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High posterior tibial slope increases anterior cruciate ligament elongation in unicompartmental knee arthroplasty during early-flexion of lunge

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Abstract

Background In-silico and in-vitro studies have revealed an appropriate posterior tibial slope (PTS) is critical for normal anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) tension and knee biomechanical behavior of unicompartmental knee arthroplasty (UKA). However, the effects of PTS on in-vivo elongation of ACL and PCL in UKA remains unknown. The study aimed to quantify in-vivo ACL and PCL elongations during lunge and analyze their relations with PTS.

Methods Thirteen fixed-bearing (FB) and 11 mobile-bearing (MB) UKA patients were recruited. The postoperative medial PTS was defined as the angle between the tibial transverse plane (perpendicular to mechanical axis) and cut plane. Accurate knee spatial postures of UKA and contralateral native knees during single-leg lunge were measured by the dual fluoroscopic imaging system. The ACL (AM, PL bundles) and PCL (AL, PM bundles) footprints were determined based on anatomical features on femoral and tibial 3D surface model reconstructed from CT. A validated 3D wrapping method was used to measure ligament bundle length. The paired Wilcoxon signed-rank test was used to analyze the ligament elongation difference between bilateral knees. The Spearman correlation between PTS and average ligament elongation difference (ACL during 0–30° early-flexion, PCL during 60–100° deep-flexion) was calculated.

Results The elongation of FB UKA PCL double-bundle was larger than contralateral sides in most flexion range of lunge (Max-Difference: AL 7.6 ± 8.7%, PM 8.2 ± 5.1%, $p < 0.05$). In contrast, ACL double-bundle elongations of MB UKA in mid-flexion were larger than contralateral sides (Max-Difference: AM 8.0 ± 8.1%, PL 7.6 ± 9.8%, $p < 0.05$). The increased PTS was significantly relevant to the increased ACL double-bundle elongation difference of bilateral knees for both FB and MB UKA patients ($R > 0.6$, $p < 0.05$).

Conclusion There was abnormal in-vivo elongation of PCL in FB UKA and ACL in MB UKA during lunge and cause over-constraints to the contralateral knee. There was a positive correlation between PTS and ACL elongation

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difference for both FB and MB UKA, indicating excessive PTS should be avoided to preserve native ACL function in further UKA implantation.

Levels of Evidence III.

Keywords Unicompartmental knee arthroplasty, In-vivo ligament elongation, Cruciate ligament, Fluoroscopy, 2D-to-3D registration

Introduction

For treating isolated osteoarthritis (OA) in the single compartment of the knee, unicompartmental knee arthroplasty (UKA) and total knee arthroplasty (TKA) are two common surgical options, and the number of UKA procedure has increased because of several advantages [1, 2]. However, the main clinical problem of UKA is the high revision rate as three times as TKA [3, 4]. There are two types of UKA systems in clinical practice: fixed-bearing (FB) and mobile-bearing (MB), which have different bearing designs with different concepts. However, which type of bearing is more suitable for OA patients remains controversial based on existing knowledge [3, 5–13]. These contradictory results may be attributed to the current lack of biomechanical knowledge on tibiofemoral kinematics and peripheral tissue function of UKA knees in daily activities.

The normal functional anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) are for knee anteroposterior and rotation stabilities during extension-flexion motions to preserve natural biomechanical environment and prevent knee injury [14–16]. ACL tightens mainly during knee early-flexion to restrict excessive anterior tibial translation while PCL showed function mainly during knee deep-flexion range to ensure normal femoral rollback [14–16]. The ACL deficiency increases the knee instability and revision rate compared with ACL intact knee after UKA procedure, therefore the ACL deficiency is considered as a contraindication for UKA in current popular concepts [17–19]. However, recent studies have investigated that the integrity of ACL does not affect the postoperative survival rates and knee kinematics patterns during daily activities for UKA patients [20–22]. This opposite conclusion may be due to the lack of ACL function changes after UKA surgery in current knowledge. On the other hand, the normal functional PCL can ensure the natural stability of UKA knees by in-silico simulation [23], and long-term results revealed that PCL deficiency can increase tibiofemoral displacements and finally non-surgical compartmental OA development for UKA patients [24]. However, no previous in-vivo studies have verified the postoperative PCL variations so far. Therefore, it's important to investigate the in-vivo cruciate ligament elongation during activities for UKA patients to explore the biomechanical functional difference between ACL and PCL following FB and MB UKA.

The posterior tibial slope (PTS) following UKA affects long-term survivorship, clinical outcomes, and knee function. In-silico models indicated that a greater PTS increases ACL tension of FB UKA during deep knee bend and gait [25]. Another in-silico study revealed that increased PTS of MB UKA was associated with increased non-surgical compartmental contact stress and ACL stress during gait, which may result in OA progression and ligament failure [26]. Furthermore, the decreased PTS was associated with PCL injury and posterior tibial translation [27, 28]. Appropriate increased PTS in FB UKA with PCL deficiency can ensure normal posterior stability and patellofemoral contact stress [23]. Hernigou and Deschamps suggested that the PTS should be less than 7° to achieve better outcomes in a retrospective clinical review of the 99 UKA cases after a mean follow-up period of sixteen years [29]. The same PTS range was recommended by Suzuki et al. for MB UKA [30]. In a cadaveric study on FB UKA knees, the optimum PTS for knees intact ACL, partial and complete ACL deficiency were 5–8°, 5–7°, and 5–6°, respectively [31]. However, whether the current suggested PTS range is optimal for in-vivo ACL and PCL function remains unknown based on current knowledge.

The purpose of the study is (1) to quantify and compare the in-vivo elongation pattern differences of ACL and PCL during weight-bearing deep lunge between UKA and contralateral native knees for FB and MB UKA patients and (2) to analyze the effect of PTS on average ACL elongation difference during early-flexion and PCL elongation difference during deep-flexion between UKA and contralateral knees. We hypothesize that the elongation patterns of ACL and PCL of FB and MB UKA knees were different from contralateral knees, and the cruciate ligament elongation difference between bilateral knees was significantly associated with PTS.

