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A new tibial coordinate system improves the precision of anterior–posterior knee laxity measurements: a cadaveric study using Roentgen stereophotogrammetric analysis

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Abstract

Roentgen stereophotogrammetric analysis (RSA) can be used to measure changes in anterior–posterior (A–P) knee laxity after anterior cruciate ligament (ACL) reconstruction. Previous measurements of A–P knee laxity using RSA have employed a tibial coordinate system with the origin placed midway between the tips of the tibial-eminences. However, the precision in measuring A–P knee laxity might be improved if the origin was placed on the flexion–extension axis of rotation of the knee. The purpose of this study was to determine whether a center-of-rotation tibial coordinate system with the origin placed midway between the centers of the posterior femoral condyles, which closely approximates the flexion–extension center-of-rotation of the knee, improves the precision in measuring A–P knee laxity compared to the tibial-eminence-based coordinate system. A–P knee laxity was measured using each coordinate system six times in three human cadaveric knees implanted with 0.8-mm diameter tantalum markers. For each laxity measurement, the knee was placed in a custom loading apparatus and biplanar radiographs were obtained while the knee resisted a 44 N posterior shear force and 136 N anterior shear force. A–P knee laxity was determined from the change in position of the tibia, with respect to the femur, resulting from the posterior and anterior shear forces. The precision for each coordinate system was calculated as the pooled standard deviation of A–P knee laxity measurements. The precision of the center-of-rotation coordinate system was 0.33 mm, which was about a factor of 2 better than the 0.62 mm precision of the tibial-eminence coordinate system ($p = 0.006$). The 0.33 mm precision with the center-of-rotation coordinate system suggests that an observed change of either 0.56 mm (i.e. 1.7 standard deviations) or greater in A–P knee laxity over time is a real change and not due to measurement error when the new tibial coordinate system is used and other factors contributing to variability are controlled as was done in this study. Accordingly, clinicians and researchers should consider the use of this alternate tibial coordinate system when making serial measurements of A–P knee laxity using RSA because the improved precision allows for the observation of smaller differences.

Keywords: Roentgen stereophotogrammetric analysis; Precision; Laxity; Knee; Imaging; Anterior; Force

Introduction

Roentgen stereophotogrammetric analysis (RSA) is a research method that has been used to study the kinematics of human joints. RSA uses biplanar radiography to image radiopaque markers implanted in bone segments (i.e. bone markers). One application of RSA is the determination of anterior–posterior (A–P) knee laxity. A–P knee laxity is determined as the anterior

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component of the translation of the tibia relative to the femur resulting from posterior and anterior forces applied to the tibia. One primary advantage of RSA over other methods for determining A–P knee laxity is that the relative translation of the thigh and shank segments can be measured with high accuracy because errors resulting from attaching markers to soft tissue such as skin are avoided [14]. Because of the high accuracy of the method, RSA is attractive for use in longitudinal studies where changes in A–P knee laxity following reconstruction of the anterior cruciate ligament (ACL) are of interest. Documenting changes in A–P laxity may be useful for determining the effects of either surgical variables (e.g. graft type, tunnel placement, initial tensioning, and fixation methods) or rehabilitation regimens on the stability of reconstructed knees.

While the high accuracy of RSA makes the method attractive for measuring changes in A–P laxity in longitudinal studies, at the same time apparent changes in A–P knee laxity may result from variability of the measurement due to repeated testing. A recent animal study using RSA reported a 95% confidence interval of 0.23 mm in repeated measurements of A–P knee laxity [5], but the application of this value to human knees is unknown. Previous studies of A–P knee laxity in humans using RSA have reported precision values of 0.8 and 1.1 mm [6,15]. Because an increase in A–P knee laxity of 2–3 mm is considered clinically important [9,10,17,23], these precision values in humans are sufficiently poor to possibly obscure clinically important increases in knee laxity.

