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Abstrakt:

Background. The analysis of the effect of axial angle changes on the weight-bearing total knee replacements (TKR) was investigated and discussed.
Methods. A two dimensional finite element models of the femorotibial joint in the frontal and the sagital planes were developed from CT images. For computation the nonoverlapping finite element technique develped in our institute was used.
Results. The effect of axial angle changes on the weight-bearing total knee replacements is analysed in frontal and sagital planes. It is shown that optimal distribution of forces operated on the TKR in anteroposterior direction and well-balanced transition of forces in anteroposterior direction will correspond to the 7 deg. case. The results show a good agreement between theoretical and orthopadic observations.

Keywords:
Biomechanics, Knee, Total replacement, Axial changes

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Analysis of axial angle changes on the weight-bearing total knee replacements

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Abstract

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1 Introduction and clinical aspect problems

Several authors have employed finite element analyses based on different rheologies for stress and strain analyses in orthopeadic biomechanics. These techniques present a wide range of applications in orthopaedic areas like bone remodeling analysis (Weinans at al., 1993), mechanical behaviour of loaded bones and joints with or without implants (Mann et al., 1995; Vavřík et al., 2005). In (Nedoma et al., 1999, 2003; Daněk et al., 2004; Nedoma et al., 2006) it was showed that the contact problems in suitable rheology and their finite element approximations are very useful tools for analyzing these relations for knee joint and its artificial replacement.

Most of femorotibial joints models were developed based on the results of in vitro experiments, stereophotogrammetry (Athesian et al., 1991; Blankevoort et al., 1991), on the X-ray body scanner radiograph and at present on the CT and MR imaging (Keyak et al., 1990, 1993).

Investigation of axial angle changes on the weight-bearing total knee replacements has been studied by many authors (Hungerford, 1995; Ritter et al., 1994; Sparmann et al., 2003).

From the view of the orthopaedic surgeon, division load acting on the tibial component after implantation of the total knee replacement is of primary importance. The pressure ratios in the knee after the TKR are decided by the soft tissue tension (capsule, ligaments, muscular insertions) in the vicinity of the replacement and the resulting axial position of the whole limb. Both these basic factors are influenced especially by the own technique of implantation and, in a decisive way determines the survival time of the implant. From many reports it follows unambiguously that nonobservance of the balance of both compartments (medial and lateral) or possible overloading of the posterior part of the tibia plate leads to quick wear out with considerable abrasion of the polyethylene insert. No implant tolerates mistakes in surgical technique, older simple types of prostheses fail just as the more sophisticated and several-times more expensive modern implants do.

Due to asymmetrical overloading, the premature abrasion of a plastic insertion with production of a great amount of polyethylene elements, which initiate a complicated inflammatory reaction leading to loosening metallic components of the total replacement from the bone.

Soft tissue tension

Balancing of the soft tissue tension of the external and internal parts of the joint in the frontal plane is very difficult and surgeons have to accept some compromises in the cases with very severe deformity. A similar problem is the balancing in the sagittal (vertical antaroposterior) plane. The contracture of the posterior cruciate ligament (PCL) influences unfavourably the load in the posterior part of the joint. It is imperative to loosen partly the ligament very often or to resect it completely. In such cases it is necesary to use a variant of the implant with mechanical posterior stabilization. Evaluation of the joint balance is, to a large extent, subjective and measurable with difficulty and, by this time, even the most up-to-date instrumental systems with computer assisted instrumentation do not solve the problem entirely. The question what measure the precision of the soft tissues balance and of the axial deviations still has the essential importance for a long-term survival of the implant.

Axial position of the limb

Participating in the resulting axial position of a limb are many factors which must be respected and carefully evaluated clinically and radiologically before the surgery. Changes on the skeleton are: (i) acquired or congenital changes of the form of the femur or the proximal tibia, (ii) deviation of the knee joint axis under the influence of the adjoing joints.

