This paper investigated the dependence of the stability of the knee joint, when exposed to external rotational load on the lower leg, on the position of the graft of the tendon of the popliteal muscle. Estimation finite-element models of the right knee joint of an adult were constructed, which included the articular ends of the bones that form this joint, as well as its main ligaments. The models reflected a surgery to restore posterolateral angle structures and differed only in the position of the popliteal tendon graft. That position was set by the point of attachment of the graft to the posterior surface of the tibia. At the same time, the fixation point changed both vertically and horizontally, in the frontal plane. In addition, a control model was built in which the hamstring tendon was completely absent. As a result of the calculations, patterns of the distribution of the fields of movement of the points of the finite-element model were obtained. As criteria for assessing the effectiveness of the selected position of the graft, movements of the lower leg model in the horizontal plane were proposed. Analysis of the results of the calculations showed that the greatest movements in all directions were obtained in the control model, in which the hamstring of the popliteal muscle was absent. The magnitudes of the considered movements derived from the control model exceeded the same values in the model with minimal movements by 17, 37, 17, 32, and 16 %. From the point of view of the stability of the tibia under rotational load, the most effective was the fixation of the graft on the posterior surface of the tibia as laterally as possible and closer to its articular surface. This is indicated by the magnitude of the movements, which, in this case, turned out to be the smallest in all directions

Keywords: knee joint, hamstring tendon, external rotational load, graft, finite-element model, stiffness characteristics

INFLUENCE OF THE FIXATION POINT OF THE ARTIFICIAL POPLITEAL MUSCLE GRAFT ON THE STABILITY OF THE KNEE JOINT UNDER EXTERNAL ROTATIONAL LOAD

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1. Introduction

The knee joint is one of the most important and largest joints in the human body. Owing to it, people can perform a variety of movements of the lower limb. Such movements primarily include bringing the lower leg to the thigh and its removal – flexion and extension of the knee, which occur in the sagittal plane, as well as internal and external rotation – in the horizontal.

The variety of movements in the knee joint is due, first of all, to its structure, which is very sophisticated as it is a whole complex of a large number of ligaments, muscles, nerves, blood vessels, cartilage, and bones. Of course, each of these elements is important for the normal functioning of the knee joint. However, special mention should be made

of the ligamentous apparatus of the joint in question, which enables the stability of the position of the bones that form the knee, under the loads acting on it. For example, the most important ligaments of the knee joint – these are the anterior and posterior cruciate ligaments, are necessary to enable the anterior and posterior stability of the specified joint, as well as for its stability under rotational load.

It should be noted here that the combined effect of body weight and the movement carried out expose the knee joint to a significant load in the form of stretching, compression, and twisting. At the same time, in young people leading an active lifestyle and athletes, the efforts arising in the ligaments increase significantly, which leads to various damages to the ligamentous apparatus from overstretching to a complete rupture. The causes of ligament damage are high-ener-

gy mechanisms of injury, such as overextension of the lower leg, twisting along the axis, excessive withdrawal, bringing the lower leg, or direct exposure to a traumatic force on the lower leg. Therefore, traumatic damage to the ligamentous apparatus of the knee joint is one of the most common, after the ankle joint and foot, primarily among the young population and athletes.

Given the above, one can argue that the study of methods for restoring the ligamentous apparatus of the knee joint is a relevant task.

2. Literature review and problem statement

Damage to the posterolateral angle of the knee joint is a significant part of the injuries to the ligamentous apparatus of this joint. Work [1] indicates that such an injury is both isolated and can be combined with ruptures of the posterior or anterior cruciate ligaments. Actually, the work considers the description of the anatomy of the ligaments of the posterolateral angle and its biomechanics. Paper [2] states that damage to the posterolateral angle of the knee joint leads to chronic lateral and external rotational instability. In addition, it is emphasized that such injuries are often combined with injuries to the posterior cruciate ligament.

