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Automatic registration of MRI-based joint models to high-speed biplanar radiographs for precise quantification of *in vivo* anterior cruciate ligament deformation during gait

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Abstract

Understanding *in vivo* joint mechanics during dynamic activity is crucial for revealing mechanisms of injury and disease development. To this end, laboratories have utilized computed tomography(CT) to create 3-dimensional (3D) models of bone, which are then registered to high-speed biplanar radiographic data captured during movement in order to measure *in vivo* joint kinematics. In the present study, we describe a system for measuring dynamic joint mechanics using 3D surface models of the joint created from magnetic resonance imaging (MRI) registered to high-speed biplanar radiographs using a novel automatic registration algorithm. The use of MRI allows for modeling of both bony and soft tissue structures. Specifically, the attachment site footprints of the anterior cruciate ligament (ACL) on the femur and tibia can be modeled, allowing for measurement of dynamic ACL deformation. In the present study, we demonstrate the precision of this system by tracking the motion of a cadaveric porcine knee joint. We then utilize this system to quantify *in vivo* ACL deformation during gait in four healthy volunteers.

Keywords

kinematics; biomechanics; ACL; biplanar radiography; optimization; strain; knee; injury mechanism; imaging

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INTRODUCTION:

Understanding dynamic *in vivo* knee joint kinematics is crucial for revealing mechanisms of injury and disease development. For example, understanding the mechanical behavior of the anterior cruciate ligament (ACL) during dynamic activity can provide information on knee positions that increase injury vulnerability, reveal injury-related changes in biomechanics, or inform design criteria for graft function. High-speed biplanar radiography has been developed to measure *in vivo* skeletal motion during dynamic activity (Anderst et al., 2009; Bey et al., 2006; Dennis et al., 2005; Farrokhi et al.; Nagai et al.; Scott et al., 2004; You et al., 2001). Several studies have utilized radiographic tracking of markers implanted in bone to quantify skeletal motion (Jessica et al., 2010; Jonsson et al., 1989; Kärrholm et al., 1988; Scott et al., 2004; Tashman and Anderst, 2003). More recently, motion tracking has been accomplished by registering 3-dimensional (3D) bone models derived from computed tomography (CT) to radiographs (Anderst et al., 2009; Asano et al., 2001; Bey et al., 2006; Dennis et al., 2005; Nagai et al., 2017; Tang et al., 2017; You et al., 2001).

To extend these methodologies to the deformations of soft tissues, such as the ACL, 3D models of bone and soft tissue structures can be derived from magnetic resonance images (MRI) (Bischof et al., 2010; Carter et al.; Taylor et al., 2013; Taylor et al., 2011; Utturkar et al., 2013; Van de Velde et al., 2009; Wu et al., 2010). Specifically, in the case of measuring ACL deformation, the footprints of the ACL attachment sites on the femur and tibia can be visualized and modeled using MRI (Abebe et al., 2009; Taylor et al., 2013; Taylor et al., 2011; Utturkar et al., 2013). Previous work has quantified ACL deformation for static knee postures using MRI-based 3D models manually registered to radiographs (Jordan et al., 2007; Utturkar et al., 2013). More recently, biplanar radiography has been combined with optical marker-based motion capture to quantify dynamic ACL deformation during gait (Taylor et al., 2013) and jump landing (Taylor et al., 2011). ACL deformation has also been evaluated during the stance phase of gait, using MRI models manually registered to biplanar radiographs (Wu et al., 2010).

The objectives of the present study were to develop a methodology for tracking *in vivo* joint mechanics using MRI-based models of bone and soft tissue automatically registered to biplanar radiographs, and to utilize this system to measure dynamic ACL deformation during gait. For use in system development, 3D surface models of a cadaveric porcine femur and tibia were created from MRI using a previously validated technique (Van de Velde et al., 2009). Then, a calibration method was developed to define the geometry of the biplanar radiographic imaging system, and to unwarp inherent distortion in the radiographic images. Subsequently, automatic registration software was developed to register models of the bone to biplanar radiographs. We assessed the precision of this system by implanting the porcine joint with stainless steel beads, imaging it as it swung freely, and tracking the motion in two ways: 1.) by marker-based tracking of the implanted stainless steel beads and 2.) using the automatic registration methodology. Finally, we applied this novel system to quantify *in vivo* ACL deformation during gait in healthy volunteers.

