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How exactly can computer simulation predict the kinematics and contact status after TKA? Examination in individualized models

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ABSTRACT

Background: It is unknown whether a computer simulation with simple models can estimate individual in vivo knee kinematics, although some complex models have predicted the knee kinematics. The purposes of this study are first, to validate the accuracy of the computer simulation with our developed model during a squatting activity in a weight-bearing deep knee bend and then, to analyze the contact area and the contact stress of the tri-condylar implants for individual patients.
Methods: We compared the anteroposterior (AP) contact positions of medial and lateral condyles calculated by the computer simulation program with the positions measured from the fluoroscopic analysis for three implanted knees. Then the contact area and the stress including the third condyle were calculated individually using finite element (FE) analysis.
Findings: The motion patterns were similar in the simulation program and the fluoroscopic surveillance. Our developed model could nearly estimate the individual in vivo knee kinematics. The mean and maximum differences of the AP contact positions were 1.0 mm and 2.5 mm, respectively. At 120° of knee flexion, the contact area at the third condyle was wider than the both condyles. The mean maximum contact stress at the third condyle was lower than the both condyles at 90° and 120° of knee flexion.
Interpretation: Individual bone models are required to estimate in vivo knee kinematics in our simple model. The tri-condylar implant seems to be safe for deep flexion activities due to the wide contact area and low contact stress.

1. Introduction

Total knee arthroplasty (TKA) is a successful procedure for the patients with end stage knee osteoarthritis to achieve the goals of relieving pain and restoring function (Nakamura et al., 2010a, 2014a). However, the polyethylene wear has been one of the main reasons for revision surgery after TKA, which might cause osteolysis and subsequent component loosening (Casey et al., 2007; Reay et al., 2009). In addition, in posterior-stabilized (PS) TKA, the contact status of the post-cam should be considered because high contact stress at the post-cam area could be the reason for the polyethylene wear and the post fracture (Clarke et al., 2004; Mauerhan, 2003; Puloski et al., 2001). Evaluation of the contact stress on the polyethylene insert including the post-cam area might help to avoid the problems that occur after TKA.

Previously, loading simulators and robotic systems using cadaver knees have been commonly used to analyze the contact status, the force and the contact stress between femoral and tibial components (Akagi et al., 2002; Borque et al., 2015). Recently, simple modelling using KneeSIM (LifeMOD/KneeSIM 2010; LifeModeler Inc., San Clemente, CA, USA) with only one muscle bundle of quadriceps and one bundle of hamstrings has been developed, which can analyze knee kinematics and kinetics in detail in various loading conditions such as tibial component rotation, and posterior tibial slope (Kuriyama et al., 2014, 2015, 2016; Okamoto et al., 2015). In the previous studies using KneeSIM, the bone model was not individualized, and a preexisting bone model was used for the analysis (Ishikawa et al., 2015; Kuriyama et al., 2014, 2015, 2016; Okamoto et al., 2015). So far, there is little information as to whether these simulation studies accurately represent in vivo status, although some complex models with many muscle fibers and ligament fibers have predicted the knee kinematics (Fitzpatrick et al., 2014; Marra et al., 2015). If the contact status of the femorotibial joint could be evaluated for individual patients in more detail, a more accurate prediction could be made thus preventing complications.

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The tri-condylar implant was developed to fit an Asian lifestyle requiring frequent deep flexion activities (Akagi et al., 2000; Nakamura et al., 2010a, 2010b). This implant has a unique design characteristic with a ball and socket joint as a third condyle in the mid-posterior portion of the femoral component and the polyethylene insert. This third condyle replaces the function of the post-cam mechanism and induces femoral rollback (Nakamura et al., 2014b, 2014c). So far, the contact stress and the contact area on the third condyle in actual patients are still unknown.

The purposes of this study are first, to validate the accuracy of the simulation programs and then, to analyze the contact area and the contact stress of the tri-condylar implants for individual patients. Our hypotheses were that the simulation program could approximate the actual contact positions and that the tri-condylar implants had a wide contact area and a low contact stress at the third condyle during deep knee flexion.

2. Materials and methods

In the current study, three knees were implanted with the tri-condylar prosthesis, and were available for analysis (Table 1). The tri-condylar implant was the Bi-Surface Knee System (Kyocera Medical, Osaka, Japan). As described above, this implant has a ball and socket joint instead of commonly used post-cam mechanism, which allows larger rotational freedom and shares the load together with the medial and lateral condyles during deep flexion to improve knee flexion and long term durability (Nakamura et al., 2014b, 2014c, 2015b). All surgeries were carried out by a single surgeon with a measured resection technique. The posterior cruciate ligament was sacrificed, and the ligament balance was adjusted equally to coordinate medial-lateral laxity and flexion-extension gap as much as possible.

A three dimensional (3D) bone model was constructed for each patient from preoperative computed tomography (CT) images as a parasolid model, because the exact shape of the femur and tibia could not be determined from the postoperative CT images due to halation of the implant. The CT data from the pelvis to the foot was imported to MIMICS (Materialize HQ, Leuven, Belgium). Postoperative X-ray image and CT were referred for implantation for each patient. Femoral, tibial, and patellar components were implanted to coincide with sagittal, coronal, and axial alignment. In KneeSIM program, parasolid geometry was required for analysis. Parasolid models of the bone and implant were imported into the program.

