УДК 616.728.3:616.72-002.77]:612.76](045)

DOI: http://dx.doi.org/10.15674/0030-59872024432-40

Biomechanical particularities of knee joint flexion defomation in rheumatoid arthritis patients

S. I. Herasymenko, M. V. Poluliakh, A. M. Babko,
A. S. Herasymenko, D. I. Kachan, D. M. Poluliakh

SI «Institute of Traumatology and Orthopedics of NAMS of Ukraine», Kyiv

Objective. To research biomechanical particularities of knee joint flexion defomation in rheumatoid arthritis patients. Methods. The work is based on the analysis of the preoperative examination of 23 patients with RA with multi-plane deformations of the lower limb, who were treated in our clinic in the period from 2021 to 2024. The average age of the examined patients was (49.14 ± 6.37) years old. The quantitative assessment of contractures and computed tomography (CT) of the knee were performed on the patients. Results. As can be seen from the obtained results, the stresses in the contact zone on the lateral condyle of the tibial plateau increase by 36.71 % in conditions of flexion contracture of the knee 30°, on the medial condyle of the tibial plateau — by 36.64 %. It was established that the load on the condyles of the tibial plateau increased by 12.8 % in forced passive flexion of the knee joint by 6°, as a result of passive extension by 3° — the load on the condyles of the tibial plateau increased by 95.2 %, which is a critical and can lead to the degradation of bone tissue in the contact area and increase the risks of secondary arthrosis. Conclusions. The simulated solid 3D-model of the knee joint was developed and demonstrates that in the knee joint flexion position of 30°, according to the radius of curvature of the mediaal and lateral condyle, the contact area on the medial condyle of the tibial plateau is 2 times larger than that on the lateral one and is shifted slightly backwards. The standing with one support under the weight of a body 60 kg (600 N) in conditions of flexion contracture in the knee joint in a position of 30°, there is an increase in contact stresses on the external condyle of the tibial plateau by 36.71 %, on the internal condyle — by 36.64 %, which contributes to the progression of the clinical picture of RA with an increase in the phenomena of osteoarthritis specifically in the back parts of the knee joint.

Мета. Вивчити особливості біомеханіки колінного суглоба за наявності згинальної контрактури у хворих на ревматоїдний артрит (РА). Методи. Проаналізовано результати передопераційного обстеження 23 хворих на РА з багатоплощинними деформаціями колінних суглобів (КоС), які знаходилися на лікуванні в ДУ «Інститут травматології та ортопедії НАМН України» у період із 2021 по 2024 р. Середній вік обстежених пацієнтів складав (49,14 ± 6,37) року. Хворим виконували кількісне оцінювання контрактур та комп'ютерну томографію (КТ) КоС. Результати. Виявлено, що в умовах згинально-розгинальної контрактури КоС у положенні 30° напруження в зоні контакту на зовнішньому виростку плато великогомілкової кістки збільшуються на 36,71 %, а на внутрішньому — на 36,64 %. На основі біомеханічних досліджень встановлено, що унаслідок примусового пасивного згинання в колінному суглобі на 6° навантаження на виростки плато великогомілкової кістки збільшилося на 12,8 %, через пасивне розгинання на 3°, тоді як навантаження збільшилося на 95,2 %, що є критичною величиною та може призводити до деградації кісткової тканини в ділянці контакту та посилення вторинного артрозу. Висновки. Розроблено імітаційну твердотільну 3D-модель КоС, яка демонструє, що в положенні згинання колінного суглоба 30° відповідно до радіуса кривизни зовнішнього та внутрішнього виростка площа контакту на внутрішньому виростку плато великогомілкової кістки в 2 рази більше за такий на зовнішньому і зміщені дещо дозаду. За одноопорного стояння під дією ваги тіла 60 кг (600 Н) в умовах згинально-розгинальної контрактури в КоС у положенні 30° спостерігається зростання контактних напружень на зовнішньому виростку плато великогомілкової кістки на 36,71 %, на внутрішньому виростку — на 36,64 %, що спричинює прогресування РА зі збільшенням прояву остеоартрозу саме в задніх відділах КоС. Ключові слова. Ревматоїдний артрит, колінний суглоб, згинальна контрактура, скінченно-елементне моделювання, твердотільна модель, коефіцієнт Хаунсфілда.

Key words. Rheumatoid arthritis, knee joint, flexion contracture, finite element modeling, solid model, Hounsfield units

