The effect of articular malposition after total shoulder arthroplasty on glenohumeral translations, range of motion, and subacromial impingement

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The articular surface of the normal humeral head has a variable posterior and medial offset with respect to the central axis of the humeral shaft. Recreation of the normal humeral head shaft offset is postulated to be an important consideration during shoulder arthroplasty. However, the effect of humeral head malposition is unknown. The purpose of this study was to determine the effect of articular malposition after total shoulder arthroplasty on glenohumeral translation, range of motion, and subacromial impingement. Twenty-one human cadavers were dissected and tested with the use of an active or passive shoulder model. Range of motion and translation were recorded by means of an electromagnetic tracking device. The experiment was performed in 2 phases. For kinematics study, 11 cadaver shoulders were positioned both passively and actively from maximum internal rotation to maximum external rotation at 90° of total elevation in the scapular plane. Three rotator cuff and 3 deltoid muscle lines of action were simulated for active joint positioning. Passive joint positioning was accomplished with the use of a torque wrench and a nominal centering force. The testing protocol was used for the natural joint as well as for 9 prosthetic head locations: centered and 2- and 4-mm offsets in the anterior, posterior, inferior, and superior directions. Repeated-measures analysis of variance was used to test for significant differences in the range of motion and translation between active and passive positioning of the natural joint as well as all prosthetic head positions. (2) For impingement study, 10 cadaver shoulders were used in a passive model, loading the tendons of the rotator cuff with a 30-N centering force. The humerus was passively rotated from maximum internal rotation (1500 Nmm) to maximum external rotation (1500 Nmm) by means of a continuous-recording digital torque wrench. Trials were performed with the use of centered, 4-, 6-, and 8-mm offset heads in the anterior, posterior, superior, and inferior positions before and after removal of the acromion and coracoacromial ligament. The relation between change in mean peak torque (with and without acromion), passive range of motion, and humeral head offset was analyzed by means of repeated-measures analysis of variance. In the kinematics study, total range of motion and all humeral translations were greater with passive joint positioning than with active positioning ($P = .01$) except for total superior-inferior translation and superior-inferior translation in external rotation. Anterior to posterior humeral head offset was associated with statistically significant changes in total range of motion ($P = .02$), range of internal rotation ($P = .02$), range of external rotation ($P = .0001$), and total anterior-posterior translation ($P = .01$). Superior to inferior humeral head offset resulted in statistically significant changes in total range of motion ($P = .02$), range of internal rotation ($P = .0001$), anterior-posterior translation during external rotation ($P = .01$), and total superior-inferior translation ($P = .03$). In the impingement study, there was a significant increase in torque from centered to 4-mm inferior offset ($P = .006$), 6-mm inferior offset ($P < .001$), and 8-mm inferior offset ($P < .001$). There was no significant increase in torque with superior, anterior, and posterior offsets. Glenohumeral motion significantly decreased from 129° for centered head to 119° for 8-mm superior ($P = .002$), 119° for 8-mm anterior ($P = .014$),

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118° for 8-mm inferior (P < .001), and 114° for 8-mm posterior (P = .001). Humeral articular malposition of 4 mm or less during prosthetic arthroplasty of the glenohumeral joint may lead to small alterations in humeral translations and range of motion. Inferior malposition of greater than 4 mm can lead to increased subacromial contact; offset of 8 mm in any direction results in significant decreases in passive range of motion. Therefore if subacromial contact is to be minimized and glenohumeral motion maximized after shoulder replacement, anatomic reconstruction of the humeral head–humeral shaft offset to within 4 mm is desirable. (J Shoulder Elbow Surg 2001;10:399-409.)

Total shoulder replacement for conditions characterized by destruction of the glenohumeral joint surfaces produces predictable pain relief and improved function in the majority of cases.[5-8,16,19] The quality of the result is dependent on many factors, including rotator cuff integrity, the degree of soft-tissue contracture, and postoperative rehabilitation.[19] An additional factor that may influence the outcome of total shoulder replacement is the quality of the anatomic result. This is defined by the degree to which the anatomic relation between the prosthetic articular surfaces resemble the anatomic relation between the articular surfaces of the normal shoulder. The presumed goal of prosthetic reconstruction of the glenohumeral joint is recreation of normal anatomic relation between the glenoid and humeral articular surfaces.

The location of the humeral head with respect to the proximal humeral metaphysis is an important factor in determining the relation between the articular surfaces. The center of the humeral head does not coincide with the projected center of the humeral shaft in the normal shoulder.[3,4,17,20,22] The distance between the center of the humeral head and the central axis of the intramedullary canal is defined as the humeral head offset.[3,4,20,22] Although humeral head offset is undoubtedly 3-dimensional, it is commonly described in 2 planes, coronal and axial. Like most other anatomic parameters, reported humeral head offsets are variable.[3,4,17,20,22] In the coronal plane, the humeral head offset is approximately 7 to 9 mm medial to the central axis of the intramedullary canal; in the axial plane, the humeral head offset is 2 to 4 mm posterior to the central axis of the intramedullary canal.[3,4,20,22]

Many shoulder prosthetic designs include a humeral head whose center is coincident with the central axis of the humeral stem. More recently, modular designs have appeared that incorporate an offset taper that allows the humeral head to be positioned eccentrically with respect to the stem of the implant. This feature is intended to address humeral head offset and facilitates anatomic placement of the articular surface on the cut surface of the metaphysis. However, the consequences of nonanatomic placement of the prosthetic head during total shoulder replacement are unknown. The purpose of our study was to determine the effect of malposition of the humeral articular surface during total shoulder arthroplasty on glenohumeral translation, range of motion, and subacromial contact in a cadaver model.

MATERIALS AND METHODS

The study was divided into 2 phases. During phase I, the effect of humeral articular malposition on glenohumeral translation and rotation (ie, kinematics) was studied during active as well as passive joint positioning. In phase II, the effects of humeral articular malposition on subacromial contact and glenohumeral rotation were assessed before and after removal of the acromion and coracoacromial ligament, by using passive joint positioning only. In both phases, specimen preparation and arthroplasty technique were the same.

