Kinematics of the Glenohumeral Joint: Influences of Muscle Forces, Ligamentous Constraints, and Articular Geometry

*Andrew R. Karduna, Gerald R. Williams, †John L. Williams, and Joseph P. Iannotti

Departments of Bioengineering and Orthopaedic Surgery, University of Pennsylvania;

**Department of Physical Therapy, Allegheny University, Philadelphia, Pennsylvania; and*

†Department of Orthopaedic Surgery, University of Missouri-Kansas City, Kansas City, Missouri, U.S.A.

Summary: Despite recent interest in the study of shoulder kinematics, there is considerable controversy in the literature regarding translations at the glenohumeral joint. The purpose of this study was to investigate the key factors that control shoulder motions, thus leading to a better understanding of joint function. Translation and rotation patterns were studied in fresh-frozen glenohumeral joints of human cadavers with a six-degrees-of-freedom magnetic tracking device. Shoulders were positioned from maximal internal to external rotation at several arm positions (various elevations and planes of motion). In order to determine the effect of muscle forces, joints were positioned both actively and passively. Additionally, articular surface geometry and ligament origin-insertion wrap lengths were measured to assess their influences on joint kinematics. When joints were positioned passively, large translations were observed at the extremes of motion. With active positioning, muscle forces tended to limit humeral head translations, principally by restricting rotational ranges of motion. However, when data from the passive model were reanalyzed by considering only the rotational ranges of motion seen actively, no significant differences in translation were found between the two models. Joint conformity was found to have a significant influence on translations during active positioning but not during passive positioning. Glenohumeral ligament wrap lengths, however, correlated with translations when joints were positioned passively but not when positioned actively. Findings from this study emphasize the importance of muscle forces in keeping the humeral head centered in the glenoid. Although large translations are possible, they can be achieved only with increases in rotational ranges of motion associated with the removal of muscle force. Additionally, joint conformity appears to play a role in controlling translations during active motions, whereas capsular constraints become more important during passive motions.

The glenohumeral articulation has the capacity for the greatest range of motion of any diarthrodial joint in the human body. This is partially due to the shallowness of the glenoid cavity (38). One consequence of this architecture is that it allows for translations of the humeral head with respect to the glenoid. Clinically, these translations are responsible for alterations in articular contact patterns (39). Additionally, they can influence muscle loading by changing moment arms and lines of action (46). At the extremes of motions, translations affect the force distribution between muscles, ligaments, and articular surfaces (13). Finally, large translations ultimately lead to joint instability (27).

Several controlling mechanisms are thought to influence glenohumeral translations, including articular and labrum geometry, intraarticular pressure, liga-

mentous constraints, and muscle forces (28). Many studies have addressed these factors as they relate to joint stability by simulating a distracting force or displacement (21,26,27,44). These studies have helped determine the relative contribution of such factors in preventing joint subluxation but have not addressed translation patterns during physiological motions, such as the activities of daily life.

Glenohumeral translations have been studied during normal motions, with apparently conflicting results. Whereas some studies have demonstrated minimal translations of 1-2 mm (22,29,35), others have documented significant deviations from ball-and-socket kinematics, with reported translations of 3-8 mm (19, 32,47). Finally, there are reports in the literature of extreme translations as large as 15 mm (15,16). These discrepancies may be due to several factors, including differences in the plane of motion studied, *in vivo* compared with *in vitro* studies, and the experimental model used.

Studying shoulder kinematics *in vivo* is desirable, but making the required measurements is difficult due to problems associated with obtaining full six-degrees-

Received October 3, 1995; accepted July 12, 1996.

Address correspondence and reprint requests to G. R. Williams at Department of Orthopaedic Surgery, University of Pennsylvania, 39th and Market Streets, One Cupp Pavilion, Philadelphia, PA 19104, U.S.A.

FIG. 1. Experimental setup. **A:** Schematic of axis system. t = Fastrak transmitter; r = Fastrak receiver; a and b = points on the medial border of the scapula; c = center of the glenoid; and d = center of the humeral head. Points a, b, and c define the plane of the scapula. **B:** Picture of active joint-positioning technique.

of-freedom information. Only Harryman et al. have recorded three-dimensional glenohumeral translations *in vivo*, to our knowledge, but this was in a limited protocol with only two subjects (14). Other *in vivo* studies have reported three rotational degrees of freedom but no translational information, possibly due to insufficient accuracy (8,18,43).

