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Verification of mathematical model of pressure distribution in artificial knee joint

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Abstract. The paper deals with pressure distribution measurement in knee arthroplasty, which is an artificial replacement of human knee joint. The scope of the article is to verify the accuracy of a mathematical model by real measurements. The calculated pressure values basing on the mathematical model are compared with actually measured pressure values in the contact area of the joint. Hereby maximal load the in the contact area, the distribution of the pressure and any potentially dangerous pressure deviations during the walk cycle are checked. To enable accurate pressure distribution measurement without interfering into human's body, a sophisticated measuring setup was created: the contact area of the joint was equipped with several pressure sensors and a machine simulating the human walk cycle was used. The measured pressure data are finally compared with those from the mathematical model and with the strength limit of the used material, to verify the accuracy of the mathematical model experimentally.

Key words: knee arthoplasty, force sensor, artificial joint, pressure distribution, strain gage, tibial plateau.

INTRODUCTION

In some cases of a serious injury, joint disease or poor functionality of a human knee joint it is necessary to perform a total replacement of the joint with an artificial one – so called knee arthoplasty. The knee joint itself is the most complex one in human body due its complicated anatomic structure and multi-dimensional motion. To prevent a direct contact of two metallic parts within the artificial joint and to enable reasonable friction between individual parts, a polymer layer is placed within the joint, which comes into contact with the metallic part, for more information see Zach et al. (2004). As the polymer layer is especially susceptible to mechanical wear, it is necessary to check the actual pressure distribution in the contact layer and compare it with the strength limit of the material. The second scope of the measurement is to compare the measured pressures with calculated theoretical values, according to mathematical model by Zhu & Chen (2004).

In order to maintain the functionality of the leg, the replacement should meet the kinematical requirements on a healthy joint, and – as an implanted part of human's body

– it should exhibit excellent reliability to avoid repeating interventions into the body of the patient. Measurements of an already implanted arthroplasty would be virtually impossible as it would represent an excessive intervention into the patient's body. To enable measurements under nearly real conditions, a specific machine which simulates human step cycle with prescribed load was used. This way the actual pressure within the artificial joint may be determined without interfering into the body of the patient, so that a potential danger or malfunction of the artificial joint may be discovered before implanting the artificial joint into human's knee.

Unlike the previous work as presented in Volf et al. (2005), where only the maximal pressure in four points was measured, this measurement covers the entire contact area between the femoral component and tibial plateau with 22 probe positions. The scope of this work is not only to determine the possible pressure peaks, but to verify the pressure distribution in the contact area during the entire walk cycle, i.e. under changing knee angle flexion and thus under changing knee geometry.

There are possible alternative measuring techniques using a plate with matrix of capacitive pressure sensors or using a pressure sensitive foil that changes its colour according to the pressure. The advantage of relative easiness of such measurements is connected with necessarily influencing of the actual contact area, which yields unavoidable errors of the measurement. Therefore this measurement is performed without putting any material into the contact area, which ensures no geometry changes of the contact area.

MATERIALS AND METHODS

Composition of the knee arthoplasty

Most of the vital parts of the knee arthoplasty are made from metallic material (cobalt alloy – Vitalium), however, to prevent a direct contact between two metallic parts and to enable friction in the contact area between the femoral and the tibial part, the tibial plateau itself is made from Ultra-High Molecular Wide Polyethylene (UHMWPE).

New femoral components from oxide ceramic (Zirconium dioxide ZrO_2 and Aluminium dioxide AlO_2) are being developed. Metal materials are used because of their strength and elasticity, but they are not abrasion-proof and their life-cycle is shorter. Ceramic materials are bio-inert and exhibit good friction characteristic; however, their disadvantage is their enhanced fragility. Ceramic femoral component has also different geometric parameters, for more information see In the measurements, the described polyethylene - metallic combination is used, i.e. UHMWPE tibial plateau and metal femoral component. The composition of the artificial knee joint is shown in Fig. 1. Konvičková & Valenta (2000).

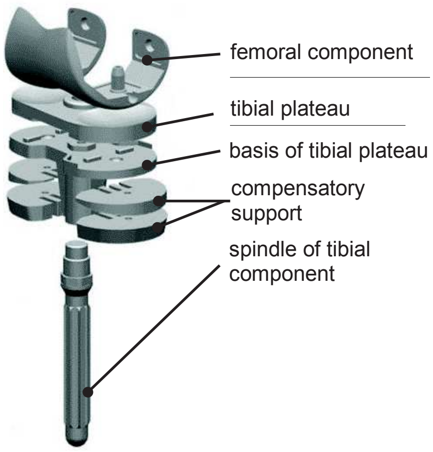


Figure 1. Composition of the knee arthroplasty.

