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Kinematically aligned total knee arthroplasty limits high tibial forces, differences in tibial forces between compartments, and abnormal tibial contact kinematics during passive flexion

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Abstract

Purpose Following total knee arthroplasty (TKA), high tibial forces, large differences in tibial forces between the medial and lateral compartments, and anterior translation of the contact locations of the femoral component on the tibial component during passive flexion indicate abnormal knee function. Because the goal of kinematically aligned TKA is to restore native knee function without soft tissue release, the objectives were to determine how well kinematically aligned TKA limits high tibial forces, differences in tibial forces between compartments, and anterior translation of the contact locations of the femoral component on the tibial component during passive flexion.

Methods Using cruciate retaining components, kinematically aligned TKA was performed on thirteen human cadaveric knee specimens with use of manual instruments without soft tissue release. The tibial forces and tibial contact locations were measured in both the medial and lateral compartments from 0° to 120° of passive flexion using a custom tibial force sensor.

Results The average total tibial force (i.e. sum of medial + lateral) ranged from 5 to 116 N. The only significant average differences in tibial force between compartments occurred at 0° of flexion (29 N, $p = 0.0008$). The contact locations in both compartments translated posteriorly in all thirteen kinematically aligned TKAs by an average of 14 mm ($p < 0.0001$) and 18 mm ($p < 0.0001$) in the medial and lateral compartments, respectively, from 0° to 120° of flexion.

Conclusions After kinematically aligned TKA, average total tibial forces due to the soft tissue restraints were limited to 116 N, average differences in tibial forces between compartments were limited to 29 N, and a net posterior translation of the tibial contact locations was observed in all kinematically aligned TKAs during passive flexion from 0° to 120°, which are similar to what has been measured previously in native knees. While confirmation in vivo is warranted, these findings give surgeons who perform kinematically aligned TKA confidence that the alignment method and surgical technique limit high tibial forces, differences in tibial forces between compartments, and anterior translation of the tibial contact locations during passive flexion.

Keywords Tibiofemoral joint · Tibial force sensor · Intra-articular load · Anterior sliding · Kinematic alignment · Contact force imbalance · Soft tissue balancing

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Introduction

The goal of kinematically aligned (KA) TKA is to closely restore knee function to native. To achieve this goal, the femoral and tibial components are aligned to restore both the native joint lines (i.e. distal and posterior femoral and proximal tibial joint lines) and the native alignments of

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the limb and knee [26] (Fig. 1). Consequently, soft tissue releases are largely avoided [13, 24, 26, 28]. During passive flexion, knee function is determined by the interaction between the articular surfaces and the soft tissue restraints [10, 57, 58]. Thus, by striving to both restore the native joint lines and avoid soft tissue releases, KA TKA should closely restore to native three important metrics of knee function: the tibial forces, differences in tibial forces between the medial and lateral compartments, and native tibial contact kinematics (i.e. anterior–posterior translation of the contact locations of the femur on the tibia during flexion).

The total tibial forces, differences in tibial forces, and tibial contact kinematics have been measured previously in the native knee [31, 40, 52]. In the native knee, the tibial forces during passive flexion are low and are greatest in extension and decrease with knee flexion [52]. This pattern agrees with the pattern that can be inferred from the distraction laxity throughout flexion [46, 49] and tension in the ligaments that are highest in extension where a majority of the soft tissue restraints are most tight and lower in flexion where most of the soft tissue restraints become less tight [1, 2, 9, 17, 23]. Although there is variability from knee to knee, the differences between tibial forces in the medial and lateral compartments are small (i.e. balanced), and the medial force is generally greater than the lateral force [52]. This pattern agrees with the pattern that can be inferred from the laxities of the native knee where there is varus, valgus, and distraction laxity throughout the full arc of flexion [46, 49] indicating that neither compartment is overly tight compared to the other. Furthermore, the varus laxity is greater than the valgus laxity [46, 49], indicating that the medial compartment is tighter than the lateral compartment. In the native knee, the tibial contact locations in both the medial and lateral compartments translate posteriorly, lateral more than medial, during passive flexion [31, 40].

Although high patient satisfaction and high function have been reported at 6 months to 6 years after KA TKA [13, 14, 27], no study has determined whether KA TKA limits high tibial forces, differences in tibial forces between compartments, and anterior translation of the contact locations during passive flexion. Accordingly, the purposes of this cadaveric study were to characterize the total tibial forces (i.e. sum of the medial and lateral tibial forces), the differences in tibial forces between compartments, and the anterior–posterior translation of the tibial contact locations during passive flexion after KA TKA. Because KA TKA should restore the native joint lines while avoiding soft tissue releases, the hypotheses were that KA TKA would restore native knee function by limiting high tibial forces, differences in tibial forces between compartments, and anterior translation of the contact locations during passive flexion. If these hypotheses are supported, then these results would both provide an explanation for high patient satisfaction and high function after KA TKA [13, 14, 27] and motivate future clinical studies of tibial forces and contact kinematics after KA TKA.

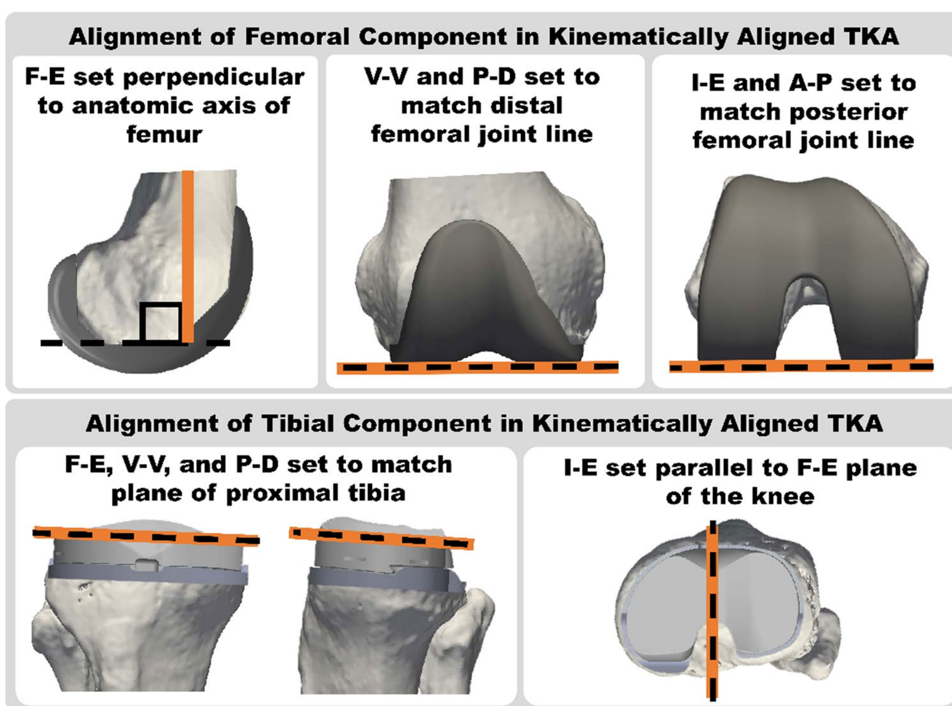

Fig. 1 Composite explains the desired alignment of both the femoral component (*top row*) in flexion–extension (F–E), varus–valgus (V–V), internal–external rotation (I–E), proximal–distal (P–D), and anterior–posterior (A–P) and the tibial component (*bottom row*) in F–E, V–V, I–E, and P–D after kinematically aligned TKA. Each *solid orange line* represents the alignment target, and each *dashed black line* represents the feature of the component being aligned to the alignment target. Not shown are the medial–lateral position of the femoral component, which is set visually by the surgeon as to centre the component on the femur and the medial–lateral and A–P positions of the tibial component, which are set to minimize overhang (color figure online)

Materials and methods

Fifty fresh-frozen human cadaveric knees were considered for inclusion. Each knee, which was procured through either the Donated Bodies Program at the University of California, Davis or Science Care Inc., was screened using an anteroposterior radiograph of the knee and a visual inspection of the articular surfaces. Thirty-seven specimens were excluded because of evidence of degenerative joint disease (i.e. marginal osteophytes, joint space narrowing, chondrocalcinosis, subchondral sclerosis, and/or cartilage lesions) and/or evidence of previous surgery to the knee. Thus, thirteen fresh-frozen human cadaveric knees were included (median age = 78 years, range 58–93 years).

As described in the following paragraphs, the in vitro set-up was designed so that the results of this cadaveric study would translate to the clinical realm. Briefly, each knee was carefully dissected to minimize disruption of the passive restraints of the knee. A six degree-of-freedom load application system [5] was used to flex and extend the knee so that non-physiologic constraints were not applied to the joint. This system is representative of a clinician flexing and extending the knee while supporting the weight of the femur and tibia and loosely holding the foot as not to constrain internal–external rotation of the tibia. KA TKA was performed using manual surgical instruments and followed the same intraoperative checks as those performed clinically [26, 28, 38]. The tibial force sensor matched the size and shape of the standard tibial component so as not to alter the articulation of the tibiofemoral joint [48].