A large number of scientific studies tackle the development and research of various methods for restoring the stability of the knee joint [3–5]. For example, [4] considers a technique of restoring the ligamentous apparatus, with combined injuries to the knee joint but only the posterior cruciate ligament is reconstructed. At the same time, work [3] indicates that when restoring the stability of the knee joint, insufficient attention is paid to the structures of the posterolateral angle. This, in turn, leads to unsatisfactory treatment results. This fact is also pointed out in study [6], which states that further expansion of knowledge about the anatomy and biomechanics of the posterolateral angle leads to improved diagnostic capabilities and successful treatment of knee instability.

To date, there are various techniques for restoring the posterolateral angle. However, not many studies have been reported in the literature related to the quantitative or qualitative assessment of factors affecting the results of treatment of knee instability in case of hamstring damage. Such work is necessary to determine the most effective methods for restoring the ligamentous apparatus of the knee joint. One of the few studies in which the results of treatment of instability of the knee joint are compared are [5, 7]. Thus, paper [7] compares the results obtained by only two methods: Arciero and LaPrade. And in [5], a study is performed to determine the optimal position of the damaged ligamentous structures of the knee joint during their recovery. However, the cited study is performed on cadaveric samples, which does not give a full opportunity to take into consideration the actual conditions of loading the joint.

It should be noted here that there are practically no papers in which the analysis of factors affecting the results of treatment of instability of the knee joint during rotational load, as well as the degree of influence of these factors, is reported. Thus, it can be concluded that in most of the works related to the considered area of research, data obtained on the basis of the results of surgical treatment of instability of the knee joint, or the results of experiments on cadaveric samples, are given. However, it is obvious that the use of such approaches to assess the effectiveness of a particular method of treatment is time-consuming. In addition, they do not make it possible to fully identify the patterns of influence of certain factors on the stability of the joint in question.

The use of modern computer technology will make it possible to perform a detailed analysis of the parameters affecting the results of the treatment of damage to the ligamentous apparatus of the knee joint. In addition, based on studies of numerical models of the knee joint, it is possible to develop recommendations for the introduction of new methods for restoring the stability of this joint. An example of successful use of this approach is work [8], in which the assessment of the forces arising in the anterior cruciate ligament when the angle of inclination of the tibial plateau changes is given.

From [9], it follows that one of the main structures of the posterolateral angle of the knee joint, which enables its stability, is the tendon of the popliteal muscle. The course of surgical intervention on the plastic of this tendon involves drilling a channel in the outer condyle of the tibia from front to back. The channel exit point on the posterior surface of the tibia is the geometric location of the graft attachment. Obviously, the position of fixation of such a transplant depends on the result of treatment to restore the stability of the knee joint. However, exact recommendations for its choice in the literature have not been found. That is, the unsolved issue is to determine the optimal position of the attachment point of the graft of the hamstring tendon muscle while restoring the stability of the knee joint, which is proposed to be solved in the current study.

3. The aim and objectives of the study

The purpose of this study is to determine the optimal position of fixation of the tendon graft on the posterior surface of the tibia during external rotation of the tibia. This will make it possible to enable the necessary stability of the knee joint when performing an operation to restore the functions of the popliteal muscle.

To accomplish the aim, the following tasks have been set:

– to determine the magnitude of movements that occur in the knee joint during rotational load;

– to investigate the nature of the distribution of displacement fields arising in the elements of the knee joint model.

4. The study materials and methods

The object of this study was the knee joint and its ligamentous apparatus.

The main hypothesis of the study assumes that the position of the graft of the tendon of the popliteal muscle affects the stability of the knee joint during rotational load.

The study was carried out using a software package based on the finite-element method in the ANSYS software environment (USA). In order to simplify the calculations, the geometric model consisted only of the articular ends of the tibia, fibula, and femur bones that form the knee joint (Fig. 1).

