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REHABILITATION

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Smith machine squats pose high risk to ACL graft integrity after the ACL reconstruction and conventional squats are a safer alternative

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Abstract

Purpose: The purpose of this study was to evaluate the impact of squats after the anterior cruciate ligament (ACL) reconstruction on the ACL graft, considering new data on biomechanics, posterior tibial slope (PTS) and anterolateral ligament (ALL).

Methods: Utilising finite element analysis on the new 14-component knee joint model, we have evaluated stresses on the knee elements separately for the knee with a native double-bundle ACL and with a single-bundle ACL graft for the 5° and 14° PTS variants during both conventional and Smith machine horizontal squats.

Results: Replacing a native ACL with a single-bundle graft causes an overstrain on the graft compared to the intact ACL under all conditions. Stresses on the ACL, ACL graft and ALL are much higher during the Smith machine squats compared to the conventional ones. The stress on the menisci is 3.6–4.9 times higher with conventional squats. PTS at the squats' lowest point minimally affects ACL stress but impacts menisci.

Conclusions: The single-bundle ACL reconstruction (ACLR) does not reproduce the biomechanics of the native ACL and increases stresses in most knee joint elements, according to the current study. Conventional squats are relatively safe for the ACL graft at their lowest point. Passing the half-squat position is the most dangerous point. Smith machine horizontal squats produce stress on the ACL graft several times higher than its estimated breaking load and dangerous stress levels on the ALL. During the rehabilitation following ACLR, it is advisable to prioritise the conventional squats over Smith machine squats until ligamentisation is complete.

Level of Evidence: Level III.

KEYWORDS
ACL, ALL, finite element analysis, knee, squat

Abbreviations: ACL, anterior cruciate ligament; ACLR, anterior cruciate ligament reconstruction; ALL, anterolateral ligament; m., muscle; PCL, posterior cruciate ligament; PTS, posterior tibial slope.

INTRODUCTION

Significant advancements have been made in anterior cruciate ligament reconstruction (ACLR) since its initial attempts, and ongoing progress continues to enhance the procedure further [3]. Modern methods of ACLR may provide patients with greater stability and a prompt return to activities and sport [28]. Rehabilitation approaches have also evolved [34]. Nevertheless, some athletes fail to reach the preoperative level and face frequent repeated injuries of the anterior cruciate ligament (ACL) [29]. Knee laxity and bone tunnel widening after ACLR in accelerated weight-bearing rehabilitation programs also pose a problem [7]. Squats are widely utilised in rehabilitation as they are among the leading and most effective sports exercises for strength development and enhancing core stability. They are fundamental in the training programs of many sports, are a competitive exercise in powerlifting and are necessary for a number of everyday activities. Squats are especially desirable closed kinetic chain exercises for post-ACLR rehabilitation because they engage multiple large muscles within a single movement [2, 5, 6]. The stress on the anatomical structures of the knee joint varies depending on the type of squat and the technique employed [32]. After ACLR, highly motivated professional athletes often enquire whether they can change their squat technique to return to squats earlier.

The literature describes the possibility of even a healthy ACL injury during squatting [8]. An ACLR or refixation/reinsertion temporarily weakens this area for up to 24 months [27]; so, understanding the stresses on the ACL in different squatting exercises and techniques is essential for building effective and safe rehabilitation programs.

The standard for this exercise is the horizontal squat. However, in the case of a temporary suspension or restriction of traditional squats, injured athletes may look for alternative exercises that could on appearance be perceived as safer. One of these exercises is the Smith machine squat. However, these squats have been studied even less than the traditional ones [19]. The recent recognition of the significance of the PTS [35] and anterolateral ligament (ALL) [9] for ACL injuries challenges many previous biomechanical studies.

Extreme variability about the normal range of PTS has been found in the literature [26]. However, there is limited research available regarding the impact of high PTS angles on ACL and ACL graft loads and the stresses carried by the anterolateral ligament (ALL) during squats.

It was hypothesised that different types of squats cause varying stresses on the ACL graft, ALL and other anatomical structures of the knee joint and that high PTS angles may exacerbate these stresses, with potential implications for post-ACLR rehabilitation programs.

The understanding of these stresses may enhance post-ACLR rehabilitation programs and mitigate the risk of graft failure.