Each knee was prepared for testing using the following dissection procedure. First, the fibula was rigidly fixed to the tibia using a transverse screw 12 cm distal to the joint line to retain the rigidity of both the tibiofibular joint and the insertions of the lateral collateral ligament and biceps femoris tendon. Second, the thigh was transected 20 cm proximal to the joint line, and the shank was transected 25 cm distal to the joint line. Third, all soft tissues more than 15 cm proximal and 12 cm distal to the joint line were removed. Fourth, the fibula was transected just distal to the transverse screw fixing it to the tibia. Fifth, all skin and subcutaneous adipose tissue were removed. Sixth, the tendons of insertion of the biceps femoris, semimembranosus, semitendinosus, and quadriceps were isolated, and the semimembranosus and semitendinosus tendons were sutured together. Seventh, cloth loops were sutured to the three tendons. Finally, intramedullary rods were cemented into the medullary canals of both the femur and tibia, and each knee was wrapped in saline-soaked gauze to prevent dehydration of the tissues.

Following dissection, each knee was aligned in a six degree-of-freedom load application system (Fig. 2) [5] using alignment fixtures so that the flexion–extension (F–E) and internal–external rotation (I–E) axes of the load application system were coincident with the F–E and longitudinal rotation axes of the tibiofemoral joint (Fig. 2). These alignment fixtures connected the intramedullary rods to the load application system, but allowed for six degree-of-freedom adjustments of both the femur and tibia relative to the load application system. Proper alignment of the knee with respect to the load application system was achieved when the coupled motions to both F–E rotation (i.e. anterior–posterior (A–P) and proximal–distal translations and varus–valgus (V–V) rotation) and I–E rotation (i.e. A–P and medial–lateral translations and V–V rotation) were minimized. Once proper alignment was achieved, the femur and tibia were potted within square aluminium tubes using methyl methacrylate to fix the position and orientation of each bone relative to the load application system during testing.

Following alignment, each knee was subjected to a preconditioning protocol consisting of cycling the knee five times between ±2.5 Nm in F–E. After completing the preconditioning, the knee was extended under 2.5 Nm to define 0° of flexion (i.e. full extension) [34]. After full extension was defined, the knee was removed from the load application system, wrapped in fresh saline-soaked gauze, placed in a sealed plastic bag, and stored in a refrigerator at 4° C overnight.

The following day, KA TKA was performed on the knee by a surgeon with expertise in the technique using manual instruments [26, 28, 38] (Fig. 3). The knee was exposed through a mid-sagittal osteotomy of the patella (i.e. the transpatellar approach [36]). Correct alignment of the femoral component (Zimmer Persona CR) in V–V, proximal–distal (P–D), I–E, and A–P was confirmed based on the following quality assurance check [26]. When the thickness of each of the four femoral bone resections (distal medial, distal lateral, posterior medial, and posterior lateral) measured using callipers was within 0.5 mm of the thickness of the corresponding region of the femoral component after correcting for the kerf of the saw blade [26], the femoral component was kinematically aligned in V–V, P–D, I–E, and A–P. Because cartilage wear was not present in these specimens, loss of cartilage thickness present in osteoarthritic patients did not have to be accounted for during these in vitro TKAs. Correct alignment of the tibial component (Zimmer Persona CR) in V–V, P-D, and F–E was also confirmed based on the following quality assurance checks [26]. When the knee was stable (i.e. had negligible V–V laxity indicated by <0.5 mm of gapping medially and laterally, which matches that of the native knee under applied V–V torques [46, 49]) at 0° of flexion and had the same A–P offset of the distal medial condyle of the femur from the anterior cortex of the tibia at 90° of flexion as that before distal femoral resections were made [26], the tibial component was kinematically aligned in V–V, P–D, and F–E. No soft tissues were released. Prior to cementing the components in place, the distal surfaces

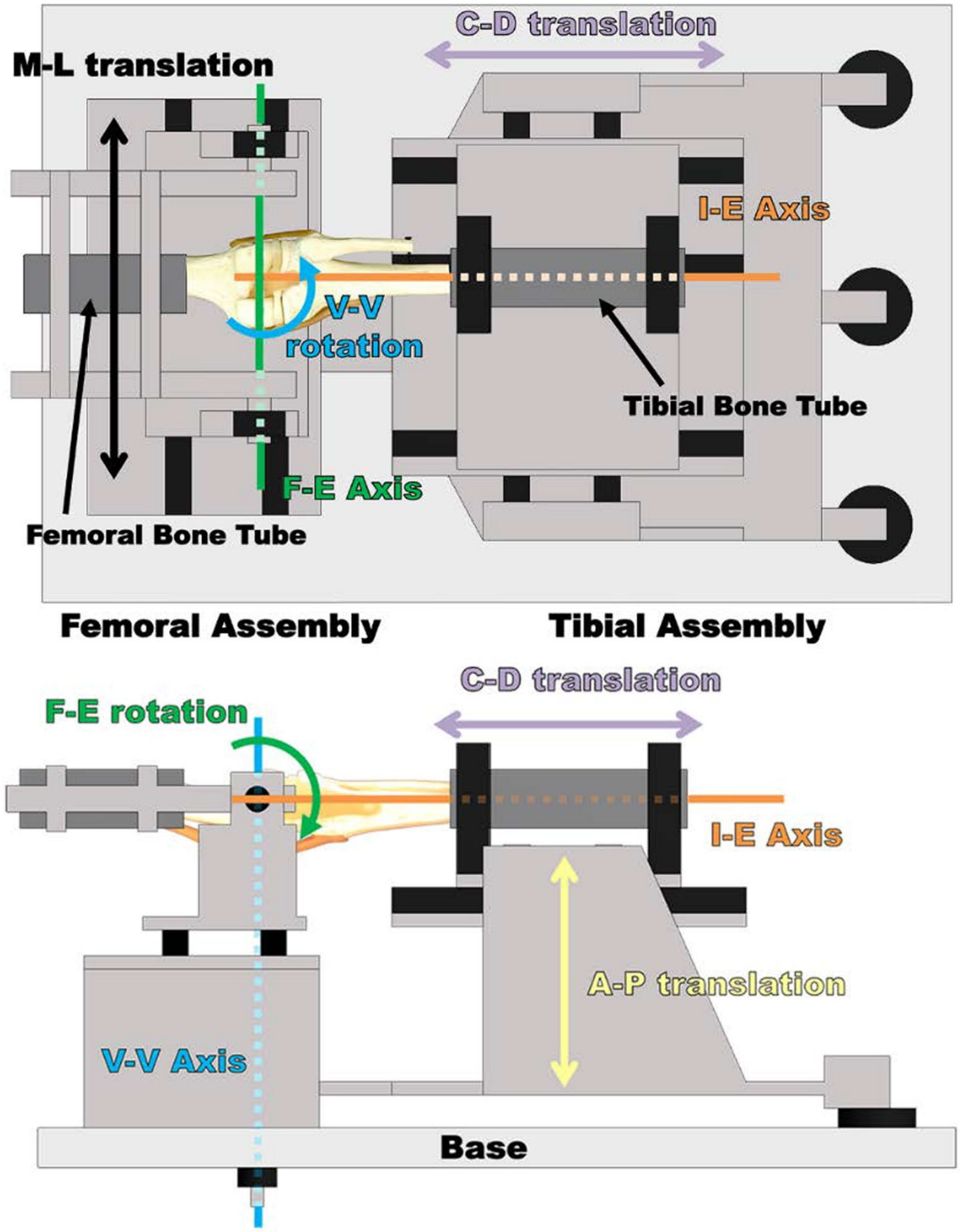

Fig. 2 Schematic of the six degree-of-freedom load application system [5] used to flex and extend the knee. A knee specimen is included to show its orientation in the load application system (i.e. patella towards the base). The degrees of freedom follow the coordinate system of Grood and Suntay [18] so that the flexion–extension axis is fixed to the femoral assembly, and the longitudinal rotation axis is fixed to the tibial assembly. Accordingly, the femoral assembly provides two degrees of freedom, flexion–extension (F–E) rotation and medial–lateral (M–L) translation. The tibial assembly provides the remaining four degrees of freedom including internal–external (I–E) and varus–valgus (V–V) rotations and anterior–posterior (A–P) and compression–distraction (C–D) translations. Knee specimens are aligned so that the F–E and longitudinal rotation axes of the tibiofemoral joint are coincident with the F–E and I–E axes of the load application system, respectively. Stepper motor actuators (omitted for clarity) are used to apply loads in all degrees of freedom except M–L translation. Unconstrained motions in all degrees of freedom are enabled through the use of low-friction bearings

of a standard tibial baseplate were coated with a release agent (petroleum jelly), which allowed for removal of the standard tibial component after the cement had cured. After the cement had cured, the standard tibial component was replaced by a custom tibial force sensor (Fig. 4) [48] with the conversion tray and insert size and thickness to match the size and thickness of the standard tibial baseplate and insert selected by the surgeon. The exposure was closed using two transverse screws in the patella.

