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# Ultrasound Experimental Approach for Knee Intra-Articular Femur-Tibia Movements at Different Loads

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**Abstract:** The purpose of the present study was to develop an experimental model for studying of the intra-articular knee movements depending on the work of the knee joint and applied muscle groups in isometric stretching conditions with different loads. The experimental procedure includes ultrasound examination of a knee joint after isometric stretching of healthy men (n=32). The changes in millimeters of the distances between the femur and tibia were measured with ultrasound sonographer at three stages. The first stage was performed on ten (n=10) healthy men at five different sitting and upright positions. In the second and third experimental model stages the lower limbs loading was on 22 participants. Our hypothesis which was confirmed, was that as a result of increasing extraloads on the participants back (2, 5, 10, 15, 17, 20 kg), a intra-articular decrease in the femur-tibia cartilage surface distance will be observed. The accuracy of the created experimental model was improved over its three stages from 30% to 9%. Quantitative model data can help to create a mathematical model for the mechanical effects during deformation of the knee joint bone cartilage, as well as to outlines the some future tasks: increasing loading weights; enlarge participant groups; comparison men and women; healthy and pathology comparison.

**Keywords:** knee joint biomechanics; ultrasound scanning; femur-tibia distance

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## 1. Introduction

The specific anatomy of each human joint determines the limits in which the quantitative parameters of the performed movements, their variety and characteristics change. Here it is necessary to list the main factors affecting the mobility of the joints, such as: the difference in the sizes of the contact cartilage surfaces; the volume and shape of the intra-articular cartilages; the peculiarities in the construction of the joint capsule; the amount and location of the ligaments strengthening the joints; the muscle groups involved (agonists, antagonists, fixators and neutralizers) in the joint moving; the presence of intra-articular bone synovial fluid and formations, etc.

Over 70% of the traumas that occur in the human musculoskeletal system are of joint origin, like in some specific categories of work, including sports, it is significantly higher [1]. From this point of view, the joint study of intra-articular bone movements in norm and pathology as a result of different loads, from inner and outher forces, can give some benefits as:

- finding optimal physical exercises related to individual sportsman fitness status;
- joint rehabilitation;
- modelling and design of artificial joints, supporting human movements;
- creation of general purpose technical devices (robots, mobile lever systems in lifting equipment, etc.);
- intra-articular biomechanical processes mathematical modeling;
- information for intra-articular cartiliage deformation at different loads.

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## 1.1. Mechanical response to physical loads

### 1.1.1. Composition and structure of cartilage

Stenular cartilage represents a layer of soft tissue, providing low friction and surface load, that covers the articulating bone surface in the synovial joint. It allows to establish basic biomechanical functions, such as resistance to wear, resistance to load and shock absorption. From a biomechanical point of view, these important functional characteristics are related to the multiphasic nature of the state [2]. From an engineering point of view, porous tissue is a porous, viscoelastic material, consisting of three main phases: 1) a rigid phase, which is composed primarily of a densely woven, tough collagenous (mainly type II) fibrous network (15–22% of wet weight). , covered with proteoglycan macromolecules (4–7% of wet weight). The mesh of collagen and proteoglycans represents the pores, reinforced with fibers forming a rigid matrix. The interstices on this porous rigid matrix are filled with water and dissolved ions. The "average" size on the pores is approximately 60 angstroms; 2) wet phase, which is water (usually 80% of wet weight); and 3) ionic phase, which has many types of ions – dissolved electrolytes with positive and negative charges [2,3]. These three phases react together to strengthen the tissue, which is remarkable in its ability to withstand enormous load pressure (several times that of body weight), and the high shear stresses associated with them. It is reported that the stress pressure reaches up to 20 MPa in the overburdened structure [2]. The bone cartilage ability to withstand such high compressive loads without being torn is due to the multiphasic nature of the cartilage tissue and the unique combination of related properties of the cartilage material [4].

