Patellar kinematics after knee arthroplasty: experimental validation of a numerical model

Latypova A. [1], Arami A. [2], Pioletti D.P. [1], Jolles B.M. [3], Aminian K.[2], Terrier A.[1]

[1] Laboratory of Biomechanical Orthopedics, Ecole Polytechnique Fédérale de Lausanne, Switzerland

[2] Laboratory of Movement Analysis and Measurment, Ecole Polytechnique Fédérale de Lausanne, Switzerland

[3] Department of Orthopedics and Traumatology, CHUV and University of Lausanne, Switzerland

Abstract- Subject-specific modelling of Total Knee Arthroplasty could be an efficient method to preoperatively evaluate surgical options. In particular, the question of the necessity of patellar resurfacing is still a debatable issue. The aim of this work was to validate a numerical model of Total Knee Arthroplasty using an instrumented robotic knee simulator equipped with kinematic sensors. The patellar kinematics during a loaded knee flexion were measured with the knee simulator and compared with the values predicted by the model. The mean absolute difference between measured and predicted patellar translations and rotations was respectively 1.4 mm and 2.4^{0}. This numerical model will be later used for subject-specific predictions of patellar kinematics and strain state after Total Knee Arthroplasty.

Keywords - knee, TKA, knee simulator, kinematics, numerical modelling

1. Introduction

Despite relatively high rate of success of Total Knee Arthroplasty (TKA), patients still face postoperative complications, especially related to the patellar resurfacing. Osteonecrosis, patellar fracture, implant failure, polyethylene wear, extensor mechanism rupture, and anterior knee pain remain one of the most disputable and currently unsolved issues associated to patellar resurfacing in TKA.

Changes in patellar kinematics and strain after TKA are suggested to be a cause of complication development. However, it is difficult to measure patellar kinematics *in vivo*, and impossible to measure patellar strain *in vivo*. Therefore, numerical modelling can be an alternative to predict preoperatively patellar mechanics after TKA, in order to diagnose patient-specific complications.

Several numerical models of TKA are predicting patellar kinematics, but only a few focused on patellar strain. Currently, the most advanced numerical model of TKA for prediction of patellar strain includes the tibiofemoral joint, the patellofemoral joint, the quadriceps muscle, and the most important soft tissues attached to the patella [1]. This model, however, constrain the kinematics of the femur and the tibia, as the muscle forces.

Therefore, we have developed an alternative numerical model of TKA with less kinematics and force constrains [2]. The model is decoupled in two levels: joint and bone. At the joint level, the patellar kinematics is predicted by using a control algorithm for the muscular activation. At the bone level, patellar strain is predicted using boundary conditions obtained from the joint level. This numerical model will be later extended for patient-specific clinical applications. The goal of the present study was to validate the joint model using an instrumented robotic knee simulator.

2. Material and Methods

Numerical model

The joint model was developed in Abaqus/Standard (Simulia, Providence, RI) [2]. The model included the femur, the tibia, the patella and the four parts of the quadriceps muscle (Fig. 1.a). The bone geometry was reconstructed by the segmentation of a cadaveric CT scan. A cemented ultra-congruent mobile-bearing knee prosthesis (FIRST, Symbios, Switzerland) was inserted into the bones according to manufacturer recommendations. The femur, the tibia, the patella, the cement and the metallic components were assumed rigid. The polyethylene tibial insert and patellar components were considered as linear elastic materials. A loaded squat movement was controlled by a feedback algorithm, through the elongation of the vastus intermedius, and the forces of the rectus femoris, vastus lateralis and vastus medialis. The model predicted the kinematics of the patella and the forces (quadriceps and patellar tendons, patellofemoral contact) acting on it. To validate the joint model, it was adapted to the following robotic knee simulator.

Robotic knee simulator

A knee simulator has been designed based on a main hydraulic load unit (MTS Bionix® MTS Bionix Kinematic Knee Simulator) and additional prismatic actuators. The simulator includes parts replicating the femur, the tibia, the patella and the quadriceps muscle (Fig. 1.b). The hip is attached to a load unit, a long-stroke axial actuator with 15 kN maximum axial force and 457 mm stroke that simulates body weight and gross motion. It allows the hip to rotate and translate in the sagittal plane. The ankle is attached to the ground and can only rotate in the sagittal plane. The hip and the ankle are aligned. A smaller actuator is attached to the femoral part and replicates the quadriceps force (maximum force of 4.5 kN retraction, 5.8 kN extension and +/- 75mm stroke). The quadriceps and patellar tendons are represented as a solid webbing strap attached to the patella and linking the quadriceps actuator and the tibia. The simulator is instrumented with an ultra-congruent mobile-bearing knee prosthesis (FIRST, Symbios, Switzerland). The system is controlled by MTS FlexTest software, which provides control, data acquisition, and multi-axial testing precision.