Given the complex geometry of the bone surface, the articular ends were built from computed tomography (CT) data of the knee joint of an adult, using specialized software for processing CT images (3D Slicer). The resulting model

in the format "stl" was used for further calculations. Consequently, the shape and size of the model corresponded to the actual knee joint of a person. At the same time, the overall (maximum) dimensions of the model in the horizontal plane were 100*70 mm. In order to reduce the total number of finite elements, the model was limited in height to 15 cm.
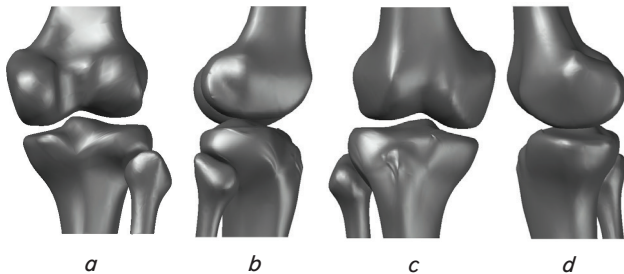

Fig. 1. Three-dimensional model of the knee joint:
a — rear view; *b* — view from the outside; *c* — front view;
d — view from the inside

It should be noted that in this study, the estimation model of the knee joint corresponded to the position of extension in its vertical orientation.

When modeling the stability of the knee joint, it is quite difficult to fully reflect its structure and the behavior of individual ligaments, both from a geometric and physical point of view. Each ligament consists of fibers that form individual branches, which, in turn, are attached to the articular ends of the bones at several points, occupying some area on the surface of the bone. In this case, the ligament along its length can vary in thickness, expanding to the places of attachment, and due to the different directions of the fibers to have a rotation of the cross-section around the longitudinal axis of the ligament, that is, "twisting". In addition, the ligaments have not only elastic physical and mechanical properties but also such as creep and relaxation, determining of and accounting for which is a laborious task.

Obviously, taking into consideration the structural features and properties of the ligaments would lead to a significant increase in the time for building a model (the geometric aspect of the problem) and for performing the calculation (the physical aspect of the problem), and often it is impossible at all.

Taking into consideration the above, the ligament was considered as an element, in terms of the resistance of materials, which works only on stretching, that is, only longitudinal tensile forces can occur in it. Therefore, it was proposed to replace the actual shapes of the ligaments with cylindrical elements with low bending stiffness (by reducing the size of the cylinder diameter). The bases of the cylinders were fixed in the places of the beginning of the ligaments, which were defined as the center of the contact area of the ligament with the surface of the bone. The diameters of all cylinders were equal to 2 mm, and the lengths were due to the places of attachment of the ligaments (Fig. 2).

Table 1 gives the actual dimensions of the cross-sections and the length of the ligaments of the knee joint.

Regarding the physical and mechanical properties of the elements of the model (bones and ligaments of the knee joint), we note the following. The material of the elements was supposed to be elastic, homogeneous, and isotropic. Note that the subject of this study is primarily the behavior of the ligamentous apparatus of the knee joint when it is loaded. Therefore, given the significant difference in the elastic properties of bone tissue and ligaments, the articular ends of the bones, in the entire volume, were given the properties of cortical bone tissue (spongy tissue was not separated). In the calculations, the following values were accepted: the Young modulus of bone tissue – $2*10^4$ MPa, Poisson coefficient – 0.25 [15]. The mechanical values of the ligamentous apparatus used in the calculations, which correspond to the average values of the properties of the ligaments of the knee joint of an adult, are given in Table 2.
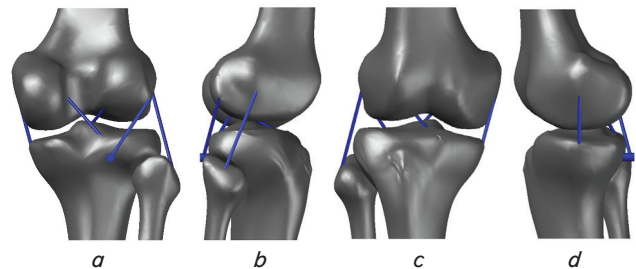

Fig. 2. Scheme of the location of the ligaments of the knee joint: *a* — rear view; *b* — view from the outside; *c* — front view; *d* — view from the inside

Note that the replacement of the anatomical shape of the ligaments with their simplified models in the form of cylinders leads to a change in the stiffness characteristics of the models:

– *EA* is the cross-sectional stiffness of the rod (ligament) during stretching;

– *E* is the Young modulus;

– *A* is the cross-sectional area.