In determining A–P knee laxity using RSA, the precision may depend on the location of the origin of the tibial coordinate system used to describe the relative motion between the tibia and femur. In general, the motion of a rigid body can be described by the combination of translational and rotational displacements. For a base point or coordinate system origin in a body, the translation of this point depends on its location when the body undergoes both translation and rotation [19]. Because the knee flexion angle changes during the application of posterior and anterior shear loads [14], the A–P knee laxity will be affected by the rotation of the tibia if the origin of a local tibial coordinate system does not lie at the flexion–extension center-of-rotation. In which case the rotation of the tibia relative to the femur will introduce additional variability into repeated laxity measurements when changes in flexion angle are not consistent.

Previous studies of A–P knee laxity using RSA have placed the origin of the tibial coordinate system at the midpoint between the tips of the tibial-eminences [4–6,12,15]. However, the flexion–extension axis of the knee traverses the femoral condyles [2,8]. To minimize variability due to changes in knee flexion and hence variability in A–P knee laxity measurements, an alternative origin could be chosen on a line passing through the centers of the posterior femoral condyles which has been shown to closely approximate the flexion–extension axis of the knee [2]. Therefore, the objective of this study was to determine whether a tibial coordinate system with the origin placed midway between the centers of the posterior femoral condyles improves the precision of measuring A–P knee laxity with RSA compared to a tibial coordinate system whose origin lies at the midpoint between the tips of the tibial-eminences.

Materials and methods

Specimen preparation and experimental setup

Three human cadaveric legs (average age 63.7 years, range 38–79 years) were tested using RSA to measure A–P knee laxity. Prior to the study, both lateral and A–P radiographs were obtained for each specimen. Based on a preliminary analysis of the radiographs, these specimens were free from arthritis (as indicated by joint space narrowing) and other anomalies (e.g. osteophites).

Specimens were prepared for RSA by implanting radiopaque markers in the distal tibia and proximal femur. Six markers were placed in each distal tibia and proximal femur. The markers (0.8 mm diameter tantalum balls, Biomet Orthopedics Inc., Warsaw, IN) were implanted in the bones using a bead injector device (Tilly Medical Products AB, Lund, Sweden). Although a minimum of only three markers are required to describe three-dimensional rigid body motion, the use of additional markers created an overdetermined system and reduced the error in determining the position of each segment [21].

A custom loading apparatus was designed to apply posterior and anterior forces to the proximal tibia of the specimens (Fig. 1). The loading apparatus was constructed with pneumatic and electronic components. Pneumatically, the loading apparatus consisted of a portable air tank (model 688, Milton Industries Inc., Chicago, IL), manual pressure regulators (NAR 2060-N01, SMC Corporation of America, Indianapolis, IN), manual toggle valves (H11-30-44, Pneumadyne, Inc., Plymouth, MN), and a dual-acting pneumatic cylinder (RLF06A-DAN-AA00, Norgren, Inc., Littleton, CO). The pressure of the portable air tank was set to 0.34 MPa. The manual pressure regulators were capable of setting the pressure between 0.0 and 0.85 MPa to the toggle valves and pneumatic cylinder. These components allowed load magn-

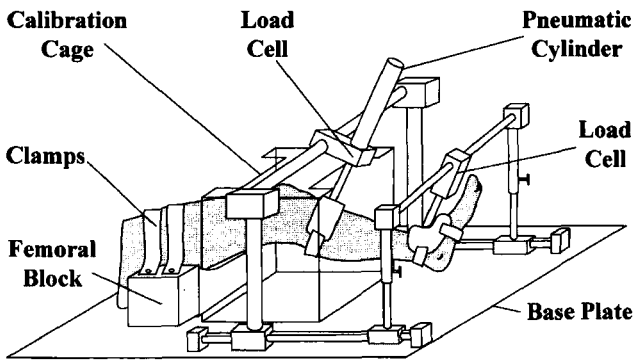

Fig. 1. Diagram of custom electro-pneumatic loading apparatus designed to apply posterior and anterior forces to the proximal tibia of human legs. Pneumatically, the system consisted of a portable air tank (not shown), manual pressure regulators (not shown), manual toggle valves (not shown), and a dual-acting pneumatic cylinder. Electronically, the system enabled monitoring of load cell outputs on a data acquisition computer (not shown).

itudes up to 388 N to be applied. An adjustable loading frame allowed for positioning of the legs. The loading frame supported the pneumatic cylinder and connected with the cylinder to the tibia via a contoured plastic cuff.