Add (i) Changes in the form of both bones forming the knee joint can be the causes of incorrect position of the component. In the case of the distal femur it is very often a gradually developing deformity, mostly hypoplasia of the lateral condyle causing a valgus deformity of the knee. Even a certain extent of the rotation round the vertical axis outside is a component of the valgus deformity.

A similar situation arises on the post-traumatic axial deformities of the femur when gradual damage of the overloaded components of the femur and development of the joint deformity come to light. That means that in on the valgus deformity of the femoral diaphysis it comes to overloading of the external side of the joint, of its gradual damage and, in the final phase, to valgus deformity knee. The varus knee position aries similarly in healing diaphysis in varus.

As for the proximal tibia, the big difficulties bring on the defect in some of the articular parts with consequential deformity of the knee. Very often is on the medial compartment (varus deformity), in postero-medial part (combination with soft-tissue contracture in flection) or less frequent defect in the lateral compartment (valgus deformity). We encounter also with the specific external rotation deformity by patients after osteotomy of the tibia.

Add (ii) Axial deviations of the lower limb can be also be an effect of the infliction of the adjoining joints. They show themselves by a compensatory deformity in the area of the knee joint.

In the acquired deformities the varus position of femoral neck results to lateralization of diaphysis and valgus position to its medialisation. Thereafter, compensatory mirror deformities emerge in the area of the knee joint.

The tibia, primarily its lower half appears usually at men, is a noticeable varus position which cannot be revised within the resection of the proximal tibia. In order to make it possible to walk with plantigrade position of the foot, the compensatory position of the ankle joints emerge in the mentioned situations. Alternatively, rigid deformities of the leg and talocrural (ankle) joint can unfavourably influence the position of the once healthy knee joint by the formation of a compensatory deformity.

Before the operation TKR of the knee joint, we evaluate the real axis of the lower limb from the X-ray body scanner radiograph, where not only the knee joint but also the head of the femur and talocrural joint must be taken. The conjuctiva of the femur center with the head of the femur in the center of the ankle joint determines so-called mechanical axis of the lower limb. After the adjustment of the mechanical axis, the angle with the anatomic femoral axis determines the degree of physiological valgus, in which the resection of the distal end of the femur should be performed. If we maintain the correct technique of implantation, restoring mechanical axis of the limb and collateral soft tissue balance and preventing too tight flexion gap of the knee: we will create the basic conditions for good function of the knee replacement. The results of the mathematical simulation of the load by a finite element method with numerical statement of deviations in various degrees of resection of the femur lower end are shown by individual models in frontal and sagital planes (Fig. 1 a,b,c).

2 The method

The success of the artificial replacements of the knee joint depends on many factors. The mechanical factor is an important one. The idea of a prosthesis being a device that transfer the knee joint loads to the bone allows explaining the mechanical factor in terms of the load transfer mechanism. A complex relation exists between this mechanism, the magnitude and direction of the loads, the geometry of the bone-joint prosthesis configuration, the elastic properties of the materials and the physical connections at the material connections. In (Nedoma et al., 2003; Daněk et al., 2004; Nedoma et al., 2006) the authors showed that the contact problems in suitable rheology and their finite element approximations are very useful tools for analyzing these relations for knee joint and its artificial replacement. As the method used the contact problem with friction in elasticity, the finite element method and the nonoverlapping domain decomposition method (Daněk et al., 2005a) were used.

The model was composed of four different materials for bone, femoral component, tibial component and tibial insert. The material properties chosen are linear elastic isotropic to describe the cortical bone and both components of the total knee replacements, manufactured by the titanium and $CoCrMo$ alloys and the tibial insert component from the ultra high molecular polyethylene - UHMWPE. The following material parameters are used:

(i) bone: Young's modulus of elasticity $E = 1.71 \times 10^{10} [Pa]$, Poisson constant $\nu = 0.25$,

(ii) anchoring plate - Ti6A14V: Young's modulus of elasticity$E = 1.15 \times 10^{11} [Pa]$, Poisson constant $\nu = 0.3$,

(iii) Polyethylen: Young's modulus of elasticity $E = 3.4 \times 10^{8} [Pa]$, Poisson constant $\nu = 0.4$,

(iv) CoCrMo: Young's modulus of elasticity $E = 2.08 \times 10^{11} [Pa]$, Poisson constant $\nu = 0.3$.