Table 1

Size of the ligaments of the knee joint [10–14]

Ligament	Length, mm	Cross-sectional area, mm^2
Anterior cruciate ligament	32	37.4
Posterior cruciate ligament	35	64.05
Lateral collateral ligament	48.15	8.76
medial collateral ligament	68.99	24.54
Hamstring tendon	34.3	21.9

Table 2

Elastic properties of the ligaments of the knee joint [11, 14, 16]

Ligament	Young modulus, MPa	Poisson coefficient	Reduced Young modulus, MPa
Anterior cruciate ligament	123	0.4	1464
Posterior cruciate ligament	168	0.4	3425
Lateral collateral ligament	280	0.4	781
Medial collateral ligament	224	0.4	1750
Hamstring tendon	130.9	0.4	913

In this case, it becomes necessary to determine such moduli of elasticity, the value of which, taking into consideration the size of the cross-sections of the models of ligaments, would give the models of ligaments stiffness values similar to the actual ones. For this purpose, the calculation of the reduced modulus of elasticity of the ligaments was performed, based on the equality of the stiffness values for the sprain of the ligament and the cylinder modeling it, that is

$$E_l A_l = E_c A_c,$$

$E_l A_l$ is the stiffness of the ligament for stretching, $E_c A_c$ is the stiffness of the cylinder for stretching.

Hence, the reduced modulus of elasticity of the ligament:

$$E_c = E_l A_l / A_c.$$

The results of these calculations are given above in Table 2.

The technique of setting the load was determined as follows. As is known [17], on the side of the lower leg, the knee joint is formed by the articular ends of two bones: the tibia and the fibula, which are present in the estimation model. These bones can be mobile relative to each other, and in the case of applying a load to the lower leg, the effect on both bones is transmitted simultaneously. In order to exclude their mutual displacement, the articular ends at the bottom of the model were rigidly connected to each other by a cylindrical element (platform), 10 mm high, and 120 mm in diameter (Fig. 3, *a*, *b*, *d*). The mechanical properties of the platform corresponded to those of the cortical bone. In addition, at the top of the tibia model, the tibia and fibula were connected by a cylindrical element with a diameter of 10 mm and the mechanical properties of the hamstring of the popliteal muscle (Fig. 3, *b*, *c*).

The present study is based on the evaluation of one of the functions of the popliteal muscle – ensuring the stability of the lower leg during external rotation of the lower leg. Therefore, the loading of the model was carried out by torque, which was applied to the lower base of the cylindrical platform in the direction of the outside (Fig. 3, *d*). In other words, we performed an external rotation of the tibia. The magnitude of the torque was chosen arbitrarily and, after preliminary calculations, it was determined at the level of 15 N*m.

Regarding boundary conditions, note the following. In the actual knee joint, contact interaction takes place between the femur and tibia. In this case, contact is realized through such joint structures as cartilage and menisci. The presence of these structures in the joint gap indicates that there is no direct contact between the bones, and there is a certain distance directly between the articular surface of the femur and tibia. Obviously, the presence and consideration of these structures of the joint can lead to an increase in its stiffness during rotational load, primarily due to the occurrence of frictional forces. However, it should be noted here that the contacting surfaces of the knee joint are congruent and smooth, which leads to a decrease in the friction force in it, and in addition, synovial fluid is present in the joint bag itself, which also reduces the level of friction in the joint. Due to these features of the joint under study, we can say that friction on the contacting articular surfaces is practically absent and can be ignored when performing calculations.

Thus, for the study, boundary conditions of the calculation model were proposed, which reflected the presence of a distance between the articular surfaces of the femur and tibia and simulated the absence of friction in the joint. It was forbidden to move the fragment of the femur in all directions. For the model of the lower leg, vertical movement was prohibited, in other directions, free displacements were allowed.

