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Original Article

Kinematic analysis of anterior cruciate ligament reconstruction in total knee arthroplasty

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Abstract: Background: This study aims to retain normal knee kinematics after knee replacement surgeries by reconstructing anterior cruciate ligament during total knee arthroplasty. Method: We use computational simulation tools to establish four dynamic knee models, including normal knee model, posterior cruciate ligament retaining knee model, posterior cruciate ligament substituting knee model, and anterior cruciate ligament reconstructing knee model. Our proposed method utilizes magnetic resonance images to reconstruct solid bones and attachments of ligaments, and assemble femoral and tibial components according representative literatures and operational specifications. Dynamic data of axial tibial rotation and femoral translation from full-extension to 135 were measured for analyzing the motion of knee models. Findings: The computational simulation results show that comparing with the posterior cruciate ligament retained knee model and the posterior cruciate ligament substituted knee model, reconstructing anterior cruciate ligament improves the posterior movement of the lateral condyle, medial condyle and tibial internal rotation through a full range of flexion. The maximum posterior translations of the lateral condyle, medial condyle and tibial internal rotation of the anterior cruciate ligament reconstructed knee are 15.3 mm, 4.6 mm and 20.6 at 135 of flexion. Interpretation: Reconstructing anterior cruciate ligament in total knee arthroplasty has been approved to be an more efficient way of maintaining normal knee kinematics comparing to posterior cruciate ligament retained and posterior cruciate ligament substituted total knee arthroplasty.

Keywords: Kinematics, total knee arthroplasty, computer simulation, anterior cruciate ligament reconstruction

Introduction

Total Knee Arthroplasty (TKA) is one of the most successful treatments in orthopaedic surgeries [1]. The long-term survival rate of TKA has been reported at over 90% at 10-years [1, 2]. However, good or excellent clinical outcomes with endurable longevity does not correlate well with patients satisfaction [3]. Related study [4] demonstrated that over 50% of patients in North American were reported to be having difficult in performing high knee flexion activities such as kneeling and squatting which are routine but important movements in people's normal life. Patients are seeking for better functional restoration to maintain normal daily activities and good life qualities after TKA.

Surgeons and researchers are paying more attention to better results after TKA. Recently there are many research efforts having been done on TKA kinematics to disclose the relationships between patient function and artificial implants. Related studies [5, 6] demonstrated that currently many prostheses have been used in clinical practices. However, inadequate femoral rollback and insufficient tibial internal rotation caused by replacing natural knees with artificial implants may result in significantly in-consistent with normal performance in postoperative kinematics. Normal knee kinematics is crucial for patients to achieve functional activities in daily life. For example, greater femoral rollback would facilitate higher knee flexion [7] and flexible tibial rotation permits deep-flexion knee postures, such as squatting and kneeling. Therefore, to achieve superior clinical and functional results, it is important to improve the TKA design to retain natural knee mechanics.

