Biomechanics of distal femoral fracture fixed with an angular stable LISS plate

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Fractures of the distal end of the femur are infrequent and constitute less than 1% of all fractures. Only 3% to 6% of femoral fractures occur at the distal end. The two groups most at risk of the said fractures are young men and older women. The aim of treatment of fractures of the distal femur is to restore normal function of the knee joint. The authors asked themselves whether, following fixation of a 33-C2 fracture (according to the AO classification) with a LISS plate, a rehabilitation program can be undertaken immediately after surgery with the implementation of active movements in the knee joint of the operated limb. In order to answer this question, we created a digital model of a fractured femur fixed with the LISS method. The model was subjected to loads corresponding to the loads generated during active lifting of a limb extended in the knee joint and during flexing of a limb in the knee joint to the 90° angle. Interfragmentary movement (IFM) is one of the key parameters taken into account in the treatment of bone fractures. It allows classification of the treatment in terms of its quality both from the mechanical and histological points of view. We analyzed interfragmentary movement in all fracture gaps. The largest recorded displacement reached in our model was 243 μm, which, in the light of the literature data, should not interfere with bone consolidation, and thus implementation of active movement in the operated knee joint (keeping in mind the simplifications of the experimental method used) is possible in the early postoperative period.

Key words: *femoral bone, distal femur fractures, finite element method*

1. Introduction

Fractures of the distal end of the femur are infrequent and constitute less than 1% of all fractures. Only 3% to 6% of fractures of the femur are fractures of its distal end [1]–[3]. Such fractures usually occur as a result of two mechanisms of injury. One of them is a high-energy trauma, which most often occurs during a traffic accident or after a fall from a height. The second mechanism is a low-energy trauma due to a fall from one's own height, occurring in elderly people with osteoporotic bone [4]. The elderly are also often affected by periprosthetic fractures after previous knee joint replacement. Statistically, the groups most at risk of the said fractures are young men and older women [2], [5].

The distal end of the femur consists of the supra- and intercondylar region and the articular surface of the femur [2]. The articular surface of the femur is valgus to the shaft of the bone. Physiological valgus amounts to approx. 6–7°. Depending on the direction and magnitude of the force acting during the trauma, different types of fractures occur in the distal end of the femur.

There are different classifications of fractures of the distal end of the femur. Recently, however, the most important role has been played by the AO classi-

fication [6]. Fractures are divided into three groups: (A) extraarticular, (B) partial articular, and (C) complete articular. Each group is divided into subgroups. The subgroups 1 to 3 in the groups A and C classify the fractures from simple to multifragmentary. The B1 fracture in group B corresponds to the fracture of the lateral condyle of the femur, the B2 fracture corresponds to the fracture of the medial condyle, and the B3 fracture concerns condylar fractures in the coronal plane.
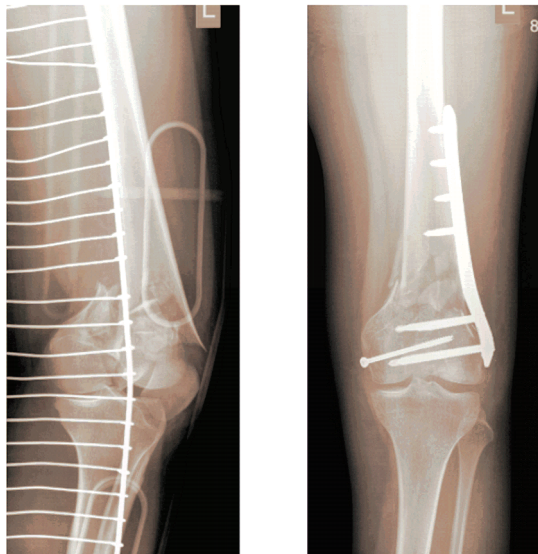

Fig. 1. The C2 fracture and its stabilization
with the LISS technique

The main aim of treatment of fractures of the distal end of the femur is to restore normal function of the knee joint. This can only be achieved if the articular surface is properly restored and the fracture fragments are fixed in a way that ensures bone consolidation.

