CHAPTER 11

Understanding Knee Arthroplasty Kinematics: News You Can Use

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Radiographic imaging and shape-matching techniques have been used since the late 1980s to quantify the motions of knee replacements in vivo. These studies have shown how knee implants move in vivo, how implant design affects knee kinematics, and how different surgical and design factors influence knee mechanics and patient function. In general, knee implants that definitively control the anteroposterior (AP) position of the femur with respect to the tibia achieve greater weightbearing flexion and exhibit kinematics that are more likely to result in better patient function and implant longevity.

THREE-DIMENSIONAL KINEMATICS FROM TWO-DIMENSIONAL IMAGES

By the late 1980s, total knee arthroplasty (TKA) had become a fairly routine procedure for the treatment of severe knee arthritis. A wide variety of implant designs were being used with predictable success and reasonable durability. The focus of designers was shifting from basic knee function and implant fixation to improvement in knee performance and implant longevity. In part, what was needed to continue evolving knee replacements was more precise information on how knee replacements moved once implanted. Unfortunately, the gait laboratories and computed tomography (CT) scanners of the day could not provide accurate three-dimensional kinematic information of knee replacement motion during weightbearing dynamic activities.

In 1988, my surgeon colleague W. Andrew Hodge and I set out to develop a better method for measuring knee arthroplasty kinematics. Having failed to use the gait laboratory motion capture system to accurately measure implant motion, Hodge suggested that we should directly image the joint with x-ray fluoroscopy and develop an image-based measurement technique. I developed a "shape-matching"–based measurement approach that worked well,[1,3] and this technique and evolved forms have been used since to provide a better understanding of knee replacement function.

The details of shape-matching–based motion measurement are beyond the scope of this volume, but the process follows logically: radiographic images are produced when x-rays pass through space and are attenuated by the patient's anatomy before striking a sensitive medium and causing a chemical or electrical reaction. The x-ray beam emanates from a single point in space with rays diverging in all directions to create a perspective projection of the object—in essence a shadow (Fig. 11–1). The location of the x-ray source with respect to the image plane can be measured so that the same projection can be reproduced on a computer. Computer-aided design information is available for knee implant components, and bone surfaces can be reconstructed from CT or magnetic resonance imaging (Fig. 11–2), thus making it a simple process to synthesize on the computer images of implants at any possible position. These synthetic views can be iteratively modified until they match the views obtained from patients. Once matched, the positions and orientations of the models represent the physical position and orientation of the patients' implants that created the radiographic projection.

Many groups the world over have used shape-matching techniques for determining implant motion from single-plane radiographic views and have studied a range of activities, including gait,[12] stair climbing,[13] and deep knee bends.[19] Although details of the methods vary, measurement precision for each moving segment is typically 0.5 to 1.0 mm for implant motions parallel to the image plane and 0.5 to 1.0 degree for rotations. Importantly, this is monocular vision, not stereo or binocular, and all these techniques have much reduced accuracy for determining translation perpendicular to the image plane, where precision is typically 3.0 to 6.0 mm. Propagating these measurement errors to the articular surfaces, one can typically expect measurement uncertainties of greater than 1.2 mm for single observations of condylar contact or separation.

POSITIONAL FINDINGS

Findings from image-based TKA studies can be organized into positional and dynamic observations. Positional observations relate closely to how implant design and surgical alignment influence articular contact and knee function at the extreme ranges of motion.

