Knee Kinematics of Healthy Adults Measured Using Biplane Radiography

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A dataset of knee kinematics in healthy, uninjured adults is needed to serve as a reference for comparison when evaluating the effects of injury, surgery, rehabilitation, and age. Most currently available datasets that characterize healthy knee kinematics were developed using conventional motion analysis, known to suffer from skin motion artifact. More accurate kinematics, obtained from bone pins or biplane radiography, have been reported for datasets ranging in size from 5 to 15 knees. The aim of this study was to characterize tibiofemoral kinematics and its variability in a larger sample of healthy adults. Thirty-nine knees were imaged using biplane radiography at 100 images/s during multiple trials of treadmill walking. Multiple gait trials were captured to measure stance and swing-phase knee kinematics. Six degrees-of-freedom kinematics were determined using a validated volumetric model-based tracking process. A bootstrapping technique was used to define average and 90% prediction bands for the kinematics. The average ROM during gait was 7.0 mm, 3.2 mm, and 2.9 mm in anterior/posterior (AP), medial/lateral (ML), and proximal/distal (PD) directions, and 67.3 deg, 11.5 deg, and 3.7 deg in flexion/extension (FE), internal/external (IE), and abduction/adduction (AbAd). Continuous kinematics demonstrated large interknee variability, with 90% prediction bands spanning approximately ± 4 mm, ± 10 mm, and ± 5 mm for ML, AP, and PD translations and ± 15 deg, ± 10 deg, and ± 6 deg in FE, IE, and AbAd. This dataset suggests substantial variability exists in healthy knee kinematics. This study provides a normative database for evaluating knee kinematics in patients who receive conservative or surgical treatment. [DOI: 10.1115/1.4047419]

Introduction

A database of knee kinematics during gait in healthy, uninjured individuals can serve as a reference for comparison when evaluating the effects of injury, surgical repair, rehabilitation, and age on knee kinematics. However, accurate measurement of tibiofemoral kinematics during the full gait cycle has proven to be a challenging task. Full gait cycle kinematics are most often recorded using skin-mounted markers [1–5], although inertial measurement units (IMUs) are becoming more popular [6,7]. These technologies that attach measurement devices to the skin are known to be inaccurate due to skin motion artifact, which introduces errors of up to 14 mm in translation and 16 deg in rotation, depending on the marker location and activity [8–11]. Although skin motion artifact reduction is an active area of research [12,13] a generalizable solution remains elusive. Improved accuracy may be achieved by attaching markers to bone pins [11,14], however, use of this technique is limited due to its invasive nature. A less invasive technique is biplane radiography, which has demonstrated submillimeter accuracy in tracking bone motion during dynamic activities [15,16]. Processing biplane radiography data are extremely time-consuming, typically requiring hours to track motion from a single movement trial, which limits sample size using this technology. Biplane radiography also has a limited field of view which permits only a portion of the gait cycle to be imaged [15,17–19] unless a mobile radiography system that moves with the participant is used [16,20].

The existing datasets that describe 6DOF knee kinematics during the full gait cycle in healthy adults and are not affected by skin motion artifact comprise 5–15 participants [11,14,20]. Knee kinematics from a larger group of healthy knees will allow us to better characterize the population and improve our ability to

identify abnormal knee kinematics that may be associated with pathology. Therefore, the purpose of this study was to characterize 6DOF knee kinematics and variability in knee kinematics waveforms during treadmill walking in healthy knees over the entire gait cycle using dynamic biplane radiography.

Methods

Participants. Following Institutional Review Board approval from University of Pittsburgh (PRO16070246), 20 participants gave written informed consent to participate in this study. All participants were healthy adults with no history of knee injuries (Table 1).

Data Collection. Data were collected using dynamic biplane radiography. The system was positioned with one beam pointing in the posterior to anterior direction, and the second beam offset 55 deg, with the emitter to image intensifier distance of 1800 mm (Fig. 1). The X-rays were produced by two independent X-ray generators (CPX3100CV SP, EMD Technologies, Saint-Eustache, QC, Canada) and X-ray tubes (G1080, Varian, Palo Alto, CA), and images were captured using a high-speed camera (Phantom V10, Vision Research, Wayne, NJ) attached to 16 in. image intensifiers (TH 9447 QX H694 L VR70, Thales, La Défense, France). The recorded images were captured at 1824×1800 pixels and down sampled to 512×512 for image processing. After the participant trials were concluded, images were distortion corrected using a grid attached to the image intensifier, and the imaging volume was calibrated using a cube with embedded beads of known distances.