Methods

Patient demographic data

The study protocol was approved by the Institutional Ethics Committee of Shanghai Sixth People's Hospital Affiliated to Shanghai Jiao Tong University School of Medicine (No.2017-084). Thirteen-three patients suffering from medial compartmental OA in unilateral knee underwent randomly medial FB or MB UKA. The inclusion criteria were as follows: (1) isolated medial compartments OA with Kellgren-Lawrence (KL) grade 3–4 [32]; (2) aging from 50 to 80 years; (3) intact functional ACL and PCL. The exclusion criteria are as follows: (1) severe postoperative knee pain or significant muscle weakness; (2) severe neurological deficit or symptoms; (3) any postoperative complications and musculoskeletal disease; (4) any surgical history of the contralateral lower limb; (5) contralateral knee KL grade 2–4; (6) postoperative flexion contracture or knee full extension inability to 0°; (7) maximum weight-bearing knee flexion angle less than 100°. The integrities of the ACL and PCL were examined before and after UKA implantation in surgery. Besides, the ACL laxity was examined by manual front drawer and Lachman tests during the follow-up period, while the PCL laxity was examined by posterior drawer and Godfrey's tests [33, 34]. Due to the COVID-19 epidemic and exclusion criteria, nine patients were excluded from postoperative follow-up and experiment. Finally, we enrolled 13 FB UKA patients (3 males, 10 females) implanted by Restoris MCK partial knee system (Stryker, USA) and 11 MB UKA patients (4 males, 7 females) implanted by Oxford Phase 3 system (Zimmer Biomet, USA). All surgeries were performed by one senior surgeon. FB UKA was implanted under the guidance of MAKO surgical robot system (Styker, USA), while MB UKA was implanted by suited microplastic instrumentation. All knee dynamic function and clinical functional scores were evaluated 6 to 12 months after UKA surgery.

The CT-based knee model reconstruction

All patients underwent a computed tomography (CT) scan (Sensation 64, Siemens, Germany) to acquire CT images of hip, knee, and ankle joints before surgery. Knee CT scans were performed with a slice thickness of 0.625 mm, whereas hip and ankle CT scans were done with a slice thickness of 1 mm. The CT scan was performed again for bilateral knees 6 to 12 months after UKA surgery. The three-dimensional (3D) surface models of preoperative and postoperative bones and components were eventually created using a watershed algorithm in Amira 6.7.0 (Thermo Fisher Scientific, Rockford IL, USA). The anatomical coordinate systems of the preoperative femur and tibia were created according to bony landmarks on the hip, knee, and ankle [35]. Based on the hypothesis that UKA didn't affect the anatomical morphology of non-surgical compartments, the coordination systems of the postoperative UKA and contralateral knees were determined by aligning the preoperative models using iterative closest points [36, 37]. The effect of modeling error caused by OA morphological deformation, CT metal artifact, and osteotomy should be reduced by the exclusion of meshes of the knee models in the medial compartment. The Root Means Square Error (RMSE) of the aligning method was 0.27 ± 0.06 mm for the femur and 0.32 ± 0.09 mm for tibia in UKA side, while RMSE was 0.41 ± 0.16 mm for the femur and 0.43 ± 0.10 mm for tibia in contralateral side. The submillimeter 3D surface models of femoral and tibial components were used to track knee kinematics; therefore, the 3D models of components were also aligned to CT-reconstructed models to determine the positions relative to femoral and tibial coordinate systems in UKA knees. The RMSE of the aligning method was 0.24 ± 0.08 mm for the femoral component and 0.31 ± 0.07 mm for the tibial component.

The posterior tibial slope (PTS) of the medial tibial plateau in the UKA knee was measured in the sagittal plane as previously described [38]. The tibial mechanical axis was defined as the line between the ankle center and tibial spine, i.e. Y-axis of the postoperative tibial coordinate system, and a transverse plane was set perpendicular to the mechanical axis of the tibia. The lower surface of the tibial component matched with the cut plane on the medial tibial plateau in ideal UKA surgery. Then, the PTS was defined as the angle between the tibial transverse and cut planes (Fig. 1). The posterior tibial slope in the medial tibial plateau was defined as a positive value, while the anterior tibial slope was set as a negative value.

The ACL and PCL footprints were determined based on anatomical features of the 3D surface model of the femur and tibia, and the in-vivo elongations of the cruciate ligament during lunge were quantified. The ACL was divided into anteromedial (AM) and posterolateral (PL) bundles, and the PCL was divided into anterolateral (AL) and posteromedial (PM) bundles according to their anatomy [39–42]. The femoral footprint of ACL and PCL was identified using the quadrant method based on the intercondylar features and Blumensaat line [39, 41]. The lateral and medial condyles were exposed by the femoral sagittal plane and the intercondylar roof line (i.e. Blumensaat line) as the High line (Fig. 2). The Low line parallel with the Blumensaat line and with maximum intercondylar notch height. The Deep-Shallow direction was defined as the lines perpendicular to Blumensaat line and across the femoral cortex. The quadrant area was built based on the High-Low and Deep-Shallow direction. The ACL AM and PL bundle femoral footprints were defined as 21%, 49% from High line and 25%, 33% from Deep line on the lateral condyle [39], and the PCL AL and PM bundle femoral footprints were defined as 16%, 35% from High line and 62%, 51% from Deep line on the medial condyle [41]. Then, the femoral footprints on the quadrant area were mapped on the intercondylar notch along the Z-axis of femoral coordinate system. Besides, the tibial footprint of ACL was determined using the anterior ridge, lateral groove, and intertubercular fossa of 3D tibia models [40]. Analogously, the tibial footprint of PCL was located on the posterior intercondylar fossa of the tibia [42]. The