Two load cells rated at a maximum load of 222 N (L1650, FUTEK Advanced Sensor Technology, Irvine, CA) were used to monitor forces transmitted by the pneumatic cylinder at the tibia and at the ankle joint. One load cell was connected to the loading frame between the pneumatic cylinder and tibia while the other load cell was connected between the loading frame and ankle. The two load cells were used to standardize an A–P shear force transmitted at the knee (discussed subsequently). Load cell outputs were monitored on a computer running custom LabVIEW and National Instruments data acquisition software (version 4.1 and NI-DAQ, respectively, National Instruments Corporation, Austin, TX). The computer continuously displayed the force values calculated from the load cell outputs that were changed by altering either the manual pressure regulators or toggle valves.

The RSA system included a calibration cage and two portable X-ray machines. The calibration cage (Tilly Medical Products AB, Lund Sweden) contained markers at known positions to be used for system calibrations. Modifications to the calibration cage were made to hold two X-ray cassettes and two scatter grids (Medical X-ray Enterprises, Inc., Culver City, CA) placed at right angles to each other (A–P and lateral views). The scatter grids were used to minimize exposure of the radiographic film from scattered X-rays and thus improve image quality. The calibration cage surrounded the tibio-femoral joint such that all markers could be seen from each view. Each portable X-ray machine (model HF80, MinXray Inc., Northbrook, IL) was positioned at a distance between 85 and 110 cm from its respective film plane so that the direction of X-rays was approximately orthogonal to this plane. The X-ray machine parameters were initially set to 3 mAs and 80 kVp, and the exposure was adjusted by altering these parameters as needed depending on the bone density and thickness of the leg. Three additional markers were fixed in a line along the pneumatic actuator to enable the identification of the loading axis from the radiographs.

Specimens were placed in the loading apparatus using a standardized protocol to enable highly repeatable laxity measurements. Initially, the knees were placed within the RSA calibration cage and visually centered such that the joint line was in the view of the radiographs. Rotation about the long axis of the femur for each specimen was visually oriented such that the line passing through the femoral condyles was parallel to the surface of the base plate. The proximal thigh was securely clamped to the femoral block of the loading apparatus (Fig. 1). Knees were then placed in 25° of flexion, as determined by a goniometer that measured the angle between the midlines of two leg segments in the sagittal plane. Previous studies measuring anterior laxity have used knee flexion angles of 20–30° [3,6,7,11,12,15,20]. The loading frame at the tibia was adjusted and secured such that the axis of the pneumatic actuator was perpendicular to the anterior tibial surface approximately 8 cm distal to the joint line [6,15]. The loading frame at the ankle was adjusted such that the axis of the load cell was parallel to the loading axis at the tibia and the ankle was secured via an adjustable strap. Once the leg was secured in the loading apparatus, the positions of all loading frame components were measured so that the leg could be precisely repositioned in the loading apparatus for subsequent trials. Once positioned in the loading apparatus, simultaneous radiographs (A–P and lateral) were taken of the leg in this "neutral" position with no tibial load applied.

A–P knee laxity was measured under the application of standardized forces. The force at the knee joint (hereafter termed *A–P shear force transmitted at the knee*) was standardized instead of the force applied by the pneumatic cylinder because the A–P shear force transmitted at the knee determines the displacement of the tibia. A static model of the forces acting on the tibia (Fig. 2) was used to define the A–P shear force transmitted at the knee. A summation of forces acting in the A–P direction (i.e. x-direction) gives