The geometry was obtained from the X-ray body scanner radiograph of the knee joint after the implantation of artificial knee joint prosthesis; an inserts producted of ultra high molecular polyethylene-UHMWPE form the contact areas with the femoral component of the total knee replacement.

The boundary conditions correspond to the parts of the artificial femorotibial joint where it is loaded, fixed and where the femoral and tibial parts are in contact. The loading forces were assumed to be corresponding to the normal weight-bearing human body. It is assumed that the tibia and fibula can be fixed in a suitable distance from the contact boundary betwen both parts of the artificial femorotibial joint.

The behaviour of femorotibial joint on the contact boundary is described by the contact conditions with Coulombian friction of Tresca type. The boundary of these contact zones is unknown a priori and, therefore, such boundaries are called as free boundaries.

During the deformation of the artificial femorotibial joint the contact points will be displaced in a different way, but colliding components of the artificial femorotibial joint cannot penetrate, i.e. the sum of normal components of the displacement vector in both part of the artificial femorotibial joint is equal or less than zero. The behaviour of the contact forces on the contact boundary follows from the law of action and reaction. Since normal components of contact forces cannot be positive (tractions), then the normal stress forces are less or equal to zero. During the deformation of colliding components of artificial femorotibial joint the colliding parts are in contact or are not in contact. If they are in contact, then on the contact boundary exist non zero contact forces. If they are not in contact, then the contact forces are equal to zero. These conditions are known as the Signorini contact conditions.

Further, if both colliding parts of the artificial femorotibial joint is in contact, then on the contact boundary act the frictional forces in the Coulombian sense, which in the absolute value are proportional to the absolute value of acting normal forces. The coefficient of proportionality is the coefficient of the

Coulombian friction. Due to the acting frictional forces we have the following cases:

If the absolute value of the tangential forces is less than the frictional forces, then the frictional forces preclude the mutual shifts of both colliding parts of artificial femorotibial joint. If the tangential forces are equal in their absolute value to the frictional forces, then both parts of artificial femorotibial joint mutually shift. At the same time points on the opposite sides of the contact boundary change their position in the direction opposite to the acting tangential forces.

In the Tresca model of friction the contact forces are limited by the friction bound (friction limit), i.e. by the magnitude of the limiting friction traction at which slip begins. Then if the absolute value of tangential forces is strictly less than this friction bound the points of the artificial femorotibial joint from the contact boundary is in the stick zone; when it is equal to this friction bound, the points of both parts of artficial femorotibial joint from the contact boundary are in the slip zone. The main feature of the Tresca friction law is the assumption that the friction bound is known.

The investigations are meditated in the frontal cut (MODEL I for 5deg., MODEL II for 7 deg. and MODEL III for 9 deg.) where it is possible to analyze the influence of the axial deviation. The sagittal cut (MODEL IV, MODEL V) tells only little about the influence of the axial deviation, but it tells something about the overload of the posterior part of the tibial plate in the sagittal (anteroposterior) direction.

3 The model

In the next the models of axial angle changes of the weight-bearing total knee replacement in the frontal plane and in the sagittal plane are defined.

The frontal plane

MODEL I matched up with the angle of the valgus in the resection of the lower end of the femur 5 degree, MODEL II 7 deg. and MODEL III 9 deg. (Fig. 1a).

