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Original Article

Systematic effects of femoral component rotation and tibial slope on the medial and lateral tibiofemoral flexion gaps in total knee arthroplasty

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A R T I C L E I N F O

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Physiological knee motion
Surgical technique
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A B S T R A C T

Purpose: To quantify the effects of the systematic internal and external femoral component rotations and tibial slope on the medial and lateral tibiofemoral gaps in total knee endoprostheses.
Methods: Nineteen knee cadaver specimens with an intact ligament apparatus were fixed in a custom frame, facilitating physiological flexion motion. Virtual total knee arthroplasty (TKA) was performed on three-dimensional models obtained from computed tomography scans (0° and 90° flexions) with systematically altered femur rotations and tibial slopes.
Results: Both the femur rotation and the tibial slope influenced the medial and lateral tibiofemoral flexion gaps ($p < 0.001$), and the effects differed between the medial and lateral sides ($p < 0.001$). The medial tibiofemoral flexion gap increased by 2.90 ± 0.34 mm and decreased by 2.66 ± 0.26 mm for 7° external and internal femur component rotations, respectively (both with $p < 0.001$). The lateral tibiofemoral flexion gap decreased by 3.11 ± 0.31 mm and increased by 3.29 ± 0.33 mm for 7° external and internal femur component rotations, respectively (both with $p < 0.001$).
Conclusion: For established surgical methods, we recommend a neutral femur rotation for a 0° tibial slope and a 3° external femur rotation for a tibial slope of 9–10°. The translational potential of this research shows that while the rotation of the femoral component in extension has no effect on the gap size, owing to the axis of rotation being perpendicular to the gap, for a 90° flexion, we not only observe differences in the gap size between the medial and lateral but also unequal differences on either side depending on the inward or outward rotation. The main reason for this is the position of the axis of rotation, which is not precisely half way between the lateral and medial contact points. The results show that rotation of the femoral component always creates an unbalanced flexion gap.
The translational potential of this article: The article points out the differences in the tibiofemoral gap in total knee endoprostheses due to the systematic internal and external femoral component rotation. While in lower leg extension there are no differences seen, in 90° knee-flexion there are unequal differences within the medial and lateral compartment that show a mathematical relationship towards the femoral compartment rotation which needs to be intraoperatively considered.

Introduction

In recent years, arthroplasty has become a standard treatment for degenerative changes in the knee. The incidence of knee osteoarthritis is higher than that of hip osteoarthritis [1], and hence, increasing numbers of knee arthroplasties are to be expected. In contrast to the stable numbers of arthroplasties for treating posttraumatic knee degeneration and rheumatoid arthritis, the number of arthroplasties for the treatment of osteoarthritis has increased exponentially [2].

Arthroplasties are typically performed with the goal of reducing pain, correcting deformities and improving functionality, and outcomes are influenced by several factors, including sex, age, body weight, surgical indication, surgical technique, technical equipment and surgeon skills [3, 4]. Moreover, the orientation of the prosthetic components plays a

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critical role in the lifetime of a prosthetic implant [5]. The poorer outcomes of knee arthroplasties compared with hip arthroplasties [6] are presumably related to the greater complexity of the anatomy and consequently the biomechanics of the knee compared to the hip.

Most knee endoprostheses are designed to replicate the native human knee. The prosthetic components replace the articulating surfaces within a maintained system, comprising ligaments, muscles and tendons. The component positioning and size influence the interplay between the artificial implant and the native system, and any malpositioning affects the loading of the bone–prosthesis interface and the tension of the knee ligaments. Moreover, malpositioning results in abnormal knee kinematics and ultimately stiffness, instability and implant loosening [7]. Previous studies have described the consequences of implant malpositioning and the relationship between component positioning and surgical outcomes. For instance, the rotation of the femoral component may influence the knee stability, tibiofemoral and patellofemoral kinematics and alignment during flexion. These changes may result in diminished flexibility and anterior knee pain [8–11]. For instance, an asymmetric tibiofemoral flexion gap leads to instability during flexion [12–14]. However, to date, the relationships between the degrees of the femoral component rotation and tibial slope on the tibiofemoral gap have not been systematically investigated.