For ease of orientation, a rectangular coordinate system was introduced (Fig. 3, *a*). With respect to the model, the *x*-axis was directed from the inside out; *y*-axis – perpendicular to the *x*-axis, rear forward; the *z*-axis is perpendicular to the *x*0*y* plane, from bottom to top. That is, the *z*-axis coincided with the vertical axis of the lower limb model, and the *x*0*y* plane was perpendicular to this axis.

To perform a study to determine the most optimal position of fixation of the popliteal muscle graft on the posterior surface of the tibia, 9 estimation schemes were constructed, which differed in the place of its attachment. Control model 10 was also built, in which the hamstring tendon was missing. The graft was modeled with a circular cylinder with a diameter of 2 mm. The upper edge of the graft was attached to the anatomical point of the beginning of the hamstring on the outer condyle of the femur, and the point of its fixation on the tibia changed vertically and horizontally. For the convenience of orientation of the lower part of the graft on the tibia, its connection was carried out through cylindrical elements created on the posterior surface of the outer condyle, which set the position of attachment of the graft to the tibia (Fig. 4, *a*, *b*).
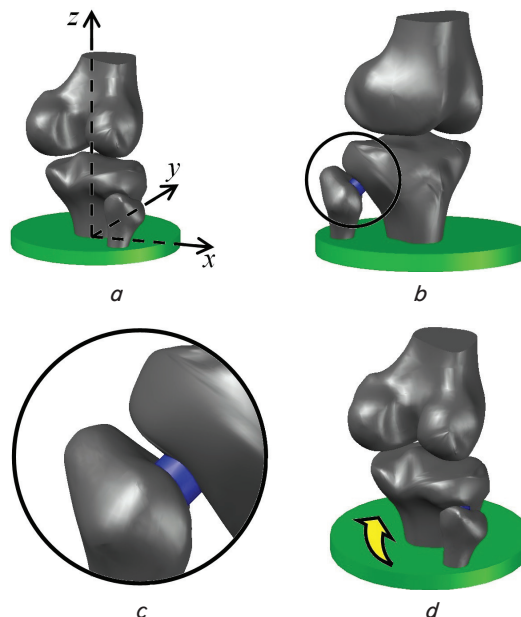

Fig. 3. Model of the knee joint: *a* — coordinate system used; *b*, *c* — the connection of the tibia and fibula in the proximal epiphysis; *d* — direction of application of rotational load to the platform
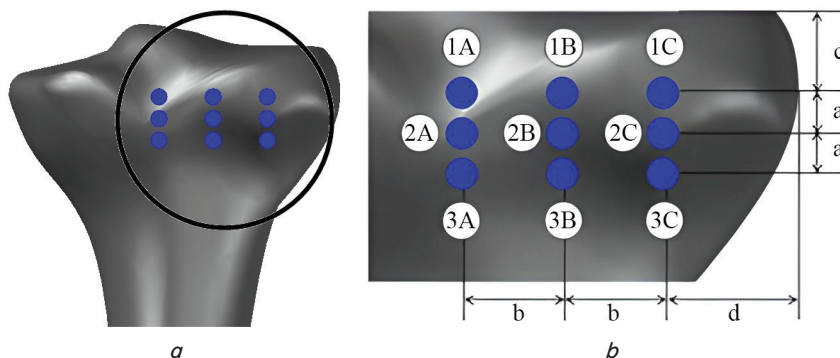

Fig. 4. Scheme of the location of the graft fixation points for the reconstruction of the hamstring of the popliteal muscle (rear view): *a* — general view of the tibia from behind; *b* — designation of graft fixation points

Distances between the centers of the bases of cylindrical elements: a=5 mm – in the vertical direction, b=12.5 mm – in the horizontal direction. As a reference point, to determine the fixation positions of the hamstring tendon graft in the computational models, point 1C was chosen, which was located laterally with respect to the tibia at the upper level of fixation. The position of this point in the frontal plane was given by distances: c=12 mm – from the articular plateau of the tibia, d=17 mm – from the lateral edge of the tibia (Fig. 4, b).