The examined femorotibial area of the knee joint occupies the region created by the femur, the tibia and the fibula in the frontal plane. The boundary is created by parts 1-2 and 3-4, where the fibula and tibia are fixed, by parts 7-8 and 9-10, which are contact boundaries between both parts of the femorotibial joint, by part 11-12, where the fibula is in contact with the tibia, by part 5-

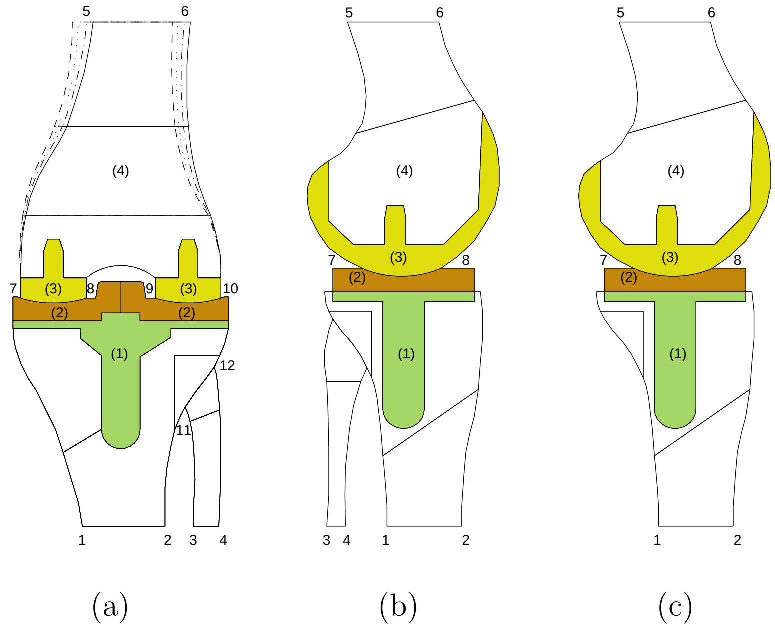

Fig. 1. The models: a) MODEL I-III, b) MODEL IV and c) MODEL V

6, where the load is prescribed, on the remaining parts of the boundary the femoral joint is unloaded.

Boundary conditions are prescribed on parts of the boundary of the femorotibial joint denoted by 1-2,3-4, 5-6, 7-8, 9-10 and 11-12. Zero displacement is prescribed between points 1-2 (fixed tibia) and 3-4 (fixed fibula), between points 5-6 the femur is loaded by a loading $0.215 \times 10^7 [Pa]$, the unilateral contact conditions are between points 7-8 and 9-10, the bilateral contact condition is between points 11-12.

Discretization statistics are characterized by

MODEL I: 13 subdomains of domain decomposition, 3737 nodes, 7132 elements, 31+31+15 unilateral and bilateral contact nodes, 344 interface elements between subdomains of domain decomposition,

MODEL II: 13 subdomains of domain decomposition, 3780 nodes, 7208 elements, 31+31+15 unilateral and bilateral contact nodes, 350 interface elements between subdomains of domain decomposition,

MODEL III: 13 subdomains of domain decomposition, 3763 nodes, 7180 elements, 31+31+15 unilateral and bilateral contact nodes, 34 interface elements between subdomains of domain decomposition.

The sagittal plane

In the sagittal plane two cross-sections are investigated. Corresponding two models of the total knee replacements in linkage on the axial deviations 5, 7, 9 deg. across both condyles, where by MODEL IV the cut across outer condyle (on the figure marked by the fibula) and by MODEL V the cut across the inner condyle (on the figure without fibula) and in three variants denoted as A corresponding to 5 degree, B 7 deg. and C 9 deg., are constructed at Figs 1b,c. The sagittal cut tells anything about the influence of the axial deviation, it tells something about the overload of the posterior part of the tibial plate in the sagittal (anteroposterior) direction, and therefore, it informs us about a relevant strong wear of the ultra high molecular polyethylene - UHMWPE insert. In both models the same material parameters were considered as above.

The investigated femorotibial area of the knee joint in the sagittal plane occupies the region created by the femur, the tibia and the fibula in the sagittal plane cross-section. The boundary is created by parts 1-2 and 3-4, where the fibula and the tibia are fixed, by the part between points 5-6, where the loading is prescribed, by the part between points 7-8, representing the contact boundary between both parts of the femorotibial joint, by part 9-10, where the fibula through the tissues is jointed with the tibia, on the remaining parts of the boundary the femorotibial joint is unloaded.