Introduction

Rheumatoid arthritis (RA) is a chronic progressive inflammatory disease that manifests as symmetrical polyarthritis of small and large joints and leads to damage to intra- and periarticular structures due to systemic inflammation [1, 2]. Despite the continuous development of medicine, RA research, the emergence of new effective drugs and treatment strategies, patients with RA continue to complain of persistent pain, swelling in the joints and functional limitations of the limbs [3]. RA is an autoimmune disease that is accompanied by proliferation of the synovial membrane and destruction of articular cartilage, which leads to disability [4, 5]. Biomechanical factors in RA may play an important role in the onset and development of degenerative changes in the joints, secondary processes are manifested through the inflammatory process [6]. The decrease in muscle strength observed in clinical studies is associated with the activity of the inflammatory state, radiological abnormalities and the degree of functional disorders [7]. The disease is also characterized by contracture of the periarticular soft tissues, the formation of valgus deformity and flexion contracture of the knee joint (KJ), a decrease in bone density and its constant progression [8–10]. The sequence of biomechanical and biochemical processes that regulate these mechanisms *in vivo* is still not sufficiently understood [11, 12]. Studying the magnitude of such loads on the articular surfaces in conditions of contracture of the joints of the lower extremities in RA and the participation of muscle forces in this process may contribute to the development of new views and approaches to the tactics of therapeutic measures specific for each stage of the disease [13]. Even a slight limitation of extension in KJ has a negative effect on gait and functional ability. Flexion contracture increases energy expenditure and places excessive strain on the quadriceps femoris [14]. It also causes pain and impairment of the limb as a whole. If the biomechanics of the lower extremities are altered, the effect of flexion contracture on the KJ extends beyond the affected joint.

Compensatory mechanisms of the musculoskeletal system, in the presence of flexion contracture in the knee joint, affect the biomechanics of other joints of both lower extremities during gait [15]. Clinicians must have a clear idea of the characteristics of damaged tissues, and treatment and rehabilitation accordingly require knowledge of the characteristics of tissue healing, biomechanics of the knee joint and lower extremities, neuromuscular physiology, activity associated with the performance of certain tasks under axial load and without [16].

Purpose: to study the features of the biomechanics of the knee joint in the presence of flexion contracture in patients with rheumatoid arthritis.

Materials and methods

The results of the preoperative examination of 23 RA patients with multiplanar deformities of the knee joint, who were treated in the clinic of joint diseases in adults of the State Establishment "Institute of Traumatology and Orthopedics of the National Academy of Medical Sciences of Ukraine" in the period from 2021 to 2024 were analyzed. The average age of the examined patients was (49.14 ± 6.37) years. Patients underwent quantitative assessment of contractures and computed tomography (CT) of the KJ.

The materials of the article were reviewed at a meeting of the Bioethics Commission of the State Establishment "Institute of Traumatology and Orthopedics of the National Academy of Medical Sciences of Ukraine" and recognized as received in accordance with the requirements of the Helsinki Convention of the Council of Europe on Human Rights and Biomedicine, the relevant Laws of Ukraine and recommended for publication (Protocol No. 4 dated 10.07.2024). Informed consent was obtained from all patients for processing the results of treatment.

The methodology for quantitative assessment of contractures of large joints is based on the task of determining their condition, susceptibility to external corrective action, and elasticity of joint tissues. The determination of the tissue response to the applied force is based on the registration of the resistance force to stretching of the altered articular and periarticular tissues under contracture conditions, as well as changes in the joint angle under dosed load.

At the initial stage, for further calculations, both analytical and finite element modeling, a simulated solid 3D model of the knee joint was created, which included the femur and both tibias, cartilage of the femoral condyles and the tibial plateau, and menisci (Fig. 1). The analytical calculations were based on the indicators of quantitative assessment of KJ contracture (susceptibility to external corrective action) and the value of joint forces that arise during the gait of a patient with RA. Their measurement was carried out using a manual muscle tester (MMT) and a goniometer. MMT was developed based on a mechanical force sensor — a PMT-1 strain gauge (portable mass transducer). Applying force to a limb segment using MMT leads to changes in the strain gauge indicators.

The change in current strength in the electrical circuit reflects the changes in the forces applied to the device, that is, the measured non-electrical quantity is converted into an electrical signal, which allows measuring the mechanical quantity. During the research, each moment of time is characterized by the degree of deformation of the sensor under the influence of the forces applied to it. The obtained values were recorded in the software environment “EXPANDER” (approved at the meeting of the Academic Council of the State Establishment “Institute of Traumatology and Orthopedics of the National Academy of Medical Sciences of Ukraine” dated 22.12.2010, protocol No. 16, certificate of measuring capabilities No. PT - 107/21 dated 09.03.21, SE “Ukrmetrteststandart”).

When measuring the compliance of the contracture with the use of MMT, the resistance of the joint tissues under the conditions of contracture of corrective movement in the range of limits from active to passive movement in the joint was determined, namely, the force that must be applied to the limb segment from the maximum active to passive angle of flexion or extension, respectively, was found. The MMT was connected to a software and computer complex via an analog-to-digital converter. Before starting the measurements, the tester was calibrated.

During the establishment of the contracture, a change in the angle of rotation in the joint was detected under the action of an applied dosed standardized force of 5 kg (50 N). The obtained measurement parameters made it possible to calculate the value of the compliance of the contracture, which limits free movement in the KJ. The operator, applying MMT with a dosed force of 5 kg (50 N) to the distal segment of the patient's limb, performed the maximum possible passive movement in the KJ from the position to which the patient was able to perform active movement. The range of passive movements was recorded according to the goniometer data. The dosed force was provided according to the MMT indicators in the “EXPANDER” software environment. The resistance of the contracture was measured in both directions (flexion and extension) with the determination of the angle by which the position in the joint changes under the action of a dosed force of 5 kg.