Specimen preparation

Twenty-one glenohumeral joints were harvested from fresh-frozen cadavers. Shoulders were dissected to the level of the rotator cuff, with the distal humeral condyles and deltoid insertion site preserved. Those shoulders found to be arthritic by radiography or stiff by physical examination were eliminated from the study. All joints were vented, eliminating any effects of intra-articular pressure. Specimens were kept moist with a protease inhibitor solution during experimentation. Joints were stored frozen and thawed at room temperature before experimentation.

In phase I specimens, simulated deltoid muscle forces were applied with strands of low-stretch braided Dacron, which were attached to the deltoid insertion site on the humerus by a transosseous bolt. These cords were fed through 3 eyelets on the scapula that roughly corresponded to the mid-anatomic origin of the anterior, middle, and posterior deltoid (tip of the coracoid, anterior acromion, and posterior acromion, respectively). Simulation of the deltoid was not required in phase II specimens.

For simulation of rotator cuff muscles, Dacron cord was sewn directly into the tendinous insertion of the subscapularis, supraspinatus, and combined infraspinatus/teres minor complex in a Bunnell suture pattern. The infraspinatus and subscapularis fossae were cleaned of all soft tissue, and the muscle centroids were marked corresponding to the mid-axis of the fossae. Pulleys were placed on the potting base so that rotator cuff lines of action were aligned with the marked muscle centroids.

We have previously tested normal cadaver shoulders in our laboratory during elevation in the scapular plane as well as during maximum internal to maximum external rotation in the scapular plane, anterior to the scapular plane, and posterior to the scapular plane in multiple degrees of elevation.[14] The humeral translation seen during maximum internal to maximum external rotation at 90° of elevation in the scapular plane was representative of the data as a whole. Moreover, translation with the arm at 90° of total elevation was found to correlate with increasing length of various portions of the inferior glenohumeral ligament during internal (posterior portion of the inferior glenohumeral

ligament) and external rotation (anterior portion of the inferior glenohumeral ligament). Therefore, in a subsequent study, the effect of articular conformity after total shoulder arthroplasty was investigated with the arm in 90° of total elevation to isolate the effects of the inferior glenohumeral ligament. The current experiment was performed in the same position. The arthroplasty was performed with the use of a technique that spared the inferior glenohumeral ligament, as described below.[12-14]

Because the entire protocols for both phase I and phase II were performed at 90° of elevation, the scapula was potted in Bondo (Dynatron/Bondo Corp, Atlanta, Ga), with its medial border aligned at 30° of scapular elevation to achieve a 2:1 ratio of glenohumeral to scapulothoracic motion. To ensure proper alignment, the scapula was held in its desired position with alignment bolts and the Bondo was injected into the base with Monoject catheter tip syringes (Sherwood Medical, St Louis, Mo). Muscle forces were simulated with a hand crank system discussed elsewhere.[12-14]

Each total shoulder implant (DePuy, Warsaw, Ind) consisted of a humeral stem, a humeral head, and a glenoid component. A titanium-alloy humeral stem of appropriate size (8 or 10 mm in diameter) with a reverse locking taper fitting at its neck was chosen for each specimen. For all phase I specimens, 3 cobalt-chromium head components with 0-, 2-, and 4-mm offset were used. Four cobalt-chromium alloy humeral head components with offsets of 0, 4, 6, and 8 mm for each size were used in each phase II specimen. All humeral head components had the same neck length to maintain consistent capsular tensioning. Pegged glenoid components were made of ultra–high-molecular-weight polyethylene, and the articular surface had a radius of curvature 3 mm greater (6 mm of diametrical mismatch) than the respective humeral head component.

Humeral head components were sized according to the surgical technique recommended by the manufacturer (DePuy). The glenoid size is first estimated with the use of glenoid-sizing disks. A corresponding humeral head is chosen that provides 6 mm of diametrical mismatch between the glenoid and humeral articular curvatures, with the glenoid diameter of curvature being 6 mm larger than the humeral diameter of curvature. As a second check, the size of the resected head was compared with the implanted head. In all cases, the diameter of curvature of the implanted humeral head was 44 mm, 48 mm, or 52 mm. Offset humeral heads were custom manufactured so the center of the morse taper was 2, 4, 6, and 8 mm from the center of the humeral head. A peg was machined into the head to allow for accurate placement of the offset in the anterior, posterior, superior, and inferior positions.

The implantation procedure used for the current study was a modification of the standard clinical technique and is described in detail by Karduna et al.[12] In the standard technique, the subscapularis tendon is incised and repaired and the anterior/inferior capsular ligament complex is often excised during prosthetic implantation. This requires the tendon and ligament to heal to obtain a functional structure.[23] Because cadavers have no capability for healing and it was desirable to minimize surgical alterations to the normal capsular ligament tissue length, an alternative procedure was developed to minimize soft tissue damage. This consisted of an incision through the rotator interval capsule and a longitudinal split through the entire humerus through this incision. The prosthetic implantation was performed through the split humerus and the humeral head changes made through the divided rotator interval, which was repaired anatomically between each trial.

Joint motions were monitored with a 6-degrees-of-freedom magnetic tracking device (3Space Fastrak System: Polhemus, Colchester, Vt). The transmitter of the device was fixed to the lateral border of the scapula, and a scapular axis was defined on the basis of the plane of the scapula (2 points on the medial border and 1 at the center of the glenoid curvature). The receiver of the device was fixed to the medial aspect of the humerus, and a humeral axis was defined so that it was aligned with the scapular axis when the humerus was in its neutral position (center of the humeral head coinciding with the center of the glenoid, the shaft parallel to the medial border of the scapula with the condyles aligned in the plane of the scapula). The protocol for determining these axes and digitizing the points has been previously described.[13,14] Euler angles were used to represent 3 sequence-dependent rotations: the plane of elevation, the degree of elevation, and internal-external rotation, as described by An et al.[1] Translations were recorded along the superior/inferior (SI), medial/lateral (ML), and anterior/posterior (AP) axes.

