Table 3. Total Amount of Tibiofemoral Motion for 22 Images in Three Subjects (Maximum Value–Minimum Value)

Tibiofemoral Kinematics	Subject 1	Subject 2	Subject 3	Overall
Anterior–posterior translation (mm)	7.4	11.2	12.2	10.3
Superior–inferior translation (mm)	5.3	3.2	7.2	5.2
Medial–lateral translation (mm)	4.7	6.3	8.4	6.5
Flexion–extension (°)	138.4	115.8	141.5	131.9
Internal–external rotation (°)	32.76	26.48	37.70	32.31
Varus–valgus (°)	6.03	5.54	8.07	6.55

Translations were measured as the femoral coordinate origin moving with respect to the tibial coordinate origin. The amount of motion corresponds to 22 frames of data, which do not necessarily include the entire range of squat motion from full extension to full flexion.

Comparison of the bone models derived from the same subject using CT and MRI showed areas where the surfaces differed by several millimeters (Fig. 2). Several factors probably contributed to these shape differences, which result in different shape-matching performance with the CT– and MRI–derived bone models. First, the fact that different shapes are obtained from the CT and MRI scans introduces bias placing the coordinate systems in the two models. These slight offsets in coordinate system origin and orientation result directly in bias when comparing the measurements from the two models, slightly reducing the ability to isolate differences solely due to bone reconstruction fidelity. Second, bone boundaries identified in CT result directly from X-ray projections, while bone boundaries in MRI result from different physical properties. This consideration is particularly relevant at the distal femur and proximal tibia, where articular regions and ligament insertions present structures with graded properties, where the boundaries are likely to differ between CT and MRI modalities. Thus, we should expect that bone models derived from CT scans will provide superior correspondence when used for shape matching with radiographic projections.

Shape matching with in vivo images showed significant RMS differences comparing kinematics from CT– and MRI–derived bone models (Table 1). When matching the in vivo images with CT–derived bone models, no visible discrepancy was noted between the bone edges in the image and the superimposed edges of the model. With the MRI–derived models, small discrepancies between image and model edges were visible after pose optimization in most cases. Kinematics measured with synthetic X-ray projections uniformly showed less bias and better precision when CT–derived bone models were used (Table 2). Because the synthetic images were created using the CT–derived models, the accuracy and precision figures represent an absolute best-case measurement performance for similar projection geometries using the nonlinear least squares optimization method. The RMS errors figures with the MRI–derived models represent the lower boundary of measurement error one might expect using models based on different physical properties.

When using MRI to create bone models, each MRI scanner will perform differently. For this study, a 0.3 T scanner was used with a gradient echo sequence, and this provided images with sufficient bone/soft tissue contrast to identify the bone boundaries. The gradient echo sequence was used to achieve good resolution for bone segmentation,[4] but spin echo sequences are better for spatial distortion if the contrast is sufficient to detect bone boundaries.[18] Distortion increases with higher magnetic fields.[10,11,17] Higher magnetic fields increase signal intensity for better tissue resolution, but chemical shift and susceptibility artifacts also contribute to geometric distortion. Smaller magnetic fields permit narrower signal bandwidths and consequent reductions in noise. Magnetic field inhomogeneity is another source of geometric distortion that decreases with decreasing magnetic field strength.

Magnetic field inhomogeneity depends on more than field strength, being a function of materials and their spatial distribution within the object being scanned. In biological tissues, MRI signals are generated by hydrogen atoms, with water and fat content accounting for the majority of the signal. All soft tissues and cancellous bone contain a large fraction of water, so the magnetic susceptibility can be approximated by that of water. In contrast, cortical bone and air do not generate significant MRI signals. Nevertheless cortical bone can distort magnetic fields in nearby tissues that do generate

MRI signals, thereby resulting in geometric distortion near these interfaces.[16]

Finally, small motions of the patient during scanning can degrade boundary resolution and spatial integrity of the resulting models. This is of particular concern when sequences requiring long scan times are used, when the anatomy of interest is affected by normal breathing movements, and when immobilization of the area is not easily accomplished. We took great care to reduce motion artifacts while subjects were being scanned, yet it is likely small motion artifacts affected the shape of the MRI-derived bone models. Investigators should attend carefully to positioning and immobilization of subjects to produce high fidelity bone models with MRI.

Useful kinematic measurements can be obtained from single-plane fluoroscopy and shape matching using bone models derived from CT or MRI. Because the fidelity of MRI-derived bone models is degraded by a variety of technical and practical factors, shape-matching results typically will be inferior to those obtained with CT-derived bone models. However, many clinical and research situations exist in which bone model creation using MRI is highly desirable. In these cases, investigators should maintain keen awareness of the factors influencing the fidelity of bone models, and they should incorporate these technical limitations into the interpretation of their findings. Carefully done and cautiously interpreted, we should be able to expand the range of useful kinematic observations using MRI-derived bone models.

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