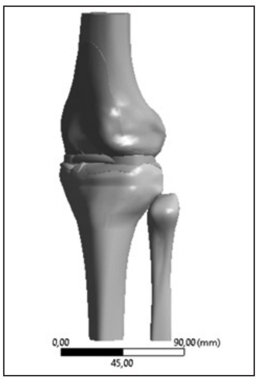


Fig. 1. SolidWorks solid model

Measurements were performed in 5 patients with RA with typical clinical manifestations with predominant damage to the hip and knee joints, expressed by pain syndromes. In patients, the body position was typical for severe forms of RA with combined contractures in the joints of the lower extremities, which determine the type of gait.

Determination of stresses and loads in the KJ was carried out with a flexion-extension contracture of 30°. By measuring the indicators of the susceptibility of the KJ contracture to external corrective action, the average values $\Delta\alpha$ of the angle of passive movement $\Delta\alpha$ were obtained, by which the position in the joint changes under the action of a dosed force $F = 50$ N: in the direction of flexion — $\Delta\alpha = 3°$, extension — $\Delta\alpha = 6°$.

Joint forces during walking of a patient with RA weighing 60 kg (600 N) in conditions of flexion-extension contracture in the 30° position and normal values obtained as a result of our own simulated musculoskeletal modeling in the AnyBody software package are given in Table 1.

The study was performed on the basis of the Biomechanics Laboratory of the State Establishment “Institute of Traumatology and Orthopedics of the National Academy of Medical Sciences of Ukraine”.

Based on CT scans, the spatial geometry of the knee joint of a patient with RA was reproduced using the Mimics 10 software package (Materialise, Ann Arbor, MI) in automatic and semi-automatic modes (Fig. 2).

For analysis, 14 layered sections with a thickness of 1 mm of the subchondral area of the tibia were selected, with their subsequent segmentation using Mimics 10 software and automatic measurement of the tissue absorption coefficient (Hounsfield, HU), in the defined heterogeneous areas of cortical and cancellous tissues 1–6 in Fig. 3. Basic segmentation

Table 1

Joint force during walking of the patient under the action of his weight - 60 kg (600 N)

Joint force	Contracture 30°	Norm
Antero Posterior (X)	1464.19	804.20
Medio Lateral (Y)	564.50	479.33
Proximo Distal (Z)	2265.62	2512.32

in the Mimics software is performed by selecting zones with the same or similar brightness values, with the designation of their range.

At the next stage, solid-state models of layered sections of the proximal tibia were created using the SolidWorks software package, taking into account the topography of their heterogeneous structure (Fig. 3).

The following distally located layers were considered as cancellous and cortical bones with a homogeneous structure (Fig. 4).

Measured using the polylinear bone absorption coefficient (HU), converted into the values of the mechanical properties of bone tissue according to the formulas:

$$\rho = 1{,}067HU + 131, \quad (1)$$

$$E = 3{,}64\rho - 506, \quad (2)$$

where ρ is the density in kg/m^3, E is the elastic modulus (MPa), HU is the bone absorption coefficient. The indicators of the Hounsfield coefficient and the corresponding values of the Young's elastic modulus are given in Table. 2.

The solid samples prepared using SolidWorks were exported to the ANSYS software environment, where a 3D computational model was assembled from them and the boundary conditions for fastening and loading were set.

The latter was carried out uniformly along the plane of the upper cut from the side of the tibial

Table 2

Hounsfield coefficient values and corresponding Young's modulus values

Cut	Area						
	1	2	3	4	5	6	7
Hounsfield bone absorption coefficient (HU)							
1	931.8	436.6	582.8	700.9	536.5	47.8	—
2	775.2	344.3	616.9	565.7	519.4	69.6	—
3	656.3	273.9	567.8	401.2	490.6	70.5	—
4	570.5	232.8	489.2	287.9	434.5	66.5	—
5	513.7	213.0	338.3	289.2	390.8	50.9	351.8
6	467.2	192.5	222.7	298.4	351.3	32.1	313.9
7	401.6	170.1	147.9	319.8	305.9	15.8	299.6
8	336.0	143.9	119.5	343.1	260.3	11.9	308.7
9	249.9	103.3	117.3	569.6	571.4	6.2	315.8
10	175.0	60.4	189.5	473.4	247.7	25.9	292.4
11	280.1	64.9	427.9	423.6	401.2	54.9	301.2
12	253.4	67.5	469.0	389.3	258.7	62.6	290.6
13	224.5	71.0	464.2	319.4	772.6	79.8	344.6
14	234.5	73.5	426.2	314.2	808.6	82.3	393.3
The value of Young's modulus of elasticity (MPa)							
1	3590	1667	2234	2693	2055	156	—
2	2982	1308	2367	2168	1988	241	—
3	2520	1035	2176	1529	1876	245	—
4	2187	875	1871	1089	1658	229	—
5	1966	798	1285	1094	1489	169	1337
6	1785	718	836	1130	1335	96	1190
7	1531	631	545	1213	1159	32	1134
8	1276	530	435	1303	982	17	1170
9	941	372	426	2183	2190	–5	1197
10	651	205	707	1809	933	71	1106
11	1059	223	1633	1616	1529	184	1141
12	955	233	1792	1483	976	214	1099
13	843	247	1774	1211	2972	281	1309
14	882	256	1626	1191	3111	290	1498

plateau, in the direction of the support surface with a force of 750 N, which corresponds to the load of an average body weight of 75 kg during single-support standing.

