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Modeling the behavior of the acetabular axis and the axis of the ischial tuberosities during the transition from a standing to a sitting position

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Abstract

Introduction The success of the treatment of patients with degenerative diseases of the spine and concomitant damage to the hip joint depends on the understanding of the biomechanics of movements in the spinal-pelvic segment. After a thorough analysis of the biomechanical processes occurring in the spine-pelvis system during the transition from a standing to a sitting position, it becomes clear that the acetabular axis of rotation of the pelvis in space is not the only one. The **purpose** of the study was to develop and test a virtual model of the pelvis to study the kinematics of the movement of the spinal-pelvic complex with a description of the emergence of the iscial axis of rotation by changing the position from standing to sitting. **Materials and methods** The problem was solved using the finite element method. The bones were modeled as absolutely rigid bodies. The main ligaments and muscles were modeled using finite element springs: elastic fragments with specified rheological characteristics. The study of contact interaction was carried out for pairs: "femoral head – acetabulum" and "ischial tuberosities - chair surface". **Results** A new axis of rotation was revealed, the ischial axis, which corresponded to the points of initial contact of the ischial tuberosities with the surface of the chair. The axis of the acetabulum rotated by 7.1° relative to the ischial axis and at the final moment shifted in the horizontal direction relative to the acetabular axis by 8.83 mm. The gap between the surfaces of the femoral head and the acetabulum was about 8 mm. **Discussion** The study shows that the pelvis rotates depending on the position around two axes: acetabular and ischial ones, hence it follows that the acetabular axis oscillates back and forth during ante- and retroversion, that is, it is non-static. Shortcomings of the model: 1) muscles and ligaments were modeled using FE springs, the end and beginning of which were set by two points, and the muscles and ligaments in the real body are attached along the entire surface of the bones; 2) soft tissues were not modeled in real volume. The merit of the study is the contact interaction of the pelvis with the chair and its rotation relative to the ischial axis, while other studies consider the rotation of the pelvis only relative to the acetabular axis. **Conclusion** A new axis of rotation arises due to the contact interaction of the pelvic bone with the surface of the chair when the skeleton moves from a standing position to a sitting position, the ischial axis. The gap between the surfaces of the femoral head and the acetabulum was about 8 mm. It is advisable to conduct a clinical study.

Keywords: axis of rotation, sagittal balance, motion biomechanics, parameters, pelvis, modeling

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INTRODUCTION

The understanding of the sagittal spine balance has been widely introduced into the scientific and practical activity of spinal surgeons. It was initiated by J. Dubusset, who proposed the concept of the "cone of economy" as part of the study of the global balance [1].

Instrumental studies that investigate changes in the sagittal and global balance, as well as examinations conducted at the stage of preoperative planning in order to assess the required amount of intervention to correct the detected disorders, are carried out only in patient's standing position without considering other body positions.

The straight line drawn through the two centers of rotation of the pelvis (acetabulum) is the axis of rotation of the pelvis in the sagittal plane, from which further indicators are calculated, assessed as part of the study of spinal-pelvic relationships and global balance as a whole. The main parameters of the spinal-pelvic relationship are linked to the rotation center: pelvic index (pelvic incidence, PI), pelvic tilt (PT), femoroacetabular angle, full balance index (FBI) and other calculated formulas, which, in turn, change in regard to patient's posture, age, presence of concomitant orthopedic pathology and as a result of surgical treatment [2, 3, 4, 5, 6, 7, 8].

The established center of rotation and the axis of rotation of the pelvis are determined by biomechanics that was studied much by the French orthopedic surgeon A. Kapandzhi. He described in detail the structure of the femur and pelvis that perfectly illustrates an example of the law of action and reaction, based on a special internal bone tissue structure which forms special groups of force lines that oppose the mechanical forces acting on the bone [9]. The described structure of the proximal femur and acetabulum is explained by their role as a central biomechanical node that bears the entire vertical load and is the center of rotation in the standing position.

In the sitting position, there is no support in the hip joint. Therefore, the above-mentioned law of action and reaction is not realized. The forces acting on the hip joint are shifted to the area of the ischial tuberosities, where the load is shunted. It means that this law moves to another support, the ischial tuberosities, which form a new center of rotation, and two points in the ischial tuberosities form a new axis of rotation. The new point of rotation of the pelvis in the space is located on the ischial tuberosities, and the former generally accepted point of rotation, the acetabulum with the head of the femur, will rotate around it.

It is assumed that in the standing position, the acetabular axis of rotation of the acetabulae and the femoral head coincide, and the action of forces on this axis is in a state of equilibrium. Therefore, the points of the center of the acetabulum, lying on the axis of rotation, are not shifted relative to the axis of rotation.