Synchronized biplane radiographs were collected (100 images/s for 1.0 s, maximum 90 kV, 160 mA, 1 ms pulse width) during walking at a self-selected speed (average 1.3 ± 0.2 m/s) on an instrumented treadmill. The walking speed for each participant was determined by averaging the velocity of a reflective marker

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Table 1 Participant age, height, weight and BMI. Mean values $\pm$ standard deviation and range of values (show in parentheses) for each measurement

Sex	Age (years)	Height (cm)	Weight (kg)	BMI
10 Females	31.0 $\pm$ 6.8 (23–42)	165.2 $\pm$ 5.6 (156–173)	65.5 $\pm$ 11.7 (46.7–80.7)	24.0 $\pm$ 3.6 (17.8–28.7)
10 Males	30.5 $\pm$ 6.2 (22–40)	179.2 $\pm$ 7.6 (170–192)	77.4 $\pm$ 8.5 (62.1–90.7)	24.2 $\pm$ 2.7 (18–27.5)

Fig. 1 The two configurations used to capture the full gait cycle in separate segments for the right leg. The first set of images were acquired from just prior to foot strike through midstance (left), and the second set of images were acquired from midstance through terminal swing (right). The imaging system was rotated to avoid occlusion from the contralateral leg for the gait cycle portion of interest.

located on the back of the neck (C7) from four trials of the participant walking across the testing room (12-camera Vicon Vantage V5, Vicon Motion Systems, Inc., Oxford, UK). The subjects required less than 60 s to acclimate to treadmill walking at the desired speed. For each knee, two separate portions of the gait cycle were imaged. The first set of images were acquired from just prior to foot strike through midstance, and the second set of images were acquired from midstance through terminal swing (Fig. 1). Two trials were collected for each imaging configuration, resulting in four total trials per knee that were collected over four separate steps. Concurrently, ground reaction forces were recorded at 1000 Hz using a dual-belt instrumented treadmill (Model: TM-07-B, Bertec Corp, Columbus, OH).

Bilateral computed tomography (CT) scans of each participant's knees were obtained ($0.6 \times 0.6 \text{ mm}^2$ in-plane resolution, 1.25 mm slice thickness). The images were resliced to generate $0.6 \times 0.6 \times 0.6 \text{ mm}^3$ cubic voxels, and a combination of automated and manual segmentation of bone tissue was performed using commercial software (MIMICS, Materialize, Leuven, Belgium). Three-dimensional models of each femur and tibia were created from the segmented bone tissue [21]. A previously validated volumetric model-based tracking technique was used to match digitally reconstructed radiographs created from the

participant-specific CT scans to the biplane radiographs [22]. This tracking system has a validated in vivo accuracy of 0.7 mm or better in translation and 0.9 deg or better in rotation for tracking the tibiofemoral joint [15] (Fig. 2). The maximum radiation exposure related to this study was estimated to be 1.32 mSv, comprised of 1.13 mSv from the CT scan, and 0.19 mSv from biplane radiographic imaging (estimated using PCXMC, STUK—Radiation and Nuclear Safety Authority, Helsinki, Finland).

Data Processing. A total of 39 knees were included in the dataset (one participant did not have a complete dataset for one knee due to an error in calibrating the biplane radiography system). Anatomic coordinate systems were determined using an automated algorithm based upon articular surface geometry [23] and single CT slices through the proximal femur and distal tibia [24] on the right-side bones. For the left-side, the right femur and tibia was mirrored and coregistered to the left-side bone model using a rigid body coherent point drift algorithm [25]. Key landmark locations were then copied to the left-side and the anatomical coordinate system was defined (180 deg rotated along the superior/inferior axis from right-side). Tibiofemoral rotational kinematics (internal/external (IE), abduction/adduction (AbAd), and flexion/

Fig. 2 Data collection and model-based tracking workflow. (a) The participant walked on a dual-belt instrumented treadmill within a biplane radiographic imaging system. (b) Synchronized biplane radiographs were collected at 100 Hz. (c) Participant-specific three-dimensional (3D) bone models of the bilateral femur and tibia were created from the CT scan. (d) Anatomical landmarks were placed on the 3D bone models to establish an anatomic coordinate system. (e) The volumetric matching process to track 3D bone motion was performed using an automated algorithm to maximize the correlation between the Digitally Reconstructed Radiographs (DRR) and the distortion-corrected biplane radiographs. (f) Joint kinematics were calculated according to the anatomic coordinate systems and bone motion.

extension (FE) rotation) were calculated following the method of Grood and Suntay [26]. Translational kinematics (anterior/posterior (AP), proximal/distal (PD) and medial/lateral (ML) translation) were calculated using the vector from the femur origin (midpoint between the center of the femoral condyles) to the tibia origin (midpoint between the most medial and lateral points of the tibial plateau), expressed in the tibial coordinate system. Left knee kinematic results were mirrored to match the sign convention for the right knee. Knee kinematic results were filtered at 10 Hz using a fourth-order Butterworth filter, with the optimal filter frequency determined through residual analysis [27].