The area which is the subject of the measurements is the contact area between the metallic femoral component and the tibial plateau made from polyethylene. This is the most vulnerable part of the artificial joint – it is the place where the most pressure concentrates, the two individual parts move relative to each and the tibial plateau is made from polyethylene which is susceptible to mechanical wear.

Mathematical model

The subject of pressure distribution in knee joint was reflected by Zhu & Chen (1999; 2004), who created a mathematical model of pressure distribution in a knee joint, more advanced model was subsequently presented by Zhu (2007). Calculations of the pressure distribution basing on the model are provided by Donát (1997) and Zach et al. (2004).

Some simulations of pressure distribution in knee joint deal with a physiologic knee, i.e. with a complete knee with muscles and fibrous apparatus. For this first verifying study, a simplified model of knee replacement which does not comprehend ligaments was used; thus the results might be slightly different according to Konvičková & Valenta (2000). The geometric model of knee replacement was created by finite element method, detailed description of this method is provided by Donát (1997).

Measurement procedure

To approximate the conditions of a real knee as much as possible and to avoid any further interventions into patient's body, a special measurement procedure was developed, consisting of two main distinctive techniques: use of movement simulator depicted in Fig. 2 and specific sensor placement.

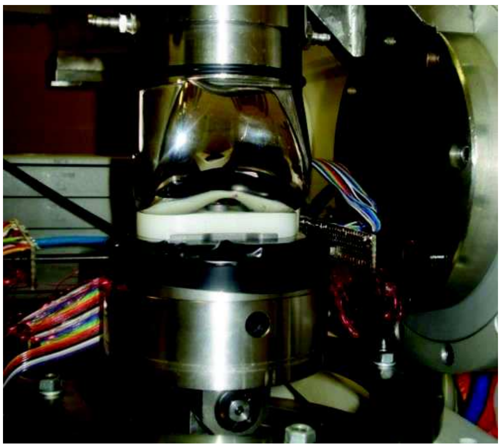


Figure 2. Knee movement simulator machine.

The movement simulator is a PLC controlled machine that models the movements of a human knee joint under real conditions. The measurements were performed according to norm ISO 14243-3:2004(E) that prescribes exactly the movements of individual parts of the joint in relation to each other and the exerted axial force; further details about the simulation in Zhu & Chen (1999). The norm gives a relation between the walk cycle in percent with associated pressure and flexion angle values as well as the associated force value. A graphic description of individual phases of human step is depicted in Fig. 3. The measurements were carried out in steady state, gradually for each phase of the walk cycle, i.e. separately for any individual flexion angle value.

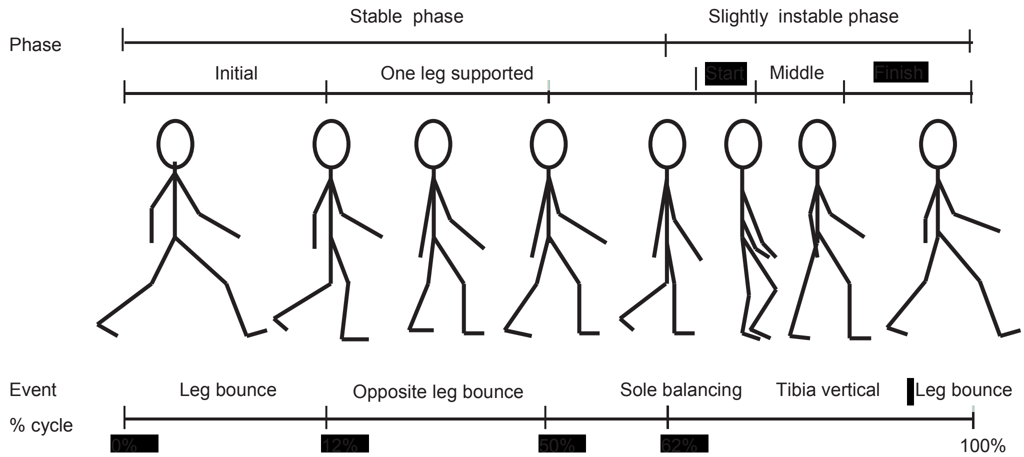
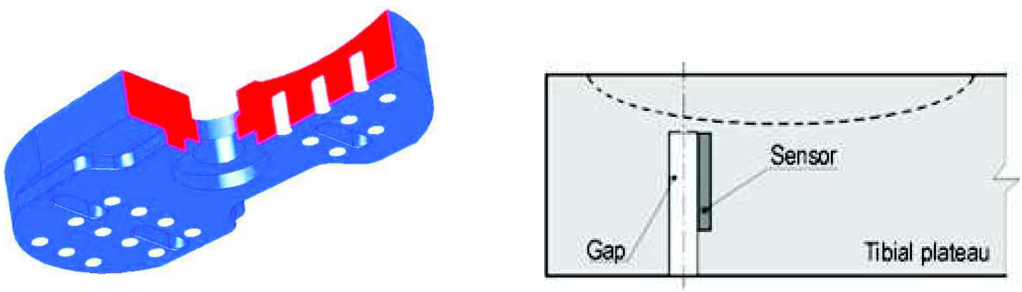