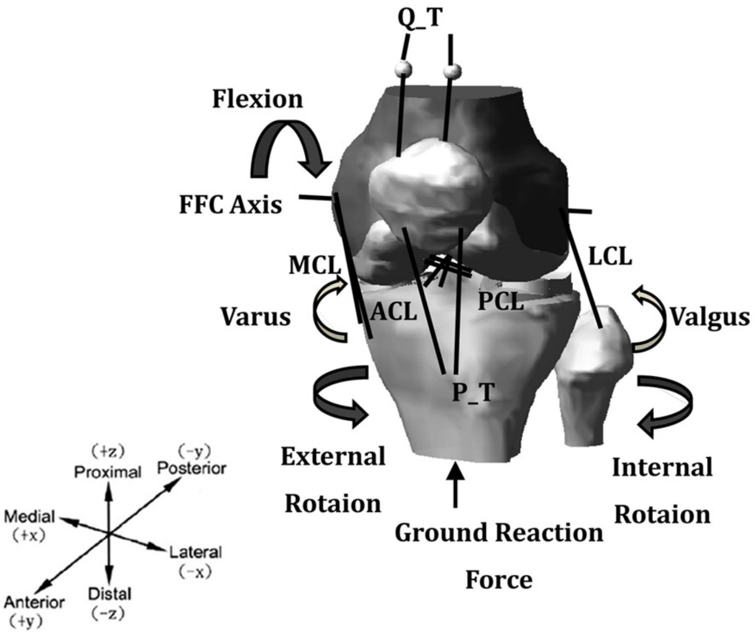

Figure 1. Dynamic intact knee model developed in MSC. ADAMS (MSC Software Corporation, Santa Ana, CA). The distal femur was driven to rotation about the flexion axis. Tibial component was free except for the flexion-extension. Attachments of ligaments and tendons were sourced from MR images of the subject and considered as tensile spring- like force elements. Note: P T=patellar tendon, Q T=quadriceps tendon, MCL=medial collateral ligament, LCL=lateral collateral ligament, ACL=anterior cruciate ligament, PCL=posterior cruciate ligament.

Anterior Cruciate Ligament (ACL) has to be removed from patients for implanting TKA prostheses. Related studies demonstrated that ACL deficiency diminishes knee stability and functionality [8, 9] and reduces patient's satisfaction. Ka rrholm et al. [10] showed that internal rotation and adduction of the tibia were reduced in ACL-injured knees. Berchuck et al. [11] demonstrated so-called quadriceps-avoidance gait in patients with ACL deficiency. Pritchett [12] pointed out femoral paradoxical motion after TKA.

Although patients benefit from retaining ligament function by either retaining or reconstructing ACL after TKA, there are few designs have been reported. Stiehl et al. [13] concluded that ACL-retaining TKAs revealed gradual posterior femoral rollback and limited anteriorposterior translation. Komistek et al. [14] concluded that patients receiving an ACL-retaining TKA experienced kinematic patterns more similar to the normal knee during gait.

This study aims to retain ligament function by reconstructing the ACL in TKA. Computational simulation tools have been used to measure the performance of reconstructed artificial ACL comparing with the performance of normal knees kinematics.

Method

The entire lower extremities of a healthy female volunteer (age: 28 years old, weight: 55 kg, height: 160 cm) were scanned by using magnetic resonance (SIEMENS MAGNETOM Trio A Tim SYSTEM 3 T, Siemens, Germany). The proximal tibial bone and the distal femoral bone were transected approximately 75 mm from the natural joint line. The slice interval of MR images was 1 mm with a resolution of 480 512 pixels. The intact model was smoothed using Geomagic Studio v9.0 (Parametric Technology Corporation, Needham MA, USA) and then imported into Pro/Engineer WildFire 5.0 (Parametric Technology Corp., Needham, MA, USA) for bone preparation. Dynamic simulation was carried out using MSC. ADAMS R3 (MSC Software, Santa Ana, CA, U.S.A.).

The anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL), lateral collateral ligament (LCL), and patellar and quadriceps tendons were reconstructed from magnetic resonance images. The origin and insertion points of ligaments and tendons were referenced from relevant anatomical literatures and confirmed by a senior surgeon (Yan Wang) (**Figure 1**). In accordance with the characterization of ligaments in the literature [15], both the ACL and the PCL were considered as anterior and posterior fiber bundles; the MCL was considered with anterior, deep and oblique fiber bundles and the LCL was considered as a single fiber bundle. All ligaments were applied to the intact model and simulated as force components with parabolic and linear regions according to the following equations:

