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Effect of deltoid tension and humeral version in reverse total shoulder arthroplasty: a biomechanical study

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Background: No clear recommendations exist regarding optimal humeral component version and deltoid tension in reverse total shoulder arthroplasty (TSA).
Materials and methods: A biomechanical shoulder simulator tested humeral versions (0°, 10°, 20° retroversion) and implant thicknesses (−3, 0, +3 mm from baseline) after reverse TSA in human cadavers. Abduction and external rotation ranges of motion as well as abduction and dislocation forces were quantified for native arms and arms implanted with 9 combinations of humeral version and implant thickness.
Results: Resting abduction angles increased significantly (up to 30°) after reverse TSA compared with native shoulders. With constant posterior cuff loads, native arms externally rotated 20°, whereas no external rotation occurred in implanted arms (20° net internal rotation). Humeral version did not affect rotational range of motion but did alter resting abduction. Abduction forces decreased 30% vs native shoulders but did not change when version or implant thickness was altered. Humeral center of rotation was shifted 17 mm medially and 12 mm inferiorly after implantation. The force required for lateral dislocation was 60% less than anterior and was not affected by implant thickness or version.
Conclusion: Reverse TSA reduced abduction forces compared with native shoulders and resulted in limited external rotation and abduction ranges of motion. Because abduction force was reduced for all implants, the choice of humeral version and implant thickness should focus on range of motion. Lateral dislocation forces were less than anterior forces; thus, levering and inferior/posterior impingement may be a more probable basis for dislocation (laterally) than anteriorly directed forces.
Level of evidence: Basic Science Study.

Keywords: Shoulder; reverse arthroplasty; deltoid tension; humeral version; biomechanical simulator

Reverse total shoulder arthroplasty (TSA) has revolutionized the treatment of arthritic, rotator cuff–deficient shoulders by reliably improving shoulder pain and function.[14,33] Successful outcomes ideally maximize range of motion and minimize instability and the risk of acromial fractures. Despite widespread use since Grammont's original report,[12] few data exist regarding the effects of key components of the surgery. Boileau et al[5] reported that the intraoperative determination of deltoid tension is difficult and guided mostly by surgical experience. There are also no consistent recommendations regarding humeral component version.[9,20]

Investigational Review Board approval was not required for this study.

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Several aspects of reverse TSA have been biomechanically investigated. The effects of glenosphere size, vertical position, version and tilt, as well as humeral component version and inclination angle on range of motion and stability, have been examined in biomechanical cadaver models.[9,15,27] To our knowledge, no authors have simultaneously investigated the effects of humeral version and deltoid tension on kinematics or stability in a cadaver model. Favre et al[9] examined stability of the humeral and glenoid components without implantation, neglecting potential stabilizing effects of the soft tissue, specifically deltoid tension. Ackland et al[1] evaluated reverse TSA in a biomechanical simulator, but only reported changes in center of rotation and abductor moment arms with respect to the intact shoulder.

The present study evaluated the kinematics and dislocation forces of a reverse TSA in a soft tissue−constrained cadaver model using a shoulder simulator. The objective was to determine if proximal humeral version and deltoid tension (implant insert thickness) in reverse TSA significantly affected the (1) range of motion in glenohumeral abduction, (2) degree of external rotation of a flexed arm, (3) force required to abduct the arm, and (4) force required to dislocate the implant in a lateral or anterior direction.