The null hypothesis of the study is that in reality, by transition to a sitting position, there is a contact interaction between the ischial tuberosities and the surface on which the person sits. As a result, support reaction forces arise in the area of the contact of ischial tuberosities with the surface of the chair, and a second axis of rotation is formed, which passes through two points of initial contact of the ischial tuberosities with the chair, the ischial axis. This axis is the new axis of rotation around which the pelvic bone and spine rotate. As a result, the axis of the acetabula shifts posteriorly relative to the acetabular axis of the femoral head, which coincided in the standing position. Therefore, the acetabula, together with the pelvis, deviate backwards to a certain magnitude. This shift creates a gap between the acetabulum and the femoral head.

Purpose To develop and to test a virtual model of the pelvis to study the kinematics of the movement of the spinal-pelvic complex and describe the emergence of the ischial axis of rotation by changing the initial standing position to a sitting position

Due to the high complexity of the geometry and trajectories of the mechanical system (spine – pelvis – femur), an analytical model cannot be built. Therefore, to describe the scenarios of the movement process, it is proposed to use the methods of three-dimensional modeling and numerical methods. In the problems of biomechanics, the most convenient and applicable is the finite element method [2, 10, 11, 12, 13, 14].

MATERIALS AND METHODS

Geometry of 3-D model

The universal software package ANSYS was used in order to build a three-dimensional model of the biomechanical system under the study and solve the task of the study.

The original geometric models of the pelvic bone, femur, and spine, obtained by MSCT and presented by faceted bodies, were stored in the STL format. To analyze the stress-strain state of the model, it was necessary to convert the original model into a solid model format. The conversion of facet models to solid models (SAT format) was performed semi-automatically using ANSYS SpaceClaim Direct Modeler (SCDM) software, which is a CAD system that implements a direct approach to modeling.

Materials and assumptions, loading diagram

Due to the complexity of the structure of ligaments, bone, soft tissue rheology, as well as the forces that occur in muscles, tendons and ligaments, it is extremely difficult to build a detailed model of a biomechanical system. Moreover, the higher the detail of the model, the longer the calculation takes. It is necessary to find a balance in order to most approximately describe the model, but at the same time not to lose the speed of calculation. Therefore, a number of reasonable assumptions were made.

An important issue while building a model is an issue of mechanical characteristics of materials. Regularly in biomechanical problems, the bone is represented either as a continuum material with homogeneous properties, or as a discrete cell structure with detailed bone microarchitecture, or as a poroelastic material [15, 13]. The last two approaches significantly complicate the finite element model and significantly increase the computational time, especially in cases with a large number of composite bodies in the model, as in the problem under consideration. Moreover, describing the mechanical characteristics of bone tissue, there is a problem with the determination of the ultimate strength and modulus of elasticity for cortical and spongy bone types. The problem is that the range of data for the mechanical characteristics indicated is significant among studies [16, 17, 18].

The values of tensile strength and modulus of elasticity depend not only on the research method, the place from which the test specimen was cut, its condition (dry or wet bone), but also significantly depend on bone density, which changes with age. Since in the problem under consideration, we were primarily interested in the kinematics of motion, therefore, in the first approximation, the assumption was made that the bones were absolutely rigid bodies, i.e. mechanical stresses and deformations in the bones were not calculated. The main ligaments and muscles were modeled using finite element (FE) springs, i.e. elastic fragments with specified rheological characteristics.

The direction of action of the ligaments was modeled in accordance with the diagrams of ligaments presented in the works of Kapanji A. [9]. The pubofemoral ligament was modeled with two springs (Fig. 1c: springs c and d). The circular zone in the ischiofemoral ligament was not modeled, because it is attached only to the femur. In total, six ligament springs were modeled around each joint. The stiffness of the ligaments in different sources also differs. Thus, the book by Martin et al. [19] reports the stiffness of the ligaments in the knee joint of monkeys, equal to 18.49 kg/mm (≈ 185000 N/m), and in the article of Kubo et al. [20] ligament stiffness for a middle-aged man is 26.1 N/mm (≈ 26100 N/m). This smallest value was taken as an approximate value for the stiffness of the ligaments in the hip joint. Since six FE springs were modeled, this value was divided by 6 accordingly. Thus, each spring was given a stiffness value equal to 4350 N/m.

The direction of muscle action was also modeled by FE springs in accordance with the diagrams presented in Kapanji's book [9]. A complete diagram of the flexor muscles is shown in Figure 2. There is not much data on muscle stiffness in the literature; moreover, the stiffness of each muscle is different. Thus, several articles [20, 21] presented the graphs of muscle stiffness depending on the angular velocity using the example of the medial gastrocnemius muscle. The passive and active action of the muscle was considered. An article by Gervasi [22] reported the effect of treatment in cyclists for muscle tension in the thigh. Stiffness data were given for tensed muscles in an active and relaxed state for the rectus and biceps femoris muscles.