The boundary and contact conditions in the case of MODEL IV were prescribed on parts of the boundary denoted by 1-2, 3-4, 5-6, 7-8, 9-10. Zero displacement is prescribed between points 1-2 (fixed tibia) and 3-4 (fixed fibula), between points 5-6 the femur is loaded by a loading $1.46 \times 10^6 [Pa]$ in the case of MODEL IV-A, $1.52 \times 10^6 [Pa]$ in the case of MODEL IV-B and $1.61 \times 10^6 [Pa]$ in the case of MODEL IV-C, the unilateral contact condition is between points 7-8 and the bilateral contact condition is between points 9-10.

The boundary and contact conditions in the case of MODEL V are prescribed on the parts 1-2, 5-6, 7-8. Zero displacement vector is prescribed on the part 1-2 (fixed tibia), on the part 5-6 a loading $1.2 \times 10^6 [Pa]$ in the case of MODEL V-A, $1.0 \times 10^6 [Pa]$ in the case of MODEL V-B and $0.86 \times 10^6 [Pa]$ in the case of MODEL V-C are prescibed, the unilateral contact condition is between points 7-8 and the bilateral contact condition is between points 9-10, the remaining part of the boundary is unloaded.

Discretization statistics are characterized by

MODEL IV: 10 subdomains of domain decomposition, 2867 nodes, 4740 elements, 33 unilateral and bilateral contact nodes, 299 interface elements between subdomains of domain decomposition,

MODEL V: 8 subdomains of domain decomposition, 2474 nodes, 4104 ele-

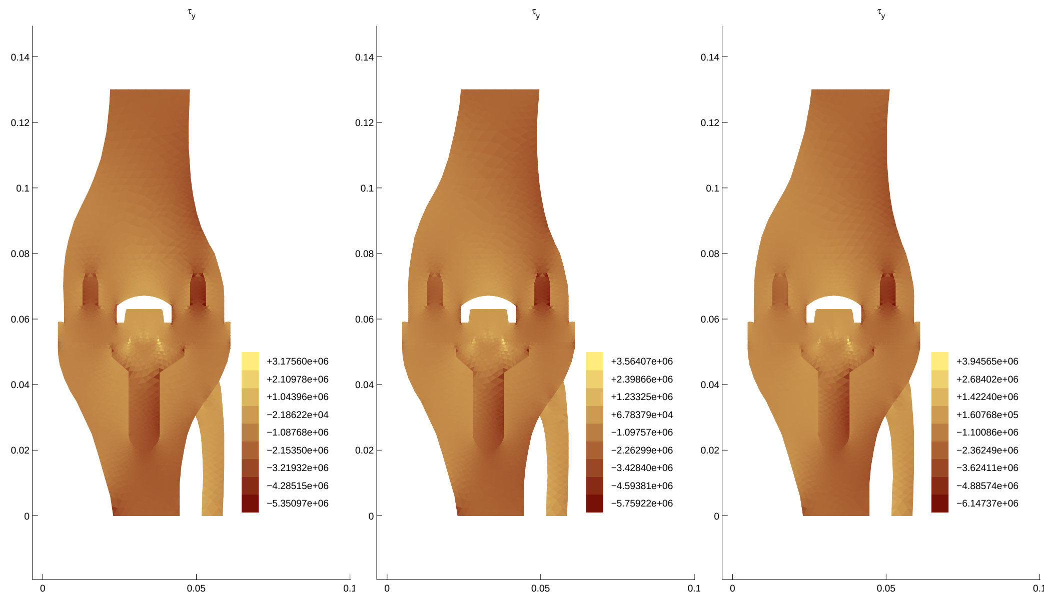

Fig. 2. The vertical component of the stress tensor for MODEL I, II, III

ments, 33 unilateral contact nodes, 290 interface elements between subdomains of domain decomposition.

4 Results

Based on our calculations the stress-strain fields for axial deviations 5, 7 and 9 degrees in frontal and sagittal cross-sections are analysed. Normal and tangential displacement components and normal and tangential stresses on the contact boundary between both components of femorotibial joint are also analysed.

