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Three dimensional finite element analysis of the influence of posterior tibial slope on the anterior cruciate ligament and knee joint forward stability

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Abstract.
OBJECTIVE: To explore the biomechanical influence of posterior tibial angle on the anterior cruciate ligament and knee joint forward stability.
METHODS: The left knee joint of a healthy volunteer was scanned by CT and MRI. The data were imported into Mimics software to obtain 3D models of bone, cartilage, meniscus and ligament structures, and then Geomagic software was used to modify of the image. The relative displacement between tibia and femur and the stress of ACL were recorded.
RESULTS: ACL tension was 12.195 N in model with 2° PTS, 12.639 N in model with 7° PTS, 18.658 N in model with 12° PTS. the relative displacement of the tibia and femur was 2.735 mm in model with 2° PTS, 3.086 mm in model with 7° PTS, 3.881 mm in model with 12° PTS. In the model with 30° flexion, the maximum tension of ACL was 24.585 N in model with 2° PTS, 25.612 N in model with 7° PTS, 31.481 N in model with 12° PTS. The relative displacement of the tibia and femur was 5.590 mm in model with 2° PTS, 6.721 mm in model with 7° PTS, 6.952 mm in model with 12° PTS. In the 90° flexion models, ACL tension was 5.119 N in model with 2° PTS, 8.674 N in model with 7° PTS, 9.314 N in model with 12° PTS. The relative displacement of the tibia and femur was 0.276 mm in model with 2° PTS, 0.577 mm in model with 7° PTS, 0.602 mm in model with 12° PTS.
CONCLUSION: The steeper PTS may be a risk factor in ACL injury.

Keywords: Posterior tibial slope, anterior cruciate ligament, knee joint, finite element analysis

1. Introduction

The injury of anterior cruciate ligament (ACL) is one of the common sports injuries, of which about more than 70% are caused by non-contact events [1]. The risk factors of non-contact ACL injury include: movement types, female, intercondylar fossa of the form, fatigue, etc. In recent years, the influence of posterior tibial slope (PTS) on non-contact ACL injury is gradually concerned. However, most researches focused on the clinical multiple factors [2,3], and the consensus on this problem was not achieved. The basic research on the role of the changes of the PTS in the ACL injury is limited [3–6]. To further investigate the relationship between PTS and knee biomechanics, the three dimensional finite element analysis of knee joint was employed to analyze the biomechanical influence of posterior tibial angle on the anterior cruciate ligament and knee joint forward stability.

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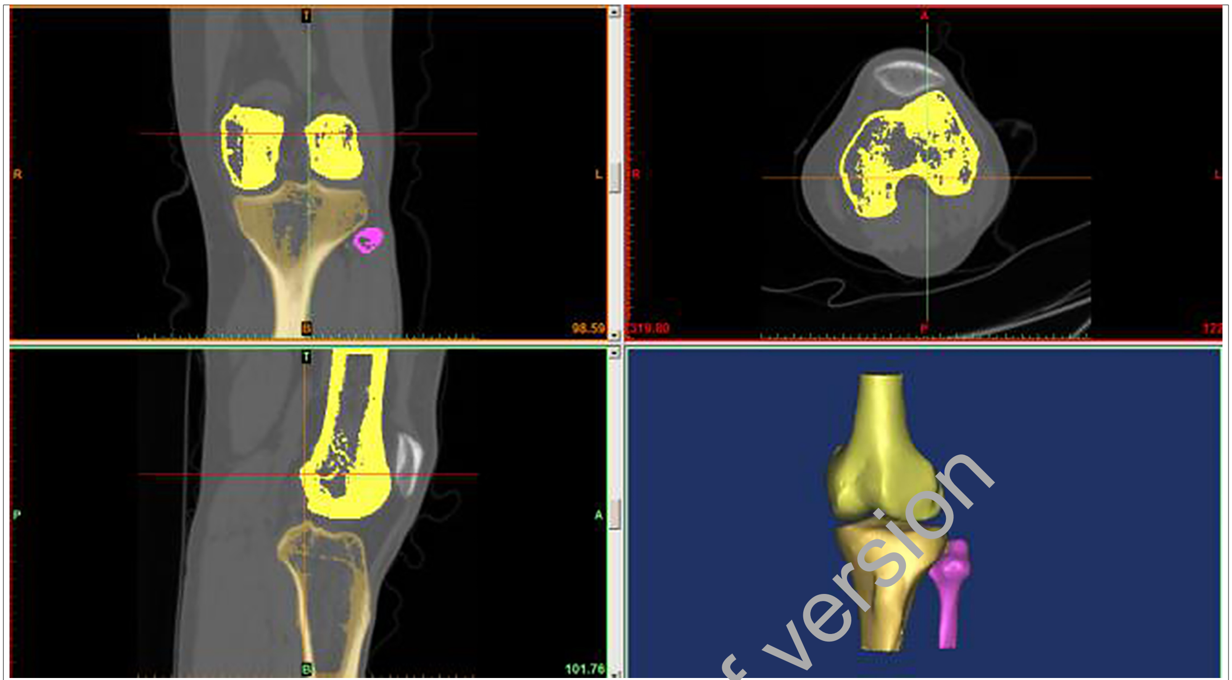