An important factor considered in the process of fracture healing is the movement taking place between bone fragments. In 1979, Perren proposed a theory of interfragmentary displacements according to which the process of fracture healing can take place under specific conditions of displacement and certainly does not occur when the displacements in the gap reach or exceed the fracture gap size [7]. As demonstrated by Klein, the initial phase of fracture healing is specifically sensitive to mechanical conditions such as mobility in the fracture gap or mutual displacement of bone fragments, which ultimately has a crucial impact on the overall healing process [8]. Subsequent studies indicate that excessive displacement values in the fracture gap can lead to inhibition of the rate of fracture healing, which occurs e.g., in the case of unstable

application of fracture fixators [9]. In extreme cases large displacement values may prevent consolidation of bone fragments [10], [11].

In addition, it is important to ensure normal, or at least close to normal, range of motion of the knee joint. Until recently, this goal was very difficult to achieve. Imperfect treatment methods made it impossible to obtain sufficiently stable fixation of fracture fragments. Conservative cast treatment forced the patient to remain in an uncomfortable hip cast for several weeks. The articular surface could not be restored. Stability of immobilization of the fracture fragments in a plaster cast was questionable. Prolonged immobilization resulted in union of the fracture but the function of the knee joint was often permanently impaired. Major progress was made when the concept of treatment of those fractures was overhauled by Prof. Tylman [12], [13]. Functional treatment introduced first passive and then active movement of the knee joint. This ensured that union of the fracture was accompanied by satisfactory restoration of the function of the knee joint. However, normal articular surface could not always be restored during functional treatment. Surgical treatment makes it possible to restore normal articular surface of the distal end of the femur. Over the years methods of surgical treatment have been systematically improved. Initially, minimum fixation was used, which involved osteosuturing with Kirschner wires. Such fixations were very unstable and required additional immobilization in a cast. This gave a result similar to that of the conservative treatment. The joint function was very limited after prolonged immobilization. Other methods of fixing distal femoral fractures used angled plates and the DCS (dynamic condylar screw) plate. Good fixation stability was obtained thanks to the use of angular stability between condylar and diaphyseal components of the plate. The diaphyseal part of the plate was fixed to the shaft of the femur with screws driven into the bone. By pressing the plate against the bone, the screws increased friction between the plate and the bone and thus stabilized fixation. However, this stability deteriorated due to ischemic changes that occurred in the bone pressed upon by the plate. Often, despite the solid build of the angled plate and the DCS plate, the fixation became destabilized and the limb had to be additionally immobilized in a cast. This drawback did not exist in the case of locked intramedullary fixations. Here, the stabilizing element is located in the marrow cavity. It is inserted into it through the articular surface of the distal femur. This means that the already badly damaged articular surface of the femur has to be re-injured. Moreover, the

number of locking screws in the distal end of the femur is sometimes insufficient to stabilize comminuted fracture and enable implementation of early movement of the knee joint. The optimal solution seems to be a combination of plate and intramedullary osteosynthesis. In 1990, a patent was taken out on a system of locking plate and a guide allowing for a minimally invasive surgical technique with the LISS (less invasive stabilization system) approach. It enabled percutaneous locking of screws in the plate without the need to expose the operated bone. The screws lock on a threaded connection between the plate and the screw head [14]. Thanks to that, the plate is not pressed against the bone and, consequently, does not interfere with its blood supply. The first implantation of the LISS plate took place in 1995 and the LISS technique was approved by the AO Technical Commission as a new surgical method [14]. Development of new surgical techniques aims to achieve a method that meets the requirements of optimal osteosynthesis, i.e., allows the articular surface to be restored and fracture fragments to stabilize well enough to enable implementation of movement in the knee joint practically two or three days after the surgery. Such is the promise of the LISS method.