$$P_{ix} + R_{ix} = A_{ix} + G_x \quad (1)$$

where $\boldsymbol{P_i}$ was the force applied by the pneumatic cylinder with i being either anterior or posterior depending on the loading direction, $\boldsymbol{R_i}$ was the reaction force at the ankle, $\boldsymbol{A_i}$ was the reaction force at the knee, and $\boldsymbol{G}$ was the force due to the weight of the specimen. The values for A_{ix} and G_x were unknown in this experimental setup while P_{ix} and R_{ix} were measured from the load cells. Because the weight of the tibia for each specimen was unknown, the value of G_x was estimated using anthropometric data for human males and was equal to 46 N [18]. The value of A_{ix} was then estimated from the sum of moments about the point $\boldsymbol{O}$ using either an 89 N posterior force or a 134 N anterior force as the value for P_{ix} [4] and anthropometric data for human males to determine the distances between forces [18]. These calculations resulted in a −90 N posterior shear force and a +90 N anterior shear force for A_{ix}. Similar calculations from female anthropometric data gave A_{ix} values of −86 N and +92 N for posterior and anterior shear forces respectively, and were considered similar enough to justify the sole use of male anthropometric data. Thus, using our model, the sum of A_{ix} and G_x was equal to −44 N for the posterior applied force and +136 N for the anterior applied force. Accordingly, the sum of the two load cell values monitored during data collection, which also equalled the sum of A_{ix} and G_x in the model, was standardized to −44 N for the posterior applied force and +136 N for the anterior applied force. The magnitude of load used in this study was similar to those magnitudes used previously to study A–P knee laxity via RSA [6,11,12,15].

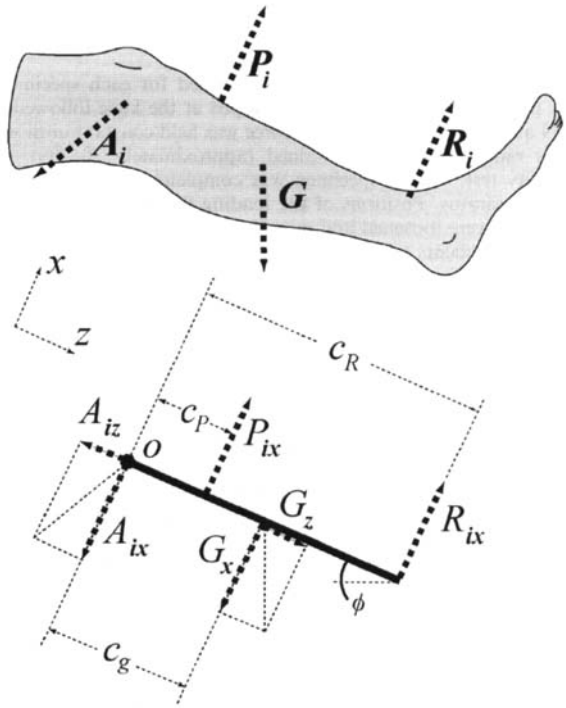

Fig. 2. Static model of the loads acting on the tibia used to define the A–P shear load transmitted at the knee. In this model, positive x is in an anterior direction and positive z is in a distal direction. Model variables were as follows: $\boldsymbol{P_i}$ is the load applied by the pneumatic cylinder where i is either anterior or posterior, depending on the loading direction; $\boldsymbol{R_i}$ is the reaction load at the ankle; $\boldsymbol{A_i}$ is the reaction force at the knee; A_{ix} and A_{iz} are components of the reaction force at the knee; G_x and G_z are components of the load due to gravity; c_p, c_g, and c_R are the distances between points of load application; and ϕ is the angle of the tibia with respect to the ground. The A–P shear load transmitted at the knee was defined as the summation of $\boldsymbol{P_i}$ and $\boldsymbol{R_i}$ (i.e. a summation of the two load cell values).

Each specimen was preconditioned prior to A–P knee laxity tests. A–P shear forces in both posterior and anterior directions were cyclically developed alternating 10 times in each direction with magnitudes of −44 N and +136 N respectively. The elapsed time during and between each force was approximately 2 s. The manual pressure regulators were adjusted during each force application such that the computer output matched the desired magnitudes. The specimens reached a steady-state response (i.e. no adjustment of the pressure

regulators was needed) after 7–8 loading cycles. The repeatability of the applied force was important because tibial displacement is influenced by the magnitude of the force applied [3].

