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ORIGINAL PAPER

Effect of tibial drill-guide angle on the mechanical environment at bone tunnel aperture after anatomic single-bundle anterior cruciate ligament reconstruction

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Abstract

Purpose The tibial drill-guide angle in anterior cruciate ligament (ACL) reconstruction influences the tunnel placement and graft-tunnel force, and is potentially associated with post-operative tunnel widening. This study aimed to examine the effect of the drill-guide angle on the stress redistribution at the tibial tunnel aperture after anatomic single-bundle ACL reconstruction.

Methods A validated finite element model of human knee joint was used. The tibial tunnel with drill-guide angle ranging from 30° to 75° was investigated. The post-operative stress redistribution in tibia under the compressive, valgus, rotational and complex loadings was analysed.

Results Compressive loading played a leading role on the stress redistribution at intra-articular tibial tunnel aperture. After ACL reconstruction, stress concentration occurred in the anterior and posterior regions of tunnel aperture while stress reduction occurred in the lateral and posteromedial regions under the compressive loading. Stress redistribution was partially alleviated by using the drill-guide angle ranging from 55° to 65°.

Conclusions The present study quantified the effect of bone tunnel drill-guide angle on the post-operative stress redistribution. This phenomenon potentially contributed to tunnel widening. A tunnel drill-guide angle ranging from 55° to 65° was proposed based on the biomechanical rationale. It could serve as a helpful surgical guide for ACL reconstruction.

Keywords ACL reconstruction · Tunnel creation · Tibial plateau · Stress redistribution · Finite element analysis

Introduction

Anterior cruciate ligament (ACL) reconstruction is a common operation in orthopaedic surgery to restore the stability of the ACL deficient knee [1]. Although the immediate success rates were up to 90 % regarding the postoperative stability [2], problems of tunnel widening after surgery could be up to 40 % [3, 4]. The enlarged bone tunnel would delay graft-bone incorporation, decrease graft-bone fixation strength, and complicate revision surgery [5, 6]. Tibia tunnel widening was more severe than femoral tunnel [6]. Tibial tunnel widening and graft-bone healing is one of the most important concerns for the long-term outcome of ACL surgery.

Jie Yao and Chun Yi Wen contributed equally to this work.

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The drill-guide angle, which is the angle between the tibial plateau and tibial tunnel, is a critical factor for the tunnel placement and graft fixation in anatomic ACL reconstruction. A small drill-guide angle reduced the graft-tunnel bonding [7], attenuated the bone between the tibial plateau and tunnel roof, and might cause micro-fracture in the tibial plateau [8].

Conversely, a large drill-guide angle may cause mismatch between the graft and tunnel. A large drill-guide angle is also associated with a distal positioned extra-articular starting point, which may injure the surrounding muscle tendons. In consideration of these factors, there is still a wide variety of choices in the drill-guide angle. A previous cadaver study reported the effect of the drill-guide angle on the size and morphology of the intra-articular tunnel aperture [9]. Numerical studies also suggested the important role of tunnel orientation in the graft tension and graft-tunnel in situ force [4, 10]. Yet the influence of the tunnel drill-guide angle on the post-operative stress redistribution in tibia has not been clearly understood.

Tunnel creation could cause local stress reduction and concentration adjacent to the tunnel wall [11, 12]. According to the Wolff's law, a low bone stress may trigger bone resorption, and an overloading bone stress may cause micro-damage [13]. Therefore, the tunneled knee with altered non-physiological stress redistribution may undergo undesirable bone remodeling and predispose to tunnel widening. A previous review has proposed the contribution of bone resorption to the tunnel widening at intra-articular aperture [14], at which the graft-bone healing is poorer than that at the extra-articular tunnel aperture [15]. Since the drill-guide angle could influence the tunnel orientation and graft-tunnel force, a proper drill-guide angle may manipulate the post-operative stress redistribution at tunnel aperture, and facilitate better mechanical environment for the bone-graft healing.

The objectives of this study were (1) to quantify the effect of tibial drill-guide angle on the stress redistribution at bone tunnel aperture after anatomic single-bundle ACL reconstruction; and (2) to identify the optimal drill-guide angle with an aim to minimize the post-operative stress redistribution at tunnel aperture.

Materials and methods

Magnetic resonance (MR) scanning

A male subject, aged 30 years and weighing 65 kg, volunteered for this study. The subject reported no history of knee injury, which was confirmed with physical and MR examinations. The right knee joint at full extension and at 120° flexion were scanned with MR machine (1.5 T, Siemens, Germany), at 2-mm interval, 0.47×0.47 mm^2 resolution, and T2-signal weight. A finite element (FE) model was constructed via the MR images of the full extension knee, while that of 120° flexion was used to determine the position of bone tunnels. Ethical approval was granted from the authority and the subject signed the consent with the experimental procedures explained.

