Knee Kinematics in Medial Osteoarthritis during In Vivo Weight-Bearing Activities

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ABSTRACT: Dynamic knee kinematics were analyzed for medial osteoarthritic (OA) knees in three activities, including two types of maximum knee flexion. Continuous x-ray images of kneeling, squatting, and stair climbing motions were taken using a large flat panel detector. CT-derived bone models were used for the model registration-based 3D kinematic measurements. Three-dimensional joint kinematics and contact locations were determined using two methods: bone-fixed coordinate systems and by interrogation of CT-based bone model surfaces. The femur exhibited gradual external rotation with knee flexion for kneeling and squatting activities, and gradual internal rotation with knee extension for stair climbing. From 100° to 120° flexion, contact locations showed a medial pivot pattern similar to normal knees. However, knees with medial OA displayed a femoral internal rotation bias and less posterior translation when compared with normal knees. A classic screw-home movement was not observed in OA knees near extension. Decreased variability with both activities and methods of calculation were demonstrated for all three activities. In conclusion, the weight-bearing kinematics of patients with medial OA differs from normal knees. Pathological changes of the articulating surfaces and the ligaments correspond to observed abnormalities in knee kinematics. © 2009 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 27:1555–1561, 2009

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For about 15 years, 3D-to-2D model registration techniques have been used to determine knee arthroplasty motions from fluoroscopic image sequences.[1,2] Kinematic information is now available while subjects with multiple designs of total knee arthroplasty (TKA) perform functional activities.[3–12] These single-plane fluoroscopic techniques have recently been applied for motion measurement in joints without implants, where 3D surface models of the bones are created from computed tomography (CT), and magnetic resonance (MR) imaging.[13–22] Komistek et al. reported kinematic measurements using single-plane fluoroscopic projections and CT-derived bone models.[15] Moro-oka et al. recently assessed the accuracy of their shape matching technique using bone models created from CT and MR, and analyzed 3D dynamic knee kinematics in kneeling, squatting, and stair climbing.[18,19] Single-plane x-ray imaging and model-based shape matching appear to provide kinematic measurements with sufficient certainty to assess normal and pathological knee motions using either CT- or MR-derived models.

Three-dimensional kinematic studies have focused on patients with TKA,[1–12] normal knees,[13–19,23] or anterior cruciate ligament (ACL) deficient knees.[20–22] To our knowledge, no studies have employed 3D-to-2D model registration techniques to study knees with osteoarthritis (OA). This information is critical to achieve a better understanding of how OA affects knee mechanics, and the state of the knee prior to TKA. Kinematics of OA knees have been evaluated with surgical navigation systems, radiostereometry, and MRI,[24–26] but these techniques have not been used to study weight-bearing, active, large range of motion activities. In particular, pre-TKA knee mechanics in deeply flexed postures has become more important, as patients expect to perform these activities post-TKA.

Our purpose was to analyze kinematics of knees with medial OA. We sought to answer three questions: (1) Do kinematics of knees with medial OA differ from healthy knees during weight-bearing activities? (2) Do kinematics of knees with medial OA vary with activity? (3) Does the method for quantifying femoral rotation differ between the bone-embedded coordinate system and the contact location from CT-derived bone models?

METHODS

From December 2004 to September 2005, 12 subjects were recruited before undergoing unilateral primary TKA for treatment of severe medial compartment knee OA. Subjects were defined according to the American College of Rheumatology criteria for the diagnosis of knee OA: medial knee pain, radiographic osteophytes at the medial joint space, and at least one of the following: age >50 years, morning stiffness <31 min in duration, or crepitus on motion.[27] Inclusion criteria were: the ability to perform at least one of three activities: kneeling, squatting, and stair climbing; and consent to participate in the study (as approved by the institutional review board). The subjects included 11 women and 1 man (age 74 ± 8 years, range, 62–86; height 151 ± 8 cm, range, 136–167; and weight 60 ± 13 kg, range, 43–81; Table 1). One knee had previous arthroscopic partial medial menisectomy. The radiographic level of OA was classified using the Kellgren–Lawrence System[28] and averaged 3.9 ± 0.3 (1 knee had grade 3, 11 had