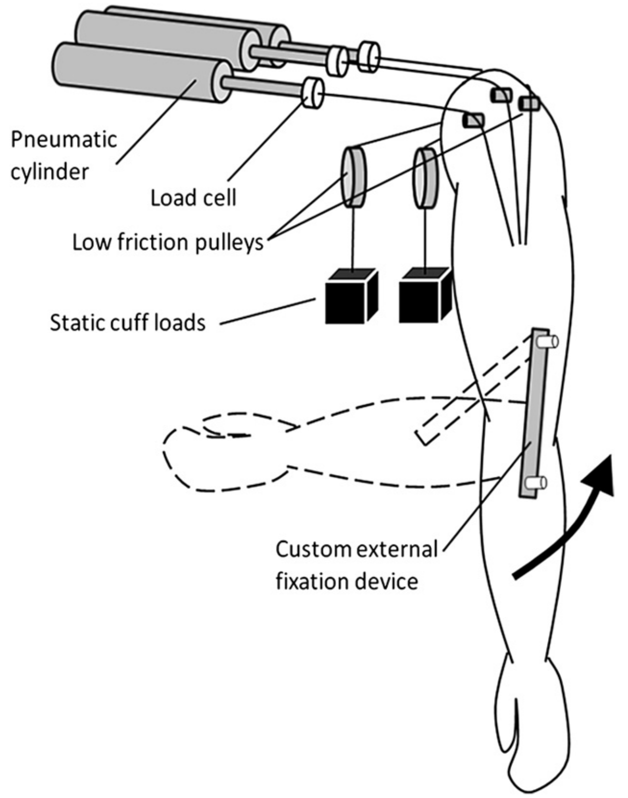

Figure 1 Schematic of the shoulder simulator. The scapula was potted and rigidly mounted to the simulator such that the glenohumeral joint approximated anatomic orientation. Pneumatic cylinders applied displacement to the deltoid insertion to abduct the arm in the scapular plane while load cells recorded force. Static loads were applied to the insertions of the rotator cuff muscles to seat the humerus on the glenoid. The elbow was locked in straight or bent positions with custom external fixation. Arm kinematics were quantified by 3-dimensional optically tracking diode arrays on the fixation pins.

Materials and methods

Specimen preparation

Six fresh frozen, nonpaired, upper extremities (3 men, 3 women; 3 right, 3 left) were obtained from deceased donors who were a mean $\pm$ SD age of 63.7 $\pm$ 5.3 years and had a body weight (BW) of 73.6 $\pm$ 19.2 kg. Arms were thawed at $\sim$20° C for 18 hours and kept hydrated with normal saline during preparation. The scapula was exposed from the medial border to the suprascapular notch before being embedded in a rectangular block of 2-part catalyzed polymer resin (3M, St. Paul, MN, USA).

Computed tomography (CT) scans of each specimen were acquired to verify the orientation of the scapula within the embedding block. CT slices were reconstructed into 3-dimensional models with Mimics (Materialise, Leuven, Belgium), and the anatomy was referenced to the planar surfaces of the block. The plane of the scapula was defined as connecting the most dorsal aspect of the inferior angle, the intersection of the scapular spine and medial border, and the center of the glenoid (intersection of the vertical and horizontal glenoid midlines). Glenoid tilt was defined from the superior to inferior margins of the glenoid.[8] Anterior scapular tilt was defined by the plane connecting the superomedial and inferior angles of the scapula with the glenoid center.[34,36]

Bicortical pins were placed distal to the deltoid tuberosity and proximally in the ulna for external elbow fixation. Through a small incision, 3 lines of 300# Spectra (WSK, Pittsburgh, PA, USA)[16] were anchored to the deltoid tuberosity with bicortical screws to simulate the anterior, middle, and posterior heads of the deltoid. A deltopectoral approach accessed the subscapularis (SSc) and a modified posterior approach, parallel to the deltoid muscle fibers, accessed the supraspinatus (SS) and infraspinatus/teres-minor (IS/TM).[19] FiberWire (size 2, Arthrex, Naples, FL, USA) was used to suture the Spectra lines to the insertions of the SSc, SS, and IS/TM on the proximal humerus. Rotator cuff lines were routed along the midline of the muscle bellies and maintained by pulleys fixed to the embedding block.[18,36] The native rotator cuff was preserved, but 1 specimen presented with a ruptured SS at preparation.

Shoulder simulator

Specimens were tested on a custom biomechanical shoulder simulator (Fig. 1). Anatomic orientations calculated from CT data were used to mount each scapula to the simulator on a 6 degree-of-freedom jig, with the glenoid tilted 10° superiorly,[3,8] the scapula tilted 10° anteriorly,[18,36] and the plane of the scapula parallel to the applied deltoid load.