Six A–P knee laxity tests were performed for each specimen. A −44 N posterior shear force was developed at the knee followed by a +136 N anterior shear force. Each force was held constant until simultaneous radiographs were obtained (approximately 20–30 s). After each laxity test, the leg specimen was completely removed from the loading apparatus. Positions of the loading frames and other system components were loosened and moved, and then repositioned according to measurements taken after the initial placement of the leg specimen. The leg specimens were then replaced in the loading apparatus. The preconditioning cycle was repeated immediately preceding each of the subsequent laxity tests.

Data analysis and laxity calculations

Analysis of the radiographs was performed using a customized RSA data analysis system. A digital image was obtained for each radiograph using a back-lit scanner (Epson 1600, Epson America Inc., Long Beach, CA). The radiographs were scanned at a resolution of 300 dpi. The appearances of the digital images were modified using contrast and threshold controls from image editing software (Photoshop 5.0, Adobe Systems Inc., San Jose, CA) to optimize the identification of the markers in the image. The two-dimensional centroid coordinates of the markers were measured from the digital image using a software program (Scion Image 1.0, Scion Corporation, Frederick, MD).

A customized computer program written in Matlab (version 6.0, The Mathworks Inc., Natick, MA) was used to determine A–P knee laxity. This program initially computed the transformation of image coordinates to the calibration cage, the positions of the Roentgen foci, and the three-dimensional position coordinates of all the markers in a laboratory coordinate system defined by the calibration cage [19]. The program also computed the calibration error for the markers in the calibration cage and the radial error for all object points (i.e. "fictive" points and implanted markers not in the calibration cage) defined by the mutually perpendicular distance between crossing lines used to reconstruct the points in space. A value of 0.2 mm was used as a maximum acceptable value for the radial error.

Two tibial coordinate systems were established using anatomical landmarks seen from the A–P and lateral radiographs with the leg in the neutral position. The first tibial coordinate system (hereafter termed *tibial-eminence coordinate system*) was established using the midpoint of the line connecting the tips of the tibial-eminences as the origin [4,5]. For the second tibial coordinate system (herein termed *center-of-rotation coordinate system*), the origin was the midpoint of the line passing through the centers of the posterior femoral condyles (Fig. 3). From the lateral view, a circle was fit to the contour of each posterior edge of the lateral and medial femoral condyles using a software program (Scion Image 1.0). The two-dimensional coordinates of the center of each circle were determined and the origin was defined from the lateral view as the midpoint between the centers (O_L). From the A–P view, the long axis of the femur ($L_{A/B}$) was drawn through the midpoint (A) between the medial and lateral cortices at the proximal section of bone, which was located at a height above the tibial plateau approximately equal to the epicondylar width, and the peak of the intercondylar notch (B). A line traversing the articular surfaces of the tibial plateau (L_P) also was drawn. The origin (O_{AP}) was defined from the A–P view as the point on $L_{A/B}$ at the same height above the articular surface as the origin O_L from the lateral view. To account for differences in imaging parameters (e.g. magnification) between the two views, the position of O_{AP} along the long axis of the femur was adjusted in the A–P view to achieve a radial error within the acceptable limit (0.2 mm). The axes of both tibial coordinate systems were defined from the line formed by markers implanted in the housing of the pneumatic actuator ($+x$ was anterior), the cross product of x and the line connecting the tips of the tibial-eminences ($+z$ was distal), and the cross product of z and x ($+y$ was lateral for a right knee).