2. Aim of the study

The aim of the study was to numerically determine the stability of fixation of a 33-C2 intraarticular fracture with the LISS plate according to the AO classification, Fig. 2. Description of the distribution of displacements in the analyzed area should help to determine whether the strength and stability of a fixation with the LISS plate allows for implementation of active movements in the knee joints without compromising the fracture union process and whether it does not destabilize the fixation. Thus it will be possible to develop an algorithm for postoperative management.
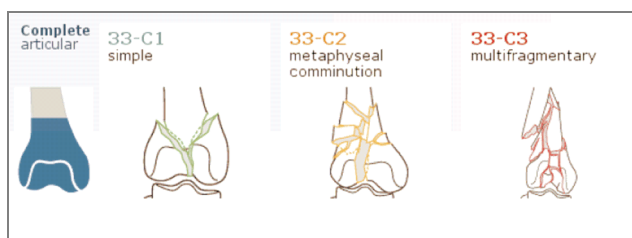

Fig. 2. Classification of periarticular fractures of long bones acc. to the AO Foundation [15]

The scope of the project covered:

• development of a geometrical model of a 33-C2 fracture of the distal femur according to the classification of the AO Foundation,

• development of a geometrical model of a LISS implant,

• combination of models into a set and its discretization,

• determination of boundary conditions (material of compact and spongy bone tissue as well as the implant material, the loading model),

• calculation of distributions of displacements using the method of finite elements.

3. Materials and methods

As part of the developed simulation we prepared a numerical model of the bone and the used fixator. The geometrical model of the bone was developed with the Ansys v. 11 software on the basis of the data derived from CAT scans of the femur of a 40-year-old patient weighing 83 kg. On the basis of the tomographic measurements we obtained a set of data recorded in the DICOM format. The data were transferred to the Ansys system using Mimics software. The obtained initial geometrical model reflected the shape and dimensions of the external surface of the distal end of the femur as well as the shape of the surface constituting the border between compact tissue and spongy tissue.

The modeled bone fracture was stabilized with a steel LISS plate. The geometrical model of the implant was developed indirectly based on the catalogue sheet and directly based on our own measurements carried out with an accuracy of up to 0.1 mm. Simplified representation of the fixator plate model omits two bevels in the main part that facilitate positioning of the plate along the femur but do not affect distribution of interfragmentary movements. In addition, some holes in the distal part of the plate were omitted from the representation, whose small diameter makes their impact insignificant in terms of plate stiffness and, consequently, interfragmentary movements. The geometrical models of bone screws did not include the bone thread because fixing of screws in the bone would be provided by properly joining the meshes of finite elements on the bone and screws. The developed geometrical models of bone and implant were then subjected to a discretization process using a solid tetrahedral element of a higher order with 10 nodes and 3 degrees of freedom at each node. In the bone

model, a division was made into compact and spongy bone tissues in order to better approximate the characteristics of the modeled bone. We assumed isotropic nature of the model-building materials, whose characteristics are summarized in Table 1. In the case of the implant model, we adopted typical material characteristics used in the manufacture, i.e., of austenitic steel. Figures 3–4 show a model of the stabilizing plate and its position relative to the bone.

Table 1. Values of the mechanical properties of the materials and tissues represented in the models [16]

Type of material	Modulus of elasticity (MPa)	Poisson's ratio (–)
Compact bone tissue	18,600	0.3
Spongy bone tissue	480	0.42
Fracture gap tissue	5	0.3
Austenitic steel	210,000	0.3

A model of bone with the inserted fixator was analyzed in 3 loading scenarios:

• model representing load on a lower limb in horizontal extension in the supine position;

• model representing flexion of a limb at the knee at a 45° angle in the supine position;