The purpose of this study was to quantify the effects of systematic internal and external femoral component rotations and the tibial slope on the medial and lateral tibiofemoral gaps in total knee endoprostheses.

Materials and methods

Specimen and frame fixation

In this investigational cadaver study, as a continuative project of the study of Nowakowski et al. from 2014 [15], the same samples consisting of 19 formalin-fixed knee cadaver specimens (all Caucasian) were further analysed from the repository of the Department of Anatomy. The convenience sample included 10 male and nine female donors [five left and five right knees for men, four left and five right knees for women; mean $\pm$ 1 standard deviation (SD), age 81.0 $\pm$ 12.1 years; body mass 68.7 $\pm$ 13.9 kg; height 165.2 $\pm$ 10.4 cm]. Specimens with scars, previous surgery, malalignment or deformities or signs of osteoarthritis were excluded from the study. This study was approved by the Institutional Review Board and conducted in accordance with the Declaration of Helsinki.

In each specimen, the femur and tibia were severed at 15 cm proximally and distally from the joint space, and skin, subcutaneous and muscle tissue were mostly resected. The popliteus muscle was saved. A 6-mm hole was drilled in the femur, 30 cm distal of the femur stump, parallel to the joint axis in the medio-lateral direction. Two 6-mm holes were drilled into the tibia, proximal of the tibia end, with a 30-mm interhole distance parallel to the joint space in the mediolateral direction. These holes were utilised to attach a custom acrylic-carbon frame to fix the specimen during computed tomography (CT) imaging (Figure 1A). Two hinges in the frame centre allowed the frame to be flexed in the sagittal plane and set to any position by using two octagonal bolts (Figure 1B). The proximal femur was fixed with a carbon rod and fed into a plastic casing (Figure 1C), which allowed for internal/external rotations and medial/lateral and proximal/distal translations, facilitating physiological ligament guided knee motion. The tibia was statically fixed to the frame using two carbon rods (Figure 1C). The quadriceps tendon was attached to the frame using an elastic rubber band to allow sliding in the patellofemoral joint during flexion.

Computed tomography

A 16-row CT scanner (Lightspeed 16, GE Healthcare, Little Chalfont, United Kingdom; slice thickness 0.63 mm) was utilised to obtain CT scans of each knee, both in 0° extension and in 90° flexion. Three-dimensional (3D) reconstructions of the scan images were obtained using VGStudio MAX (Volume Graphics GmbH, Heidelberg, Germany) [16], to enable the calculation of virtual cuts at the joint lines in extension and flexion. Because we could not establish the mechanical axes in the specimens, we defined a coordinate system according to Grood and Suntay [17] and McPherson [18]. The rotation is smallest in full extension, and the coordinate system was defined in this position. Two circles were fit to the medial and lateral dorsal femur condyles, respectively, in the sagittal plane, and the line connecting the centres of these circles (the flexion/extension axis) and a tibia reference point (the intersection of a line parallel to the posterior aspect of the femur cortex, with the posterior tibial cortex at the height of the proximal tibia hole [18] were used to define the frontal plane). The sagittal plane was defined orthogonally to the frontal plane containing the tibia reference point, the transverse plane by the flexion/extension axis orthogonal to the other two planes and the origin as the intersection of all planes. Pairs of data sets in extension and flexion were rendered by matching the identical voxel data of the tibia to obtain a combined data set oriented along the coordinate system [15]. The medial and lateral tibiofemoral contact points were identified in both extension (0°) and flexion (90°) according to Wretenberg et al. [19].