METHODS:

3D modeling of the porcine joint

A cadaveric porcine knee joint was obtained from a local abattoir, with the femur, tibia, and all soft tissues intact. Sagittal plane MRIs were acquired using a double-echo steady-state (DESS) sequence (voxel size: $0.3 \times 0.3 \times 1.0 \text{ mm}^3$, flip angle: 25° , TR: 17 ms, TE: 6 ms) on a Siemens 3T Trio Tim scanner (Abebe et al., 2009; Okafor et al., 2014; Sutter et al., 2015). The outer contours of the femur and tibia were manually segmented from the sagittal MR images using solid-modeling software (Rhinoceros 4.0, Robert McNeel and Associates). Surface models, consisting of uniformly distributed vertices and corresponding faces, were created from these contours (Coleman et al., 2013; Van de Velde et al., 2009; Widmyer et al., 2013).

Calibration of the high speed radiography system and image unwarping

The biplanar radiography system consists of two x-ray generators, (EMD technologies), two x-ray tubes (G296, Varian), and two image intensifiers (41cm diameter, TH 9447 QX, Thales) which are coupled to two high-speed cameras (Phantom v9.1, Vision Research). Several calibration steps (Figure 1) were necessary to define the geometry of the imaging environment and unwarp inherent distortion from the radiographic images (Reimann and Flynn, 1992). Briefly, a source beam alignment tool (Figure 1A) was imaged on each intensifier (Figure 1B) and used to determine where the source beams intersect with their respective intensifiers (Figure 1C, red arrow). A spatial calibration plate was then imaged on both planes simultaneously. In post-processing, a model of the spatial calibration plate was positioned such that its projections onto the intensifiers matched the radiographs; this determined the relative positions of the intensifiers and sources and defined the geometry of the imaging environment (Figure 1D).

To correct inherent distortion in the radiographic images, an acrylic calibration plate consisting of 4,500 regularly-spaced stainless steel beads was imaged (Figure 1E, before (left) and after (right) unwarping). Distortion correction software written in MATLAB (Mathworks, Natick, MA) was used to align beads in the distorted image with their known locations. Beads in each distorted image were coarsely identified with an edge detection filter, and the centroid of each bead was refined by fitting the bead pixel intensities with a bivariate Gaussian function. The bead centroid was then defined as the mean of the Gaussian function (the position of the peak signal intensity). The image was then divided into 25 sections and, within each section, 2-dimensional (2D) image distortion was quantified by calculating the error between distorted bead positions and actual bead positions (Figure 1F, before and after unwarping). A set of orthogonal polynomials was fit to the error data and used to map the distorted bead positions to their actual positions. This solution was then applied to the radiographic images of the joints in motion to correct distortion. Twenty trials were performed to evaluate the precision of the unwarping procedure; the average error between the unwarped bead positions and the actual bead positions was 0.10 ± 0.02 pixels ($31 \pm 6 \mu\text{m}$, mean $\pm$ standard deviation) over these trials.

Automatic registration of 3D models to biplanar radiographic data

Software was developed in MATLAB to enable automatic registration of the surface models of the bones to biplanar radiographs (Figure 2). The purpose of the software is to calculate the 6 degrees of freedom (6DOF) parameters necessary to translate and rotate a bone surface model to match biplanar radiographs. The interface (Figure 2) allows the user to manipulate the MRI model manually to provide an initial condition for the optimization algorithm used in automatic registration (Figure 2, Manual Adjustment). The interface also allows the user to process the radiographs to facilitate registration by providing contrast enhancement, filtering, and edge detection options.

The registration consists of an optimization that iteratively applies combinations of 6DOF parameters to move the model in 3D space until its projections onto the intensifiers from the perspective of the sources match the biplanar radiographs. The combinations of 6DOF parameters tested at each iteration are guided by a constrained (Figure 2, Parameter Limits) Nelder-Mead simplex search algorithm (Lagarias et al., 1998). The edges of the biplanar radiographic images are detected using an edge detection filter (Figure 2, red points). At each iteration, the vertices of the surface model are projected onto the radiographs (Figure 2, blue points). The optimization cost function calculates the distance between each projected model vertex and their nearest radiographic edge points. The goal of the optimization is to minimize this cost function by positioning the model such that the distances between the projected edge points and their nearest neighbor radiographic edge points are minimized on both radiographs simultaneously. This algorithm is feasible because the density of the projected model vertices is greatest around radiographic edges. Projected vertices that fall outside the field of view (FOV) or outside a user-specified distance (Figure 2, Edge Distance) from a radiograph edge point are ignored. This prevents the model from deviating from reasonable solutions, and allows for precise registration even when the joint is partially outside the FOV, occluded by another limb, or when the edge information of the radiograph does not include the entirety of the edge of the bone. The stopping criteria for the optimization is that both the 6DOF parameters and cost function change by less $< 0.1\%$ of their value between iterations. An example optimization is shown in Figure 3.

