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Neither Anterior nor Posterior Referencing Consistently Balances the Flexion Gap in Measured Resection Total Knee Arthroplasty: A Computational Analysis

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ABSTRACT

Background: Whether anterior referencing (AR) or posterior referencing (PR) produces a more balanced flexion gap in total knee arthroplasty (TKA) using measured resection remains controversial. Our goal was to compare AR and PR in terms of (1) medial and lateral gaps at full extension and 90° of flexion, and (2) maximum medial and lateral collateral ligament (MCL and LCL) forces in flexion.

Methods: Computational models of 6 knees implanted with posterior-stabilized TKA were virtually positioned with both AR and PR techniques. The ligament properties were standardized to achieve a balanced knee at full extension. Medial-lateral gaps were measured in response to varus and valgus loading at full extension and 90° of flexion; MCL and LCL forces were estimated during passive flexion. *Results:* At full extension, the maximum difference in the medial-lateral gap for both AR and PR was <1 mm in all 6 knee models. However, in flexion, only 3 AR and 3 PR models produced a difference in medial-lateral gap <2 mm. During passive flexion, the maximum MCL force ranged from 2 N to 87 N in AR and from 17 N to 127 N in PR models. The LCL was unloaded at >25° of flexion in all models. *Conclusion:* In measured resection TKA, neither AR nor PR better balance the ligaments and produce

symmetrical gaps in flexion. Alternative bone resection techniques and rotation alignment targets are needed to achieve more predictable knee balance.

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Anterior referencing (AR) and posterior referencing (PR) are the most common techniques to position and size the femoral component in total knee arthroplasty (TKA) using a measured resection approach [1,2]. Historically, the benefit of AR is reduced risk of notching the anterior femoral cortex and overstuffing the patellofemoral (PF) joint, but the resulting variation in the posterior bone resection may lead to less control of the flexion gap and joint instability [3–6]. Conversely, PR ensures a constant posterior condylar

offset [3], but PR runs the risk of notching the anterior cortex and overstuffing the PF joint [3–7]. Previous studies comparing the clinical outcomes between AR and PR reported no significant differences between the 2 techniques [3,5,6,8]. Therefore, surgeons choose a referencing technique according to implant design or personal preference. Recently, the introduction of more component sizes into some implant systems has reduced the risk of anterior notching and PF overstuffing. Nonetheless, whether the AR or the PR techniques better balances the collateral ligaments and produces symmetrical gaps in flexion and extension remains unknown.

Cadaveric experiments could be used to evaluate the biomechanical impact of each referencing technique [9,10]. The wide variations in both ligament properties and ligament balancing techniques, however, make it difficult to isolate the effect of AR and PR. In contrast, computational models enable control of ligament properties and balancing techniques to better identify the impact of AR and PR on knee mechanics. Therefore, the objective of our study was to compare the biomechanical effects of AR vs PR in terms of

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Fig. 1. Distal and posterior bony cuts of the femur and proximal cut of the tibia are shown for knee models with anterior referencing (AR) and posterior referencing (PR).

(1) symmetry between the medial and lateral gaps at full extension and 90° of flexion and (2) maximum medial and lateral collateral ligament (MCL and LCL) forces in passive flexion. Computational models of the tibiofemoral joint with a posteriorly stabilized (PS) TKA using a measured resection approach were used. We hypothesized that PR techniques would result in more reproducible and balanced collateral ligament forces and more equal flexion gaps during varus and valgus loading.

Methods

Computational models of the tibiofemoral joints from 6 cadaveric knees were virtually implanted with TKAs based on our previously developed framework [11–13]. We previously showed that our model of the native tibiofemoral joint predicted the kinematics and ligament forces measured in a cadaveric experiment [11,12]. We also used the same framework to assess the biomechanical impact of external rotation of the femoral component in PS TKA [13]. In the present study, we quantified differences in medial-lateral gaps and collateral ligament forces following AR and PR techniques by simulating both varus and valgus moments and passive flexion. We standardized the bony cuts for AR and PR, the ligament properties, and the knee balance at full extension to isolate the effect of AR and PR on knee mechanics.