For the calculation, the case was considered when the stiffness of the muscles is the smallest in the passive state. In this case, the stiffness is 250 N/m for the rectus femoris muscle and 270 N/m for the biceps muscle. Since the stiffness of each thigh muscle is different and data for each muscle are not presented in the literature, an approximate generalized case was considered as an assumption in the calculation, when the stiffness of all thigh muscles corresponds to the stiffness of the rectus muscle. Accordingly, in the calculation, the stiffness of each spring simulating the muscle was set to 250 N/m.

In the first assumption, the vertebrae are rigidly connected to each other. In detailed models, the influence of the spine can be considered, for example, by introducing elastic fragments with certain rheological characteristics into the model of the spine.

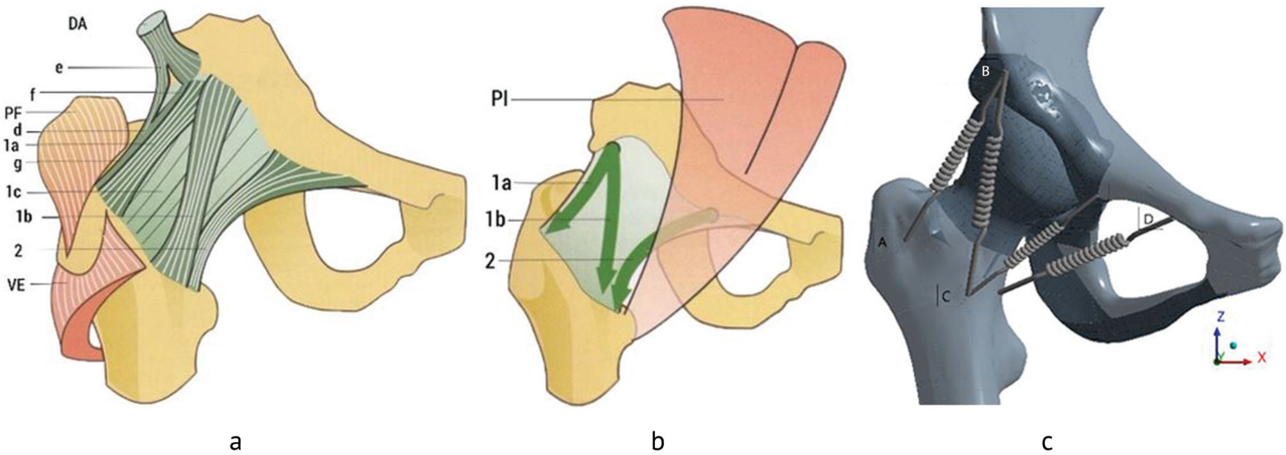

Fig. 1 Ligaments of the hip joint: *a* schematic image [9]; *b* direction of action forces in these ligaments [9]; *c* modeling of the action of ligaments by means of FE springs in the model. Designations: 1 and 2: pubic-femoral ligaments; 1a and 1b: iliofemoral ligaments

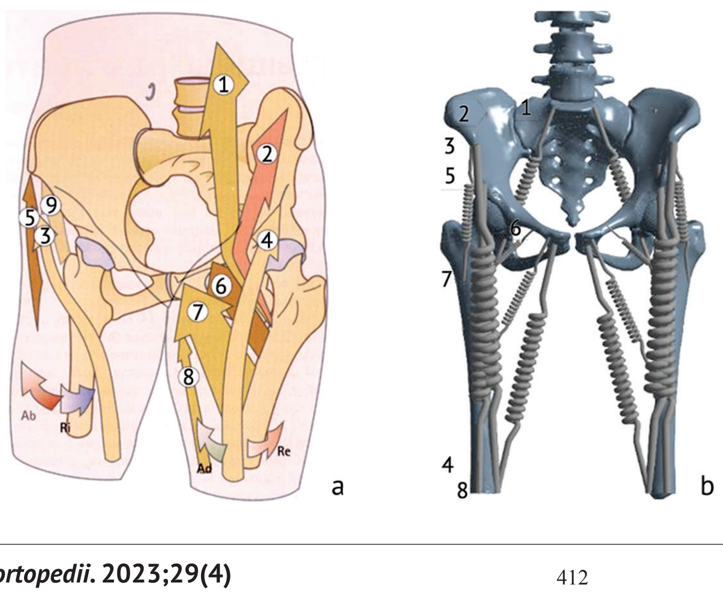

Fig. 2 Flexor muscles: *a* schematic image [9]; *b* modeling of muscle action by means of FE springs in the model. Designations: 1: psoas muscle; 2: iliac muscle; 3: sartorius muscle; 4: rectus femoris; 5: tensor fascia latae; 6: pectineous; 7: adductor longus; 8: gracilis; 9: the most anterior fibers of the gluteus minimus and medius

Joints are modeled using models of mechanical contact of bodies (friction-sliding contacts). The contact interaction of the femur with the pelvic bone, as well as the contact interaction of the endoprosthesis components, is described using contact interfaces that determine the friction-sliding contact. Correct modeling of the contact interaction between the femoral head and the acetabulum required surface smoothing (approximation of the acetabulum surface to a sphere with a diameter of ~23 mm).