Fig. 1. The tissue was segmented and extracted automatically after the data of CT scans were imported into Mimics software to obtain accurate 3D model reconstruction.

2. Materials and methods

2.1. Objects and equipment

In this study, a healthy male volunteer was selected as research object, with the height of 175 cm, the weight of 75 kg, the body mass index BMI was 24.5, and the knee joint had no chronic pain or trauma history.

The equipment used in this study were 1.5 T dual gradient nuclear magnetic resonance (Philips Achieva), Philips ingenuity core 128 slice spiral CT.

The working platform was LENOVO workstation with CPU: I7-3770, memory: 32 G, operating system: Windows 7.

The analysis software were Mimics10.01, Geomagic2013, SolidWorks 2012, ANSYS14.5.

2.2. Establishment of three-dimensional finite element model of knee joint

2.2.1. Basic data acquisition

Informed consents were obtained from all the included patients. The left knee set in extension position (genuflex at 0 degrees) was scanned with CT and MRI. CT and MRI images of the upper and lower 15 cm range of the knee joint were obtained. Layer thickness of CT scanning was 0.625 mm, which of MRI scanning was 0.8 mm. The image is saved as DICOM format.

2.2.2. Establishment of a preliminary model of bone, cartilage, ligament and meniscus tissue

CT and MRI scanning data in DICOM form were imported into Mimics10.01 software. By using the image segmentation and automatic extraction of the system, the accurate 3D segmentation and model reconstruction were performed.

The 3D model of bone tissue is obtained from the CT scan data (Fig. 1). After that, the MRI scan data were imported into the Mimics software, the sagittal plane, coronal plane, cross section and a three-dimensional view of knee joint could be observed. The outline of cartilage, meniscus, medial and lateral collateral ligament as well as the cruciate ligament were described, and the data were obtained. The preliminary data of knee joint cartilage, meniscus and ligament in the model was obtained (Fig. 2).

Surface mesh editing tool of Geomagic2013 software was used to build a 3D model for each part. In particular, the 3D model reconstructed by MRI was modified to make the model more smooth, supple, and get the model with high quality surface. Registration alignment of the three dimensional model of soft tissue surface was performed in Solidworks 2012 soft-