The KA TKA was remounted in the load application system for testing. Small loads were applied to the tendons of the biceps femoris (15 N), the semimembranosus/semitendinosus (26 N), and the quadriceps (80 N) along their lines of action [3] using constant force springs. Both the quadriceps and medial and lateral hamstrings were loaded to maintain the inherent stability to the joint [29, 32]. The relative magnitudes of these loads were proportional to the average physiologic cross-sectional area of each muscle, and each was about 3% of the maximum isometric force of that muscle group assuming a specific tension of 30 N/cm^2 [16, 53]. The KA TKA was preconditioned by cycling five times in F–E between 0° and 120° of flexion using the full extension reference determined prior to TKA. After preconditioning, the knee was flexed from 0° to 120°. At 0°, 10°, 30°, 45°, 60°, 90°, and 120° of flexion, the tibial force and tibial contact location in each compartment were determined using the voltage outputs of the tibial force sensor in the coordinate system of each compartment (Fig. 5) [48]. To

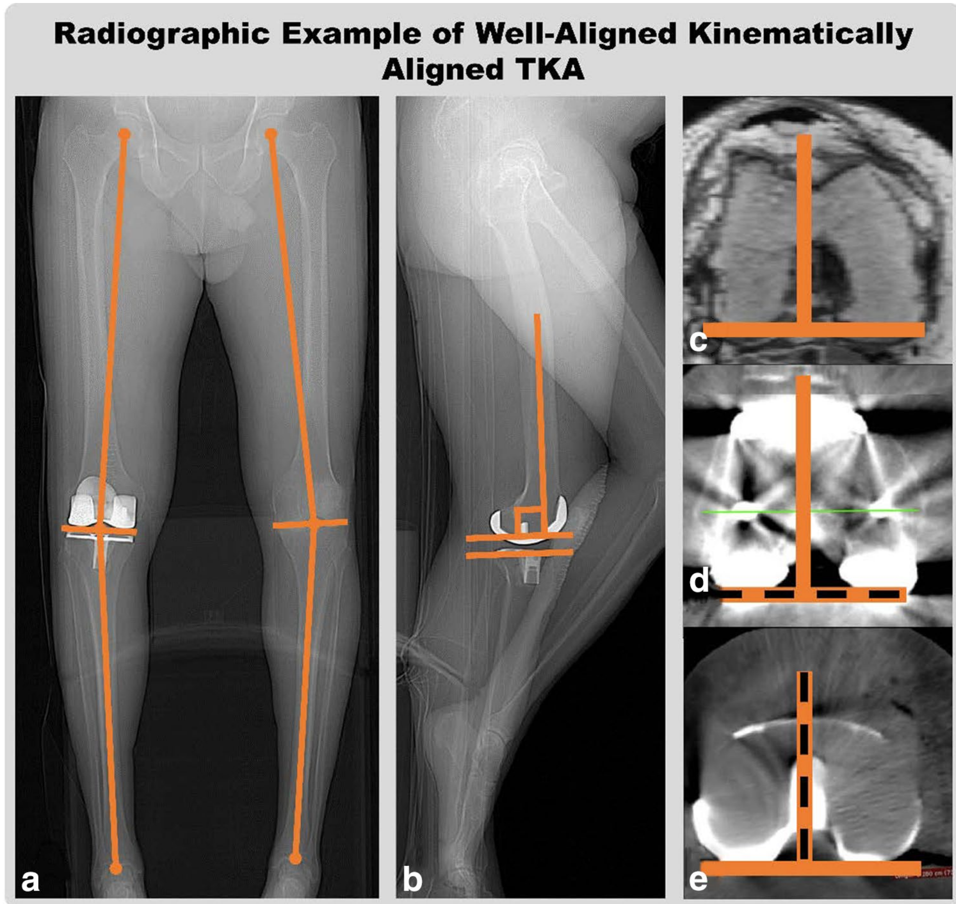

Fig. 3 Composite shows a representative set of post-operative images of a well-aligned kinematically aligned TKA. The coronal scanogram (**a**) shows that both the limb and joint line alignments of the kinematically aligned TKA closely match those of the native contralateral limb. The sagittal scanogram (**b**) shows that the femoral component is aligned perpendicular to the femoral anatomic axis and that the tibial component is aligned with a normal posterior slope. The pre-operative magnetic resonance (MR) image (**c**) shows the posterior condylar axis (*solid orange line*), which is closely perpendicular to the flexion–extension plane of the knee. The post-operative computed tomography (CT) images of the femoral component (**d**) and tibial component (**e**) show that the components (*dashed black lines*) are aligned parallel to the posterior condylar axis (*solid orange line*) and the flexion–extension plane of the knee (*solid orange line*), respectively (color figure online)

account for errors introduced by friction, the tibial forces and tibial contact locations at a particular flexion angle were the average of those computed during flexion from 0° to 120° and those computed during extension from 120° back to 0°. To account for errors introduced by the curvature of the tibial articular surface, the tibial contact locations were corrected using error correction functions developed previously to minimize the errors in computed tibial contact location caused by the curved articular surfaces [47]. The average total tibial force caused by the tension in the soft tissues was computed as the difference between the average measured total tibial force (i.e. sum of medial + lateral) and the average contribution of the applied muscle loads to the total tibial force (see "Appendix 2: Anterior–posterior and compression–distraction components of applied muscle loads" section). The differences in tibial forces between compartments were computed as the difference between the medial and lateral tibial forces. Thus, a positive difference indicated that the medial tibial force was greater than the lateral tibial force.

Following University of California policies, this study did not require institutional review board (IRB) approval because de-identified cadaveric specimens were used.

Statistical analysis

To determine how well KA TKA limited differences in tibial forces between compartments, a two-factor repeated measures analysis of variance (ANOVA) including interaction was performed (JMP version 11.2.0; SAS Institute Inc., Cary, NC; www.jmp.com). The first factor was compartment of the knee at two levels (medial and lateral), and the second factor was flexion angle at seven levels (0°, 10°, 30°, 45°, 60°, 90°, and 120°). When an important interaction was observed in the ANOVA, post hoc pairwise comparisons between the treatment means ($\bar{Y}_{ij.}$, averaged over the 13 knees) of the medial ($i = 1$) and lateral ($i = 2$) compartments were made at each flexion angle ($j = 1, 2, 3, 4, 5, 6, 7$ for 0°, 10°, 30°, 45°, 60°, 90°, and 120° of flexion, respectively) using the Bonferroni method [39]. For the ANOVAs, the level of significance (α) was set at 0.05. For the post hoc pairwise comparisons using the Bonferroni method, the level of significance ($\alpha_{\text{Bonferroni}}$) was set at 0.007 ($\alpha_{\text{Bonferroni}} = \frac{\alpha}{g}$, where $g = 7$; 1 comparison/flexion angle × 7 flexion angles).

To determine how well KA TKA limited anterior translation of the tibial contact locations in either or both compartments, a one-factor repeated measures ANOVA was