Figure 3. Phases of human step.

Sensor placement and parameters

The pressure sensors themselves placed in the artificial knee joint mustn't influence the surface shape of knee components. If sensor changes geometry of tibial plateau, it would change also the contact areas and thus contact pressures, as by Mootanah (2006) and Anderson & Lim (2006). In order to avoid any geometry changes in the contact area as discussed above, a new approach was chosen: there are bored at specified places into the polymer tibia plate several holes with 3 mm diameter and 2.9 mm bottom distance from the contact surface. A three-dimensional view on the plateau holes is displayed in Fig. 4a, and a cross-section of a hole with an attached sensor is displayed in Fig. 4b. The position of the individual holes with corresponding labeling is presented also on the top view in Fig. 5.



a) b)

Figure 4. a) tibial plateau with bored holes for the sensors; b) cross-section of a hole with an attached sensor.

By selecting the exact number and location of individual probe holes, there are two contradictory requirements; on the one hand it was required to measure the pressure in the whole contact surface, at as many measuring points as possible. On the other hand, the number of holes and their diameter are limited in order not to influence the tension course within the material. Therefore, a compromise consisting of 22 holes spread evenly over the contact area was implemented.

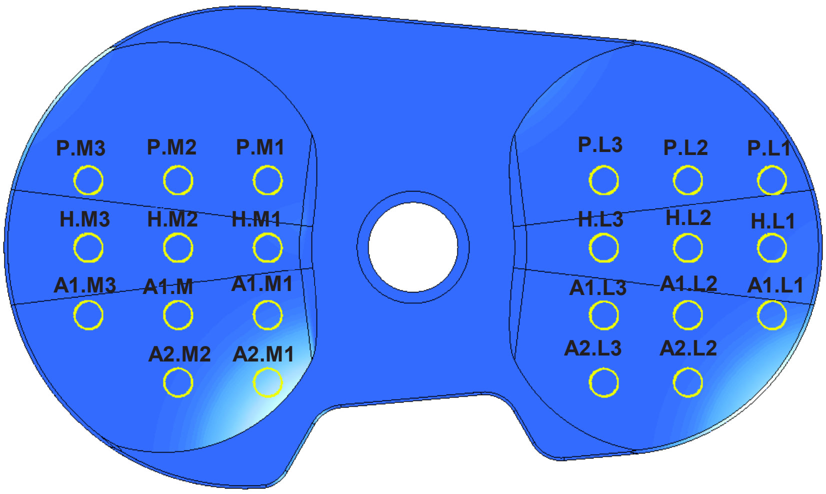


Figure 5. Position of individual holes in the tibial plateau.

For the measurements of the deformation were used monocrystalline semiconductor strain gages with N conductivity that exhibit better, more linear dependency on press deformation; further details about biomechanical measurements are in Volf et al. (2002). The used sensors were developed specially for this measurement by producer VTS Zlín, Czech Republic. This is a semiconductor strain gages of type AP120-2-12/Au/BP with a length of 2 mm.

The strain gages were placed into the prepared holes, glued and covered by silicone. Although they exhibit significant dependency of electrical resistance on the temperature and a non-linear dependency of the measured resistance on the deformation, their negatives are compensated by their accuracy and stress-fatigue resistance. The temperature error of the semiconductor was compensated by small thermometers placed into the holes.

RESULTS AND DISCUSSION

According to the output from the mathematical model by Zach et al. (2004), which is depicted in 3D view in Fig. 6, the maximal calculated pressure value is 7.04 MPa. However, the named peak pressure is concentrated in a very small area, in the close surroundings the pressure drops rapidly. Due to the setup of the experiment, where there was only limited number of probe holes, the exact point with the peak pressure cannot be hit exactly.