Bone cuts and virtual slope variation

Virtual bone cuts in accordance to the average prostheses space needed during total knee arthroplasty (TKA) were performed on the 3D data sets following the method of Cheng et al. [20] and as previously described by Nowakowski et al. [15]. Briefly, the tibia was sectioned orthogonally to the coordinate axis defined above, at a resection depth of 6 mm (the average tibia plateau component thickness), which was defined as a 0° slope. The distal femur was cut in extension parallel to this plane, with a resection depth of 9 mm from the most distal point of the medial femoral condyle (average femoral component thickness). The gap between the sectioned femur and tibia was defined as the prosthesis gap,

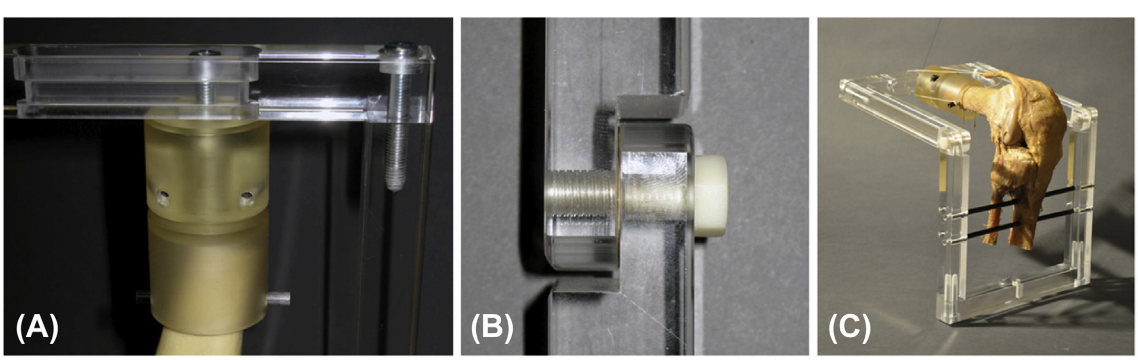

Figure 1. Experimental setup. (A) Plastic casing with three degrees of freedom (rotation, translation and separation) holding the femur stump. (B) Hinge of the acrylic frame with octagonal bolts. (B) Knee specimen in 90° flexion.

which differed from the usual extension and flexion gap in that it was not defined in distraction and corresponded only to the space required for the implanted prosthesis. This gap was identical in extension and flexion with a 0° tibial slope in all specimens. Similarly, the posterior cut was performed in 90° flexion, at the same resection depth of 9 mm. The longest mediolateral diameter in the 0° tibial slope plane was identified, and its midpoint was measured and projected orthogonally onto the ventral tibia edge to obtain the centre of rotation for the simulated tibial slopes. Virtual tibial slopes of 0°, 3.5°, 7° and 10° were generated [15].

Virtual femur component rotation

The femur component is usually placed in reference to an intramedullary rod. Based on the literature, we assumed that the anatomical axis passes 3 mm medially to the femoral notch (literature: 2–5 mm medial [2, 63, 64]). We identified the most distal two-dimensional axial slice of the extended femur showing the anterior and posterior cortical bone and translated the midpoint of the transcortical line by 3 mm to the medial in the frontal plane. The axis of rotation (anatomic axis of rotation) was defined using this point and the midpoint of the mark canal 80 mm proximal. The intersection of the axis of rotation and the distal cut was defined as the centre of rotation for rotating the section planes. This point was transferred to the flexed femur using the volumetric data. The perpendicular projection of this point onto the posterior section plane corresponded to the lever arm of the rotation of the posterior section plane. Hence, we could apply the virtual cuts to the prostheses according to the intraoperative procedure. We virtually rotated the femur by −7°, −5°, −3° (external rotations), 0°, 3°, 5° and 7° (internal rotations) and calculated the medial and lateral tibiofemoral gaps as the distances between the medial and lateral tibiofemoral contact points, respectively, in flexion but not extension, because the femoral rotation does not affect the tibiofemoral gap in extension. Fourteen data points were obtained for each specimen (seven for each of the lateral and medial tibiofemoral gaps in 90° flexion) for each of the tibial slopes 0°, 3.5°, 7° and 10° for each specimen (Figure 2). The accuracy of our measurements was less than 0.01 mm.

Statistical analysis

All statistical tests were conducted with SPSS version 22.0 (IBM Corporation, Armonk, NY). All data were tested for normalcy and homogeneity using Kolmogorov–Smirnov tests. The means and standard deviations (SDs) were calculated. Differences in the magnitude of the tibiofemoral flexion gap were detected using an analysis of variance for repeated measurements of the sides (medial and lateral), tibial slope and femur component rotation as subject factors. Linear regression analyses were utilised to test for a significant relationship between the magnitude of the tibiofemoral flexion gap and the femur component rotation angle for each tibial slope. Differences in the regression slopes between the medial and lateral compartments were detected using analyses of covariance. The significance level for all statistical tests was set a priori to 0.05.