4.1 Evaluation of numerical results in the frontal plane

In the numerical results we can see that in the knee joint the stress gradients become equal in epiphysis and metaphysis, and diaphysis is already strained evenly. For the vertical components of stress (Fig. 2), predominantly the pressure stresses are observed in the whole area, and the area incisura intercondylica and the medial margin are reduced and the stress grows further in the lateral direction, similar situation is observed also in the tibia. As shown by the numerical results, transmission of the pressure stresses is transmitted over the external lateral and internal medial parts of the contact area. Location of the maximal stresses in the direction of axis x is nearly identically situated with the area of the minimal stresses in the direction of axis y. In

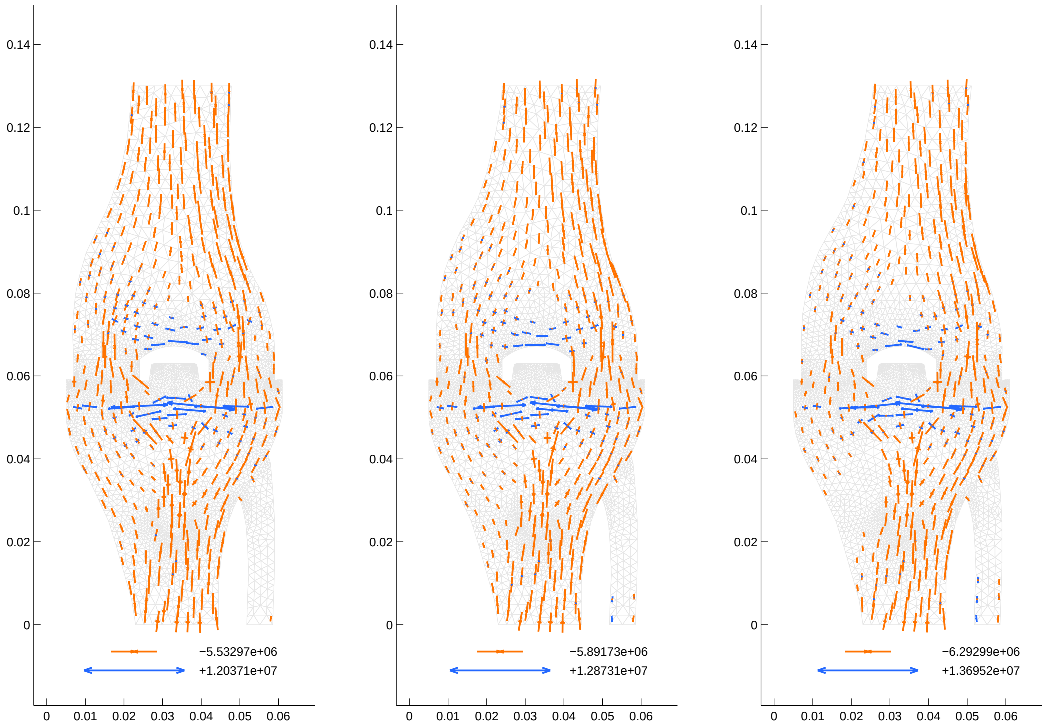

Fig. 3. The principal stresses for MODEL I, II, III

the course of a vertical component of stress and the principal stresses (Fig. 3) we can see that concentration of the pressure stresses is more located into the area of the external condyles of femur and tibie, the smaller part across the internal condyles, tensile stresses are located into the area incisura intercondylica. The directions of the power stream agree to the timber structure in epiphysis. The normal and tangential displacement components (Fig. 4a) and the normal and tangential stresses τ_n and τ_t (Fig. 4b) on the contact boundaries of both condyles have a testing value for the analysis of the total replacement of the knee joint in dependence on the axial deviation. The normal component DUn characterizes receding of both parts of the joint from one to another in the load of the knee joint and its subsequent deformation. From the analysis of the normal displacement component on the overload of the knee joint it results that during the loading both parts of the joint in a certain phase of weighting, for example during walking with a load, etc., the contact boundary retreat from one to the other in some points, even though the distance in the opposite points is relatively small. In the bigger overloading of the femorotibial joint, and similarly in the case of all human joints, it can give rise to a number of different types of cysts (Nedoma et al., 2006).