We developed our model based on KneeSIM, which includes femorotibial and patellofemoral contact, lateral collateral ligament (LCL), medial collateral ligament (MCL), elements of the knee capsule, the quadriceps muscle and tendon, patellar tendon, and the hamstring muscles. A squatting activity in a weight-bearing deep knee bend was simulated using the developed model which was established according to an Oxford-type knee rig. In the current study, a constant vertical force of 4000 N was applied at the hip and loaded on the bicondylar joint of the knee. The hip joint was modeled as a revolute joint parallel to the flexion axis of the knee and was allowed to slide vertically, but constrained in the mediolateral and anteroposterior directions. The ankle joint was modeled as a combination of several joints that combined to allow free translation in the medial-lateral direction and free rotation in flexion, axial, and varus-valgus directions. A closed-loop controller monitored knee flexion and compared it with the prescribed input. Quadriceps and hamstrings loads were adjusted accordingly to obtain the prescribed flexion angle at each time point. The knee model was flexed in the same range of motion of each patient.

Table 1
Patient demographics.

	Case 1	Case 2	Case 3
Age (years)	73	85	75
Sex	Female	Male	Female
Diagnosis	Osteoarthritis	Osteoarthritis	Osteoarthritis
Time point assessed after surgery (months)	11	12	18
Femoro-tibial angle (degrees)	175	170	173
Flexion (degrees)	120	120	90
Extension (degrees)	0	0	0

The origins of the insertion points of MCL and LCL were determined for each bone model from the relevant anatomical studies (LaPrade et al., 2005; Park et al., 2005; Wijdicks et al., 2010). The MCL comprised 2 bundles, which consisted of the anterior and posterior bundles. All ligament bundles were modeled as nonlinear springs with material properties obtained from a published report (Blankevoort et al., 1991). Stiffness coefficients of the LCL, MCL-anterior, and MCL-posterior were determined as 59, 63, and 63 N/mm, respectively (Anderson et al., 2012; Harner et al., 1995; Kennedy et al., 2013; Sugita and Amis, 2001; Wijdicks et al., 2010). Medial and lateral gaps were measured in axial radiographs, and the amount of each gap was defined as the slack of the MCL and LCL (Kanekasu et al., 2005; Tokuhara et al., 2006). In the simulation program, relative positions of the femoral and tibial components were calculated, and then the anteroposterior (AP) contact positions for medial and lateral condyles were identified with respect to the tibial component at 30° increments of knee flexion. AP contact positions were defined as the medial and lateral femorotibial lowest points, which were based on the distance between the plane of the tibial tray (the base of the tibial insert) and the respective femoral condyles. The femorotibial contact forces were computed by our developed models during deep knee bending, at 30° increments of knee flexion.

Each patient was asked to perform a weight-bearing deep knee bend activity under fluoroscopic surveillance. Kinematic measurements were performed at 30° increments of knee flexion. Using a previously reported 3D to two-dimensional registration approach, AP contact positions were determined from a single-perspective fluoroscopic image (Mahfouz et al., 2003). An error analysis for this process has been previously performed for TKA components, documenting a translational error of less than 0.5 mm and a rotational error of less than 0.5° (Mahfouz et al., 2003). The absolute values of the differences in AP contact positions of both methods were averaged at all flexion angles to assess the accuracy of our developed model.

The calculated position of the components and the magnitude and direction of each force using the musculoskeletal model at 30° increments of knee flexion were further used to predict the contact stress and area using finite element (FE) model. FE simulations were performed using an ANSYS Workbench (ANSYS, Inc., Canonsburg, PA, USA). The femoral component was modeled as a linear elastic body. The Young's modulus of the femoral component was set at 240 GPa, which is consistent with data for Co-Cr-Mo alloy femoral components. The tibial insert, which is consisting of ultra-high molecular weight polyethylene, was modeled as a nonlinear elastoplastic body. A true stress versus strain curve of GUR 4150 HP was obtained from the previous study because of the similarity of the molecular weight (Kurtz et al., 1999), which was imported to FEA. Poisson's ratio was set at 0.46. The thickness of the tibial insert was 11 mm, 11 mm and 15 mm for case 1, 2 and 3, respectively. Concerning mesh sensitivity analysis, the first analysis was performed at element size of 1.0 mm in a representative model. Afterwards the element size was decreased by 20% and the analysis was repeated. Finally the element size was applied when the change of maximum stress provided by the analysis in a representative model converged in less than 5%. The mesh of the femoral component and the tibial insert were generated based on 10-node quadratic tetrahedral elements sized at 0.8 mm. The generated mesh contained a total of 221,317, 254,854 and 216,148 nodes for the femoral condyle and 112,456, 125,902 and 128,100 nodes for the tibial insert, as a result of 136,855, 157,525 and 133,523 total elements for femoral condyle and 70,085, 78,577 and 80,165 total elements for tibial insert for case 1, 2 and 3, respectively. Connections were defined by using ‘contacts’ tool