The deltoid lines were routed using custom 6 mm-diameter Delrin (DuPont, Wilmington, DE, USA) pulleys rigidly suspended from the simulator frame. A slotted Delrin guide allowed the deltoid lines a lateral degree of freedom (<5 mm) along the pulley-bearing surface to prevent binding or dislocation of the dynamically changing line of action. The anterior deltoid pulley was positioned 5 mm lateral to the anterolateral corner of the coracoid. The middle deltoid pulley was positioned 5 mm lateral to the acromion, midway between the anterolateral and posterolateral corners.[18,25,36] The posterior deltoid pulley was positioned 5 mm superior to the scapular spine midway along the insertion of the posterior deltoid. Anatomic landmarks were located by palpation. The rotator cuff lines of action were suspended over pulleys, and static weights were applied (described subsequently).[28,38]

The arm was prepared with custom external fixation of the elbow to test the influence of straight and bent arms (90° elbow flexion) on native and TSA cases. In addition, the wrist was splinted and wrapped in Coban (3M) to stabilize the forearm.

Pneumatic cylinders (Bimba, Monee, IL, USA) with 250-lb load cells (Omega Technologies, Stamford, CT, USA) were attached to the deltoid lines of action. Electromechanical encoders (Celesco, Chatsworth, CA, USA) quantified the cylinder position throughout the loading cycle while the load cells recorded the applied force. Custom code that was written using LabVIEW software (National Instruments Corp, Austin, TX, USA) maintained a constant force, allowing for changes in cylinder stroke, or followed a cylinder displacement trajectory.

The control scheme set the contribution of each cylinder as a percentage of the total desired force, as measured by the load cells. In the load-control mode, each cylinder retracted until its relative percentage of the total load was reached. In the position-control mode, an equal tonic load was applied to each cylinder (using load control), and the arm was manually articulated through glenohumeral abduction in the scapular plane. The cylinders retracted and extended as the excursion of the deltoid lines changed, and their paths were recorded. The displacement trajectories were played back to repeat the desired motion path of the arm.

Optical tracking diode arrays (Optotrak 3020, Northern Digital Inc, Waterloo, ON, Canada) were mounted to the distal humerus and ulna anchors to record abduction and internal/external rotation. The array profiles were fit to a sphere to calculate the humeral center of rotation using least-squares minimization.[16,34] Two arrays were mounted to the simulator frame and the embedding block to verify that the structural components of the simulator were rigid. Three digitized points were collected from each surface of the embedding block so data from specimens could be transformed back to the respective CT coordinate systems. This allowed the center of rotation of both native and implanted arms to be compared with respect to the glenoid and scapular plane.

Experimental protocol

Specimens were first tested in the native state. A 2% BW load was applied to each rotator cuff line (SSc, SS, IS/TM), using static weights, to seat the humeral head in the glenoid. In load control, a 2% BW load was applied to each of the deltoid actuators. Cuff loads of 10.7 to 19.3 N were similar to those used by other investigators (5-40 N).[2,11,18,21,28]

The arm was then manually articulated through the physiologic range of motion for calculation of the humeral center of rotation. In position control, a trajectory was recorded from the resting abduction position to 65° of glenohumeral abduction in the scapular plane. The trajectory was played back, and the force/position profile was recorded. This procedure was repeated for the 90° elbow. Pilot testing determined 16% BW on the IS/TM resulted in slight external rotation of native arms from the neutral position at 60° of abduction. Thus, for elbow flexion, 16% BW (85.6-154.1 N) was applied to the IS/TM. The testing order (straight vs flexed elbow) was randomized using the random permuted blocks method.[24]

After native specimens were tested, they were implanted with a Tornier Aequalis Reverse Shoulder prosthesis consisting of a 36 mm glenosphere and a 6.5 mm humeral stem (Tornier, Edina, MN, USA), following Tornier's recommended surgical technique. The SSc line of action was retained at the insertion, although the muscle was resected for implantation. The SS tendon was resected to simulate a disrupted rotator cuff.