Results

The tibiofemoral flexion gaps were identical in the medial and lateral compartments for a 0° rotation and 0° slope (Tables 1 and 2). The effect of the femur component on the medial and lateral tibiofemoral flexion gaps increased with an increasing rotation and increasing tibial slope (Figure 3; Tables 1 and 2). The medial tibiofemoral flexion gap increased by 2.90 ± 0.34 mm and decreased by 2.66 ± 0.26 mm for 7° external and internal femur component rotations, respectively (both with $p < 0.001$; Table 1). The lateral tibiofemoral flexion gap decreased by 3.11 ± 0.31 mm and increased by 3.29 ± 0.33 mm for 7° external and internal femur component rotations, respectively (both with $p < 0.001$; Table 2). The effect of the tibial slope on the tibiofemoral flexion gap was stronger than that of the femur component rotation ($p < 0.001$).

The effect of the tibial slope on the tibiofemoral flexion gap was stronger on the medial compartment than on the lateral compartment ($p < 0.001$). Furthermore, the effect of the femur component rotation on the tibiofemoral gap in the medial compartment acted in the opposite direction than that in the lateral compartment ($p < 0.001$). Hence, we utilised separate linear regression models for the medial and lateral compartments.

The linear regression equation for the medial tibiofemoral flexion gap ($TFG_{flexion,\ medial}$) was

Table 1
Mean (one standard deviation) medial tibiofemoral flexion gap (mm) depending on the tibial slope and femur component rotation.

Medial tibiofemoral flexion gap [mm]			Tibial slope			
			0°	3.5°	7°	10°
Femur component rotation	External rotation	−7°	22.19 (2.88)	23.41 (3.84)	25.36 (4.70)	28.16 (2.86)
		−5°	21.39 (2.89)	22.56 (4.34)	24.51 (5.26)	27.36 (2.87)
		−3°	20.58 (2.91)	21.69 (4.87)	23.64 (5.38)	26.55 (2.88)
	Neutral	0°	19.29 (2.95)	20.32 (5.73)	22.27 (5.77)	25.26 (3.32)
	Internal rotation	3°	17.94 (2.98)	19.21 (5.41)	21.16 (6.13)	23.92 (2.91)
		5°	16.99 (3.02)	18.42 (5.20)	20.37 (5.81)	22.97 (2.95)
		7°	16.00 (3.06)	17.59 (5.00)	19.55 (5.59)	21.97 (2.98)

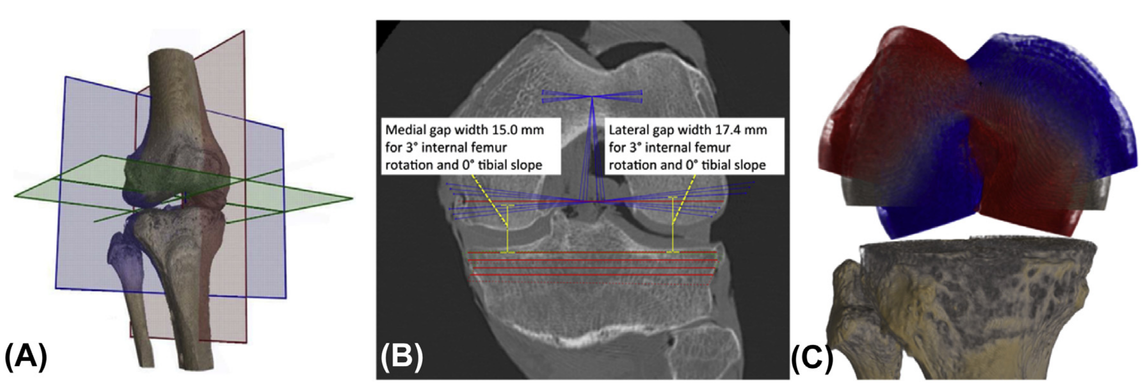

Figure 2. (A) Coordinate system defining the axes of rotation. (B) Measurement of medial and lateral tibiofemoral flexion gaps (yellow) in virtually rotated femur cuts (blue) with different tibial slopes (red). (C) Illustration of the posterior femoral cut in neutral, internal (red) and external (blue) femur rotation in flexion.