The kinematic measurements from each trial were input into a custom MATLAB (MathWorks, Natick, MA) program to interpolate the results to percent gait cycle. Three discrete points were used to interpolate the data to percent gait cycle; first foot strike, toe off, and second foot strike. The timings of all three events were extracted from the vertical ground reaction force recorded by the treadmill, with a 50 N threshold. The stance phase was mapped to 0–60% of the gait cycle and the swing phase mapped to 60–100% of the gait cycle for each participant. Kinematic results over multiple trials for each knee were then averaged to calculate knee-specific average kinematic curves. To determine the within knee trial-to-trial variability, standard deviations of the kinematic parameters were calculated for the overlapping portions of the multiple trials for each knee.

For some participants, the contralateral knee occluded the knee of interest being imaged during late stance to early swing in the oblique view. Due to these occlusions, kinematics were not calculated at these portions of the trial. These gaps in the continuous kinematic curves were filled using a Piecewise Cubic Hermite Interpolating Polynomial (pchip) fit. Pchip fit was used due to its shape-preserving nature, as some gaps were located at or near inflection points, and using a normal cubic spline function at these gaps could result in erroneous estimations. Across the entire dataset, there was kinematic data from a minimum of 30 knees at each 1% interval of the gait cycle (Fig. 3). To quantify potential errors due to gap filling, data were removed from the knees that had full gait cycle data, using the average location and duration of missing data from the knees with incomplete gait cycle data, and the same gap filling technique was applied. The accuracy of the gap fill technique was estimated by the average absolute difference between true data and the gap filled data.

Range of motion (ROM) for each of the 6DOF kinematics was calculated for each knee using the maximum and minimum values

Fig. 3 The number of knees with measured data at each 1% interval of the gait cycle. The number of knees with measured data at each percent of the gait cycle is represented as the solid line. The dotted line represents the minimum number of knees with measured data for any 1% interval of the gait cycle.

of the continuous kinematics curves and averaged for the entire group during the stance phase, swing phase, and over the full gait cycle. Standard deviation of the kinematic ROM was assessed as a measure of interknee variability.

To determine variability in the continuous kinematics curves, 90% confidence interval prediction bands for the entire kinematics waveform were determined following methods described by Lenhoff et al. [28]. To assess the variability in the shape of the kinematic waveforms after accounting for knee-specific offsets, each kinematics curve was shifted on the vertical axis to minimize the differences between the individual kinematics curves and the average curve, and the 90% confidence interval of the shifted curves was calculated.

Results

A total of 81 gait cycle portions for the right-side and 80 cycle portions for the left-side were tracked. With the 161 total gait cycle portions, 39 full gait cycles were calculated, of which 22

Fig. 4 Six degree-of-freedom kinematics for 39 knees during a full gait cycle. Thin lines indicate kinematics for each of the 39 knees. Thick black lines indicate group mean curves. The dotted lines represent the 90% prediction bands calculated using the bootstrapping technique. The shaded area represents swing phase.

required interpolations in late stance to early swing phase (average of 11% of gait cycle) to obtain a continuous kinematic curve. The average differences between gap filled and true data values were 0.4 ± 0.3 , 0.5 ± 0.3 , and 1.1 ± 0.7 mm in the LM, PD, and AP translations, respectively. The average differences in rotations were 1.6 ± 1.0 deg, 0.4 ± 0.2 deg, and 1.0 ± 0.4 deg in FE, AbAd, and IE rotations, respectively. For within-knee trial-to-trial variability, the maximum standard deviation across trials was in IE rotation at 1.1 deg and in AP translation at 1.1 mm.

The 90% prediction bands had a width of approximately ± 15 deg, ± 6 deg, and ± 10 deg in FE, AbAd, and IE rotations,

and ± 4 mm, ± 10 mm, and ± 5 mm for ML, AP, and PD translations, respectively (Fig. 4). The individual FE waveforms followed the group mean pattern, whereas the individual AbAd and IE waveforms displayed a wider variability in pattern. For translations, in general, the individual ML, AP, and PD translation waveforms were similar in pattern to the group mean curve, but with a knee-specific offset. The average Fourier series fit of the translational and rotational kinematic waveforms for each individual knee were all greater than $R^2 > 0.95$ and $R^2 > 0.96$, respectively.

