cases} \quad (16).$$

From this system of equations, we determine the values of the forces and moment acting on the support 2:

$$\begin{cases} x_4 = F_Y - x_1 \\ x_5 = F_X - x_2 \\ x_6 = F_Y a - F_X c - x_4 \end{cases} \quad (17).$$

Substitute in the third line the value of the force x_4 from the equation in the first line of the system (17), we have:

$$\begin{cases} x_4 = F_1 - x_1 \\ x_5 = F_2 - x_2 \\ x_6 = F_1 a - F_2 c - F_1 + x_1 \end{cases} \quad (18).$$

The model of loading the femoral condyles with articulated fixation of the support mainly corresponds to the situation after endoprosthetics of the knee joint. Therefore, the next step was to study the components of the reaction forces of the support and their torques depending on the angle of inclination of the femoral axis in the presence of support for the femoral condyles of the «hinge» type. The results of calculations for the model with the slope of the femoral axis towards a healthy condyle are shown in Table 1.

In the absence of deviation of the femoral axis, the vertical load is distributed evenly between its condyles, and it is 350 N for each of them. Later, when the femoral axis deviates towards a healthy condyle, with an increase in the angle of deviation, an increase in the load value to 401 N is observed, particularly on a healthy condyle. Herewith, the vertical load on the endoprosthesis is proportionally reduced to 162 H.

It should be noted that there are no horizontal components of the reaction forces of the support under their uniform load. But already with 2° deviations of the femoral axis towards a healthy condyle, quite significant horizontal loads (about 300 H) occur on both condyles. An increase in the angle of inclination of the femoral axis leads to an increase in the horizontal component of the support reaction force on a healthy condyle to 467 H, and a proportional decrease on the endoprosthesis — to 189 H.

Data on the values of the components of the support reaction forces for a scheme with articulated fixation of supports, when the femoral axis is tilted towards the endoprosthesis, are shown in Table 2.

According to the articulated fixation of the support, when the femoral axis deviates towards the endoprosthesis, the picture of changes in the values of the components of the reaction forces of the support highlights that one in case of the deviation towards a healthy condyle.

Table 1 Values of the components of the reaction forces of the support for a scheme with articulated fixation of the supports, when the femoral axis is tilted towards a healthy condyle

Value of the angle g, °	Components of the reaction forces of the support, H			
	Nx_2	Nx_1	Ny_2	Ny_1
0	0	0	350	350
2	287	311	335	363
4	275	323	321	377
6	261	335	305	390
8	248	345	290	403
10	234	355	273	415
12	220	365	257	427
14	206	375	240	438
16	192	384	224	448
18	177	393	206	458
20	162	401	189	467

Table 2 Values of the components of the reaction forces of the support for the scheme with articulated fixation of the supports, when the femoral axis is tilted towards the endoprosthesis

Value of the angle g, °	Components of the reaction forces of the support, H			
	Nx_2	Nx_1	Ny_2	Ny_1
0	0	0	350	350
2	311	287	363	335
4	323	275	377	321
6	334	261	390	305
8	345	248	403	290
10	355	234	415	273
12	365	220	426	257
14	375	206	438	240
16	384	192	448	224
18	393	177	458	206
20	401	162	467	189

Discussion

Thus, based on the data of the above tables, it can be generalized that the deviation of the femoral axis in the direction of varus or valgus leads to proportional changes in the load of the condyles, namely, an increase in the load on the condyle in the direction of which the slope is made, and unloading the opposite. Especially we should note the occurrence of significant horizontal components of the support reaction forces on both condyles already when the femoral axis deviates by only 2° in different directions.

In 2006, in order to achieve optimal anatomical and physiological kinematics of the knee joint, a method of kinematic restoration of the axis during total knee replacement was developed. Sections of bone surfaces according to the technique allow restoring the axial ratios and natural shape of the joint, which leads to less trauma oa the capsule-ligamentous apparatus. The results of using mechanical and kinematic axis restoration in total knee replacement have been analyzed in a number of studies. Thus, most researchers are inclined to believe that the method of kinematic restoration of the axis during total knee replacement allows to achieve better functional and clinical results, to reduce pain and the frequency of instability [4, 5].