FE model development

A three-dimensional (3D) FE model of the knee joint was developed by the commercial package Abaqus (Simulia Inc., USA) and was previously validated [12]. The model consisted of meniscus, cartilage, and bones meshed with four-node tetrahedral elements. The ligaments and meniscal attachments were modeled by multiple nonlinear one-dimensional elements. The material properties of the tissues were adopted from the literature (Table 1). Cartilage, ligament, and meniscal attachments were fixed at their insertion sites on the bone. Frictionless finite sliding contact algorithm was applied among femur, tibia, and meniscus.

Simulation of ACL reconstructions

The positions of the bone tunnels were determined by the knee joint model at 120° knee flexion (Fig. 1a). The femoral tunnel was drilled from the accessory anteromedial portal and through the centre of the ACL femoral footprint (Fig. 1b). The tibial tunnel was drilled from the medial side of the anterior tibial tuberosity and through the centre of the ACL tibial footprint [19]. The angle between the posterior–anterior direction and the tibial tunnel long axis in the transverse plane (transverse angle) was fixed at 25° (Supplementary material 1). The drill-guide angle was defined as the angle between the tibial plateau and the tibial tunnel long axis (Fig. 1c). The effect of different drill-guide angles—30°, 45°, 55°, 65° and 75°—were evaluated, with the tunnel diameter fixed at 9 mm. The tunneled knee at 120° flexion was manipulated to full extension according to the knee model at full extension through a geometry registration technique in Rapidform (3D Systems Korea, Inc., Korea). The tunneled knee at full extension is shown in Fig. 1d.

Tendon graft and Endobutton tape were placed in the bone tunnels. The tendon graft was fixed in the tibial tunnel 10 mm away from the tibial intra-articular tunnel aperture, which was a recommended protocol to decrease the tunnel widening [20]. The Endobutton tape was fixed at the femoral extra-articular tunnel aperture. The "Endobutton tape–graft" complex is modeled as multiple nonlinear one-dimensional elements with the same material property as the native ACL. The FE model after ACL reconstruction is shown in Fig. 1e. The interface condition between the bone tunnel wall and graft body was simulated with slip-ring algorithm in ABAQUS, which allowed the graft to move along the femoral and tibial tunnel axes.

Table 1 Material properties for the finite element (FE) model of human knee joint [11, 16–18]

Tissue	Material property	Parameters
Cortical bone	Homogeneous, linear, isotropic, elastic	Young's modulus (E)=17 GPa, Poisson ratio (ν)=0.33
Subchondral bone	Homogeneous, linear, isotropic, elastic	Young's modulus (E)=1.15 GPa; Poisson ratio (ν)=0.25
Cancellous bone	Homogeneous, linear, isotropic, elastic	Young's modulus (E)=0.4 GPa; Poisson ratio (ν)=0.33
Cartilage	Homogeneous, linear, isotropic, elastic	Young's modulus (E)=5 MPa; Poisson ratio (ν)=0.35
Meniscus	Homogeneous, transversely, isotropic, linear elastic	Circumferential modulus (E_θ)=125 MPa, radio modulus (E_R)=axial modulus (E_Z)=27.5 MPa, shear modulus ($G_{\theta R}$ and $G_{\theta Z}{}^{)}$=2 MPa, Poisson ratios: $\nu_{\theta R}$, $\nu_{\theta Z}$ and ν_{RZ} are 0.1, 0.1 and 0.33
Meniscal attachments	Homogeneous, no compression elastic	Tensile modulus (E_T)=600 MPa, compressive modulus (E_C)=0 MPa
Ligaments	Homogeneous, nonlinear, hyperelastic	$f = \begin{cases} 0\ , & \varepsilon \le 0\ , \\ \frac{1}{4}k\varepsilon^2/\varepsilon_1, & 0 \le \varepsilon \le 2\varepsilon_1, \\ k(\varepsilon - \varepsilon_1), & 2\,\varepsilon_1 \le \varepsilon\ , \end{cases}$ where f is the in situ force of the ligament, ε is the strain, ε_1 is the nonlinear strain level parameter assumed to be 0.03; the stiff parameters (k) of ACL, PCL, MCL, and LCL are 10,000, 18,000, 6,000, and 8,250 N, respectively

Loading and boundary conditions

Four loading conditions were applied. (1) A compressive force of twice the body weight was applied at the proximal femur along the axis of the femoral shaft, which commonly occurred during level walking [21]. The distal tibia was fixed in all six degrees of freedom (DOFs). (2) A valgus torque of 1.0 % body weight times meter was applied at the distal tibia, which was the maximum value in a normal gait cycle [21]. The proximal femur was fixed in all six DOFs. (3) An internal rotational torque of 1.1 % body weight times meter was applied at the distal tibia, which was also the maximum value in a normal gait cycle [21]. The proximal femur was fixed in all six DOFs. (4) A complex loading including the above-mentioned compressive, valgus, and rotational loadings together was applied on the knee joint.