Knee implants are typically designed to maximize the tibiofemoral contact area with the knee in extension and to accommodate 10 to 15 degrees of hyperextension. Implant wear testing is performed such that the implants reach 0 degrees relative flexion at simulated early stance. Yet neither context takes into account the fact that surgical alignment may place the implants in positions that differ from 0 degrees relative flexion. Femoral components implanted with intramedullary rods or extramedullary

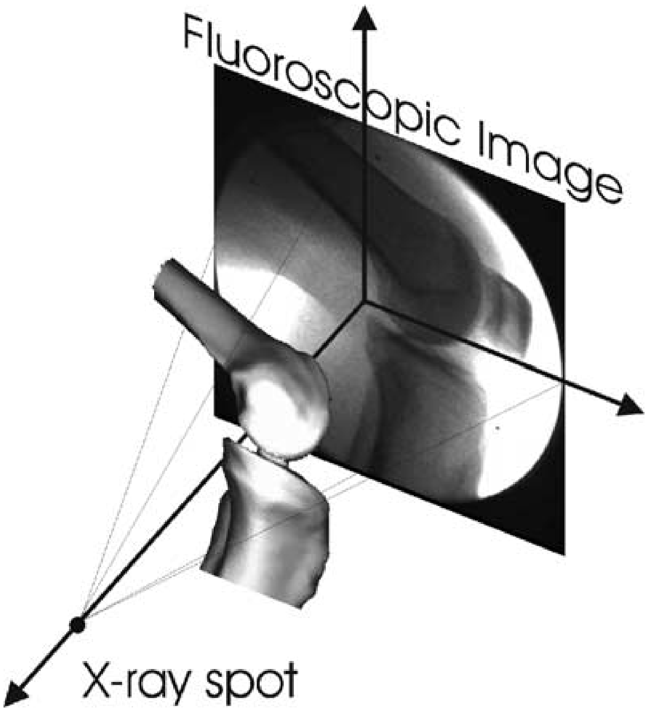


Figure 11–1. Fluoroscopic and radiographic projections are created by a spot source of rays so that the image is a "perspective" projection, or shadow, that is a three-dimensional function of the projection geometry and the position and orientation of the bones. This geometry allows three-dimensional kinematics to be derived from sequences of two-dimensional radiographic images.

techniques are aligned to the distal femur. The anterior bow of the femoral shaft results in the femoral implant component being flexed forward in the sagittal plane by 5 to 7 degrees. Similarly, tibial implant techniques range from alignment perpendicular to the long axis of the tibia to alignment matching the normal posterior slope of the tibial plateau. The net result of typical surgical placement is that the implants are in 5 to 12 degrees of relative hyperextension. Simultaneous measurements of skeletal flexion using goniometry or motion capture (Fig. 11–3) and implant flexion using fluoroscopy have shown an average of 9.5 degrees implant hyperextension when compared with the skeletal flexion angle.[2]

This simple and intuitive observation has at least three important ramifications. First, implants that have hyperextension stops will probably experience much greater contact and possible wear than the designers anticipated.[2,11] Posterior-stabilized designs with tibial posts and some posterior cruciate ligament (PCL)-retaining designs accommodate limited hyperextension, often 5 to 15 degrees. With the implants routinely placed in almost 10 degrees of hyperextension at 0 degrees knee flexion, many of these designs will experience anterior impingement during routine activity. Second, standard evaluations of TKA designs, whether by computer or machine, do not account for implant alignment. The evaluations assume that the straight leg corresponds to 0 degrees of implant flexion. Given that many designs have surfaces with changing curvatures in early flexion, it is possible that these tests will predict performance differing from the clinical experience. Third, implant features designed to guide implant motions at particular flexion angles will engage later in the flexion arc. Post and cam mechanisms in posterior-stabilized knees will engage at approximately 10 degrees greater anatomic flexion than anticipated by the design. In very deep flexion, there is some concern that the proximal "edge" of the femoral condyles (where the articular and bone-cut surfaces meet) will dig into the tibial articular surface. Normal implant alignment means that this phenomenon will occur 10 degrees later in the flexion arc, if at all.