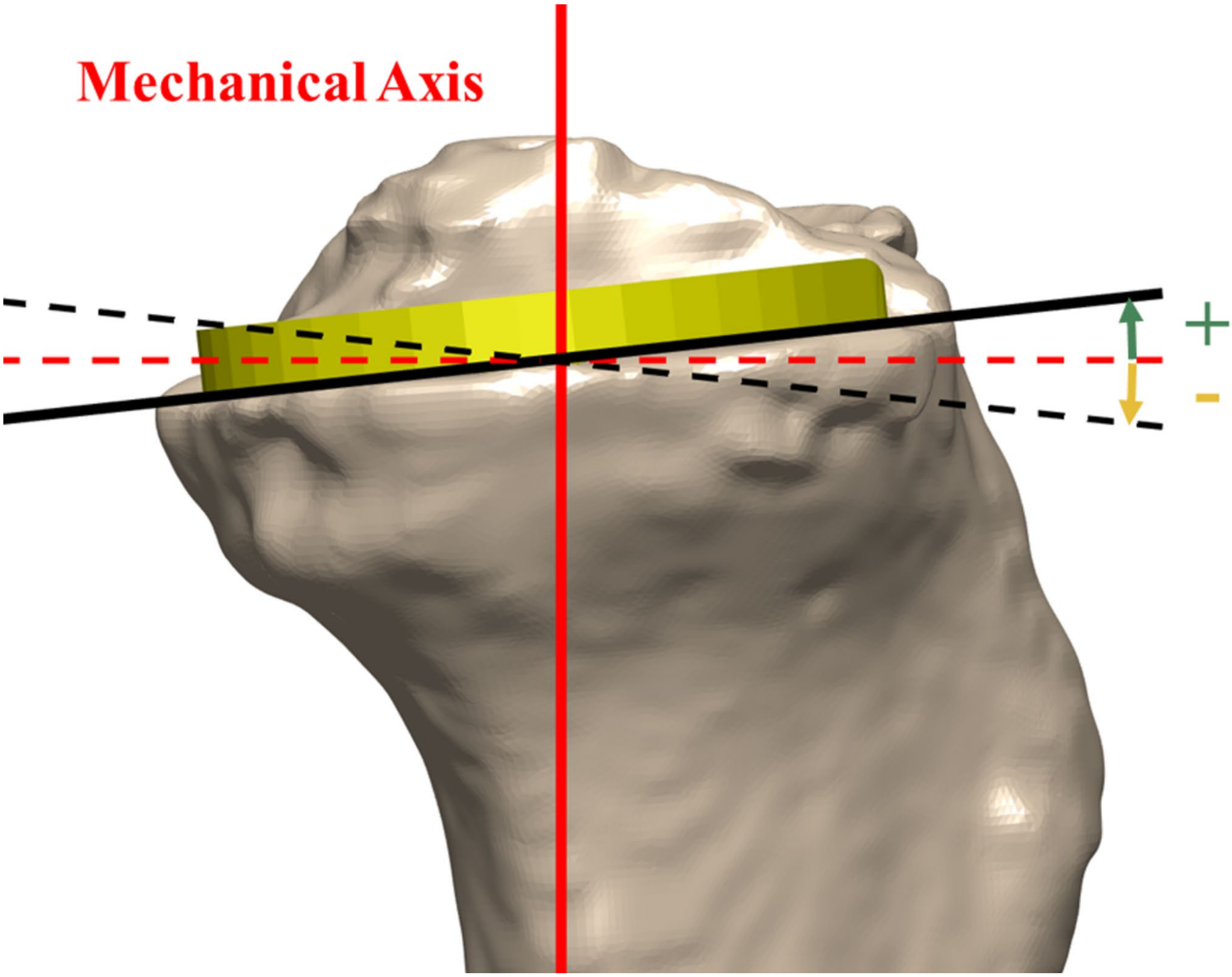

Fig. 1 The measurement of posterior tibial slope (PTS) of medial tibial plateau in sagittal view for UKA knees. The mechanical axis was defined by the ankle center and tibial spine, and the transverse plane was set perpendicular to the mechanical axis. The PTS was identified by the angle between transverse and cut planes in sagittal view. The posterior slope was positive value (green vector) while the anterior slope was negative value (yellow vector)

footprints of ACL and PCL bundles on preoperative femoral and tibial models were marked and mapped onto postoperative knee models on the UKA side to avoid the effect of CT metal artifact on ligament footprint identification, while the corresponding ligament footprints were directly determined on contralateral knee models. The start and end points of each bundle of ACL and PCL were considered as the center of corresponding ligament footprint areas.

In-vivo knee kinematics and cruciate ligament elongation measurement

Both UKA and contralateral knees were scanned under the dual fluoroscopic imaging system (DFIS) surveillance for all unilateral UKA patients. The DFIS consisted of two mobile fluoroscopes (BV Pulsera, Philips Medical, Netherlands), which were set in approximately orthogonal positions and kept continuous acquisition during motion. The radiation pulse was 30 snapshots per second and the pulse width was 8 milliseconds. All patients underwent DFIS surveillance in static standing posture and during single-leg lunge and the two-dimensional (2D) dynamic fluoroscopic images with size of 280×280 mm (1024×1024 pixels) was acquired. The ligament length in static standing posture was set as reference. The lunge was selected as a representative weight-bearing motion as it benefits postoperative bone and muscle strength with highly-demanding knee flexion. At the beginning of the lunge, the bilateral knee kept full extension; then, patients performed step back with the non-experimental leg and experimental knee squatted following the trend until maximum knee flexion angle. During lunge motion, most knee joint tissue should be captured under the configuration of DFIS to increase the accuracy of motion tracking. Every single lunge trial was completed in 5 s with 3–5-minute reset after motion. Any patient who

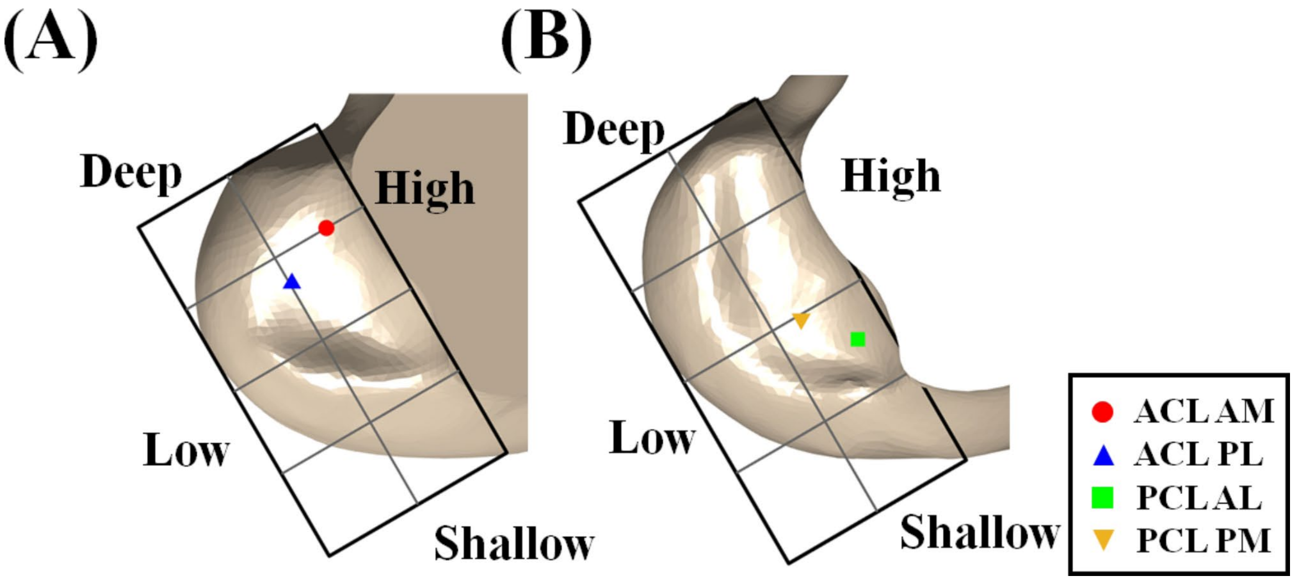

Fig. 2 Footprints of ACL and PCL on femoral surface models in knee lateral view. (**A**) ACL AM and PL bundles on femoral lateral condyle. (**B**) PCL AL and PM bundles on femoral medial condyle. The High line indicated the Blumensaat line. The Low line was parallel to the High line with maximum intercondylar notch height. The Deep and Shallow lines were perpendicular to the High line and crosses the borders of posterior and anterior cortexes