The elastic properties of the graft for the restoration of the popliteal muscle such as the Young modulus and its reduced value, as well as the Poisson coefficient, which were accepted in the calculations, are given in Table 2.

Regarding the calculations, it should be noted that in this study it was proposed to consider the task from the point of view of physical and geometric linearity.

5. Results of studies of the finite-element models of the knee joint in plastic surgery of the hamstring of the popliteal muscle

5. 1. Determining the magnitude of movements arising at the points of the finite-element model of the knee joint

Based on the results of our calculations, patterns of the distribution of stresses, deformations, and movements in the elements of the model of the knee joint (articular ends of bones and ligaments) were established. Recall that this work reports a study to determine the optimal position of the graft to restore the popliteal muscle on the back surface of the outer condyle of the tibia. Since the main goal of the operation is to enable the stability of the lower leg under rotational load, the movements of parts of the finite-element model were identified as criteria for assessing the effectiveness of choosing the graft fixation point. The main values were the maximum movement of the points of the cylindrical platform in the directions of the coordinate axes x, y, as well as the magnitude of the total movement of the platform. Our results are given in Table 3.

Table 3

Moving platform points

Point location	Displacement, mm				Maximum total displacement (d)
	Along the x-axis		Along the y-axis		
	outwards x_{max}	inward x_{min}	anteriorly y_{max}	posteriorly y_{min}	
1A	9.9995	−4.3242	8.3687	−5.955	10.245
1B	9.7025	−4.1251	8.1362	−5.6914	9.9586
1C	9.1472	−4.0146	7.7101	−5.4518	9.3846
2A	9.9544	−4.4223	8.3349	−6.0418	10.183
2B	9.487	−4.1489	7.9625	−5.6734	9.722
2C	9.4306	−4.2919	7.9493	−5.7733	9.6514
3A	9.9472	−4.4712	8.3361	−6.0823	10.17
3B	9.672	−4.365	8.1271	−5.9099	9.8943
3C	9.6444	−4.5298	8.1272	−6.0471	9.8478
Without tendon	10.696	−5.4907	8.9943	−7.1925	10.848

Table 3 gives the largest and smallest, taking into consideration the sign, displacement along the coordinate axes, as well as the magnitude of the total displacement. The follow-

ing designations are introduced: x_{max}, x_{min} – movement along the x-axis in its positive and negative directions (x_{max} – outwards, x_{min} – inward), y_{max}, y_{min} – movement along the y-axis (y_{max} – anteriorly, y_{min} – posteriorly), d is the maximum full displacement.

Table 3 demonstrates that the outward displacements of the tibia in the frontal plane were more pronounced compared to its positive movements in the sagittal plane with all variants of fixation of the popliteal muscle graft. At the same time, the values of outward movements were greater than the displacements inwards, and the movements to the front, there were more movements to the rear. This fact indicates that with a rotational load on the knee joint, the predominant displacement of the lower leg model occurs outwards and anteriorly.

Regarding movements along the x-axis (in the frontal plane), the following is worth noting. Regardless of the height of the graft fixation (levels 1–3), when this point is shifted from the outside to the inside (from points C to points A, Fig. 4, b), the x_{max} values change upwards. At the same time, the movements of x_{min} during graft fixation farther from the articular surface (points 2ABC and 3ABC) are reduced during the transition from the lateral fixation point (point C) to the middle (point B). Then these displacements increase again with the transition from the midpoint B to the medial point A. However, at the upper level (1ABC), when the fixation point is shifted from the outside to the inside (from C to A), the movement of the points of the x_{min} platform (in the direction of the x-axis inwards), like x_{max}, gradually increases.

Table 3 shows that the nature of the distribution of the magnitudes of the movements y_{max} and y_{min} in the direction of the coordinate axis y was similar to the displacements along the x-axis.

With respect to the magnitude of the total movements (d), one can see that their changes depending on the position of the graft fixation point were fully correlated with changes in the movements in the frontal plane x_{max}, x_{min} (along the x-axis).