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Table 1. Data for All 12 Subjects

Patient	Age/ Sex	Previous Surgery	KL Grade	Range of Motion (Extension/Flexion)	KSS (KS/FS)	WBR (%)	ACL/PCL Status
1	84/F	–	IV	–10/125	66/45	–13	Ruptured/attenuated
2	77/F	–	IV	–15/115	60/60	11	Attenuated/attenuated
3	64/F	+	III	0/115	64/45	64	Intact/intact
4	62/F	–	IV	–10/120	55/65	12	Attenuated/intact
5	71/F	–	IV	0/140	70/70	14	Intact/intact
6	86/M	–	IV	0/130	70/65	–26	Attenuated/attenuated
7	73/F	–	IV	–35/120	47/65	15	Attenuated/attenuated
8	72/F	–	IV	–10/90	59/45	–2	Ruptured/intact
9	84/F	–	IV	–20/130	57/45	–13	Ruptured/attenuated
10	62/F	–	IV	–10/115	54/65	26	Intact/intact
11	73/F	–	IV	–10/100	37/55	–6	Intact/intact
12	77/F	–	IV	–20/115	59/60	0	Attenuated/attenuated

KL grade, Kellgren–Lawrence grade; KSS, Knee Society score; KS, knee score; FS, function score; WBR, weight-bearing ratio; ACL, anterior cruciate ligament; PCL, posterior cruciate ligament.

grade 4). The average passive range of motion (extension/flexion) was $-12^\circ \pm 10^\circ$ (range, -35°–0°)/$118^\circ \pm 13^\circ$ (range, 90°–140°). The average Knee Society knee score/function score[29] was 58 ± 9 (range, 37–70)/56 ± 9 (range, 45–70). Limb alignment in the coronal plane was measured by drawing a mechanical axis, from the center of the femoral head to the center of the ankle, on each limb depicted in a full-standing radiograph. The weight-bearing ratio as a percentage was calculated by measuring the distance from the medial edge of the proximal tibia to the point where the mechanical axis intersects the proximal tibia, then dividing that measurement by the width of the proximal tibia[30] and multiplying by 100. The average ratio was $7\% \pm 23\%$ (range, -26%–64%). The integrity of the ACL and posterior cruciate ligament (PCL) were assessed during surgery. Both were intact in four knees, both were attenuated or ruptured in six knees, and the ACL was attenuated or ruptured, but the PCL was intact in two knees. In all knees, cartilage destruction in the medial compartment on the femur and tibia was severe with exposed subchondral bone, and mildly or severely degenerated medial menisci. The lateral compartment showed low- or mild-grade degeneration with cartilage fibrillation.

Bone models of the femur and tibia/fibula were created from CT (Toshiba, Aquilion, Tochigi, Japan) scans with a 512×512 image matrix, a 0.35×0.35 pixel dim, and a 1-mm thickness spanning about 150 mm above and below the joint line, and 2-mm slices through the centers of the hip and ankle joints. Cortical bone edges were segmented using commercial software (SliceOmatic, Tomovision, Montreal, CA), and these point clouds were converted into polygonal surface models (Studio, Raindrop Geomagic, Research Triangle Park, NC). Anatomic coordinate systems were embedded in each bone following published conventions.[18,19] The mediolateral (Z) axis of the femur and tibia/fibula was defined by fitting a cylinder to the posterior femoral condyle.[19] The mid-point of the cylindrical axis was defined as the coordinate system origin for both femur and tibia.

Continuous sagittal x-ray images of kneeling, squatting, and stair climbing were taken using a flat panel detector (Hitachi, Clavis, Tokyo, Japan): 3 frames/s, image area 397×298 mm, 0.20×0.20 mm/pixel resolution (Fig. 1). For kneeling, subjects started in about 90° flexion and applied their body weight to achieve maximum knee flexion. For squatting, subjects were asked to squat bilaterally from a fully extended standing position to maximum knee flexion. For stair climbing, subjects ascended a two-step staircase with knee motions from flexed to extended positions recorded on the first step. A total of 496 images were used for the analysis: 163 images for kneeling (10 knees), 136 images for squatting (8 knees), and 197 images for stair climbing (12 knees).