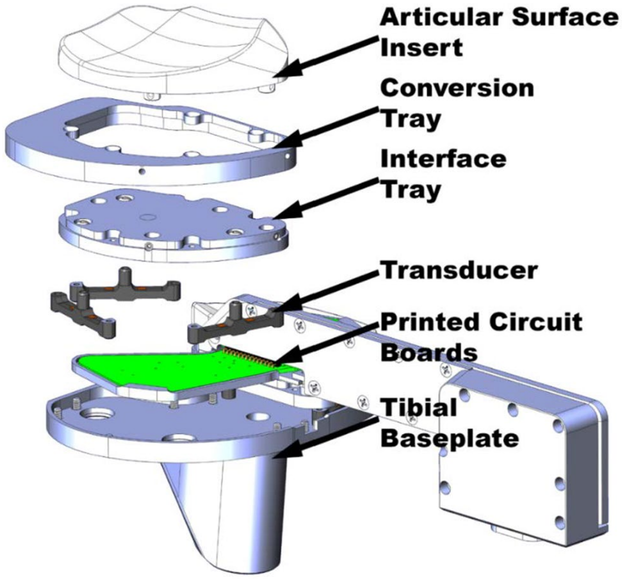

Fig. 4 Image showing an isometric view of the custom tibial force sensor [48] with the medial compartment exploded to show the five layers. The first layer, which is the most distal, is a modified tibial baseplate (Persona CR size D, Zimmer, Inc.) that has been hollowed out from the proximal surface. The second layer consists of printed circuit boards that are used to complete the Wheatstone bridge circuit of each of the six transducers. The third layer consists of two triangular arrays of three custom transducers each; one array is in the medial compartment, and the other is in the lateral compartment. The fourth layer consists of the medial and lateral trays. The interface trays provide a rigid connection between the transducers and the tibial articular surface inserts, which make up the fifth layer. Conversion trays can be attached to the interface trays to accommodate larger articular surface inserts. The fifth and most proximal layer consists of independent medial and lateral tibial articular surface inserts, which are 3D printed (Grey 60, Stratasys Ltd, Eden Prairie, MN). These inserts have the same articular shape as the standard tibial articular surfaces and come in different sizes and thicknesses so that the overall size and thickness of the tibial force sensor match those of the standard tibial component with the proper thickness articular surface insert. Once assembled, the internal cavity between the hollowed-out baseplate and interface trays was filled with a low stiffness dielectric gel (SYLGARD™ 527 Silicone, Dow Corning, Midland, MS) to seal the electrical components but not interfere with the load transfer. The root-mean-squared errors (RMSEs) in tibial force and tibial contact location are ≤6.1 N and ≤1.6 mm, respectively [48]. The repeatability of the measurement of tibial force and net A–P translation of the tibial contact locations were determined based on the pooled standard deviation of five repeated trials in three KA TKAs. The pooled standard deviations of the tibial forces were ≤5.3 N at 10° through 120° of flexion (≤12.9 N at 0° of flexion). The pooled standard deviations of the net A–P translation from 0° to 120° of flexion were ≤1.6 mm

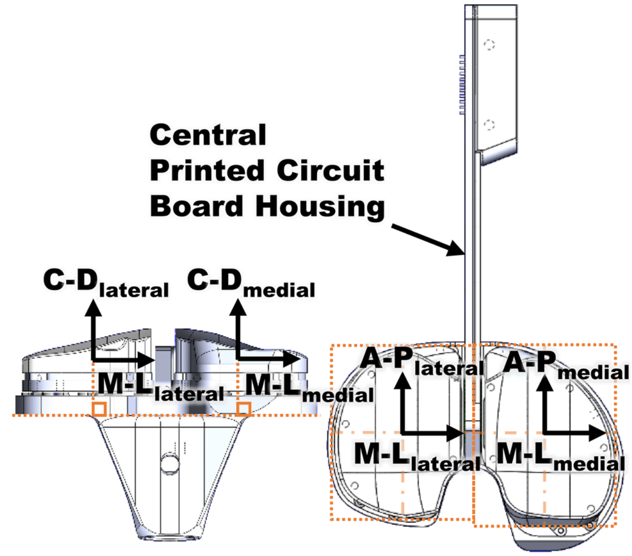

Fig. 5 Images show posterior (*left*) and proximal (*right*) views of the tibial force sensor with the coordinate system for each compartment. In both compartments, the compression–distraction (C–D) direction is normal to the transverse surface of the tibial baseplate. The anterior–posterior (A–P) direction is perpendicular to the C–D direction and parallel to the central printed circuit board housing, which is parallel to the central mating features of the tibial articular surface and is designed to be the A–P direction by the manufacturer. The medial–lateral (M–L) direction is the cross-product of the C–D and A–P unit vectors. The origin in the lateral compartment is located one-quarter of the M–L width of the articular surface from the lateral edge of the articular surface in the M–L direction and one-half the depth of the lateral compartment of the articular surface in the A–P direction. The origin in the medial compartment is located one-quarter of the M–L width of the articular surface from the medial edge of the articular surface in the M–L direction and at the same A–P location as in the lateral compartment in the anterior–posterior direction

performed for each compartment (JMP version 11.2.0; SAS Institute Inc., Cary, NC; www.jmp.com). The one factor was flexion angle at seven levels (0°, 10°, 30°, 45°, 60°, 90°, and 120°). Tukey's tests were used to compare the mean A–P tibial contact location in each compartment between flexion angles. The level of significance (α) was set at 0.05.

A power analysis confirmed that with thirteen knees, differences in tibial forces between compartments as small as 38 N could be detected with $\alpha = 0.05$ and $(1 - \beta) \geq 0.80$ using the maximum standard deviation of the differences in tibial forces between compartments for all flexion angles (43 N at 0° of flexion, Table 1).

Results

The total tibial forces varied with flexion (Fig. 6). The total tibial force was greatest at 0° of flexion (mean = 116 N), decreased to a minimum at 30° of flexion (mean = 5 N), and then increased during flexion from 30° to 120° (mean = 59 N at 120° of flexion).

The average differences in tibial forces between compartments after KA TKA also varied with flexion (Fig. 7). The greatest average difference in tibial forces between

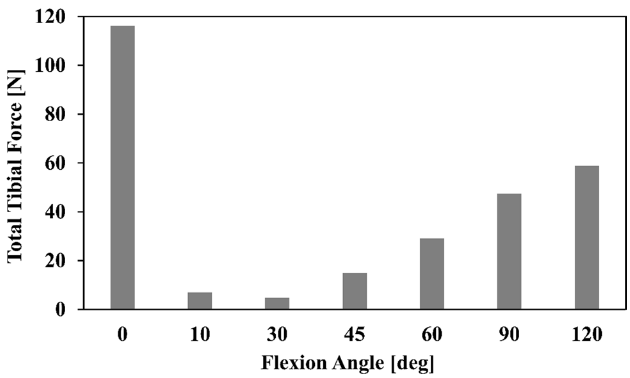

Fig. 6 *Vertical bar chart* shows the average total tibial force caused by the soft tissue restraints during passive flexion (i.e. the average difference between the sum of the measured medial and lateral tibial forces and the compressive component of the muscle loads (see "Appendix 2: Anterior–posterior and compression–distraction components of applied muscle loads" section, Fig. 14 for details)

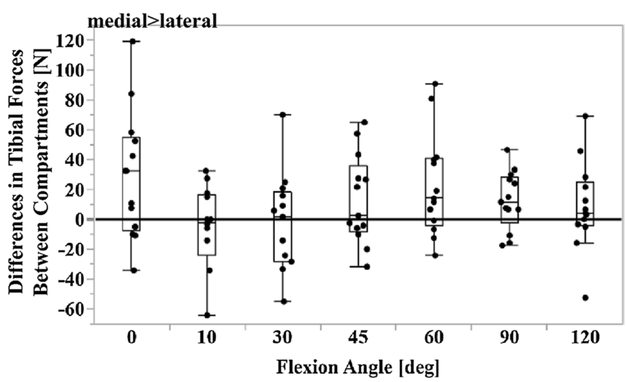

Fig. 7 The *box-and-whisker plot* shows the differences in tibial forces between compartments for each of the thirteen KA TKAs at each flexion angle. The *lower* and *upper lines* defining the box indicate the 1st quartile and the 3rd quartile, respectively, and the whiskers extend from the ends of the box to the outermost data point that falls within the distances computed as follows: 3rd quartile + 1.5 × (interquartile range) and 1st quartile − 1.5 × (interquartile range)

compartments (29 N) occurred at 0° of flexion, and 0° of flexion was the only flexion angle where the difference in tibial forces between compartments was statistically significant ($p = 0.0008$) (Fig. 7; Table 1). On average, the medial tibial force was greater than the lateral tibial force from 45° to 120° of flexion (range of average imbalance = 9–23 N) (Fig. 7; Table 1). The difference in tibial forces between compartments in the thirteen KA TKAs across all flexion angles ranged from 119 N (medial > lateral) at 0° of flexion to −64 N (lateral > medial) at 10° of flexion (Fig. 7; Table 1).