• model representing flexion of a limb at the knee at a 90° angle in the supine position.
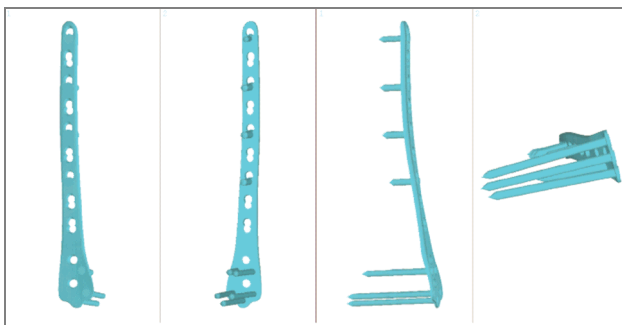

Fig. 3. LISS locking plate. Lateral, medial, anterior, and distal views

Fig. 4. LISS locking plate mounted on a model of the femur. Anterior, posterior, lateral, and medial views

Due to the lack in the literature of an appropriate model representing load on the knee joint in the lying position, we had to create a model of loading of the lower limb in the lying position and during the movement of raising the lower leg with simultaneous flexion of the knee joint. Such a model was developed on the basis of a simplified two-dimensional analysis, as a result of which, thanks to the use of the rule of three forces for systems of converging forces, we obtained three loading scenarios corresponding to the assumptions and the prepared numerical models (load on the lower limb in horizontal extension in the supine position, flexion of the limb at the knee at a 45° angle in the supine position, and flexion of the limb at the knee at a 90° angle in the supine position). Figure 5 shows the concept of creating a loading model.
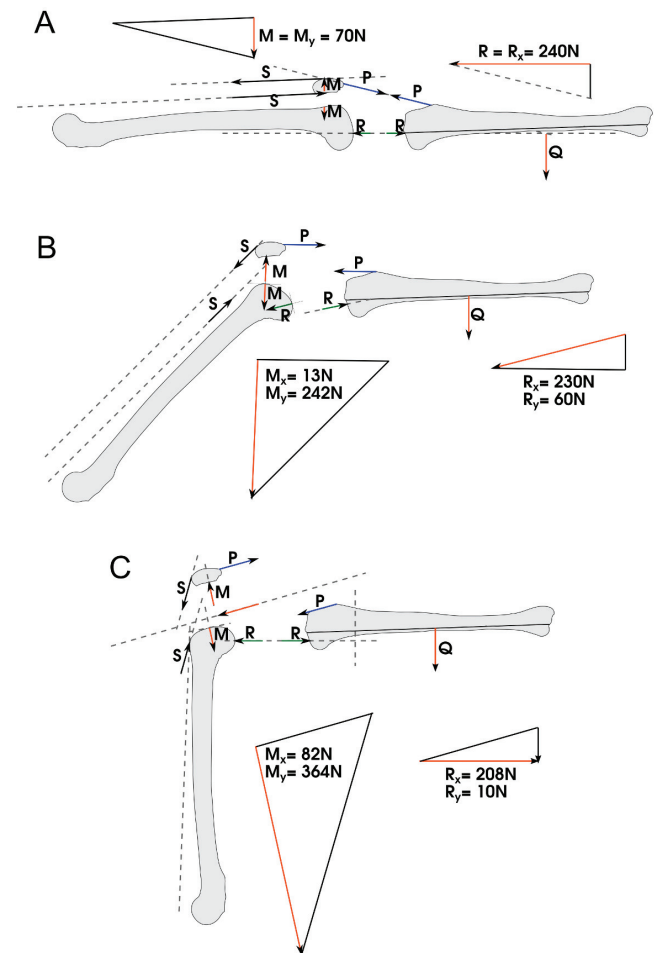

Fig. 5. Model of femoral loading in the supine position, in extension (A), in flexion at 45° (B), in flexion at 90° (C)

The bodyweight of the patient was the model parameter determining the values of the forces involved in loading of the numerical model. Next, in accordance with the assumptions about the weight of the

lower extremities ([17] p. 159), it was assumed that the weight of the lower leg was 7.5% of the body-weight. This provided the basis for the calculation of the forces acting on the infrapatellar region (force M, area A) and the articular surface area (force R, areas B and C).