A–P knee laxity was determined in a sequence of steps. First, motion of the tibia and femur for each trial was determined relative to the fixed laboratory coordinate system defined by the calibration cage. Specifically, tibia and femur rigid body motion transformations from a posterior position to an anterior position were obtained using the markers and a least-squares algorithm [4,5,22]. Second, motion of

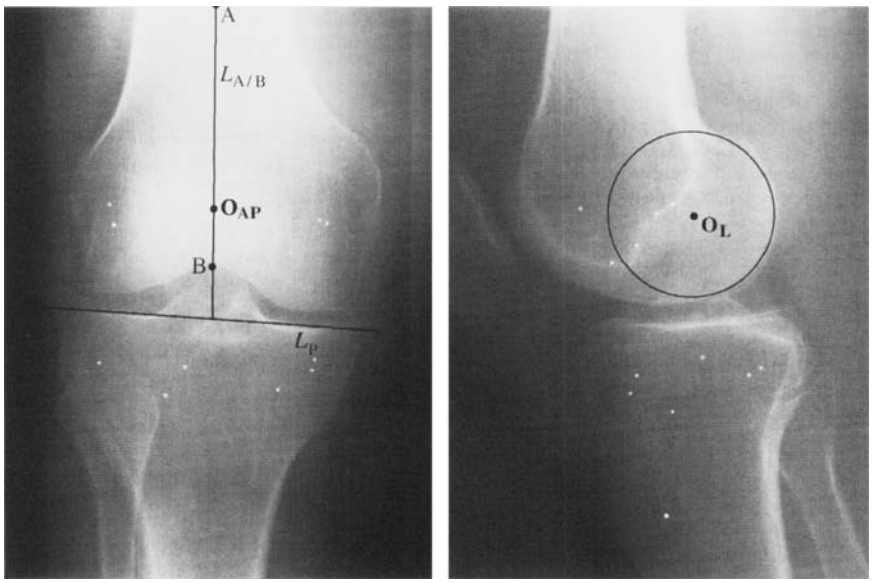

Fig. 3. A–P and lateral radiographs of a human knee joint showing the method used to locate the origin for the center-of-rotation coordinate system. From the lateral view, a circle was fit to the contour of the posterior edge of both the medial and lateral condyles (only the circle for the lateral condyle is shown for clarity). The location of the origin from the lateral view (O_L) was defined as the midpoint between the centers of the two circles. From the A–P view, $L_{A/B}$ is the line connecting the midpoint of the proximal femur (A) to the intercondylar notch (B) and L_P is the line traversing the articular surfaces of the tibia. The location of the origin in the A–P view (O_{AP}) was the point on $L_{A/B}$ at the same height above the articular surface as O_L. Also visible are the tantalum markers implanted in the distal tibia and proximal femur.

the tibia was described relative to the femur using one of the two anatomically-defined tibial coordinate systems. The "fictive" points defining the origin of each respective coordinate system were determined once from radiographs of each knee in the neutral position and were located in all other radiographs using motion transformations and the positions of the tantalum markers in the tibia. The complete relative motion between the tibia and femur was computed including the linear displacement of the tibial coordinate system origin and the rotational displacement of the tibial coordinate system. A–P knee laxity was defined as the anterior component of linear displacement (along the x-axis). Additionally, the change in knee flexion angle was defined as the rotation about the y-axis.

A statistical test was used to determine whether the precision of A–P knee laxity measurements using the tibial-eminence and center-of-rotation coordinate systems was different. A–P knee laxity precision, defined as the standard deviation of repeated measurements, was computed for each coordinate system using a pooled variance defined as

$$s^2_{\text{pooled}} = \frac{\sum_{j=1}^{3} s^2_{\text{subject } j}}{3} \tag{2}$$

where $s^2_{\text{subject } j}$ is the sample variance for the jth subject. The A–P knee laxity precisions, computed using each coordinate system, were compared using an F-test. The computed value of the F-variate was given by

$$f = \frac{s^2_{\text{tibial-eminence pooled}}}{s^2_{\text{center-of-rotation pooled}}} \tag{3}$$

where $s^2_{\text{tibial-eminence pooled}}$ was the pooled specimen variance for the tibial-eminence coordinate system and $s^2_{\text{center-of-rotation pooled}}$ was the pooled variance for the center-of-rotation coordinate system.