In vitro cadaver models offer the advantage of allowing invasive attachment of markers or measurement devices directly to bones for accurate measurement of three-dimensional translations. However, there are various ways to simulate the desired motion. Reports in the literature can be grouped into two categories: active models, where muscle forces are simulated; and passive models, where no muscle forces are simulated and an external force or torque is applied. Some of the discrepancies in translation patterns observed in the literature may be related to these model differences.

The present study was undertaken to help resolve some of the confusion regarding translations of the humeral head with respect to the glenoid. The object was to determine if glenohumeral translations were significantly influenced by the chosen experimental model. To this end, experiments were conducted using both active and passive models on the same human cadaver joints to determine the effect of muscular control on glenohumeral kinematics. A secondary objective was to determine if ligamentous constraints and articular joint conformity influenced humeral head translations. Ligament wrap length and articular surface geometry patterns were therefore measured in all specimens.

MATERIALS AND METHODS

Normal fresh-frozen human glenohumeral joints were harvested from cadavers and were dissected to the level of the rotator cuff. The condyles and the deltoid insertions were preserved. Each potential specimen was examined for signs of gross pathology (e.g., arthritis or rotator cuff tears) and was manually manipulated to

verify stability and range of motion. A radiograph was taken to help confirm the condition of the joint. Those specimens that were found to be unacceptable, by radiographic or physical examination, were eliminated from the study. Specimens were stored frozen until needed and were then thawed at room temperature prior to experimentation. Once thawed, they were kept moist with a protease-inhibitor solution (34). Data were collected from a total of seven joints (mean age 72 years; range 61-88 years).

The infraspinatus and subscapularis fossae were cleaned of all soft tissue, and several bolts were inserted along the inferior border of the scapula to add stability in the potting process. The scapula was potted with its medial border aligned vertically in a plastic cup containing Bondo car body filler (Dynatron/Bondo, Atlanta, GA, U.S.A.). This pot was bolted to a hinge joint that allowed for scapular tilt about the anterior-posterior axis to simulate scapular rotation during elevation. The axis was aligned with respect to the scapular plane, which was approximately 30° from the clinical coronal plane of the trunk.

The active experimental model was based on the work of Soslowsky et al. (39). A static-active model was used, in which muscle forces were applied sequentially until an equilibrium position was achieved, and then a single kinematic datum point was recorded. This is in contrast to a dynamic-active model, in which muscle forces are continuously applied and kinematic data are recorded while the joint is being positioned (6,48).

To simulate deltoid muscle forces, four strands of low-stretch braided Dacron (Berkley, Spirit Lake, IA, U.S.A.) were tied to a plastic washer that was bolted to the humerus at the deltoid insertion site. Lines of action for the muscle were simulated by running the cord through four corresponding eyelets on the scapula. These eyelets represented the edges of the deltoid origin site and were located on the tip of the coracoid, the anterior acromion, the posterior acromion, and the middle of the scapular spine. For simulation of rotator cuff muscles, Dacron cord was sewn directly into the tendinous insertions of the subscapularis, supraspinatus, and combined infraspinatus-teres minor complex in a Bunnell suture pattern. Lines of action for these muscles were maintained with alignment holes at the centroids of their scapular origins. It was not possible to consistently maintain capsular integrity during the dissection and rotator cuff preparation, so all joints were vented for consistency.

Plastic tubing was inserted into the alignment holes at all of the origin sites to help reduce friction. Each cord was then run through its respective origin alignment hole and fastened to a spring scale (Chatillon, Kew Gardens, NY, U.S.A.) with a 15 kg maximum load and 0.25 kg increments. The scale was in turn connected to a hand crank. This hole/scale/crank system allowed

for the manual application of individual loads to each muscle insertion site on the humerus.

Kinematics were monitored with a magnetic tracking device (Polhemus 3Space Fastrak; Colchester, VT, U.S.A.) (2,33) consisting of a system electronics unit wired to a transmitter that was fixed to the lateral border of the scapula with a plastic mount, and a receiver attached directly to the medial side of the humerus, as close as possible to the center of the humeral head (Fig. 1). To record anatomically relevant data, the reference system developed by An et al. (3) was adapted for the present study. The coordinate axis of the transmitter was transformed into a fixed axis on the scapula, and the coordinate axis of the receiver was transformed into a moving axis on the humerus.