The 3D position and orientation of the tibia/fibula and femur were determined using previously reported shape matching techniques[1,17–19] (Fig. 2). A region of the flat panel x-ray image was extracted and scaled to 512×512 square pixels for 3D shape registration. The CT models were projected onto the images and manually aligned with the bone projections. An automated matching algorithm using nonlinear least squares optimization and an image edge-to-model edge distance criteria was used to refine registration. RMS errors for this method were 0.53 mm for in-plane translation, 1.6 mm for out-of plane translation, and 0.54° for rotations in a preliminary study.[18]

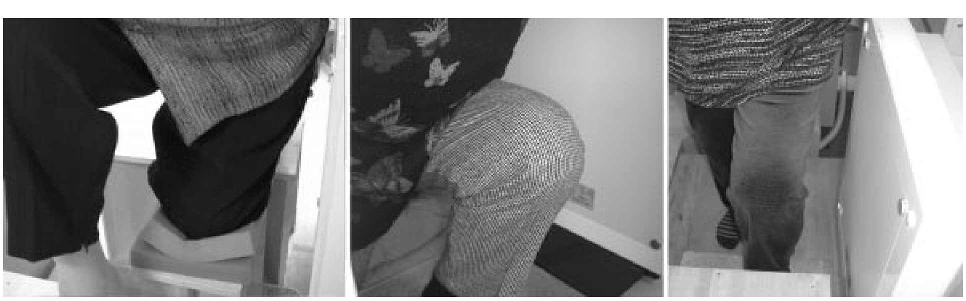

Figure 1. Subjects performed kneeling (left), squatting (middle), and stair climbing (right) while their knee motions were observed using a large flat-panel x-ray image detector. The images show a left knee. Kneeling was performed from about 90° to maximum knee flexion. Squatting was performed from a fully extended standing position to maximum flexion. Stair climbing captured the ascent phase from about 80° to 20° flexion.

Figure 2. 3D-to-2D CT model-to-flat panel image registration was used to determine the kinematics of medial OA knees during three weight-bearing activities.

Joint kinematics were determined from the 3D position of each bone model using Cardan angles.[31] Femoral external tibial rotation was analyzed as a function of flexion angle. Spline interpolation with 5° flexion increments was used to create average kinematics for the group. The "screw-home" movement was defined as a sharp femoral external rotation from full extension to 15° flexion and internal rotation from 15° flexion to full extension.[19,32–34] Stair climbing proceeded from flexion to extension, while kneeling and squatting proceeded from extension to flexion. Some subjects did not participate in kneeling (two knees) and squatting (four knees) because of limited range of motion and knee discomfort. Each subject began and completed each activity at different flexion angles, so data representing fewer than three knees were eliminated from group averages. The average maximum extension/flexion angle was $92 \pm 8°$ (range, 83–103)/$121 \pm 13°$ (range, 106–143) during kneeling, $22 \pm 16°$ (range, 3–53)/$93 \pm 19°$ (range, 72–132) during squatting, and $17 \pm 19°$ (range, 2–74)/$82 \pm 11°$ (range, 65–106) during stair climbing using the model registration-based 3D kinematic measurement technique. Repeated measures ANOVA and post hoc tests (Tukey–Kramer) ($p < 0.05$) were used to compare means for different activities and measurement methods.

The surface of each CT-derived tibial model was divided into medial and lateral compartments and represented as a point cloud. For every image, a surface separation map was created by computing the minimum distance between each point on the tibial surface and all points on the femoral surface.[19] Lateral femoral condylar contact locations were computed as the geometric centroid of the region having <6 mm separation, which acknowledges uncertainty about cartilage thickness,[35,36] cartilage deformation, and measurement errors.[19] Medial femoral condylar contact locations were computed as the geometric centroid having <2 mm separation, which was based on the intraoperative confirmation of eburnated bone in the medial compartment. The angle between the tibial transverse axis and the line connecting the contact centroids represented the rotational kinematics of femoral condylar contact.