MATERIALS AND METHODS

The impact of conventional horizontal squats and Smith machine horizontal squats with varying PTS values on the ACL, ACL grafts and other knee joint elements was evaluated with the help of finite element analysis. This method has been chosen, recognising its progress, increasing use in orthopaedics [33] and accuracy comparable with a cadaver knee joint experiment (discrepancy less than 11 %) [30].

Conventional horizontal squats are both a rehabilitation exercise and a competitive one. So, the squat technique was determined by the Technical Rules Books of the largest and oldest International Powerlifting Federation. The squat is performed until the top surface of the legs at the hip joint is lower than the top of the knees. Therefore, we have chosen the horizontal plane as the reference level.

The neutral-zero method was utilised to set the angles in the model and to describe the angles in the knee and ankle joints later in the article. Considering research on the lower extremity angles during squats, we have chosen the 113° angle of flexion in the knee joint and the 23° angle of dorsal flexion (extension) in the ankle joint (Figure 1a) [14].

Since Smith machine squats are not a competitive exercise, there are no clear and generally accepted recommendations for their performance, and the activity may be practised in a wide range of variations. Less ankle dorsiflexion is commonly observed in Smith machine squats. Therefore, considering the literature [10, 14, 19], the angle of knee flexion in the Smith squat

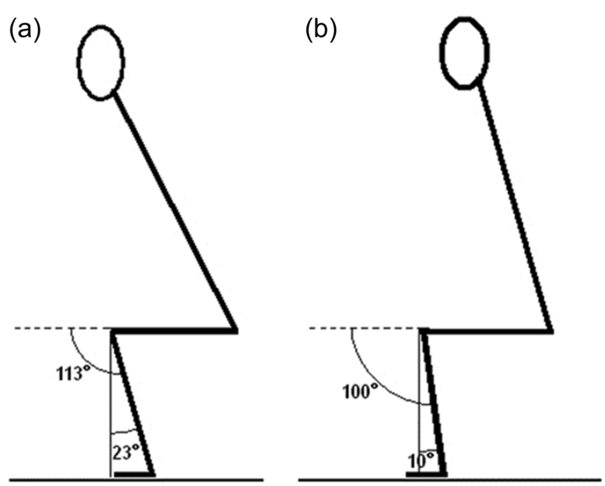

FIGURE 1 Schematic drawing of the angles of the lower extremity joints in conventional (a) and Smith machine (b) squats.

was taken as 100°, and the angle of dorsal flexion (extension) in the ankle was taken as 10°.

SolidWorks software was utilised to create a unique knee joint model consisting of 14 intricate, curved shape components, including the anterolateral ligament (ALL), frequently omitted in older models. It was constructed using CT images with a 0.5 mm interslice interval. To ensure accuracy, literature data [20, 22] for average size parameters of bones and menisci were consulted, cross-referencing the CT images to ensure they align with these averages.

Within each model, we conducted stress analysis on various anatomical elements, including the ACL, ACL graft, ALL, menisci, posterior cruciate ligament (PCL), medial and lateral collateral ligaments and cartilage of the femoral condyles. These calculations were carried out separately for two scenarios: the knee with a native 6 mm double-bundle ACL and the knee with an 8 mm single-bundle ACL graft.

Two different PTS variants, 5° and 14°, were considered. Our objective was to investigate the impact of PTS variations within normal limits on the stress experienced by the ACL and ACL graft. The 5° PTS angle represents the minimal average PTS based on the study by Medda et al. [25]. The 14° option was chosen on the basis of the research by Matas et al. on the upper limit of PTS [24] and by Mandalia et al. [23] suggesting the danger of PTS values exceeding 12°.

To evaluate the effects of exercise and loads on the intact ACL, we opted for a 6-mm diameter for the native double-bundle ACL because the average ACL diameter in patients with ACL injuries is 6.2 mm [21], and the average diameters of the ACL in the population range from 4.8 to 8.3 mm [4, 15]. The diameter of the ACL graft varies depending on the type used. An 8-mm diameter for the single-bundle ACL graft was selected, representing the minimum acceptable diameter for commonly used grafts such as the semitendinosus and gracilis tendons [1]. However, thinner grafts are occasionally utilised in practice.