The procedure was used in the collection of data for both cartilaginous and reconstructed joints in the first phase of the experiment. Joint arthroplasty did not alter the alignment of the medial border of the scapula, nor the neutral orientation of the humerus. This was not the case for the centers of the glenoid and humeral head; therefore it was necessary to define different coordinate systems for each reconstructed joint. This procedure was performed once the entire experimental protocol was completed because it was first necessary to completely excise the joint capsule. All data were thus saved in an uncalibrated format and subsequently calibrated with the appropriate axis system determined for each individual reconstructed joint.

Once the entire experimental protocol was completed, the capsule was excised. To find the center of the implanted glenoid component, a humeral head with the same radius of curvature as the implanted glenoid was placed on the humeral stem. The humerus was rotated through a central range of motion with the head centered in the glenoid. The point that moved the least in the scapula reference frame was defined as the glenoid center. Because the same glenoid component was used for all arthroplasty experiments, this center was used for all implant coordinate axis systems for a given specimen.

To find the center of a humeral head component, it needed to be articulated on a conforming glenoid component. Because the head components had a nonconforming radius of curvature with the implanted glenoid component, additional calibration glenoid components were machined to fit snugly over the implanted glenoid component. Each calibration glenoid matched the radius of curvature of the implanted, experimental humeral heads. In turn, each head was passively rotated on its matching glenoid, and the point that moved the least in the humeral reference frame was defined as the center of that head component.

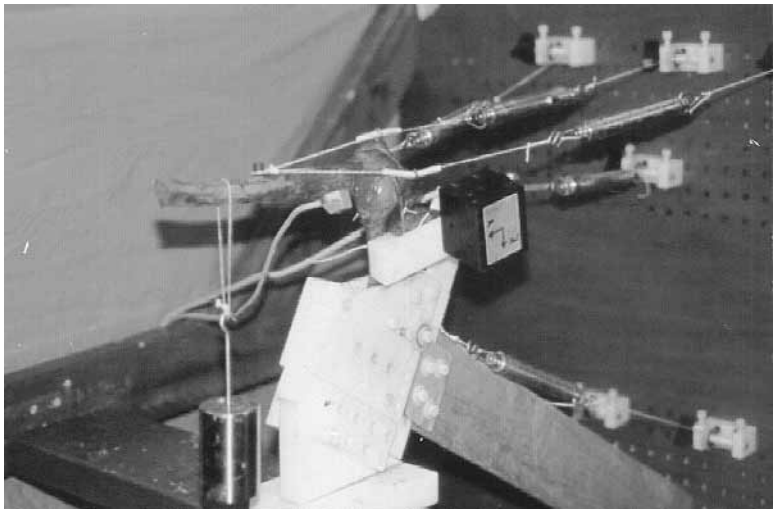

Figure 1 Photograph of the active model setup (model 1).

Because the magnetic tracking device is susceptible to errors when in the vicinity of metal objects, it was desirable to subtract out systematic errors to increase the accuracy.[9] This involved articulating the heads under conditions of no translations. The passive experimental protocol was then repeated for each head by articulation on its corresponding calibration glenoid after capsular excision. The translations recorded during these motions were then subtracted from those recorded during the actual experiment.

After capsular excision, the passive experimental protocol was repeated by using a humeral head with a radius of curvature that matched that of the implanted glenoid component. There should not have been any translation during this motion, therefore recorded translations after subtraction of systematic errors were taken as an indication of random error. Previous experiments have shown that translational error is approximately 0.4 mm at the 95% confidence level.[12,14] This is consistent with previous authors.[2,9,11] However, the measurement of the sensor was not relied on, and the systematic error was subtracted from the experimental results, as described in the section on subtracting out systematic error.

A total of 11 shoulders, with a mean age of 66 years (range, 61 to 81), were tested in phase I. The testing protocol consisted of positioning the glenohumeral joint in maximal internal rotation and then externally rotating it in 10° increments until maximal external rotation was achieved. This procedure was performed in one position: 90° of total elevation in the plane of the scapula. At each increment, plane and elevation were maintained within ±5° of their desired values, whereas the rotation was maintained within 1° of its target. This was done through the use of custom-written software that allows real-time feedback of joint position.[13,14] Once the desired position was reached, a data point was taken and saved in an uncalibrated format. Experiments were first conducted on the unreconstructed joint. Once this was completed, the arthroplasty procedure was performed and the experiment was repeated for all 9 humeral head positions in a random fashion.

Active and passive motions were performed in a random order for the natural as well as each implant joint. Active motions were achieved by applying forces to muscle insertion sites with hand cranks with spring scales to determine the amount of force (Figure 1). A 2.3-kg weight was hung on the distal humerus to approximate the weight of the arm.[21,25] Maximal rotations during active motions were set with a 60-N force on either the infraspinatus/teres minor complex or the subscapularis. Muscle forces were adjusted as necessary to achieve the desired arm position, with a maximum of 60-N force on each muscle unit. The sequence began with maximal internal rotation by applying a 60-N force to the subscapularis muscle and proceeded with progressive external rotation by applying increasing forces to the infraspinatus/teres minor while decreasing the force on the subscapularis. A completed range of motion was achieved when a 60-N force was reached by the infraspinatus/teres minor complex.

For passive motion, a medial applied centering load was achieved by applying 10 N of force to each of the rotator cuff tendons by means of hanging weights. Maximal rotations during passive motions were set by applying a torque wrench to the distal humerus. This was accomplished by attaching the hand-held torque wrench to a bolt that had been cemented within the intramedullary canal of the distal humerus. To approximate the torque achieved actively by the rotator cuff (60 N force × 25 mm), a maximal torque of 1500 Nmm was set for both internal and external rotation. The humerus was then brought from maximal internal rotation to maximal external rotation, with each end point being reached at 1500 Nmm of torque.