Table 1 Summary statistics of differences in tibial forces between compartments after KA TKA during passive flexion

Flexion angle	Differences in tibial forces between compartments (medial–lateral) mean ± SD (range)
0°	29.2 ± 42.6 N (−34.4–119.1 N)
10°	−5.1 ± 26.9 N (−64.0–32.4 N)
30°	−2.7 ± 32.6 N (−55.2–69.7 N)
45°	13.1 ± 29.9 N (−31.4–65.0 N)
60°	22.9 ± 34.4 N (−23.8–91.2 N)
90°	12.7 ± 19.5 N (−17.1–46.5 N)
120°	8.8 ± 29.3 N (−52.1–68.8 N)

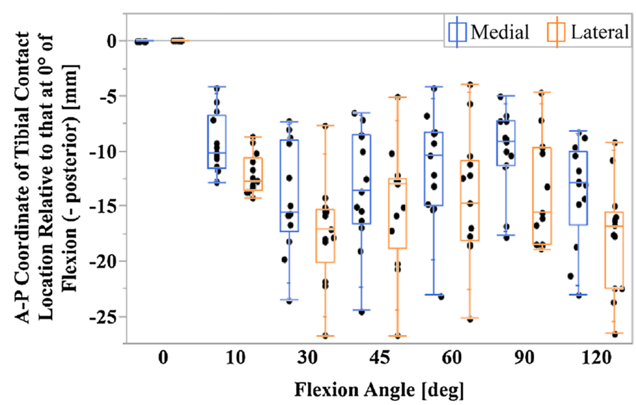

Fig. 8 *Box-and-whisker plot* show the A–P coordinate of the contact location in each compartment during passive flexion from 0° to 120°. The *lower* and *upper lines* defining the box indicate the 1st quartile and the 3rd quartile, respectively, and the whiskers extend from the ends of the box to the outermost data point that falls within the distances computed as follows: 3rd quartile + 1.5 × (interquartile range) and 1st quartile − 1.5 × (interquartile range)

In all thirteen KA TKAs, the tibial contact locations between 30° and 120° of flexion were posterior to those at 0° of flexion in both compartments (Fig. 8; Table 2). The average A–P coordinates of the tibial contact location at 120° of flexion were −13.8 mm ($p < 0.0001$) and −17.8 mm ($p < 0.0001$) more posterior than those at 0° of flexion in the medial and lateral compartments, respectively. The net posterior translation from 0° to 120° of flexion ranged from −8.2 to −23.0 mm in the medial compartment and from −9.2 to −26.6 mm in the lateral compartment (Fig. 8; Table 2).

Discussion

There were three important findings in the present study. The first important finding was that KA TKA limited the average

Table 2 Summary statistics of A–P translation of medial and lateral tibial contact locations after KA TKA during passive flexion

Flexion Angle	A–P translation of tibial contact location relative to that at 0° of flexion (−posterior translation)	
	Lateral compartment mean ± SD (range)	Medial compartment mean ± SD (range)
10°	−9.4 ± 2.8 mm (−4.2 to −12.8 mm)	−12.1 ± 1.8 mm (−8.7 to −14.2 mm)
30°	−17.4 ± 4.6 mm (−7.7 to −26.8 mm)	−14.3 ± 4.9 mm (−7.3 to −23.5 mm)
45°	−15.0 ± 5.4 mm (−5.1 to −26.8 mm)	−13.4 ± 5.2 mm (−6.5 to −24.5 mm)
60°	−14.1 ± 5.7 mm (−4.0 to −25.5 mm)	−11.5 ± 4.9 mm (−4.2 to −23.1 mm)
90°	−13.7 ± 4.9 mm (−4.7 to −18.9 mm)	−10.0 ± 3.7 mm (−5.0 to −17.7 mm)
120°	−17.8 ± 4.7 mm (−9.2 to −26.6 mm)	−13.8 ± 4.6 mm (−8.2 to −23.0 mm)

total tibial forces to 116 N from 0° to 120° of flexion without soft tissue release. The second important finding was that KA TKA limited the average differences in tibial forces between compartments to 29 N from 0° to 120° of flexion. The third important finding was that KA TKA enabled a net posterior translation of the contact locations during passive flexion from 0° to 120° of flexion in both compartments.

The magnitudes of the average total tibial forces after KA TKA are somewhat lower than those measured previously in the native knee [52] (Fig. 9). In the study that measured the tibial forces in the native knee [52], the measured tibial forces included contributions from both the soft tissue restraints and the passive stretch of muscles crossing the knee. Accordingly, it is not unexpected that the total tibial forces measured in the present study are less than those measured previously in the native knee [52] (Fig. 9) because the contribution of the soft tissue restraints was isolated in the present study (see "Appendix 2: Anterior–posterior and compression–distraction components of applied muscle loads" section) and passive muscle forces were absent. The low average total tibial forces indicate that KA TKA reduces the risk of overly tight soft tissue restraints that might cause persistent pain, stiffness, and limited range of motion [4, 22].

The average differences in tibial forces after KA TKA closely matched those measured in a previous study in the native knee [52] throughout flexion (Fig. 9). This similarity indicates that KA TKA closely restores the balance between the tensions in the medial and lateral soft tissue restraints to native. In KA TKA, the soft tissues are balanced by adjusting the V–V, P–D, and F–E of the tibial resection [26, 38]. Thus, when the plane of the proximal tibia and the joint lines of the femur are closely restored (Figs. 1, 3), the soft tissues are also balanced [45]. Because the tibial forces in the medial and lateral compartments are determined by the interaction between the articular surfaces and the soft tissue restraints [10, 57, 58], these results indicate that the components were aligned to closely restore both the articular surfaces and the tension in the soft tissue restraints to native.

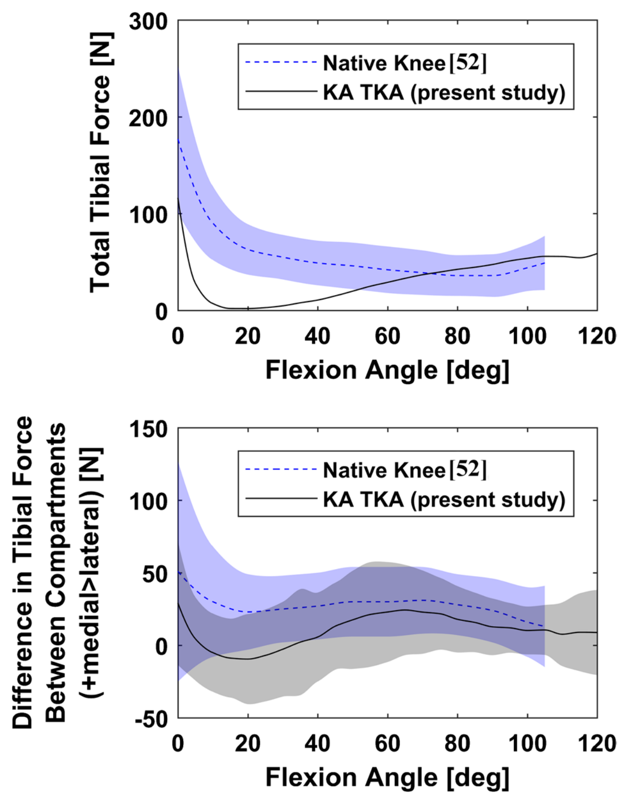

Fig. 9 Comparison between total tibial force (*top*) and difference in tibial forces between compartments (*bottom*) measured previously in the native knee [52] and measured in the present study after KA TKA. The lines represent the mean, and the shaded regions represent ±1 standard deviation. Note that there is no shading for the total tibial force in the present study because only the average measured total tibial force was adjusted to remove the effect of the applied muscle forces (see "Appendix 2: Anterior–posterior and compression–distraction components of applied muscle loads" section). The total tibial forces in the present study were somewhat lower than those measured previously in native knees from full extension to mid-flexion and comparable from mid-flexion to deep flexion. The differences in tibial forces between compartments were similar in the native knee and after KA TKA because the shaded regions overlap throughout flexion

Additionally, these results demonstrate that the quality assurance checks [26, 38] used to confirm that the femoral and tibial components were correctly aligned are effective at helping the surgeon closely restore the articular surfaces and tensions in the soft tissue restraints to native.

It is important to note that KA TKA limited the differences in tibial forces between compartments without either soft tissue releases or the use of an intraoperative sensor. While comparisons must be made with caution because different alignment goals were used in the previous studies that measured tibial forces intraoperatively, it is striking that 13–42% of the TKAs after mechanically aligned TKA had differences in tibial forces between compartments that the authors considered undesirable [19, 20, 35] compared to 0% after KA TKA in the present study based on the same criteria. Further, in previous studies of mechanically aligned TKA [19, 20], patients were left with differences in tibial forces between compartments considered undesirable by the authors because the surgeons were concerned that further releases could cause post-operative instability. Thus, by striving to restore the native soft tissue balance by aligning the components congruent with the native joint lines using simple quality assurance checks [26, 38], KA TKA closely restored to native the differences in tibial forces between compartments (Fig. 9). While confirmation in vivo is warranted, these findings indicate that KA TKA might better limit differences in tibial forces between compartments than mechanically aligned TKA, which provides one possible explanation for why KA TKA has high rates of patient satisfaction and function [13, 14, 25, 27, 28, 37] and why patients are three times more likely than those with mechanically aligned TKA to report that their knee feels normal [37].