Indices of stress redistribution

The post-operative von Mises stress in the tibial plateau was compared with that in the intact knee. Since the stress level in the tibial intercondylar region was lower than that in the tibial condyles, and the tunnel aperture was located in the tibial intercondylar region, the stress alteration adjacent to the tunnel aperture may be submerged. Therefore, the percentage change in stress (Δ) was used to normalize the stress alteration and indicate the degree of stress redistribution:

$$\Delta = \frac{\sigma_r - \sigma_i}{\sigma_i} \times 100\%$$

Where σ_r is the nodal von Mises stress after ACL reconstruction and σ_i is the nodal von Mises stress in the intact knee. The distribution of Δ in tibia was calculated.

There has been increasing interest in the correlation between strain energy density (SED) and bone remodeling. A low SED would cause bone loss, a high SED would promote bone growth, and an SED over a threshold may lead to bone micro-damage [22]. Therefore, the SED in the post-operative tibia was also calculated to consolidate the result of stress.

Results

Stress redistribution in tibial plateau

Post-operative stress redistribution occurred around the intra-articular tunnel aperture (Fig. 2 and Supplementary material 2). Under compressive and complex loadings, the stress was increased in the anterior and posterior regions of the tunnel aperture, whereas the stress was decreased in the lateral and posteromedial regions. Under the valgus and rotational loadings, the stress was increased in the anteromedial region of the tunnel aperture, whereas it was decreased in the posterior region. The trend of the SED redistribution was similar to the stress redistribution under all loading conditions.

Effect of drill-guide angle

The magnitude of stress redistribution was influenced by the drill-guide angle (Fig. 3). Under compressive and complex loadings, the Δ in the anterior region of the tunnel aperture decreased with increasing drill-guide angle when less than 55°, and it slightly decreased when the drill-guide angle was greater than 55°. The Δ in the posterior region of the tunnel aperture increased with increasing drill-guide angle when the drill-guide angle was less than 65°; and it slightly increased