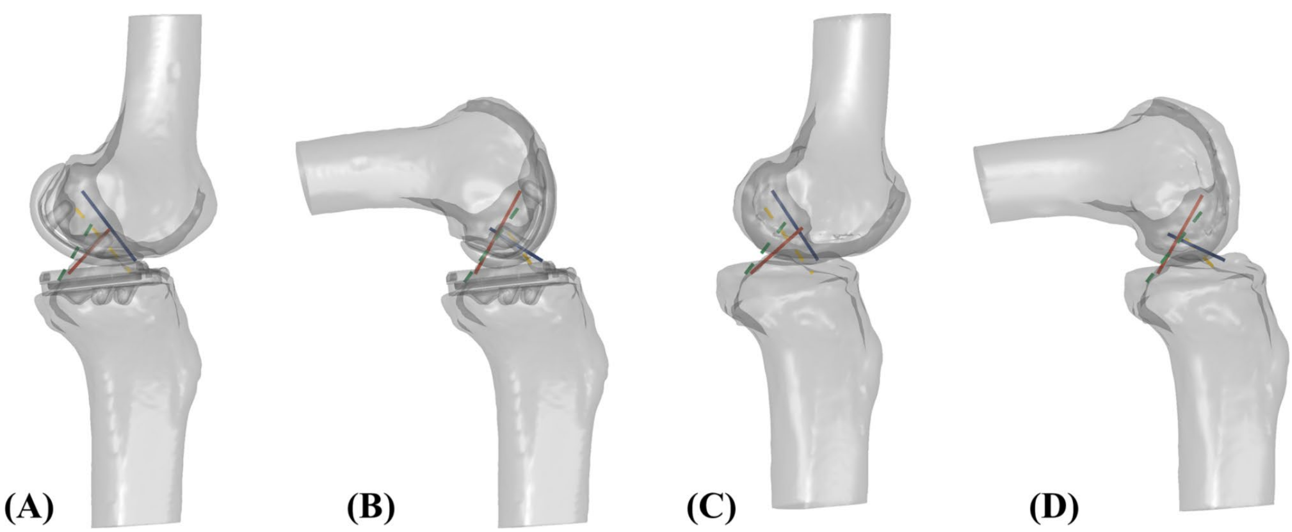

Fig. 3 The cruciate ligament direction of UKA knees in extension (**A**) and flexion (**B**) postures and contralateral knees in extension (**C**) and flexion (**D**) postures. The blue and yellow lines indicated AM and PL bundles of ACL, while the red and green lines indicated AL and PM bundles of PCL, respectively

cannot perform knee full extension or maximum flexion angle less than 100° has been excluded. The virtual DFIS system was built in MATLAB (MathWorks, MA, USA) to reconstruct real spatial positions during the experiment, and the accurate positions of bones and UKA components of bilateral knees by matching the projection of 3D surface models and the corresponding silhouette of 2D images. According to a previous study, the positional error of the 2D-3D matching procedure was < 0.2 mm in translation and < 0.4° in rotation [43, 44]. Then, the tibiofemoral 6 degrees-of-freedom (6-DOF) were calculated by femoral transepicondyle axis center translation relative to the tibia and tibial rotations relative to the femur.

The in-vivo ACL and PCL elongations during lunge were measured by the previously published algorithm [45, 46]. The dynamic spatial positions of 3D femoral, tibial, and component models and footprint points of ACL and PCL in both UKA and contralateral knees were determined at static standing and each flexion angle during lunge. An optimization procedure was applied to find the shortest 3D wrapping path of AM, PL bundles of ACL and AL, and PM bundles of PCL to avoid the direct line connecting femoral and tibial footprint points penetrating the surface models of bones and UKA components (Fig. 3). Then, each bundle length was quantified by the length of the 3D wrapping path.

Statistical analysis

The older adults perform different knee kinematics from young group during weight-bearing activities [47], and it's difficult to recruit the subjects with completely normal and healthy knees, therefore we set the contralateral native knees with intact structure and without OA symptom of unilateral UKA patients as self-control group. The ligament length during three trials were averaged for both UKA and contralateral knees for every patient to decrease the motion variation. The ligament relaxation can be precisely quantified based on current non-invasive motion capture method; therefore, we selected the length (l_0) of ACL and PCL in static standing posture as reference, and the ligament elongation with 1° increment during 0-100° flexion of lunge was evaluated by the following formula: $(l-l_0)/l_0 \times 100\%$. The paired Wilcoxon signed-rank test was performed to examine the difference in ligament elongation between UKA and contralateral knees at every knee flexion angle during lunge for FB UKA and MB UKA patients, respectively. Considering the knee flexion range of cruciate ligament to perform primary function, the average elongation differences of ACL and PCL bundles between UKA and contralateral knees were calculated during early-flexion (0–30°) and deep-flexion (60–100°), respectively. Then, Spearman rank correlation coefficient was calculated to determine the correlation between postoperative PTS and ligament elongation differences for FB UKA and MB UKA patients. The linear regression curve was fitted for the statistical correlation between ligament elongation difference and PTS. The level of significance was set as 0.05, and all statistical analyses were conducted using MATLAB (R2023b, MathWorks, Natick, USA). A post-hoc statistical power analysis based on the average, standard deviation of elongation difference between bilateral knees and sample size of each UKA group was performed to assess the effect size (G*Power 3.1.9.7, https://stats.idre.ucla.edu/other/gpower/), and the power was larger than 0.9.

Results

The patient characteristics data are shown in Table 1. The PTS difference between FB and MB UKA was due to respective surgical guideline. The average flexion angle in static standing pose was 2.3 ± 3.8° for implanted side and − 0.5 ± 3.9° for contralateral side in FB UKA group (p = 0.38), and 1.7 ± 4.0° for implanted side and 0.1 ± 4.3° for contralateral side in MB UKA group (p = 0.51).

Cruciate ligament elongation of FB-UKA

Both ACL AM and PL bundles of the FB UKA and contralateral native knees shortened in knee flexion from 0 to 100° during lunge, and the maximal bundle elongation was 1.3 ± 2.0% for AM, 3.7 ± 3.3% for PL of FB UKA knees and 2.3 ± 4.2% for AM, 3.4 ± 5.8% for PL of contralateral knees at knee extension, respectively (Fig. 4A-B). The elongation differences between FB UKA and contralateral knees at extension were − 1.1 ± 4.1% for the AM bundle (p = 0.64) and 0.3 ± 6.1% for the PL bundle (p = 0.89), which were not statistically significant.