In our case, both components of the knee joint are constantly distributed in a close contact and there is no receding (removal, estrangement) during the loading in dependence on the axial deviation because the knee joint is not overloaded. The tangential component of the displacement vector DUt vector

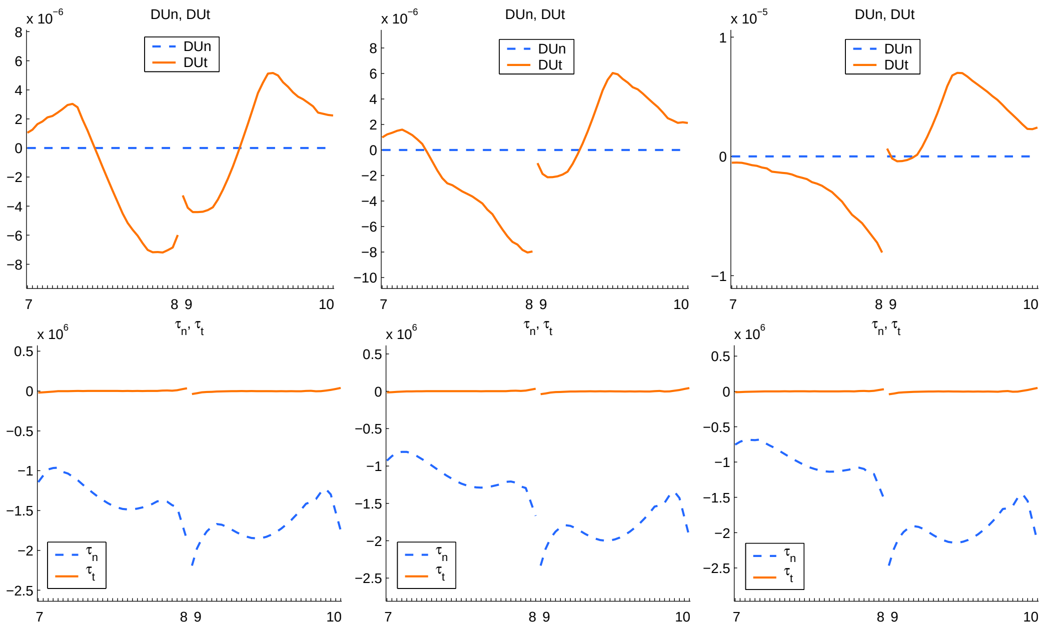

Fig. 4. Normal and tangential components of the displacement and of the stress vectors on the contact boundary in the area of both condyles for MODEL I, II, III

characterizes mutual shift of the contralateral points of both components of the knee joint in the tangential direction. The analysis of the tangential component of the displacement vector points a relatively small shifts in both condylar components of the joint. The character of their process is a little different for the studied axial deviation. From the values of both components of the displacement vector we get then real mutual shift of the contralateral points of the contact in the space. Normal and tangential stresses on the contact in both condylar parts of the knee joint characterizes loading relations on the both condylar parts of the knee joint. We see that on the contact between both femorotibial parts of the knee joint the pressure is observed in both condyles.

The analyses of numerical results deduce that optimal distribution of forces operated on TKR in anteroposterior direction and well balanced transition of forces in anteroposterior direction will corresponded to the 7 deg. case, which is also declared by analyses of principal stresses.

4.2 Evaluation of numerical results in the sagittal plane

The same computations as in the previous case in the frontal plane were made also for the cross-section in the sagittal plane.