The scapular coordinate system was defined by three points: the superior and inferior points of the medial border of the scapula and the center of the sphere defined by the glenoid articular surface (Fig. 1A, points a, b, and c). The axis system was determined as follows: the superior-inferior axis orientation was defined by the vector connecting the two points on the medial border; the anterior-posterior axis orientation was defined by a vector perpendicular to the plane that was defined by all three points; and the medial-lateral axis orientation was defined by a vector orthogonal to the superior-inferior and anterior-posterior axes. The origin of this scapular coordinate system was located at the center of the glenoid sphere. The orientation of the humeral coordinate system was defined as in Fig. 1A so that it was aligned with the scapular coordinate system when the humerus was in its neutral position, with the arm at the side and the condyles aligned with the scapular plane. The origin of the humeral coordinate system was located at the center of the humeral articular surface (Fig. 1A, point d).

The points on the scapular medial border were located with a digitizer consisting of the Fastrak receiver mounted on a plastic stylus. The humerus was passively positioned through a central range of motion where the ligaments were not taut, ensuring that the head was centered in the glenoid with a manually applied, nominal compressive load. The center of the head was defined as the point in the transmitter axis system that moved the least during this motion (smallest root mean square motion as determined by an iterative searching algorithm). The center of the glenoid was defined as the location of the center of the head in the receiver axis system when the head was centered in the glenoid.

As the humerus deviated from its neutral position, three rotations and three translations were recorded. Euler angles were used to represent three sequence-dependent rotations: plane of elevation, degree of elevation, and internal-external rotation. It was not possible to distinguish between plane of elevation and internal-external rotation when the humerus was at 0° of elevation; therefore, the plane of elevation was considered undefined when the humerus was not elevated. The translations occurred along the three previously defined orthogonal axes: superior-inferior, medial-lateral, and anterior-posterior. These parameters were monitored on-line by interfacing the Fastrak with custom-written software.

The testing protocol consisted of positioning the joint in maximal internal rotation and then externally rotating it in 10° increments until maximal external rotation was achieved. This procedure was performed in the plane of the scapula with the arm at 0, 30, and 60° of glenohumeral elevation. With the arm at 30 and 60° of elevation, testing was also performed with the arm positioned at 30° posterior and 30° anterior to the plane of the scapula, yielding a total of seven experiments. At each increment, elevation and plane were maintained within 5° of their target values, while internal-external rotation was maintained within 1° of its desired value.

Each motion was carried out with both active and passive positioning. For active motions, the cranks were used to simulate muscle forces that moved the humerus (Fig. 1B). A 2.3 kg concrete mass was hung at the distal end of the humerus to approximate

the weight of the arm. Also, for every 2° of glenohumeral elevation, the scapula was rotated 1° (7,9,32). Muscle forces were determined from clinical knowledge and electromyographic studies (25,31). The specific muscle-simulation strategy was as follows: the supraspinatus and middle deltoid forces were used to achieve the proper elevation, the anterior and posterior deltoid forces maintained the desired plane, and the subscapularis and combined infraspinatus-teres minor forces controlled the internal-external rotation. For passive motions, the humerus was manually rotated with a small, medially directed centering force (13).

When the experimental protocol was completed, the anterior band of the inferior glenohumeral ligament was identified and was marked with two sutures (origin and insertion sites). The entire capsule was excised and these sites were digitized with the Fastrak stylus (17). For each experimental position, the so-called origin-insertion wrap length, w , was calculated as $w = r * \cos^{-1}(1 - d^2 / 2r^2)$ where r is the radius of curvature of the humeral head and d is the straight-line distance between the origin and insertion sites (see Appendix). Due to the subjective nature of determining zero-stress lengths, ligament data were analyzed as wrap lengths and not as ligament strains. The wrap length was not necessarily the absolute ligament length, since this analysis did not account for infolding that occurs when a ligament is lax.