Building the computational models required 3 steps [13]. First, under institutional review board approval, 3-dimensional bony geometries from the femoral head to the foot of 6 neutrally aligned, nonarthritic, male cadaveric legs (ages 33 ± 15 years) were reconstructed from computed tomography scans (Biograph mCT; Siemens, Erlangen, Germany) using image processing software (Mimics; Materialise, Leuven, Belgium).

Second, the geometries of the femoral and tibial components of a PS implant (Optetrak Logic; Exactech, Gainesville, FL) were obtained from computer-aided design files. The distal and posterior thicknesses of the femoral component were 8 mm, while the implanted tibial insert was 9 mm thick. Each component was virtually positioned to simulate TKA installation with measured resection using reverse engineering software (Geomagic, Morrisville, NC) [13].

Both AR and PR were simulated in each of the 6 knee models. The only difference between the 2 methods was the anterior and posterior bony cuts of the femoral condyles. In AR, the anterior cut was made first such that the finished cut surface was flush with the anterior femoral cortex (Fig. 1) [5]. In PR, 8 mm of bone was resected from the most posterior point of the medial condyle (Fig. 1). The femoral component was externally rotated in the transverse plane to align it parallel with the surgical transepicondylar axis (sTEA), which connected the lateral epicondyle to the center of the medial sulcus [14]. The resulting external rotation of the femoral component with respect to the posterior condylar axis ranged from 0.6° (Knee 2) to 4° (Knee 5) among the knees (Fig. 2). The femur was then sized, and the anterior femoral cut was made to accommodate the size of the implant [3]. For each knee, implant sizes were the same between AR and PR.

Across all 6 models, the amount of posterior bone resected both medially and laterally in AR and PR ranged from 10.8 to 4.2 mm and from 8 mm to 4.2 mm, respectively (Fig. 2). All other bony cuts and implant orientation were the same (Fig. 1). Specifically, 8 mm of bone were resected from the most distal condyle perpendicular to the femoral mechanical axis in the coronal plane [15]. A maximum of 9 mm of bone was resected from the highest point of the proximal tibia perpendicular to the tibial mechanical axis in the coronal plane [15]. The tibial component was internally rotated such that its center was aligned with the medial one-third of the tibial tubercle [16].

In the third step of building each model, a soft tissue envelope consisting of 20 fibers that represented the collateral ligaments and joint capsule was added to the tibiofemoral joint (Fig. 3). Specifically, the medial and lateral posterior capsule were each represented by 3 fibers, while the oblique popliteal ligament was represented by 2 fibers (Fig. 3A). The anterolateral, fabellofibular, and the LCLs were each represented by a single fiber (Fig. 3B). The MCL consisted of 3 proximal and 3 distal fibers wrapping around the medial aspect of the proximal tibia, while the posterior oblique ligament consisted of 3 fibers (Fig. 3C). Each fiber was defined as a tension-only, nonlinear force element using mean structural

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Fig. 2. The amounts of bone resected from the posterior condyles between AR (red) and PR (blue) for the 6 knee models are shown along with the external rotation of the posterior femoral cut relative to the posterior condylar axis, which varied from knee to knee.

properties reported in the literature [17,18]. Particular attention was given to define the proximal insertions of the MCL as the MCL force in flexion was found to be very sensitive to the insertion site. Specifically, the proximal insertion of the MCL resides posterior and

proximal to the medial epicondyle with an insertion area of approximately 10 mm in diameter [19,20]. Therefore, the locations of the proximal MCL fibers were defined to span this insertion area within the medial sulcus [19]. The central MCL fiber was placed at



Fig. 3. Fibers representing the ligaments in the knee model with TKA: (A) posterior lateral capsule (PLC, 3 fibers), posterior medial capsule (PMC, 3 fibers), oblique popliteal ligament (OPL, 2 fibers); (B) anterolateral ligament (ALL, 1 fiber), lateral collateral ligament (LCL, 1 fiber), fabello fibular ligament (FFL, 1 fiber); (C) superficial medial collateral ligament (sMCL, 3 fibers), posterior oblique ligament (POL, 3 fibers).