For the purposes of statistical analysis, 9 variables were identified: total range of motion, range of internal rotation, range of external rotation, total anterior-posterior translation, anterior-posterior translation in internal rotation, anterior-posterior translation in external rotation, total superior-inferior translation, superior-inferior translation in internal rotation, and superior-inferior translation in external rota-

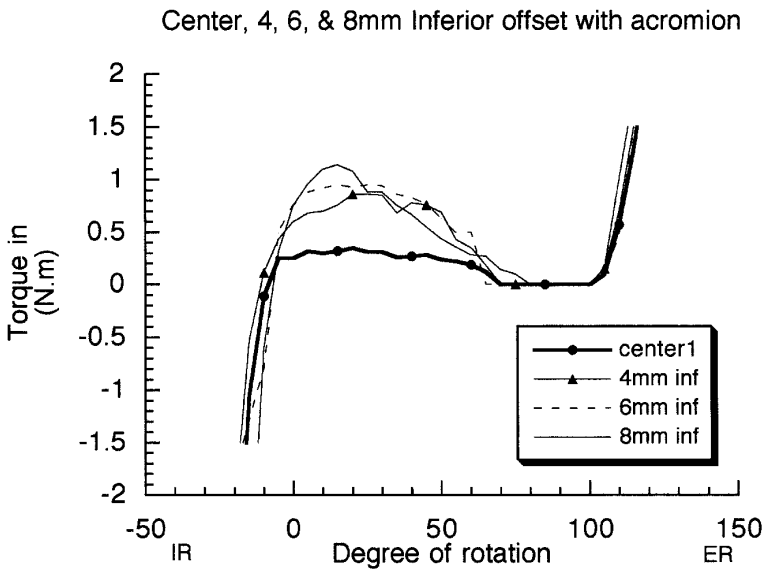

Figure 2 Graph of a representative specimen showing the relation between the amount of torque required to overcome impingement and varying degree of inferior offsets.

tion. Translation was defined as the difference in position of the center of the humeral head between maximum internal rotation and maximum external rotation, maximum internal rotation and neutral rotation, and maximum external rotation and neutral rotation. For translation and rotation (ie, range of motion), repeated-measures analysis of variance tests were performed with 2 between-subjects factors (active and passive models) and 5 within-subject factors offset (implants × 5). Two repeated-measures analysis of variance tests were run for each variable—one for superior to inferior offset and another for anterior to posterior offset. The Wilcoxon signed rank *t* test was used to test for differences between the natural shoulders and the centered arthroplasties. The acceptable rate for a type I error was chosen as 5% ($P = .05$) for all tests.

Impingement

Joint motions were monitored with a 6-degrees-of-freedom magnetic tracking device (3Space Fastrak System: Polhemus, Colchester, Vt) with the same protocol as in the kinematic portion of the experiment. The transmitter of the device was fixed to the lateral border of the scapula and a receiver was fixed to the medial aspect of the humerus, and an axis was defined for each.

A total of 10 shoulders were tested in this phase (average age, 54 years; range, 41 to 75 years). For passive glenohumeral motion, a centering load of 30 N for each of the simulated rotator cuff muscles was applied by using hanging weights and pulleys. Real-time feedback from the magnetic tracking device was used to help maintain the humerus at 90° of total elevation (30 scapula and 60 glenohumeral) in the plane of the scapula (±5°) throughout the range of motion. The humerus was then rotated from maximum internal rotation of 1500 Nmm to maximum external rotation of 1500 Nmm by using a continuous recording digital torque wrench (Computorq II, City of Industry, Calif) attached to the distal end of the humerus. Thirteen trials were performed with 0-, 4-, 6-, and 8-mm offset heads in superior, anterior, inferior, and posterior positions. Afterward, the acromion and coracoacromial ligament were removed, and the trials were repeated in a similar manner. Subacromial contact was quantified by measuring the difference in peak torque during the impingement zone of motion with and without the acromion (Figure 2). Passive glenohumeral motion was measured from maximum internal rotation to maximum external rotation with the magnetic tracking device.

A repeated-measures analysis of variance was performed for each group (superior, anterior, inferior, and posterior) with 4 within-subject factors: center, 4-, 6-, and 8-mm offsets. If the overall analysis was significant, an appropriate multiple comparison procedure was done. Comparisons were made between offsets of the same direction (ie, superior) but not between offsets of different directions. The acceptable rate for a type I error was chosen as 5% ($P = .05$) for all tests.

RESULTS

Phase I: Kinematics

The model effect: Active versus passive kinematics. Means of range of motion and translation were all significantly greater for the passive model when testing offset in the anterior to posterior direction, with the exception of 2 variables: superior-inferior translation in external rotation ($P = .09$) and total superior-inferior translation ($P = .5$) (Table I). The mean glenohumeral range of motion during passive positioning (144°) was significantly greater than the range of motion achieved during active positioning (113°) ($P = .0005$) (Figure 3). Total anterior-

Table I Active and passive rotation and translation means

	4 mm Anterior	2 mm Anterior
Anterior-posterior offset		
AP in IR	–0.7 ± 0.8: 2.8 ± 1.6	–1.1 ± 0.8: 2.5 ± 1.5
AP in ER	0.2 ± 1.3: –1.3 ± 1.9	0.8 ± 1.1: –0.8 ± 1.7
AP total*	1.7 ± 1.2: 4.7 ± 2.4	2.0 ± 0.9: 3.8 ± 1.1
SI in IR	0.9 ± 1.0: 0.2 ± 2.0	1.5 ± 1.0: –0.3 ± 2.2
SI in ER	1.1 ± 1.2: –0.5 ± 3.2	1.0 ± 0.6: –0.6 ± 2.8
SI total	2.5 ± 1.3: 3.4 ± 3.4	2.4 ± 1.1: 3.0 ± 3.0
IR*	51 ± 16: 66 ± 23	56 ± 15: 69 ± 24
ER*	63 ± 13: 74 ± 14	63 ± 14: 75 ± 12
ROM*	114 ± 15: 140 ± 23	119 ± 15: 144 ± 24
	4 mm Superior	**2 mm Superior**
Superior-inferior offset		
AP in IR	–0.7 ± 0.7: 4.6 ± 3.6	–0.7 ± 1.1: 3.3 ± 3.9
AP in ER*	1.0 ± 0.9: –1.0 ± 0.9	1.4 ± 1.1: –0.9 ± 0.8
AP total	1.7 ± 1.2: 5.8 ± 3.5	2.3 ± 1.5: 5.1 ± 2.8
SI in IR	0.7 ± 0.6: 0.2 ± 1.7	1.4 ± 1.6: –0.1 ± 1.9
SI in ER	0.1 ± 1.3: 0.2 ± 1.2	0.8 ± 1.3: –0.1 ± 1.9
SI total*	1.7 ± 1.1: 2.3 ± 1.1	2.7 ± 1.6: 2.5 ± 2.5
IR*	48 ± 17: 66 ± 22	53 ± 15: 71 ± 21
ER	58 ± 16: 73 ± 13	61 ± 13: 73 ± 13
ROM*	105 ± 17: 139 ± 26	114 ± 12: 143 ± 23