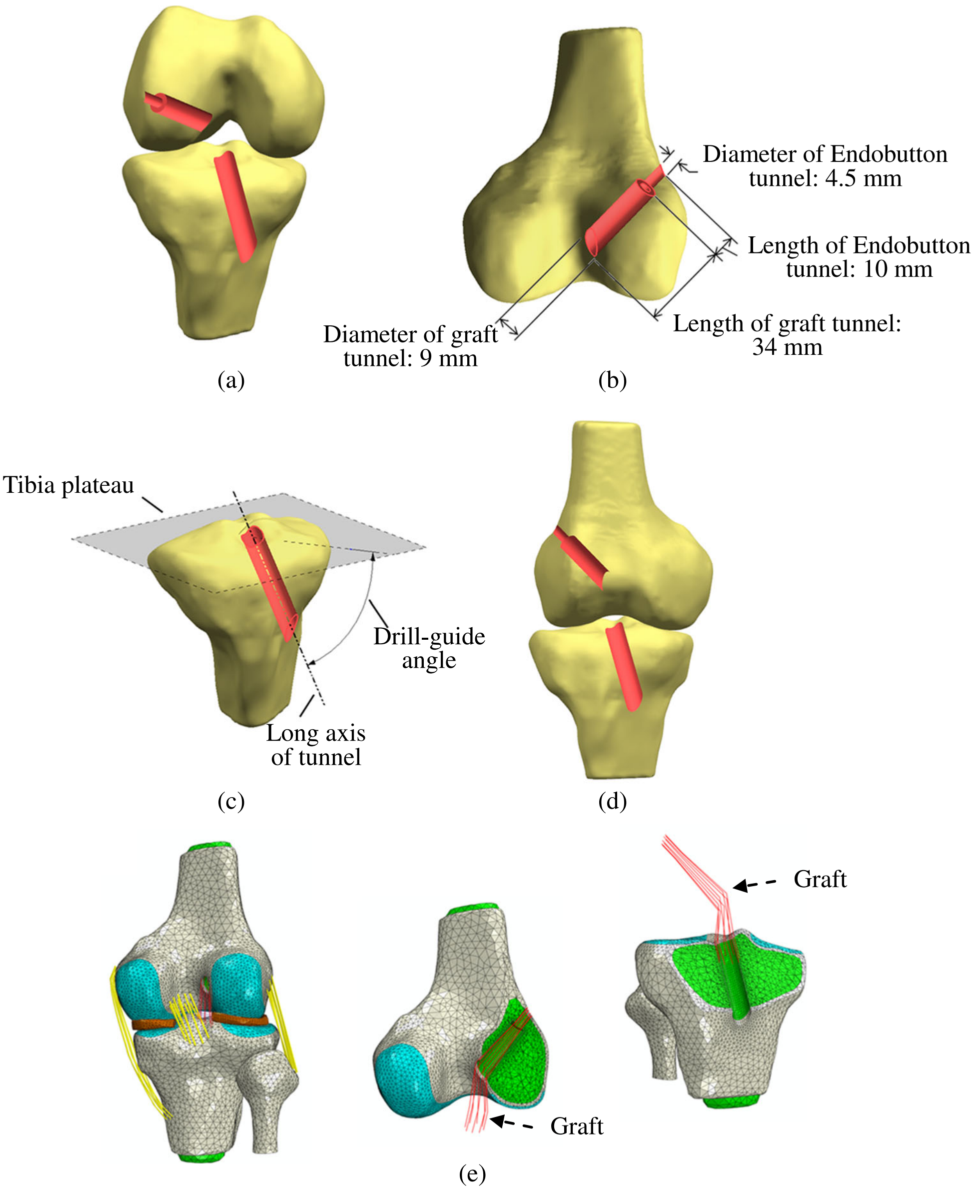


Fig. 1 Tunnel creations according to clinical specifications in anatomic single-bundle ACL reconstruction. **a** Tunnel creations on the 3D model of knee joint at 120° flexion. **b** Femoral tunnel. **c** Definition of drill-guide angle for tibial tunnel creation. **d** The tunneled knee at 120° flexion was manipulated to full extension according to the geometrical knee joint model at full extension through a geometry registration technique in Rapidform (3D Systems Korea, Inc., Korea). **e** Finite element model of knee joint after ACL reconstruction. Elements sizes of 1 mm, 2 mm and 5 mm were applied at the articular surface, the proximal tibia and distal femur, and the distal tibia and proximal femur, respectively

when the drill-guide angle was greater than 65°. At the tunnel aperture, the maximum stress and SED constantly occurred at the anterior side, and both of them decreased with increasing drill-guide angle (Supplementary material 3). The minimum stress and SED migrated from the posteromedial towards the medial side. Under valgus and rotational loadings, the Δ in the anteromedial region of the tunnel aperture decreased with increasing drill-guide angle; whereas the Δ in the posterior region of the tunnel aperture changed slightly.

The area of stress redistribution as well as the stress orientation was also influenced by the drill-guide angle. The areas of Δ≥100 % decreased with increasing drill-guide angle under all conditions. The decreasing slope became small when drill-guide angle was greater than 55° (Fig. 4). In comparison with the intact knee, the tensile stress migrated from the tibial plateau to the bone tunnel wall after surgery, and its direction varied with the drill-guide angle. The compressive stress in the tibial plateau decreased with increasing drill-guide angle when the drill-guide angle was less than 55°, whereas it maintained when the drill-guide angle was greater than 55° (Fig. 5).

The effects of the drill-guide angle on the cartilage and graft are shown in Supplementary material 4. In the medial cartilage, the stress in the anterolateral region increased, whereas the stress in the lateral region decreased. These redistributions were partially reduced with increasing drill-guide angle. The graft tension slightly changed with drill-guide angle. The bone-graft force was minimized at 55° drill-guide angle. The drill-guide angle has no influence on the meniscal attachments.

Fig. 2 Distribution of Δ in tibial plateau after ACL reconstruction. Drill-guide angle is 65°. **a**–**d** Distributions of Δ under compressive, valgus, rotational, and complex loadings, respectively