To analyze the contact interaction on the surface in the area of contact between the femur and the acetabulum and in the area of the contact between the ischial tuberosities and the surface on which the skeleton sat, the following mechanical characteristics were set for the trabecular bone. The modulus of elasticity was assumed to be 389 MPa, Poisson's ratio was 0.3, and the density was 0.5 g/cm^3 [19, 23].

Prior to constructing the loading diagram, the angles PT and SS were compared between the three-dimensional model and CT images of the patient (Fig. 3).

In the 3D model, a "chair" was additionally built shaped as a platform, on which the skeleton sat in order to correctly model the contact between the pelvic bone and the chair.

In order to exclude the "set" movement as much as possible, only the dead weight of the body and the end position SS of the endplate angle in the sitting position were indicated in the loading diagram. The force was applied perpendicular to the surface of the endplate and measured 1000 N, what corresponds to a weight of 100 kg. In the sitting position according to the CT image of the patient, the SS angle is 7°, and in the model in the initial standing position, the SS angle is 27°. Based on this, a rotation angle of 34° was applied to the endplate in the model (as a loading condition), assuming that in the final position of the sitting model, the angle SS of the model should correspond to the angle SS of the CT image.

As boundary conditions, a fixation in space was made for the lower surface of the "chair" platform, where movements in all directions were prohibited: $U_x = U_y = U_z = 0$. In addition, movements in the knee area were prohibited in the directions $U_x = U_y = U_z = 0$; but rotations were allowed about the axis x: $Rot_x = free$ and prohibited relative and z: $Rot_y = Rot_z = 0$. The diagram of loading and fixing the model is shown in Figure 4.

The mathematical model included a system of equations of the theory of elasticity: equilibrium equations, defining relations, Cauchy equations for small deformations. Contact pairs of contact and target finite elements were created at the contact boundary between the acetabulum and the femoral head, as well as between the ischial tuberosities and the "chair" surface. Contact pairs were specified by friction. The coefficient of friction was 0.2. Thereby, contact conditions were implemented in each unit at the contact boundary: gap; compression contact forces $RN \leq 0$. The condition of the integrity of the parts of the model was also fulfilled, so the condition of the connectedness of the contact between the spine and the pelvic bone $U_{x1} = U_{x2}$; $U_{y1} = U_{y2}$; $U_{z1} = U_{z2}$, where movements with index 1 mean movements at the contact boundary in the spine, and with an index of 2 movements at the contact boundary in the pelvic bone. The Newton-Raphson step-by-step load incremental method was used to solve nonlinear problems with an additional linear search algorithm; method of penalty functions for solving the contact problem.

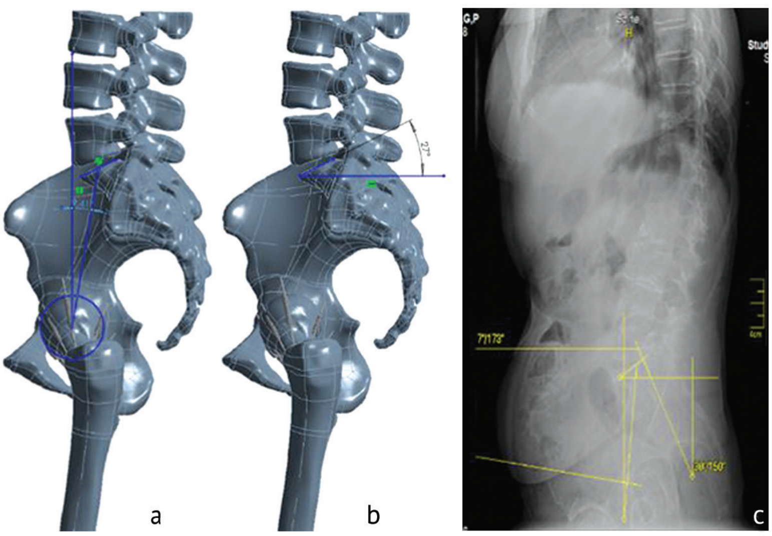

Fig. 3 Comparison of PT and SS angles in the standing position: *a* PT for the original 3D model 9°; *b* SS for the original 3D model 27°; *c* angles PT = 7° and SS = 30° according to the CT image of the patient