Table 1. The stiffness of each ligament

	K1 (N/mm)	K2 (N/mm)
ACL-Anterior	22.48	83.15
ACL-Posterior	26.27	83.15
PCL-Anterior	31.26	125.00
PCL-Posterior	19.29	60.00
MCL-Anterior	10.00	91.25
MCL-Oblique	5.00	27.86
MCL-Deep	5.00	21.07
LCL	10.00	72.22

$$F = \begin{cases} 0 & \varepsilon_j \leq 0 \\ K1_j \left(L_j - L_{0J}\right)^2; & 0 < \varepsilon_j \leq 2 \leq 2\varepsilon_1 \\ K2_j \left[L_j - \left(1 + \varepsilon_1\right) L_{0J}\right] & \varepsilon_j \leq 2\varepsilon_1 \end{cases} \qquad (1)$$

where εj is the strain in the jth element, K1j and K2j are the stiffness coefficients of the jth spring element for the parabolic and linear regions, respectively, and Lj and L0j are its current and slack lengths, respectively. The linear range threshold is specified as ε1 = 0.03. **Table 1** shows the stiffness of each ligament. Moreover, both quadriceps and patellar tendons were defined as medial and lateral fiber bundles and simulated as purely elastic tensile springs. The stiffness coe [16] and 1142 N/mm [17].

During knee flexion, each of medial and lateral menisci was considered as three portions in simulating its movements, including the anterior portion fixing on the tibial plateau, with posterosuperior and posteroinferior portions connecting to the anterior menisci portion by elastic tensile spring elements. The stiffness coefficients of medial and lateral menisci were simulated as 200 (N/mm) and 5 (N/mm), respectively. A damping coefficient of 0.5 (N s/mm) was applied to lateral meniscus.

The medial/laterial femoral movements were validated with an in-vivo study performed by [18], and meniscal translations were validated with an existing study describing the meniscal kinematics of the intact knee [19].

The TKA models used for computational simulation in this study were deviated from previous works proposed and validated by the co-authors [20]. The TKA models of the ACL reconstructing knee (ACL), 6 PCL retaining knee (CR) and PCL substituting knee (PS) are shown in **Figure 2**. The bone models included the distal femur, proximal tibia, fibula, and patella. Except for the ACL, the conditions of ligaments and tendons of ACL, CR and PS knees were consistent with the intact knee. The intact knee was designated as a control model. The symmetrically traditional U2 Total Knee System-PCR type (United, Co., Hsinchu, Taiwan) was used to simulate the CR model. And the PS model was simulated by the United Posterior- Stabilized Knee System (United Orthopaedic Corp., Taiwan). The distal femur, proximal tibia and patella of the intact knee bone models were resected and replaced with a medium sized CR, PS and ACL knee prosthesis under its standard surgical procedure. The ACL was reconstructed in the ACL model by anatomical single bundle techniques. The ACL knee was deviated from the CR model. The geometry of the femoral component of ACL model was identical to the CR model. The polyethylene insert of ACL model was modified to leave a hole on the articular surface of the insert through which the reconstructed ACL ligament could pass. The tibial component of the ACL model was identical to the CR model, except there was a tunnel from the post to the tibial plateau designed for the ligament direction.

Lines connecting the extension facet centers (EFCs: full extension to 15 flexion) and flexion facet centers (FFCs: 15 135 flexion) were designated as femoral flexion/extension axes [21]. The femoral flexion axes of TKA models were in compliance with the condylar radii of the femoral component. An averaged ground reaction force (1.5 body weight = 750 N) [22] was applied to the center of mass on the tibial (**Figure 1**). The friction coefficients of cartilage-to-cartilage and metal-to-polyethylene surfaces were designated as frictionless and 0.04, respectively [23]. Multiple beads connected by springs were used to simulate the wrapping of quadriceps tendon around the trochlear groove at higher knee flexion [24, 25] (**Figure 1**). All of the above mentioned criteria had been published in previous research [20].