After the ligament digitization process, the glenoid and the humeral head were returned to the freezer. These specimens were subsequently thawed, repotted in Bondo, and mounted on a translation table, which was fixed to a mechanical testing machine (MTS, Minneapolis, MN, U.S.A.). The glenoid and humeral head articular surfaces were aligned for digitization of an arc in the anterior-posterior direction. Only one arc was measured on each surface, since it was assumed that both surfaces were spherical (38). A 10 mm diameter metal ball was run along the deepest arc of the glenoid and the highest arc of the humeral head. A force of 1 N was applied to maintain contact between the ball and cartilage. This contact point was recorded accurately in two dimensions by the MTS linear variable differential transformer and another orthogonally mounted linear variable differential transformer (Transcoil, Valley Forge, PA, U.S.A.).

These points were then used to find the best-fit radii of curvature of the articular surfaces, taking into account the curvature of the metal ball. Joints were divided into two groups: nonconforming articulations, where the humeral head curvature was less than the glenoid curvature; and conforming articulations, where the humeral head curvature was greater than or equal to the glenoid curvature. Cases in which the humeral head curvature was greater than the glenoid curvature were included in the conforming group since the flexible glenoid labrum deforms to conform to the humeral head surface.

The accuracies of the measurement techniques were established as follows. The Fastrak transmitter and receiver were mounted to the base and arm of a plastic ball-and-socket joint, respectively. A coordinate system was established as with the shoulder experiments, and the joint was rotated 120°, with Fastrak data recorded every 10°. This experiment was repeated eight times, with the orientation of the transmitter changed between repetitions. The same technique used for measuring the radii of curvature of the humeral head and glenoid was used to determine the radius of curvature of a metal ball with a 25.4 mm radius in 10 trials. Additionally, the amount of indentation of articular cartilage attributed to this technique was measured. The effects of these indentations were determined by introducing these errors into the calculation of the radius of curvature of the 25.4 mm metal ball. Finally, the method of determining changes in ligament lengths by digitizing ligament origin and insertion sites with the Fastrak stylus was compared with readings from a Hall-effect strain transducer (Micro Strain, Burlington, VT, U.S.A.). Internal to external rotation with the arm at 0° of elevation was tested in a cadaver specimen.

TABLE 1. Net translations (mean $\pm$ SEM) for all experiments

Experiments	Axis	Translations for each model (mm)			P values	
		Active (A)	Passive (P1) full range of motion	Passive (P2) limited range of motion ^a	Case 1: A vs P1	Case 2: A vs P2
0° elevation; plane undefined	Anterior-posterior	3.9 $\pm$ 1.0	7.5 $\pm$ 0.7	4.0 $\pm$ 0.9	0.035	> 0.4
	Superior-inferior	2.6 $\pm$ 0.4	4.0 $\pm$ 0.5	2.6 $\pm$ 0.5	0.070	> 0.4
30° elevation; posterior plane	Anterior-posterior	1.6 $\pm$ 0.2	9.7 $\pm$ 1.0	3.4 $\pm$ 0.8	0.003	> 0.4
	Superior-inferior	1.6 $\pm$ 0.4	4.3 $\pm$ 0.5	1.9 $\pm$ 0.4	0.001	> 0.4
30° elevation; scapular plane	Anterior-posterior	1.7 $\pm$ 0.4	9.4 $\pm$ 0.9	1.7 $\pm$ 0.4	0.002	> 0.4
	Superior-inferior	1.3 $\pm$ 0.2	4.6 $\pm$ 0.7	1.9 $\pm$ 0.3	0.005	> 0.05
30° elevation; anterior plane	Anterior-posterior	2.1 $\pm$ 0.4	7.3 $\pm$ 0.5	2.0 $\pm$ 0.3	0.001	> 0.4
	Superior-inferior	1.4 $\pm$ 0.3	4.2 $\pm$ 0.7	1.6 $\pm$ 0.4	0.005	> 0.4
60° elevation; posterior plane	Anterior-posterior	2.0 $\pm$ 0.4	9.8 $\pm$ 1.6	2.1 $\pm$ 0.3	0.005	> 0.4
	Superior-inferior	2.0 $\pm$ 0.6	4.0 $\pm$ 1.0	2.1 $\pm$ 0.5	0.040	> 0.4
60° elevation; scapular plane	Anterior-posterior	2.1 $\pm$ 0.4	7.4 $\pm$ 1.1	1.4 $\pm$ 0.3	0.035	> 0.4
	Superior-inferior	1.6 $\pm$ 0.6	4.4 $\pm$ 0.8	1.9 $\pm$ 0.6	0.003	> 0.4
60° elevation; anterior plane	Anterior-posterior	1.5 $\pm$ 0.3	6.1 $\pm$ 1.0	2.7 $\pm$ 1.2	0.030	> 0.4
	Superior-inferior	1.4 $\pm$ 0.4	4.0 $\pm$ 0.8	1.8 $\pm$ 0.5	0.010	> 0.4

^aSame range of motion as corresponding active experiment.