For uniform load distribution over the cut surface, the pressure value was calculated as follows:

$$P = \frac{M \cdot g}{S} = \frac{75\ \text{кг} \cdot 9{,}81\ \text{м/с}^2}{670{,}5\ \text{мм}^2} = 1{,}1\ \text{MPa}. \qquad (3)$$

Furthermore, a finite element (FE) model of the proximal tibia was generated in a semi-automatic mode, which included 776,422 nodes and 325,030 elements. Under this condition, tetrahedral elements with quadratic approximation of functions prevailed. In order to increase the accuracy of calculations in critical zones, the finite element mesh was densified. It consisted mainly of tetrahedral elements (Tetrahedrons), the size of which on the main model does not exceed 1 mm, in

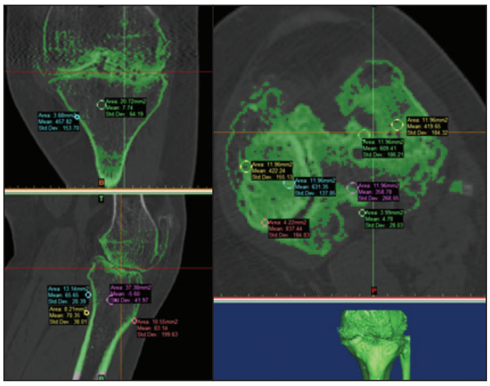


Fig. 2. CT scan: spatial geometry of the knee joint of a patient with RA

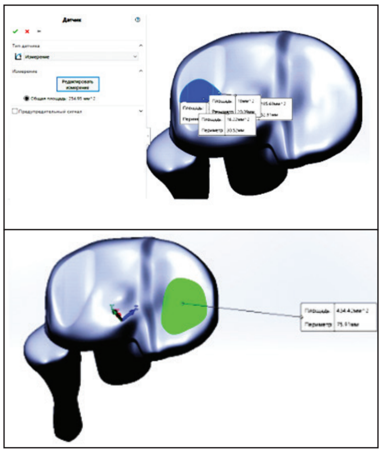


Fig. 5. Model of the contact surface area for the internal and external condyles of the tibial plateau

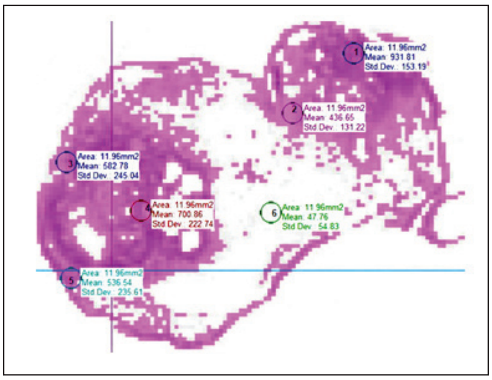


Fig. 3. CT scan: measurement of the Hounsfield coefficient in specific heterogeneous areas of cortical and cancellous tissues

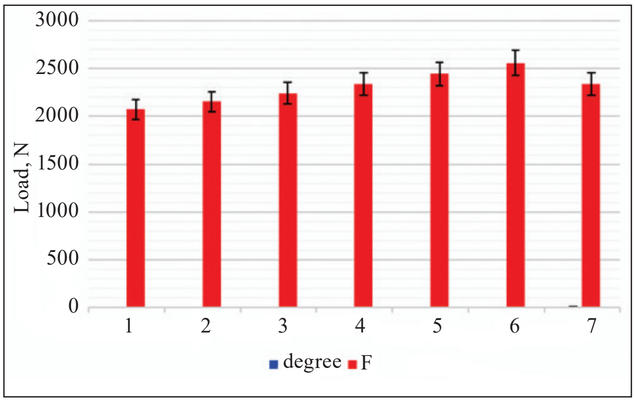


Fig. 6. Diagram of the load on the knee joint during passive flexion at 6°, taking into account standard deviations

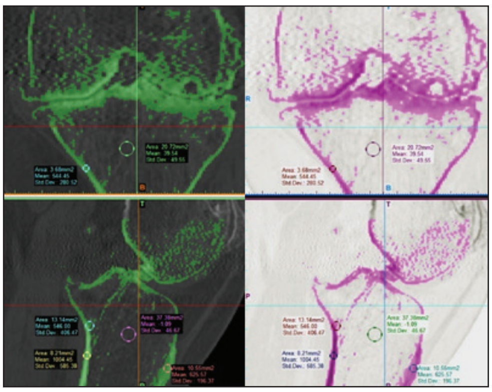


Fig. 4. CT scan: distally located layers of the proximal tibia with a homogeneous structure