Because the precision of the change in knee flexion angle that occurred between application of a posterior and anterior force can affect the precision of A–P knee laxity measurements, this value was also computed. The precision was determined for each specimen as the standard deviation of the rotation about the y-axis, computed over the six trials, and the overall precision was computed from the pooled variance as was done previously for the laxity measurements using Eq. (2).

Results

The precision of A–P knee laxity measurements using the center-of-rotation coordinate system was significantly less than the precision using the tibial-eminence coordinate system ($p = 0.006$, Table 1). The precision using the center-of-rotation coordinate system was 0.33 mm and was 0.29 mm smaller than the precision using the tibial-eminence coordinate system. This difference indicates nearly a factor of 2 improvement in the precision of A–P knee laxity measurements when the center-of-rotation coordinate system was used.

Table 1
Mean and standard deviation of A–P knee laxity measurements for each tibial coordinate system and the change in knee flexion angle that occurred between the application of posterior and anterior forces

Specimen number	A–P knee laxity (mm)		Change in knee flexion angle (°)
	Tibial-eminence coordinate system	Center-of-rotation coordinate system	
1	7.75 ± 0.61	11.24 ± 0.30	14.5 ± 3.6
2	1.61 ± 0.63	8.16 ± 0.31	23.6 ± 2.2
3	0.32 ± 0.62	4.17 ± 0.37	13.2 ± 1.2
Overall	3.22 ± 0.62	7.85 ± 0.33	17.1 ± 2.5

A significant difference was found between the pooled knee laxity precisions ($p = 0.006$).

The knee flexion angle changed between the application of posterior and anterior forces. On average, the knee flexed 17.1° when moving from a position of posterior applied force to a position of anterior applied force (Table 1). The precision of the change in flexion angle varied for each specimen between 1.2° and 3.6°. The overall precision of the change in flexion angle was 2.5°.

Discussion

The objective of this study was to determine whether the center-of-rotation coordinate system whose origin was located approximately at the flexion–extension center-of-rotation of the knee improved the precision of A–P knee laxity measurements compared to the tibial-eminence coordinate system. The key finding of our study was that the precision of A–P knee laxity improved by nearly a factor of 2 for the center-of-rotation coordinate system.

One usefulness of our key finding is that smaller changes in A–P knee laxity can be detected in a single subject being followed in a longitudinal study. For a precision of 0.33 mm, an increase of either 0.56 mm (i.e. 1.7 standard deviations) or greater in A–P knee laxity observed over time would be considered a statistically significant increase rather than an increase due to measurement error because the probability of this value occurring in the absence of any increase is 0.05 or less. Thus when other factors contributing to variability are controlled as was done in this study, relatively small increases in laxity can be detected.

Another usefulness of our key finding is that the improved precision when using the center-of-rotation coordinate system allows for a smaller number of subjects in a study which involves testing a subject sample to draw more general conclusions regarding the population. To demonstrate this, the sample size n can be found from [1]:

$$n = (z_\alpha + z_\beta)^2 \times (\sigma/\delta)^2 \tag{4}$$

where z_α and z_β are the values of the Z-variate for the probabilities of Type I (i.e. α) and Type II (i.e. β) errors respectively, σ is the standard deviation (i.e. precision) of the measurement, and δ is the increase in anterior laxity to be detected. For a given desired difference to detect and probabilities of Type I and Type II errors, the sample size would decrease by a factor of 4 for the center-of-rotation coordinate system.

Because the A–P laxity for the two tibial coordinate systems was computed in the same specimen in the same experiment, the observed difference in precision is due solely to the different location of the origin for each coordinate system. Because knee flexion changes during

the application of anterior and posterior forces as documented previously [14] and as demonstrated in this study (Table 1) and because the translation of the coordinate system origin is affected by these flexion changes [19], these changes in knee flexion contribute to A–P laxity. Accordingly any variability in the changes in knee flexion introduces an additional source of variability into the determination of A–P laxity. From the results presented herein where the precision for the tibial-eminence coordinate system was a factor of 2 greater than that for the center-of-rotation coordinate system, it can be inferred that the variability in A–P laxity from the variability in knee flexion was approximately equal to the variability in the A–P laxity from the variability in translation.

