Original Article

Kinematic Analysis of a Posterior-stabilized Knee Prosthesis

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Abstract

Background: The goal of total knee arthroplasty (TKA) is to restore knee kinematics. Knee prosthesis design plays a very important role in successful restoration. Here, kinematics models of normal and prosthetic knees were created and validated using previously published data.

Methods: Computed tomography and magnetic resonance imaging scans of a healthy, anticorrosive female cadaver were used to establish a model of the entire lower limbs, including the femur, tibia, patella, fibula, distal femur cartilage, and medial and lateral menisci, as well as the anterior cruciate, posterior cruciate, medial collateral, and lateral collateral ligaments. The data from the three-dimensional models of the normal knee joint and a posterior-stabilized (PS) knee prosthesis were imported into finite element analysis software to create the final kinematic model of the TKA prosthesis, which was then validated by comparison with a previous study. The displacement of the medial/lateral femur and the internal rotation angle of the tibia were analyzed during 0–135° flexion.

Results: Both the output data trends and the measured values derived from the normal knee’s kinematics model were very close to the results reported in a previous in vivo study, suggesting that this model can be used for further analyses. The PS knee prosthesis underwent an abnormal forward displacement compared with the normal knee and has insufficient, or insufficiently aggressive, “rollback” compared with the lateral femur of the normal knee. In addition, a certain degree of reverse rotation occurs during flexion of the PS knee prosthesis.

Conclusions: There were still several differences between the kinematics of the PS knee prosthesis and a normal knee, suggesting room for improving the design of the PS knee prosthesis. The abnormal kinematics during early flexion shows that the design of the articular surface played a vital role in improving the kinematics of the PS knee prosthesis.

Key words: Arthroplasty; Knee; Kinematics; Posterior-stabilized Prosthesis; Replacement

INTRODUCTION

Osteoarthritis of the knee is a very common joint disease that primarily afflicts older people and leads to a series of symptoms, including pain, decreased range of motion (ROM), and loss of function, among others. Over the past 30 years, the total knee arthroplasty (TKA) procedure has proven to be one of the most effective surgeries for patients with degenerative knee disease, and TKA is accepted as the gold standard for the treatment of osteoarthritis.[1,2] According to one report, the number of annual TKA procedures in the United States is about 300,000–500,000. Based on the proportional population, an estimate of the number of annual TKA procedures in China is about 1,000,000–1,500,000. The clinical statistics over the past decade show that 90% of patients obtain good clinical results after receiving a TKA.[3,4] However, good clinical results do not necessarily indicate high postoperative satisfaction rates in the patients.[5,6] Abnormal knee kinematics is an important factor that influences postoperative satisfaction rates. For example, deficient rollback of the femur or abnormal rotation of the tibia will lead to a decreased range of flexion angles for patients after TKA.[7,9] Therefore, the in vivo kinematics after an operation is a key element for evaluating the design of knee prostheses.

The posterior cruciate ligament (PCL) must be excised intraoperatively during TKA with a posterior-stabilized (PS) knee prosthesis. The PS knee prosthesis substitutes a cam and postinteraction for the stability normally offered by the PCL. This interaction can also help with the rollback of the femur and control the backward movement of the tibia, reducing instability during flexion. Several studies on in vivo knee kinematics using radiologic analysis have been published,[10–13] but studies that apply computer simulation to analyze and predict knee kinematics are limited. The in vivo knee kinematics studies using radiologic analysis show that
with increasing knee flexion, the lateral femur rollback tends to exceed that of the medial femur and the internal rotation of the tibia. This is called “screw-home.” To improve the postoperative satisfaction rates of patients, knee prostheses should reproduce the “screw-home” effect.

In the present study, a model of normal knee kinematics was created. We simulated the entire process of TKA using this model, and analyzed and predicted the in vivo knee kinematics of PS knee prosthesis. The results of the study may provide a useful kinematics reference for the design of knee prostheses.