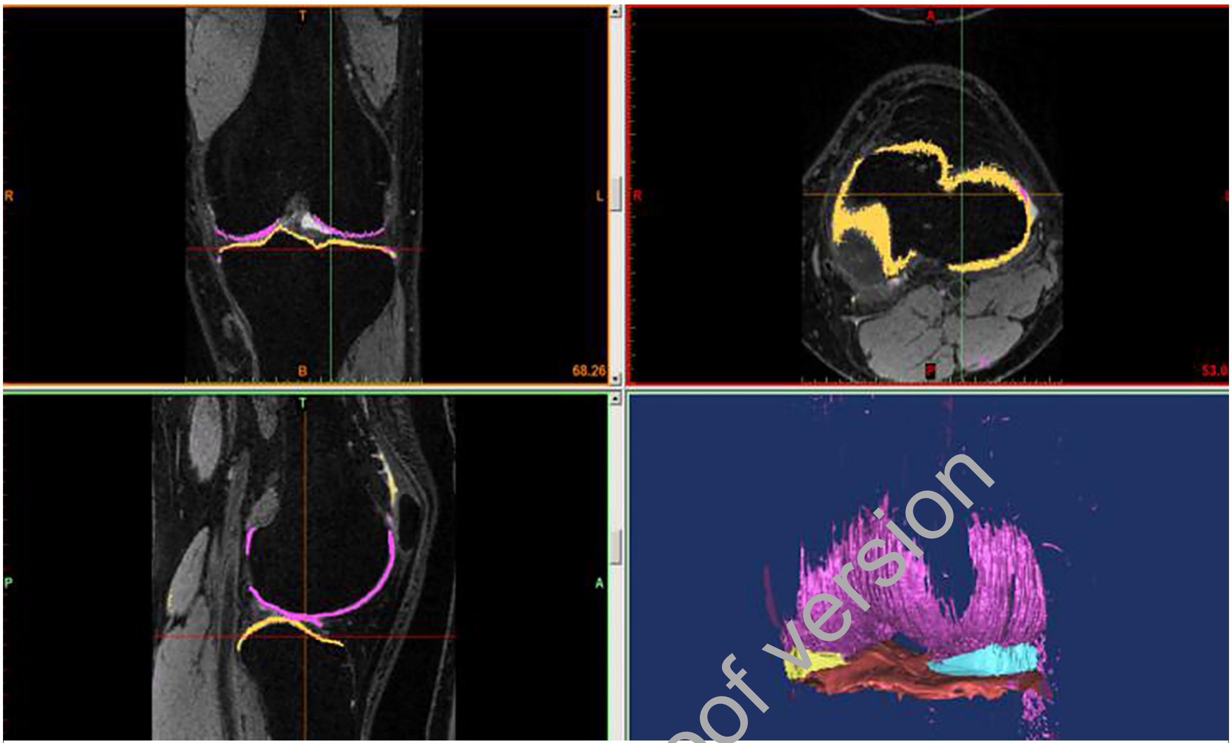

Fig. 2. The tissue of cartilage and meniscus were outlined and extracted automatically after the data of MRI scans were imported into Mimics software to get 3D model reconstruction.

ware based on two different modal data. The three-dimensional model of main ligament, meniscus reconstructed by MRI was converted to CT scanning data space. The biomechanical finite element mesh of Solidworks is used to directly partition the 3D solid model in to high quality body mesh and to integrate the three-dimensional finite element model of the knee joint.

The ANSYS software was used to import the 3D finite element model of knee joint into ANSYS 14.5 finite element analysis software. Tetrahedral units were classified to improve numerical accuracy. The model was divided into 48909 unit, 81593 nodes (Fig. 3).

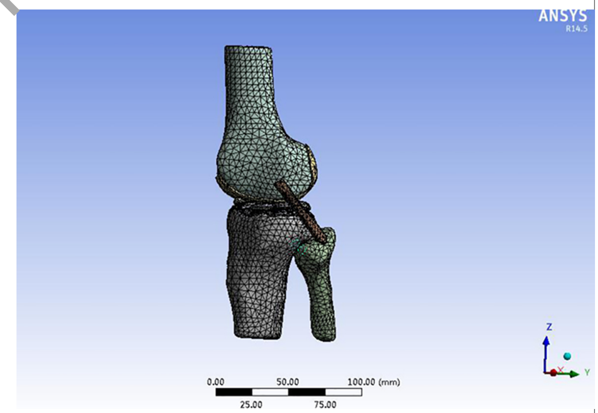

Fig. 3. The tetrahedral elements were parted with ANSYS14.5 software, the model is divided into 48909 units and 81593 nodes.

2.2.3. The establishment of different flexion angle model and different angle of posterior slope of the medial tibial plateau

According to the previous report [7], the PTS of the posterior aspect of the tibia through MRI sagittal image. The posterior slope of the medial tibial plateau was 7°. The cutting function of Solidworks 2012 image processing software was used to cut the bone under the tibial plateau. After adjusting the posterior slope angle of the platform by rotating the proximal end of the osteotomy, the clearance was filled with bone of the same attribute material. The knee joint model of 12 degree and 2 degree were reconstructed respectively. Three kinds of knee joint model with different posterior tilt angle were obtained, which were classified as, group A with posterior tilt angle of 2°, group B with posterior tilt angle of 7°, group C with posterior tilt angle of 12°.