Numerical results show that the greatest testifying values have on one hand horizontal component of the displacement vector U_x (Fig. 5) and on the other hand numerical results of normal and tangential components of displacement

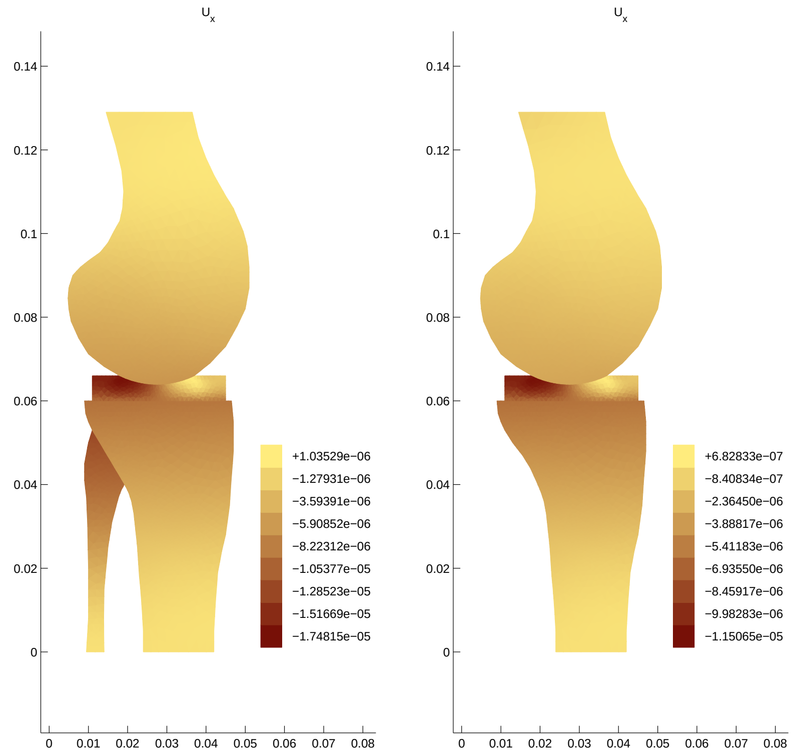

Fig. 5. The horizontal component of the displacement for MODEL IV - B and MODEL V - B

vector DUn, DUt and of normal and tangential stresses τ_n and τ_t on the contact between both components of the knee joint. The contact between both parts of the joint is at the part of the boundary between points 7-8. The point 7 corresponds to the investigated posterior part of the tibial plato. The figure 5 shows on the shifts within the bounds approx. $-1.748 \times 10^{-5}[m]$ in the case of the outer condyle and within approx. $-1.15 \times 10^{-5}[m]$ in the case of the inner condyle for the axial deviations 7 deg. in the area of the tibial plato. From these results it is seen that the polyethylene inlay is press out in the posterior part of the tibial plato, which is greater in the case of the outer condyle than in the case of inner condyle, and therefore, that the polyethylene inlay is deformed and worn out.

The results demonstrate certain overloading of the posterior part of tibial plato, which indicate the possibility of worn out and resulting wear of polyethylene insert of TKR.

Numerical results show that the $3D$ models will be necessary analysed. Such models then clear up certain antagonistic conclusions of the $2D$ models in the posterior area of tibial plato and a certain tendency of rotation of the polyethylene inlay towards the tibial plato.

5 Conclusions

From the first analyses of numerical results deduce that optimal distribution forces operated on TKR in anteroposterior direction and well-balanced transition of forces in anteroposterior direction will corresponded to the 7 deg. case, which declarate also the distribution of the primal stresses. Optimal transfer of forces in anteroposterior direction with maximum in the place of amplification of the posterior corticalis of the femur, fixation elements of femoral and/or of stem of tibial component is also observed in practice. On the contrary the overloading of the posterior part of the tibial plato in anteroposterior direction with unloosen soft posterior structures or incorrect inclination of the resection of the proximal tibia suggest our numerical results for the 9 deg. case of valgus. Analyses of results of cuts across the condyles in the sagittal plane document present certain overloading of the posterior part of the tibial plato, which suggest the possibility of worn out and resulting wear of polyethylene inlay of TKR. Numerical results indicate the necessity to analyse the $2D$ and $3D$ models of the knee joint system with joint capsule and ligaments. Convincing biomechanical analyses can give the analyses of $3D$ models of knee joint system only.

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