All values given as active model: passive model with standard deviations. Translation values are in millimeters. Rotation values are in degrees. Positive translation values are anterior or superior.
AP, Anterior-posterior translation; *IR,* internal rotation; *ER,* external rotation; *SI,* superior-inferior translation; *ROM,* range of motion.
*$P < .05$ offset effect.

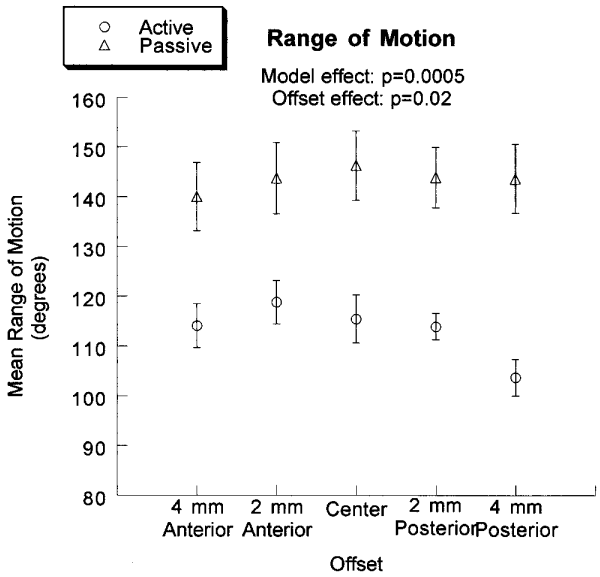

Figure 3 Graph depicting mean total range of motion for anterior to posterior offset of the humeral head for both active and passive models. *Error bars* represent standard error of the mean.

posterior translation was significantly greater during passive positioning (5.0 mm) when compared with active positioning (2.1 mm) ($P = .0001$) (Figure 4). The model effect (active versus passive positioning) was also significant with respect to internal rotation ($P = .01$), external rotation ($P = .02$), anterior-posterior translation in internal rotation ($P = .0001$), anterior-posterior translation in external rotation ($P = .0001$), and superior-inferior translation in internal rotation ($P = .03$).

As with anterior-posterior offset of the humeral head, analysis of the translation and range of motion achieved with superior-inferior offset yielded significantly greater values for the passive positioning than for active positioning, with the exception of superior-inferior translation in external rotation ($P = .2$) and total superior-inferior translation ($P = .8$). Total range of motion during passive positioning averaged 143° and for active positioning averaged 113° ($P = .001$). Total anterior-posterior translation was significantly greater for passive positioning (4.9 mm) than for active positioning (2.3 mm) ($P = .0003$). The model effect (active versus passive positioning) was also significant for internal rotation ($P = .01$), external rotation ($P = .03$), anterior-posterior translation in internal rotation ($P = .0001$), anterior-posterior translation in external rotation ($P = .0001$), and superior-inferior translation in internal rotation ($P = .01$).

The offset effect. When the 9 variables representing rotation and translation were analyzed with regard to anterior to posterior offset, statistically significant associations were noted in 4 of the categories: range of motion ($P = .02$) (Figure 3), internal rotation ($P = .02$), external rotation ($P = .0001$), and total anterior-posterior transla-

Centered	2 mm Posterior	4 mm Posterior
–1.6 ± 1.2: 3.2 ± 1.9	–1.3 ± 1.6: 4.2 ± 2.0	–1.5 ± 1.8: 3.1 ± 5.5
0.3 ± 0.6: –1.5 ± 1.1	1.0 ± 0.7: –1.3 ± 1.4	0.7 ± 0.5: –1.3 ± 1.0
2.1 ± 1.2: 4.4 ± 2.0	2.6 ± 1.3: 5.6 ± 1.6	2.3 ± 1.8: 6.5 ± 3.0
1.8 ± 2.1: –0.3 ± 2.1	1.7 ± 2.1: –0.4 ± 2.6	1.0 ± 1.7: –0.4 ± 2.7
0.4 ± 1.2: –0.7 ± 2.5	0.8 ± 1.2: –0.2 ± 1.3	0.2 ± 1.2: –0.6 ± 2.4
3.0 ± 1.8: 3.3 ± 3.2	2.7 ± 2.2: 2.9 ± 2.2	2.0 ± 1.5: 3.5 ± 3.5
54 ± 15: 74 ± 21	56 ± 13: 74 ± 18	50 ± 15: 78 ± 20
61 ± 14: 72 ± 12	58 ± 11: 70 ± 13	53 ± 13: 66 ± 13
115 ± 15: 146 ± 23	114 ± 9: 144 ± 20	104 ± 12: 144 ± 23
Centered	**2 mm Inferior**	**4 mm Inferior**
–1.6 ± 1.2: 3.2 ± 1.9	–1.5 ± 1.7: 2.9 ± 2.3	–2.2 ± 1.7: 2.4 ± 1.7
0.3 ± 0.6: –1.5 ± 1.1	0.6 ± 0.7: 1.1 ± 1.8	–0.01 ± 0.9: –1.5 ± 1.6
2.1 ± 1.2: 4.4 ± 2.0	2.4 ± 1.7: 4.8 ± 1.7	2.8 ± 1.7: 4.3 ± 1.6
1.8 ± 2.1: –0.3 ± 2.1	1.5 ± 1.4: –0.5 ± 1.9	2.2 ± 1.6: 0.5 ± 2.1
0.4 ± 1.2: –0.7 ± 2.5	1.0 ± 0.6: –0.6 ± 3.1	1.1 ± 0.7: 0.2 ± 3.1
3.0 ± 1.8: 3.3 ± 3.2	2.6 ± 1.5: 2.9 ± 3.7	3.3 ± 1.5: 3.8 ± 3.1
54 ± 15: 74 ± 21	57 ± 13: 73 ± 19	58 ± 14: 71 ± 19
61 ± 14: 72 ± 12	59 ± 20: 72 ± 13	57 ± 13: 69 ± 14
115 ± 15: 146 ± 23	116 ± 19: 144 ± 19	115 ± 10: 140 ± 19