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Fig. 4. Medial and lateral gaps in response to 40 Nm of varus and valgus moments at 90° of flexion are shown in (A) AR and (B) PR knees. Only 3 knees produced differences between the medial and lateral gaps that were <1 mm for both AR and PR. The other 3 knees produced gap differences >3 mm.

the deepest point of the medial sulcus. The insertions of the anterior and posterior fibers were positioned midway between the center fiber and the anterior and posterior borders of the medial sulcus, respectively.

All ligament insertions and material properties were consistently defined across the 6 models for AR and for PR. In addition, because each knee model has a different geometry, the slack lengths, or the reference lengths at which the ligament fibers became taut, were standardized to achieve a balanced knee at full extension. Specifically, ligament slack length was defined using a previously developed optimization algorithm to produce in situ ligament pretension measured experimentally in the native knee at full extension [11]. To have a common reference for comparing AR and PR in each knee model, we used the slack length determined with the femoral component positioned at a standard posterior cut of 3° external rotation to the posterior condylar axis as defined in our previous work and following the manufacturer's surgical technique guide [13]. The slack lengths obtained from this standard component positioning were then applied to the corresponding AR and PR models of each knee.

All the model components were incorporated into a multibody dynamics program that generated and solved the equations of motion (MSC software; Adams, CA). The contact force at the articulation of the femoral component and the tibial insert was modeled as a nonlinear function of the penetration depth of the metallic femoral component into the polyethylene insert [12,13].

Two examinations that are commonly used to evaluate knee balance intraoperatively were simulated. First, varus and valgus moments of 40 Nm were applied at full extension and 90° of flexion with 10 N of compression applied across the tibiofemoral joint. This magnitude of moments represented a surgeon subjecting the ankle to a respective 80 N of medial and lateral force, assuming an average tibial length of 0.5 m [21]. The medial and lateral gaps were determined at full extension and 90° of flexion. We assumed that the knee that produces a difference between the medial and lateral gap ≤ 1 mm in extension and ≤ 2 mm in flexion is considered balanced, while the knee that produces larger gaps is considered unbalanced [22]. Second, passive flexion was applied from full extension to 90° of flexion with 500 N of compression across the tibiofemoral joint [23]. The resulting MCL and LCL forces were calculated throughout the range of flexion.



Fig. 5. Maximum MCL forces are shown for each of the 6 knee models for anterior AR and PR. Neither referencing system consistently produced higher or lower maximum MCL force.

Results

Gaps in AR and PR in Response to Varus/Valgus Moments

At full extension, AR and PR produced similar medial and lateral gaps in response to varus and valgus moments. The medial gaps were consistently larger than the lateral gaps across all knees. Specifically, the maximum difference between the medial and lateral gaps across the 6 knees was 1 mm (in Knee 3), and the minimum was 0.1 mm (in Knees 1 and 5).

At 90° of flexion, the difference between the medial and lateral gaps for 3 of the 6 knees (Knees 1, 2, and 6) with both AR and PR models was <1 mm (Fig. 4). In contrast, the gap differences for Knees 3, 4, and 5 reached 10, 3.5, and 6 mm, respectively, in both AR and PR models (Fig. 4). Unlike at full extension, in flexion, the lateral gaps were consistently larger than the medial gaps across all knees.

MCL and LCL Forces in AR and PR During Passive Flexion

During passive flexion, the maximum MCL force varied across the 6 models, ranging from 2 to 87 N with AR and from 17 to 127 N with PR (Fig. 5). This maximum force occurred at 90° of flexion in every knee. Neither AR nor PR models consistently produced the highest maximum MCL force. In particular, the maximum MCL force in the PR models of Knees 1, 3, and 6 was higher than that in the AR models by as much as 40 N. In contrast, the maximum MCL force in the PR model of Knee 4 was 15 N lower than that in the AR model. In Knees 2 and 5, however, similar maximum forces, differing by <2 N, were produced in AR and PR (Fig. 5). The LCL forces decreased to zero before reaching 25° of flexion in all knees.