In the three groups after the reconstruction, the femur was rotated backward to 30° and 90° respectively. The skeleton model of genuflex with 30° and 90° was

Table 1
Models with different knee flexion angle and with different PTS

Posterior tibial slope (PTS)	Extension position	Knee flexion at 30°	Knee flexion at 90°
2°	A1	A2	A3
7°	B1	B2	B3
12°	C1	C2	C3

Table 2
Biomechanical parameters of each tissues in the 3D finite element model of knee joint

Organization structure	Elastic modulus E (MPa)	Poisson ratio
Bone	2.06×10^5	0.30
Cartilage	5.0	0.46
Ligament	215.3	0.40
Meniscus	59.0	0.49

set. Adjustment of the spatial position of the ligaments of the knee joint was made in order to achieve a better spatial anastomosis. The stress concentrated and the potential interference zone for late calculation were corrected. In this way, three different state model of knee flexion with 0°, 30° and 90° were established (Table 1).

2.3. Material properties and boundary conditions

According to previous reports [8,9], experiments and analysis, the deformation of the bone tissue structure is smaller than soft tissue structure, including articular cartilage and meniscus. Therefore, we set the material properties of femur, tibia and fibula as isotropic elastic material with elastic modulus of 2.06 $\times 10^5$ MPa, poisson ratio of 0.30, set the material properties of articular cartilage and meniscus as homogeneous, continuous, isotropic elastic material with elastic modulus of 5 MPa, poisson ratio of 0.46 and with elastic modulus of 5 MPa, poisson ratio of 0.49 respectively [8,9]. At last, we set ligament as homogeneous, continuous, isotropic elastic material with elastic modulus of 215.3 MPa, poisson ratio of 0.4 (Table 2).

In order to make the model closer to the entity, the two ends of the main ligaments and anatomical attachment points were set to be connected to the common nodes. Setting the surface of the articular cartilage fixed with the surface of the bone tissue, the anterior horn and posterior horn of the meniscus and the outer edge of the medial meniscus were fixed with the edge of the tibial plateau, to simulate the attachment of the meniscus in the tibial plateau. The contact between cartilage and meniscus was surface-surface contact and the contact property was nonlinear no friction contact [8,9].

2.4. Loading condition

In the genuflex model with 0°, the X, Y and Z axes of tibia were fixed. In the femoral condyle, the medial collateral ligament is given vertical stress of 1150 N to observe the relative displacement of the femur and tibia and the tension of anterior cruciate ligament [8,9].

In the genuflex model with 30°, the X, Y and Z axes of tibia were fixed. Vertical ground stress of 750 N and external rotation stress of 10 Nm were given at the attachment point of medial collateral ligament upper the femoral condyle.

In the genuflex model with 90°, the X, Y and Z axes of tibia were fixed. A point fix method was performed on the distal end of tibia. The anterior forward stress of 134 N was given at the lower surface of the tibia [8,9].

The load of 1150 N with the extension of the knee joint is the maximum load of one side of the knee joint in the 70 kg person's normal walking gait. The load of 750 N along the femoral load and 10 Nm torque in the genuflex of 30° is to simulate the condition of the knee joint during the take-off or landing.

3. Results

In the genuflex model with 0°, the maximum displacement of the femur in the three groups were 2.735 mm of A1, 3.086 mm of B1, 3.881 mm of C1 respectively (Fig. 4A–C). The maximum tension of the anterior cruciate ligament were 12.195 N of A1, 12.639 N of B1 and 18.658 N of C1 respectively (Fig. 4D–F).

In the genuflex model with 30°, the maximum displacement of the femur in the three groups were 5.590 mm of A2, 6.721 mm of B2, 6.952 mm of C2 respectively (Fig. 5A–C). The maximum tension of the anterior cruciate ligament were 24.585 N of A2, 25.612 N of B2 and 31.481 N of C2 respectively (Fig. 5D–F).

In the genuflex model with 90°, the maximum displacement of the femur in the three groups were 0.276 mm of A3, 0.577 mm of B3, 0.602 mm of C3 respectively (Fig. 6A–C). The maximum tension of the anterior cruciate ligament were 5.119 N of A2, 8.674 N of B2 and 9.314 N of C2 respectively (Fig. 6D–F).