RESULTS

The femur rotated externally from an extended to a flexed position in squatting and kneeling, and rotated internally from a flexed to an extended position in stair climbing (Fig. 3A). Based on the bone-embedded coordinate systems, femoral external rotation in minimum and maximum flexion (20° and 100°, respectively) during squatting averaged $4° \pm 2°$ (range, 1°–5°) and

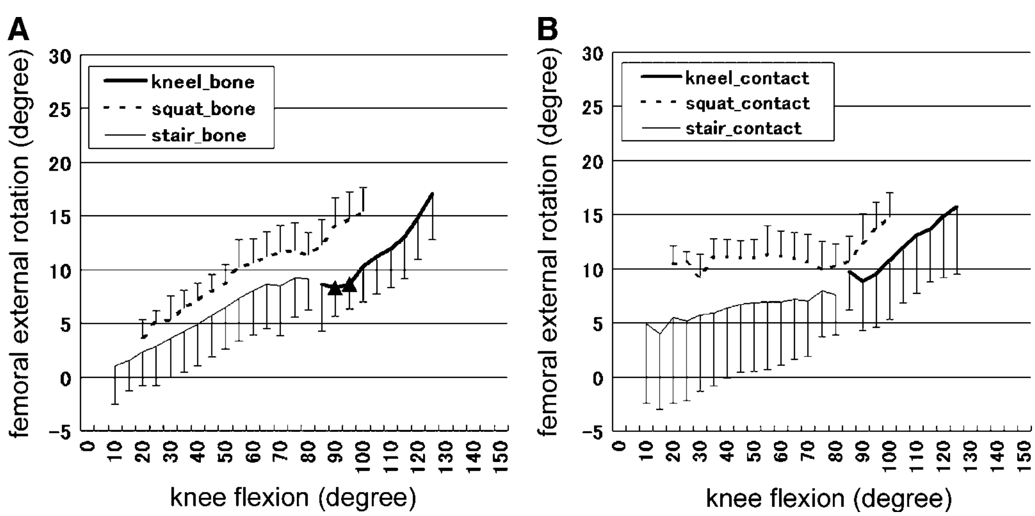

Figure 3. (A) Femoral external rotations based on bone-embedded coordinate systems. Femoral external rotation increased with flexion for squatting and kneeling, and decreased with extension for stair climbing. No significant differences occurred between rotations for squatting and stair climbing ($p = 0.05$). Solid triangles on kneeling represent significant differences from the corresponding rotation during squatting ($p < 0.05$). (B) Femoral external rotations based on contact location determined from CT-derived bone models. The femur showed net external rotation with flexion for squatting and kneeling, and net internal rotation with extension for stair climbing. No significant differences occurred between rotations for squatting and stair climbing ($p = 0.08$), and for squatting and kneeling ($p = 0.26$). No significant differences occurred between rotations determined by the CT model-based contact locations and the rigid body bone motions in graph (A) (squatting: $p = 0.23$, stair climbing: $p = 0.79$, kneeling: $p = 0.86$).

$15^\circ \pm 2^\circ$ (range, 13°–18°). External rotation during squatting was greater than stair climbing ($p > 0.05$) and kneeling ($p < 0.01$) at all flexion angles, with significant pair-wise differences between squatting and kneeling at 90° and 95° ($p < 0.05$). Similar absolute values of femoral axial rotation were observed in squatting from 15° to 80° flexion and stair climbing from 80° to 15° flexion ($p = 0.50$). A sharp external rotation with flexion or internal rotation with extension, the "screw-home movement," was not seen in either squatting or stair climbing because the subjects' arthritic knees did not reach full extension.