Fluoroscopic evaluations have elucidated the mechanics of TKAs in deeply flexed postures. It has long been assumed that greater posterior femoral translation on the tibia permits greater knee flexion.[29] In a study of 16 different TKA designs in patients with excellent clinical outcomes, there was a significant linear relationship between the amount of posterior femoral translation and maximum weightbearing flexion.[9] This relationship, 1.4 degrees greater flexion for each additional millimeter of posterior femoral translation, held true for all types of TKA design (Fig. 11–4). Implant designs that definitively controlled tibiofemoral position in flexion achieved greater femoral "rollback" and demonstrated greater weightbearing flexion than did designs that required the

Figure 11–2. Three-dimensional measurement of dynamic knee motion using fluoroscopy and shape-matching techniques has been performed for natural knees *(left)*, knees with partial arthroplasty *(middle)*, and knees with total arthroplasty *(right)*. The bone surface models can be created from computed tomographic and magnetic resonance imaging scans, and the implant models are obtained from the manufacturer or three-dimensional laser scans.

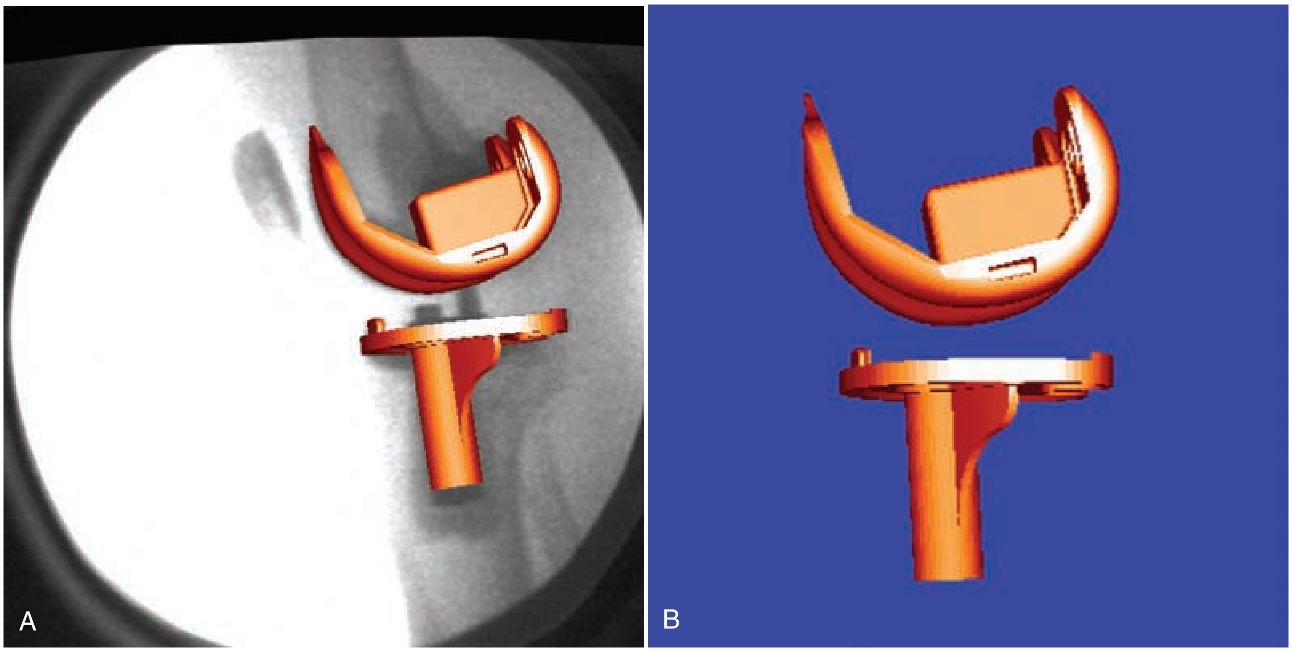


Figure 11–3. Knees with well-aligned implants commonly show implant hyperextension. Anterior bow of the femur and posterior slope of the tibial plateau bias implant alignment by an average of 10 degrees hyperextension. Thus, when the knee is fully extended at toe-off during gait (**A**), the implants are in hyperextension (**B**).