The mechanical properties of the knee joint's anatomical elements, such as compact and cancellous bone, cartilage, menisci and ligaments, were assigned using Young's modulus and Poisson's ratio values based on relevant literature [17, 18]. Ansys software was employed to evaluate the stresses and strains on these structures under normal physiological loads during squats. A finite element grid was generated for all models, including those with ACL and ACL graft while considering different PTS angles. The grid consisted of 2 079 911 nodes and 1 377 642 elements, with additional refinement in contact areas to ensure more accurate calculations. Further grid refinement did not significantly impact the results according to the tests conducted. The average size of the elements was ≤1 mm, with a mean orthogonal quality of 0.85 and a mean skewness of 0.23. The relevant forces were utilised to replace the cut parts of the knee joint elements, and muscles were modelled using springs. The forces were applied at the spring's attachment points, aligned with the corresponding muscle direction, and based on tabulated values (Table 1) [11]. Using the constructed model, the total deformation, equivalent elastic deformation and equivalent stress in the anatomical elements of the knee joint were determined during horizontal squats with total weights of 75, 100, 125 and 150 kg.

TABLE 1 Muscle exertions while weight bearing in horizontal squats [11].

	Muscle exertions [N] at weight			
Muscles	**75 kg**	**100 kg**	**125 kg**	**150 kg**
m. rectus femoris	130.5	131.0	85.0	132.0
m. vastus lateralis	225.0	375.0	585.9	878.9
m. vastus medius	75.0	125.0	195.3	293.0
m. vastus intermedius	39.0	72.0	110.0	154.5
m. semitendinosus	1.50	3.0	5.00	10.5
m. semimembranosus	24.0	78.0	166.3	348.0
m. biceps femoris short head	6.0	7.0	8.8	10.5
m. biceps femoris long head	162.8	245.0	318.8	378.0
m. soleus	60.0	132.0	246.3	388.5
m. gastrocnemius lateralis	20.3	29.0	41.3	57.0
m. gastrocnemius medialis	41.3	56.0	87.5	129.0

RESULTS

The load distribution in the knee with the single-bundle ACL graft in 14° PTS at a body weight load (75 kg), while weight bearing in the Smith machine and conventional horizontal squats, is represented visually in Figure 2. The obtained numerical values of the stress in the ACL and other anatomical elements of the knee joint under the different conditions are summarised in Table 2.

Table 2 shows that during horizontal squats, both in the Smith machine and conventional squats, the stress on the ACL predictably increases proportionally to the increasing weight loading. Replacing a double-bundle ACL with a single-bundle graft causes an overstrain on the ACL graft compared to the intact ACL under all conditions studied. In the case of 5° PTS, this may lead to an ACL graft stress increase of up to 43% during squats in the Smith machine and up to 23% during conventional horizontal squats. The mathematical model demonstrates that single-bundle ACL reconstruction

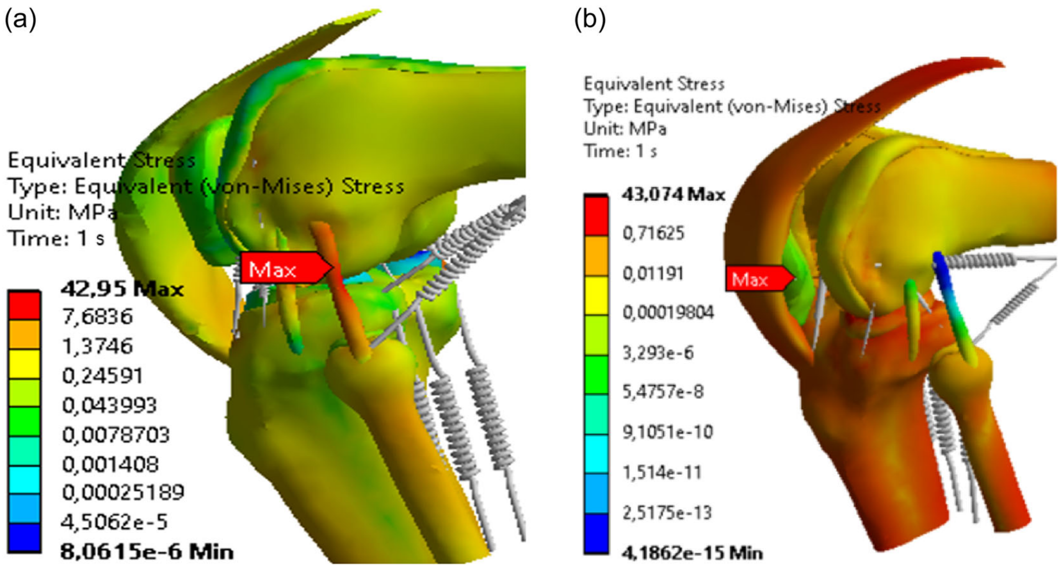

FIGURE 2 Equivalent (von Mises) stress distribution in the knee with the single-bundle ACL graft in PTS 14° at a body weight load (75 kg) while weight bearing in Smith machine horizontal squat (a) and conventional squats (b).

does not reproduce the same biomechanics as native double-bundle native ACL. To a greater or lesser extent, replacing a native ACL with a single-bundle graft increases stresses in most of the studied knee joint elements.