Femoral external rotation based on contact location determined from CT-derived bone models also increased from an extended position to a flexed position for squatting and kneeling activities, and decreased from a flexed position to an extended position for stair climbing activity (Fig. 3B). Femoral external rotation in minimum and maximum flexion (20° and 100°, respectively) during squatting averaged $10^\circ \pm 8^\circ$ (range, 2°–20°) and $15^\circ \pm 2^\circ$ (range, 13°–17°). External rotation during squatting was larger than stair climbing and kneeling at all flexion angles, but no significant differences were found between rotations determined from contact points for squatting and stair climbing ($p = 0.08$), or for squatting and kneeling ($p = 0.26$). The magnitude of external rotation from contact points was half as much as from rigid body bone kinematics in all activities (Fig. 3), but these differences were not significant (squatting: $p = 0.23$; stair climbing: $p = 0.79$; kneeling: $p = 0.86$).

Average AP translations of the femoral condylar contact points were smaller than previously reported for healthy knees (Fig. 4). For stair climbing, average translation was <1 mm from 80° to 20° flexion. From 20° to 10°, contact translated several millimeters anteriorly with little femoral internal rotation. For squatting, the femoral condylar contact translated <1 mm from 20° to 80°. From 80° to 100°, the contact points rotated externally with 4 mm posterior translation of the lateral compartment. No significant differences in contact locations occurred between squatting and stair climbing (medial: $p = 0.37$; lateral: $p = 0.32$). For kneeling, the contact points rotated externally with posterior translation of the lateral compartment from 100° to 120°. Significant differences in lateral compartment contact locations were found between squatting and kneeling at 90°–100° flexion ($p < 0.01$).

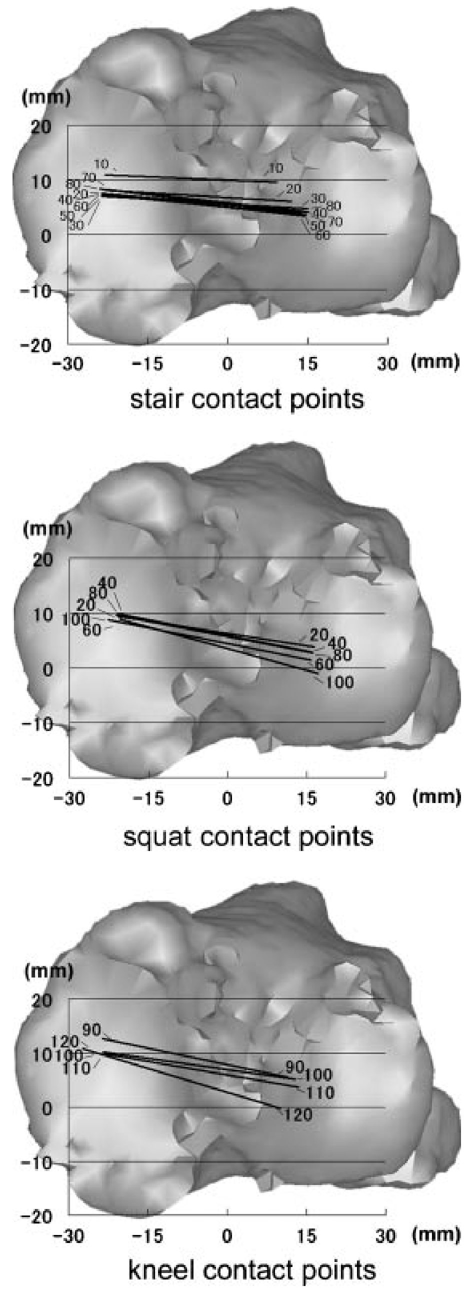

Figure 4. Average contact locations superimposed on a right tibial surface model of a representative knee with medial OA. Average contact locations are shown for 10° flexion increments for stair climbing and kneeling, and 20° flexion increments for squatting. Knees with medial OA revealed small kinematic differences in all activities. For squatting, the contact positions showed little posterior femoral translation with flexion from 20° to 80°. For stair climbing, the contact positions showed little anterior femoral translation with extension from 80° to 20°. For squatting from 80° to 100° and kneeling from 80° to 120°, the contact position in the lateral compartment showed posterior translation with flexion.