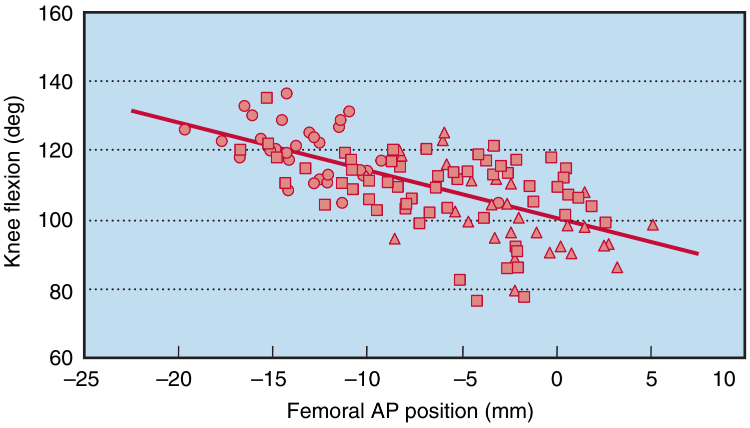
soft tissues and muscles to control tibiofemoral position. These findings suggest that the flexion space, particularly in PCL-retaining TKA, ought not to be made too loose because the additional laxity may allow unwanted anterior translation of the femur and a concomitant decrease in maximum weightbearing flexion.

Similar analyses have demonstrated the importance of posterior condylar geometry on knee flexion range. Bellemans et al[10] showed a significant linear relationship between changes in the posterior condylar offset, or the maximum AP distance from the femoral shaft to the most posterior point on the condyles, and changes in passive range of motion. They found that reducing the posterior condylar offset by 1 mm from its anatomic value decreased passive range of motion by 6 degrees (Fig. 11–5). This finding is particularly relevant for surgeons using anterior referencing instrumentation: when a knee measures in-between component sizes, common practice argues for selecting the smaller component. This will typically reduce the anatomic posterior condylar offset by several millimeters and thus potentially reduce the flexion range by 10 degrees or more! Using the larger femoral component, when possible, or adjusting the position of the smaller femoral component can reduce the effect on the posterior condylar offset and provide the patient with the best possible range of motion.

DYNAMIC CHARACTERISTICS

Early fluoroscopic studies of TKA kinematics demonstrated that dynamic motions could differ markedly from those of a normal knee.[8] These and subsequent studies

Figure 11–4. Maximum weightbearing knee flexion as a function of femoral anteroposterior (AP) position for 121 knees. Femoral posterior positions are negative, anterior is positive, and zero represents the AP midpoint of the tibial component. *Circles* represent posterior-stabilized knees; *asterisks* represent posterior cruciate–retaining, fixed-bearing knees; and *triangles* represent mobile-bearing knees. The *solid line* shows the linear regression with a slope of 1.4 degrees more flexion per millimeter femoral posterior translation (r = 0.64, $p < 0.001$). (From Banks SA, Bellemans J, Nozaki H, et al: Tibio-femoral translation and maximum weight-bearing flexion in fixed and mobile bearing knee arthroplasties. Clin Orthop 410:131-138, 2003.)