Discussion

Both AR and PR models predicted neither symmetrical gaps in flexion nor consistent MCL and LCL forces with measured resection TKA. The flexion gaps and the collateral ligament forces were also variable from knee to knee in both AR and PR models despite balancing the knee in extension and aligning the femoral component parallel to the sTEA (Figs. 4 and 5). Interestingly, the range of maximum MCL force across knees (2 to 87 N in AR knees and 17 to 127 N in PR knees) was greater than the range of maximum MCL force between AR and PR for each knee (0 to 40 N; Fig. 5). It was also found that the knees that produced symmetrical gaps in flexion with AR models also produced symmetrical gaps with PR models.

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Fig. 6. The isometry of the MCL was assessed by checking the distance of the femoral insertions of the anterior fibers of the MCL to the posterior (r_p) and the distal (r_d) cuts. In Knee 3, r_p was greater than r_d . In Knee 6, r_p was equal to r_d .

Moreover, increased MCL forces in flexion corresponded to asymmetric gaps in both AR and PR knees. Conversely, the LCL was consistently unloaded at $>25^{\circ}$ of flexion in all knees. Our results support previous clinical studies that showed no differences between AR and PR in clinical outcomes such as postoperative range of motion and Knee Society scores [3,8].

To explain the variations in the flexion gaps and the MCL forces between AR and PR (Figs. 4 and 5), we measured the size of the resected bone from the posterior condyles (Fig. 2). It was found that the referencing technique that resected more posteromedial bone yielded less MCL force in flexion. In Knee 3, for example, 9.4 and 8 mm of posteromedial bone were resected in AR and PR, respectively (Fig. 2), and the maximum MCL force in AR (87 N) was lower than in PR (127 N). In Knee 4, conversely, 7 and 8 mm of posteromedial bone were resected in AR and PR, respectively, and the maximum MCL force in AR (84 N) was higher than in PR (69 N). In both of these examples, however, MCL forces remained elevated in flexion independent of the referencing technique. Therefore, the amount of posteromedial bone resection only predicted which of the 2 referencing techniques produced greater MCL force in flexion, but did not predict knee-to-knee variations in peak MCL force.

To further understand factors driving knee-to-knee variations in maximum MCL force, we assessed the isometry of the MCL through flexion after TKA. In the native knee, the MCL exhibits nearisometric behavior from 0° to 120° of flexion [24], but the effect of TKA on MCL isometry is not well understood. We focused on the isometry of the anterior fiber of the MCL because it carries most of the MCL load in flexion [25]. In Knee 3, where the MCL force had the greatest magnitude in flexion, the distances from the proximal insertions of the anterior fiber to the posterior cut were greater than to the distal cut (Fig. 6). In contrast, in Knee 6, where the MCL force was the least and similar in flexion and extension, the distance to the posterior and distal cuts was the same. Accordingly, in Knee 3, the anterior fiber elongated more during flexion resulting in less isometric MCL behavior leading to higher forces in this ligament. This pattern was consistent across all 6 knee models for both AR and PR. Therefore, the anisometric behavior of the MCL may explain variations in maximum MCL force across the knees following TKA.

Our findings also suggest that aligning the femoral component parallel to a computed tomography—based sTEA in measured resection TKA guarantees symmetrical flexion gaps in neither AR nor PR. Specifically, at 90° of flexion, 3 knees (Knees 3, 4, and 5) produced asymmetric gaps with a difference in medial-lateral gap exceeding 2 mm (Fig. 4), despite producing symmetrical medial and lateral gaps at full extension (differing by <1 mm). This finding highlights the importance of identifying targets for femoral rotation that lead to more balanced flexion gaps including those that consider the isometry of the collateral ligaments.