5. 2. Studying the nature of the distribution of displacement fields arising in the model of the knee joint

There were the greatest movements at the points of the platform farthest from its center (Fig. 5). Fig. 5 shows the distribution pattern of movements along the x-axis (Fig. 5, a), along the y-axis (Fig. 5, b), and the total displacement d (Fig. 5, c). Arrows show the points of occurrence and direction of the specified maximum movements. In addition, the dotted line marks the initial position of the cylindrical platform of the model. Note that the nature of the distribution of displacement fields in all models of fixation and control model was identical.
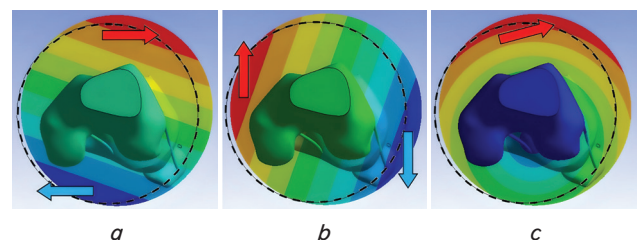

Fig. 5. Distribution pattern of displacement fields in the model (top view): a — in the direction of the x-axis; b — in the direction of the y-axis; c — full displacements d

Fig. 5, *a* demonstrates that the greatest positive movements in the direction of the *x*-axis (outward, red arrow) occurred on the front border of the platform, and the greatest negative (inward, blue arrow) – on the rear. At the same time, the greatest positive movements in the direction of the *y*-axis (anterior, red arrow) occurred on the leftmost border, and the greatest negative (rear, blue arrow) – on the right (Fig. 5, *b*). The greatest total displacements (*d*) are obtained at platform points between the points at which the greatest x_{max} and y_{max} displacements occurred (Fig. 5, *c*). The direction of the total displacement *d* can be determined by the parallelogram rule, which is used to add vectors (the direction of the red arrow, Fig. 5, *c*).

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6. Discussion of results of studying the effect of graft position on the stability of the knee joint

General analysis of the data given in Table 3 showed that when the graft fixation is shifted from points C to points A (from the periphery of the tibia plateau to the center), there is an increase in the magnitude of movements in all directions. That is, the stability of the knee joint of external rotation deteriorates. At the same time, there was either a gradual increase or a decrease in values during the transition from point C to point B and then again an increase with the displacement of fixation from point B to point A. However, in all calculated cases, the magnitudes of the movements when fixing the graft at point A were higher than the same values at point C. An exception was obtained only for movements x_{min} (inwards, in the frontal plane) for models of graft fixation at the level of 3ABC. In this case, the movements of x_{min} at point A were slightly less than at point C.

This increase in the magnitude of movements when the graft fixation point is shifted from the outside to the inside is explained by the fact that in this case, the length of the graft increases. In this case, the graft is under conditions of axial stretching. And according to the formula for determining absolute elongation derived from Hooke's law [18]:

$$\Delta l = \frac{N \cdot l}{E \cdot A},$$

it follows that under all equal other conditions, the elongation will be greater for such an element whose length is longer.

The nature of the changes in the magnitudes of the movements of the points of the model indicates that the lower and more centrally the graft is fixed on the tibia during popliteal muscle plastic surgery, the less rotationally stable the knee joint will be.

It should be noted that among all the estimation schemes in which the presence of a restored popliteal muscle was simulated, the smallest movements in all directions were obtained in one model, in which the graft was fixed at point 1C – as much as possible outward and upwards near the articular surface. And the maximum x_{max} and y_{max} were obtained in a model with graft fixation at the upper level in the center of the tibia (point 1A). The greatest movement of x_{min} is when fixing on the lower level from the outside (point 3C), and the maximum y_{min} is also when fixing at the lower level but in the center of the tibia (point 3A).

Among all the calculated values, as expected, the maximum values of movements were obtained from the control model, which did not take into consideration the influence of the hamstring of the popliteal muscle, that is, it was absent.