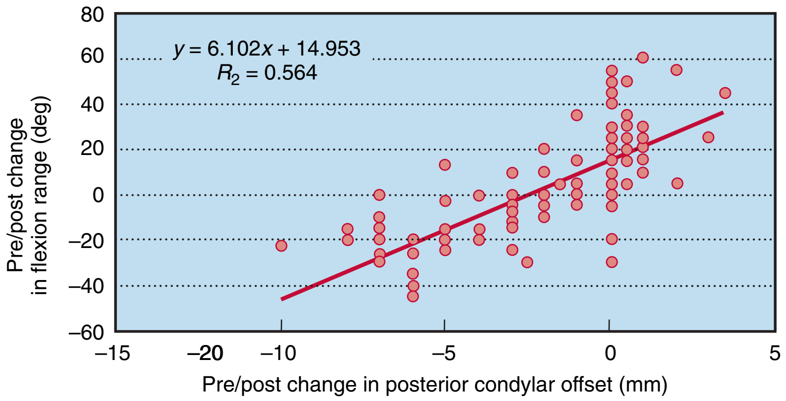


Figure 11–5. Correlation of restoration of posterior condylar offset (postoperative minus preoperative) with postoperative flexion gain (+)/loss (–) for 150 consecutive knees. Overlapping points are not shown. (From Bellemans J, Banks SA, Victor J, et al: Fluoroscopic analysis of deep flexion kinematics in total knee arthroplasty: The influence of posterior condylar offset. J Bone Joint Surg Br 84:50-53, 2002.)

showed that in knees lacking the anterior cruciate ligament and menisci, there is a tendency for the femur to slide forward on the tibia with flexion and backward with extension. However, tibial rotations were normal, with the tibia rotating inward with flexion. A simple method to quantify these translations and rotations is to consider the average center of rotation: in a healthy knee, posterior translation of the femur and internal rotation of the tibia with flexion result in a medial center of rotation. The lateral condyle moves posterior with flexion about a relatively stationary medial condylar position. In unconstrained TKA, the medial condyle is observed to slide forward with flexion about a relatively stationary lateral condylar position. Thus, a lateral center of rotation has been observed in unconstrained TKA designs. An analysis of stair-climbing motions in 25 different TKA designs showed a significant relationship between the intrinsic constraints of the implant and the average center of rotation (Fig. 11–6): designs with greater intrinsic control had central or medial centers of rotation, whereas up to 86% of unconstrained devices had lateral centers of rotation.[4] This analysis included only patients with high satisfaction and excellent clinical scores and showed that a wide range of knee motion patterns are compatible with good clinical results. Hence, implant designers and surgeons may have wide latitude to modulate knee motion patterns to achieve further improvement in patient strength, range of motion, and implant longevity.

Fluoroscopic studies of TKA kinematics comparing different activities have shown that TKA motions can vary dramatically depending on the activity and implant design.[5] Many implant designs have articular surfaces with varying curvatures or mechanisms that engage at different parts of the flexion arc. Intuitively, these changing constraints might result in different kinematic patterns depending on the flexion range of the activity. Conversely,