### 1.1.2. Mechanical cartilage response under physical loading with different profiles

Several authors considered cartilage as a viscoelastic material [3,5,6]. Hayes and Mockros [5] on the basis of generalized Kelvin solid evaluated shear and bulk creep compliances of human articular cartilage from independent creep tests in torsion and uniaxial strain. Linearity of the compliance coefficients in the loading range tested indicated that the results are applicable to visco- elastic analyses of synovial joint mechanics. The measured compliances for normal and degenerative tissue are compared and found to differ significantly. Preliminary investigations also suggest that flow processes are not important in the initial stages of the deformation of normal tissue. Some authors considered cartilage to be porousviscoelastic material. For instance, Wouters et al., 2015 [6] applied Burger's model and reported that the data revealed nonlinear relations between the applied force and the resulting deformation, with time and frequency dependence.

The cartilage, like many fine connective tissues, functions mechanically across a wide range of daily frequency loads, from <1 Hz for slow activities such as walking, to 1000 Hz for high-speed activities such as jumping and impact sports [7]. Poroelasticity is known to be the basic mechanism underlying the mechanical cartilage functions. Poroelasticity is manifested by friction between the synovial fluid and the hardened matrix of the cartilage and when the synovial fluid is injected under pressure into the cartilage tissue. These two elements of poroelasticity are at the basis of important mechanical cartilage functions, such as self-reinforcement, energy dissipation and hydraulic permeability [7].

Nia et al., 2013 estimated the high frequency cartilage nanomechanics on normal cartilage and a cartilage with denatured glycosaminoglycans (GAG) [7]. The last is due to the degradation of the cartilage matrix, which occurs in the earliest stages of osteoarthritis. In this research, high-frequency measurements that simulate velocity under load during activities such as running and jumping are made possible by developing a combined high-frequency nanorheological system coupled to an atomic-force microscope. Through this system the authors investigate the interaction between synovial fluid and the stiffness of the cartilage matrix at the molecular level, which occurs mainly at high frequency loads. The authors get some important results. The first is that GAG chains play a major role in the permeability of the synovial fluid, which is known to protect the tissue from wear and tear during activities with high load. Secondly, GAG depletion occurs in the early stages is more vulnerable to high-frequency and rapid loading than to high-force loading. These derivatives have a

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direct relationship with the influence of the frequency and speed of loading on the endurance of the muscles in certain sports disciplines [7].

*1.2. Methods for joint biomechanical characterization*

The wide penetration of the test of nuclear-magnetic resonance (NMR) in scientific research made it possible to evaluate some processes in the joint capsule *in vivo* under fixed loads. For example, Cotofana et al., found that cartilage thickness decreased to 5.2% when loading the knee with a force equal to 50% of body weight [8]. Herberthold et al. with in situ measurement of articular cartilaged compression in intact femoro-patellar joints loading observed a mean in situ deformation of 44% in patellar cartilage after 3.5 h of loading (mean contact pressure 3,6 MPa) [9]. As well as they make the conclusion that the femoral cartilage showing a smaller amount of deformation than the patella. They determine for the first time and the amount of fluid flow from the crust into the capsular cavity. In the last research, however, the load is on the foot and does not involve the participation of muscle groups.

In an other review [10] it was found that physical activity (sliding on stairs, running and climbing) leads to mild deformation of the articular cartilage, which recovers after 90 minutes. In an other research type Kubo's group studies [11,12] show that isometric exercises increase up to 7% of muscle group volume as well as their elasticity and Young's modulus.

Recent research [13] has been reported on changes in the volume of the knee joint capsule with isometric stretching. These data convincingly show that during active isometric exercise there is a change in the distance between the articular surface of the large bone in the femur, which is associated with a change in the size of the muscle-tendon system of the adjacent muscle group as a result of its contraction. Additionally, Ranchev et al. [1] performed preliminary studies on the change in length of the kinematic chain of the upper extremity during isometric stretching on a group of 10 men and 10 women. The results show that the increase in chain length reaches up to 40 mm in some individuals.

So it can be established what processes take place inside the joint capsule – the size change of the contact surface of the participating bone, a movement of the synovial fluid, bone cartilage deformation etc. [14].