tion (P = .01) (Figure 4). Posterior offset of the humeral head decreased active total range of motion and internal rotation by 10% (115° to 104° and 56° to 50°, respectively) and decreased active external rotation to 53° from 61°.

The effect of superior to inferior offset on translation and rotation was only significant for 4 variables: range of motion (P = .02), internal rotation (P = .0001), anterior-posterior translation in external rotation (P = .01), and total superior-inferior translation (P = .03). In the active model, superior positioning of the humeral head as compared with the centered head decreased range of motion from 115° to 105°. Both anterior and posterior positioning of the head decreased passive range of motion as compared with the centered position. The effect of offset on internal rotation was similar in both the active and the passive models. As the humeral head was moved from the superior offset to inferior offset, the amount of internal rotation increased.

Natural versus centered head. In all categories of range of motion and translation in the active model, the natural joints showed no statistical differences when compared with the means of the centered implants (Table II). The centered arthroplasty showed a trend toward increasing total superior-inferior translation (3.0 mm) when compared with the natural shoulders (1.4 mm), although this result was not statistically significant (P = .055).

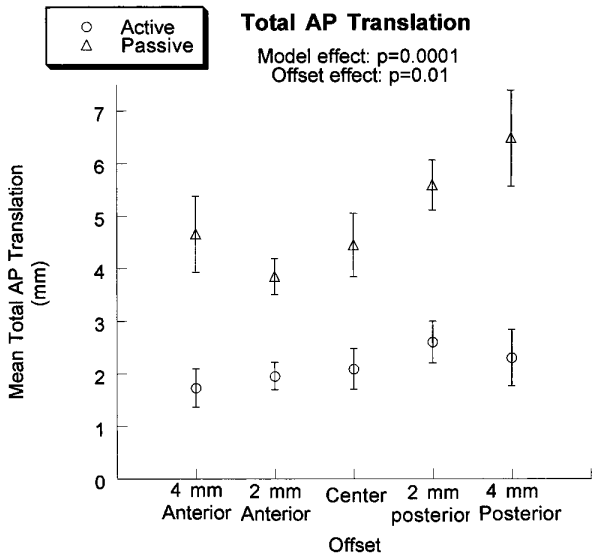

Figure 4 Graph demonstrating mean total anterior-posterior *(AP)* translation of the humeral head for anterior to posterior offset of the humeral head for both active and passive models. *Error bars* represent standard error of the mean.

There was no significant difference between the mean values for range of motion and translation of the natural shoulder and the centered arthroplasty, with two exceptions. Mean internal rotation of the natural