A Cartesian coordinate system was defined to measure kinematics movements by the mediolateral axis (x, flexion and extension axis), the anteroposterior axis (y, varus and vaglus rotation axis), and the longitudinal axis (z, internal and external rotation axis) (**Figure 1**). During knee flexion, movements of medial/lateral con-

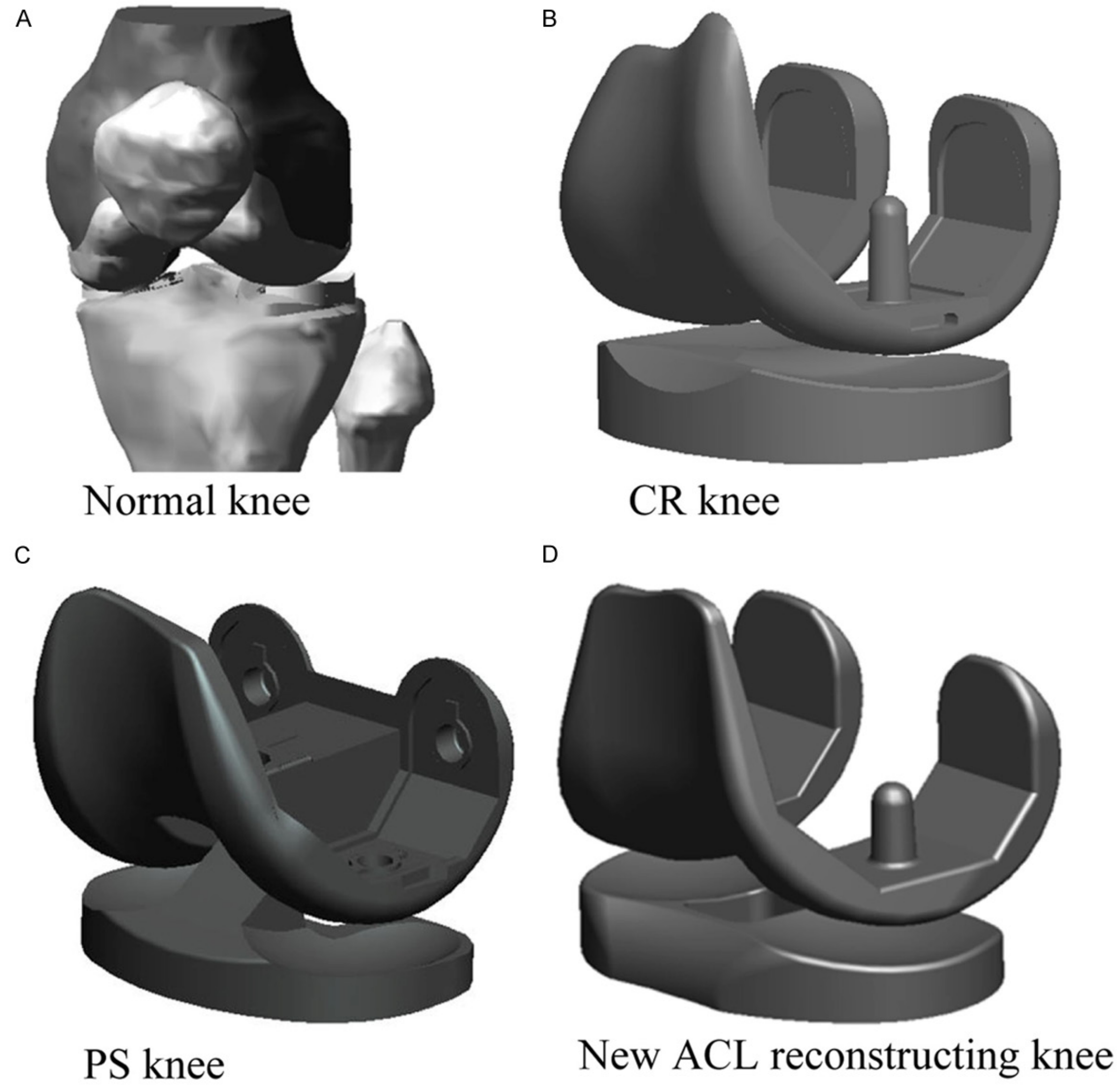

Figure 2. Normal knee and TKA models.

dyles, as well as the translations of lateral and medial menisci, were defined as the distance between original and present positions of each centers in y direction of local coordinates. Internal/external tibial rotation was defined as the angular change of the tibial local coordinate point in in z direction. All data of femoral and meniscal movements and tibial rotations were recorded every 15 from full extension to 135 of flexion.