The spatial model was loaded by applying forces to the nodes lying on the surface of the model in specific regions. Figure 6 shows loaded areas of the femur in extension. Thus, in the model we can distinguish force acting on the infrapatellar area with the strength calculated in a 2-D model; the other two forces exert equal load on the articular surfaces of both femoral condyles. All forces were converted from a 2-D model and transferred taking into account changes in the system and values of the forces. The applied forces and their values changed for each loaded area.
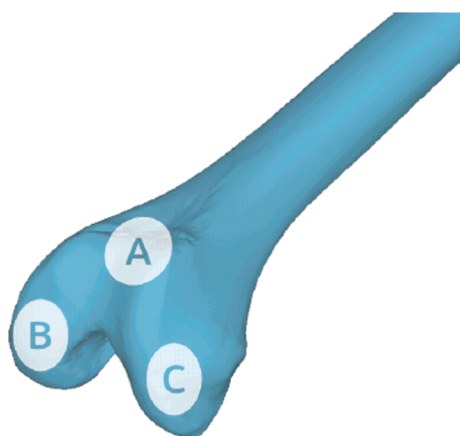

Fig. 6. Loaded areas of the femur in extension

4. Results

As a part of the conducted comparative analysis of the state of distribution of the bone–implant system, distributions of the displacements were specified along all axes of the global coordinate system. The *x*-axis is the horizontal axis in the coronal plane, directed laterally. The *y*-axis is the vertical axis directed upward. The *z*-axis is the horizontal axis in the sagittal plane, directed forward. Figure 7 shows seven gaps highlighted in the general form of fracture, which were subjected to a more detailed analysis of displacements.

Analysis of displacements of the tibia with a type 33-C2 fracture showed that, for the model representing extension of the knee joint, the loaded bone becomes bent forward in the sagittal plane and this ten-

dency grows at the distal end. The bone is also bent in the coronal plane. Displacements along the *X* axis show that femoral condyles are displaced medially. The diaphyseal region of the bone becomes slightly and gradually bent laterally by a value of 0.07 mm. Isolines become increasingly dense as they move towards the distal end past the first fracture gap. Also, the displacement direction changes – the distal end tends to move medially. Analysis of displacements along the *Y* axis shows that the forces acting on articular surfaces cause stretching of the anterior section of the diaphysis; the *UY* displacements in this bone region reach an absolute value of 0.12 mm. The distal end of the bone behaves differently, i.e., its posterior part becomes raised by a maximum of 0.7 mm but this value is reached only on the medial condyle while the posterior part of the lateral condyle is characterized by the *UY* displacement of 0.42 mm. The *UZ* displacements indicate bone bending in the sagittal plane, which is expressed by forward bending of the bone shaft by a maximum of 1.6 mm with simultaneous medial rotation. Isoline density is uniform in the diaphysis and becomes slightly denser towards the distal end.

A more detailed analysis was conducted in each fracture plane, which gives a more complete picture of nodal displacements in the model and makes it possible to determine if the set boundary conditions adversely affect bone tissue regeneration, disturbing or even preventing reconstruction processes.
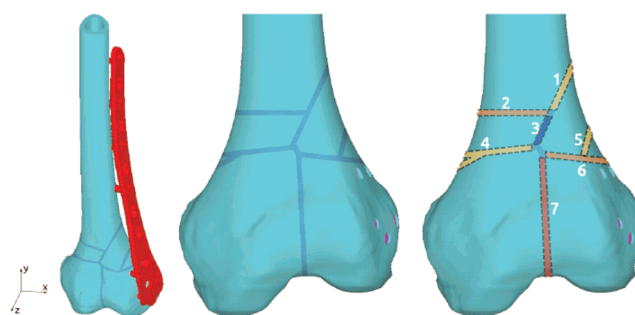