Systat 5.2 (Systat, Evanston, IL, U.S.A.) for the Macintosh was used for all statistical analyses (42). To distinguish between translation patterns of active and passive models, a repeated measures analysis of variance (ANOVA) was performed with two within-subject factors: models (active and passive) and experiments (seven total). Contrasts were used to make planned comparisons between means. Since these contrasts were not orthogonal, a Bonferroni correction factor was used to maintain the desired experiment-wise error rate (24,36). The effect of conformity on translation was tested with a repeated measures ANOVA with one within-subject factor, experiments (seven total), and one between-subject factor, conformity (conforming and nonconforming joints). Correlations between ligament wrap lengths and translations were made with dummy variables to account for between-subject differences. This approach fits the data from every specimen with the same slope but allows for different intercepts for each specimen (11). For all tests, the acceptable rate for a type-I error was set at 5% ($p = 0.05$).

RESULTS

Since there were no physical translations with the plastic ball-and-socket joint, the absolute magnitude of any translation recorded by the Fastrak was considered the error of that measurement. The data for all three axes were pooled, yielding a total of 312 points. The errors were found to have a mean of 0.25 mm and a SD of 0.17 mm. The uncertainty of these measurements at the 95% confidence level was 43 mm. For each of the 10 trials, the calculated radius of curvature of the metal ball was within 0.2 mm of the actual radius of curvature. On cartilage surfaces, however, the 1 N load was found to produce small indentations that were on the order of the linear variable differential transformer accuracy (approximately 0.02 mm). When these errors were taken into account, the radius of curvature calculated for the metal ball was within 0.5 mm of its value in all cases. For the purposes

FIG. 2. Comparison of ligament lengths with a Hall-effect strain transducer (HEST) and a Polhemus Fastrak. Motions are for external rotation with the arm at the side for a representative ligament (middle glenohumeral ligament). The zero ligament-length change was set as the smallest value of the strain-transducer reading.

FIG. 3. Comparison of anterior-posterior (AP) translation with passive and active models. Data are from a representative specimen during maximum internal to external rotation, at 60° of elevation, in the plane of the scapula.

of comparing the techniques for measuring ligament elongation, a zero distance for each motion was defined as the smallest reading from the strain transducer during that motion. The results from the two techniques (Hall-effect strain transducer and Polhemus Fastrak) matched qualitatively for percentage increases in lengths (Fig. 2). However, while the strain-transducer reading stopped decreasing once the ligament became slack, the Fastrak continued to record a decrease in lengths.

The net anterior-posterior translation was defined as the distance between the most posterior and anterior positions of the center of the head for a given motion. The net superior-inferior translation was defined in an analogous manner. The mean net anterior-posterior translation for all of the experiments was approximately 4-fold larger for the passive model (8.2 mm) than for the active model (2.1 mm) ($p = 0.0003$). Similarly, the mean net superior-inferior translation for all experiments during passive motions (4.2 mm) was approximately 2.5 times larger than for active motions (1.7 mm) ($p = 0.0002$). For all individual experiments, mean net translations for passive motions were greater than for active motions for both the anterior-posterior and superior-inferior axes. Individual contrasts between active and passive translations were significant for all but one experiment ($p < 0.05$) (Table 1, Case 1).

The mean internal-external rotational range of motion during passive positioning (144°) was significantly greater than during active positioning (103°) ($p < 0.0001$). The range of motion was greater in the passive model in 48 of the 49 trials for the seven specimens. The passive data, therefore, were reanalyzed by considering only the range of motion that occurred during the corresponding active motion. The new mean of net anterior-posterior translations for all passive experiments was 2.5 mm, and for superior-inferior translations, it was 2.0 mm. When the planned contrasts were repeated, there were no significant differences between the mean net active and the limited range of motion passive data ($p > 0.05$) (Table 1, Case 2). Representative translation patterns during active and passive motions are presented in Fig. 3 to illustrate these results. For passive motions, there was a trend toward anterior translations with internal rotation and posterior translations during external rotation. These translations typically occurred outside the ranges of active motion.