4. Discussion

Knee joint anterior cruciate ligament injury caused by knee joint injury was considered as a hot topic in

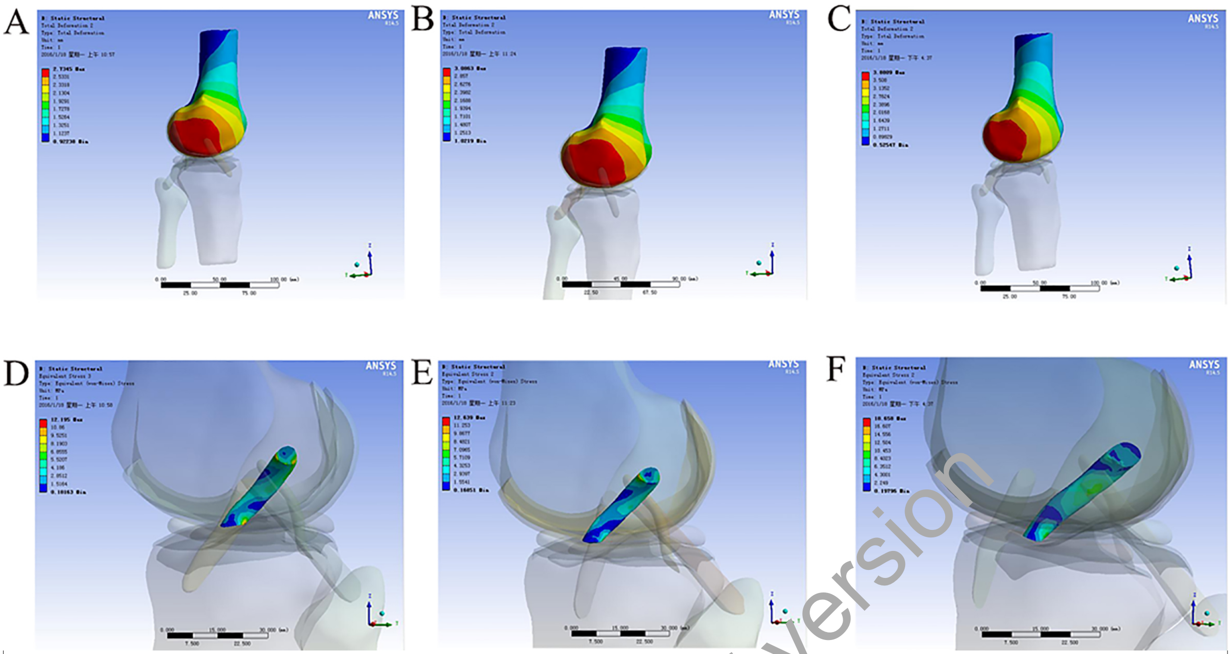

Fig. 4. The relative displacement of the tibia and femur and ACL tension in the genuflex model with 0°. A. The relative displacement of the tibia and femur was 2.735 mm in model A1; B. The relative displacement of the tibia and femur was 3.086 mm in model B1; C. The relative displacement of the tibia and femur was 3.881 mm in model C1; D. The ACL tension was 12.195 N in model A1; E. The ACL tension was 12.639 N in model B1; F. The ACL tension was 18.658 N in model C1.