Stresses on the ACL, ACL graft and ALL are higher in the Smith machine squats compared to the conventional ones by 3.0–3.6 times on the ACL and by 3.5–4.5 times on the ACL graft depending on the PTS and by 98–149 times on the ALL depending on the state of the ACL and the PTS angle. However, increasing PTS from 5° to 14° in both studied types of horizontal squats has little effect on the stress of the ACL or its graft at the lowest point.

On the contrary, the situation is different for menisci. The stress on the meniscus is 3.6–4.9 times higher with traditional squats compared to the Smith machine, depending on the PTS and the state of ACL. PTS raising from 5° to 14° increases the stress on the menisci at the lowest point in both studied types of horizontal squats.

DISCUSSION

The most important finding of the present study was a relatively low stress on the ACL and ACL graft at the lowest point of conventional horizontal squats (less than in half squats). Nevertheless, reaching this point is only possible through the half squat position, which is risky for the ACL graft because of the stresses 2.3-4.6 times higher than standing with the knees extended [13].

For lower extremity muscle strengthening, the deeper squats are more effective than half squats [31]. Therefore, patients cleared for half squats post-ACL surgery can safely progress to deeper conventional squats.

In contrast, Smith machine horizontal squats pose significant stress on the ACL graft, especially with a more vertical tibial shaft position (46.7 MPa at a patient weight of 75 kg). It is 18.7 times higher than the stress on the ACL graft in the standing position with knee joints extended and 20.0 times higher than the stress on the intact ACL while standing [12].

The breaking load for the double-folded semitendinosus and gracilis tendon ACL autograft within the first weeks after surgery, according to our estimates, is at least 17.7 MPa (depending on the graft thickness) and 12.9 MPa 6 weeks after surgery due to graft degradation [13]. So, several times higher stresses on the ACL graft are induced by Smith machine horizontal squats compared to conventional horizontal squats, suggesting they should be avoided in ACLR rehabilitation until ligamentisation is complete.

Replacing a native double-bundle ACL with a single-bundle graft increased stress at the lowest point of both squat types studied. Additional weight load leads to a proportional increase in stress on all elements of the knee joint. Therefore, it is easy to calculate the stresses for heavier patients or under additional weight bearing and compare them with the strength characteristics of the ACL, ACL graft and other anatomical elements of the joint.

Considering available data from other studies on the influence of PTS on ACL stress and the risk of ACL damage [35] and taking into account the data on the increase in ACL stress with increasing PTS in the standing and semi-squatting positions [12, 13], some impact of PTS was expected on the ACL and ACL graft

TABLE 2 Stress in some knee joint elements while weight bearing in Smith machine and conventional horizontal squats depending on the load conditions, posterior tibial slope (PTS) and the state of the anterior cruciate ligament (ACL) (native double-bundle ACL or single-bundle ACL graft).