It is also necessary to know the specific force (in newtons), which acts in the joint. These are muscular forces and joint reactions. The magnitude and direction of the erect response depend greatly on the position of the extremity and on the acting muscle force. They cannot be calculated directly, because the numbers on the unknown muscle force and reaction at the joint are much larger than the equations for equilibrium that can be written down. Therefore, in most cases modeling and optimization methods are used, choosing the optimization function according to the preferences of the researcher or from some physiological considerations [15]. For example, different mathematical and biomechanical models for knee joint muscle joint inclusion were obtained [16–21]. For verification of modeled results in the some of enumerated papers surface electromyographic signals (EMGs) are often used [22]. The direct relationship between these signals and the developing force is very complex and varies with different motor tasks. Nevertheless, some authors use processed EMGs as the input excitation signal [23,24] to the so-called Hill-type model of the muscle. Also unclear and discussed is the question how to process EMG signals - they are low-amplitude, noise-like, depending on many individual characteristics of the person being examined - gender, age, training, skin resistance, electrode placement, and others [25].

From what has been said so far, some effects of isometric muscle work and stretching on the muscle-tendon complex can be seen. However, the influence of isometric muscle work and stretching on the functions and biomechanics of joints and the processes within them is very poorly studied. In other words, what happens inside the joint during the intra-joint bone movements in norm and pathology, as well as the effects of different loads (from inner and outer forces) on the joint elements kinematics and why, as far as we know, has not been investigated and clarified.

Ultrasound Echography (ultrasonography, sonorography) is an aproach often applied for diagnostic of different muscle, tendon and joint injures [26–28]. Ultrasound method for the lower

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limbs joint visualisation is important in medicine, kinesitherapy and biomechanics from diagnostic, therapeutic and prophylactic standpoint [29–34]. The easy work with ultrasound and lack of ionizing radiation are benefits in comparisson with a computer tomography (CT) and fuoroscopy [35].

The aim of the present study was to create an experimental *in vivo* ultrasonographycal model for evaluation the change in the distance between the bony cartilaginous surfaces of the tibia and femur inside the knee capsule under different vertical external loads with isometric muscle work of applied muscle groups.

## 2. Materials and Methods

The knee joint is complex joint and is the commonly injured joint now a day because of increased vehicular trauma and sports related injuries. Being frequently loaded joint and with relative big size, more exposed to external forces, this joint easily gets injured [36]. An intraarticular fracture is a bone fracture in which the break crosses into the surface of a bone cartilage. Intraarticular fractures ideally should be reduced anatomically and fixed securely so that early joint movement can be allowed. Complex kinematics of knee weight bearing position and complex ligamentous stability and articular congruency are the main reason why these fractures are of concern to surgeon and cause disability to the patients [36].

The presented experimental model includes an ultrasound examination on the change in the distance between the bony surfaces of the tibia and femur inside the knee capsule under different inner and vertical external loads with isometric muscle work of applied muscle groups of 32 healthy men (n=32). Whole group age was between 19 and 24 years (Table 1). The subjects were athletic students from the National Sports Academy “Vassil Levski”, Sofia, Bulgaria. They did not report any health problems. They completed an injury record and were informed in detail about the aim of the experiments and the procedure. All participants gave informed consent. The experimental procedure was approved by the Scientific Council of the Institute of Biophysics and Biomedical Engineering, Sofia, Bulgaria. For some of the participants, the ultrasound scans were with low quality because of knee movement and/or ultrasound transducer position changes around the knee joint centre. Therefore, these data were excluded from Table 1 and from further statistical analyses of bone-to-bone distance.

The changes in millimeters of the distances between the femur and tibia were measured with portable ultrasound system Vinno 6, China, 8 - 10 MHz transducer frequency, musculo-sceletal and thyroid test mode (good intraarticular knee visualization). The RadiAnt DICOM Viewer 2022.1.1 was used in obtaining the distances in millimeters. The statistical analysis was conduct with Sigma Plot 10. All ultrasound scanning was made from the same medical physicist.

For all experiments the ultrasound transducer was laterally placed outside the right knee joint with long axis coaxially oriented with femur-tibia line (Figure 1). The transduser vision field was focused between iliotibial band and long head biceps femoris tendon on tibia (Figure 1, Figure 2, Figure 3, Figure 4).