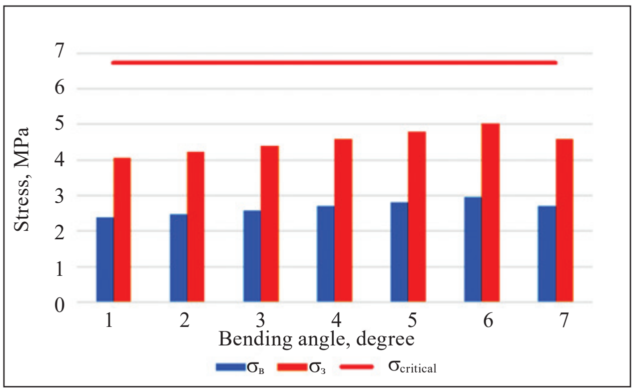


Fig. 7. Stress diagram on the contact surfaces of the tibial plateau during passive flexion at 6°

places of densification 0.1–0.5 mm. Further calculations were carried out using the finite element method, which allows us to study the evolution of the deformation process under the load of the elements of the simulation model. The analysis of the stress-strain state (SSS) was performed on the main elements of the model.

Calculation of the load on the articular surface of the tibial plateau in the case of forced passive flexion at 6° in the KJ with a force of 50 N

According to the calculations, the total force was calculated by the formula:

$$\sum F_{сум} = \int_{25}^{31} \frac{F_{50} \cdot l_4}{l_3 \cdot \sin 30 \cdot \sin x} dx, \quad (4)$$

where x is the angle of action of the total force on the tibial plateau; Ftot is the total force acting in the KJ with a flexion contracture in the position of 30°.

Next, the total force was calculated during additional passive movement of the limb segment in the flexion range from 1° to 6°.

To determine the stress indices on the internal and external condyles of the tibial plateau, the following ratio was used:

$$\sigma_в = F_{вi} / S_{вi}, \quad (5)$$

where F_{ii} is the force acting on the internal condyle (i = 6); S_{ii} = 434 mm^2 is the contact area of the internal condyle.

And the ratio $\sigma = F_{зi} / S_{зi}$, (6)

where F_{ei} is the force acting on the external condyle (i = 6); S_{ei} = 255 mm^2 — the contact area of the external condyle.

The obtained values of forces and stresses are given in Table 3.

Results

By measuring the resistance of the contracture of the KJ to external corrective action in 5 patients with RA, the average values of the angle of passive movement Δα, by which the position in the joint changes under the action of a dosed force F = 50 N, were obtained: in the direction of flexion — Δα = 3°, extension — Δα = 6º, namely:

– for changing the angle from the final position of the active movement to extension, under the action of a force F = 50 N, which acted at a distance l = 350 mm from the center of rotation in the joint, the passive movement was Δαp = 12°;

– for changing the angle from the final position of the active movement to flexion, under the action of a force F = 50 N, which acted at a distance l = 350 mm from the center of rotation in the joint, the passive movement was $\Delta\alpha_з$ = 10°.

On a solid 3D model using SolidWorks tools, the contact surface on the tibial plateau was measured (Fig. 5).

According to the radius of curvature of the external and internal epicondyle, the contact area on the internal condyle of the tibial plateau was 434 mm^2, while on the external one 255 mm^2. Both contact points in the position of 30° flexion of the KJ are slightly shifted posteriorly.

Thus, because of passive flexion in the KJ by 6°, the load on the condyles of the plateau increased by 12.8 % (Fig. 6).

The compressive strength limits in this case vary from 17.5 to 93.4 MPa, conservatively (according to the most dangerous option), we have adopted 17.5 MPa, then:

$$[\sigma]\min = R_m^T / n_m = 6{,}73 \text{ MPa}. \quad (7)$$

The strength condition for the model is fulfilled when [σ]/σmax ≥ n = 1.

Based on this, we obtain the minimum safety margin n_{min} = 6.93 / 5.02 = 1.38 for the position of passive flexion by 5° on the external condyle, which is approaching the critical one (Fig. 7).

Thus, as a result of passive extension in the KJ by 3°, the load on the condyles of the tibial plateau increased by 95.2 % (Fig. 8).

Based on this, we obtain the minimum safety margin nmin = 6.93 / 7.93 = 0.87 for the position to passive extension by 3° on the external condyle, which is critical for it and can lead to degradation of bone tissue in the contact area (Fig. 9–11).