Following an initial manual positioning of the model to generate an initial condition for the 6DOF parameters, optimization is performed. This procedure is performed for the first two biplanar radiographs in the series. Subsequent registrations for the remainder of the series are performed automatically. The initial condition for each of these optimizations is predicted by numerically estimating the derivative of the 6DOF parameters from the two prior optimizations.

System precision measurement

The precision of the imaging system was assessed using a cadaveric porcine joint implanted with stainless steel beads (3 each in the femur and tibia). The joint was frozen to preserve the relative position of the femur and tibia. Images were acquired at a frame rate of 120 Hz, a per-frame exposure time of 1.5 ms, and a radiographic protocol of 75 kVp and 160 mA. Images had a matrix size of 1152×1152 pixels, with a pixel size of 0.3×0.3 mm. Three imaging trials consisting of 200 biplanar radiographs each were collected as the joint swung

freely. For each trial, the positions of the femur and tibia were calculated in two ways: 1.) using marker- based tracking of implanted steel beads and 2.) using the automatic registration methodology.

Marker-based tracking provided an estimate of the overall tracking precision of the imaging system, and served as benchmark of precision to compare with the automatic registration method. Marker-based tracking was performed using an optimization procedure to position each bead within the imaging environment such that its projections onto the intensifiers matched the beads as they appeared in the radiograph. The centroids of these bead positions were calculated for both the femur and tibia, and the precision of the imaging system was defined as the standard deviation between these centroids. For the automatic registration precision measurement, the software was used to position the models of the femur and tibia. The series of 6DOF parameters obtained from automatic registration were filtered using a zero-phase fourth order low-pass Butterworth filter with a cutoff frequency of 10 Hz to reduce noise (Goyal et al., 2012), and the model was repositioned using the filtered parameters. Precision of automatic registration was defined as the standard deviation of the distance between the centroids of the femur and tibia over the set of images.

***In vivo* ACL deformation during gait**

Institutional Review Board (IRB) approval was obtained for this study and subjects provided written consent prior to participation. Subjects had no history of injury or surgery to either knee. Right knees of 3 female and 1 male participant were imaged using MRI and high speed biplanar radiography (age range: 23–34 years, body mass index (BMI) range: 21.3–24.9). Sagittal and coronal plane MRIs were acquired using the previously described DESS sequence, with subjects lying supine with their knees relaxed (mean flexion angle: $15.1^{\circ} \pm 1.3^{\circ}$). The outer contours of the femur and tibia were outlined on the sagittal (Figure 4A) and coronal images (Figure 4B). ACL attachment sites were outlined on the coronal images (Figure 4B). These sagittal (Figure 4C) and coronal (Figure 4D) contours were compiled into wireframe models and converted into 3D surface models consisting of uniformly spaced vertices and their corresponding faces. The coronal model was registered to the sagittal model to generate a surface model with an appropriately positioned ACL attachment site (Figure 4E). Orthogonal image sets were used to confirm the shape and position of the ACL. Previous validation studies have demonstrated that this approach can locate the center of the ACL footprint to within 0.3 mm (Abebe et al., 2009; Taylor et al., 2013). Additionally, the ACL attachment sites on the femur and tibia were divided into anteromedial (AM) and posterolateral (PL) bundles as described previously (Jordan et al., 2007; Utturkar et al., 2013).

During biplanar radiography, participants were positioned on a dual belt treadmill such that the knee joint was approximately centered within the FOV of both intensifiers. The same imaging speed and resolution as in the porcine joint validation study were used. Each *in vivo* experiment used a radiographic protocol not exceeding 110 kVp/200 mA. Two 3 second trials were performed; this resulted in 540 ms of total x-ray exposure per trial. To assess radiation risk to subjects, the radiation effective dose (a weighted average of absorbed doses to bone surfaces, skin and soft tissues) was calculated by Duke Radiation Safety from the

total skin entrance exposure and energy absorption by the tissues. The total effective dose was less than 0.14 millisievert per participant, which results in minimal risk (NCRP, 2010). Participants walked on the treadmill 5 minutes in order to practice keeping their joint within the FOV prior to imaging. Data was then collected as subjects walked at a speed of 1 m/s. Automatic registration was used to position the femur and tibia, thereby reproducing *in vivo* gait kinematics for each subject (Figure 5).