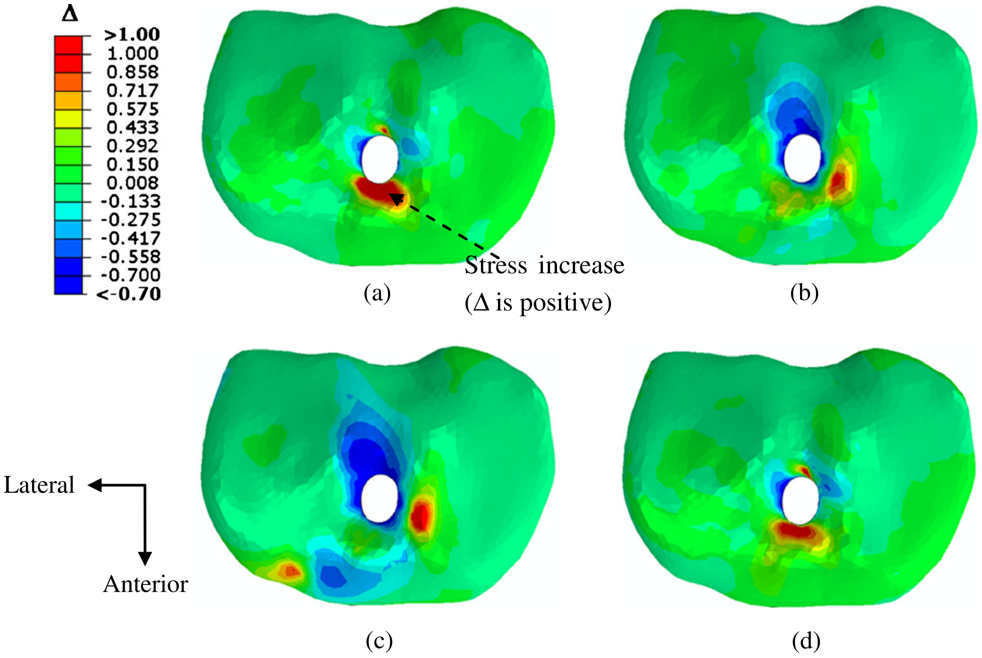
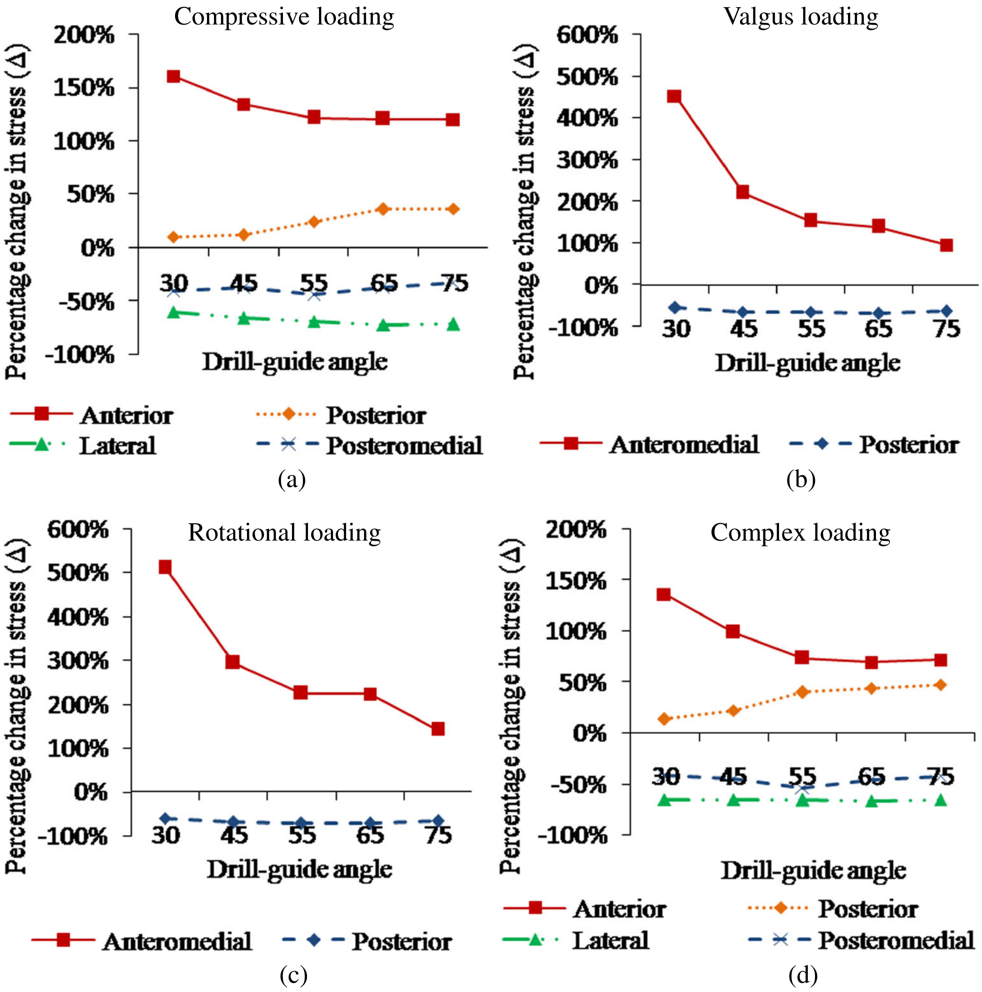