Results

Lateral femoral condyle movements

Figure 3 shows the comparison of the lateral femoral condyle movements of the normal knee, CR, PS and new ACL models. The ACL model presents larger posterior movement of lateral condyle similar to the normal knee model during a full range of flexion. The PS model has no significant increase in posterior movement until 90 of flexion. However, paradoxical anterior translation was evident in the CR model beyond 90 of flexion. The posterior movement of lateral condyle for CR model increases gradually to 4.8 mm at 90 of flexion and then reduces to 3.6 mm at 135 of flexion. The maximum posterior movement of the lateral condyle was 4.8 mm at 90 of flexion for the CR model, 10 mm at 135 of flexion for the PS model, and 15.3 mm at 135 of flexion for the ACL model. The ACL model enhanced posterior movement of the lateral condyle by 11.7 mm

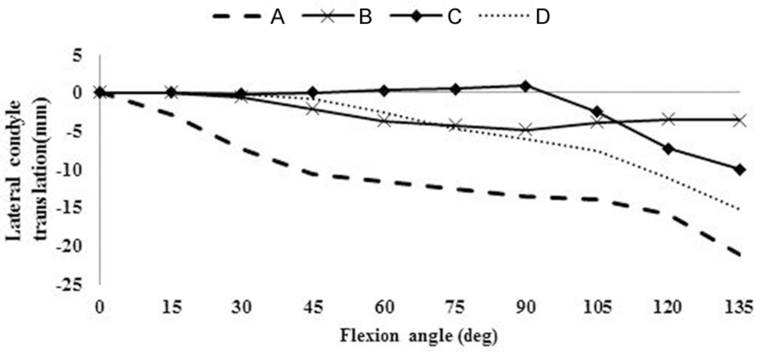

Figure 3. Comparison of lateral condyle translation of (A) normal knee, (B) CR knee, (C) PS knee and (D) ACL knee models (+: anterior; : posterior).

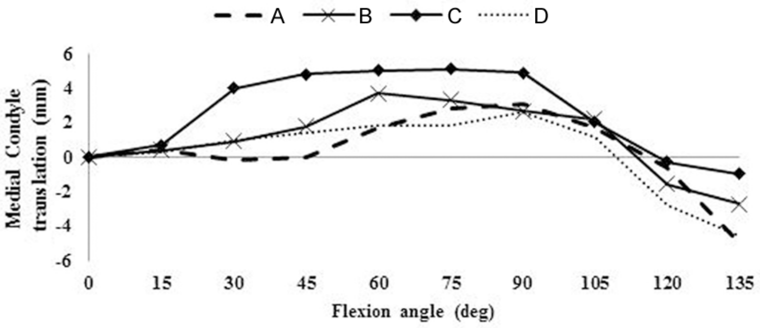

Figure 4. Comparison of medial condyle translation of (A) normal knee, (B) CR knee, (C) PS knee and (D) ACL knee models (+: anterior; : posterior).

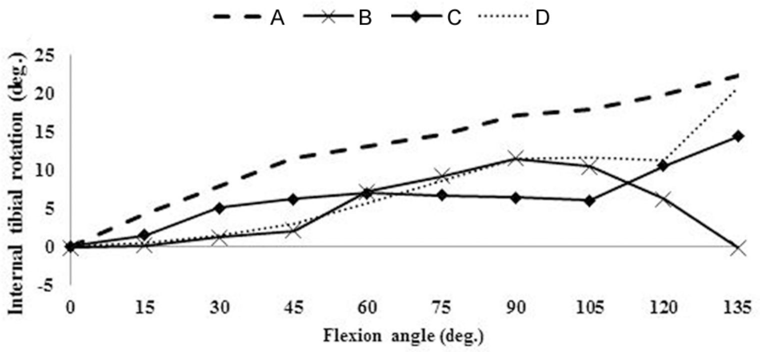

Figure 5. Comparison of internal tibial rotation of (A) normal knee, (B) CR knee, (C) PS knee and (D) ACL knee models (+: anterior; : posterior).