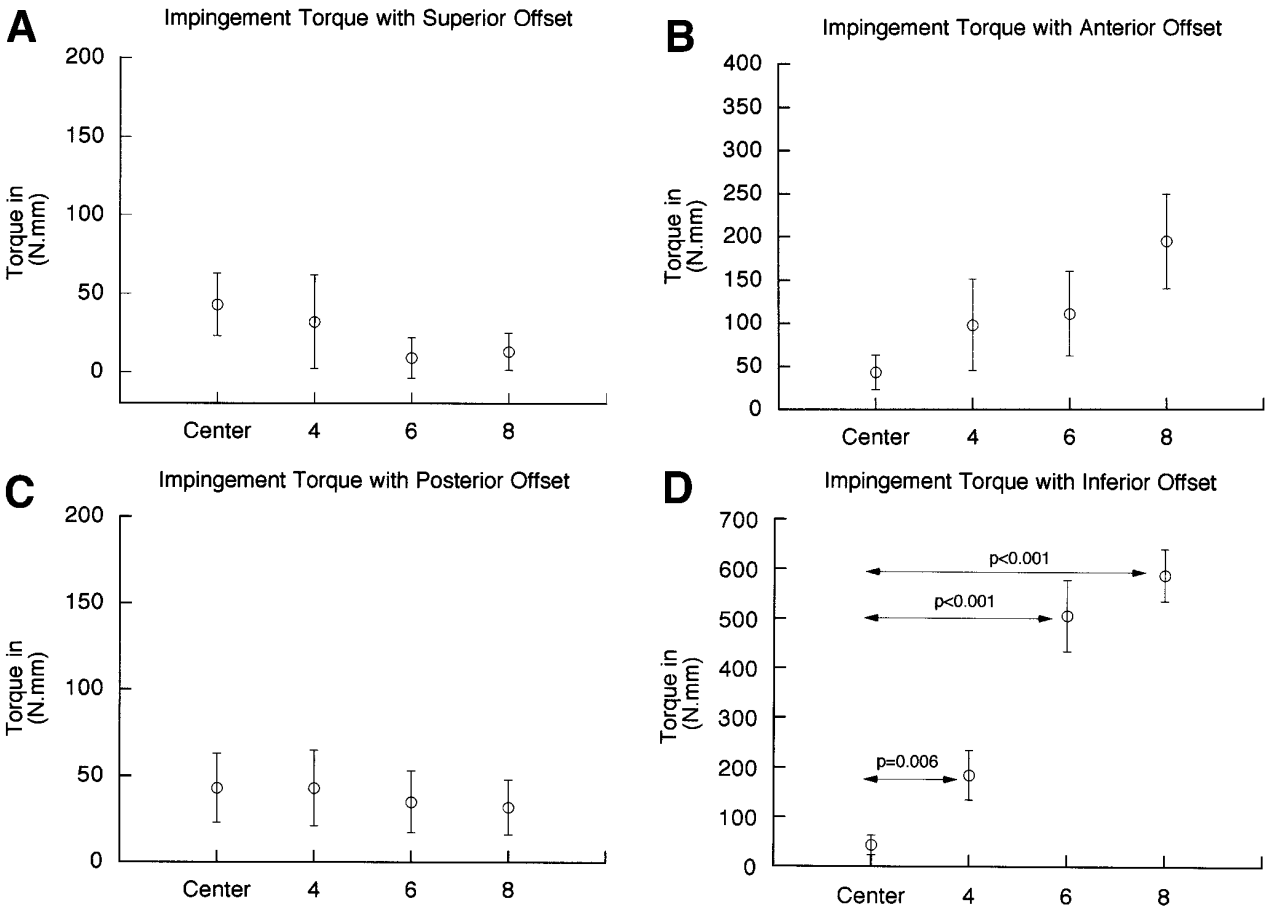

Figure 5 A through **D,** Graphs show the difference in impingement torque (Nmm) between center, 4-mm, 6-mm, and 8-mm offsets. *Error bars* represent standard error of the mean. **A,** Impingement torque with superior offsets. **B,** Impingement torque with anterior offsets. **C,** Impingement torque with posterior offsets. **D,** Impingment torque with inferior offsets. *P* values are shown for specific pairwise comparisons.

Table II Intact joint compared with centered arthroplasty

	Intact joint	Centered arthroplasty
AP in IR	–0.9 ± 1.5: 2.3 ± 1.9	–1.6 ± 1.2: 3.2 ± 1.9
AP in ER	0.2 ± 1.5: –4.1 ± 2.9*	0.3 ± 0.6: –1.5 ± 1.1*
AP total	2.2 ± 1.3: 6.9 ± 3.3	2.1 ± 1.2: 4.4 ± 2.1
SI in IR	0.5 ± 1.0: 0.3 ± 1.5	1.8 ± 2.1: –0.3 ± 2.1
SI in ER	–0.3 ± 0.6: –1.1 ± 2.8	0.4 ± 1.2: 0.7 ± 2.5
SI total	1.4 ± 0.6: 3.6 ± 1.9	3.0 ± 1.8: 3.3 ± 3.2
IR	50 ± 14: 65 ± 14*	54 ± 15: 74 ± 21*
ER	57 ± 17: 76 ± 14	61 ± 14: 72 ± 12
ROM	107 ± 14: 141 ± 11	115 ± 15: 146 ± 23

All values given as active model: passive model with standard deviations. Translation values are in millimeters. Rotation values are in degrees. Positive translation values are anterior or superior. *AP,* Anterior-posterior translation; *IR,* internal rotation; *ER,* external rotation; *SI,* superior-inferior translation; *ROM,* range of motion.
*$P < .05$.

shoulders was 65° and for the shoulders with a centered arthroplasty was 74° ($P = .01$). Anterior-posterior translation in external rotation was also significantly greater for the natural shoulders (4.1 mm posterior) than for the shoulders with a centered arthroplasty (1.5 mm posterior) ($P = .03$).

Phase II: Impingement

Effects of malposition on subacromial impingement. The data were analyzed, comparing the difference in mean peak torque with and without the acromion, which is defined as the impingement torque. Any increase in torque was attributed to an increase in subacromial contact because that increase disappeared when the acromion and coracoacromial ligament was removed. The impingement torque with superior offsets was 43 ± 20 Nmm, 32 ± 30 Nmm, 9 ± 13 Nmm, and 13 ± 12 Nmm for center, 4-, 6-, and 8-mm offsets, respectively

(Figure 5, *A*). There was no significant difference in torque between the center position and the offsets position. The impingement torque for anterior offsets was 98 ± 53 Nmm, 111 ± 49 Nmm, and 195 ± 55 Nmm for 4-, 6-, and 8-mm offsets, respectively (Figure 5, *B*). Again, there was no significant difference between the center position and the offset position. With posterior offsets, the change in mean peak torque was 43 ± 22 Nmm for 4-mm offset, 35 ± 18 Nmm for 6-mm offset, and 32 ± 16 Nmm for 8-mm offset (Figure 5, *C*). There was no significant change in impingement torque comparing center and the offsets. However, with inferior offsets, the change in mean peak torque was 185 ± 50 Nmm for 4-mm offset, 507 ± 72 Nmm for 6-mm offset, and 589 ± 52 Nmm for 8-mm offset (Figure 5, *D*). There was a significant increase in torque from the centered position to 4-mm inferior offset (P = .006), 6-mm inferior offset (P = .001), and 8-mm inferior offset positions (P = .001).

Effect of malposition on passive motion. The range of motion for a given experiment was defined as the angle between the points of maximum internal and external rotations. For superior offsets, the total range of motion was 129°, 125°, 127°, and 119° for center, 4-, 6-, and 8-mm offsets, respectively. There was no significant difference in range of motion between center, 4-, and 6-mm offsets, but there was a significant decrease in range of motion between center and 8-mm superior offset (P = .002). For anterior offsets, the total range of motion for 4-, 6-, and 8-mm offsets was 128°, 126°, and 118°, respectively. There was a significant decrease in motion from center to 8-mm anterior offset (P = .014) but not from center to 4- and 6-mm offsets. For inferior offsets, the total range of motion for 4-, 6-, and 8-mm offsets was 127°, 126°, and 118°, respectively. There was a significant decrease in motion from center to 8-mm inferior offset (P = .001). As for posterior offsets, the range of motion for 4-, 6-, and 8-mm offsets was 125°, 123°, and 114°, respectively. There was no significant difference between center, 4-, and 6-mm offsets, but there was a significant decrease in motion between center and 8-mm posterior offset (P = .001).