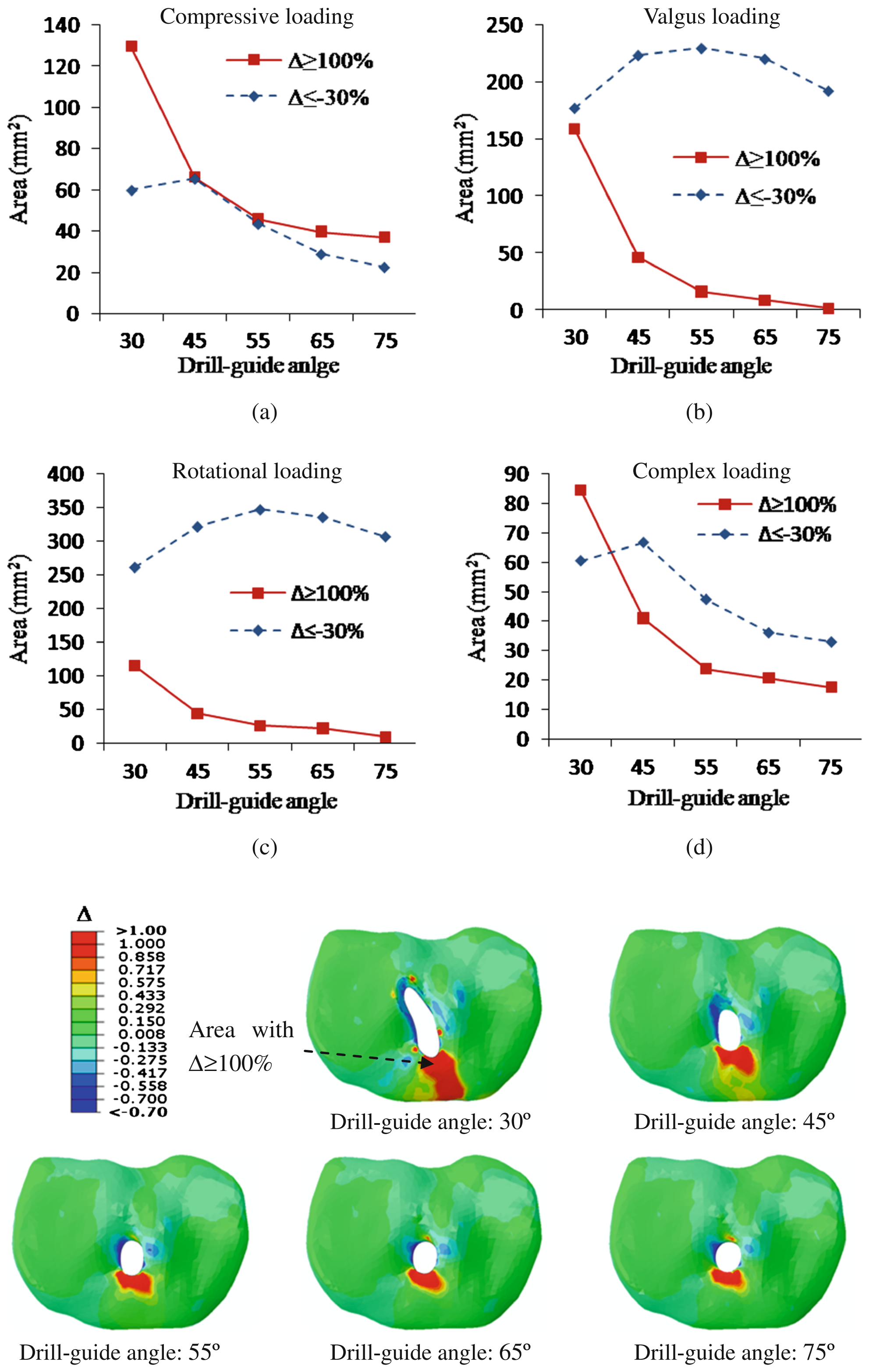