It should be noted that the magnitudes of the considered movements (x_{max}, x_{min}, y_{max}, y_{min}, *d*) obtained from the control model exceeded similar values in the model with minimal movements by 17, 37, 17, 32, and 16 %. At the same time, the largest movements among models with the installation of a popliteal muscle graft exceeded similar minimum values by 9, 13, 8, 11, and 9 %. In addition, the values of movements of the points of the platform of the control model exceeded the largest similar values obtained during calculations with the installation of a popliteal muscle graft by 7, 21, 7, 18, and 6 %.

It should be noted that the x_{min} values were less than x_{max} depending on the graft fixation point, by 53–57 %. At the same time, y_{min} compared to y_{max} is less by 26–30 %. On the other hand, y_{max} movements were less than x_{max} by 15–16 %, but y_{min} was higher than x_{min} by 33–38 %. In a model that did not take into consideration the presence of a popliteal muscle, the difference between the values of these movements was lower and these values were: 48 % ($x_{min}–x_{max}$), 20 % ($y_{min}–y_{max}$), 31 % ($y_{min}–x_{min}$), however, for $y_{max}–x_{max}$ remained at 16 %.

The data given in Table 3, as well as the patterns of the distribution of displacements shown in Fig. 5, evidence that for all fixation models, the displacement of the platform points mainly occurred in the directions outwards and forwards.

Based on the results of our study, it is possible to formulate recommendations for choosing the position of fixing the graft of the tendon of the popliteal muscle. Thus, obviously, the points that are located laterally and closer to the articular plateau are optimal for fixing the graft on the posterior surface of the tibia. This is indicated by the analysis of data obtained as a result of a numerical study, which is absent in similar works.

It should be noted that these recommendations are of a theoretical nature. From the point of view of medical practice, it can be said that the application of the results of this study may be limited by the peculiarities of the anatomical structure of the knee joint in each individual case.

In addition, the disadvantages of this study include the fact that it was performed on simplified models of ligaments both from the point of view of geometry and physics. This did not take into consideration the dynamic loading of the joint. The factors described above, which were not taken into consideration in this work, indicate the possibility to continue research in this direction, with the construction of more complex computational models.

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7. Conclusions

1. Analysis of the calculation results showed the following. From the point of view of the stability of the tibia under rotational load, the most effective is the fixation of the graft for plastic surgery of the popliteal muscle on the back surface of the tibia as laterally as possible and closer to its articular surface. This is indicated by the magnitude of the movements, which, in this case, turned out to be the smallest in all directions.

The greatest values of movements in all directions were obtained from a control model in which the hamstring of the popliteal muscle was absent. Thus, the studied movements: x_{max}, x_{min}, y_{max}, y_{min}, *d*, obtained from the control model, exceeded similar indicators in models with minimum values by 17, 37, 17, 32, and 16 %.

When the fixation point of the popliteal muscle graft is displaced on the posterior surface of the tibia from the outside to the inside, an increase in movement occurs, which means a decrease in the stability of the lower leg during external rota-

tion. This fact is explained by the fact that the displacement of the fixation point during modeling leads to an increase in the length of the graft, and this, in accordance with Hooke's law, in turn, leads to an increase in its deformations at the same load.

2. The distribution of displacement fields was as follows. The greatest positive displacements of the model points along the x-axis (outward-facing relative to the knee joint) occurred at the extreme front of the cylindrical platform. These values gradually decreased as they moved along the diameter of the platform along the line from front to back, reaching zero in the center of the model. The displacement values then increased again, but in the opposite direction, reaching maximum negative values (in the direction inwards) at the extreme rear of the platform.

Regarding the movements of the model along the y-axis, one can see that their nature was similar to the displacements along the x-axis. At the same time, the maximum positive movements of y_{max} (in the direction anteriorly) occurred on the far-left part of the model, and the largest negative values of y_{min} (rear) – on the far-right side of the platform.

The largest total displacements d occurred on the platform section between the x_{max} and y_{max} points while the vector of these movements was directed to the front outer part of the model.

Conflict of interest

The authors declare that they have no conflict of interest in relation to this research, whether financial, personal, authorship or otherwise, that could affect the research and its results presented in this paper.

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