and 5.3 mm at maximum flexion in comparison to the CR and PS models. Compared with the normal knee through a full range of flexion, posterior movement of lateral condyle in the CR, PS and ACL models could be restored by 17.1%, 47.4% and 72.5%, respectively.

Medial femoral condyle movements

Medial femoral condyle movements of the normal knee, CR, PS and new ACL models are shown in **Figure 4**. All models displayed anterior translation during early flexion, and followed by gradual medial condyle posterior translation. Both normal knee and ACL models reached their maximum anterior translation at 90 of flexion, and then turn to medial condyle posterior translation, but anterior movements of the medial condyle of CR and PS models stopped increasing at 60 of flexion. The maximum anterior movement of the medial condyle was 3.7 mm at 60 of flexion for the CR model, 5.1 mm at 75 of flexion for the PS model, and 2.6 mm at 90 of flexion for the ACL model. The maximum posterior movement of the medial condyle at 135 of flexion was 2.8 mm for the CR model, 1 mm for the PS model, and 4.6 mm for the ACL model. Compared with the normal knee through a full range of flexion, posterior movement of the medial condyles in the CR, PS and ACL models could, on average, be restored by 56.0%, 20.0% and 92.0%, respectively.

Tibial internal rotations

Tibial internal rotations of the normal knee, CR, PS and ACL models are shown in **Figure 5**. The ACL model shows an obvious increase in tibial internal rotation when compared against the CR and PS models throughout a full range of flexion. The CR model displays external rotation from 90 of flexion through to maximum flexion. The maximum tibial internal rotations were 11.4 at 90 of flexion for the CR model, 14.3 at 135 of flexion for the PS model, and 20.6 at 135 of flexion for the ACL model. Compared with the CR and PS models, the ACL model enhanced tibial internal rotation by 20.7 and 6.3 at maximum flexion. Compared with the normal knee through a full range of flexion, he average restorations of tibial internal rotation of the CR, PS and ACL knee models were -0.01%, 64.1%, and 92.4%, respectively.

Discussion

The most important finding of this study was that ACL reconstruction could have a positive effect on postoperative TKA kinematics. ACL deficiency has been approved to be the reason of resulting in loss of femoral rollback, femoral condyle paradoxical anterior translation and other abnormal TKA kinematics by other studies [13, 14, 26, 27]. However, ACL reconstructing with TKA has not been well investigated in kinematic analysis. Researchers seemed more interested in ACL retaining or substituting designs [28, 29], but didn't pay much attention to the significance of ACL reconstruction.

The computational simulation results show that comparing with the PCL retained knee model and the PCL substituted knee model, reconstructing ACL improves the posterior movement of the lateral condyle, medial condyle and tibial internal rotation through a full range of flexion.

Tibiofemoral translation patterns of the ACL reconstructing model were similar to natural knees. The lateral condyle moved progressively posterior, while the medial condyle pivoted from full extension to maximum flexion. However, paradoxical anterior translation of the the medial condyle was observed in all knees at initial flexion angles, which was commonly observed in clinical results [30-32]. In the ACL deficient or sacrificed knees, the tibia slid forward, placing the femur too far posteriorly and resulting in forward sliding of the femur as a compensatory mechanism [33]. The lateral condyle of the CR knee showed paradoxical anterior translation. In vivo studies have often observed that PCL knees show paradoxical anterior translation, the anterior femoral translation during terminal flexion, which indicates an abnormal function of the posterior cruciate ligament [30, 34, 35].