All surgeries were performed by orthopedic surgeons and coauthors (R.B. or R.T.), who together have performed more than 150 clinical reverse procedures. A baseline of 10° of humeral component retroversion was chosen for each specimen by aligning the humeral insertion guide with the forearm. To test multiple versions, the humeral stem was removed and reimpacted to the same level between versions. To limit damage to the bone, the humeral component was stabilized with removable 2-part vinyl polysiloxane impression material (Examix NDS, GC, Alsip, IL, USA).

The baseline polymer insert (for deltoid tension) was selected subjectively to provide secure reduction of the joint that minimized gap formation (<2 mm) and implant levering throughout the arm range of motion. The chosen insert was assigned as "baseline (0 mm)" and spacers of −3 mm and +3 mm from baseline were selected. A 9-mm spacer was used as baseline in 5 cases and a 12 mm spacer was used in 1 case. This study design intended to mimic small differences in the starting joint tension experienced intraoperatively, while capturing the gross effect of changing insert thickness and humeral version in the laboratory.

A randomized test procedure was also used for implanted arms. Each condition was repeated for 5 cycles and averaged. First, all inserts were tested for the baseline version for the straight and flexed elbow (order randomized). Again, 16% BW was applied to the IS/TM for the flexed arm. The SSc was loaded as in the native case, but the resected SS was not loaded. The humeral component was removed, the version was changed to 0° or 20° retroversion (order randomized), and the polysiloxane stabilizing media was replaced. All inserts were again tested in randomized order for straight and flexed elbows. The procedure was repeated for the final humeral version for a total of 18 implant cases (3 insert, 3 version, straight and flexed elbow).

The force to dislocate the implant was tested after each flexed elbow condition was completed for a given humeral version. A Spectra line was fastened around the proximal humerus at the metaphysis of the humeral component. One investigator (H.H.) applied a manual force, through a cable and in-line load cell, to dislocate the implant while the elbow and wrist of the specimen were restrained. Again in random order, the flexed arm was dislocated laterally and anteriorly for each insert. Lateral dislocation held the arm at neutral external rotation in the resting abduction position with a 2% BW load applied to the SSc, IS/TM, and deltoid. The load cell recorded the laterally applied force and the implant was said to have dislocated when a gap of ∼5 mm formed between the glenosphere and insert. For anterior dislocation, the arm was manually held in 90° external rotation at resting abduction under a 2% BW load. The load cell recorded the anteriorly directed force, and dislocation occurred when the humeral

Table I Change in joint center of rotation after reverse arthroplasty*

First author	Implant	X (mm, anterior +)	Y (mm, superior +)	Z (mm, medial +)
		Mean ± SD	Mean ± SD	Mean ± SD
Present study	Tornier	3.0 ± 2.0	−12.3 ± 3.6	17.4 ± 4.3
Ackland (2010)[1]	Zimmer	...	−9.5 ± 4.1	20.9 ± 3.9
De Wilde (2005)[8]	Delta	0 ± 0	−5 ± 1	28 ± 1
Saltzman (2010)[30]	Delta	0.2 ± 1.3	−6.9 ± 3.1	19.3 ± 2.5
Saltzman (2010)[30]	Encore	−0.4 ± 1.3	−2.0 ± 3.0	28.0 ± 3.3

SD, standard deviation.
* Values reported with respect to preoperative center of rotation.

component released around the glenosphere. Eighteen dislocations were performed for each arm.

Data analysis

The outcome measures were humeral center of rotation, resting abduction angle (an increase vs native is considered adduction deficit), cumulative deltoid force (sum of anterior, middle, and posterior deltoid), external rotation at 60° abduction (flexed elbow, deviation from neutral), and force to dislocate the implant. Coefficients of variance over 5 cycles were 1% to 3% for resting abduction angle, 5% to 10% for deltoid force, and 2% to 10%, for external rotation angle. Therefore, with the exception of dislocation, 5 cycles of abduction were averaged to generate a single representative data set for each condition. All statistical comparisons were performed with paired *t* tests at a significance level of $P \leq .05$. Holm's step-down correction adjusted for multiple comparisons. These tests allowed for multiple comparisons to be made in lieu of analysis of variance and post hoc analysis. All data are presented as mean ± standard deviation, unless otherwise noted.