Although A–P knee laxity differed between the two coordinate systems, the results were consistent within specimens (Table 1). The rank of specimens (i.e. greatest to least) according to laxity was independent of the RSA technique used. When calculated using the center-of-rotation coordinate system, the A–P knee laxity values were within the range of values previously reported in studies using RSA [4,13,14]. The difference in magnitude between coordinate systems can be explained by the negative bias that occurs as a result of the increased flexion of the knee between the application of posterior and anterior forces (Fig. 4).

In this study, the shear force transmitted at the knee was standardized rather than the load applied to the tibia. This standardization method was preferred because the shear force transmitted at the knee may vary depending on the placement of the load actuator on the tibia. The loading method used by this study provided a repeatable shear force transmitted at the knee as evidenced from the high precision that was achieved. Although differences in the magnitude of shear force transmitted at the knee may exist between specimens, it is more important to standardize the shear force transmitted at the knee in each individual for the purposes of documenting changes in A–P knee laxity in a longitudinal study.

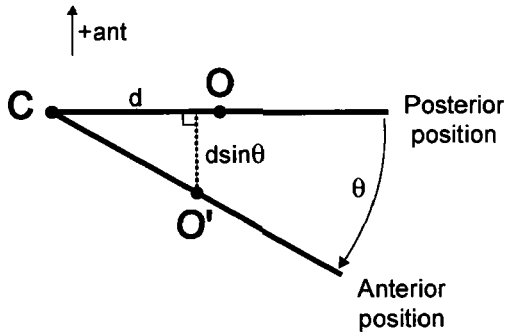

Fig. 4. Simplified model showing the negative bias of A–P knee laxity measurements as a result of the increased knee flexion when the chosen tibial origin is distal to the actual flexion–extension center-of-rotation. During the A–P knee laxity tests, the knee flexed between application of a posterior force (posterior position of tibia) and anterior force (anterior position of tibia). Thus the tibia rotated with respect to the femur in the direction shown. This model represents the case when the tibia rotates but does not translate, where C is the center-of-rotation, O is the chosen tibial origin, d is the distance of the chosen origin from the center-of-rotation, and θ is the change in flexion. The relative position of the tibia (as described by the motion of its origin, from O to O′) appears to translate in the posterior direction by an amount equal to $d \sin \theta$.

While the results of this study are intended to be applicable to longitudinal studies which involve the use of live patients, they are limited in that the measurements were not made in vivo. To obtain the most complete estimate of precision, our in vitro model included many sources of variability, such as variability due to differences in load magnitude, load direction, location of load application, flexion angle, and preconditioning of the knee. However, this study did not include variability due to muscle contraction, as might be expected to influence repeatability when measuring in vivo A–P knee laxity. Because muscle contraction may affect A–P knee laxity [16], it is important to minimize the effect of this variable as much as possible to achieve the best precision (e.g. through subject monitoring and minimization of EMG activity using visual computer feedback).

Although the results of this study are intended to be used toward an in vivo study documenting changes in the A–P knee laxity of ACL-reconstructed knees, the precision was obtained from knees with an intact ACL. Because the kinematic behavior of the ACL-reconstructed knee may be different than that of the ACL-intact knee, the question arises as to whether the same precision would be obtained for ACL-reconstructed knees. One previous study documented small differences (less than 5°) in flexion–extension during anterior and posterior loading between ACL-intact and ACL-deficient knees [15]. Accordingly the precision determined in this study should apply equally well to ACL-reconstructed knees.

In summary, our study has demonstrated that the precision in repeated measurements of A–P laxity depends on the location of the origin of the tibial coordinate system and that a new center-of-rotation coordinate system improved the precision by a factor of 2 over that of a previously used tibial-eminence coordinate system. One important consequence of these results is that smaller differences in A–P laxity can be detected as statistically significant in longitudinal studies that require repeated measurements of A–P laxity.

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