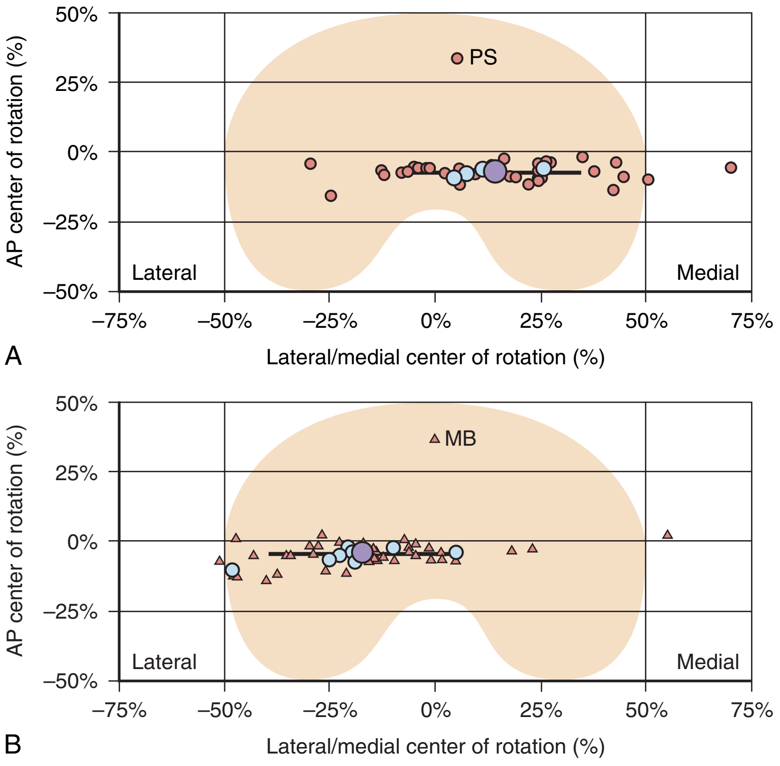


Figure 11–6. The average center of rotation is strongly influenced by the intrinsic constraints of the implant design for stair step activities. Posterior-stabilized knees, which force the femur posterior on the tibia with flexion, mostly show medial centers of rotation (**A**). Gait-congruent, mobile-bearing knee designs allow relatively free anteroposterior translation of the femur in flexion, with most knees showing a lateral center of rotation (**B**). Each red circle and red triangle represents the center of rotation in one knee; the blue circles represent the average center of rotation for a specific implant design, and the large purple circle represents the average center of rotation for all knees of that type. (From Banks SA, Hodge WA: Implant design affects knee arthroplasty kinematics during stair-stepping. Clin Orthop 426:187-193, 2004.)

implant designs with consistent intrinsic constraint over the flexion arc might be expected to show similar motion patterns across the range of activities. Comparison of TKA motions during the stance phase of gait and during stair activities confirms these concepts (Fig. 11–7). For example, posterior-stabilized, fixed-bearing TKA designs consistently showed more medial centers of rotation during stair activities than during gait (Fig. 11–7A). During stair climbing, the post-cam mechanism controls motion and forces posterior femoral translation with flexion. During gait, the post-cam mechanism is not engaged and the femur tends to slide posterior with *extension*, more so on the medial side. The opposite situation is observed in rotating platform, mobile-bearing knees with gait-congruent articulations (Fig. 11–7B). During gait, the tibiofemoral articulation is fully conforming and allows only axial rotation with flexion/extension. Stair climbing flexes the knee beyond the range of tibiofemoral congruency, and the femur slides forward on the tibia with flexion, more on the medial side. Implants with condyles that have the same sagittal radius from 0 to 75 degrees flexion and correspondingly consistent tibiofemoral constraint did exhibit similar motion for the gait and stair activities (Fig. 11–7C).