Patterns between ligament wrap lengths and translations were found to account for some of the discrepancies between active and passive translations. For the active model, there was no significant correlation ($p = 0.48$, $r^2 = 0.02$) between posterior translation and the wrap length of the anterior band of the glenohumeral ligament at maximum external rotation (Fig. 4A). To

FIG. 4. Ligament wrap length-translation relationships. These plots show the correlation between the anterior band of the inferior glenohumeral ligament origin-insertion wrap length and posterior translation during external rotation. To account for between-subject differences, dummy variables were used so that each specimen was fit with the same slope but different intercepts. Normalized ligament lengths were then calculated by shifting data from each specimen so that each specimen data set had the same y-axis intercept. **A:** Translation at maximum external rotation for the active model. **B:** Difference in translation between passive and active models during maximum external rotation.

FIG. 5. Effects of conformity on translation. Grand mean $\pm$ SEM of net anterior-posterior (A/P) translation for conforming and nonconforming joints, in both active and passive models. Conforming joints are defined as specimens in which the humeral head curvature is greater than or equal to the glenoid curvature. Nonconforming joints are defined as specimens in which the humeral head curvature is less than the glenoid curvature.

account for any changes between the active and passive models, the difference between passive and active anterior-posterior translations at maximum external rotation was calculated. When this difference was compared with the ligament wrap length at maximum passive external rotation, a significant correlation was observed ($p < 0.0001$, $r^2 = 0.52$) (Fig. 4B).

Three specimens were found to be nonconforming articulations, since each had a larger glenoid radius of curvature compared with that of the humeral head. Four specimens were found to be conforming articulations, since each had a larger humeral head than glenoid. For active motions, net anterior-posterior translations were significantly larger in nonconforming joints than in conforming ones ($p = 0.049$) (Fig. 5). For passive motions, however, there was no significant difference between these two groups ($p = 0.22$) (Fig. 5).

DISCUSSION

Several *in vitro* studies have reported minimal glenohumeral translations during simulated active elevation, concluding that for the most part, the joint behaves in a ball-and-socket fashion (22,29). Other investigators, however, have demonstrated very large translations during *in vitro* motions, signifying gross deviations from ball-and-socket kinematics (13,16,47). *In vivo* results appear to fall between these two extremes, with limited translations observed during certain arm positions (19,32).

During active positioning in the present study, there were minimal humeral head translations. This was in contrast to the large translations observed during passive positioning, especially at the extremes of motion. The nature of these translations was such that when passive motions were considered only over the same range of motion obtained during active motions, so that the extremes were not considered, translations were not significantly different from those observed during active motions.

The translations seen in the present study during full passive range of motion are in agreement with those reported by Harryman et al. (13). Additionally, the translations during active motions are in agreement with Kelkar et al. (22) and McMahon et al. (29). In contrast to Howell et al. (19), large posterior translations during external rotation were seen only during passive motions in the present study. Similarly, the large translations reported by Wuelker et al. (47) during active elevation are difficult to explain. The differences may be attributable to different muscle forces, potential initial inferior subluxation of the specimens, or different methods of locating the center of the humeral head.

The amount of discrepancy between posterior translation observed in the active and full range of

motion passive models during external rotation was found to correlate with changes in the wrap length of the anterior band of the inferior glenohumeral ligament. As this length increased, larger passive posterior translations were observed. It is presumed that lengthening of this ligament, at these extremes, corresponds with the development of increased tension in it, and that these translations are related to this tension. During active motions, there is an additional force provided by the rotator cuff muscles that tends to compress the head into the glenoid. This joint centering may help to offset any tension developed in the capsular ligaments, thus limiting glenohumeral translations by restricting the achievable range of motion. The concept that ligaments only support loads at the extremes of motion has been demonstrated previously for both rotations (13) and translations (27). This is apparently the first study, however, to demonstrate a direct correlation between translations and ligament lengthening patterns.

