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Validation of Imaging-Based Quantification of Glenohumeral Joint Kinematics Using an Unmodified Clinical Biplane Fluoroscopy System

Joseph D. Mozingo^{1,2}, Mohsen Akbari Shandiz, Ph.D.², Felicia M. Marquez², Beth A. Schueler, Ph.D.³, David R. Holmes, Ph.D.⁴, Cynthia H. McCollough, Ph.D.³, and Kristin D. Zhao, Ph.D.²

¹Biomedical Engineering and Physiology Graduate Program, Mayo Clinic Graduate School of Biomedical Sciences, Mayo Clinic, Rochester, MN, USA

²Department of Physical Medicine and Rehabilitation, Mayo Clinic, Rochester, MN, USA

³Department of Radiology, Mayo Clinic, Rochester, MN, USA

⁴Department of Physiology and Biomedical Engineering, Mayo Clinic, Rochester, MN, USA

Abstract

Model-based tracking, using CT and biplane fluoroscopy, allows highly accurate quantification of glenohumeral motion and changes in the subacromial space. Previous investigators have used custom-built biplane fluoroscopes designed specifically for kinematic applications, which are available at few institutions and require FDA approval prior to clinical use. The aim of this study was to demonstrate the utility of an off-the-shelf clinical biplane fluoroscope for kinematic applications by validating model-based tracking for measurement of glenohumeral motion using an unmodified clinical system. Biplane images of each shoulder of a cadaver torso were acquired at various joint positions and during simulated movements along anatomical planes of motion. The pose of each humerus and scapula was determined using model-based tracking and compared to a bead-based gold standard. Error due to a temporal-offset between corresponding biplane images, characteristic of clinical biplane systems, was determined by comparison of measured and known relative position of 2 bead clusters of a phantom that was imaged while moved throughout the fluoroscopy image volume. Model-based tracking had global kinematic mean absolute errors of 0.27 mm and 0.29° (static), and 0.22 – 0.32 mm and 0.12 – 0.45° (dynamic). Glenohumeral mean absolute errors were 0.39 mm and 0.45° (static), and 0.36 – 0.42 mm and 0.41 – 0.48° (dynamic). The temporal-offset was predicted to add errors of 0.06 – 0.85 mm and 0.05 – 0.28° for cadaveric trials for the speeds examined. For defined speeds, sub-millimeter and sub-degree kinematic

*Corresponding Author: Kristin D. Zhao, Ph.D., Assistive and Restorative Technology Laboratory, Rehabilitation Medicine Research Center, Department of Physical Medicine and Rehabilitation, Mayo Clinic, Rochester, MN 55905, Phone: 507-284-8942, zhao.kristin@mayo.edu.

Conflict of interest statement

All authors declare no conflict of interest.

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accuracy and precision were achieved using an unmodified clinical biplane fluoroscope for quantification of glenohumeral motion.

Keywords

Shoulder; Kinematics; Fluoroscopy; Clinical Biplane; Computed Tomography; Radiostereometric Analysis; Model-based tracking; Accuracy; Validation

1. Introduction

Glenohumeral and scapular kinematics are key extrinsic factors that contribute to a narrowing of the subacromial space and resultant shoulder pain secondary to mechanical impingement of underlying structures (Braman et al., 2014; Mackenzie et al., 2015). It is suggested that a 1 – 2 mm change in the relative position of the scapula and humerus may constitute as great as a 67% reduction in the overall subacromial space over the range of glenohumeral elevation commonly associated with impingement (Bey et al., 2007). In fact, from 55 – 70° of glenohumeral elevation, a difference as small as 1 – 1.3 mm in the size of the subacromial space between individuals with and without rotator cuff pathology has been shown to be statistically significant (Baumer et al., 2016). Thus, a technique capable of providing sub-millimeter and sub-degree relative kinematic accuracy is requisite for detecting clinically relevant changes in the subacromial space and further understanding the biomechanical basis of rotator cuff damage.

Model-based tracking, using computed tomography (CT) and biplane fluoroscopy, has become the gold standard for *noninvasive in vivo* quantification of bony motion. Despite the utility of this technique as a research tool, it is not yet used routinely in clinical practice. Investigators have achieved sub-millimeter and sub-degree kinematic accuracy in applications of this technique to the shoulder, but have done so using custom-built biplane fluoroscopy systems or two modified single-plane C-arm systems, which are not widely available and require FDA-approval prior to clinical use (Bey et al., 2006; Massimini et al., 2011; Zhu et al., 2012; Giphart et al., 2012). Clinical biplane fluoroscopy systems are an alternative that could extend the reach of this technique due to their availability at many medical centers. Furthermore, such systems do not require modification prior to use, are FDA-approved for clinical use, and are ideally situated for patient imaging.