Methods

Establishing the three-dimensional model of normal knee kinematics

The sample was a healthy, anticorrosive female cadaver (age: 40 years; height: 164 cm; weight: 50 kg). Computed tomography (CT) scans (Siemens SOMATOM Emotion 16, Siemens Ltd., Munich, Germany) were taken from 5 cm above the tip of the femoral head to the ankle joint. The basic settings during the CT scan included: A scan interval of 3 mm, the apparent plane as the main plane, and a scanning resolution of 512 × 512 pixels. In addition, the bony structures 10 cm above and below the knee joint line were scanned using a magnetic resonance imaging (MRI) device (Siemens Avanto, Siemens Ltd., Munich, Germany). The interval was 0.5 mm, and the scanning resolution was 512 × 512 pixels. All the data obtained from the CT and MRI scans were saved as Digital Imaging and Communications in Medicine format files. After inputting these data, the medical modeling software Mimics 13.0 (Materialise Ltd., Leuven, Belgium) was used to establish the model of normal knee kinematics.

The CT images were used to establish a model of the entire lower limbs, including the femur, tibia, patella, and fibula. The normal attenuation coefficient range of human skeletal bone is 226–1701 Hu; this threshold range was chosen to distinguish the bony structures 10 cm above and below the knee joint line. The different colors indicate the masks of different bone models. Manually divided and repaired images were applied to the scanned images for processing. First, a partial division was made for the scanned images of the connection structures among the femur, tibia, and fibula. Next, the “region growth” function of Mimics was used to further divide the selected image, and different bone structures were separated. Finally, the “three-dimensional (3D) calculation” function of Mimics was used to reconstruct each individual mask. After this process, the 3D bone structures were clearly visualized [Figure 1a].

Magnetic resonance imaging scans were used to establish the model of the distal femur cartilage and meniscal and medial and lateral menisci. The scanned MRI images were imported into Mimics 13.0. The same image processing described above was used to divide, repair, and reconstruct 3D models of the distal femur cartilage and menisci [Figure 1b].

Because the coordinate frame of the CT images was different than that of the MRI images, 3D rectification was performed to reconstruct the structure of the CT and MRI images. This helped the reconstructed 3D bone model based on the CT images match to the reconstructed model based on the MRI images. Our main idea was to perform a coordinate transformation on the 3D bone model based on the CT images using Geomagic software (Parametric Technology Corporation, Needham, MA, USA), and adapt the new coordinate frame to the frame of the MRI images. After splicing the CT and MRI models, the new data were imported into the 3D design software PRO/E (Parametric Technology Corp., Needham, MA, USA), finally establishing an intact 3D model of the knee joint model of knee joint was established [Figure 1c].

Simulation of the ligaments around the normal knee joint and muscle strength

The anterior cruciate ligament (ACL), PCL, medial collateral ligament (MCL), and lateral collateral ligament (LCL) were simulated in the present study. Springs with different stiffness coefficients (SCs) were used to simulate the force of each ligament. Based on the findings of Abdel-Rahman and Hefzy,[14] the ACL and PCL can be simulated using two branches: An anterior and a posterior branch. The MCL can be simulated using three branches: Anterior, deep, and oblique branches, while the LCL can be simulated using a single branch. The piecewise function used to define the force of ligaments was:

$$F_j = \begin{cases} 0; & \varepsilon_j \leq 0 \\ K_1 (L_j - L_{o0})^2; & 0 < \varepsilon_j \leq 2\varepsilon_0 \\ K_2 (L_j - [1 + \varepsilon_j] L_{o0}); & 2\varepsilon_0 < \varepsilon_j \end{cases}$$