DISCUSSION

Previous studies of the natural glenohumeral joint have demonstrated the effects of muscle forces, ligament lengths, and joint conformity on translation patterns during both active and passive joint positioning.[12,14] With active positioning, the humerus nearly exhibits ball-and-socket kinematics, with only minimal translations that are dependent primarily on joint conformity.[12,14,15,24] Passive joint positioning causes much larger translations that are primarily dependent on ligamentous constraints.[12,14] This increased translation occurs at the extremes of motion not achieved by active joint positioning.

The differences in joint kinematics seen in the active and passive models in both normal shoulders and after total shoulder arthroplasty are expected and have been reported by previous authors.[9,12,14,15] In agreement with Karduna et al,[12,14] we observed significantly greater ranges of motion with the passive model than the active, even when total motion was broken down into its component parts. Total anterior-posterior translation, as expected, was also significantly greater for the passive model. Total superior-inferior translation and superior-inferior translation in external rotation, however, were not statistically different between the two models. It is unclear why total superior-inferior translation and superior-inferior translation in internal rotation were not statistically different between the active and passive models. This may be caused by a greater percentage of total superior-inferior translation that was found to occur in internal rotation in the active model.

The differences in range of motion in the active model seen in shoulders with offset head positions as compared with those with centered implants were small but may be clinically relevant. The posterior position of the humeral head limited active total range of motion by 10% when compared with the centralized component, whereas the anterior position had no effect. The same holds true when the data were analyzed only in external rotation, with a loss of 13% of active motion with the head in the posterior position. One possible explanation for the loss of external rotation and total range of motion with posterior head positioning in the active model is the decreased moment arm of the external rotators as well as their altered lines of action.

The effect of humeral head position on glenohumeral translation was statistically significant for 1 of the 6 translational variables tested with anterior to posterior offset and 2 of the 6 variables tested for superior to inferior offset. The anterior to posterior offset effect was significant for total anterior-posterior translation; the superior to inferior offset effect was significant for anterior-posterior translation in external rotation and total superior-inferior translation. Humeral head offset could be expected to affect translation during active positioning because of alterations in rotator cuff lines of pull and moment arms. It is possible that a significant offset effect may have been encountered for more of the translational variables analyzed than we encountered in our study if greater degrees of offset had been tested. Our data indicate that articular malposition of 4 mm or less results in small alterations in glenohumeral translations. The optimal amount of translation after total shoulder arthroplasty is not known. It would seem intuitive that a recreation of normal anatomy after shoulder replacement would allow for optimum translation, therefore maximizing range of motion and longevity of the prosthesis.

During passive joint positioning, glenohumeral translations and ranges of motion are dependent on capsuloligamentous tensions.[9,12,14] In the current study, the location of the head was varied, possibly

causing selective anterior, posterior, superior, or inferior tightening. This could result in increased translation in the direction opposite of the tightened ligament and decreased range of motion because of early tightening. Our phase I data support mildly diminished passive range of motion with the offset humeral heads. There was a 9% decrease in external rotation with posterior offset of the humeral head of 4 mm and a 10% loss of internal rotation with superior offset of 4 mm.

Inferior malposition of the humeral head component below the level of the greater tuberosity can lead to subacromial impingement, but the threshold at which this occurs is not known. Phase II of our study demonstrated that inferior malposition of as little as 4 mm leads to a significant increase in subacromial contact. An increase in subacromial contact can be undesirable after total shoulder arthroplasty because it could lead to pain, decreased range of motion, and rotator cuff degeneration.[18] Moreover, subacromial impingement could also potentially cause eccentric loading of the glenoid component and early glenoid component loosening.[10]

In our model, we were unable to demonstrate a significant increase in subacromial contact with superior, anterior, and posterior offsets of up to 8 mm of malposition. With superior offsets, the distance between the acromion and greater tuberosity is increased, making impingement highly unlikely. With anterior offsets, the tuberosities are displaced posterior in the neutral rotation but superior in the elevated and internally rotated position. Although subacromial impingement was observed in 5 of 10 specimens, the data failed to show a statistical increase in mean peak torque with increasing offsets. With posterior offsets, the tuberosities are displaced anteriorly in neutral rotation and superiorly during abduction and external rotation. Removing the acromion and coracoacromial ligament did not affect the mean peak torque value with 4-, 6-, and 8-mm posterior offsets. During the experiment, the coracoid was left intact to preserve the coracohumeral ligament. Therefore any differences in torque related to contact of the tuberosities with the coracoid could not be identified, even when the acromion and coracoacromial ligament were removed.

The small effect of articular malposition on glenohumeral motion seen in phase I of this study was seen to be much greater in phase II, presumably because of the greater degree of malposition. The results demonstrated that malposition of up to 6 mm in any direction from the center of the osteotomy surface did not cause a significant decrease in passive range of motion under the testing conditions present in phase II. However, 8 mm of malposition in any direction led to a significant decrease in glenohumeral motion. A decrease in range of motion from abnormal soft tissue tensioning or from mechanical impingement after total shoulder arthroplasty is undesirable. In addition to compromising the functional result, it could lead to early capsular tightening and obligate glenohumeral translation with potential for increased glenoid component wear.[10]

In conclusion, the results of this study demonstrate that humeral head malposition of 4 mm or less during total shoulder arthroplasty has small but statistically significant effects on glenohumeral translations and motion. Inferior malposition of as little as 4 mm causes increased subacromial contact. Furthermore, articular malposition of 8 mm in any direction results in a significant decrease in passive range of motion. If subacromial contact is to be minimized and glenohumeral motion maximized after shoulder replacement, anatomic reconstruction of the humeral head–humeral shaft offset to within 4 mm is desirable.

Our experimental results and conclusions are limited by the fact that we chose only one glenohumeral position and one type of motion (maximum internal to maximum external rotation) to study. Rotational motion with the shoulder positioned in 90° of elevation tests the inferior capsule, which has not been disturbed by the arthroplasty technique. Despite these limitations, we hope that the data will provide some guidance regarding the tolerable limits of humeral articular malposition during total shoulder arthroplasty.

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