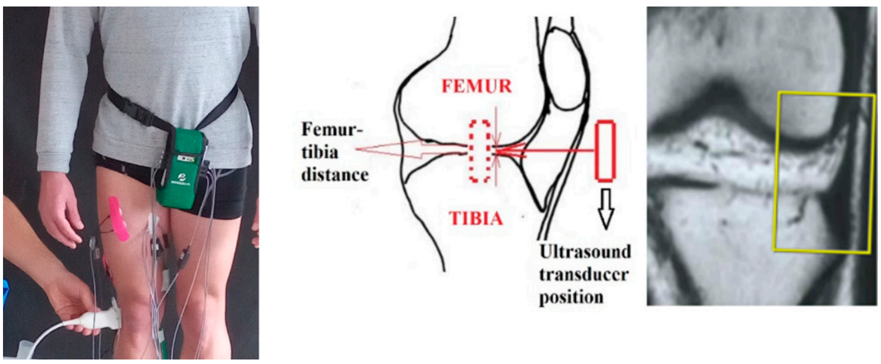


**Figure 1.** Ultrasound tranducer position to the knee joint.

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The first stage of ultrasound testing was performed on ten (n=10) healthy men at two different sitting positions and at three different upright positions (Figure 2) - 5 different lower limb pose to achieve different knee joint response:

1. at rest, femur-tibia angle 140˚, leg on the floor;
2. own weight stretched, femur-tibia angle 140˚, leg in the air;
3. straight, femur-tibia angle 140˚, leg without extraload;
4. straight, femur-tibia angle 140˚, leg with 4 kg extraload;
5. straight, femur-tibia angle 140˚, leg with 8 kg extraload.

The second stage was performed on fifteen (n=15) healthy men (Figure 3), and the third stage – on seven (n=7) healthy men (Figure 4).

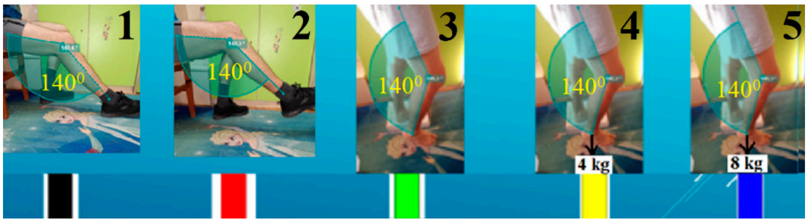



**Figure 2.** Stage 1 experimental setup with used lower limb poses with loads.

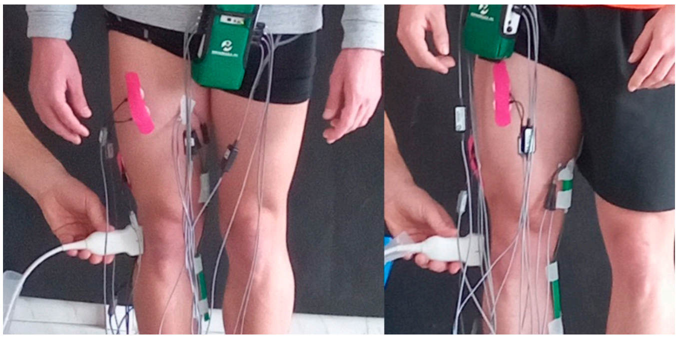


**Figure 3.** Experimental setup in stage 2 [37].

**Figure 4.** Experimental setup in stage 3 with ultrasound transducer muff for immobility of the transducer regarding the knee joint.

## 3. Results and discussion

The model development was carried out throw three stages. The first stage was performed on ten (n=10) healthy men at two different sitting positions and at three different upright positions – all with a knee angle between femur and tibia of 140 degrees (Figure 2) for better femur-tibia distance visualisation. In two of the three upright positions, extra loads of 4 and 8 kg were applied vertically down to the lower right limb to induce isometric stretching, by footpack (Figure 2). Three quantitative parameters - distance up ($D_{up}$, femur-tibia distance nearest to measurement transduser surface), distance down ($D_{down}$, femur-tibia distance in deapth to measurement transduser surface), and an area (A) from ultrasound pictures were introduced (Figure 5). The defined two displacements $D_{up}$ and $D_{down}$ were separated with 2,5 mm (Figure 5). The parameter Dup was measured for every participant in depth of knee joint space interval (D, mm) showen in column 7 at Table 1.