The knee simulator replicated a loaded (bodyweight = 600 N) squat movement from 10 to 60° of flexion. The knee flexion was controlled by the quadriceps and hip forces. For each angle of flexion, a constant load was applied on the hip. The simulator searched for equilibrium through a control of the quadriceps actuator force with a help of a custom-designed self-learning algorithm [3]. The muscle force was measured by a force sensor (MTS 661.19F-01) attached to the quadriceps actuator. Another force sensor (MTS 661.19F-03) was used to measure reaction force in the ankle.

The patellar kinematics was measured via a motion capture system including four Mx3+ cameras (Vicon, UK) and 18 reflective markers (8 on the femur, 6 on the tibia and 4 on the patella). During the simulated movement, the position of the markers was recorded after the system reached equilibrium at the set angle. At least 1000 frames per angle were used to record a position of markers. The averaged measurements were used in the analyses.

Simplified model

The TKA joint model was simplified to replicate the robotic simulator (Fig. 1.c). The geometry and position of the components were extracted from the CAD model of the knee simulator. The 4 parts of the quadriceps muscle were replaced by one, modelled as 2 parallel connector elements. The quadriceps tendon was modelled as a fiber reinforced 3D elastic band (E=100 MPa, ν =0.3). And the patellar tendon was modelled as 2 parallel fibers with the same properties.

The model replicated loaded (bodyweight = 600 N) squat movement from 10 to 60° of flexion. During the movement, the ankle could only rotate in the sagittal plane, while the hip could rotate and vertically translate in the sagittal plane. The patella had all degrees of freedom. The knee flexion was controlled by the quadriceps elongation. For each angle of flexion, the model calculated the muscle force using a feedback algorithm aimed on maintaining system in equilibrium.

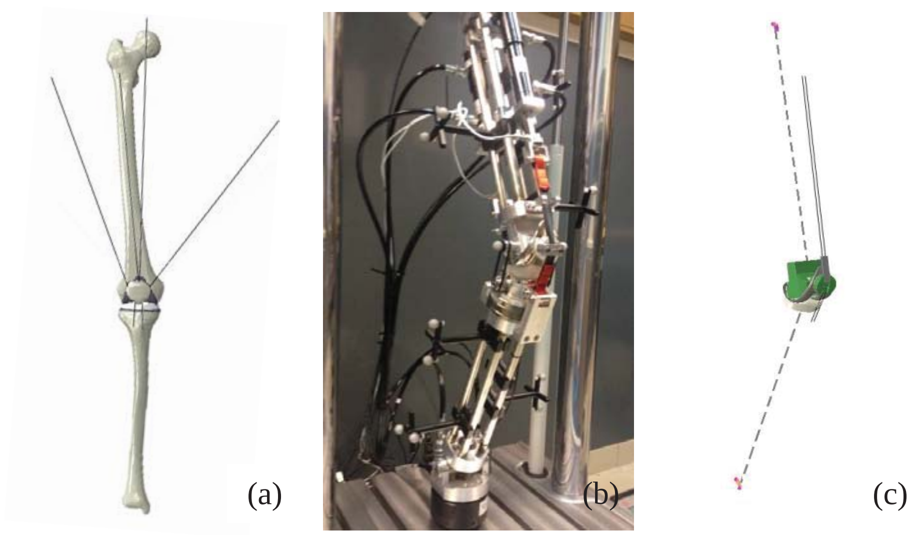

Figure 1. Numerical TKA model (a), knee simulator(b) and simplified TKA model(c).

Model validation

The patellar translations and rotations were measured with the robotic knee simulator and compared to the predictions of the simplified numerical model. The patellar kinematics was measured as the motion of patellar coordinate system (P-XYZ) relative to the femur coordinate system (F-XYZ). F-XYZ was defined as following: the X-axis was parallel to the axis of flexion of the femoral component; Y-axis passed through the center of rotation of the femoral component and the center of rotation of the femoral head; Z-axis was orthogonal to X and Y axes; the origin of F-XYZ was set at the geometric center of the femoral component. P-XYZ was defined as following: X and Y axes were parallel to the flat part of the patellar component, and X-axis was perpendicular to patellar tendon at full knee extension; Z-axis was orthogonal to X and Y axes; the origin of P-XYZ was set at the center of the flat part of the patellar component.

The patellar translations were defined as the coordinates of the origin of P-XYZ in F-XYZ. The patellar flexion was defined as the angle between the axes F-Y and P-Y projected on the plane F-XY. The patellar rotation was defined as the angle between the axes F-X and P-X projected on the plane F-XY. The patellar tilt was defined as the angle between the axes F-X and P-X projected on the plane F-XZ.

We evaluated the minimum, maximum and average difference between the measured and predicted values, throughout the knee flexion.

3. Results

In the simulator, the patella shifted laterally, reaching 1.3 mm at 27° of flexion (Fig. 2). The lateral shift of the patella predicted by the model reached 0.9 mm, also at 27° of flexion. The minimum, maximum and average differences were respectively 0, 0.5, and 0.2 mm. The patella translated inferiorly from 16.5 to 3.8 mm in the simulator and from 18.5 to 6.5 mm in the model. The minimum, maximum and average differences were respectively 1.8, 2.8, and 2.5 mm. The patella translated posteriorly from 41.4 to 35.8 mm in the simulator and from 42.8 to 37.3 mm in the model. The minimum, maximum and average differences were respectively 1.2, 2.1, and 1.5 mm.