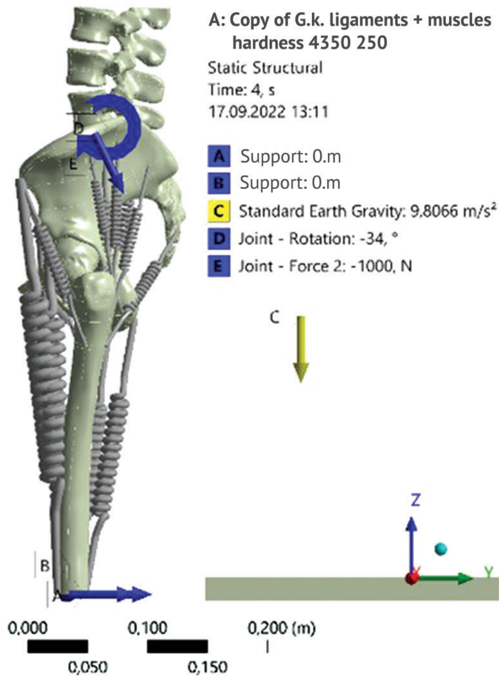

Fig. 4 Diagram of loading and fixing 3D-model

RESULTS

The solution of the problem resulted in the kinematics of the movement of the hip joint by its changing the position from standing to sitting, considering the contact interaction of the chair surface with the pelvic bone. Diagrams of the movement of the femoral head and the angles PT and SS in the sitting position were obtained (Fig. 5).

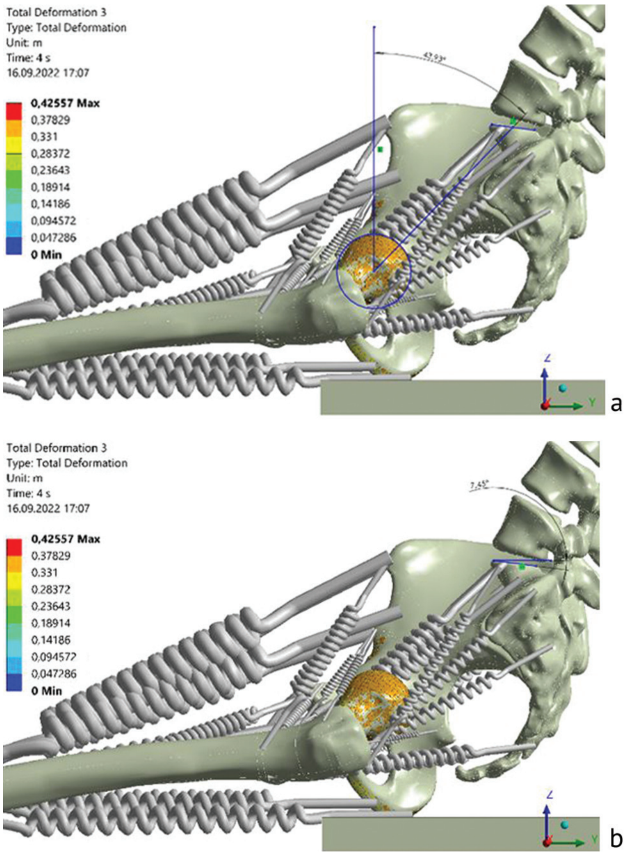

Fig. 5 Diagram of the amount of movement (in meters) of the femoral head from a standing position to a sitting position. Angles PT and SS in the final sitting position: *a* PT angle = 43.9°; *b* SS angle = 7.45°

Comparison of the PT and SS angles of the original virtual model with the angles obtained on images of a real patient in two positions is given in Table 1. The difference between the angles for CT images and the 3D model before and after the calculation corresponds to 2-3°, which indicates a good correlation of the models with a real patient.

Table 1

Comparison of 3D-model and CT scan of a patient

Study	Parameter of sagittal balance			
	Standing position		Sitting position	
	PT	SS	PT	SS
3D-model	9°	27°	43.9°	7.45°
CT image	7°	30°	41°	7°

In accordance with the null hypothesis, we were interested in the displacement of the axis of rotation of the acetabulum relative to the acetabular axis of rotation of the femoral head in the sitting position. Therefore, at the initial moment of time for the model in the standing position, the projection of the acetabular axis of rotation onto the acetabulum was performed. The projection point corresponded to the finite element mesh node (Fig. 6). The coordinates of this point (node), which belongs to the surface of the acetabulum, were recorded at the initial moment of time, at the moment of contact of the pelvic bone with the surface of the chair, and at the final moment of time, when the skeleton was in the final sitting position.