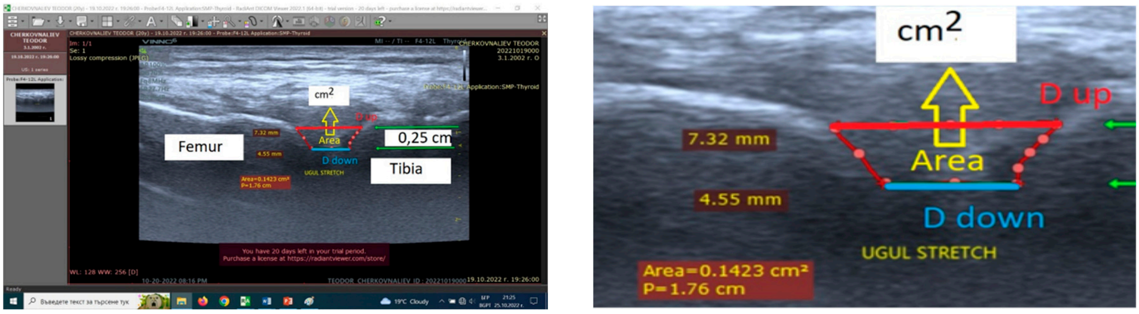
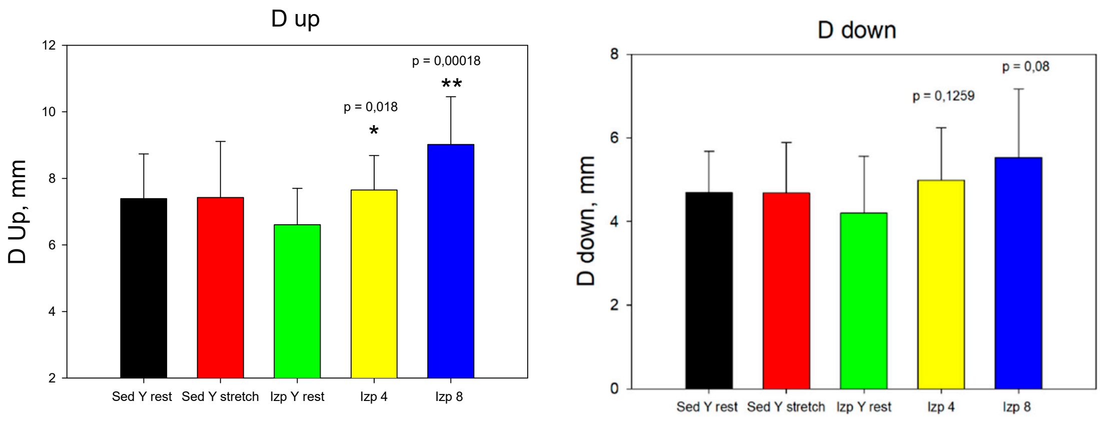



**Figure 5.** The screen view of the echograph VINNO 6 with measured distances between the femur and tibia bones in the knee joint for one participant [37].

The results from stage 1 (Figure 6) show that applying extra loads statistical significantly increase Dup at 4 and 8 kg, $D_{down}$ only at 8 kg and A only at 8 kg. The obtained results for the change of the intra-articular geometry under loads can serve as a quantitative assessment of the internal joint kinematics and determination of the individual joint mobility of the participants in the experiment [1,37,38].

The results obtained in stage one showed that the undetermination (accuracy or error in %) in obtaining the distances in millimeters between the femur and the tibia in the knee joint is high. The reason for this was mainly the unaccuracy at ultrasound transducer positioning about the knee joint for the same participant, despite of the used marker outline for ultrasound transducer on the knee joint skin. The authors concluded that the reproducibility of the obtained ultrasound pictures was poor. For this reason, it was decided to minimize the undetermination (inaccuracy) by testing the same distance in the knee joint, but in a straight posture with increasing loads on the back (Figure 3) to induce a lower limb shortenning. Some of the results of this second stage have been published yet [37–40].