Both PCL AL and PM bundles of the FB UKA and contralateral knees elongated during mid- and deep flexion from 30 to 100°. The maximal bundle elongation was 34.9 ± 7.3% for AL, 18.9 ± 9.1% for PM of FB UKA knees and 30.1 ± 11.4% for AL, 12.3 ± 9.8% for PM of contralateral knees at knee flexion, respectively (Fig. 4C-D). There were significant differences in PCL elongation between FB UKA and contralateral knees during knee flexion 6–86° for the AL bundle with an average of 6.2 ± 6.8% (p < 0.05) and 2-100° for the PM bundle with an average of 6.5 ± 8.6% (p < 0.05). The maximal PCL bundle elongation differences were 7.6 ± 8.7% for AL (p = 0.02) and 8.2 ± 5.1% for PM (p = 0.01).

Table 1 Patient demographics and postoperative knee function

	FB UKA [Range]	MB UKA [Range]	*p*-value
Sex	3 M, 10 F	4 M, 7 F	\
Age/years	64.7 [52 to 72]	65.1 [52.to 72]	0.77
Weight/kg	67.4 [51.2 to 80.4]	68.9 [51.8 to 97.2]	0.73
Height/cm	157.6 [148.6 to 169.5]	158.7 [145.3 to 178.2]	0.95
Body Mass Index/kg*m^{-2}	27.2 [22.6 to 31.7]	27.2 [22.6 to 31.8]	0.99
Follow-up Period/months	7.1 [6.0 to 10.5]	7.6 [6.0 to 10.9]	0.32
Posterior Tibial Slope/°	4.2 [0.3 to 7.2]	6.8 [3.7 to 10.7]	0.02*
Oxford Knee Score	16.6 [12 to 24]	15.4 [12 to 20]	0.41
Knee Society Score	90.5 [84.0 to 99.5]	92.5 [81.5 to 100]	0.37
Knee Score	91.8 [88 to 99]	93.1 [83 to 100]	0.49
Functional Score	89.2 [80 to 100]	91.8 [80 to 100]	0.45
Forgotten Joint Score	81.4 [52.1 to 100]	80.1 [29.2 to 100]	0.88

M: Males; F: Females

* indicated statistically significant difference (p < 0.05)

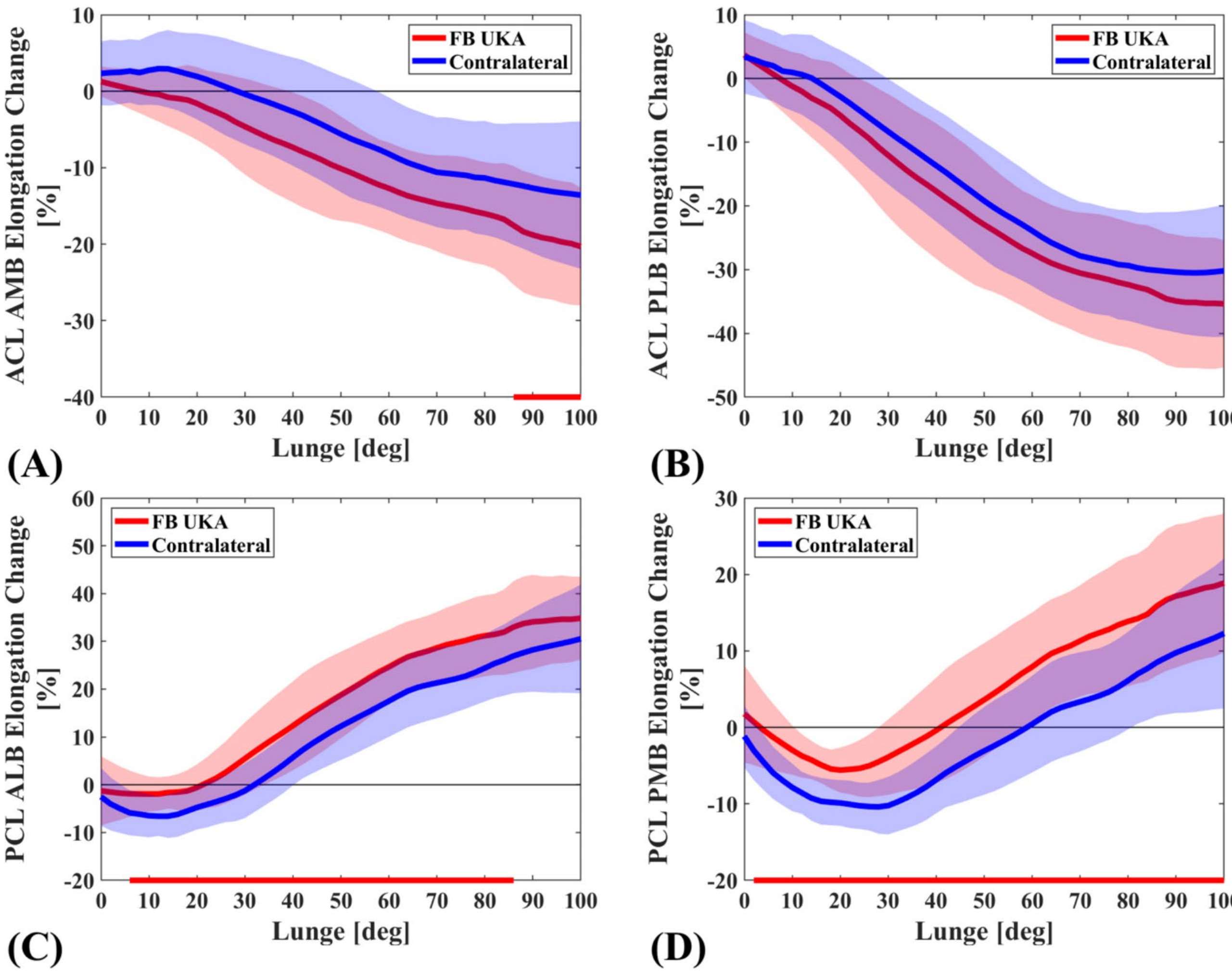

Fig. 4 Average and standard deviation of ligament elongation of the ACL AM (**A**), ACL PL (**B**), PCL AL (**C**) and PCL PM (**D**) in UKA and contralateral native knees for unilateral FB UKA patients. Significant differences between FB UKA and contralateral knees were marked with bold red line on x-axis of each graph

Cruciate ligament elongation of MB-UKA

The ACL AM bundle length of MB UKA knees remained nearly constant in early flexion (0–30°), and then shortened in mid- and deep flexion (30–100°). The ACL PL bundle of MB UKA knees and ACL double bundles of contralateral knees displayed a lax tendency in the full range of knee flexion during lunge (Fig. 5A-B). The maximal bundle elongation was 2.0 ± 2.4% for AM, 3.8 ± 3.2% for PL of MB UKA knees, and 2.3 ± 4.2% for AM, 3.0 ± 5.6% for PL of contralateral knees at knee extension, respectively. The length of ACL AM and PL bundles in MB UKA knees were significantly larger than contralateral sides during knee flexion of 34–72° and 38–54° ($p < 0.05$), and the maximal difference was 8.0 ± 8.1% at flexion 48° ($p = 0.01$) and 7.6 ± 9.8% at flexion 46° ($p = 0.02$).