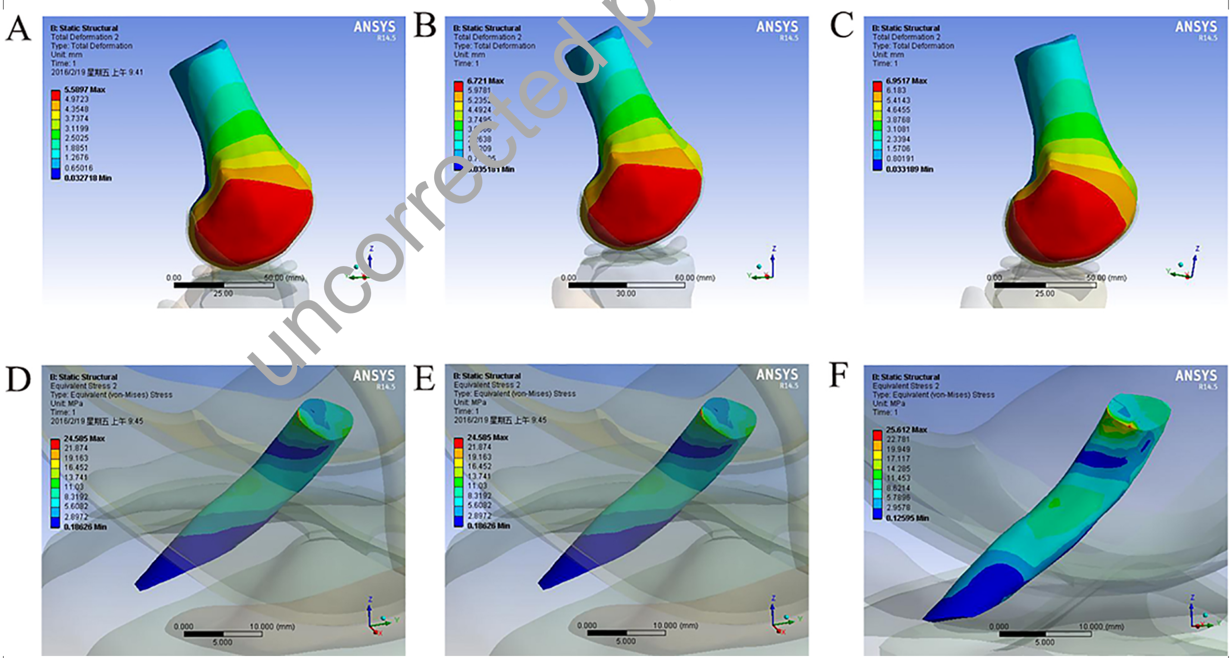

Fig. 5. The relative displacement of the tibia and femur and ACL tension in the genuflex model with 30°. A. The relative displacement of the tibia and femur was 5.590 mm in model A2; B. The relative displacement of the tibia and femur was 6.721 mm in model B2; C. The relative displacement of the tibia and femur was 6.952 mm in model C2; D. TheACL tension was 24.585 N in model A2; E. The ACL tension was 25.612 N in model B2; F. TheACL tension was 31.481 N in model C2.

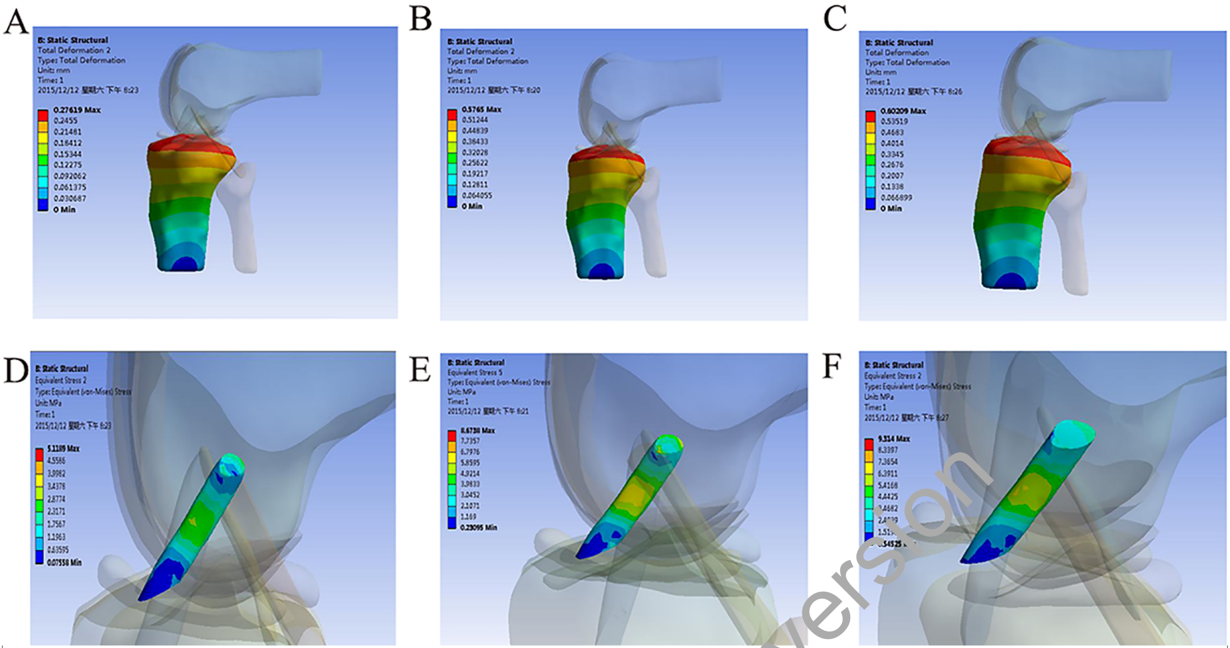

Fig. 6. The relative displacement of the tibia and femur and ACL tension in the genuflex model with 90°. A. The relative displacement of the tibia and femur was 0.276 mm in model A3; B. The relative displacement of the tibia and femur was 0.577 mm in model B3; C. The relative displacement of the tibia and femur was 0.602 mm in model C3; D. The ACL tension was 5.119 N in model A3; E. TheACL tension was 8.674 N in model B3; F. The ACL tension was 9.3141 N in model C3.

the research field of orthopaedic sports medicine. The former research focused mainly on the functional damage of the knee joint after ligament injury and the operation method of the repair and reconstruction. In recent years, the research on knee joint anterior cruciate ligament injury accessed to the mechanism of damage. In particular, the mechanism of non-contact ACL damage of the knee joint had been made a few progresses. Female, menstrual cycle, exercise fatigue, genetic genes, and so on, are thought to be associated with ACL damage [10–13].