Our study has limitations. First, the virtual environment enabled us to place the femoral component in an ideal, well-controlled manner; in practice, this may be challenging to implement, especially with commonly available rotational alignment guides. Our virtual approach, however, eliminated this source of uncertainty, which allowed us to better isolate the impact of AR and PR on knee balance and soft tissue tensioning. Moreover, the virtual bony cuts and component rotations were supervised and approved by our team of orthopedic surgeons. Second, the ligaments in the TKA models were balanced at full extension using target pretensions of a native knee [11]. In contrast, ligaments in TKA are typically balanced by targeting symmetrical gaps in extension and flexion [26]. Our method, however, yielded clinically acceptable differences in medial and lateral gaps at full extension that were all <1 mm across the 6 knees [27]. Hence, our method of defining ligament properties reflected a balanced knee in extension. Third, we developed models based on the geometries of nonarthritic, neutrally aligned knees, while TKA is typically performed in arthritic knees that could be stiffer. These geometries were used to control for the bone deformity that might be found in arthritic knees to help isolate the impact of AR and PR on knee mechanics. Hence, the ligament forces predicted in this study would represent a lower limit of those obtained clinically. Fourth, the PF joint and the surrounding muscle-tendon units were not included. Since our goal was to simulate a passive, intraoperative examination in an anesthetized patient, the role of the quadriceps muscle-tendon unit is likely mitigated. Fifth, a difference of 2 mm between the medial and lateral gaps in flexion was considered a balanced gap, which contradicts the common notion of creating rectangular gaps in TKA [28]. Our clinical observations, and

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according to data from recent studies, however, suggest that maintaining a difference of 1 to 2 mm in the medial-lateral gaps in flexion is comparable to that measured in native knees [22,29,30]. Sixth, the anterior-posterior dimension of the femoral component increases in 4-mm increments for each whole size in the implant system that we used. This 4-mm increment increased the chances of either notching the anterior cortex or overhanging of the anterior aspect of the femoral component in the PR models. Because our focus was to evaluate tibiofemoral mechanics via measurement of flexion gaps and MCL forces, notching the anterior cortex or overhanging of the anterior aspect of the implant would not affect the conclusion of this study. Interestingly, these discreet size increments act to maximize differences between AR and PR; even so, no consistent biomechanical difference between AR and PR emerged. Finally, we subjected the ankle to 80 N of medial and lateral shear force to simulate varus and valgus moments; this value was the average force obtained from 6 surgeons applying varus loading in a single cadaveric lower limb [21]. Because the gaps depend on the magnitude of the applied force, we repeated the varus and valgus simulation in each knee with the maximum (110 N) and minimum (40 N) loads previously measured in the clinic to test whether the magnitude of the applied load would affect the conclusion of this study (See Appendix 1). We found that each knee remained balanced in full extension with the difference in medial-lateral gaps being <1 mm and the 3 knees that were unbalanced in flexion remained unbalanced. Therefore, variations in the applied load would not impact the conclusions of this study.

In conclusion, neither AR nor PR produced consistent maximum MCL forces or flexion gaps even with idealized component placement and a balanced extension gap. In fact, variations in knee mechanics (flexion gaps and maximum MCL forces) were larger across knees than between AR and PR. This key finding suggests that current methods of bone resection and for placing the femoral component do not predict consistent biomechanical outcomes of the knee. An alternative target for placing the femoral component that considers knee-to-knee variations in MCL isometry after measured resection TKA might help predict MCL forces and flexion gaps.

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Appendix 1

To account for variability in the shear force applied at the ankle in the varus and valgus test across the surgeons and how it may affect the conclusion of this study, we repeated the varus and valgus simulation with different magnitude of forces. Specifically, for all AR and PR models, the varus and valgus test was simulated by applying shear forces equal to 110 N (maximum), 80 N (average), and 40 N (minimum), which were measured in a cadaveric study reported by Meere et al [21]. The distance from the point of load application at the ankle to the center of the knee was assumed to be 0.5 m. The medial and lateral gaps in response to each loading condition were measured. The results showed that, at full extension, all knee models remained balanced under the 3 loading conditions with gaps of <1 mm. Moreover, in flexion, the knees that were unbalanced (Knees 3, 4, and 5) remained unbalanced and the knees that were balanced (Knees 1, 2, and 5) remained balanced for the 3 loading conditions (Fig. A1). Therefore, for the knee models developed in this study, knee balance was independent of the loading magnitude in the varus and valgus test.



Fig. A1. Medial and lateral gaps in response to applying 40, 80, and 110 N of shear force in varus/valgus test at 90° of flexion for the 6 AR knee models. Only 3 knees (Knees 1, 2, and 6) produced differences between the medial and lateral gaps that was <2 mm. The other 3 knees (Knees 3, 4, and 5) produced gap differences >3 mm. PR knee models produced similar pattern for the medial and lateral gaps (data not shown).