A characteristic trait of clinical biplane systems is alternating pulsing of the 2 x-ray sources (Bushberg et al., 2012), which is not heavily advertised by manufacturers likely due to the fact that its effect is not perceivable when used for its primary application of surgical visualization. Due to this property, corresponding biplane images are acquired with a temporal offset that is equal to approximately half the inverse of the frame rate. While the offset preserves image quality by reducing cross-scatter (Bushberg et al., 2012; Cross et al., 2015), object motion within this interval hinders convergence of model-based tracking, which assumes synchrony of acquired images. To reduce this effect, the object can be imaged in static positions or at movement speeds for which accurate registration is still possible. The maximum image capture rate of such systems, which is typically 30 – 60 frames/s, also dictates the movement speeds that may be assessed using these systems. For

these reasons, it is critical to characterize the performance of a clinical biplane system across the range of movement speeds that will be used in future studies.

The purpose of this work was to demonstrate the accuracy and feasibility of a standard unmodified clinical biplane fluoroscope for glenohumeral kinematic measurements in order to accelerate translation of this technique to clinical practice. To this end, the study objectives were: (1) To characterize the performance of Radiostereometric Analysis (RSA) (Selvik, 1989; Hallström and Kärrholm, 2006) and verify the sufficiency of this technique as a kinematic gold standard for validation of model-based tracking; (2) To determine the accuracy and precision of model-based tracking for quantification of glenohumeral motion and establish the acceptable range of speeds for which it may be used for a given imaging setup and protocol. First, a phantom experiment was conducted in order to quantify the error in RSA-based measures with (dynamic) and without (static) a temporal-offset between corresponding biplane images. Next, the pose of each humerus and scapula of a cadaveric specimen was determined using model-based tracking and compared to RSA for a series of joint positions and simulated motions. This approach was deemed accurate contingent on meeting the following criteria: (1) Error in RSA of one or more orders of magnitude less than that anticipated for model-based tracking; (2) Sub-millimeter and sub-degree relative kinematic accuracy of model-based tracking. The feasibility of this approach was determined by defining the range of speeds for which the total relative kinematic accuracy (model-based tracking error + temporal-offset induced error) was sub-millimeter and sub-degree.

2. Methods

2.1 Cadaver Prep and Imaging

With approval from our institutional biospecimens committee, 1 intact fresh-frozen cadaver torso (sex: male, age: 82 yr.) was obtained from our anatomical bequest program. A minimum of 4 radio-opaque beads (tantalum, 1.0 mm diameter) were implanted into both the humerus and scapula of the left and right shoulders. The specimen was imaged with a clinical CT scanner (128-slice SOMATOM Definition Flash; Siemens Healthcare) and clinically used musculoskeletal shoulder protocol at 120 kVp and 250 quality reference mAs, a 200–290 mm FOV (512×512 image matrix), 0.75 mm slice thickness, and 0.35 mm slice increment. 3D surface models of each humerus and scapula were created by segmentation (AnalyzePro; Mayo Clinic) (Robb et al., 1989).

Next, the glenohumeral joint was centered in the image volume of a clinical biplane fluoroscope (30×40 cm flat-panel detectors; 960×1240 pixel image size) (Artis zee biplane; Siemens Healthcare). The specimen was seated upright and positioned as shown (Fig. 1). The fluoroscope used in this study is a true integrated biplane system designed for orthogonal neuro-angiographic imaging. Imaging was performed at a detector entrance air kerma of 1820 nGy/frame using a 48 cm field of view. Pulse width was minimized in order to reduce image blur (3.2 ms). Images were acquired at 15 frames/s per plane (representative images shown in Fig. 1). For the specified parameters, the temporal-offset between pulses from each x-ray source, and therefore corresponding biplane images, was consistently 33.3 ms (half the inverse of the frame rate), confirmed via an oscilloscope connected to the x-ray

generator. Tube voltage and current were dynamically modulated by automatic exposure rate control, with trial averages that ranged between 70–100 kVp and 255–370 mA. Total effective dose for the dynamic trials was calculated as 1.4 mSv using a Monte Carlo simulation (PCXMC; STUK) (Tapiovaara et al., 1997); less than half the average annual dose to an individual living in the U.S. that comes from naturally occurring sources (Broga, 2009).