In addition to limiting high tibial forces and differences in tibial forces between compartments, KA TKA also prevented a net anterior translation of the contact locations during passive flexion in both compartments. The net posterior translation of the tibial contact locations during passive flexion from 0° to 120° in all thirteen TKAs (Fig. 8, Table 2) demonstrates that KA TKA is unlikely to limit flexion due to impingement between the femur and the tibia [6, 7, 50]. This net posterior translation of the tibial contact locations despite the net posterior force on the tibia due to the muscle loads from 30° to 120° of flexion (Fig. 13) indicates that KA TKA properly tensioned the posterior cruciate ligament, which engages between 60° and 90° of flexion and drives posterior translation of the tibial contact locations [2, 9, 21, 54]. While comparison to the results of prior studies must be made with caution due to different loading conditions, the average posterior translations from 0° to 120° of flexion in the medial and lateral compartments (16 mm and 18 mm, Fig. 8, Table 2) are within the ranges measured in the medial and lateral compartments of native knees in previous studies of 11–20 mm and 17–22 mm [31, 33, 40], respectively.

These three important findings have clinical relevance related to soft tissue balancing in TKA. First, these findings demonstrate that striving to restore the native joint lines and adjusting component alignment to balance the soft tissues rather than performing releases can limit both high tibial forces that might cause stiffness and pain [4] and differences between tibial forces that might lead to patient dissatisfaction [19, 20]. Avoiding unnecessary releases is important because it reduces operative times, reduces the risk of over-release that might lead to instability, and reduces trauma to the soft tissues that should lead to quicker recovery. Second, because the tibial forces were generally low (Figs. 6, 9), the tibial forces are not a good metric for detecting overly loose TKAs (i.e. instability). Instability is arguably better detected based on knee laxities where excessive separation between the components will be apparent. However, previous studies have indicated that surgeons have difficulty detecting overly tight knees based on laxities [12, 15]. Thus, the tibial forces are likely a better metric to detect an overly tight knee because the tibial forces should be sensitive to over-tightness due to the stiffness of the soft tissue restraints [30, 43, 44].

Two limitations should be considered when translating these findings into the clinical realm. One is that the muscle loads applied were chosen to maintain the stability of the joint and not to represent any specific physiologic loading condition; hence, the muscle loads might cause non-physiologic tibial forces and contact kinematics. Regarding the total tibial forces, the applied muscle forces contributed substantively to the measured tibial forces (Fig. 14), which is why their contribution was removed from the total tibial forces reported (Fig. 6). Surgeons using intraoperative sensors should carefully consider the external loads being applied to the knee during measurements because these loads are likely to influence the measured forces especially because the contributions of the soft tissue restraints are small (Fig. 6). Regarding the difference in tibial forces between compartments, the applied muscle forces had a negligible effect (<2 N) on the measured differences in these forces between compartments (Fig. 11). Regarding the contact kinematics, the anterior translation of the tibial contact locations between 30° and 90° of flexion was likely caused by the posterior component of the net muscle loads (Fig. 13).

The second limitation is that KA TKA was performed on native knees rather than knees with osteoarthritis (OA). In clinical practice, patients' knees will have end-stage OA, which might include contracture, lengthening, and/or stiffening of the soft tissue restraints [8, 41, 51]. These changes to the soft tissue restraints must be accounted for during soft tissue balancing. Traditionally, soft tissue balancing involves releases of the soft tissue restraints [55], and the amount of soft tissue releases might differ between native and OA knees due to the possible changes in the soft tissue restraints due to OA. Thus, because the difference in tibial forces between compartments is determined by the relative tensions in the soft tissue restraints, different amounts of soft tissue releases would lead to altered differences in tibial forces between compartments. However, KA TKA largely avoids soft tissue releases during soft tissue balancing by adjusting the alignment of the tibial component to achieve the desired balance [26, 28, 38]. Therefore, because KA TKA largely avoids soft tissue releases, the differences in tibial forces between compartments measured in the present study should be representative of those occurring intraoperatively.

Conclusions

After KA TKA, average total tibial forces due to the soft tissue restraints were limited to 116 N, average differences in tibial forces between compartments were limited to 29 N, and a net posterior translation of the tibial contact locations was observed in all knees during passive flexion from 0° to 120°. These results are consistent with those measured in previous studies of native knees [31, 40, 52]. While confirmation in vivo is warranted, these findings give surgeons who perform kinematically aligned TKA confidence that the alignment method and surgical technique limit high tibial forces, differences in tibial forces between compartments, and anterior translation of the tibial contact locations during passive flexion. These findings support previous clinical studies that reported that KA TKA leads to high patient satisfaction and function [13, 14, 25, 27, 28, 37], has a low risk of failure [25, 27], and has a low prevalence of abnormal contact kinematics during kneeling [24].

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Authors contributions JDR, MLH, and SMH collectively conceived of and designed the study. JDR carried out data collection and analysis. JDR and MLH drafted the manuscript. JDR, MLH, and SMH read and approved the final manuscript.

Compliance with ethical standards

Conflict of interest Two of the authors received research support from Zimmer Biomet related to this study (MLH and SMH). Two of the authors received research support from Think Surgical for unrelated studies (JDR and MLH). One of the authors is a paid consultant of Zimmer Biomet and Think Surgical (SMH). One of the authors received royalties from Saunders/Mosby-Elsevier and Zimmer Biomet (SMH).

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Ethical approval Following University of California policies, this study did not require institutional review board (IRB) approval because de-identiied cadaveric specimens were used.

Informed consent The informed consent does not apply to this study because de-identified cadaveric specimens were used.

Appendix 1: Contribution of applied muscle loads to differences in tibial forces between compartments

The differences in tibial forces between compartments ($F_{\mathrm{diff},i}$) created by the applied muscle loads at each flexion angle (i = 0°, 30°, 60°, 90°, and 120°) can be estimated using average anatomy and lines of action [3, 11] (Fig. 10, Eq. 1) where $\vec{r}_{\mathrm{BF}}$, $\vec{r}_{\mathrm{SMST}}$, and $\vec{r}_{\mathrm{Q}}$ are the vectors from the centre of

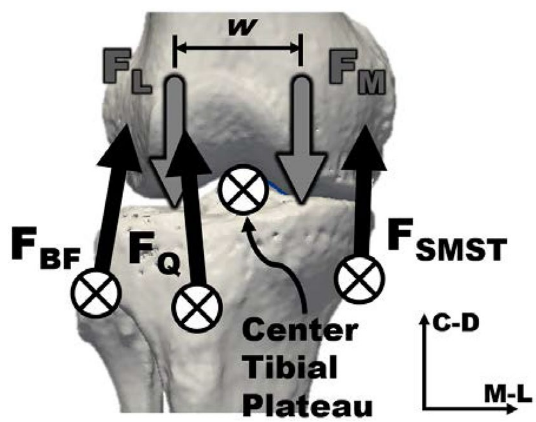

Fig. 10 Coronal view of free body diagram of the knee with applied muscle loads

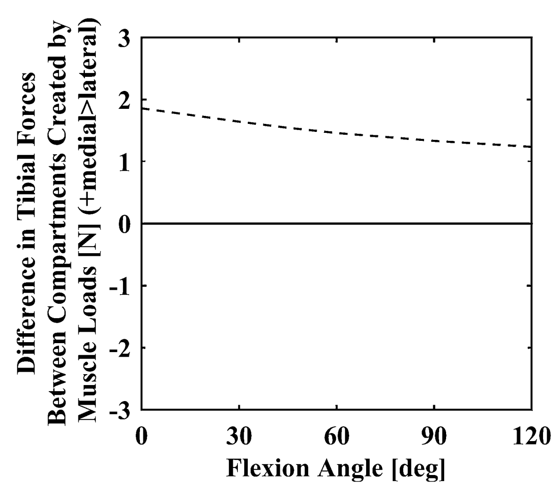

Fig. 11 Line plot shows that applied muscle forces had a negligible effect on the differences in tibial forces between compartments from 0° and 120° of flexion measured in the present study

the tibial plateau to the tibial insertions of the biceps femoris (BF), semitendinosus/semimembranosus (SMST), and the patellar tendon (Q), respectively; $\vec{F}_{\mathrm{BF},i}$, $\vec{F}_{\mathrm{SMST},i}$, and $\vec{F}_{\mathrm{Q},i}$ are the vectors whose orientation is set by the line of action and magnitude is set by the applied loads at the tibial insertions of the biceps femoris, semitendinosus/semimembranosus, and the patellar tendons, respectively; w is the average medial–lateral spacing between the contact locations in the medial and lateral compartments. The magnitude of $\vec{F}_{\mathrm{Q},i}$ was computed using the average ratio of the load in the quadriceps tendon to that in the patellar tendon [42].

$$\vec{F}_{\mathrm{diff},i} = \frac{\vec{r}_{\mathrm{BF}} \times \vec{F}_{\mathrm{BF},i} + \vec{r}_{\mathrm{SMST}} \times \vec{F}_{\mathrm{SMST},i} + \vec{r}_{\mathrm{Q}} \times \vec{F}_{\mathrm{Q},i}}{w} \tag{1}$$