In the above function, $F_j$ represents the force of ligament; $K_1$ and $K_2$ are the SCs of the spring unit; $L_{o0}$ is the initial length of the ligament; $L_j$ is the stretched length; $\varepsilon_0$ is defined as 0.03; and $\varepsilon_j$ is the deformation of the ligament. To precisely simulate the forces of the ligaments, a hypothesis was applied to the spring unit: The force of the ligaments is a tensile stress with no force during compression. The SC and compensation coefficient (CC) of different ligaments [Table 1] prevented the ligaments around the knee joint from experiencing...
compressive forces. The CC was further interpreted in a study by Blankevoort et al.[15] Each ligament was identified by the CC in each test, which helped to simulate the forces of the ligaments. To better simulate the physiological mechanisms within the knee joint, the patellar tendon and quadriceps tendons were also simulated. Both the patellar and quadriceps tendons consist of medial and lateral branches, which can also be simulated using spring units.[16] The SC of the patellar tendon is 1000 N/mm,[17] and that of the quadriceps tendon is 521 N/mm.[18] The attachment points of all the ligaments and the patellar and quadriceps tendons were identified by experienced orthopedic surgeons after repeated observations of the bone model and MRI data.

### Establishing the kinematics model of the normal knee

The data from the 3D model of the normal knee joint were imported into MD Adams R3 software (MSC Software, Newport Beach, CA, USA), and the properties of different bone structures were assigned. Next, the simulation unit was established to simulate the forces of the ligaments and muscle strength based on the settings described above. Because the MD Adams software could not simulate deformation, we designed the study to better simulate the real physiology. First, the medial and lateral menisci were divided into anterior and posterior parts based on the location of the centroid. These two parts of the menisci were then connected to the spring unit, and a damping coefficient of 0.5 N s/mm[19] was applied to the spring unit to ensure that the distal femur cartilage was in contact with the meniscus at the same time. The rotational axis during flexion of the femur (femur flexion center [FFC] axis) was based on the medial and lateral FFC. The rules of the coordinate frame were defined as the FFC axis as the X-axis, the mechanical axis of the lower limb as the Y-axis, and the Z-axis was identified using the “right-hand rule.” Together, the X, Y, and Z axes constitute a cartesian coordinate system in a certain space. As flexion increased, the displacements of the medial and lateral femur (forward and backward) were identical to the displacements of the medial and lateral femur along the Y-axis. The internal rotation of the tibia was reflected by the rotation around the Z-axis. Thus, the model of normal knee joint kinematics was established [Figure 2a]. With this model, the knee could be simulated within a ROM of 0–135° of flexion. The output of the model consisted of the displacements of the medial and lateral femur and the internal rotation angle of the tibia.

### Establishing the kinematics model of the total knee arthroplasty prosthesis

A set of PS knee prostheses was selected as the study objects. A 3D model derived from the 3D scanner information was created, and this model was developed using Geomagic software. The entire TKA surgery procedure was simulated using the 3D model of the normal knee with the PS knee prosthesis implanted in it. The 3D knee model of the TKA prosthesis was then imported into MD Adams R3 software. The femur prosthesis was fixed on the femur; the tibia prosthesis was fixed on the tibia plateau, and the polyethylene inlay was well-seated on the tibia prosthesis. The femur was set in contact with the high molecular weight polyethylene inlay. In addition, the properties of the ligaments and muscle strength were set according to the methods described above, as well as the settings of the rotational axis and the coordinate system of measurement parameters. Given that the meniscus was already excised during the TKA, it was not necessary to simulate the parameters of the meniscus. Thus, the final kinematics model of the TKA prosthesis was established [Figure 2b], simulating a ROM (0–135° of flexion). The output of the model consisted of the displacements of the medial and lateral femur and the internal rotation angle of the tibia.

### Results

#### Validation of the model of normal knees

Validation was performed to verify the accuracy of the target model by analyzing the trends of the output data and the values measured. There are two methods of validation: Experimental verification and literature comparison. Because it was difficult to find in vivo kinematics data, the method of literature comparison was used to validate the model in the present study. The displacements of the medial and lateral femur predicted by the normal knee model were compared with data from an in vivo study by Johal et al.[13] Because only the displacements of the medial and lateral

### Table 1: The SC and CC of different ligaments

| Ligament       | K1 (N/mm) | K2 (N/mm) | CC  |
|----------------|-----------|-----------|-----|
| ACL-Anterior   | 22.48     | 83.15     | 1   |
| ACL-Posterior  | 26.27     | 83.15     | 1.051|
| PCL-Anterior   | 31.26     | 125.00    | 1.004|
| PCL-Posterior  | 19.29     | 60.00     | 1.05 |
| MCL-Anterior   | 10.00     | 91.25     | 0.94 |
| MCL-Oblique    | 5.00      | 27.86     | 1.031|
| MCL-Deep       | 5.00      | 21.07     | 1.049|
| LCL            | 10.00     | 72.22     | 1.05 |