Results

With respect to the scapular plane, the native humeral head center of rotation (COR) was 3.4 ± 2.0 mm posterior, 4.0 ± 2.2 mm superior, and 21.2 ± 5.9 mm lateral to the glenoid center as measured on the CT reconstruction. Reverse TSA centered the COR on the glenoid in the anterior/posterior direction (3-mm anterior shift, $P = .063$), and moved it significantly inferior and medial vs native shoulders (both $P < .001$, Table I). This shift agreed with previous reports from reverse TSA (Table I).[1,8,30] The native glenoid was 7.8° ± 4.3° retroverted with respect to the scapular plane, 27.5 ± 3.7 mm wide, and 35.6 ± 3.6 mm tall. These parameters were in good agreement with previous anatomic reports.[7,17]

Resting abduction angles increased for implant cases (9 combinations) compared with native (all $P \leq .049$, Fig. 2). For a given insert, 10° retroversion exhibited the highest resting abduction angle (up to 40°, Fig. 2, *A*). Increasing or decreasing retroversion by 10° decreased the resting abduction (all $P \leq .044$). For a given humeral version, increasing insert thickness increased resting abduction (all $P \leq .035$, Fig. 2, *B*). Peak resting abduction was highest for the +3/10° case.

When 16% BW was applied to the IS/TM, native arms externally rotated 20° from neutral at 60° scapular abduction (Fig. 3). Implanted arms exhibited net internal rotation

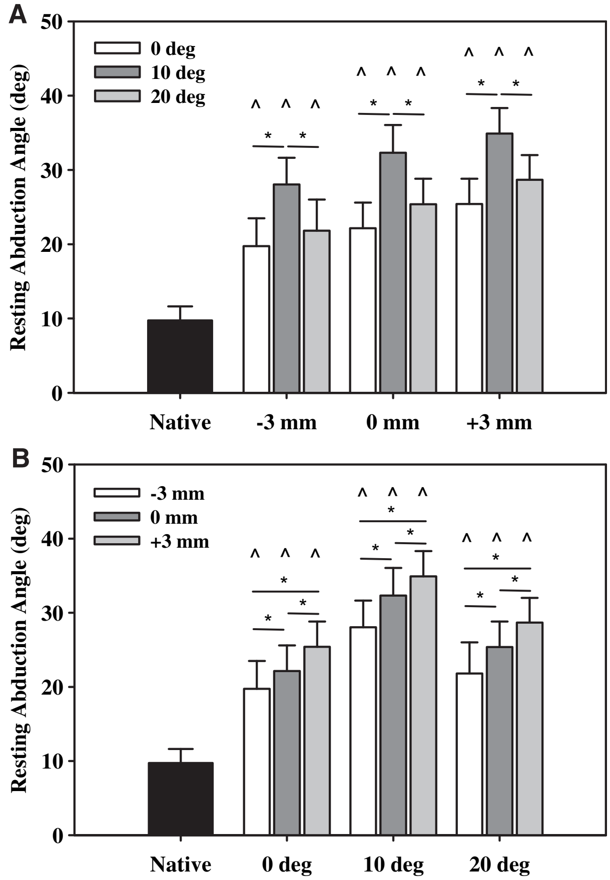

Figure 2 Resting abduction angle. Reverse arthroplasty resulted in significantly increased resting abduction angles versus native arms (^), where tare loads and scapular orientation were otherwise constant. Resting abduction increased a minimum of 10° vs native. (**A**) As a function of insert thickness, 10° of retroversion resulted in the highest resting abduction (*). (**B**) As a function of version, incremental increases in insert thickness resulted in incremental increases in resting abduction (*). Mean data are shown with the standard error of the mean.