Appendix 2: Anterior–posterior and compression–distraction components of applied muscle loads

The total tibial force ($F_{\mathrm{total},i}$) can be decomposed into the contribution of the muscle loads ($F_{\mathrm{muscle},i}$) and the contribution of the soft tissue restraints ($F_{\mathrm{soft\ tissue},i}$) at each flexion angle ($i = 0°, 10°, 30°, 45°, 60°, 90°$, and $120°$) (Eq. 2). The total tibial force is calculated as the sum of the medial and lateral tibial forces computed using the tibial force sensor. The anterior–posterior (A–P) and compression–distraction (C–D) components of the muscle loads are computed based on applied load to each muscle, the alignment of the muscle loads relative to the tibia [3, 11], and the ratio of the load in the quadriceps tendon to that in the patellar tendon ($r_{Q/\mathrm{Pat},i}$) [42] (Fig. 12).

$$\vec{F}_{\mathrm{total},i} = \vec{F}_{\mathrm{muscle},i} + \vec{F}_{\mathrm{soft\ tissue},i} \tag{2}$$

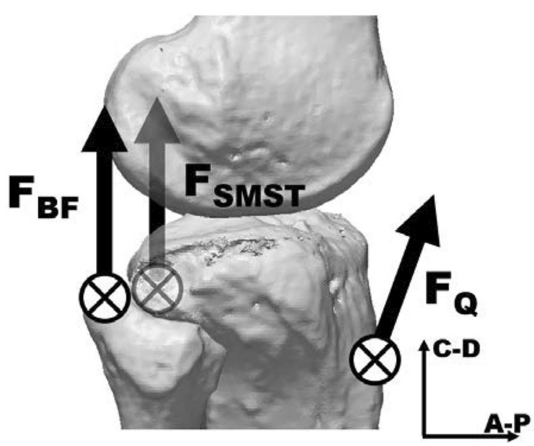

Fig. 12 Sagittal view of a free body diagram of the knee with applied muscle loads

Between 0° and 30° of flexion, the net A–P force component of the muscle loads is directed anteriorly, whereas between 30° and 120° of flexion, the net A–P force component of the muscle loads is directed posteriorly (Fig. 13). The anterior translation of both tibial contact locations between 30° and 90° (Fig. 8; Table 2) is likely driven by the muscle forces pulling the tibia posteriorly on the femur. Because the soft tissues are minimally loaded between these flexion angles (Fig. 14), the small A–P forces applied by the muscles can drive the A–P translation of the tibial contact locations.

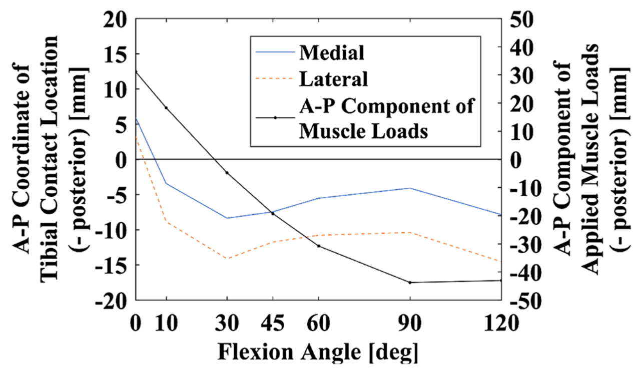

Fig. 13 Line plot show the A–P force component of the muscle loads overlaid on the average A–P coordinates of the tibial contact locations. The initiation of anterior translation of the tibial contact locations (Fig. 8; Table 2) coincides with the switch from an anteriorly directed component of the muscle forces to a posteriorly directed component. The posteriorly directed force would pull the tibia more posterior on the femur, which would cause the tibial contact locations to translate anteriorly. The anterior translation begins to level off between 60° and 90° of flexion when the posterior cruciate ligament begins to tighten [2, 9], which limits the posterior translation of the tibia

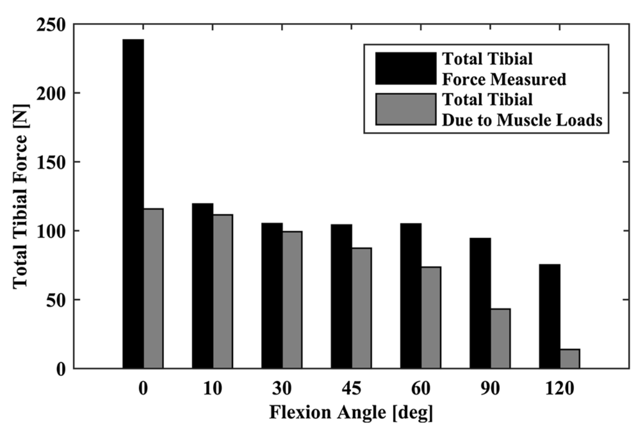

Fig. 14 *Vertical bar graph* shows the average total tibial force measured and the compressive force component of the applied muscle loads