(a) (b)

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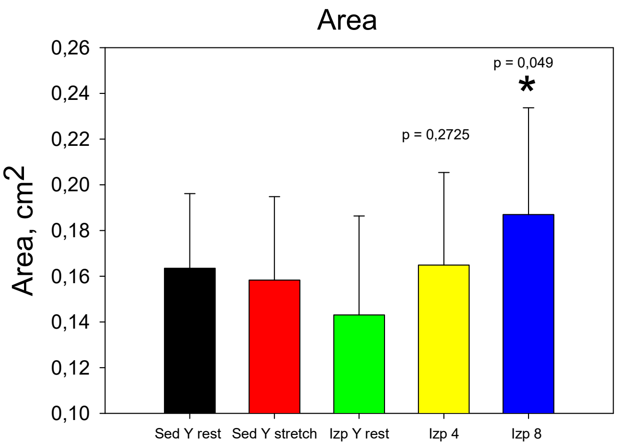


(c)

**Figure 6.** Data comparison for the first stage: a). Dup; b). Ddown; c). Area.

The second stage was performed on fifteen (n=15) healthy men at stright upright body position with increasing of loads on the back – 0, 2, 5, 10, 15, 17 and 20 kg [37](Figure 3).

Our hypothesis which was confirmed, was that as a result of increasing extraloads on the participants back (2, 5, 10, 15, 17, 20 kg), a intra-articular decrease in the femur-tibia cartilage surface distance will observed during static straight participant pose. This change in the distance between the bones indicates the presence of intra-articular processes, which in turn depends on the factors as:

- the extraload levels;
- the biomechanical properties of the knee joint components – femur, tibia, fibula and patela cartilage defformability, knee joint ligaments and tendons viscoelastisity, knowing that they are individual for each person and depend on many other factors (age, gender, height, level of training, etc.);
- amount and viscosity of synovial fluid;
- the lower limp pose when is extraloading;
- age, sex, weight.

Regression values for femur-tibia distances D for the seven tested participants were determined from the individual linear regression equations for all subjects. All collected data fell within the 95% confidence interval around the regression line. The relative reduction in femur-tibia distances in percentage was calculated based on the obtained regression model with individual angle coeficient (Figure 7, Table 1).

The third stage was carried out to further model optimization by increasing its accuracy. For this purpose it was made ultrasound transducer muff for immobility of the transducer regarding the knee joint (Figure 4). The results obtained in the third stage were shown in Table 1 (N26-N32). The mean percent decrease for this seven participants (column % decrease in Table 1) is smaller than the others N1-N25. This is indirect measure for the aimed model optimization.

**Table 1.** Results.

| Participants | | | | | | D, mm | 0 kg | 2 kg | 5 kg | 10 kg | 15 kg | 17 kg | 20 kg | Angle coeficient | %, decrease | Stage |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Number | Age | Height, cm | Weight, kg | BMI (kg/m²) | BSA (m²) | | | | | | | | | | | |
| N11 | 20 | 175 | 70 | 22,86 | 1,85 | 1 | 1,5 | 1,58 | 1,3 | 1,39 | 1,26 | 1,23 | 1,36 | -0,0117 | 15,74 | |
| N12 | 22 | 171 | 70 | 23,94 | 1,82 | 1 | 1,65 | 1,57 | 1,6 | 1,55 | 1,46 | 1,7 | 1,37 | -0,0072 | 8,81 | |
| N13 | 21 | 176 | 77 | 24,86 | 1,93 | 1 | 1,28 | 1,24 | 1,14 | 1,13 | 1,09 | 1,06 | 1,02 | -0,0116 | 18,55 | |
| **N14** | **22** | **188** | **103** | **29,14** | **2,29** | **1,5** | **2,21** | **1,9** | **2,05** | **2,06** | **1,53** | **1,46** | **1,6** | **-0,0327** | **30,35** | II |
| N15 | 20 | 183 | 76 | 22,69 | 1,98 | 1 | 1,45 | 1,02 | 1,09 | 1,16 | 0,996 | 1,03 | 1,05 | -0,0119 | 19,18 | |
| N16 | 20 | 181 | 86 | 26,25 | 2,07 | | 2,05 | 2 | 1,85 | | 1,8 | 1,78 | 1,71 | -0,0147 | 14,61 | |
| N17 | 20 | 183 | 90 | 26,87 | 2,12 | 1 | 1,78 | 1,54 | 1,52 | 1,47 | 1,62 | 1,71 | 1,49 | -0,0032 | 3,94 | |

| | | | | | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| N18 | 19 | | | | | 1 | 0,895 | 0,651 | | 0,673 | 0,456 | 0,439 | | -0,022 | 53,92 | |
| N19 | 19 | 170 | 78 | 26,99 | 1,89 | 1 | 1,41 | 1,36 | 1,4 | 1,32 | 1,25 | 1,19 | 1,17 | -0,0121 | 17,08 | |
| N20 | 20 | 177 | 96 | 30,64 | 2,13 | 1,25 | 1,66 | 1,58 | 1,51 | 1,47 | 1,57 | 1,38 | 1,37 | -0,0112 | 13,89 | |
| N21 | 22 | 175 | 64 | 20,90 | 1,78 | 0,75 | 1,62 | 1,44 | 1,51 | 1,52 | | 1,44 | | -0,0094 | 11,76 | |
| N22 | 20 | 190 | 80 | 22,16 | 2,08 | 0,9 | 1,80 | | 1,71 | 1,57 | 1,55 | 1,56 | 1,54 | -0,0132 | 14,9 | |
| N23 | 20 | 185 | 90 | 26,30 | 2,14 | 1 | 2,13 | 1,87 | 2,05 | 1,75 | 1,65 | 1,71 | 1,52 | -0,0256 | 24,78 | |
| N24 | | 191 | 85 | 23,30 | 2,14 | 1 | 1,84 | | | | | | 1,39 | -0,0225 | 24,46 | |
| N25 | 21 | 167 | 70 | 25,10 | 1,79 | 1,25 | 1,93 | 1,82 | | 1,76 | | 1,54 | 1,41 | -0,0233 | 24,27 | |
| N26 | 22 | 184 | 73 | 21,56 | 1,95 | 0,875 | 1,09 | 1,08 | 1,07 | 1,03 | 0,979 | | 0,99 | -0,0059 | 10,737 | |
| N27 | 20 | 184 | 69 | 20,38 | 1,90 | 1,5 | 0,595 | | | | | | 0,489 | -0,0053 | 17,81 | |
| N28 | 21 | 183 | 86 | | | 1 | 1,25 | 1,22 | 1,2 | 1,25 | 1,21 | | 1,18 | -0,0022 | 3,54 | |
| N29 | 24 | 185 | 79 | 23,08 | 2,03 | 1 | 1,86 | 1,8 | 1,77 | 1,75 | 1,77 | | | -0,0052 | 5,7 | III |
| N30 | 20 | 193 | 79 | 21,21 | 2,09 | 1,125 | 1,84 | 1,8 | 1,89 | 1,66 | 1,61 | | 1,61 | -0,0139 | 15 | |
| N31 | 24 | 178 | 80 | 25,25 | 1,98 | 1 | 1,19 | 1,18 | 1,14 | 1,05 | 1,07 | | 1,08 | -0,0064 | 10,92 | |
| N32 | 19 | 187 | 74 | 21,16 | 1,99 | 1,5 | 0,648 | | 0,656 | 0,629 | 0,619 | | | -0,0684 | 6,87 | |

The obtained results through the three stages for the change of the intra-articular geometry under load and stretching serve as a quantitative assessment of the internal joint kinematics and determination of the individual joint mobility of the participants in the experiment [1,37,38].

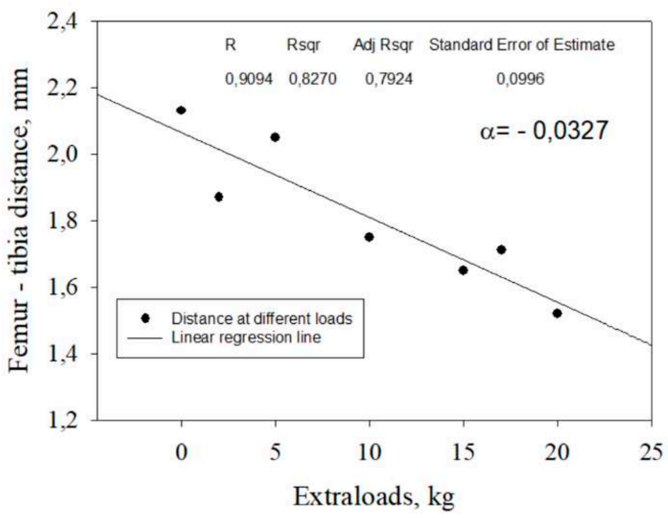


**Figure 7.** Femur – tibia distance versus extraloads for the participant N14 with the highest femur – tibia distance decreasing – 30,35% and the highest weight 103 kg.