Following data collection, a calibration cube was imaged with the cube centered in the image volume in order to determine the position of the x-ray sources and detectors and establish the global coordinate system. The cube was comprised of 4 evenly spaced interconnected acrylic sheets (232×232×4.5 mm sheets), each embedded with a 4×4 grid of 3.0 mm diameter steel spheres (machining tolerance ± 0.005 mm).

2.2 RSA: Validation and Temporal-Offset Error

Static images of the calibration cube, which here functioned as a phantom, were obtained using the fluoroscopy setup and settings previously described, with the phantom positioned at various locations throughout the image volume. 3D bead positions were determined by RSA (XMA Lab; Brown University) (Knorlein et al., 2016). Inter-bead distance was measured for bead pairs that were visible in both image planes (190 image pairs). The mean of the absolute difference between the measured and machined distance and the standard deviation of the absolute differences were quantified and used as metrics of RSA accuracy and precision respectively, similar to Tashman and Anderst (2003) and Brainerd et al. (2010).

Precision was also determined *in situ* using 3D bead positions measured from the static and dynamic cadaveric trials. Specifically, pairwise inter-bead distance was measured for 3 beads within the humerus for each shoulder. The angles between the legs of the triangle formed by the vectors joining the beads were also quantified (Tashman and Anderst, 2003). Provided the beads are rigidly fixed, the measured distances and angles should remain constant. The standard deviation of the measured inter-bead distances and angles was determined for all static positions (45 image pairs) and for each dynamic trial (6 trials, 340 image pairs).

To measure temporal-offset error, dynamic images of the phantom were acquired by suspending and releasing it from a wound cord, allowing it to rotate and translate within the image volume, thus providing a range of speeds (angular velocities from 0.5 – 140°/s) in order to determine an upper limit for the cadaveric trials. Coordinate systems were assigned to 2 non-coplanar bead clusters comprised of 3 beads each. The relative translation and rotation (XZ'Y'' Euler sequence) of the coordinate systems was determined in 250 dynamic frames. The absolute difference between each measured and machined relative position was quantified, and the relationship between the absolute difference and object velocity was determined via simple linear regression.

2.3 Model-Based Tracking Validation

Dynamic scapular-plane elevation, sagittal-plane elevation, and internal/external rotation were simulated by manually manipulating the specimen's arms. Scapular-plane elevation

started in neutral anatomical position and ended at maximum achievable glenohumeral elevation. Sagittal-plane elevation started with the distal humerus in maximum extension, then brought to maximum flexion, and returned to maximum extension. Internal/External rotation was performed in the scapular plane and started with the humerus in maximum external rotation, then rotated to maximum internal rotation, and returned to maximum external rotation. Average trial duration was 3.75 s. Average angular velocity of individual trials ranged from 12 – 35°/s, with a combined trial average of approximately 20°/s and maximum frame-to-frame velocity of 60°/s. One trial of each simulated motion was analyzed per shoulder, yielding a combined 6 trials. 125 total dynamic image pairs were analyzed based on an increment that corresponded to 5% of the total number of frames for a given trial. Additionally, 20–25 static positions comprising the full range of motion were imaged and analyzed per shoulder (45 total image pairs) in order to assess agreement with no temporal-offset.

For each bone, 3D coordinates of at least 3 anatomical landmarks and the center of 3 implanted beads were determined directly from the CT volume and used to construct bone-based (Wu et al., 2005) and bead-based local coordinate systems. To quantify 6 degree-of-freedom (DOF) kinematics, the humerus and scapula bone models were each registered to fluoroscopic images pairs (corresponding biplane images offset by 33.3 ms) (Autoscooper; Brown University) (Miranda et al., 2011). The 3D pose of the bead-based coordinate systems was determined from the same images via RSA. The relative transformation from bead- to bone- based systems was determined from the CT volume and used to transform each individual pose of the bead-based coordinate systems measured from RSA, enabling direct comparison between techniques.

For a given trial, the accuracy and precision of model-based tracking were evaluated by computing: (1) The mean of the absolute difference, as well as the root-mean-square error, of all global bone positions measured by model-based tracking compared to those measured by RSA; (2) The standard deviation of the absolute difference of the aforementioned position measures. Relative (glenohumeral) kinematic parameters were also computed and compared using an XZ'Y'' Euler sequence (glenohumeral angle of elevation, angle of horizontal adduction/abduction, and axial rotation) (Phadke et al., 2011).