The average tibiofemoral ROM was smaller during stance than during swing in all six degrees-of-freedom (Table 2). The ROM

Table 2 Average individual knee tibiofemoral ROM during stance, swing and over the entire gait cycle, and the ROM of the group average kinematic curves over the entire gait cycle

	Average of all individual knees			Group average		
	Stance ROM	Swing ROM	Total ROM	Stance ROM	Swing ROM	Total ROM
Medial–lateral (mm)	2.1 ± 0.8	2.8 ± 1.0	3.2 ± 1.2	1.4	2.1	2.6
Anterior–posterior (mm)	4.6 ± 1.8	6.4 ± 2.4	7.0 ± 2.2	2.5	4.8	4.8
Proximal–distal (mm)	2.1 ± 0.8	2.6 ± 1.0	2.9 ± 1.0	1.6	1.6	1.7
Flexion–extension (deg)	29.6 ± 10.1	66.2 ± 11.9	67.3 ± 12.1	30.8	66.1	66.1
Abd–adduction (deg)	2.2 ± 1.0	3.4 ± 1.5	3.7 ± 1.6	1.4	2.5	2.5
Internal–external (deg)	8.8 ± 3.1	10.2 ± 4.1	11.5 ± 3.7	6.7	7.3	7.3

All values shown are mean ± 1 SD.

determined from the group average kinematics curves underestimated the average ROM for all knees by up to 2.2 mm in AP translation and by up to 3.1 deg in IE rotation (Table 2).

The 90% prediction bands were reduced to a width of approximately ±12 deg, ±2 deg, and ±7 deg in FE, AbAd, and IE rotations, and ±2 mm, ±4 mm, and ±1 mm for ML, AP, and PD translations, respectively, after knee-specific offset minimization (Fig. 5).

Discussion

This study provides, to our knowledge, the largest dataset of 6DOF full gait cycle knee kinematics not affected by skin motion artifact, using an amalgamation technique to overcome the limitations inherent to conventional biplane radiography. In addition to calculating kinematics for 39 knees, this study also determined 6DOF tibiofemoral ROM during stance, swing, and the full gait cycle, along with 90% prediction bands and normalized kinematics waveforms that can be used in tandem to identify abnormal kinematics over the entire gait cycle.

Overall, the rotational kinematics patterns observed in this study were most similar to those reported by Gray et al. [20], who used a mobile biplane radiography system to track tibiofemoral motion during over ground walking (Fig. 6). Comparison of our results to those of Guan et al. [29] suggest that after TKA, the knee remains more extended during support, and that the TKA knees tested do not allow the small adduction that occurs during early support or the internal rotation during push-off and early swing that occurs in healthy adult knees. The largest rotational differences between our results and previous studies were found in the IE rotation component reported by Andriacchi and Dyrby [1], who used a skin surface marker system. However, differences in IE rotation between Andriacchi and Dyrby [1] and other studies may not be entirely explained by the use of skin surface markers. Kadaba et al. [5] also used a skin surface marker system and reported IE rotation kinematics during the stance phase that were similar to those found using bone pins [11,14] and biplane radiography [30]. These differences may be due to different marker sets and/or skin motion artifact suppression techniques.

The translational kinematics observed in this study were also most similar in pattern to Gray et al. [20] for the medial–lateral component (Fig. 6). The medial–lateral offsets among studies are explained by differences in the methods used to establish the anatomical origins. For example, Gray et al. defined the femoral origin as “the foot of the perpendicular dropped from the intercondylar notch apex to the X-axis” [20], while the current study defined it as the midpoint along the line connecting the medial and lateral condyle centers. Another example of differences due to origin definition is that Lafortune et al. [14] defined the tibial origin as the most proximal point on the medial tibia spine, which placed the tibia origin slightly medial to the center of the tibia, in contrast to the present study which placed the tibia origin closer to the center of the tibia. The two studies that used bone pins [11,14] as well as Kozanek et al. [30] observed similar patterns in lateral shift at the beginning of the gait cycle followed by

a medial shift during mid-to-late stance. For the anterior–posterior translations, Guan et al. [29], Gray et al. [20], and the present study observed similar patterns with an anterior shift at 70–80% of the gait cycle that was not observed by Lafortune et al. [14]. The shift in value is again likely due to differences in how the anatomical origin was defined. The similarity between the present study and Guan et al. [29] suggests that the translational kinematics after TKA closely replicate healthy knee kinematics. The anterior–posterior component observed by Andriacchi and Dyrby [1] again differs drastically compared to all other studies, potentially due to skin motion artifact.