Calculation of the load on the articular surface of the tibial plateau under the conditions of single-support standing under the action of a body weight of 60 kg (600 N) in conditions of flexion-extension contracture in the position of 30°

The total force acting on the knee joint has the following form:

$$\sum F_{сум} = \sqrt{F_x^2 + F_y^2 + F_z^2} = 2756 \text{ H}. \quad (8)$$

Table 3

Results of passive bending in the KJ with a force of 50 N

Bending angle, °	F_{tot} (H)	σ_i (MPa)	σ_e (MPa)
0	2072.36	2.3875	4.0635
1	2153.29	2.4807	4.2221
2	2241.50	2.5824	4.3951
3	2337.99	2.6935	4.5843
4	2443.93	2.8156	4.7920
5	2560.75	2.9502	5.0211
6	2337.99	2.6935	4.5843

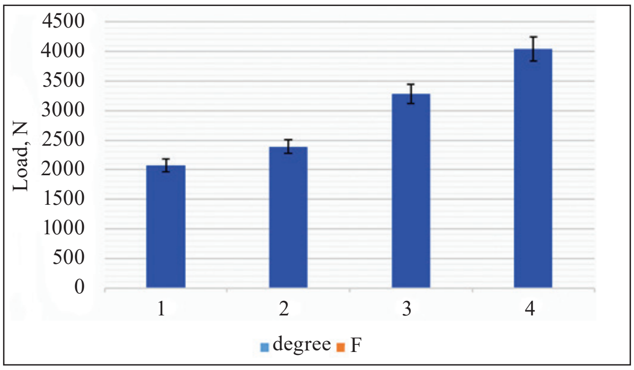


Fig. 8. Diagram of the load on the knee joint during passive extension by 3° taking into account standard deviations

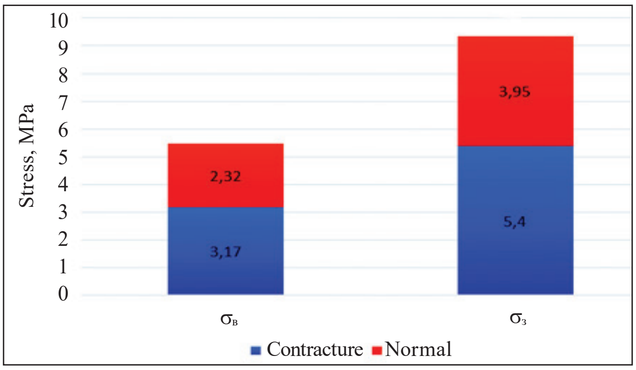


Fig. 12. Diagram of the comparative analysis of stress indicators during gait under conditions of KJ contracture and in the norm

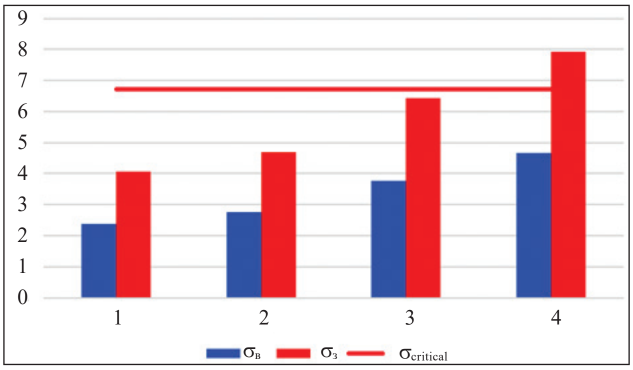
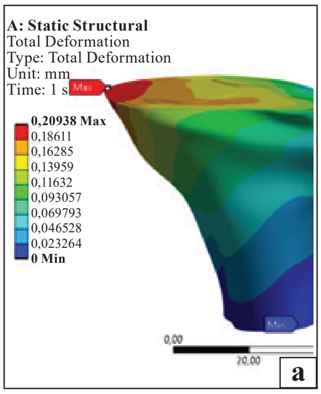
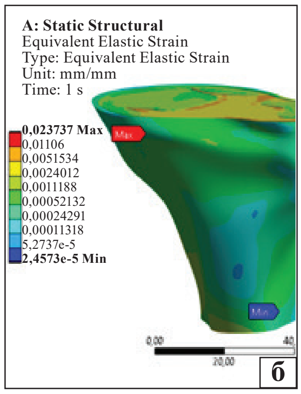


Fig. 9. Diagram of the stress on the contact surfaces of the tibial plateau during passive extension by 3°





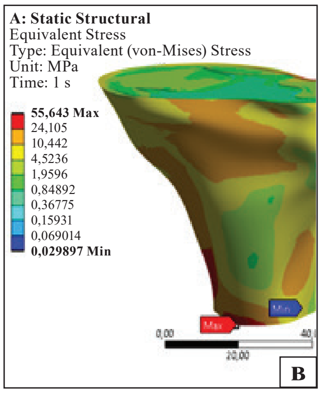


Fig. 13. Indicators of the SSS model: a) displacement (Δ_{max} = 0.209 mm), b) deformation (ε_{max} = 0.0237 mm), c) stress (σ_{max} = 55.64 MPa)

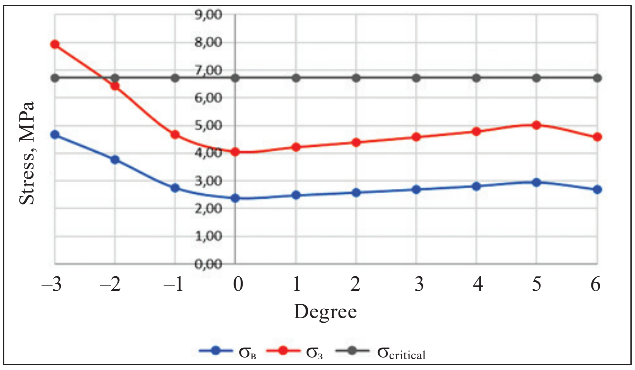


Fig. 10. Graph of the dependence of stresses on the angle in the KJ under conditions of flexion-extension contracture in the 30° position