3. Results

The accuracy and precision of RSA under static conditions were 0.05 ± 0.04 mm (Table 1). *In situ* precision of RSA was 0.06 mm and 0.14° for inter-bead distance and inscribed angle measures respectively (Table 2). For simulated motions, precision was as great as 0.40 mm and 0.84° (Table 2). For dynamic phantom data, which incorporates error associated with a temporal-offset between biplane images, the maximum absolute error (AE) in RSA relative kinematic measures was 1.11 mm and 0.35° for the range of speeds measured *in situ* (Fig. 2, data points enclosed by dashed lines). The coefficients of determination for the regression of relative kinematic error on angular velocity were 0.80 (translation) and 0.91 (rotation) (Fig. 2, inset equations).

The static kinematic accuracy and precision of model-based tracking were sub-millimeter and sub-degree, independent of bone and DOF. Root-mean-square error (RMSE), Mean AE (MAE), and standard deviation (SD) did not exceed 0.55 mm/0.60°, 0.43 mm/0.49° and 0.36 mm/0.33° respectively (Table 3). Largest differences between bones were in rotation, with MAE (averaged across DOFs) and SD of $0.15 \pm 0.12^\circ$ (scapula) and $0.42 \pm 0.27^\circ$ (humerus). Under dynamic conditions, RMSE, MAE, and SD did not exceed 0.52 mm/0.58°, 0.42 mm/0.48° and 0.32 mm/0.33° respectively across motions (Table 4). Global MAE and SD incorporating both bones ranged from 0.26 – 0.28 mm/0.26 – 0.29° (MAE) and 0.19 – 0.21 mm/0.23 – 0.26° (SD) for the different simulated motions.

4. Discussion

The first objective of this study was to characterize the performance of RSA and to verify the sufficiency of this technique as a kinematic gold standard for validation of model-based tracking. Accuracy and precision of RSA were 0.05 mm and 0.04 mm respectively (Table 1), representing the uncertainty in bead-based distance measures caused by factors other than the temporal-offset. RSA *relative kinematic* errors are predicted to be 0.05 mm and 0.04° (Fig. 2, inset equations, velocity set equal to 0), which are one order of magnitude below biplane model-based tracking errors reported in the literature, as expected (Bey et al., 2006; Massimini et al., 2011; Zhu et al., 2012; Giphart et al., 2012). Precision was similarly assessed *in situ* to better simulate *in vivo* conditions. The mean static precision of inter-bead distance and inscribed angle measures were 0.06 mm and 0.14° respectively (Table 2). These results are in line with Brainerd et al. (2010), who reported a MAE of 0.037 mm and 0.084 mm for phantom-based and *in situ* measures respectively, and Tashman and Anderst (2003), who reported *in situ* precision of 0.053 – 0.085 mm and 0.31 – 0.35°. As expected, precision decreased with bone movement (Table 2, Dynamic).

For dynamic studies, it is important to quantify the amount of error due to temporally-offset images based on the selected frame rate, pulse width, image field of view, and object speeds. To do so, the dynamic phantom trial was conducted over a range of angular velocities that included those measured *in situ* (cadaveric trials ranged between 0 – 60°/s). Using the regression equations (Fig. 2, inset) and maximum cadaveric velocity, the maximum predicted offset errors in glenohumeral measures for the cadaveric trials are 0.85 mm and 0.28°. Importantly, this approach enables quantification of offset error without the need for a separate motion capture system, and can be easily adapted for studies that include different speeds and/or frame rates. The offset error can then be summed with the measured model-based tracking error to provide a direct estimate of the maximum aggregate error of this technique.