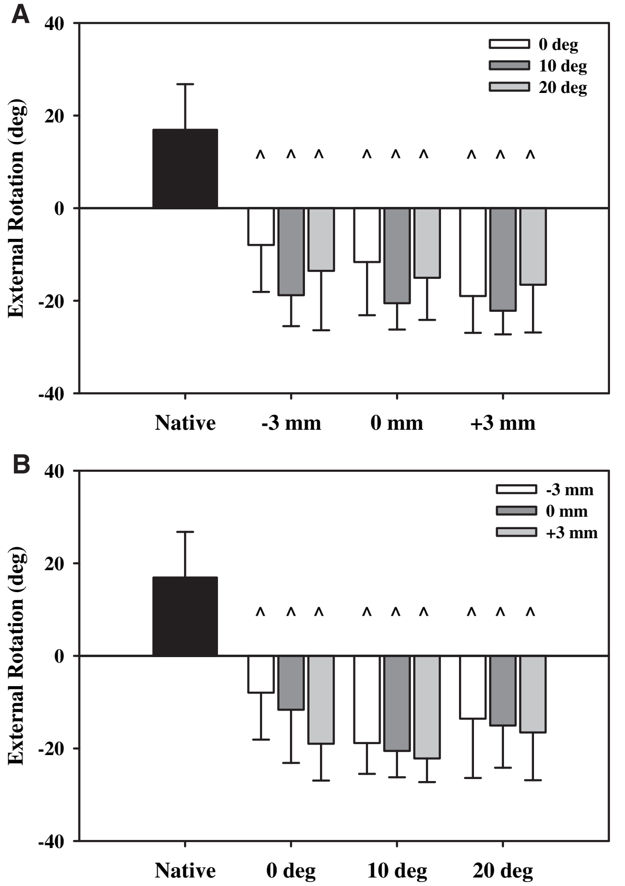

Figure 3 External rotation at 60° scapular abduction with 16% body weight applied to the infraspinatus/teres-minor. Native arms externally rotated up to 30°, whereas all reverse arthroplasty cases were significantly internally rotated (^). (**A**) By insert, version did not have a significant effect on arm rotation. (**B**) By version, insert did not have a significant effect on arm rotation. Mean data are shown with the standard error of the mean.

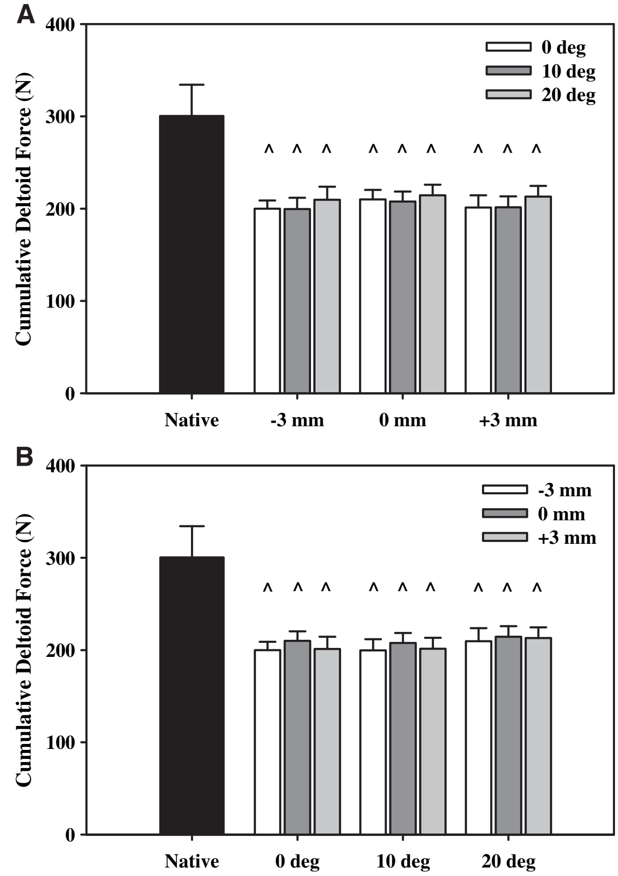

Figure 4 Cumulative deltoid force at 60° scapular abduction. Reverse arthroplasty cases required significantly less force to abduct than the native arm (^). **(A)** No significant differences were detected by insert across all versions tested. **(B)** No significant differences were detected by version across all inserts tested. Mean data are shown with the standard error of the mean.