Both PCL AL and PM bundles of the MB UKA and contralateral knees elongated during mid- and deep flexion from 30 to 100°, and the maximal bundle elongation was 35.3 ± 19.7% for AL, 15.7 ± 15.2% for PM of MB UKA knees and 37.2 ± 16.8% for AL, 13.2 ± 14.7% for PM of contralateral knees at knee flexion, respectively (Fig. 5C-D). There was no significant difference in PCL AL and PM bundle elongation between MB UKA and contralateral knees during knee flexion ($p > 0.24$).

Correlation between ligament elongation and posterior tibial slope

The average PTS in the medial UKA compartment was 4.2 ± 2.1° for FB UKA knees and 6.8 ± 2.2° for MB UKA knees, respectively. The PTS in FB groups was significantly less than that in MB UKA knees ($p = 0.02$, Table I). A significant positive correlation was found between the increased ligament elongation difference of ACL AM ($R = 0.63$, $p = 0.02$) and PL ($R = 0.75$, $p = 0.003$) bundles between UKA and contralateral sides in extension posture and PTS increase for FB UKA patients

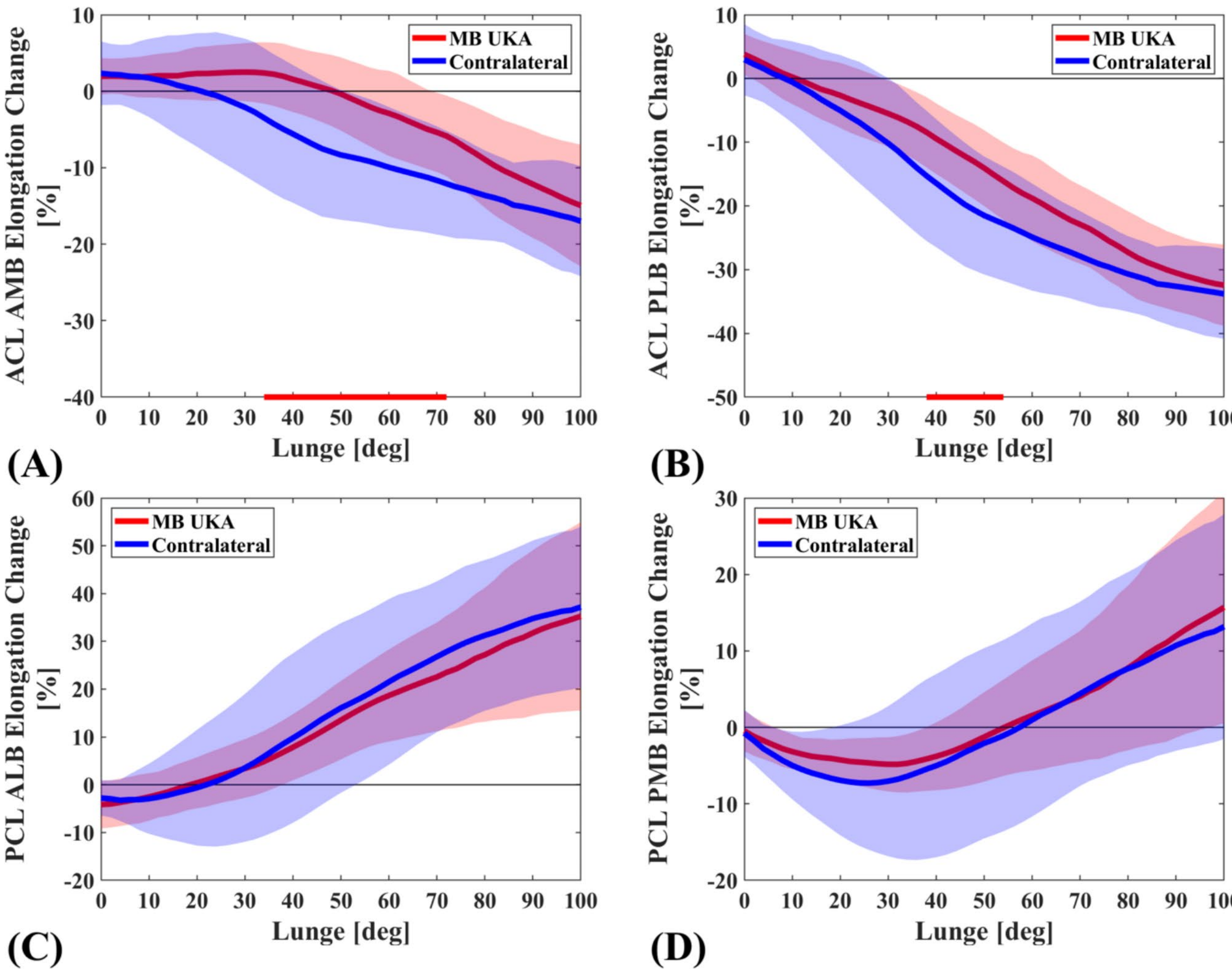

Fig. 5 Average and standard deviation of ligament elongation of the ACL AM (**A**), ACL PL (**B**), PCL AL (**C**) and PCL PM (**D**) in UKA and contralateral native knees for unilateral MB UKA patients. Significant differences between MB UKA and contralateral knees were marked with bold red line on x-axis of each graph

(Fig. 6A-B). Similarly, there was significant positive relevance between the ACL AM ($R = 0.77$, $p = 0.005$) and PL ($R = 0.73$, $p = 0.01$) elongation difference and PTS for MB UKA patients (Fig. 6C-D). In knee deep flexion, no correlation was found between PCL elongation difference and PTS in either FB or MB UKA knees ($p > 0.5$).