The second objective of this study was to determine the accuracy and precision of model-based tracking for quantification of glenohumeral motion and to establish the acceptable range of speeds for which it may be used. For direct comparison to the literature, we highlight the results of our static cadaveric validation, which are free of error due to a temporal-offset, similar to custom-built systems. Zhu et al. (2012) reported MAE of 0.12 – 0.30 mm and 0.09 – 0.58° for global kinematic measures (reported ranges for the humerus, scapula, and various DOFs considered separately). These results are in close agreement with

the global MAE of 0.18 – 0.34 mm and 0.12 – .48° obtained in the present study (Table 3). Massimini et al. (2011) reported a global MAE $\pm$ SD of 0.27 $\pm$.019 mm and 0.49 $\pm$.36° (reported errors averaged the humerus and scapula and various DOFs). In the present study, values of 0.27 $\pm$ 0.20 mm and 0.29 $\pm$ 0.25° were obtained. Bey et al. (2006) reported global RMSE that ranged from 0.29 – 0.39 mm (scapula) and 0.22 – 0.37 mm (humerus) for the various DOFs. Similarly, we obtained global RMSE that ranged from 0.26 – 0.40 mm (scapula) and 0.23 – 0.39 mm (humerus) (Table 3). We found rotational errors to be greater for the humerus than the scapula across DOFs (Table 3). Additionally, RMSE was greatest for glenohumeral axial rotation (Table 3). These findings are corroborated by other investigators (Bey et al., 2006; Massimini et al., 2011; Zhu et al., 2012; Giphart et al., 2012) and are likely due to the symmetrical shape and lack of defining features of the humerus geometry as relates to model registration. Based on our measured MAEs of 0.39 mm and 0.45° (glenohumeral errors averaged across bones and DOFs), using the regression equations to estimate added offset error (Fig. 2, inset), the maximum speeds for which sub-millimeter and sub-degree total relative kinematic errors are obtained are approximately 43°/s and 136 °/s respectively for the imaging setup and protocol used.

Glenohumeral MAE of 0.36 – 0.42 mm and 0.41 – 0.48° (Table 4) were obtained for 3 simulated dynamic motions (representative kinematic profiles shown in Fig. 3). These results must be evaluated in light of the added error in RSA-based measures caused by a temporal-offset between corresponding images (Fig. 2), predicted to account for 0.85 mm and 0.28° error for the fastest activity segments. For each simulated motion, scapula rotational measures were more accurate and precise than the humerus. In addition to the humerus morphology, this discrepancy is likely due to the fact that the humerus often underwent faster angular velocities relative to the scapula for a given simulated motion, and was therefore susceptible to greater offset error.

An important limitation of this study was the use of 1 cadaveric specimen, which likely does not fully capture variation in model-based tracking accuracy that might result from individual differences in bony geometry. However, the main focus of this study was on quantifying the impact of the temporal-offset, which is expected to be the primary source of error in kinematic measures when using a clinical biplane system. Importantly, the impact of this parameter was well characterized in this study (Fig. 2) and is not expected to vary across specimens.

Another limitation of this study was the rate of simulated motions (within trial averages of 12 – 35°/s, and maximum frame-to-frame velocity of approximately 60°/s across trials). Previous patient studies that have characterized glenohumeral kinematics during arm elevation at self-selected slow and fast speeds report average angular velocities of 11.5 – 64.1°/s (scapular plane elevation) and 19 – 39.4°/s (sagittal plane elevation) for slow trials (Michiels & Grevenstein, 1995; Fayad et al., 2006). Thus, the speeds examined are functionally relevant. Additionally, the derived relationship between offset-induced relative kinematic error and object speed (Fig. 2, inset equations) can be used to estimate added error for activity speeds greater than those included in the present study.

Given the temporal-offset in images, we believe that the use of clinical biplane fluoroscopy systems is feasible for certain dynamic, speed-regulated, and quasi-static applications, provided the maximum system frame rate is also adequate for the object speeds examined. For applications in which glenohumeral range of motion is the primary outcome measure, as is often the case in clinical musculoskeletal exams, sub-degree rotational accuracy may be achieved for speeds as great as 136°/s for the imaging setup and protocol (15 frames/s image capture rate) used in the present study. For applications that necessitate sub-millimeter accuracy, such as measurement of the subacromial space, object speeds as great as 43°/s are acceptable. For greater object speeds, we advocate the use of an interpolation-based technique, which has been shown to improve RSA accuracy for asynchronous images (Goodsitt et al., 2017; Akbari-Shandiz et al., 2017). In summary, for defined speeds, use of an unmodified clinical biplane fluoroscope and model-based tracking for quantification of glenohumeral motion is able to provide sub-millimeter and sub-degree relative kinematic accuracy, even with a temporal offset in biplane images.