The ACL reconstructing knee and normal knees showed similar patterns in tibial rotation, moving from neutral rotation in extension to about 20 internal rotation at 135 flexion. While the CR and PS knee produced rela tively greater external rotation in comparison to the intact knees. Moreover, tibial rotation patterns of the ACL knee show similar performances to the normal knee compared to either the CR or PS knee which might cause by the absence of the ACL. Previous works [10, 36] have reported that the ACL may control internal tibial rotation. Karrholm et al. [10] reported that internal rotation and addyctio of the tibia were reduced in the injured knees compared with the normal knees. Nilsson et al. [36] demonstrated similar findings with several different posterior cruciate-retaining TKAs, reporting more externally rotated tibia in extension and less resistance to internal tibial rotation during flexion.

Differences are clearly evident between our results and other ACL-retaining and ACL-substituting studies with regard to the magnitude of medial and lateral condyle posterior translation and tibia internal rotation. Kuroyanagi et al. [37] recorded fluoroscopic measurements of 25 bi-cruciate substituted knees during deep flexion. It was reported that during activities to demonstrate maximum flexion the medial and lateral condyles moved posteriorly by an average of 5 mm and 11 mm and tibia internal rotation averaged at 10. Moro-oka et al. [28] analyzed knee kinematics of ACL preserving knees by fluoroscopy. The ACL preserving knees showed an average posterior translation of the lateral and medial condyle of 7 mm and 2 mm and tibial rotation of 10. There are many factors that could have led to the observed differences. Firstly, in our computer simulation, the movements of medial and lateral condyles were defined as the distance between the original and present positions of flexion facet centers during knee flexion. Kuroyanagi et al. [37] and Moro-oka et al. [28] determined the anterior-posterior locations of medial and lateral condyles as the lowest point on each femoral condyle relative to the transverse plane of the metal tibial baseplate.

Although ACL-retaining knees can provide kinematics similar to normal knees, there are still some defects apparent with ACL-retaining designs. Bartel et al. [38] reported that if the ACL was preserved during TKA exposure,it was easy to mistakenly misalign the prosthesis during implantation. Jenny and Jenny [29] found that to preserve the ACL tibial insertion point, the tibial baseplate design must be altered, which would decrease the metal-cement fixation strength and result in premature implant loosening and subsidence. Also, the ACL retaining design is unsuitable for ACL-deficient patients. The ACL reconstructing TKA design used in this study aims to preserve ACL function. The choosing of an autologous ACL anatomical single bundle graft for reconstruction in this study was based on two considerations. Firstly, Goldblatt et al. [39] reported that the hamstring tendons are clinically effective as a graft choice, with strength and stiffness comparable to the previous gold standard graft choice, the central-third patellar tendon graft. Secondly, Kim et al. [40] compared the long-term stability and functional score results between anatomical single bundle reconstruction and double bundle techniques and did not find any significant difference.

Although the proposed ACL reconstructing model showed similar performances to normal knees kinematics, there are several limitations need to be considered. This study is based on the anatomic-like PCR total knee system which was developed for a particular insert design and can not integrate the mechanical properties of soft tissue. In addition, the prostheses employed in other studies were different from the ones used to establish our model. Using the anatomic-like PCR knee model made our experimental results difficult to be compared with existing reports of ACL retaining knees directly. Finally, the kinematic performance of ACL reconstructing after TKA need further investigation in clinical trials to demonstrate its advantages over other studies.

Conclusion

Removing ACL in TKA results in significant kinematic differences, including inadequate femoral rollback and insufficient tibial rotation. Our study shows that reconstructing ACL improves the posterior movement of the lateral condyle, medial condyle and tibial internal rotation through a full range of flexion. Reconstructing ACL in TKA has been approved to be an more efficient way of maintaining normal knee kinematics comparing to PCL retained and PCL substituted TKA.

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Disclosure of conflict of interest

None.

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