Table 2
Mean (one standard deviation) lateral tibiofemoral flexion gap (mm) depending on the tibial slope and femur component rotation.

Lateral tibiofemoral flexion gap [mm]			Tibial slope			
			0°	3.5°	7°	10°
Femur component rotation	External rotation	−7°	16.16 (3.00)	18.58 (5.16)	21.38 (5.77)	24.86 (3.32)
		−5°	17.09 (2.98)	19.36 (5.39)	22.16 (6.01)	25.79 (3.30)
		−3°	18.00 (2.96)	20.11 (5.62)	22.91 (6.25)	26.69 (3.29)
	Neutral	0°	19.27 (2.94)	21.17 (5.96)	23.97 (6.60)	27.97 (3.28)
	Internal rotation	3°	20.46 (2.95)	22.45 (5.15)	25.25 (5.76)	29.16 (3.28)
		5°	21.21 (2.96)	23.27 (4.64)	26.07 (5.22)	29.91 (3.29)
		7°	21.94 (2.97)	24.06 (4.16)	26.87 (4.70)	30.63 (3.29)

$$TFG_{flexion,medial} = TFG_{0,0} + 0.597 \cdot TS - 0.441 \cdot FCR \quad (1)$$

where $TFG_{0,0}$ is the tibiofemoral gap in extension for a 0° tibial slope and 0° femur component rotation, tibial slope (*TS*) is the tibial slope and femur component rotation (*FCR*) is the femur component rotation (both in degrees). This regression model explained 97.3% of the variability in the medial tibiofemoral flexion gap ($p < 0.001$).

The linear regression equation for the lateral tibiofemoral flexion gap ($TFG_{flexion,\ lateral}$) was

$$TFG_{flexion,lateral} = TFG_{0,0} + 0.868 \cdot TS + 0.412 \cdot FCR \quad (2)$$

This regression model explained 97.4% of the variability in the medial tibiofemoral flexion gap ($p < 0.001$). Combining equations (1) and (2), the tibial slope and femoral component rotation combination resulting in balanced tibiofemoral flexion gaps in the medial and lateral components can be calculated using the following equation:

$$FCR = -0.381 \cdot TS \quad (3)$$

This relationship is illustrated in Figure 4.

The orientation of the anatomical transepicondylar axis did not significantly differ from the sagittal orientation of the tibia cutting plane ($p > 0.05$).

Discussion

The purpose of this study was to quantify the effects of systematic internal and external femoral component rotations on the medial and lateral tibiofemoral gaps in total knee endoprostheses. All bone cuts were performed virtually on 3D CT image sets, which were aligned to a standardised coordinate system. We resected sufficient bone to facilitate placement of the prosthetic components. The difference between the resulting gap and typical flexion-extension gaps is that the former was not measured in distraction, but only in terms of the dimensions of the endoprostheses. We defined this gap as the prosthetic flexion-extension gap. In the initial setting (0° slope and 0° femoral rotation), the medial and lateral tibiofemoral gaps in extension and flexion were identical. Averaging over all specimens, the virtually resected gap was 19.3 mm (SD 2.9 mm).

A symmetric knee prosthesis would fill this entire gap. Under distraction, the typical flexion-extension gap would occur assuming that the ligament tension would not be affected by the resection. This phenomenon can be simplified by the equation [15] flexion-extension gap = deformation of the surrounding soft tissue + bony resection gap. Assuming an ideal soft tissue mantle, perfect orientation of the components and a neutral alignment, a prosthesis would take up the entire gap around the tibiofemoral contact points throughout the entire movement cycle. Hence, the prosthetic gap corresponds to the idealised bone resection gap [15].