The association between posterior tibial slope (PTS) and non-contact ACL damage has attracted increasing attention worldwide. It was considered that the increase of posterior slope angle would lead to tibial antdisplacement and the increase of ACL load, which would affect the biomechanical properties of knee joint and increase the risk of ACL injury [14–16]. Giffin et al. [15] performed osteotomy on 10 cadaveric specimens. When the PTS increased 4.4°, the tibial antedisplacement increased 4.7 mm with genuflex of 30° under 200 N axial stress. Dare et al. [16] also found that large posterior tilt angle was rick factor for ACL injury. When the posterior tilt angle was more than 4°, sensitivity of the prediction for ACL damage was 76%, and the specificity was 75%. Patients with higher PTS had a relatively high failure rate of the reconstruction of ACL. Moreover, some surgeons performed osteotomy to reduce the tension of the ACL in the ACL reconstruction [17,18].

However, the results of current study on the risk of ACL damage and PTS are not consistent. It was reported that there were no association between PTS and ACL injury [3,19]. The results of that control study showed that both medial and lateral bony PTS were not associated with ACL injury. Kostogiannis et al. compared the PTS of the patients with obvious knee instability after ACL fracture, required reconstruction surgery and the PTS of the patients without knee instability and reconstruction surgery. This prospective research showed that knee instability after ACL injury was more obvious in the patients with smaller PTS. Therefore, it was considered that the lager the PTS was, the more stable the knee joint was and it was not considered that the increase of PTS will increase the risk of ACL injury.

There is controversy about whether PTS has become a risk factor of non-contact ACL damage, which may be induced by following reasons. First, the research methods were different. Some researches were conducted by cadaver specimens and some results were obtained from clinical observation. There is an great disadvantage in cadaver specimens. Beside the difference of PTS in different specimens, other parts including the difference between the inside and outside posterior tilt angle, the size of the condylar fossa, the thickness of the meniscus etc., would interfere the results. Second, the object of study is different. Differences in race may also lead to differences in the results of the study, or even to the contrary.

To investigate the effect of PTS single factor on the biomechanics of ACL and knee stability and to reduce the interference of other factors, our study used computer inverse reconstruction method to build a digital model of normal human knee joint and a computer finite element analysis method was performed. The method used the image data to reconstruct the highly simulated knee joint model and overcome the shortcomings of cadaver specimen. In this study, we set up different posterior tilt angle and kept the consistent of other anatomic parameters. At the same time, the way to build posterior tilt angle was not the same as the osteotomy and fixation method used in the corpse. This method can avoid the influence of bone discontinuity after osteotomy and the difference between steel plate and the elastic modulus of bone.

In this study, the loading was set according to previous reports [20–22]. The load of 1150 N with the extension of the knee joint is the maximum load of one side of the knee joint in the 70 kg person's normal walking gait. The load of 750 N along the femoral head and 10 N.m torque in the genuflex of 30° is to simulate the condition of the knee joint during the take-off or landing. The load of 114 N in the genuflex of 90° is mainly in order to compare to the related research [21,22].

We have found that when the posterior tibial angle was increased, no matter the stress was applied in the extension position or in the flexion position, the displacement of the tibia and the femur will increase subsequently. Similarly, tension changes of the anterior cruciate ligament also showed the same trend. Therefore, we speculate that the larger PTS will lead to greater stress in the ACL movement, which is a high risk factor for non-contact injury of ACL. Further research with larger sample size is needed to verify this speculation.

In conclusion, our results support the conclusion of most clinical observation, that is PTS is the high risk of anterior cruciate ligament injury. However, beside PTS, other factors would also affect the ACL stress in actual motion. Therefore, the significance of PTS in the non-contact ACL damage and the possible influence on the choice of the ACL reconstruction method are still need to be further studied.

Conflict of interest

None to report.

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