The accuracy of the our measurements in the proposed experimental model by used device VINNO 6 is limited by three components. The first is related with the used transducer accuracy characteristics. The second is dependent by the accuracy in identity of the transducer-knee joint image position reproduction. The third is determined by the researcher skill at pictures scan, treatment and obtaining the distances in millimeters.

The first accuracy component is lower than 5 % and is defined and described in Vinno 6 user manual for our concrete transducer type (F4-12L) and used experimental mode. This accuracy level is the same for all three stages (Table 2).

The second component accuracy was different for the model stages. For the first stage this second component reaches to 20 %. In the second stage this component decreased to 7 %, because the identity of the transducer-knee joint image position reproduction is strictly observing and the ultrasound transducer position in relation to the knee joint at each participant is maximal stationary.

In the third stage the model accuracy decreased to 2%. The third accuracy component is minimized to 2% by the fact that all ultrasound pictures scan were made by ultrasound transducer muff for immobility of the transducer regarding the knee joint.

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At the moment the present experimental model accuracy is defined as a sum of three described components and is lower than 9 % [Table 2].

**Table 2.** Experimental model acuracy.

| Stage \ Component | 1 | 2 | 3 | Total |
|---|---|---|---|---|
| Stage 1 | ≤ 5% | ≤ 20% | ≤ 5% | ≤ 30% |
| Stage 2 | ≤ 5% | ≤ 7% | ≤ 5% | ≤ 17% |
| Stage 3 | ≤ 5% | ≤ 2% | ≤ 2% | ≤ 9% |

## 4. Conclusion

To the best of our knowledge, our experimental model is the first for investigation the reduction of the distance between the femur and the tibia in the knee joint at 0° flexion with increased load. These results will be the basis for the following: (1) determination of the change in the contact area between the femur and the tibia under different axial loads; (2) evaluation of the deformation of the cartilage tissue from the contact area between the femur and the tibia under different axial loads; (3) modelling the interaction between cartilage deformation and interstitial fluid flow from the cartilage into the joint cavity under loading conditions. The difficulties of the method for accurate measurement of the femur-tibia distance are related to ensuring the immobility of the ultrasound transducer regarding the knee joint and second, minimizing small movements in the knee of the tested participants

In the world literature there are an information that the contact surface between the femur and the tibia varies between 2 $cm^2$ and 6 $cm^2$ and it was influenced by the menisci [41,42]. Some authors concluded that the menisci may occupied 70 per cent of the total femur-tibia contact area [43].

The obtained quantitative data for femur-tibia distances combined with the femur-tibia contact surface area, will help to create in a future a mathematical model for the mechanical effects during deformation of the knee joint femur and tibia cartilages, as well as to attempt to prepare quantitative method with software program for automatic calculation of femur-tibia kinematics from ultrasound images.

On the basis of our experimental model can outlines some future tasks:

increasing loading weight;

enlarge participant groups;

comparison men and women; study of cartilage deformation at stretching loading;

looking for isometric stretching influence on knee hemorheology [44];

development/design of exercises in order to divide and estimate the contribution to the load of the joint cavity separately from stretching only and in combination with other loads;

development of practically applicable mechanical trainer for joint fitness.

**Author Contributions:** "Conceptualization, I.I.; Methodology, I.I.; Software, I.I.; Validation, I.I.; Formal Analysis, I.I.; Investigation, I.I.; Resources, I.I. and S.R.; Data Curation, I.I.; Writing – Original Draft Preparation, I.I.; Writing – Review & Editing, I.I.; Visualization, I.I.; Supervision, I.I.; Project Administration, I.I.; Funding Acquisition, I.I. and S.R.".

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