The mathematical model presumes static load 2,100 N and flexion angle 0°, the norm ISO 14243-3:2004(E) prescribes varying force and angle values according the actual human step cycle. Therefore, a direct comparison of the obtained data with the model is not possible; the measured values have to be recalculated. The maximal load obtained by the movement simulator is limited due to its construction to 1,730 N and the pressure values calculated by the mathematical model base on the force of 2,100 N; the force also changes during the walk cycle. And secondly, as discovered in previous experiments by Volf et al. (2005), the pressure does not always increase linearly with the force; this is given by the changing geometry of the contact area.

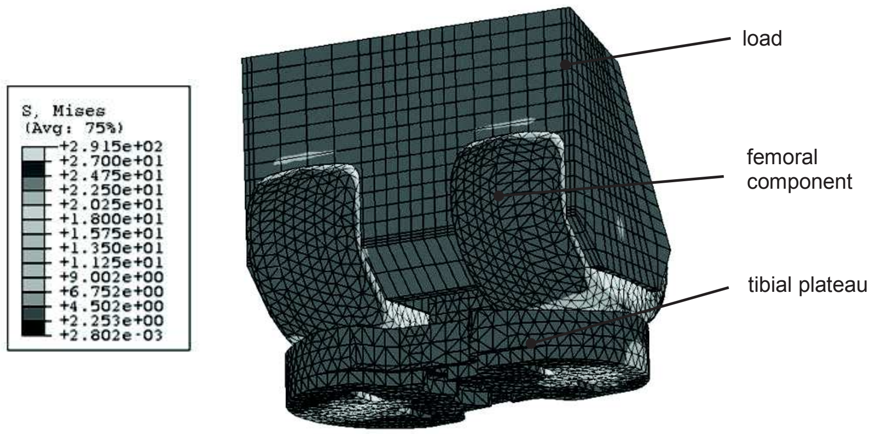


Figure 6. Geometrical model of knee arthoplasty and pressure distribution.

Considering the named limitations – varying flexion angles and exerted forces in the walk cycle vs. defined mathematical model – there have been obtained after recalculation relatively correspondent results. As the peak pressure are could not be matched exactly, the data from the nearby sensors in probe holes H.L2 and P.L3 (see Fig. 5 and Fig. 7) are presented. These recalculated pressure values are about 0.8 MPa, which is in concordance with the model that states the pressure 0.9 MPa for these points. It also has to be noted the maximal pressure values in these probe holes as seen in Figs. 7 and 8 cannot be compared with the model directly; they have to be reduced as explained above. Other probes give similar results as calculated pressure values in those points, too. Thereby the mathematical model can be considered as useful and relevant when designing joint replacements, however its accuracy is partially limited by the changing geometry of the knee joint during walk.

In the second part of our research, we focused mainly on the simulation of the movement of the joint during the walk, as the most natural and important kind of movement. The aim was to study the course of the pressure during the walk. Due to the very complex structure of the knee joint, the pressure changes significantly during the walk cycle. It is caused by combined movement of the joint, i.e. flexion, shift and rotation of the nearby parts, explained further in Zheng & Fleisig (1998). First is presented in Fig. 7 the planar distribution of individual pressure values in all sensors at 70% of the walk cycle (associated flexion is about 57.5°). The exact placement of the sensors can be seen in Fig. 5 above.

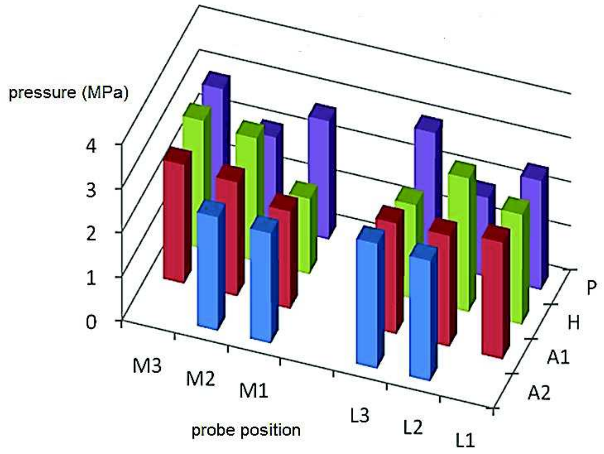


Figure 7. Measured pressure values in individual probe holes in 70% of the walk cycle.

The graph in Fig. 7 shows the planar pressure distribution over the contact area of the joint under the named conditions. The measured pressures vary from cca. 1 MPa to 2.5 MPa, the calculated pressure peak point lies between the sensors H.L2 and P.L3. The pressures are relatively high due high flexion and associated geometrical and force changes in the joint. Similar charts have been created for any individual stage of the walk according to the ISO norm, and the corresponding results are summarised in Fig. 8, which shows the course of pressure in individual probes during the walk cycle. Although the data have been measured individually for each stage of the walk cycle, because of lucidity they are connected with curves; each curve represents a data row from one sensor.