There are several potential explanations for the observed differences between our results and previously reported full gait cycle kinematics curves. First, the methods used to define anatomic coordinate systems and the location of the coordinate system origins were not identical for all research studies, which will cause shifts in the value even if kinematics patterns are similar. Second, skin motion artifact affected the results for all studies that used skin-mounted markers to estimate underlying bone motion. Finally, previous studies that used bone pins comprised only five and eight participants. As demonstrated in our data from 39 knees, substantial variability exists among knees. Therefore, mean kinematics from a small sample of knees may not adequately reflect average knee kinematics in a larger sample.

The present dataset provides a sufficient sample size to estimate variability in full gait cycle knee kinematics in the healthy population. As demonstrated by the 90% prediction bands and the individual knee kinematics curves, there are a large range of knee kinematic patterns within the healthy population. The individual and mean curves also serve to demonstrate that the group average kinematic curves do not represent the kinematics of individual knees very well. These differences between individual and mean curves highlight the importance of subject-specific kinematic analysis, computational modeling, and prosthesis design.

In contrast to the continuous kinematics curves, the ROM results were relatively consistent across knees. Interknee variability in ROM was less than 2.4 mm and 4.1 for all DOF except FE, which demonstrated by far the greatest ROM and variability among knees. These findings suggest that, with the exception of FE, knee ROM is fairly consistent across individuals during gait, however, the knee-specific “offset” among knees is quite variable. These offsets may be due to morphological differences between individual bones or differences in neuromuscular control. The presence of these offsets highlights an advantage of using ROM to identify abnormal knee kinematics. Controlling for these knee-specific offsets may be an effective strategy for comparing individual gait kinematics to an average curve to identify gait abnormalities based upon entire kinematic waveform patterns (Fig. 5).

The present study calculated average and 90% prediction bands for the 6DOF kinematics using a bootstrapping technique [28]. Beyond the present study, these prediction bands may be used to identify “abnormal” or “atypical” gait patterns, and to determine at what point in the gait cycle abnormalities may occur.

There are important limitations to keep in mind when interpreting this data. First, the participant group comprised healthy and

Fig. 5 Six degree-of-freedom kinematics for 39 knees after RMS optimization to remove knee-specific offsets and to highlight similarities of waveforms

relatively young adults, and therefore the results should not be extrapolated to apply to older individuals or individuals with injury or other pathology. It is also important to keep in mind that bone kinematics were determined using the anatomical coordinate systems established within each bone and not normalized to a standing neutral position or the knee orientation during the CT scan. By using the anatomical coordinates dictated by individual morphology, differences in the definition of “neutral” positions that can be affected by factors such as posture or scanning position in the CT scanner were removed and the natural variabilities in the kinematics were highlighted. In addition, the gait kinematics reported here were performed on a treadmill, and previous research suggests kinematics differences may exist between treadmill and over ground walking in knee arthroplasty patients [29]. The individual kinematics waveforms presented in this study are

an amalgamation of data obtained separately for the stance and swing phases from multiple trials and therefore not a true single gait cycle kinematics waveform. Considering the maximum potential errors introduced by the gap filling technique, the within-knee variability, and the uncertainty of the measurement system, this technique calculated knee kinematics to within 1.8mm and 2.5deg of the true kinematic waveforms for each knee. Although motion capture systems can more easily capture an entire gait cycle without the need for multiple trials, the knee-specific variability found in the current study suggests that it would be a considerable challenge to accurately correct for skin motion artifact across individual morphologies, kinematics, and activities. Finally, the prediction bands define a boundary for full gait kinematics of young healthy adults; however, “abnormal” kinematic patterns could exist within these boundaries.

Fig. 6 Comparison of rotational and translational kinematics from previous studies to the current study. The translational kinematic curves from LaFortune et al., Guan et al., and Benoit et al. followed conventions set forth by Grood and Suntay, whereas the Kozanek et al., Andriacchi et al., and the present study expressed translations with respect to the tibia coordinate system.

The results presented here highlight the importance of subject- and patient-specific analysis when assessing the effects of injury, surgery, rehabilitation, and age on knee kinematics. The data suggest that computational models developed using group average kinematics data are unlikely to be representative of the kinematics of any individual and highlights the need for subject-specific kinematics in musculoskeletal modeling [11]. Future work should be directed at identifying morphologic and neuromuscular factors that contribute to these individual differences in knee kinematics.

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