In the simulator, the patella tilted medially, reducing the tilt angle from approximately 8° to 1° (Fig. 2). In the model, the patellar tilt angle grew from 8 to 10° between 10 and 30° of flexion, and then gradually reduced to 4° (Fig.3). The minimum, maximum and average differences were respectively1.1, 5.3 and 3.8°. The measured patellar flexion angle grew linearly from 8° to 32°, while the predicted angle increased from 10° to 30°. The minimum, maximum and average differences were respectively 0.4, 3.9, and 2.1°. The measured patellar rotation angle remained around 2.6° during all movement. The predicted patellar rotation angle remained around 1.0°. The minimum, maximum and average differences were respectively 0.3, 1.8, and 1.2°.

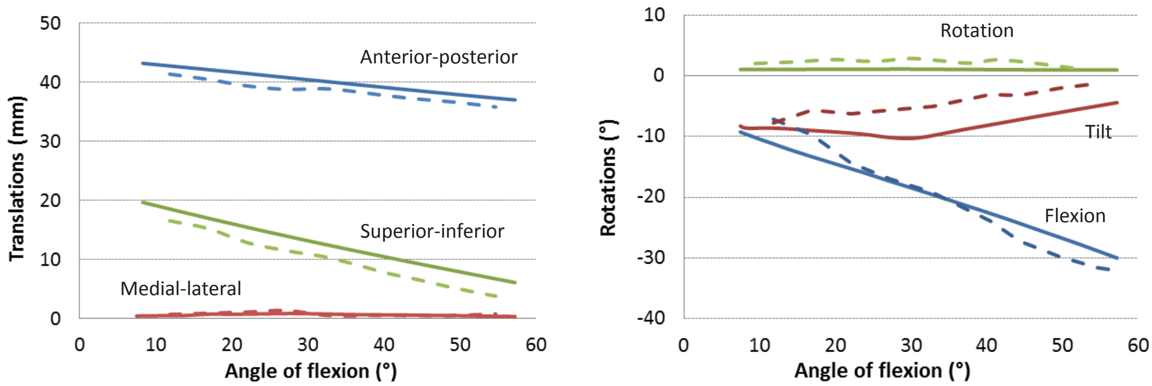

Figure 2. Predicted (solid) and measured (dashed) patellar translations and rotations.

4. Discussion

In this work, we presented a validation of a numerical TKA model using a robotic knee simulator. The patellar kinematics (translations and rotations) were measured with the knee simulator during a loaded knee flexion, and compared to the values predicted by the numerical model.

The results showed good match between experimental data and the model predictions. The maximum difference between measured and predicted values was observed for patellar tilt (5.3°) and for anterior-posterior translation (2.8 mm). Since only one actuator (quadriceps) was available in the simulator, the patella became unstable at low degrees of flexion. Slight difference of patellar attachment and its position in the model and the simulator could explain the discrepancy between measured and predicted results. The attachment of the reflective markers could also lead to a systematic error. The quadriceps force in the simulator was consistent with the predicted one (mean relative difference 7.8%), and in agreement with reported values [4].

As the continuation of the work, additional measurements and predictions are planned for different patellar positions (patella alta, baja), femoral component positions. We will also improve the attachment of the reflective markers to the patella, and measure more precisely the initial position of the patella and the patellar tendon.

The strength of this work was to propose and validate an original TKA model, which controls the knee flexion through the quadriceps muscle activation. The quadriceps force is calculated during the movement and depends on the angle of the flexion, and on subject-specific data such as the anatomy, weight, knee prosthesis, and surgical technique. It allows to reduce the model's constrains and to increase the accuracy of the predicted subject's patellar strain. This validated subject-specific model of TKA could help in improving surgical planning of TKA and reducing the associated complications.

5. Acknowledgment

This study was supported by the Inter-Institutional Center for Translational Biomechanics (EPFL-CHUV-DAL), and Symbios, Switzerland.

6. References

[1] Fitzpatrick, C.K., M.A. Baldwin, A.A. Ali, P.J. Laz, and P.J. Rullkoetter, Comparison of patellar bone strain in the natural and implanted knee during simulated deep flexion. J Orthop Res, 2011. 29(2): p. 232-9.

[2] Latypova, A. et al, A musculoskeletal numerical model of total knee replacement with the focus on the patella modelling, V International Conference on Computational Bioengineering, Leuven, Belgium, 2013

[3] Arami, A., Javan-Roshtkhari, M., and Lucas, C., A fast model free intelligent controller based on fused emotions: a practical case implementation, 16th Int. IEEE Medit. Conf. on Control and Auto., Ajacsio, France, 2008, p. 596-602.

[4] Frohm, A. et al, Patellar tendon load in different types of eccentric squats. Clinical Biomechanics, 2007. 22: p. 334-341.