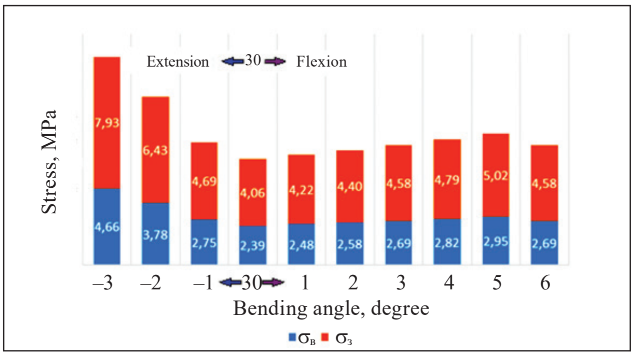


Fig. 11. Diagram of the dependence of stresses on the angle in the KJ under conditions of flexion-extension contracture in the 30° position

Taking into account the contact areas on the condyles of the tibial plateau, the stress on them is determined:

$$\sigma_в = F_в / S_в = 2{,}32 \text{ MPa} \tag{9}$$

$$\sigma_з = F_з / S_з = 3{,}95 \text{ MPa}, \tag{10}$$

where S_i = 434 mm^2 — the contact area of the internal condyle; S_e = 255 mm^2 — the contact area of the external condyle.

Comparative analysis of stress indicators during walking in conditions of flexion-extension contracture of the knee joint in the 30° position and in normal conditions is shown in Fig. 12.

As can be observed from the obtained results, in conditions of flexion-extension contracture of the knee joint in the 30° position, the stresses in the contact zone on the external condyle of the tibial plateau increase by 36.71 %, on the internal one by 36.64 %.

Load distribution on the proximal part of the tibia (mathematical modeling)

The SSS on the model as a whole (Fig. 13) and layer by layer on each slice were estimated by the indicators of displacements, deformations and stresses according to Misis.

Conclusions

A simulated solid-state 3D model of the knee joint has been developed, which demonstrates that in the knee joint flexion position of 30º, in accordance with the radius of curvature of the external and internal condyles, the contact area on the internal condyles of the tibial plateau is 2 times larger than that on the external and is shifted somewhat posteriorly.

During single-support standing under the action of a body weight of 60 kg (600 N) in conditions of flexion-extension contracture in the knee joint in the 30° position, an increase in contact stresses is observed on the external condyles of the tibial plateau by 36.71 %, on the internal by 36.64 %, which contributes to the progression of the clinical presentation of rheumatoid arthritis with an increase in the phenomena of osteoarthritis precisely in the posterior parts of the knee joint.

By measuring the indicators of the susceptibility of the knee joint contracture to external corrective action, it was established that in patients in the late stages of RA with persistent contractures in the knee joint, the average values of the angle of passive movement by which the position in the knee joint changes under the influence of a dosed force F = 50 N: in the direction of flexion 2 times greater than extension.

Based on biomechanical studies, it was established that due to forced passive flexion in the knee joint by 6°, the load on the condyles of the tibial plateau increased by 12.8%, with passive extension by 3° the load increased by 95.2%, which is a critical value and can lead to degradation of bone tissue in the contact area and increased manifestations of secondary arthrosis.

An increase in the proportion of osteoporotically altered bone tissue causes additional stresses on the entire structure of the subchondral area of the proximal tibia. The more uneven the load distribution on the knee joint, the greater the stresses will occur on the cortical layer, provoking pain and progression of deformation in this area.

The heterogeneous structure of the subchondral bone of the tibial plateau, which is confirmed by CT data of layer-by-layer automatic measurement of the tissue absorption coefficient, has a significant difference in bone density values and is a prerequisite for the formation of knee joint deformities.

Conflict of interest. The authors declare the absence of a conflict of interest.