DISCUSSION

We examined dynamic 3D kinematics of OA knees in three activities, including weight-bearing and passive maximum knee flexion, using a high resolution flat panel x-ray detector. Previous studies of healthy knees examined kinematics during these same activities, so our primary goal was to determine how severe medial OA affected kinematics. We believe this immediate preoperative assessment of knee function will be useful to understand post-TKA knee function.

Normal knees typically show femoral external rotation relative to the tibia with flexion. In our previous study with normal knees, tibiofemoral contact locations exhibited three phases: sharp femoral external rotation from 0° to 15° flexion (similar to the "screw home" movement), relatively constant rotation from 20° to 80°, and increasing rotation from 80° to 150°.[19] In knees with medial OA, femoral external rotation gradually increased with flexion for squatting and kneeling, and gradually decreased with extension for stair climbing. During squatting from 20° to 100°, similar rotations of the bone-embedded coordinate systems were observed in normal (12°)[19] and OA knees (11°). From 100° to 120° flexion, contact locations showed a medial pivot pattern similar to normal knees. However, knees with medial OA differed from normal knees. First, they displayed a femoral internal rotation bias of about 8°compared to normal knees: femoral external rotations of 4° and 15° during squatting (20° and 100° flexion, respectively) were less in knees with medial OA than the 12° and 24° observed in normal knees.[19] Second, the natural screw-home movement was not observed in knees with medial OA, perhaps because they did not reach full extension in squatting or stair climbing. Finally, femoral condylar contact in knees with medial OA did not exhibit significant AP translation between 30° and 80° for either squatting or stair climbing.[19]

Femoral internal rotation bias in patients with medial OA has been reported. Matsui et al. evaluated the femorotibial rotation using CT, and reported that the tibia tended to locate in an externally rotated position in knees with severe OA.[37] In OA knees, a femoral internal rotation bias may be caused by rotational deformity and/or insufficiency of the ACL. Siston et al. measured in vivo intraoperative passive kinematics using a surgical navigation system[25] and noted that OA knees displayed decreased tibial internal rotation between 10° and 90°compared to normal knees. Saari et al. used dynamic RSA to study knees with medial OA and found decreased tibial internal rotation between 50° and 20°, corresponding to less posterior displacement of the lateral femoral flexion facet center.[24] We also observed less posterior displacement of the lateral femoral contact point. Nagao et al. studied the rotational angle during weight-bearing in OA knees by means of ultrasound and found that tibial external rotation at maximum extension and screw-home rotation proportionately decreased with progression of medial OA.[38] We did not observe screw-home type knee kinematics, likely because our subjects with severe OA did not reach full extension. Osteophytes in the femoral intercondylar notch, and anterior tibial articulation and ligament stiffness in extension, might affect extension and the screw-home mechanism.

A consensus about AP motion in knees with medial OA has not been reached. Siston et al. reported the AP motion of the femur in knees with OA was not different from normal knees.[25] We observed patients with medial OA had less posterior femoral translation with flexion than normal knees. Siston et al. studied intraoperative passive kinematics, whereas we observed dynamic volitional motions. Scarvell et al. used MRI and reported anterior contact patterns were associated with OA severity.[26] In our study, knees with severe OA showed little AP translation between 30° and 80° during squatting and stair climbing. The medial femoral condyle was in a similar position to that reported in healthy knees, while the lateral femoral condyle remained more anterior than in normal knees during all three activities.[19] At higher flexion in kneeling (>100°), we observed additional femoral external rotation and posterior lateral translation, essentially pivoting about the medial compartment, as observed in normal knees.[19] These knees had only modest arthritic changes in the lateral compartment, and 50% had a functioning PCL, so one could reasonably expect a different kinematic result with severe tricompartmental disease or an absent PCL.