Fig. 7. Axes of the global model system and type 33-C2 fracture with the highlighted seven fracture gaps

In gap No. 1, the maximum dislocation was noted for the *Y* axis, which deviates from the fracture plane by approx. 30°. The area of maximum displacement covers approx. 5–10% of the gap surface. In gap No. 2, the maximum interfragmentary movement, by one of the greatest amounts recorded in the model, was observed on the *Y* axis perpendicular to the fracture plane; the displacement area covers, accordingly, approx. 20–25% of the fracture gap surface. In gap No. 3,

the maximum interfragmentary movement was observed on the *Y* axis, which deviates from the fracture plane by approx. 30°, while the value of the displacement is almost uniform throughout the gap – the whole bone fragment moves in parallel. Similar to gap No. 2, in gap No. 4 the largest interfragmentary movement was recorded on the *Y* axis perpendicular to the fracture plane. Interfragmentary movement of very similar value was found throughout the gap. In gap No. 5, the displacement values were negligible. In gap No. 6, the largest displacements were observed on the *X* axis, parallel to the fracture plane, which covered with their area approx. 10–15% of the entire gap area. No interfragmentary movement was found in gap No. 7.

In the model illustrating a 45° flexion of the knee joint, an analysis of *UX* displacements throughout the bone shows that the diaphysis and the proximal section of the distal femur are displaced laterally by a maximum of 0.17 mm – this tendency is observed up to fracture gaps Nos. 4 and 6. Starting from the above-mentioned gaps, the rest of the end section is displaced medially – by a maximum of 0.26 mm.

The *UY* displacements show that during bone bending under load, the entire posterior part of the distal end displaces towards the proximal end whereas the anterior part of the bone, both its diaphysis and epiphysis, are minimally displaced distally. This process is not uniform – in the case of displacements towards the proximal end, the medial condyle displaces with more intensity, by 0.72 mm on the front side and by 0.05–0.25 mm on the back side. Accordingly, the lateral condyle displaces by 0.52 mm and by –0.08 mm.

The UZ displacements demonstrate clearly the process of bone being bent forward. Isolines are characterized by horizontal distribution that becomes denser at the distal end – it displaces by a maximum of over 1.3 mm.

Smaller condylar and diaphyseal inclination backward is related to the fact of partial equilibration of the forces applied to the bone.

The resultant *FZ* force in the sagittal plane amounts to approx. –43 N compared to –70 N in the first model.

In gap No. 1, the maximum dislocation was noted for the *Y* axis, which deviates from the fracture plane by approx. 30°. The area of maximum displacement covers approx. 5–10% of the gap surface. In gap No. 2, the maximum interfragmentary movement, by the greatest amount recorded in the model, was observed on the *Y* axis perpendicular to the fracture plane; the displacement area covers approx. 10% of the fracture

gap surface. In gap No. 3, similar to gap No. 1, the maximum interfragmentary movement was identified on the *Y* axis, which also deviates from the fracture plane by approx. 30°, while the value of the displacement is almost uniform throughout the gap – the whole bone fragment moves in parallel. In gap No. 4, similar to gap No. 2, the largest interfragmentary movement was recorded on the *Y* axis, perpendicular to the fracture plane. Interfragmentary movement of very similar value was found throughout the gap. In gap No. 5, the displacement values were negligible. In gap No. 6, the largest displacements were observed on the *X* axis, parallel to the fracture plane, which covered with their area approx. 30% of the entire gap area. No interfragmentary movement was found in gap No. 7.

In a model of stabilized femur simulating a 90° flexion of the knee joint, the general form of displacement looks different than in the previous cases. Under the influence of external forces the bone becomes bent forward with simultaneous lateral rotation.

The *UX* displacements show a slight lateral bending of the diaphysis, but starting from the highest fracture gaps, the distal end is displaced medially under the influence of external forces. This tendency is strongest in the posterior part of the condyles, which change their original position by 0.47 mm.