The small anterior-posterior translations observed in the present study during active positioning, in which the humeral head was probably still in contact with the glenoid articular surface, were significantly influenced by joint conformity. These results are consistent with those of Kelkar et al., who demonstrated a correlation between translation and joint conformity during scapular plane elevation in an active cadaver model (23). Based on the number of specimens in the present study, differences in joint conformity patterns were not found to affect translations during passive motions. Power calculations indicated that this analysis had a 75% certainty of detecting a 2-fold increase of translation with lack of conformity, as seen in the active model. The similarity in translation patterns between conforming and nonconforming joints in the passive model was probably because a lack of centralizing muscle forces caused a loss of contact between the head and the glenoid articular surfaces.

Although the muscle forces used in the active model of the present study were selected on the basis of their function, other combinations of muscle forces could have produced the same rotational motions, since this was a statically indeterminate system. For example, forces could have been selected by preserving certain ratios between deltoid and rotator cuff muscles (29,35,47) or through the use of an optimization technique to determine the appropriate muscle forces (1,4). Although different muscle forces may have produced different results, it is unlikely that they could have increased the rotational ranges of motion to the extent that translations in the active model would have approached those seen in the passive model. Although the technique of passive manipulation of the glenohumeral joint has been used by other investigators (13,16), the forces and moments that

were applied in the present study were not measured. A preliminary reproducibility study and consistent results from specimen to specimen indicate that this methodology was well controlled. Additionally, each test was performed by the same individual with no graphical feedback as to the amount of translation present. However, due to the complexity of the glenohumeral joint and the experimental setup, it was not possible to completely control all of the forces and moments involved in joint positioning.

Since ligaments can only support loads in tension, one must know whether or not a ligament is longer than its zero-stress length in order to determine if it is in a state of tensile strain. Although capsular ligament strains have been reported by other investigators (30,40), this approach was avoided due to the potentially large errors associated with determining zero-tension lengths and the difficulty of attaching strain transducers to the joint capsule.

Joints were vented in the present study, eliminating the negative intraarticular pressure normally found in the glenohumeral joint (12,41). It has been found that this pressure contributes to the control of glenohumeral translations (10,45). However, Bowen et al. found that the contribution of negative intraarticular pressure in preventing translations was dramatically reduced with the application of joint compressive loads of as little as approximately 100 N (5). Finally, there were potential inaccuracies due to cartilage indentation while making radius of curvature measurements. The effects of this indentation were estimated, and the resulting accuracy was sufficient to account for the differences in curvature between the glenoids and humeral heads used in the present study. A non-contacting method, such as stereophotogrammetry, would avoid this problem (20,37).

In conclusion, this study provides the first attempt at directly comparing the kinematics of active and passive *in vitro* cadaver models for the same natural glenohumeral joints. In doing so, it confirms and reconciles several opposite views of the glenohumeral joint noted in the literature. It was shown that the muscle forces simulated in this study served to limit the rotational ranges of motion and translations in comparison with passive positioning of the joint, presumably due to the large joint compressive loads produced. When muscle forces were absent, much larger rotational ranges of motion and translations were possible. Additionally, while joint conformity was related to translations during mid-range active motions, ligament wrap lengths were found to be potentially important at the extreme of motion achieved only during passive motions.

Acknowledgment: We gratefully acknowledge the help of Dr. Van Mow, Dr. Louis Soslowsky, and Dr. Rajeev Kelkar for

their assistance in the development of our model. This work was funded by a research grant from DePuy, Warsaw, IN, U.S.A.

APPENDIX

The geometry for determining the length of a ligament that is wrapped around the humeral head is shown in the configuration below. This wrap length, w , is given by

$$w = r * \theta \quad (1)$$

where θ is the angle subtended and r is the radius of curvature of the humeral head. In order to solve for this angle, the law of cosines is used:

$$r^2 + r^2 - (2r^2) \cos(\theta) = d^2 \quad (2)$$

where d is the straight-line distance referred to in the paper. Solving for θ gives

$$\theta = \cos^{-1}(1 - d^2 / 2r^2) \quad (3)$$

Finally, combining Eqs. 1 and 3 gives

$$w = r * \cos^{-1}(1 - d^2 / 2r^2) \quad (4)$$

The geometric configuration used to relate ligament wrap length, w , to the so-called straight line distance, d , between the origin and insertion sites, looks like this:

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