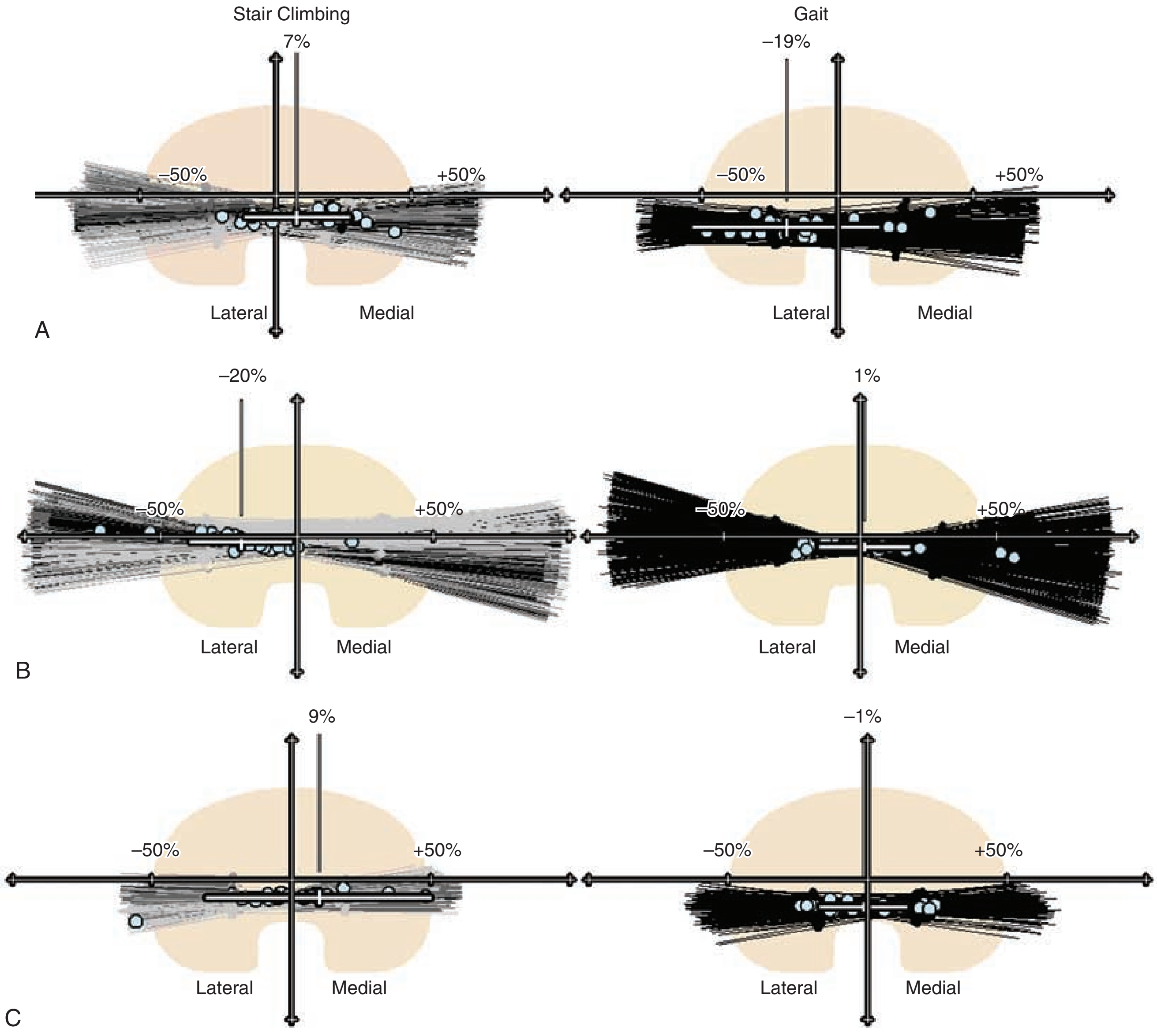


Figure 11–7. Patterns of knee motion vary with activity and implant design. **A**, Sagittally unconstrained, posterior-stabilized knees show a medial center of rotation during stair-climbing activities *(left)*, but greater femoral sliding and a lateral center of rotation during the stance phase of gait. **B**, Gait-congruent, rotating platform mobile-bearing knees show anterior sliding of the femur in flexion during stair activities and a lateral center of rotation, but they are constrained to pure internal/external tibial rotation with flexion during the stance phase of gait *(right)*. **C**, Knees that maintain similar conformity over the flexion range show more similar knee motion patterns for the gait and stair activities. Each *line* represents the location and orientation of the femoral condyles with respect to the tibial plateau for all frames of data for a group of subjects. The *gray dots* indicate the average center of rotation for a single motion trial, and the *white cross* indicates the mean and standard deviation for the group average center of rotation.

OTHER APPLICATIONS FOR IN VIVO DATA

In vivo kinematic data are unquestionably useful for understanding the interplay of implant design and surgical factors in TKA performance. In addition, these data provide important corroboration or guidance for other studies. For example, interpretation of wear patterns on retrieved tibial inserts is greatly enhanced with knowledge of that particular implant's in vivo kinematics.[18] In vivo kinematics can be used to implement increasingly realistic and more powerfully predictive mechanical wear tests. With advanced computer codes, it is now possible to input in vivo kinematics and make reasonably accurate predictions of that implant's wear performance over its service life.[14]

CONCLUSION

Fluoroscopy has provided a unique window for direct observation and measurement of dynamic knee replacement motions. Shape-matching–based measurements are a powerful tool for accurately quantifying knee motions, and they provide informative characterization of implant design and surgical factors that influence patient outcomes. In addition to providing an enhanced understanding of implant design and surgical issues, these in vivo data are a useful complement to retrieval studies, gait laboratory analyses, and computer simulations.