Fig. 3 Effects of drill-guide angle on Δ distribution in the tibial plateau. **a** Peak Δs in the anterior, posterior, lateral, and posteromedial regions of the tibial plateau under compressive loading. **b** Peak Δs in the anteromedial and posterior regions of tibial plateau under valgus loading. **c** Peak Δs in the anteromedial and posterior regions of tibial plateau under rotational loading. **d** Peak Δs in the anterior, posterior, lateral, and posteromedial regions of tibial plateau under the complex loading





(a) (b)

(c) (d)

(e)

Fig. 4 Effects of drill-guide angle on the areas of stress alterations. **a** Area of $\Delta \geq 100$ % at the anterior region of tunnel aperture (*red*) and area of $\Delta \leq -30$ % at the lateral and posteromedial regions (*blue*) under compressive loading. **b** Area of $\Delta \geq 100$ % at the anteromedial region of tunnel aperture (*red*) and area of $\Delta \leq -30$ % at the posterior region (*blue*) under valgus loading. **c** Area of $\Delta \geq 100$ % at the anteromedial region of tunnel aperture (*red*) and area of $\Delta \leq -30$ % at the posterior region (*blue*) under the rotational loading. **d** Area of $\Delta \geq 100$ % at the anterior region of tunnel aperture (*red*) and area of $\Delta \leq -30$ % at the lateral and posteromedial regions (*blue*) under the complex loading. **e** Distributions of Δ with drill-guide angles of 30°, 45°, 55°, 65° and 75° under compressive loading

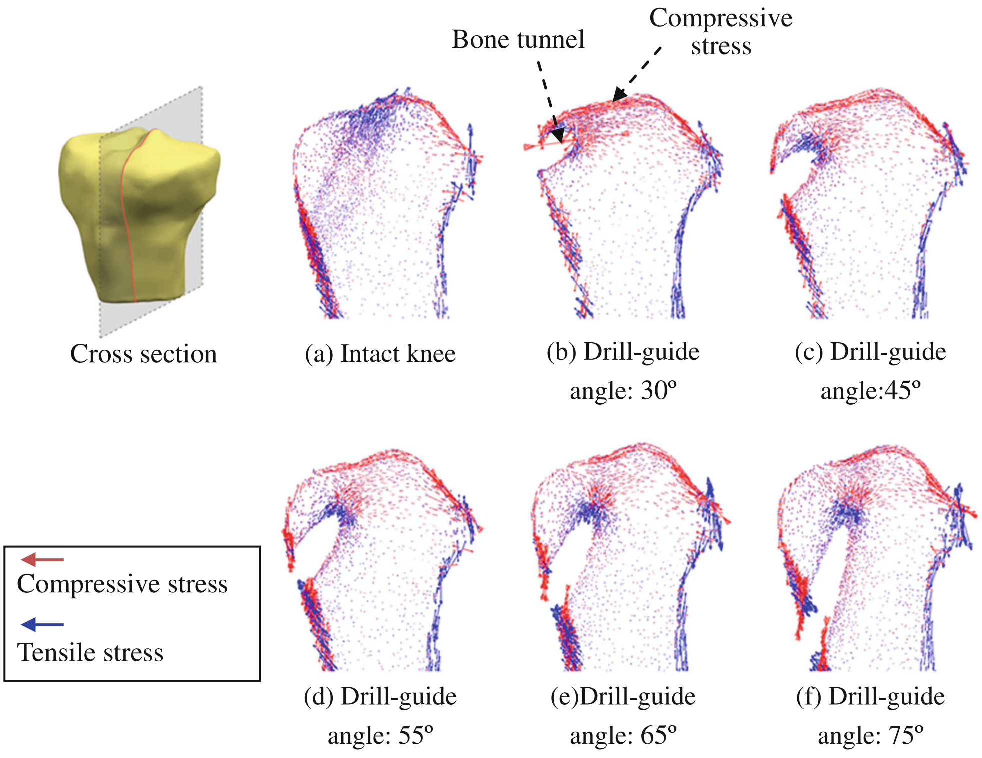


Fig. 5 Trajectories of the compressive stress (*red arrows*) and tensile stress (*blue arrows*) under the valgus loading. **a** Intact knee. The tensile stress was concentrated at the native ACL footprint in the tibial plateau. **b**–**e** Post-operative knee. The tensile stress migrated to the bone tunnel wall. The compressive stress in the tibial plateau decreased with increasing drill-guide angle when drill-guide angle was less than 55°, and it slightly changed when drill-guide angle was greater than 55°

Discussion

The present study quantified the stress redistribution in the tibial plateau after ACL reconstruction with the FE model validated. Regardless of the drill-guide angle varying, the stress redistributions under the complex loading were similar to that under the compressive loading. Since the applied loadings were maximum loadings in the normal gait cycle, these findings implied that the compressive loading played a critical role in the stress redistribution during level walking. Under the complex loading, the maximum SED in the anterior region of tunnel aperture was comparable to the threshold of bone micro-damage [22, 23]. The micro-damage may accumulate in the anterior tunnel wall, which could lead to tunnel widening in the sagittal plane. Simultaneously, the minimal SED in the posteromedial region of tunnel aperture was below the threshold of bone resorption [22, 23]. The resorption of posteromedial bone tunnel wall may lead to tunnel widening in both sagittal and coronal planes. These findings revealed a predisposing factor for the tunnel widening at tunnel aperture. These stress deprivations occurred at the anterior and posteromedial regions of the tunnel aperture, which may potentially lead to more tunnel widening in the sagittal plane compared with the coronal plane. This phenomenon is coincident with the clinical observations [4, 24].