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Fig. 1. Biplane fluoroscopy configuration and representative image views of shoulder anatomy (left and right fluoroscopic images correspond to left and right image receptors respectively). The glenohumeral joint was centered in the image volume. The specimen was seated upright and positioned as shown (torso model used to depict setup; arms of cadaveric specimen were intact). The detectors were positioned at a relative angle of 125° in the same vertical plane, at a source to detector distance of 105–115 cm, in portrait orientation which provided anteroposterior and oblique superoinferior views of the shoulder anatomy (inset images).

Fig. 2.

Temporal-offset induced relative kinematic error as a function of object movement speed. Fluoroscopic images were acquired of a moving phantom at 15 frames/s from 2 alternating x-ray sources. Each point represents the absolute difference between the measured and machined relative position of 2 bead clusters, averaged across degrees of freedom (circle = translation, diamond = rotation). Data was fit using simple linear regression (inset equations). Error measures enclosed by dashed lines correspond to range of speeds of cadaveric trials.

Fig. 3. 6 DOF glenohumeral kinematics during a dynamic trial of simulated scapular-plane elevation, determined via model-based tracking (diamond) and RSA gold standard (circle) (top row = translations, bottom row = Euler rotations). Trial started with arm beside torso (frame 1) and ended at max achievable humeral elevation (frame 61) (every 3rd frame analyzed). Data was not filtered for analysis.

Table 1

Static phantom-based accuracy and precision of RSA, defined as (1) mean absolute error (MAE) of measured and machined inter-bead distances and (2) standard deviation (SD) of absolute error for bead pairs across frames.

MAE [mm]	SD [mm]
0.05	0.04

Table 2

In situ precision of RSA, defined as standard deviation of (1) measured inter-bead distances for each of 3 humeral bead pairs and (2) inscribed angles of triangle formed by bead triplet across frames. Range provided for dynamic measures due to within and across motion speed variability.

	SD [mm]	SD [°]
Static	0.06	0.14
Dynamic	0.04 – 0.40	0.11 – 0.84

Table 3

Static kinematic accuracy (root-mean-square error; RMSE, mean absolute error; MAE) and precision (standard deviation of AE; SD) of model-based tracking relative to RSA reported in bone local coordinate system.

		RMSE	MAE	SD
Translation	<i>Humerus</i>	X	0.32 mm	0.26 mm
		Y	0.23 mm	0.18 mm
		Z	0.39 mm	0.34 mm
	<i>Scapula</i>	X	0.36 mm	0.29 mm
		Y	0.40 mm	0.31 mm
		Z	0.26 mm	0.22 mm
	<i>Glenohumeral</i>	X	0.55 mm	0.41 mm
		Y	0.40 mm	0.33 mm
		Z	0.52 mm	0.43 mm
Rotation	<i>Humerus</i>	X	0.55°	0.48°
		Y	0.54°	0.45°
		Z	0.40°	0.34°
	<i>Scapula</i>	X	0.16°	0.12°
		Y	0.19°	0.16°
		Z	0.23°	0.18°
	<i>Glenohumeral</i>	X	0.56°	0.49°
		Y	0.60°	0.49°
		Z	0.43°	0.36°

Dynamic kinematic accuracy (root-mean-square error; RMSE, mean absolute error; MAE) and precision (standard deviation of AE; SD) of model-based tracking relative to RSA for 3 simulated motions (DOFs combined) reported in bone local coordinate system.

Table 4

		RMSE	MAE	SD
Translation	<i>Scapular Plane Elevation</i>	Humerus	0.28 mm	0.22 mm
		Scapula	0.39 mm	0.32 mm
		Glenohumeral	0.52 mm	0.42 mm
	<i>Sagittal Plane Elevation</i>	Humerus	0.31 mm	0.25 mm
		Scapula	0.33 mm	0.27 mm
		Glenohumeral	0.46 mm	0.37 mm
	<i>Internal/External Rotation</i>	Humerus	0.35 mm	0.29 mm
		Scapula	0.33 mm	0.26 mm
		Glenohumeral	0.47 mm	0.36 mm
	<i>Scapular Plane Elevation</i>	Humerus	0.48°	0.41°
		Scapula	0.19°	0.15°
		Glenohumeral	0.49°	0.41°
Rotation	<i>Sagittal Plane Elevation</i>	Humerus	0.52°	0.45°
		Scapula	0.16°	0.12°
		Glenohumeral	0.58°	0.48°
	<i>Internal/External Rotation</i>	Humerus	0.47°	0.39°
		Scapula	0.16°	0.13°
		Glenohumeral	0.50°	0.41°