Regarding the technique of unicompartmental endoprosthetics of the knee joint, such studies are single.

So, Ch. Rivière et al., 2019 proved that restoration of the kinematic axis of the lower limb during the medial unicompartmental endoprosthetics (Oxford™) contributes to better clinical results due to the formation of optimal biomechanical conditions for the functioning of endoprosthetics. It is proved that the positioning of endoprosthesis components (Oxford™) during kinematic axis restoration differs significantly from its position during mechanical axis restoration, but remains within the recommended positioning limits. By modeling the conditions for restoring the kinematic axis of the lower limb, the researchers confirmed that the femoral components are more oriented towards valgus, and the lower legs have more varus orientation, which contributes to the formation of better biomechanical conditions in the «endoprosthesis-bone» system, compared to the model of mechanical axis restoration. The question about the accuracy of implant positioning came up before researchers from Ch. Rivière et al., 2019 [8]. This problem was partially solved in our study. By mathematical modeling, we have proved that the deviation of the angle of inclination in the range of ±2–3° is relatively safe, and exceeding this indicator leads to the appearance of horizontal components of the reaction forces of the support of a significant value. It is known that during mechanical restoration of the axis, the design of unicompartmental endoprostheses, in most cases, allows achieving a high degree of congruence of components, even under conditions of inaccurate implantation. In kinematic restoration of the lower limb axis, the question of load distribution on the bones, especially their edges, is open and requires further studies.

Conclusions

The deviation of the lower leg angle as a result of unicompartmental knee replacement can lead to a significant redistribution of loads between the femoral condyles in the direct proportion to the value of the angle of inclination for a stable joint or non-linearly in case of lack of stability.

We consider it relatively safe to deviate the angle in the range of ±2–3°, exceeding of this indicator leads to the appearance of horizontal components of the support reaction forces of a significant value.

During unicompartmental endoprosthetics, it is necessary to preserve all the stabilizing elements of the knee joint, since its instability leads to significant nonlinear fluctuations in the values of the components of the support reaction forces, both in the vertical and horizontal planes, in particular from the side of the endoprosthesis.

Conflict of interests

The authors declare no conflict of interest.

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Анатомо-біомеханічне обґрунтування та роль залишкового кута відхилення гомілки при монокондилярному ендопротезуванні колінного суглоба

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Анотація. ***Мета:*** розрахунок оптимальних значень залишкового кута відхилення гомілки при монокондилярному ендопротезуванні колінного суглоба шляхом математичного моделювання. ***Об'єкт і методи дослідження.*** Для математичного моделювання обрано розрахункову схему навантаження колінного суглоба з однополюсним ендопротезом, в якій стегнова кістка представлена конструкцією з двома точками шарнірної фіксації опори. Величину сили F приймали рівною 700 Н. Кут нахилу g сили F змінювали від 0 до 20° в бік варусної і вальгусної деформації з кроком 2°. При моделюванні використовували метод розрахунку статично не визначених систем. Розрахунки виконували в пакеті прикладних програм для вирішення задач технічних обчислень MathLab R2012a. ***Результати.*** Залишковий кут відхилення гомілки при монокондилярному ендопротезуванні призводить до значного перерозподілу навантажень між виростками стегнової кістки прямо пропорційно величині кута нахилу для стабільного суглоба та нелінійно у випадку відсутності стабільності. Доведено, що відносно безпечним є відхилення кута в межах ±2–3°, перевищення даного показника призводить до виникнення горизонтальних складових сил реакції опори значної величини. ***Висновок.*** Розрахунок оптимальних значень залишкового кута відхилення гомілки при монокондилярному ендопротезуванні колінного суглоба дозволяє покращити біомеханічні умови функціонування ендопротезу.

Ключові слова: залишковий кут, монокондилярне ендопротезування, колінний суглоб.

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