SC: Stiffness coefficient; CC: Compensation coefficient; ACL: Anterior cruciate ligament; PCL: Posterior cruciate ligament; MCL: Medial collateral ligament; LCL: Lateral collateral ligament.
femur were reported in Johal’s study, we had to use the data for the model validation. The data comparison [Figure 3] between the simulated model of the normal knee and the results of the in vivo study shows that both the trends of the output data and the values measured, which were derived from the normal knee kinematics model, are very close to the results from Johal’s in vivo study. Therefore, this model can be used for further analyses.

The kinematic characteristics of the posterior-stabilized knee prosthesis
Analysis was performed on the kinematic characteristics of the PS knee prosthesis based on the kinematics model of the TKA prosthesis and three other parameters: The displacement of the medial femur, the displacement of the lateral femur, and the internal rotation angle of the tibia. The kinematics characteristics of the normal knee and the PS knee prosthesis are compared below. The maximum displacement of the medial femur after implantation of the PS knee prosthesis was about 5.1 mm compared with 3.1 mm in a normal knee. This result indicates that the PS knee prosthesis underwent an abnormal forward displacement compared with the normal knee [Figure 4a]. The backward displacement of the lateral femur with the PS knee prosthesis was nearly 0 mm for flexion angles <90°, and the maximum backward displacement was about 10 mm for flexion angles more than 90°. The lateral femur of the normal knee experiences a continuous “rollback” movement, and the maximum backward displacement is about 21.1 mm. This finding indicates that the PS knee prosthesis has an insufficient, or insufficiently aggressive, “rollback” compared with the lateral femur of the normal knee [Figure 4b]. The internal rotation angle of the tibia in the PS knee prosthesis was <7° for flexion angles <105°. There was mild reverse rotation (about 1°) when the knee flexion angle was between 60° and 105°. This reverse rotation angle increased to 20.6° when the flexion angle was over 105°, while the internal rotation angle of the tibia in normal knees continuously increases until 22.3°. This finding indicates that the tibia, after implantation of a PS knee prosthesis, has insufficient internal rotation compared with the tibia in normal knees, and that there is a certain degree of reverse rotation during flexion [Figure 4c].

Discussion
The design concept of the PS knee prosthesis includes the use of the cam and postinteraction as a substitute for the function of the PCL. The contact between the post and cam in most PS knee prostheses on the market occurs when the knee flexion angle is about 70–100°. Before the cam and post mechanism were created, the entire knee movement was controlled by the MCL, LCL, and the inlay articular surface. This suggests that the design of the inlay articular surface is critical during the early flexion of the knee joint, when the inlay acts as a substitute for the roles of the ACL and PCL.

According to the results of several published in vivo studies on knee kinematics, the femur has an external rotation relative to the tibia with the medial femur being...
The abnormal kinematics predicted during early flexion showed that the design of the articular surface plays a vital role in improving the kinematics of PS knee prostheses. The simulated kinematics model of the knee joint in the present study can be used as the basis for assessing further improvements on the design of cruciate-retaining (CR) and PS knee prostheses.

Several limitations of the present study should be mentioned. The set of PS knee prostheses that were used here are not necessarily representative of all the different styles of commercial posterior substituting products. Similarly, a single sample female cadaver cannot be considered representative of the anatomical features of all patients who undergo knee prosthetic procedures. In addition, certain surrounding soft tissues, including the joint capsule, hamstring, and gastrocnemius could not be involved in our model simulation. All of these problems should be addressed in follow-up studies.

Our future work will focus on several points. We should make better use of the kinematics model established in this study. In addition, our future work should provide design proposals for the development of CR and PS knee prostheses aimed at restoring physiological kinematics after TKA.

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