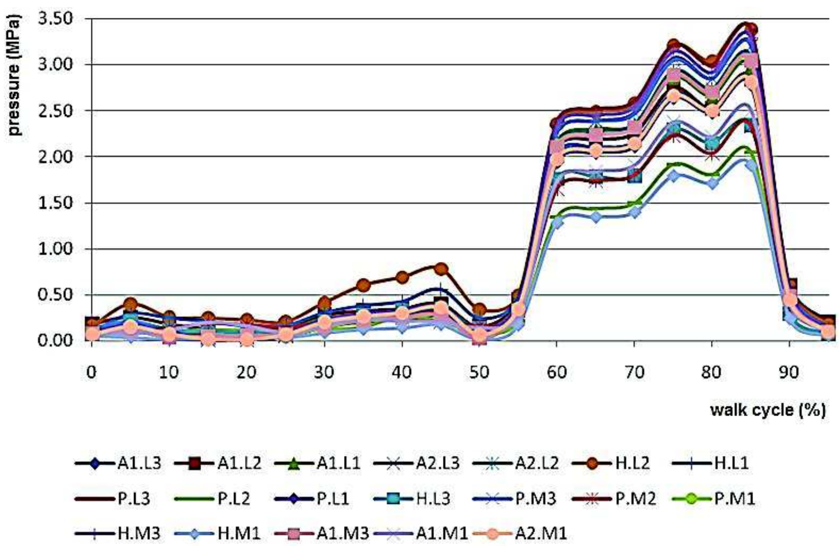


Figure 8. Measured pressure values in individual probe holes in dependency on walk cycle.

The course of pressure indicates a rapid raise at 60% of the walk cycle and a further raise at 80% of the cycle, with the peak pressure of 3.4 MPa (probe P.L3) that is the closest one to the calculated origin of force. In these stages of walk cycle the flexion increases significantly, which changes the geometry of the joint. The peak values are caused predominantly by anterior-posterior shift and by tibia rotation, which cause additional moment acting on the joint; for more detailed information see Taylor et al. (2004) and Miller & Zhang (2001).

Although the greatest load is according to ISO 14243-3:2004(E) exerted at 45% of the walk cycle (2,433.5 N), the pressure is way below its maximum at this stage. The deformation and associated pressure are compensated by low flexion angle (8.13°) and low anterior-posterior shift and tibia rotation. On the contrary, at 80% of the walk cycle, the exerted force according to the ISO norm is only 167.6 N, but the flexior reaches 47.08°, associated anterior-posterior shift is 2.38 mm and tibia rotation is 4.92°. Therefore, the peak pressure is influenced predominantly by overall kinematic changes of the joint rather than size of the force, where the flexion, shift and rotation are reflected.

According to the mathematical model, the highest calculated pressure value was 7.04 MPa, and the highest measured pressure value was 3.4 MPa, see Fig. 8. The determination of the exact point with the highest pressure would require different experiment setup and it was not the goal of our work; the aim was to determine the pressure over the entire contact area of the joint replacement components, to compare the measured values with the mathematical model and to analyse the pressure change during the walk. However, none of these values approaches to the limit stress 13 MPa of the polymer material UHMWPE from which the tibial plateau is made.

CONCLUSIONS

Pressure distribution in the knee arthoplasty – an artificial knee replacement joint, using a knee movement simulator was measured, according to the ISO 14243-3:2004(E) norm and using specific pressure sensor placement. We focused on the comparison of the measured pressure values with mathematical models developed by Zhu & Chen (1999) and calculations performed by Donát (1997) and Zach et al. (2004). The measured pressure values were in correspondence with the mathematical model given above; this mathematical model describes, with some limitations, accurately the load of knee arthroplasty and that such model can be used to design artificial joints.

Further the change of the pressure distribution in the contact area during the walk was examined. To simulate the walk a knee movement simulator was used according the norm ISO 14243-3:2004(E), which prescribes exactly the flexion and associated force values during the individual stages of the walk. Hereby were confirmed significant changes of the pressure during the walk that is given by the changing geometry of the joint.

Neither the calculated nor the measured value exceeds the limit stress of the polymer joint replacement material. Finally, no one of the sensors exhibited unexpected or unacceptable pressure peak or significant deviation from the mathematical model that could endanger the functionality of the artificial joint, and the limit stress of the material was not exceeded.

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