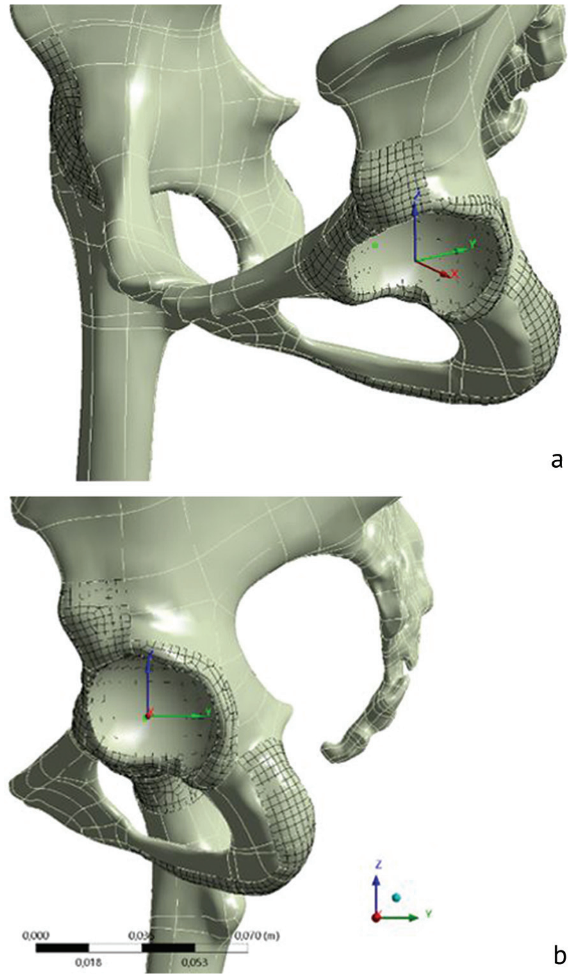

Fig. 6 Projection point of the acetabular axis of rotation on the surface of the acetabulum: *a* 3D view; *b* lateral view

Due to the fact that the shape of the ischial tuberosities is arched, these arches “roll” over the surface upon contact with the chair. Thus, the points of contact of the ischial tuberosities with the chair change in the interval between the initial contact with the chair and the final position of the skeleton in the sitting position. This leads to the conclusion that the ischial axis is static and does not belong to a fixed point on the surface of the ischial tuberosities, but to the surface of the chair, and the axis passes through two points of initial contact of the ischial tuberosities with the chair. This axis is the new axis of rotation, relative to which the rotation of the pelvic bone and spine occurs.

In order to determine where the ischial axis passes in the model, the point of initial contact of the pelvic bone with the surface of the chair was fixed. Those points were selected according to the highest contact pressure on the surface of the ischial tuberosity at the moment when the bone touched the chair.

Figure 7 clearly shows that after the pelvic bone contacts the surface of the chair, a posterior deviation of the pelvic bone occurs and a gap is formed between the femoral head and the acetabulum, while the femoral head remains in the same position.

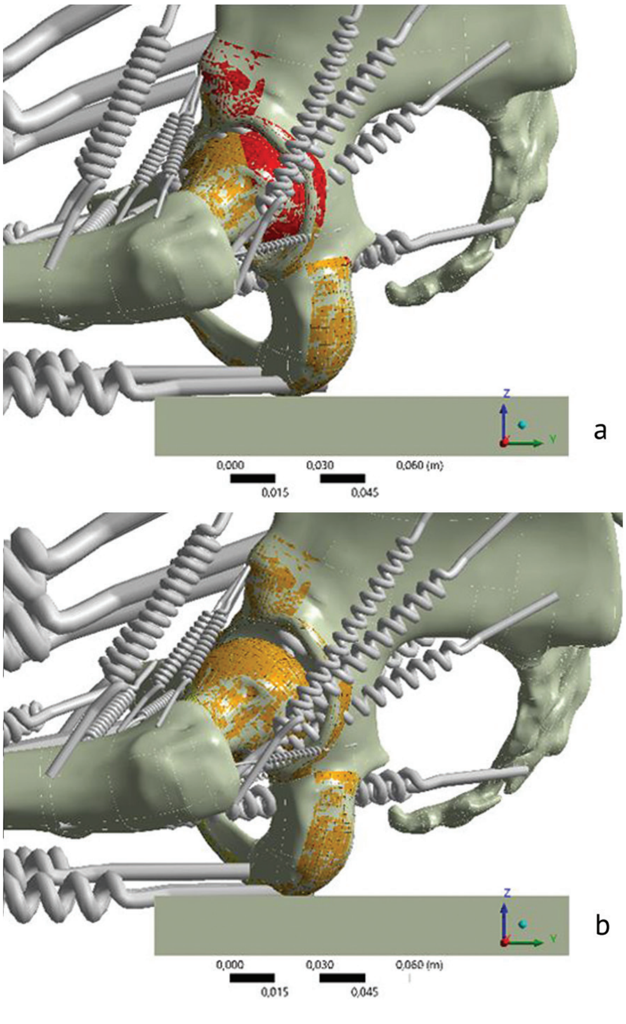

Fig. 7 Contact of ischial tuberosities with the surface of the chair: *a* the moment of contact with the chair; *b* final sitting position

Next, the coordinates of the points were plotted on the plane, and the angle of rotation of the point of the acetabular axis relative to the ischial axis was measured (Fig. 8).

Figure 8 shows that the axis of the acetabulum rotated by 7.1° and shifted along the arc by 8.83 mm.

Moreover, based on the solution of the contact problem, the following was revealed: the gap between the surfaces of the femoral head and the acetabulum, the maximum value was about 8 mm, as well as the amount of sliding of the femoral head along the surface of the acetabulum was 9.3 mm (Fig. 9).