Discussion

The current study quantified the in-vivo ACL and PCL elongation on femoral and tibial 3D surface models by DFIS during lunge motion for unilateral FB or MB UKA patients. We found that the PCL double bundles of FB UKA knees elongated more than contralateral sides in most flexion ranges of the lunge, while ACL double bundles were less affected. In contrast, ACL double-bundle elongations of MB UKA in mid-flexion during lunge were larger than contralateral sides, while there was no significant difference in PCL elongation. Besides, the increased PTS in the medial implanted compartment was significantly correlated to the increased ACL double-bundle elongation of both FB and MB UKA knees compared with contralateral knees, while there was no correlation in personal PCL elongation pattern and medial PTS.

The elongation patterns of PCL after FB UKA and ACL after MB UKA during weight-bearing single-leg lunge were different from contralateral native knees for medial UKA patients. The ACL contributes to knee anteroposterior and rotational stabilities when knee flexes between 0–30°, the PCL performs maximal tension beyond 60° of flexion, while the function of ACL and PCL diminishes during knee flexion 30–60° [14–16]. Several biomechanical studies declared that an abnormal ACL or PCL may reduce tibiofemoral joint stability and lead to long-term OA progression [19, 23, 24]. To best of our knowledge, it's the first study to quantify the in-vivo ACL and PCL length of UKA patients during weight-bearing activities

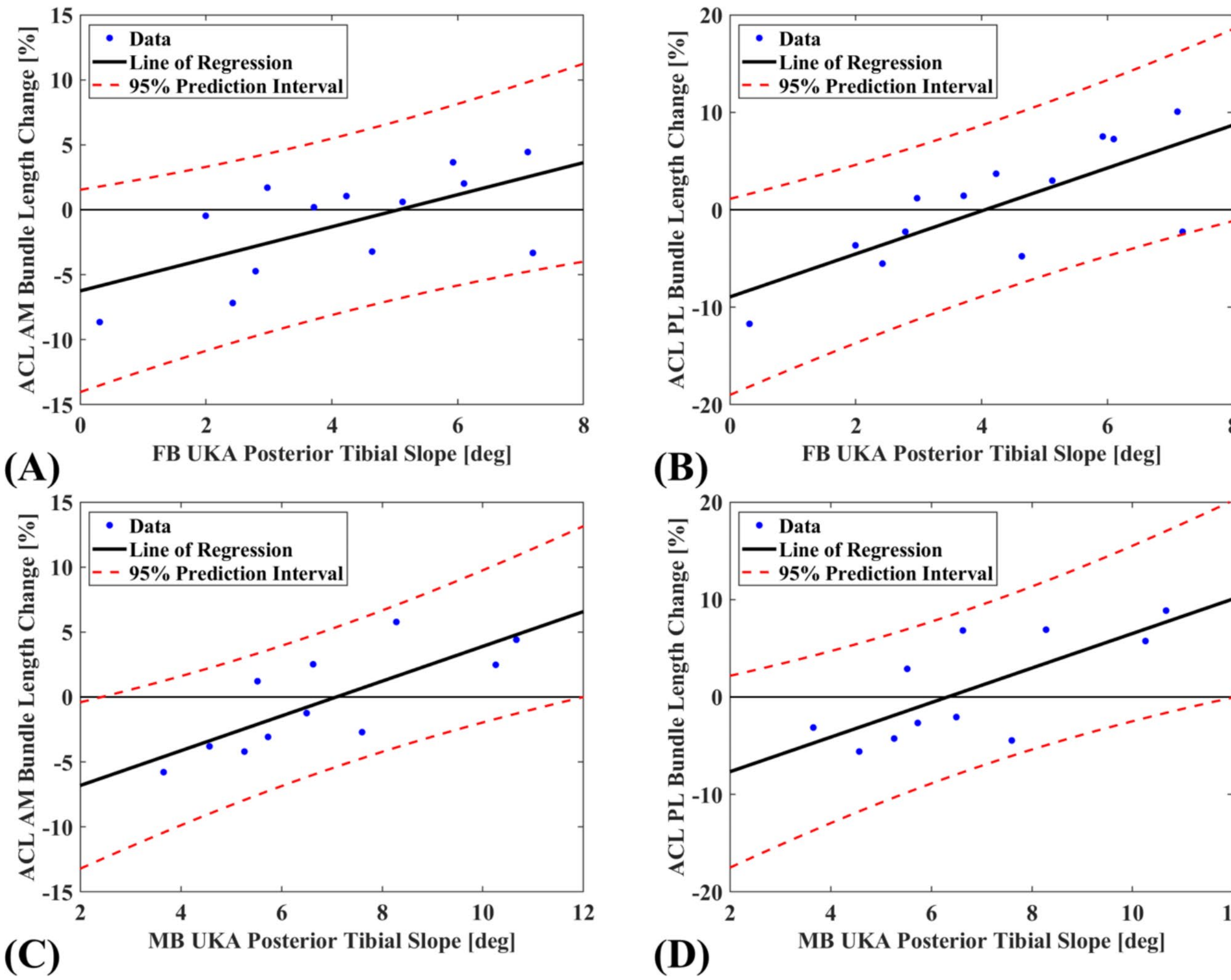

Fig. 6 Significant positive correlation between the ligament elongation increase of ACL double bundles in extension posture and posterior tibial slope in medial UKA compartment for both FB and MB UKA patients. (**A**) $R=0.63$, $p=0.02$ for ACL AM bundle in FB UKA patients. (**B**) $R=0.75$, $p=0.003$ for ACL PL bundle in FB UKA patients. (**C**) $R=0.77$, $p=0.005$ for ACL AM bundle in MB UKA patients. (**D**) $R=0.73$, $p=0.01$ for ACL PL bundles in MB UKA patients

and the effect of implant design on ligament elongation patterns. We found the elongation of PCL AL and PM bundles of FB UKA knees was $6.2 \pm 6.8\%$ ($p<0.05$) and $6.5 \pm 8.6\%$ ($p<0.05$) larger than contralateral sides during most flexion range of lunge (Fig. 4C-D). The overstretching PCL during lunge may reduce the normal femoral rollback in the same patient group after FB UKA [48]. Besides, the ACL AM bundle after MB UKA remained nearly consistent in early flexion. However, the ACL AM and PL bundle elongation were greater than contralateral knees with the maximal difference of $8.0 \pm 8.1\%$ ($p=0.01$) and $7.6 \pm 9.8\%$ ($p=0.02$) during mid-flexion of lunge, respectively (Fig. 5A-B). The elongated ACL in the mid-flexion range after MB UKA restricted the normal ACL function, which may disturb healthy knee kinematics, causing over-constraints at knee extension and finally lead to subjective instability feeling [16]. It should be closely tracked whether the altered elongation pattern of PCL after FB UKA and ACL after MB UKA can result in long-term ligament deficiency, postoperative OA progression, and revision surgery.