The asymmetric flexion-extension gap under a load should be equivalent to the prosthetic gap around the tibiofemoral contact points, with corresponding laxity or stretching of the surrounding soft tissue. As previously demonstrated [15], our model exhibited similar physiological

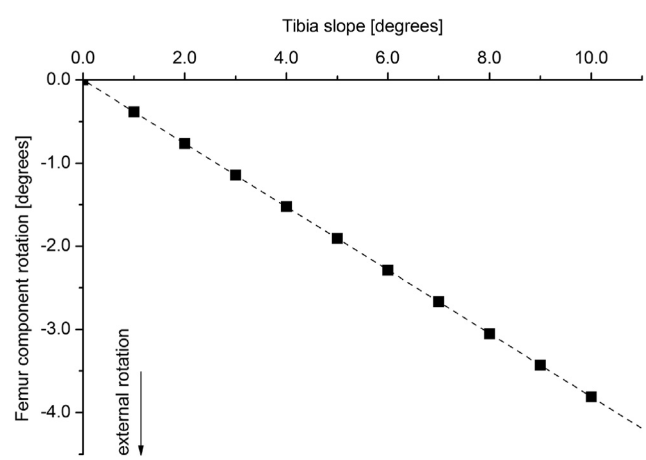

Figure 4. Illustration of the combinations of femoral component rotation and tibial slope that facilitated balanced medial and lateral tibiofemoral flexion gaps in this cadaver study.

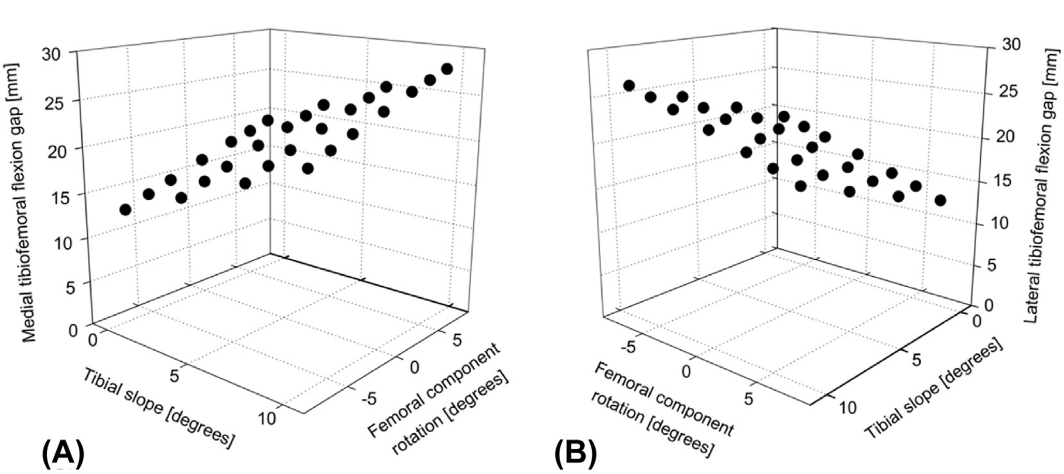

Figure 3. (A) Medial tibiofemoral flexion gaps dependent on the tibial slope and femoral component rotation. (B) lateral tibiofemoral flexion gaps dependent on the tibial slope and femoral component rotation. Exemplary specimen (specimen #1). "−" external rotation; "+" internal rotation. Note that the graphs are rotated differently for optimal visualiszation.

motion in our specimen. In agreement with the results reported by Freemen and Pinskerova [21], our joints exhibited a significantly greater rollback in the lateral compartment than the medial compartment around the tibiofemoral contact points during flexion from 0° to 90°.