References

1. Amis AA, Dawkins GP (1991) Functional anatomy of the anterior cruciate ligament. Fibre bundle actions related to ligament replacements and injuries. J Bone Joint Surg Br 73(2):260–267
2. Amis AA, Gupte CM, Bull AM, Edwards A (2006) Anatomy of the posterior cruciate ligament and the meniscofemoral ligaments. Knee Surg Sports Traumatol Arthrosc 14(3):257–263
3. Arnold EM, Ward SR, Lieber RL, Delp SL (2010) A model of the lower limb for analysis of human movement. Ann Biomed Eng 38(2):269–279
4. Babazadeh S, Stoney JD, Lim K, Choong PFM (2009) The relevance of ligament balancing in total knee arthroplasty: how important is it? A systematic review of the literature. Orthop Rev (Pavia). doi:10.4081/or.2009.e26
5. Bach JM, Hull ML (1995) A new load application system for in vitro study of ligamentous injuries to the human knee joint. J Biomech Eng 117(4):373–382
6. Banks S, Bellemans J, Nozaki H, Whiteside LA, Harman M, Hodge WA (2003) Knee motions during maximum flexion in fixed and mobile-bearing arthroplasties. Clin Orthop Relat Res 410(5):131–138
7. Bellemans J, Banks S, Victor J, Vandenneucker H, Moemans A (2002) Fluoroscopic analysis of the kinematics of deep flexion in total knee arthroplasty. Influence of posterior condylar offset. J Bone Joint Surg Br 84(1):50–53
8. Bellemans J, Vandenneucker H, Vanlauwe J, Victor J (2010) The influence of coronal plane deformity on mediolateral ligament status: an observational study in varus knees. Knee Surg Sports Traumatol Arthrosc 18(2):152–156
9. Belvedere C, Ensini A, Feliciangeli A, Cenni F, D'Angeli V, Giannini S, Leardini A (2012) Geometrical changes of knee ligaments and patellar tendon during passive flexion. J Biomech 45(11):1886–1892
10. Blankevoort L, Huiskes R, de Lange A (1988) The envelope of passive knee joint motion. J Biomech 21(9):705–720
11. Defrate LE, Nha KW, Papannagari R, Moses JM, Gill TJ, Li G (2007) The biomechanical function of the patellar tendon during in vivo weight-bearing flexion. J Biomech 40(8):1716–1722
12. Delport P, Labey L, De Corte R, Innocenti B, Vander Sloten J, Bellemans J (2013) Collateral ligament strains during knee joint laxity evaluation before and after TKA. Clin Biomech (Bristol, Avon) 28(7):777–782
13. Dossett HG, Estrada NA, Swartz GJ, LeFevre GW, Kwasman BG (2014) A randomised controlled trial of kinematically and mechanically aligned total knee replacements: two-year clinical results. Bone Joint J 96-B(7):907–913
14. Dossett HG, Swartz GJ, Estrada NA, LeFevre GW, Kwasman BG (2012) Kinematically versus mechanically aligned total knee arthroplasty. Orthopedics 35(2):e160–e169. doi:10.3928/01477447-20120123-04
15. Elmallah RK, Mistry JB, Cherian JJ, Chughtai M, Bhave A, Roche MW, Mont MA (2016) Can we really "Feel" a balanced total knee arthroplasty? J Arthroplasty 31(9 Suppl):102–105
16. Erskine RM, Jones DA, Maganaris CN, Degens H (2009) In vivo specific tension of the human quadriceps femoris muscle. Eur J Appl Physiol 106(6):827–838
17. Ghosh KM, Merican AM, Iranpour F, Deehan DJ, Amis AA (2012) Length-change patterns of the collateral ligaments after total knee arthroplasty. Knee Surg Sports Traumatol Arthrosc 20(7):1349–1356
18. Grood ES, Suntay WJ (1983) A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. J Biomech Eng 105(2):136–144
19. Gustke KA, Golladay GJ, Roche MW, Elson LC, Anderson CR (2014) A new method for defining balance: promising short-term clinical outcomes of sensor-guided TKA. J Arthroplasty 29(5):955–960
20. Gustke KA, Golladay GJ, Roche MW, Jerry GJ, Elson LC, Anderson CR (2014) Increased satisfaction after total knee replacement using sensor-guided technology. Bone Joint J 96-B(10):1333–1338
21. Harner CD, Giffin JR, Vogrin TM, Woo SLY (2001) Anatomy and biomechanics of theposterior cruciate ligament and posterolateral corner. Oper Techn Sport Med 9(2):39–46
22. Heesterbeek PJ, Verdonschot N, Wymenga AB (2008) In vivo knee laxity in flexion and extension: a radiographic study in 30 older healthy subjects. Knee 15(1):45–49
23. Hosseini A, Qi W, Tsai TY, Liu Y, Rubash H, Li G (2015) In vivo length change patterns of the medial and lateral collateral ligaments along the flexion path of the knee. Knee Surg Sports Traumatol Arthrosc 23(10):3055–3061
24. Howell SM, Hodapp EE, Vernace JV, Hull ML, Meade TD (2013) Are undesirable contact kinematics minimized after kinematically aligned total knee arthroplasty? An intersurgeon analysis of consecutive patients. Knee Surg Sports Traumatol Arthrosc 21(10):2281–2287
25. Howell SM, Howell SJ, Kuznik KT, Cohen J, Hull ML (2012) Does a kinematically aligned total knee arthroplasty restore function without failure regardless of alignment category? Clin Orthop Relat Res 471(3):1000–1007
26. Howell SM, Hull ML, Mahfouz MR (2017) Kinematically aligned total knee arthroplasty. In: Scott S (ed) Insall and Scott Surgery of the Knee, 6th edn. Elsevier, Philadelphia, PA, pp 1784–1796
27. Howell SM, Papadopoulos S, Kuznik K, Ghaly LR, Hull ML (2015) Does varus alignment adversely affect implant survival and function six years after kinematically aligned total knee arthroplasty? Int Orthop 39(11):2117–2124
28. Howell SM, Papadopoulos S, Kuznik KT, Hull ML (2013) Accurate alignment and high function after kinematically aligned TKA performed with generic instruments. Knee Surg Sports Traumatol Arthrosc 21(10):2271–2280
29. Hunt NC, Ghosh KM, Blain AP, Athwal KK, Rushton SP, Amis AA, Longstaff LM, Deehan DJ (2014) How does laxity after single radius total knee arthroplasty compare with the native knee? J Orthop Res 32(9):1208–1213
30. LaPrade RF, Bollom TS, Wentorf FA, Wills NJ, Meister K (2005) Mechanical properties of the posterolateral structures of the knee. Am J Sports Med 33(9):1386–1391
31. Li G, Most E, Sultan PG, Schule S, Zayontz S, Park SE, Rubash HE (2004) Knee kinematics with a high-flexion posterior stabilized total knee prosthesis: an in vitro robotic experimental investigation. J Bone Joint Surg Am 86-A(8):1721–1729
32. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL (1999) The importance of quadriceps and hamstring muscle loading on knee kinematics and in situ forces in the ACL. J Biomech 32(4):395–400
33. Lu TW, Tsai TY, Kuo MY, Hsu HC, Chen HL (2008) In vivo three-dimensional kinematics of the normal knee during active extension under unloaded and loaded conditions using single-plane fluoroscopy. Med Eng Phys 30(8):1004–1012
34. Markolf KL, Gorek JF, Kabo JM, Shapiro MS (1990) Direct measurement of resultant forces in the anterior cruciate ligament. An in vitro study performed with a new experimental technique. J Bone Joint Surg Am 72(4):557–567
35. Meneghini RM, Ziemba-Davis MM, Lovro LR, Ireland PH, Damer BM (2016) Can intraoperative sensors determine the "target" ligament balance? Early outcomes in total knee arthroplasty. J Arthroplasty 31(10):2181–2187

36. Merican AM, Ghosh KM, Deehan DJ, Amis AA (2009) The transpatellar approach for the knee in the laboratory. J Orthop Res 27(3):330–334
37. Nam D, Nunley RM, Barrack RL (2014) Patient dissatisfaction following total knee replacement: a growing concern? Bone Joint J 96-B(11 Supple A):96–100
38. Nedopil AJ, Howell SM, Hull ML (2016) Does malrotation of the tibial and femoral components compromise function in kinematically aligned total knee arthroplasty? Orthop Clin North Am 47(1):41–50
39. Neter J, Kutner MH, Nachtsheim CJ, Wasserman W (1996) Analysis of factor effects in two-factor studies-equal sample sizes. In: Irwin (ed) Applied linear statistical models. 4th edn. McGraw-Hill, Chicago, pp 849-874
40. Pinskerova V, Johal P, Nakagawa S, Sosna A, Williams A, Gedroyc W, Freeman MA (2004) Does the femur roll-back with flexion? J Bone Joint Surg Br 86(6):925–931
41. Pottenger LA, Phillips FM, Draganich LF (1990) The effect of marginal osteophytes on reduction of varus-valgus instability in osteoarthritic knees. Arthritis Rheum 33(6):853–858
42. Powers CM, Chen YJ, Scher IS, Lee TQ (2010) Multiplane loading of the extensor mechanism alters the patellar ligament force/quadriceps force ratio. J Biomech Eng 132(2):024503
43. Race A, Amis AA (1994) The mechanical properties of the two bundles of the human posterior cruciate ligament. J Biomech 27(1):13–24
44. Robinson JR, Bull AM, Amis AA (2005) Structural properties of the medial collateral ligament complex of the human knee. J Biomech 38(5):1067–1074
45. Roth JD, Howell SM (2017) Soft tissue balance of the native knee provides guidance for balancing a total knee arthroplasty. In: Matsuda S, Lustig S, van der Merwe W (eds) Soft Tissue Balancing in Total Knee Arthroplasty, pp 17–27
46. Roth JD, Howell SM, Hull ML (2015) Native knee laxities at 0°, 45°, and 90° of flexion and their relationship to the goal of the gap-balancing alignment method of total knee arthroplasty. J Bone Joint Surg Am 97(20):1678–1684
47. Roth JD, Howell SM, Hull ML (2017) Characterization and correction of errors in computing contact location between curved articular surfaces: application to total knee arthroplasty. J Biomech Eng 139(6). doi:10.1115/1.4036147
48. Roth JD, Howell SM, Hull ML (2017) An improved tibial force sensor to compute contact forces and contact locations in vitro after total knee arthroplasty. J Biomech Eng 139(4):041001–041008
49. Roth JD, Hull ML, Howell SM (2015) The limits of passive motion are variable between and unrelated within normal tibiofemoral joints. J Orthop Res 33(11):1594–1602
50. Sharma A, Dennis DA, Zingde SM, Mahfouz MR, Komistek RD (2014) Femoral condylar contact points start and remain posterior in high flexing patients. J Arthroplasty 29(5):945–949
51. Sharma L, Lou C, Felson DT, Dunlop DD, Kirwan-Mellis G, Hayes KW, Weinrach D, Buchanan TS (1999) Laxity in healthy and osteoarthritic knees. Arthritis Rheum 42(5):861–870
52. Verstraete MA, Meere PA, Salvadore G, Victor J, Walker PS (2017) Contact forces in the tibiofemoral joint from soft tissue tensions: implications to soft tissue balancing in total knee arthroplasty. J Biomech 58:195–202
53. Ward SR, Eng CM, Smallwood LH, Lieber RL (2009) Are current measurements of lower extremity muscle architecture accurate? Clin Orthop Relat Res 467(4):1074–1082
54. Watanabe T, Ishizuki M, Muneta T, Banks SA (2013) Knee kinematics in anterior cruciate ligament-substituting arthroplasty with or without the posterior cruciate ligament. J Arthroplasty 28(4):548–552
55. Whiteside LA (2002) Soft tissue balancing: the knee. J Arthroplasty 17(4 Suppl 1):23–27
56. Whiteside LA, Saeki K, Mihalko WM (2000) Functional medial ligament balancing in total knee arthroplasty. Clin Orthop Relat Res 380(11):45–57
57. Wilson DR, Feikes JD, O'Connor JJ (1998) Ligaments and articular contact guide passive knee flexion. J Biomech 31(12):1127–1136
58. Wilson DR, Feikes JD, Zavatsky AB, O'Connor JJ (2000) The components of passive knee movement are coupled to flexion angle. J Biomech 33(4):465–473