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Effects of posterior condylar offset and posterior tibial slope on mobile-bearing total knee arthroplasty using computational simulation

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ABSTRACT

Background: Postoperative changes of the femoral posterior condylar offset (PCO) and posterior tibial slope (PTS) affect the biomechanics of the knee joint after fixed-bearing total knee arthroplasty (TKA). However, the biomechanics of mobile-bearing is not well known. Therefore, the aim of this study was to investigate whether alterations to the PCO and PTS affect the biomechanics for mobile-bearing TKA.
Methods: We used a computational model for a knee joint that was validated using in vivo experiment data to evaluate the effects of the PCO and PTS on the tibiofemoral (TF) joint kinematics, patellofemoral (PF) contact stress, collateral ligament force and quadriceps force, for mobile-bearing TKA. The computational model was developed using ± 1-, ± 2- and ± 3-mm PCO models in the posterior direction and $-3°$, $0°$, $+3°$, and $+6°$ PTS models based on each of the PCO models.
Results: The maximum PF contact stress, collateral ligament force and quadriceps force decreased as the PTS increased. In addition, the maximum PF contact stress and quadriceps force decreased, and the collateral ligament force increased as PCO translated in the posterior direction. This trend is consistent with that observed in any PCO and PTS.
Conclusions: Our findings show the various effects of postoperative alterations in the PCO and PTS on the biomechanical results of mobile-bearing TKA. Based on the computational simulation, we suggest that orthopaedic surgeons intraoperatively conserve the patient's own anatomical PCO and PTS in mobile-bearing TKA.

1. Introduction

Total knee arthroplasty (TKA) is the most popular surgical technique for end-stage degenerative osteoarthritic disease in the knee joint [1]. Functional restoration is the first and most important purpose of TKA, as well as good pain relief and implant survival after surgery [2–4]. Many studies have been conducted to determine which variables influence the functional outcomes

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including the maximal flexion post-TKA [5–7]. The maximal flexion after TKA is influenced by many factors including patient factors, prosthesis design, surgical technique and rehabilitation. For patient factors, preoperative flexion is known to mainly influence the postoperative flexion [8].

Most surgeons have investigated methods to control certain predictable intraoperative factors for better knee motion and functionality, and it has been reported that surgical-related factors are important when attempting to restore knee flexion [9–11]. With respect to predictable intraoperative factors, there have been some studies evaluating the effects of alterations to kinematic parameters such as the posterior condylar offset (PCO), joint line and posterior tibial slope (PTS) on the maximal postoperative flexion [11–14].

Previous studies have concluded that increasing the PCO and PTS using a more posterior tibiofemoral (TF) contact point can increase the range of motion (ROM) in cruciate-retaining (CR) TKA [11, 12, 15]. However, it was found that the effects of these kinematic factors vary with the type of TKA design [16, 17]. In a fixed-bearing posterior cruciate ligament (PCL) sacrificing (PS) type, increased PCO and PTS were not shown to provide better ROM [17–19]. On the other hand, other studies showed that PCO and PTS played an important role in the optimization of active knee flexion during activity after PS TKA. In a mobile-bearing PS type, there was also no correlation found between PCO and ROM in a short-term follow-up [20]. However, a different study showed that there was a direct relationship between ROM and PCO and PTS in mobile-bearing TKA [21]. Therefore, consistent studies on the different effects of PCO and PTS on the maximal postoperative flexion in different implant types should be conducted. Theoretically, a mobile-bearing TKA has been considered to have a more anatomical contact point which provides less risk of impingement, thereby providing better kinematics and greater ROM, as well as a better Knee Society score [5, 11, 14, 15]. However, the biomechanical effect on mobile-bearing TKA with respect to changes in PCO and PTS has not yet been fully understood. The contact stress in the patellar button, the ligament force, the quadriceps muscle force, and TF (polyethylene (PE) insert) joint contact point are challenging to directly measure and evaluate; however, this can be achieved using a finite element (FE) analysis.

The aim of this study was to determine the biomechanical effects of the change in PCO, and the corresponding change in PTS, for mobile-bearing TKA under deep-knee-bend conditions. We analysed the displacement of the TF contact point, the patellofemoral (PF) contact stress, the force in the collateral ligaments and the quadriceps muscle forces. We hypothesized that increasing posterior translation PCO and PTS decreased quadriceps force for mobile-bearing TKA.

2. Materials and methods

2.1. FE model

A previously validated healthy knee joint FE model was used for this study [22–24]. A three-dimensional (3D) non-linear FE model of a normal knee joint was developed using data from computed tomography (CT) and magnetic resonance imaging (MRI) scans of a healthy 37-year-old male subject.

The reconstructed CT and MRI models were combined with a positional alignment of each model using commercial software (Rapidform version 2006; 3D Systems Korea, Inc., Seoul, South Korea) for modelling bone structures as rigid bodies using four-node shell elements [25]. Additionally, the major ligaments were modelled using nonlinear and tension-only spring elements [26, 27]. The ligament insertion points were applied with respect to the anatomy from the MRI sets of the subject and descriptions based on previous studies (Figure 1) [28–30]. The interfaces between the cartilage and bones were modelled as fully bonded. The contacts between the femoral cartilage and meniscus, meniscus and tibial cartilage, and femoral cartilage and tibial cartilage were modelled for both the medial and lateral sides, which resulted in six contact pairs [24]. Convergence was defined as a relative change of <5% between two adjacent meshes. The average element size of the simulated cartilage and menisci corresponded to 0.8 mm.

To develop models for the changes in PCO and the corresponding PTS, two experienced surgeons (S.K.K. and Y.-G. Koh) applied a surgical simulation of a TKA. Surgical simulation was performed by using Unigraphics NX (Version 7.0; Siemens PLM Software, Torrance, CA, USA). Computer-assisted design models of a mobile-bearing TKA from LCS (Johnson & Johnson — DePuy Orthopaedics, Inc., Warsaw, IN, USA) were virtually implanted into the bone geometry. A large-size femoral component and size 4 tibial baseplate for mobile-bearing TKA were selected based on the dimensions of the femur and tibia, respectively. A total of 28 configurations were considered in this study (Table 1).

The femoral component was aligned in a neutral position, such that the distal bone resection was perpendicular to the mechanical axis of the femur, and the anterior and posterior resections were parallel to the clinical epicondylar axis in the transverse plane. A PCO model identical to the original subject was developed, followed by the modified PCO model. The femoral component position was adjusted in the anterior–posterior (AP) direction to avoid notching of the anterior cortex as per the standard surgical protocol. Seven models were developed with 0 mm, ±1 mm, ±2 mm and ±3 mm in the posterior direction (Figure 2).

The default tibial alignment was rotated by 0° in relation to the anterior–posterior axis, the coronal alignment was 90° to the mechanical axis, and the sagittal alignments were −3°, 0°, +3° and +6° to the posterior slope, with an eight-mm resection below the highest point of the lateral plateau (Figure 3). This corresponds to the lowest point of the PE insert articular surface adjacent to the lowest points of the femoral articular surfaces during extension.

Contact conditions were applied between the femoral component, PE insert, and patellar button in the TKA. The coefficient of friction between the PE material and metal was selected as 0.04 for consistency with the explicit FE models proposed in previous studies [30]. The femoral component, PE insert, tibial component and bone cement were made of a cobalt chromium alloy (CoCr),

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