Based on measurements in physiologic knees, the results of this study revealed a combined effect of the femur component rotation and tibial slope on the medial and lateral tibiofemoral flexion gaps. In the native joint, the joint gaps are determined by the combination of the external femur rotation, appropriate tibial slope and femur rollback. Tokuhara et al. [22] reported that the lateral joint gap opened by 6.7 mm, and the medial joint gap opened by only 2.1 mm under varus or valgus stress. Typically, total knee endoprostheses do not facilitate femur rollback or exhibit paradoxical movement [23]. Moreover, it is generally accepted that the femur component should be implanted in a 3° external rotation. Most procedures aim to achieve balanced medial and lateral tibiofemoral flexion gaps [8–11]. Based on the results of this cadaver study, a tibial slope of 9–10° would be necessary to achieve a balanced flexion gap with a 3° externally rotated femur component, which is greater than the values recommended in the literature (5–7°) [24,25]. Recently, patient-matched implants have been introduced to the market, where preoperative CT or magnet resonance images of the leg axes and knee are utilised for surgical planning. Individual bone geometry is considered to produce cutting blocks that perfectly match the individual knee. While this technique can increase the precision of placing the components in the correct rotation, it does not solve the problem of lacking femur rollback [26].

The effects of the femur component rotation on the tibiofemoral gap during knee flexion on gap widening were stronger than on gap narrowing with the same rotation angles. These results suggest that the axis of the femoral component rotation was not located at the centre between the medial and lateral tibiofemoral contact points. In our study, we employed the intramedullary axis as the axis of rotation to match the most commonly employed surgical techniques for placing the femur component. Using this method, the point of rotation was located anterior to the posterior cut surface, and hence, the posterior section plane was pivoted perpendicular to the section plane, explaining the greater change in the tibiofemoral flexion gap on the side that is rotated. Moreover, the greatest absolute change in the tibiofemoral gap was observed in the lateral compartment for an internal femur component rotation. Hence, the surgeon should be aware of the eccentric location of the centre of rotation in the intramedullary orientation of the cutting plane, and the resulting different relative effects of modulated resection planes on the medial and lateral flexion gaps.

In our study, the tibiofemoral flexion gap was balanced for a 0° femur component rotation and 0° tibial slope. Both internal and external femur component rotations resulted in an unbalanced tibiofemoral flexion gap in the medial and lateral components, i.e., the gap was greater in either the medial or lateral compartment. During knee motion from extension to flexion, the lateral tibiofemoral contact point moved more for the dorsal (16.3 mm) than the medial tibiofemoral contact point (4.4 mm) [15,21]. Usually, the femur component is implanted in a 3° external rotation, to account for the 3° varus slope of the physiological tibia articulating surface, which is already implemented in some tibia component designs. This varus deformity is compensated for by the prominence of the medial femur condyle; this condylar twist of approximately 3°–4° corresponds to the epicondyle axis internal femur rotation relative to the tibia [27]. Because a horizontal cut is usually applied to the tibia to avoid varus or valgus loading of the tibia component, it has been generally assumed that this 3° external femur component rotation is necessary to facilitate normal patellofemoral alignment [28,29]. Owing to the greater rollback in the lateral compartment, the decrease in the lateral tibiofemoral gap is smaller with an increasing tibial slope. Hence, combinations of specific femoral component rotations and tibial slopes result in balanced mediolateral tibiofemoral flexion gaps, as shown in equation (3). Moreover, the result of a close-to-parallel orientation of the anatomical transepicondylar axis to the tibial cutting agrees with the existing literature [10,30] and suggests that the transepicondylar axis may be more reliable and accurate than other axes for determining the rotation of the femoral component of total knee endoprostheses [7].

The major limitation seen in this study is seen in the discrepancy between the used model and the presented situation of knee for TKA. The investigation performed on knee samples representing a healthy knee-joint situation does not mirror the initial situation before TKA. Our model does not resemble the smaller gap changes because of contraction of the soft tissue envelope in an osteoarthritis knee situation. Nor does it consider the effects of relaxation secondary to anesthesia and/or applied Esmarch ischemia. Nevertheless, since a biomechanically influenced behavior of bone as well as strict bone resection associate gap behavior is discussed and described, we believe to be accurate in our description of bone movement in regards.

Conclusion

In conclusion, the results of this study demonstrate that both the femur component rotation and tibial slope influence the medial and lateral tibiofemoral flexion gaps and that the effects differ between the medial and lateral sides. These effects also depend on the specific tibial slope. Hence, for established methods, we recommend a neutral femur rotation for a 0° tibial slope and a tibial slope of 9°–10° for a 3° external femur rotation.

Conflicts of interest statement

The authors have no conflicts of interest relevant to this article.

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