The degree of stress redistribution could be manipulated by adjusting the tunnel drill-guide angle. Under either of the loading conditions, a small drill-guide angle severely increased the stress in the anterior region of the tunnel aperture. At 30° drill-guide angle, both the magnitude and the area of the stress redistribution were large and extended to the cartilage. Such stress redistribution was partially reduced with increasing drill-guide angle. Under the compressive and complex loadings, the maximum Δs were minimized when the drill-guide angle was equal to or greater than 55°. Under the valgus and rotational loadings, the maximum Δ also decreased with increasing drill-guide angle. Although the stress concentration in the posterior region of tunnel aperture was promoted with increasing drill-guide angle, the stress and SED in this region were lower than that in the anterior region, and did not reach the threshold of bone micro-damage in the present study. Furthermore, the compressive stress in the tibial plateau was also recovered when the drill-guide angle was equal to or greater than 55°. These findings implied that a larger drill-guide angle (≥55°) could provide a better mechanical environment at the tibial intra-articular tunnel aperture.

The areas of $\Delta \geq 100$ % reduced when the drill-guide angle was equal or greater than 55°. The areas of $\Delta \leq -30$ % also decreased with increasing drill-guide angle when this angle was greater than 55°. A small area of $\Delta \leq -30$ % was observed when the drill-guide angle was 30°. A possible explanation was that drilling the tunnel at 30° would remove a large region of bone at the posterior side of the ACL footprint, and these removed regions could not be included in the areas of $\Delta \leq -30$ %. These findings also implied that increasing the drill-guide angle could partially reduce the stress redistribution adjacent to the tunnel aperture.

The direct relationship between the drill-guide angle and surgical outcome has not been established. A small drill-guide angle may damage tissues adjacent to the tunnel aperture [9], reduce the graft-tunnel bonding [7], attenuate the bone between the tunnel roof and tibial plateau, and cause micro-fracture trauma in the subchondral bone [8]. Conversely, a large drill-guide angle led to a long bone tunnel, which may mismatch the graft length and cause a great bone removal. Given these considerations, a drill-guide angle ranging from 55° to 65° was recommended. Future studies in clinical observation are still needed to consolidate this finding.

The graft tension slightly changed with the drill-guide angle. This was similarly reported by the previous study [10]. Both the graft tension and graft-bone force remained within a safe range of the routine surgery procedure [4, 25]. The graft-bone force would even decrease with knee flexion [4]. Therefore, both the graft tension and graft-bone force were within the safe range in the condition of this study.

The percentage change in stress, Δ, was used to quantify the stress redistribution. Δ could normalize the stress alteration in both high and low stress regions. It can facilitate the comparison between the intact knee and post-operative knee. In the previous studies, stress distribution was used to estimate the post-operative mechanical effect [10, 11]. However, the stress near the tunnel aperture is much lower than those in the tibial condyles, and the stress redistribution near the tunnel aperture would be submerged in a general stress distribution. Therefore, Δ distribution was introduced in this study, and stress redistribution could be clearly observed near the tunnel aperture (Fig. 2).

There are limitations to this study. First, the effect of the drill-guide angle was analyzed with the transverse angle and diameter fixed. Coupling effects of these three factors should be investigated in our future work. Second, this study simulated the “bungee-cord” effect of the bone-graft motion, which was a contributing factor to tunnel widening [14]. Yet the “windshield-wiper” effect was not considered. Third, the material property of the tape-graft may not be as strong as the native ACL. Both drill-guide angle and graft material contributed to the tibial stress redistribution. To focus on the effect of drill-guide angle, and to eliminate the confounding effect of graft material, the material properties of the tape-graft and ACL were assumed to be the same. Fourth, the effect of graft fixation was not analyzed. The graft was fixed at the extra-articular tunnel aperture. It had little influence on the stress redistribution near the articular surface. Finally, we investigated the knee joint at extension under four typical loading conditions. These are typical loadings in daily activities. However, muscle force and dynamic effect should be considered in the future study.

Conclusion

This study quantified the post-operative stress redistribution near the intra-articular tunnel aperture after ACL reconstruction. This phenomenon potentially contributed to tunnel widening. A drill-guide angle ranging from 55° to 65° was recommended to alleviate the stress redistributions in the tibial plateau. Our findings provided biomechanical insight into the post-operative tunnel widening. The tunnel drill-guide angle ranging from 55° to 65° was proposed from the biomechanical rationale, and could serve as a helpful surgical guide for ACL reconstruction.

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Conflict of interest We declare that we have no financial and personal relationships with other people or organizations that can inappropriately influence our work; there is no professional or other personal interest of any nature or kind in any product, service and/or company that could be construed as influencing the position presented in, or the review of, the manuscript entitled.

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