Flexion angle (°) and ACL length (mm) were measured from the models in their optimized positions for each biplanar radiograph collected during gait. Flexion angle was defined based on the orientation of the long axis of the tibia relative to the long axis of the femur, using the transepicondylar axis of the femur as the axis of rotation (Grood and Suntay, 1983; Taylor et al., 2013; Utturkar et al., 2013). The long axes of the femur and tibia were defined by cylinders fit to the shafts of the bones. The transepicondylar axis of the femur was set between the most medial and most lateral points of the femoral condyles. ACL length was defined as the straight line distance between the centroids of the ACL attachment sites on the femur and tibia. Heel strike was defined as the beginning (0%) and end (100%) of the gait cycle (Taylor et al., 2013). Flexion angle and ACL length were then interpolated to represent points ranging from 0 – 99% of the gait cycle in 1% increments to allow for comparison of data across subjects. ACL strain (%) was calculated as $\text{strain} = \frac{l - l_0}{l_0} \times 100\%$

(Fleming and Beynon, 2004) where the reference length l_0 was estimated as the ACL length measured from the MRI model in its unloaded position in the MR scan (Taylor et al., 2011). Flexion angle (°) and ACL length (mm) at points ranging from 87–100% and 0–37% of the gait cycle were averaged across subjects. This portion of the gait cycle captures the end of swing phase, heel strike, loading, mid-stance, and terminal extension.

Statistics were performed using SAS (version 9.4, SAS Institute, Cary, NC). Linear mixed models were carried out to test whether flexion angle (°) was a significant predictor of ACL length (mm). The relationship between flexion angle and ACL length was quantified using a Spearman's Rank correlation. The relationships between average overall ACL lengths and the average lengths of the AM and PL bundles were assessed using a Pearson correlation.

RESULTS:

System precision measurements using a porcine knee joint

The system precisions assessed using the marker-based tracking method were 46, 42, and 45µm for each of the three trials. For automatic registration of the 3D models of the knee, the precisions were 73, 65, and 70µm for the same three trials. These precisions include the portions of the trials where the femur and/or tibia extend partially outside of the FOV. Average matching times for each trial were 6.5, 7.8, and 8.7s per image for the femur, and 6.3, 7.0, and 8.3s for the tibia. For the automatic registration of the porcine knee joint experiments, following an initial manual positioning of the femur and tibia, all trials were matched automatically using the software with no additional user input. These results are summarized in Table 1.

ACL deformation during gait

The average ACL reference length (l_0) used in strain calculations was 26.8 ± 1.6 mm (mean $\pm$ standard deviation), and subjects were positioned at a mean flexion angle of $15.1 \pm 1.3^\circ$ at the time of MR imaging. The average resting length for the AM bundle was 29.4 ± 2.4 mm, and the average resting length for the PL bundle was 23.5 ± 1.2 mm. The means and standard deviations of flexion angle and ACL length at each percentage of the gait cycle are plotted in Figure 6. A minimum in flexion angle ($-2.2 \pm 6.7^\circ$) was observed at 97% of the gait cycle (Figure 6A), which was associated with a peak in ACL length (29.2 ± 2.9 mm) and a ACL strain of $9 \pm 5\%$ (Figure 6B). A second minimum in flexion angle ($-6.3 \pm 5.0^\circ$) was observed at 37% of the gait cycle (Figure 6A), accompanied by corresponding peaks in ACL length (29.2 ± 1.6 mm) and strain ($9 \pm 3\%$) (Figure 6B). The linear mixed model revealed that flexion angle was a significant predictor of ACL length ($p < 0.0001$), and ACL length ranked significantly ($p < 0.0001$) with flexion angle such that ACL length increased with increasing extension ($\rho = -0.68$, $p < 0.01$). Both average AM ($R^2 = 0.99$, $p < 0.01$) and average PL ($R^2 = 0.99$, $p < 0.01$) bundle lengths were linearly correlated with average overall ACL length.

DISCUSSION:

In the present study, we describe a method for quantifying skeletal kinematics and soft tissue deformations that combines MRI-based 3D surface models of the joint, high-speed biplanar radiography, and a novel automatic registration methodology. The use of MRI allows for modeling of soft tissues, such as the footprints of the ACL attachment sites on the femur and tibia, in addition to bony anatomy. We found that the automatic registration methodology presented here was able to track joint motion with high precision, even when bones extended partially outside the FOV (Table 1). Using this system, we measured the *in vivo* mechanics of the ACL during gait.