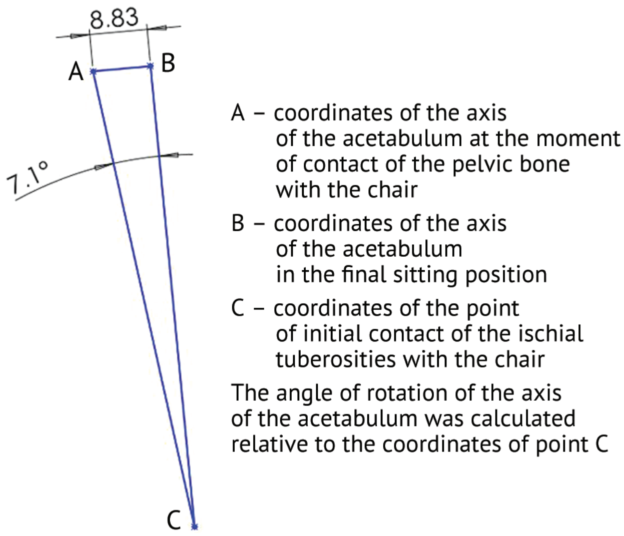

Fig. 8 Diagram of movement of the point of the acetabulum axis relative to the ischial axis

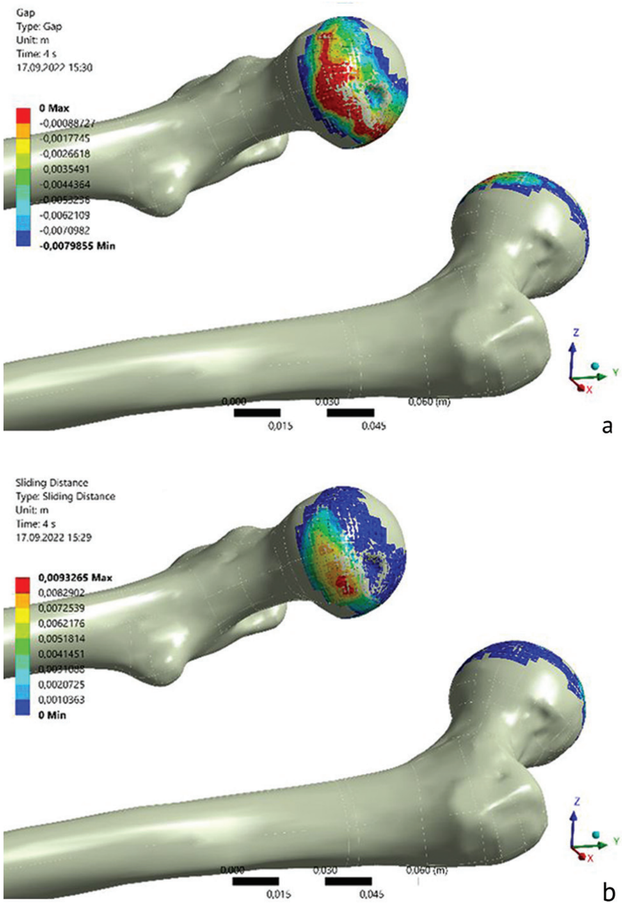

Fig. 9 Area of contact between the femoral head and the acetabulum (the color scale corresponds to meters): a gap size diagram; b the amount of sliding of the femoral head on the surface of the acetabulum

The pressure in the area of contact between the ischial tuberosities and the chair was also studied, the maximum value of which was 81.8 MPa. The amount of sliding of the ischial tubercles on the surface of the chair was 6.5 mm (Fig. 10).

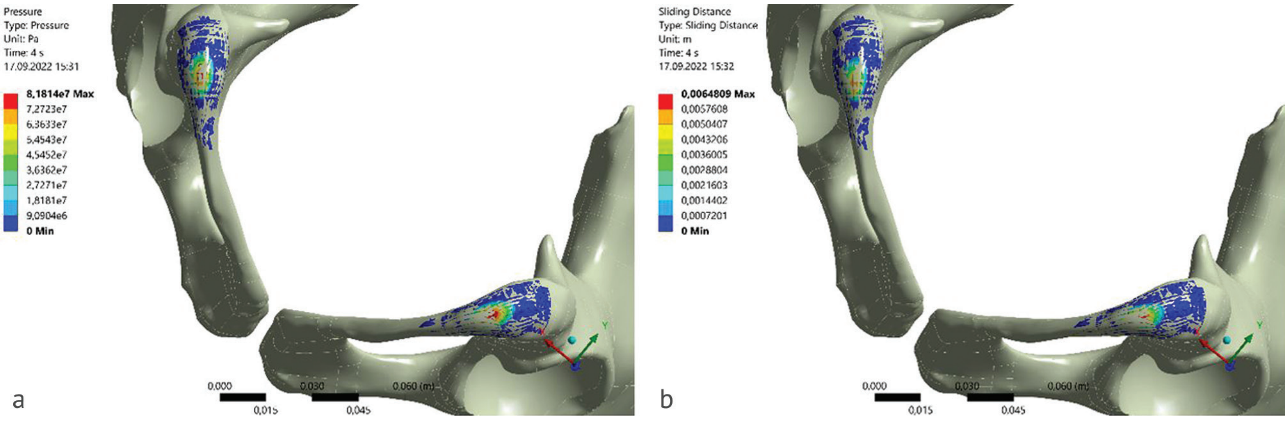

Fig. 10 Contact area of ischial tuberosities with the surface of the chair: *a* diagram of contact pressure distribution in MPa; *b* diagram of the magnitude of slip in meters

DISCUSSION

The studied problem of modeling the behavior of the acetabular axis and the ischial axis during the transition from a standing to a sitting position is relevant for understanding the kinematics of the hip joint movement and interaction with the chair surface.