Displacements along the long bone axis point to the process of forward bone bending, conversely to the previous cases of loading.

The anterior part of the condyles moves towards the attachment point by a maximum of 0.34 mm while the posterior part moves away by almost the same value. The process is particularly evident at the end of the bone, where the fragments may be compressed. This is impossible in diaphysis, which is a monolithic block.

The *UZ* displacements are characterized by a horizontal arrangement of the contour lines in diaphysis, indicating uniform displacement in the sagittal plane. Such regularity disappears in the region of condyles, where isolines become inclined in the coronal plane – forward movement is more prominent in the case of the lateral condyle – by 1.76 mm.

In gap No. 1, the maximum dislocation was recorded for the *Y* axis, which deviates from the fracture plane by approx. 30°. The area of maximum displacement covers approx. 5–10% of the gap area.

In gap No. 2, the maximum interfragmentary movement, by the greatest amount recorded in the model, was observed on the *Y* axis perpendicular to the fracture plane; the displacement area covers approx. 10% of the fracture gap surface. In gap No. 3,
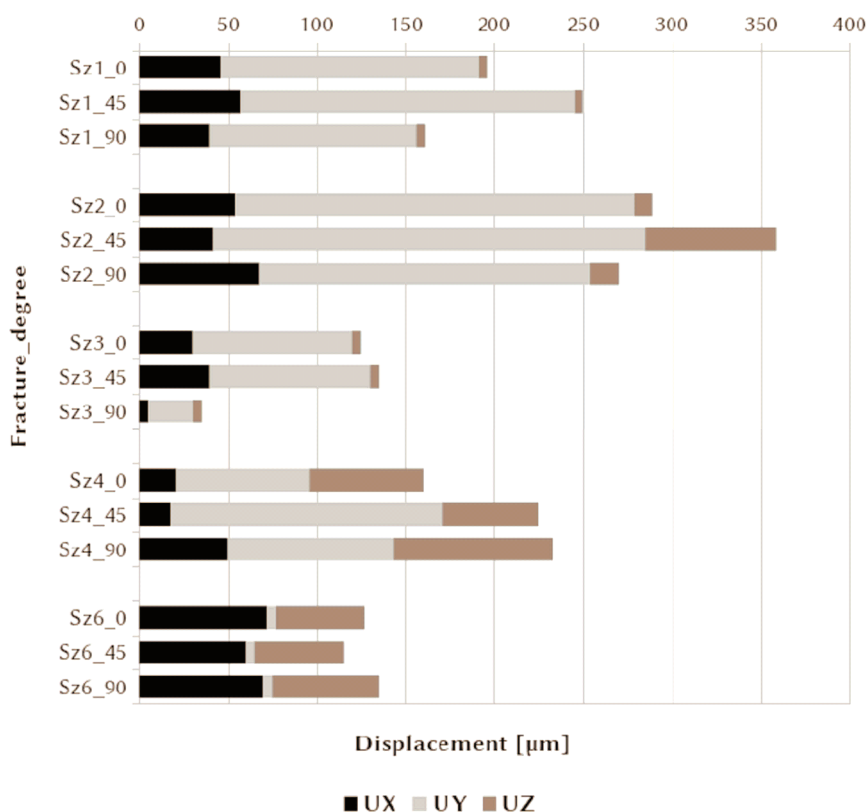
Fig. 8. The maximum value of displacements recorded in the respective gaps
for the analyzed loading scenarios – cumulative graph

similar to gap No. 1, the maximum interfragmentary movement was observed on the Y axis, which also deviates from the fracture plane by approx. 30°, while the value of the displacement is almost uniform throughout the gap – the whole bone fragment moves in parallel.