under the same loading conditions (all $P \leq .035$). There were no differences in rotation among the 9 implant combinations (all $P \geq .184$). By insert, the net internal rotation showed a similar trend as resting abduction, where 10° humeral retroversion had the highest relative internal rotation (ie, least external rotation, Fig. 3, *A*). Increasing or decreasing retroversion by 10° increased external rotation up to 10°. By humeral version, the trend was again similar to resting abduction (Fig. 3, *B*). Increases in insert thickness tended to increase the relative internal rotation and limited external rotation.

Cumulative deltoid force was ~30% lower after reverse TSA than in native arms (all $P \leq .049$, Fig. 4). There were no differences in abduction force between any combination of humeral version or insert (all $P \geq .397$). In native arms, the anterior, middle, and posterior head of the deltoid accounted for 26.3% ± 7.3%, 56.8% ± 7.0%, and 15.3% ± 8.7% of the cumulative deltoid load, respectively. In implanted arms the anterior, middle, and posterior deltoid loads were 29.1% ± 6.0%, 48.4% ± 4.9%, and 22.8% ± 3.5% of the cumulative load, respectively. There were no differences between implant combinations (all $P \geq .109$) or between native and implant cases (all $P \geq .061$), with the exception of the middle deltoid for 0°/−3 mm, 20°/−3 mm, and 20°/+3 mm (all $P \leq .044$).

For all cases, laterally directed dislocation, with the arm at neutral external rotation, required less force (~100 N) than anteriorly directed dislocation (~200-300 N) with the arm at 90° external rotation (all $P \leq .021$, Fig. 5, *A* and *C* vs *B* and *D*). Increasing retroversion and implant thickness slightly increased the average dislocation forces in both anterior and lateral directions for all combinations of insert and version, although the change was not significant (all $P \geq .079$).

Discussion

This study was conducted to determine how humeral retroversion and deltoid tension (implant thickness) in