References

1. Watson, R. S., Gouze, E., Levings, P. P., Bush, M. L., Kay, J. D., Jorgensen, M. S., Dacanay, E. A., Reith, J. W., Wright, T. W., & Ghivizzani, S. C. (2010). Gene delivery of TGF-β1 induces arthrofibrosis and chondrometaplasia of synovium *in vivo*. *Laboratory investigation, 90*(11), 1615–1627. https://doi.org/10.1038/labinvest.2010.145
2. Ouyang, X., Ghani, A., & Mehal, W. Z. (2013). Inflammasome biology in fibrogenesis. *Biochimica et Biophysica Acta (BBA) — molecular basis of disease, 1832*(7), 979–988. https://doi.org/10.1016/j.bbadis.2013.03.020
3. Canovas, F., & Dagneaux, L. (2018). Quality of life after total knee arthroplasty. *Orthopaedics & traumatology: surgery & research, 104*(1), S41–S46. https://doi.org/10.1016/j.otsr.2017.04.017
4. Oiestad, B., Juhl, C., Eitzen, I., & Thorlund, J. (2015). Knee extensor muscle weakness is a risk factor for development of knee osteoarthritis. A systematic review and meta-analysis. *Osteoarthritis and cartilage, 23*(2), 171–177. https://doi.org/10.1016/j.joca.2014.10.008
5. Herman, M. J., Martinek, M. A., & Abzug, J. M. (2014). Complications of tibial Eminence and Diaphyseal fractures in children. *Journal of the American academy of orthopaedic surgeons, 22*(11), 730–741. https://doi.org/10.5435/jaaos-22-11-730
6. Dell'Isola, A., Smith, S., Andersen, M., & Steultjens, M. (2017). Knee internal contact force in a Varus malaligned phenotype in knee osteoarthritis (KOA). *Osteoarthritis and cartilage, 25*(12), 2007–2013. https://doi.org/10.1016/j.joca.2017.08.010
7. J. George, M. (2020). Valgus deformity correction in total knee replacement: An overview. *Knee surgery — reconstruction and replacement.* https://doi.org/10.5772/intechopen.89739
8. Schiavone Panni, A., Cerciello, S., Vasso, M., & Tartarone, M. (2009). Stiffness in total knee arthroplasty. *Journal of Orthopaedics and Traumatology, 10*(3), 111–118. https://doi.org/10.1007/s10195-009-0054-6
9. Formby, P. M., Donohue, M. A., Cannova, C. J., & Caulfield, J. P. (2016). Hydraulic distension of the knee: A novel treatment for arthrofibrosis after total knee replacement (case series). *ANZ Journal of surgery, 86*(6), 480–482. https://doi.org/10.1111/ans.13540
10. Khatri, K., Bansal, D., & Rajpal, K. (2020). Management of flexion contracture in total knee arthroplasty. *Knee surgery — reconstruction and replacement.* https://doi.org/10.5772/intechopen.90417
11. Schiavone Panni, A., Cerciello, S., Vasso, M., & Tartarone, M. (2009). Stiffness in total knee arthroplasty. *Journal of orthopaedics and traumatology, 10*(3), 111–118. https://doi.org/10.1007/s10195-009-0054-6
12. Kukin, I. A Kirpichjov, I. V, Maslov, L. B, & Vihrev, S. V. (2013). Particularities of the strength characteristics of spongy bone in diseases of the hip joint. *Fundamental research*, (7), 328–333
13. Ipach, I., Mittag, F., Lahrmann, J., Kunze, B., & Kluba, T. (2011). Arthrofibrosis after TKA — Influence factors on the absolute flexion and gain in flexion after manipulation under anaesthesia. *BMC musculoskeletal disorders, 12*(1). https://

doi.org/10.1186/1471-2474-12-184
14. Kalson, N. S., Borthwick, L. A., Mann, D. A., Deehan, D. J., Lewis, P., Mann, C., Mont, M. A., Morgan-Jones, R., Oussedik, S., Williams, F. M., Toms, A., Argenson, J. N., Bellemans, J., Bhave, A., Furnes, O., Gollwitzer, H., Haddad, F. S., Hofmann, S., & Krenn, V. (2016). International consensus on the definition and classification of fibrosis of the knee joint. *The bone & joint journal, 98-B*(11), 1479–1488. https://doi.org/10.1302/0301-620x.98b10.37957
15. Babko, A. M., Herasymenko, A. S., & Mazevych, V. B. (2019). The mechanism of contracture formation in the hip and knee joints in early stages of rheumatoid arthritis Natural experiment. *Visnyk ortopedii travmatologii protezuvannia, 103*(4), 53–61. https://doi.org/10.37647/0132-2486-2019-103-4-53-61
16. Deshmukh, A. J., Rathod, P. A., Moses, M. J., Snir, N., Marwin, S. E., & Dayan, A. J. (2015). Does a non-stemmed constrained condylar prosthesis predispose to early failure of primary total knee arthroplasty? *Knee surgery, sports traumatology, arthroscopy, 24*(10), 3194–3199. https://doi.org/10.1007/s00167-014-3494-3.

The article has been sent to the editors 03.10.2024

BIOMECHANICAL PARTICULARITIES OF KNEE JOINT FLEXION DEFOMATION IN RHEUMATOID ARTHRITIS PATIENTS

S. I. Herasymenko, M. V. Poluliakh, A. M. Babko, A. S. Herasymenko, D. I. Kachan, D. M. Poluliakh

SI «Institute of Traumatology and Orthopedics of NAMS of Ukraine», Kyiv

✉ Sergiy Herasymenko, MD, Doctor in Traumatology and Orthopaedics: kievorto3@gmail.com

✉ Myhailo Poluliakh, MD, PhD, Prof.: orthoin.ua@gmail.com

✉ Andriy Babko, MD, PhD: Orthokiev@i.ua

✉ Andriy Herasymenko, MD, PhD in Orthopaedics and Traumatology: corado734@ukr.net

✉ Dmytro Kachan: d.kachanmd@gmail.com

✉ Dmytro Poluliakh, MD, PhD in Orthopaedics and Traumatology: dmpoluliakh@gmail.com