In gap No. 4, the largest interfragmentary movement was recorded in the *Y* and *Z* axes, which were, respectively, perpendicular and parallel to the fracture plane. The first interfragmentary displacement covers approx. 10–15% of the gap, while the second one covers approx. 10% of the surface. In gap No. 5, the displacement values were negligible. In gap No. 6, the largest displacements were recorded in the *Y* and *Z* axes, which were, respectively, perpendicular and parallel to the fracture plane. In both cases the area of such behavior of the fragments covers approx. 5 to 10% of the gap surface.

No interfragmentary movement was found in gap No. 7.

The values of the displacements that were recorded in the respective gaps for the analyzed loading scenarios were summarized in the graph, see Fig. 8. Due to the minimum values, no displacements were included for the gaphs 5 and 7.

5. Discussion

The LISS method (Less Invasive Stabilization System) is a relatively new method of surgery. The first implantation took place in 1995 [14]. The reports found in the literature focus on clinical assessment of the LISS method and its comparison to other, previously used methods of surgical treatment of fractures of the distal end of the femur [18], [19]. In our study, we looked at the discussed issue from a different perspective. We assessed the biomechanical strength of the broken bone–LISS fixator system. The aim of the experimental study performed on a digital model was to determine if use of active movement in the knee joint of the operated limb would not disturb union of the fracture. Because the lower limb loading models available in the literature [20], [21] concern the standing position, for the purposes of our study we created our own, original model of loading of the distal end of the femur during flexion of the knee joint in the recumbent position.

The issue of interfragmentary movement and its effect on fracture union was investigated by Claes and

Augat. In their studies [22]–[24], the authors demonstrated that fracture gaps exceeding 2 mm are slower to reconstruct than the gaps measuring less than 2 mm. Also they demonstrated a beneficial effect of interfragmentary movement on the process of callus formation. Of course, it is preferred that this movement takes place relative to an axis perpendicular to the fracture gap and is in the range of 0.15–0.34 mm [25] or even up to 0.5 mm [26]. Goodship in his report [27] allowed interfragmentary movement up to 1 mm. Shear movements taking place in the planes parallel to the fracture gap reduce the strength of the forming callus [24] and should be significantly lower in comparison to the movement taking place relative to the axis perpendicular to the fracture gap.

According to the experiment conducted, distribution of displacements in all loading models shows that the boundary value of interfragmentary movements indicated in the literature was not reached.

Such behavior of the fragments suggests that loads placed on the limb of the patient in the supine position are in the safe range [22], [25]–[28] and thus it is possible to rehabilitate the patient with the use of his or her muscle strength and load the stabilized bone with the forces resulting from the patient's limb weight.

The largest recorded displacements amount to 243 µm (33-C2, 45° flexion, gap No. 2, *UY* the axis perpendicular to the gap No. 2), which should be regarded as a safe value in the light of the literature data. What is particularly important, two condylar fragments that are bonded by means of screws show consistent and uniform behavior in all models. There is no mutual movement of the edges. With the bodyweight-only loads applied in the lying position, there are no displacements which, according to the literature, might be considered dangerous from the viewpoint of fracture union and follow-up subsequent rehabilitation of the patient.

6. Conclusions

It should be noted that the loading model only considers the femur, tibia, and patella and an idealized distribution of forces. However, it does not include the contribution of the ligaments and the surrounding tissue, which, despite the introduction of additional forces, also help to stabilize the knee joint. The presented model of the distal end of the femur and the method of its loading indicate that interfragmentary movement does not reach the values capable of disturbing fracture consolidation. Additionally, the scaling up of the forces loading the model to simulate a patient with a higher bodyweight translates into new distributions of displacements in the model. Note that these changes are linear and doubling the loads (patient weighing 166 kg) doubles interfragmentary displacements. The maximum movement between the fragments for such patient in the model described reaches approx. 0.5 mm, which appears to be safe considering the range reported in the literature (0.15–1 mm) [25], [27], [28]. In the light of this study, the LISS method has a large margin of safety. Nevertheless, it must be noted that this model simplifies reality and the theoretical study must be confronted with clinical research.

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