	Stress (MPa), while weight bearing in															
	Smith machine horizontal squat (thighs parallel to the horizontal, knee flexion angle 100°, tibial shaft forward tilt angle 10°)								Conventional horizontal squats (thighs parallel to the horizontal, knee flexion angle 113°, tibial shaft forward tilt angle 23°)							
	PTS 5°				PTS 14°				PTS 5°				PTS 14°			
The knee joint element	**75 kg**	**100 kg**	**125 kg**	**150 kg**	**75 kg**	**100 kg**	**125 kg**	**150 kg**	**75 kg**	**100 kg**	**125 kg**	**150 kg**	**75 kg**	**100 kg**	**125 kg**	**150 kg**
ACL double-bundle	32.6	42.7	49.6	63.0	30.7	39.9	49.6	59.3	9.0	12.0	15.0	18.0	9.6	13.2	15.7	18.9
Single-bundle ACL graft	46.7	57.6	68.5	82.0	43.0	52.4	61.9	73.0	10.3	14.6	15.7	22.1	10.4	15.0	16.9	21.6
ALL (with double-bundle ACL)	11.6	16.0	18.9	25.0	10.9	14.6	18.9	23.2	0.1	0.1	0.2	0.2	0.1	0.2	0.2	0.2
ALL (with single-bundle ACL graft)	15.9	20.6	25.3	30.8	14.3	18.4	22.5	27.0	0.1	0.2	0.2	0.2	0.1	0.2	0.2	0.2
Menisci (with double-bundle ACL)	8.1	11.6	16.0	18.6	8.6	12.3	16.0	19.6	37.6	50.1	62.6	75.1	41.9	55.5	65.9	79.2
Menisci (with single-bundle ACL graft)	11.0	14.8	18.5	22.8	11.3	15.2	19.1	23.2	42.5	60.4	65.9	91.2	43.1	62.3	70.2	89.9
Cartilage of the femoral condyles (with double-bundle ACL)	11.4	15.1	19.2	22.4	11.6	15.5	19.2	22.9	8.4	11.2	13.9	16.7	9.0	12.4	14.7	17.6
Cartilage of the femoral condyles (with single-bundle ACL graft)	14.8	18.5	22.2	26.7	15.0	18.8	22.6	27.3	9.4	13.3	14.7	20.1	9.5	13.7	15.5	19.8
Posterior cruciate ligament (with double-bundle ACL)	4.4	7.0	11.3	12.2	5.3	8.4	11.3	14.1	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Posterior cruciate ligament (with single-bundle ACL graft)	5.2	7.6	10.1	12.8	6.1	9.2	12.3	15.9	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Lateral (fibular) collateral ligament (with double-bundle ACL)	12.0	16.3	21.7	24.9	12.7	17.3	21.7	26.2	0.0	0.1	0.1	0.0	0.0	0.1	0.1	0.0
Lateral (fibular) collateral ligament (with single-bundle ACL graft)	15.2	19.5	23.8	28.9	16.0	20.6	25.2	30.8	0.0	0.1	0.1	0.0	0.0	0.1	0.1	0.0

TABLE 2 (Continued)

	Stress (MPa), while weight bearing in															
	Smith machine horizontal squat (thighs parallel to the horizontal, knee flexion angle 100°, tibial shaft forward tilt angle 10°)								Conventional horizontal squats (thighs parallel to the horizontal, knee flexion angle 113°, tibial shaft forward tilt angle 23°)							
	PTS 5°				PTS 14°				PTS 5°				PTS 14°			
The knee joint element	75 kg	100 kg	125 kg	150 kg	75 kg	100 kg	125 kg	150 kg	75 kg	100 kg	125 kg	150 kg	75 kg	100 kg	125 kg	150 kg
Medial (tibial) collateral ligament (with double-bundle ACL)	12.9	16.8	20.6	24.6	12.9	16.7	20.6	24.5	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Medial (tibial) collateral ligament (with single-bundle ACL graft)	17.2	21.2	25.2	30.2	17.0	20.9	24.8	29.5	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0

in the horizontal squat. However, this was not the case. PTS has minimal impact on ACL and ACL graft stress during horizontal squats. Ankle dorsiflexion angle greater than PTS may mitigate PTS-induced stress on the ACL graft and ALL. Therefore, to reduce the load on the ACL graft and ALL, we recommend slightly more ankle dorsiflexion to make the angle of the anterior tibial tilt greater than PTS.

This is the first time the stress on the ALL has been assessed during Smith machine horizontal squats.

The ALL breaking load is approximately 32 MPa [36]. The Smith machine horizontal squat may bring the load on the ALL to a dangerous level of 31 MPa for 150 kg weight. Traditional horizontal squats pose little stress on the ALL but carry risks due to the half-squat position.

Therefore, we recommend avoiding the Smith machine squats during rehabilitation after the ACLR until the ACL has regained its strength, favouring conventional horizontal squats, keeping in mind they pose the highest risk in the half-squat position.

Menisci stress is higher during traditional squats, especially with higher PTS, contrasting with lower ACL stress. This finding is relevant for meniscus suture rehabilitation, favouring Smith machine squats. However, similarly to the ACL, even in the case of the Smith machine squat, the half-squat is the most dangerous position when the meniscal stress exceeds the highest values in the deeper horizontal squat by at least several times [13].