The in-vivo ACL overstretching during early-flexion of lunge is associated with larger medial PTS for both FB and MB UKA knees relative to contralateral sides in weight-bearing extension posture. In the current popular theory, the large PTS increases the knee anteroposterior translation and sheer stress in extension pose and finally increase ACL injury risk [28, 49]. In contrast, the decreased PTS may contribute to limited posterior femoral rollback, enlarged contact stress and PCL injury risk [27, 50]. However, no previous study has clearly varied this relation in UKA patients. To our knowledge, it is the first study to analyze the quantitative relationship between PTS and the in-vivo cruciate ligament elongation pattern of UKA. We found ACL AM and PL bundles tended to overstretch in weight-bearing extension for both FB and MB UKA knees with increased medial PTS (Fig. 6). However, there was no significant relationship between postoperative PTS and PCL elongation difference between implanted and contralateral knees in flexion posture. It may attribute to the absent data during deeper knee flexion range of lunge (> 100°), where the PCL function was further challenged [14, 15]. Furthermore, the optimal PTS range is one of critical factors to determine the cut plane location during UKA surgery. Hernigou and Deschamps conducted a long-term follow-up cohort and found a steep PTS was associated with ACL rupture in UKA knees with preoperative intact ACL, suggesting that PTS less than 7° should be appropriate in tibial cut during UKA to protect ACL function [29]. Sekiguchi et al. conducted an in-silico study on FB UKA, and the result showed that the ACL tension increased, and PCL tension slightly decreased during gait and deep knee bend with larger PTS in UKA model [25]. Another in-silico study revealed that increased PTS of MB UKA was associated with increased non-surgical compartmental contact stress and ACL stress during gait, which may result in OA progression and ligament failure [26]. The regression result (Fig. 6) supported the effect of PTS on ACL elongation for MB UKA patients, however PTS ranging from 5 to 7° may elongate the ACL during early-flexion and result in ACL laxity in long-term follow-up period. Considering current findings and the effect of PTS on PCL comprehensively [23, 27, 28], the PTS around 5° for FB UKA and 7° for MB UKA may be more appropriate to restore normal cruciate ligament function during weight-bearing lunge.

There has not yet been a consensus among surgeons and researchers regarding which UKA design benefits medial OA patients better [5–13, 51]. The different UKA designs had different effects on cruciate ligament function during weight-bearing lunge. The current study revealed that the almost full-range PCL elongation increased in FB UKA knees compared with contralateral sides (Fig. 4), i.e. the increased demanding on PCL function, which may indicate higher PCL injury risk after FB UKA. In contrast, the ACL of MB UKA knees abnormally kept elongated in mid-flexion during lunge (Fig. 5), implying the preoperative medial compartmental knee OA combined with ACL deficiency was not appropriate for MB UKA procedure, which conformed to current clinical concept [17–19]. Furthermore, FB UKA (MCK partial knee system; Stryker, USA) and MB UKA (Oxford Phase 3; Zimmer Biomet, USA) have different surgical procedures and require different PTS range in surgical guidance. The FB UKA was implanted by MAKO robotic system and the target PTS range was 3–7°, while the MB UKA was implanted by suited microplastic instrumentation and the targeted PTS was 2–12°. Our findings supported current surgical guidance, i.e. the optimal PTS around 5° for FB UKA and 7° for MB UKA. On the other hand, the average PTS of UKA knees satisfied the guidance required PTS range (Table 1), however only 8 of 13 FB UKA had PTS ranging 3–7° and all MB UKA had PTS ranging 3.7–10.7°. The large individual difference emphasized the PTS should be determined more strictly in tibial cut procedure during UKA. Therefore, the current study suggests the effect of UKA design and PTS on in-vivo ACL, PCL elongation can be well considered in preoperative planning.

The several limitations in the current study should be declared. First, we defined the footprint areas of ACL and PCL by quadrant method and bony landmark on 3D surface bone models from CT according to previously published anatomical descriptions, and the true ligament footprint was not identified due to the absent MRI data. However, the insufficient resolution in conventional 3T MRI device and decreased contrast by metallic implants will greatly affect the ligament identification in fossa intercondylar femoris [52]. Besides, previous sensitivity analysis demonstrated that cruciate ligament footprint defined on surface models by CT can provide accurate measurements in in-vivo functional activities [15, 53]. Secondly, the cruciate ligament elongation pattern was merely quantified during a single-leg lunge. More weight-bearing and non-weight-bearing daily activities should be investigated in further study. Third, only one type of implant was studied for each UKA design, however, Stryker MCK knee system was one representative of FB UKA with a non-congruent bearing surface and Oxford III was one of the most commonly used MB UKA. Lastly, the follow-up period of our study limited to 6–12 months, which was relatively short. Long-term follow-up duration would be required to determine the effect of abnormal cruciate ligament elongation patterns on further postoperative complications, survival rate, and knee functional performance.

In conclusion, this is the first study that illustrates quantitative in-vivo ACL and PCL elongation patterns during single-leg lunge for patients who underwent unilateral medial FB and MB UKA procedures using dual fluoroscopy. Our results indicated increased ligament tension of PCL double bundles of FB UKA knees in nearly full-range flexion of lunge and ACL double bundles of MB UKA knees in mid-flexion compared with corresponding contralateral native knees. There was a significantly positive correlation between medial PTS and ACL elongation difference between implanted and contralateral knees for both FB and MB UKA, indicating excessive PTS should be avoided to preserve native ACL function in further UKA implantation. Furthermore, current PTS target around 5° for FB UKA and 7° for MB UKA should be appropriate in tibial cut during implantation procedure.

Author contributions
N.Z. and T.Y.T designed the experiment. N.Z., C.X., H.D. and D.Z. completed all data acquisition and analysis. Q. W. completed all UKA surgeries as surgeon. N.Z. and D. Z. wrote the main manuscript text and N.Z. prepared all figures and tables. Y.L., Z.H. and Q.W. contributed to the date interpretation and manuscript revision. All authors reviewed the manuscript and approve of submission.

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Data availability
All data generated and analyzed during this study are included in this published article.

Declarations

Consent for publication
Not applicable.

Competing interests
Author T.Y. Tsai got research support from TAOiMAGE Co., Ltd., China and MicroPort Co. Ltd., China. Other authors declared that they had no conflicts of interest.

Ethical approval
The study protocol was in compliance with the Helsinki Declaration and approved by the Institutional Ethics Committee of Shanghai Sixth People's Hospital Affiliated to Shanghai Jiao Tong University School of Medicine (No.2017-084). All patient wrote the informed consent under personal consciousness before experimental processes.

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