Our study shows that the pelvis rotates depending on its position around two axes: in the standing position around the axis drawn through the centers of rotation in the heads of the femur (acetabular), and the second, in the sitting position, around the axis drawn through the ischial tuberosities. If we accept the assumption that, on the basis of the created biomechanical model, the axis of rotation passing through the ischial tuberosities takes place, then it follows that the acetabular axis of rotation oscillates back and forth during ante- and retroversion, that is, it is not static. The described movement of the acetabulum around the new center of rotation, the ischial axis, has certain amplitude, which differs for every individual. The lever arm, the distance from the new center of rotation to the previous one, differs in men and women and it influences the amplitude of the described movement.

As a critical point of the studied model may be the point that muscles and ligaments were modeled using FE springs, the end and beginning of which were set by two points, while the muscles and ligaments in the real body are attached along the entire surface of the bones. The behavior of the model depends on the stiffness of the introduced springs. The introduction of the stiffness values of each muscle and each ligament into the model would provide more accurate results. However, such data are extremely rare in the literature and not available for all components of the ligaments and muscles; moreover, for some of them it is not possible to obtain stiffness values, so averaged values were used. However, the assumptions introduced do not contradict similar assumptions in other studies [24, 25].

Moreover, the shortcoming of the model is the fact that the real volume of soft tissues, such as muscles and skin, was not modeled. Thus, in real life there is a layer of soft tissues between the ischium and the surface of the chair, which can provide hyperelastic behavior of the muscles in the contact area and, accordingly, large displacements (sliding) of the bone, at a distance equal to the thickness of the soft layer, relative to the surface of the chair. Another shortcoming is the fact that the bones were modeled as absolutely rigid bodies, and only in the contact area there was the material corresponding to the mechanical parameters of the trabecular bone. Thus, the model of an isotropic elastic body was used in the study and inelastic deformations were not considered, although they are present in real bones. However, it should be emphasized that the authors had the task of analyzing the kinematics of the movement of the skeleton, and not the analysis of the stress-strain state of the skeleton components, and the assumptions introduced in the model were justified.

The merit of this study is that it reveals the contact interaction of the pelvic bone with the chair and describes the rotation of the pelvic bone relative to the new ischial axis, while other studies consider the rotation of the hip joint only relative to the acetabular axis [4, 10, 26, 27, 28]. Daisuke Nishiyama's study evaluated the factors influencing the pelvic tilt in the sitting position after fixation of the spine; thereby, the parameters of the sagittal balance in the standing and sitting positions were assessed, the pelvic tilt was considered, the axis of rotation of which was the acetabular axis [28]. Lazennec et al studied the effect of changes in the sagittal spinal-pelvic shift in various positions of the body on the frequency of instability of the hip joint. Complex relationship of the spine-pelvis system and the femur was studied, the tilt of the pelvis, anterior-posterior movement of the pelvis in the sagittal plane were described but without the interaction of the pelvis with the support [29]. Kanawade et al. studied the change in the spatial characteristics (tilt, anteversion) of the acetabulum in different positions of the pelvis; however, as in the studies of other authors, only one axis of rotation in th sagittal plane was taken into account [30].

CONCLUSION

A new axis of rotation arises due to the contact interaction of the pelvic bone with the surface of the chair when the skeleton moves from a standing position to a sitting position, the ischial axis. The axis of the acetabulum rotates by 7.1° relative to the ischial axis and at the final moment shifts in the horizontal direction relative to the acetabular axis by 8.83 mm. The gap between the surfaces of the femoral head and the acetabulum was about 8 mm.

The developed model enables to study the behavior of the spinal-pelvic complex during the transition from a standing position to a sitting position. The study of the biomechanical features of changes in the spinal-pelvic complex may predict the transformation of the acetabular component of the endoprosthesis in patients undergoing total hip arthroplasty when changing body position. The data revealed in the course of the study suggest that biomechanical aspects may be a possible cause of instability of the hip joint endoprosthesis in group VI according to the Glenn D. Wera classification [31]. Further clinical research is needed.

Conflict of interest *The authors declare that there are no conflicts of interests.*

Source of financing *The authors did not have any sponsorship to conduct this study.*

Ethical expertise *Not required.*

informed consent *Not required.*

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