NEWS YOU CAN USE

1. *Surgical technique matters.* Surgical balancing of the ligaments and soft tissues can have a significant effect on weightbearing knee kinematics, particularly in unconstrained arthroplasty designs.[23] When using unconstrained tibiofemoral articulations, the surgeon has the opportunity to restore near-physiological knee motion. However, many kinematic studies of unconstrained devices demonstrate abnormal knee motions wherein the femur slides forward with flexion instead of moving backward with flexion.[2,6,7,10,25-28]
2. *Axial rotation range is maintained in healthy, reconstructed, unicondylar, and total knee arthroplasties of many types.* The total amount of tibial internal/external rotation appears to be maintained after ligament reconstruction or arthroplasty. However, the specific pattern of rotation can change with reconstruction. Knees generally exhibit tibial internal rotation with early flexion, approximately 5 degrees in the stance phase of gait and 8 degrees during stair climbing.[5] These ranges of rotation are similar for healthy and reconstructed knees.[20] At a minimum, new surgical techniques or knee replacements ought to provide for these ranges of axial rotation.
3. *Axial rotation is activity dependent, not obligatory, with total knee replacements in deep flexion postures.* With TKA, early tibial rotation with flexion does not necessarily continue into deeply flexed postures, and achieving high flexion is not correlated to the amount of axial rotation. Different high-flexion postures, such as squatting, kneeling, or lunging, show rotations more closely related to the body position than to the degree of flexion achieved.[9,19] This situation may differ considerably from an intact normal knee.
4. *Restoring normal posterior condylar geometry permits greater maximum knee flexion.* Normal posterior condylar geometry elevates the femoral cortex with respect to the tibia to allow deep flexion without impingement. When this posterior condylar offset is diminished during arthroplasty, posterior impingement can occur earlier in the flexion arc and block full flexion potential.[10] Reduced condylar offset can also provide greater laxity in flexion and allow the femur to translate forward in flexion—likewise reducing flexion potential.
5. *A posterior femoral position enhances maximum weightbearing flexion.* Knee arthroplasties that achieve a relatively posterior position of the femur on the tibia exhibit greater maximum weightbearing flexion.[9] Arthroplasty designs with definitive control of tibiofemoral translation, through either conforming tibial surfaces or PCL substitution, maintain a more posterior position of the femur on the tibia in flexion. For patients desiring lifestyles involving deeply flexed postures, the surgeon might consider arthroplasty designs or modular tibial inserts that provide enhanced control of the tibiofemoral position in flexion.
6. *With normal surgical alignment, implant components are biased toward hyperextension.* Anterior bowing of the femur and posterior slope of the tibial plateau cause arthroplasty components to have approximately 10 degrees of hyperextension with neutral surgical alignment.[2] This often places components close to the design limit for allowable hyperextension and thus accounts for the very common observation of anterior impingement damage on many types of retrieved knee arthroplasty components.[2,11,15,16,18,21,22,24] Posterior translation of the femur with knee extension exacerbates anterior impingement. Anterior impingement damage is commonly observed in retrieved posterior-stabilized knees with post/cam stabilization. Surgeons using this type of arthroplasty should know and take into account the allowable hyperextension range during component alignment.
7. *The center of rotation in the knee can be changed by varying articular constraints.* Kinematic studies of clinically successful knee arthroplasties during a variety of activities demonstrate a wide range of motion patterns. Unconstrained arthroplasties with either translating, mobile-bearing or flat, fixed-bearing tibial inserts show a lateral center of rotation wherein the medial condyle moves forward with flexion. This center of rotation, or the AP movement of the femur, can be modulated by changing

tibiofemoral conformity or the AP constraints in the knee.[5,17,27,28] By knowing how tibiofemoral motions are modulated throughout the functional range of motion, one can develop better expectations for patient and device performance.

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