The results obtained in this study with regard to ACL deformation during gait are congruent with those present in the literature. Specifically, a study by Taylor et al. (Taylor et al., 2013) utilized a technique involving biplanar radiography in combination with MRI models and marker-based motion capture to measure knee flexion angle and ACL strain during gait. Minimums in flexion angle, accompanied by peaks in ACL strain occurring just prior to heel strike (~10%) and in mid-stance phase (~13%) were identified. While these results are similar in magnitude to those of the present study, the technique presented here is advantageous in that it removes potential for error introduced by the movement of the optical markers on the skin relative to the underlying bone. The data presented in this study are also in line with a prior biplanar radiography study that measured knee flexion and ACL elongation at discrete intervals during the stance phase of gait, using manual positioning of MRI-based bone models to reproduce *in vivo* kinematics (Wu et al., 2010). Using a sampling rate of 30 Hz, the study by Wu et al. found peaks in the elongation of the AM and PL bundles of the ACL (~9%) around heel strike and during mid-stance (~12%). The automatic registration methodology described in the present study allows for efficient measurement of *in vivo* kinematics at a high spatial and temporal resolution.

Similar to prior imaging studies of ACL function (Taylor et al., 2013; Wu et al., 2010), the present study identified peaks in ACL length associated with knee extension just prior to heel strike and during the mid-stance phase of gait. The finding that ACL length and strain are maximized when the knee is in extension are also supported by *in vivo* studies utilizing arthroscopically implanted strain gauges (Cerulli et al., 2003; Fleming et al., 2001). The peak in ACL length just prior to heel strike may be related to activation of the quadriceps (Li et al., 1999) in anticipation of weight acceptance. To this point, studies have shown that quadriceps activation results in increased loading on the ACL due to anterior shear forces transmitted through the patellar tendon when the knee is positioned at low flexion angles (DeFrate et al., 2007; DeMorat et al., 2004; Draganich and Vahey, 1990; Dürselen et al., 1995; Markolf et al., 1995; Nunley et al., 2003). This mechanism is consistent with the peak strains observed near the time of heel strike and in mid-stance in the work by Taylor et al. (Taylor et al., 2013), and in the present study. Thus, quadriceps activity during these portions of the gait cycle may increase strain on the ACL (Mesfar and Shirazi-Adl, 2005; Yu and Garrett, 2007). To further clarify this mechanism, future studies should evaluate *in vivo* ACL deformation in other dynamic physiological loading conditions, and examine these results in the context of musculoskeletal model studies of gait (Sharifi et al., 2017; Shelburne et al., 2004).

In some *in vitro* studies, the AM bundle of the ACL has been shown to be tight in flexion and lax in extension, while the PL bundle has been described as tight in extension and lax in flexion (Girgis et al., 1975; Petersen and Zantop, 2007). Therefore, we measured AM and PL bundle deformations in the present study, and found that the AM and PL bundle deformations followed a similar relationship to flexion angle as overall ACL deformation throughout the gait cycle (Figure 6B). This finding is congruent with previous *in vivo* studies of AM and PL bundle function (Jordan et al., 2007; Li et al., 2004; Utturkar et al., 2013; Wu et al., 2010), which also indicated that the AM and PL bundles did not appear to have reciprocal function *in vivo*.

The automatic registration methodology presented here is unique in that it utilizes MRI-based surface models of the joint, enabling modeling of soft tissue structures in addition to bony anatomy. Importantly, the tracking precisions measured in the validation portions of this study are in line with those reported for tracking methodologies that utilize CT-based models (Bey et al., 2006; Tashman and Anderst, 2003; You et al., 2001) as both CT- and MRI-based tracking allow for sub-millimeter precision. However, the use of MRI-based surface models provides both computationally efficient and precise model registration (Table 1), enabling the measurement of ACL length and strain during dynamic activity.

Understanding the *in vivo* biomechanics of the knee joint is key to revealing mechanisms of knee injury (i.e. anterior cruciate ligament rupture) and development of disease (i.e. post-traumatic osteoarthritis), as well as important for design criteria of graft function. The methodology presented here represents a comprehensive system for obtaining bone kinematics and soft tissue deformations that is both fast and precise. This system can be used in the future to quantify deformations in other soft tissues during dynamic activity.