In OA knees, reduced posterior translation during dynamic weight-bearing flexion may be caused by increased collateral stiffness, other soft tissue contractures, and osteophyte formation with cartilage-bone erosion in the medial compartment,[39,40] and may be related to the cartilage wear pattern on the tibial plateau. Moschella et al. reported that wear patterns on OA knees with intact ACLs occurred in the central region of the medial plateau,[41] corresponding to the medial condylar contact patterns in the current and previous kinematic studies.[23,24,26] Conversely, morphologic studies showed that knees with deficient ACLs had significantly larger posterior wear patterns in the medial tibial platea,[42,43] consistent with our kinematic findings in OA knees. In our study, the medial femoral condyle of the three OA knees with ruptured ACLs showed more posterior contact during stair climbing compared to the nine OA knees with intact/attenuated ACLs. OA knees with ruptured ACLs also showed less femoral external rotation of the bone-embedded coordinate systems during stair climbing than OA knees with intact/attenuated ACLs. However, further investigations with sufficient sample sizes are needed to compare kinematics in OA knees with different ligament states.

Normal knee and TKA kinematics vary significantly with activity.[6,19] Knees with medial OA showed greater femoral external rotation during squatting than other activities, similar to normal knees.[19] The wider stance width used during squatting likely biased the knee toward greater femoral external rotation in mid-range flexion. However, OA knees revealed smaller kinematic differences among activities than did normal knees. No differences between squatting and stair climbing were observed, and differences between squatting and kneeling were significant only from 90° to 95° flexion, despite contrasting passive and weight-bearing activities. Severe medial OA appears to limit significantly the range of knee motions during squatting, stair climbing, and kneeling.

Previous kinematic analyses of TKA knees showed some similarity to knees with medial OA. Dennis et al. reported that rotation magnitudes and the percentage of knees having normal axial rotation patterns decreased in TKA during a deep knee bend.[7] Knees with medial OA showed biased femoral internal rotation compared to normal knees for squatting and stair climbing. Physiologic femoral roll-back is rarely seen in PCL retaining TKA during single-leg deep knee bend and stair activities.[8–10] Knees with medial OA did not show physiological roll-back during squatting nor physiological roll-forward during stair climbing. Komistek et al. showed that tibiofemoral contact shifted anteriorly in PCL retaining TKAs during kneeling compared to a seated position.[11] Knees with medial OA showed more anterior tibiofemoral contact locations in kneeling than squatting from 90° to 100°. These observations suggest that preoperative knee kinematics might play a larger role in TKA kinematics than has been appreciated.

Iwaki et al. showed that the sagittal profile of the medial femoral condyle is composed of a larger distal and smaller posterior circular arc, while the lateral femoral condyle is well described by a single circular arc.[44] These asymmetric femoral geometries, and a distinct upward slope at the anterior tibial plateau, lead to differences in axial rotation measured from tibiofemoral contact locations or bone-embedded coordinate systems in early flexion (0°–30°).[19] These differences in axial rotation according to calculation method were not observed in knees with medial OA. Lack of full knee extension and pathological changes to the articulating surface geometries likely explain this difference from normal knees.

This study has several limitations. First, the approach to analyzing joint contact ignores the degenerated menisci, which are invisible on x-ray, but obviously affect joint contact and load distribution, especially in the lateral compartment. Second, determination of tibiofemoral contact conditions from CT-derived models requires an assumption of uniform cartilage thickness. We chose different separation thresholds for the severely arthritic medial compartment (2 mm) and modestly arthritic lateral compartment (6 mm) to better represent the contacting surfaces. We believe this leads to reasonable estimates of the centers of articular contact.[19] The optimal approach might be to perform model registration using CT-derived bone models with MR-derived cartilage surfaces, but this would impose a bigger burden on preoperative subjects. Third, single-plane imaging has much higher measurement uncertainties for out-of-plane (mediolateral) translations, so we do not report these small translations that can be measured well using bi-plane techniques.[22] Importantly, none of these limitations or uncertainties would bias our measures to show less knee motion.

In summary, we provided observations of knee kinematics in medial OA during three activities using dynamic imaging and model-image registration with CT-derived models. Weight-bearing kinematics in OA knees differ from normal knee kinematics. Pathological changes due to medial OA affect kinematics, and these changes may be predictive of post-TKA knee kinematics. Currently, we are following these patients to assess their post-TKA knee function at several postoperative intervals.

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