Smith machine squats resulted in higher stresses in articular cartilage of the femoral condyles, posterior cruciate ligament and fibular and tibial collateral ligaments compared to traditional squats. However, they remained within safe limits for intact structures. Fluctuations in PTS within the studied values also had little effect on the resulting stress in these structures.

In addition to the evaluation of the stresses on the ACL and ACL graft during different variants of horizontal squats, the importance of the ALL and the loads it carries with different PTS angles was also demonstrated, thus providing additional rationale for its reconstruction in ACL injuries.

There are several limitations to this study, including the exclusive reliance on computer simulation, the absence of biomechanical data derived from cadaveric studies (including calculations for cyclic loads) and the corresponding support from pertinent clinical trial data, which constrain the current investigation. However, there are significant limitations of the cadaveric experiment too. Therefore, computer modelling is not inferior; rather, it represents an alternative approach of stress analysis compared to the cadaveric experiment.

The other limitation of our study is that only two variants of the PTS angles were included. It is still possible that the stress on the ACL, ALL and other joint elements will differ at various PTS angles, other anterior tibial tilt angles or their combinations.

Therefore, increased stress on the ACL and ALL is still possible at higher PTS angles, especially bigger than anterior tibial tilt ones.

Cartilage on the tibial plateau and the posterior surface of the patella was not included in the model, as its importance for the task was considered relatively minor [16]. Additionally, the role of the skin, subcutaneous fat, synovial membrane, synovial fluid and synovial bursae was not taken into account. Their role in this task is small, but their cumulative effect could be significant.

The stresses in different portions of the menisci were not studied separately, resulting in the inability to specify their most vulnerable portions and suture types.

CONCLUSIONS

Smith machine horizontal squats produce stress on the ACL graft several times higher than its estimated breaking load and dangerous stress levels on the ALL. Conventional squats are relatively safe for the ACL graft at their lowest point.

According to the current study, single-bundle ACL reconstruction does not reproduce the biomechanics of the native ACL and increases stresses in most knee joint elements.

An increase in PTS from 5° to 14° has a minor effect on the stress of the ACL, ACL graft and ALL at the lowest point of conventional and Smith machine horizontal squats.

AUTHOR CONTRIBUTIONS

All authors studied literature and planned the study, contributed to the study conception and design, analysis and interpretation of data studied to a greater or lesser extent (more details below). All authors approved the version to be published and agree to be accountable for all aspects of the work in ensuring that questions related to the accuracy or integrity of any part of the work are appropriately investigated and resolved. *Conceptualisation, supervision, protocol development, getting ethical approval*: Viktor Kotiuk and Oleksandr Kostrub. *Methodology*: Viktor Kotiuk and Tomasz Ziółek *Literature analysis and, on its basis, scientific substantiation of the numerous parameters selected for the study and the computer model; participation in substantiation and preparation of responses to reviewers*: Tomasz Ziółek, Volodymyr Podik, and Dmytro Smirnov. *Creation of a new knee joint computer model*: Volodymyr Podik and Dmytro Smirnov. *Verification of accuracy of the computer model, numerous adjustments to the model*: Viktor Kotiuk and Oleksandr Kostrub. *Computer simulation of squats and loads with stress analysis separately for different scenarios with the determination of the total deformation, equivalent elastic deformation and equivalent stress in the anatomical elements of the knee*: Roman Blonskyi and Viktor Kotiuk. *Data analysis*: Viktor Kotiuk, Tomasz Ziółek, and Roman Blonskyi. *Writing—original draft preparation*: Viktor Kotiuk. *Writing—review and editing*: Viktor Kotiuk and Tomasz Ziółek. The first draft of the manuscript was written by Viktor Kotiuk, and all authors commented on previous versions of the manuscript. All authors read and approved the final manuscript.

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CONFLICT OF INTEREST STATEMENT

The authors declare no conflict of interest.

DATA AVAILABILITY STATEMENT

Data are available on request from the authors. The data that support the findings of this study are available from the corresponding author upon reasonable request.

ETHICS STATEMENT

This study was performed in line with the principles of the Declaration of Helsinki. Approval was granted by the Ethics Committee of SI 'Institute of Traumatology and Orthopedics of NAMS of Ukraine', Kyiv, Ukraine with the study research protocol approval number 5 on 12 December 2019.

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