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FIGURE 1: Calibration consists of determining the geometry of the imaging environment and unwarping the radiographic images from inherent distortion. (A) A source beam alignment tool was (B) imaged on each intensifier. (C) Using this tool, the intersections of the source beams with their respective intensifiers were determined (red arrow). (D) A spatial calibration plate was imaged on both intensifiers and used to determine the relative positions of the intensifiers and sources and define the geometry of the imaging environment. (E) To correct distortion in the radiographic images, an acrylic calibration plate consisting of 4,500 regularly spaced

stainless steel beads was imaged on each plane, displayed here before (**left**) and after (**right**) unwarping. (**F**) Distortion error was defined as the distance between the centroid each bead in the radiograph and its known location on the calibration plate, displayed here before (**left**) and after (**right**) unwarping.

FIGURE 2:

The registration software automatically positions 3D surface models of bone such that the distances between the model vertices, projected onto the radiographs (**blue points**), and their corresponding nearest neighbor radiographic edge points (**red points**) are minimized. The interface allows the user to manipulate the MRI model manually (**Manual Adjustment**) to provide an initial condition for the optimization algorithm. The interface also allows the user to process the radiographs to facilitate registration by providing contrast enhancement, filtering, and edge detection options. The **Edge Distance Limit** defines how close the projected vertices need to be to a radiographic edge point for them to be included in the optimization. The **Parameter Limits** provide constraints on how much the parameters can change from the initial condition during optimization.

FIGURE 3:

(A) Demonstrated on the cadaveric porcine femur, the projected vertices (**blue**) do not match the radiograph edges (**red**) prior to optimization (**yellow arrows**). (B) The optimization procedure is guided by a constrained Nelder-Mead algorithm which identifies the position that minimizes the weighted mean distance between the radiograph edge points and projected vertices (cost, arbitrary units). (C) After optimization, the projected vertices correspond to the edges of the radiographs. X, Y, and Z refer to translations (mm) and rX, rY, and rZ refer to rotations ($^{\circ}$).

FIGURE 4:

The outer contours of the femur and tibia were outlined on the (A) sagittal and (B) coronal plane images. (B) ACL attachment sites were outlined on the coronal plane images. These (C) sagittal and (D) coronal contours were compiled into wireframe models and converted into 3D surface models consisting of uniformly spaced vertices and their corresponding faces. (E) The coronal model was registered to the sagittal model to generate a surface model with an appropriately positioned ACL attachment site.

FIGURE 5:
3D surface models of the femur and tibia were registered to biplanar radiographs using the automatic registration software (**top**). The kinematics of the joint, as well as the elongation of the ACL, were determined from the models their registered positions (**bottom**).

FIGURE 6:

Flexion angle (°) and ACL lengths (mm) at points ranging from 87–100% and 0–37% of the gait cycle in increments of 1% were averaged across subjects, with 0% and 100% of the gait cycle representing heel strike. The ACL was divided into its anteromedial (AM) and posterolateral bundles (PL). Means and standard deviations of flexion angle and overall ACL length are plotted. (A) A minimum in flexion angle ($-2.2 \pm 6.7^\circ$) was observed at 97% of the gait cycle (just prior to heel strike), and a second minimum ($-6.3 \pm 5.0^\circ$) was observed at 37% of the gait cycle. (B) These minimums in flexion angle were associated with peaks in

ACL length of 29.2 ± 2.9 mm, corresponding to an ACL strain of $9 \pm 5\%$, and of 29.2 ± 1.6 mm, corresponding to $9 \pm 3\%$ strain, respectively. Mean values of the AM (red) and PL (green) lengths are plotted. Both the AM and PL bundle lengths had a similar relationship to flexion angle as overall ACL length over the gait cycle.

Table 1:

Precision of spatial measurements using marker-based motion tracking and automatic registration. Computations were performed on a standard 2.79 GHz 64-bit Intel Core i7 CPU with 16.0 GB of installed RAM. Matching times are presented as means $\pm$ st.dev.

Trial	1	2	3
System precision (marker-based motion tracking) (μm)	46	42	45
Automatic registration precision (μm)	73	65	70
Matching time per biplanar radiograph (femur) (s)	6.5 ± 1.0	7.8 ± 1.0	8.7 ± 1.9
Matching time per biplanar radiograph (tibia) (s)	6.3 ± 1.1	7.0 ± 1.0	8.3 ± 2.0
Femur partially outside FOV (% of biplanar radiographs)	50	43	69
Tibia partially outside FOV (% of biplanar radiographs)	60	25	80