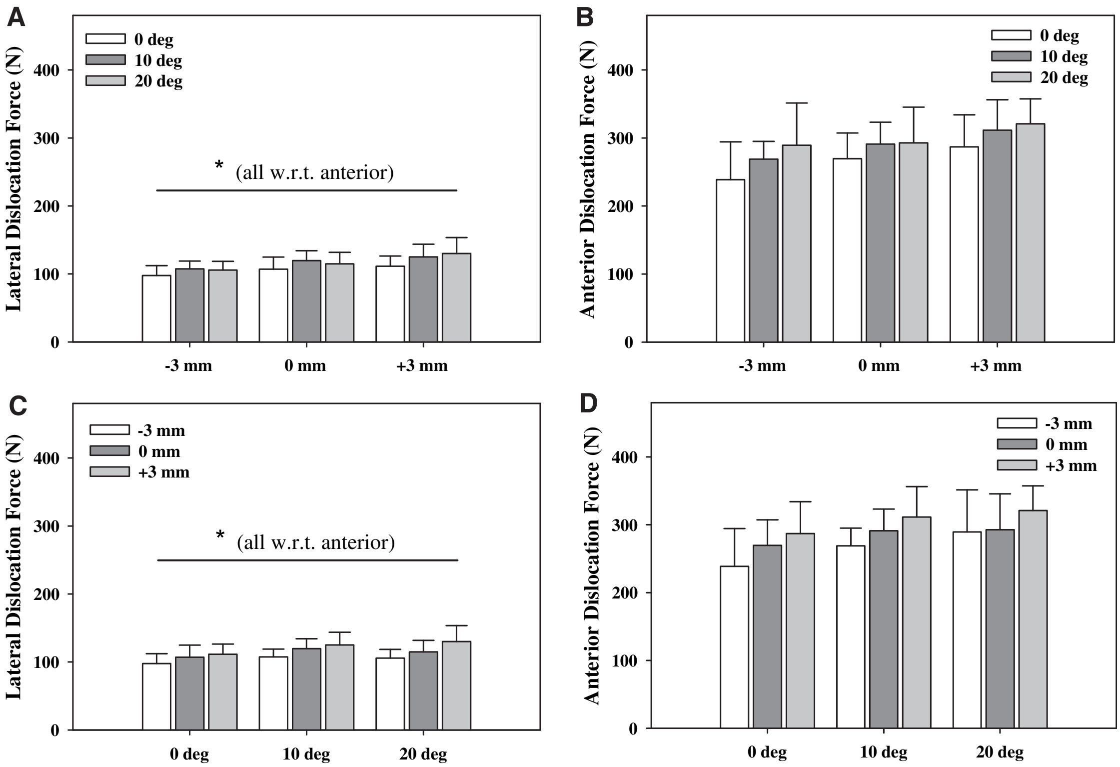

Figure 5 Force required to dislocate arms laterally in the neutral position and anteriorly at 90° of external rotation. **(A, C)** Lateral dislocation required significantly less force than **(B, D)** anterior dislocation for every combination of insert and version tested (A,C vs B,D *). No combination of version or insert was significantly different than another for either type of dislocation. Mean data are shown with the standard error of the mean.

reverse TSA affect range of motion (abduction, external rotation), deltoid abduction force, and anterior and lateral dislocation forces.

Our primary finding was that humeral version and deltoid tension have a large effect on adduction. We demonstrated that reverse TSA significantly increased adduction deficit, independent of insert thickness or humeral version. Each 3-mm increase in implant thickness increased deltoid tension and led to a significant stepwise increase in the adduction deficit (Fig. 2, *B*). Previously, increased adduction was achieved with inferior placement of larger-diameter glenospheres,[6,15,27] and a humeral neck-shaft angle of 130° (typically, 155°). Lateralized center of rotation and inferior glenoid tilt avoided adduction deficit in a computer simulation.[15]

Anatomic humeral retroversion is ~20°.[4,29] Recommendations for humeral component retroversion in reverse TSA range between 0° and 30°.[9,10,35] Limited clinical data have shown neutral version provides better outcomes for activities of daily living, strength, Constant score, radiologic loosening, and glenoid complications compared with 20° of retroversion.[26] In the present study, 10° of retroversion created the largest adduction deficit, which was reduced for 0° or 20°. A possible explanation is that 10° of retroversion created the most deltoid tension by positioning the insert on the most lateral aspect of the glenosphere. Adding or reducing retroversion therefore reduced soft tissue tension, increasing adduction.

The present model constrained the scapula, so changes in resting abduction were a direct result of the soft tissue tension within the glenohumeral joint. The large increase in resting abduction (>30°) suggests that the scapula is reoriented after surgery to compensate for altered tension, the affects of which have yet to be studied.

The second major finding was that reverse TSA resulted in less external rotation than in native arms. Increasing humeral retroversion in reverse TSA has been described to improve external rotation, with loss of internal rotation as a consequence.[13] Our data showed the contrary (Fig. 3): increasing humeral retroversion decreased external rotation. The difference in rotation may be from a reduced moment arm due to the medial/inferior shift of the center of rotation after reverse TSA or from impingement of the humeral component on posterior aspects of the scapula. Although our implant comparisons did not reach statistical significance, data trends suggest that the same variables that led to large adduction deficits (thick inserts, 10° retroversion) also led to decreased external rotation. Therefore, improving external rotation with a Grammont-style prosthesis may depend on the ‘‘tightness’’ of the shoulder and conserving a functional posterior rotator cuff rather than the choice of humeral version.

Our third finding was that deltoid abduction force was reduced after reverse TSA. It was originally determined that medializing the glenohumeral center of rotation increased the deltoid moment arm up to 20%, and an inferior move increased the efficacy of the deltoid up to 30%.[13,32] In the present study, abduction force was decreased by ~30%, regardless of humeral version or implant thickness, which has not been previously investigated (Fig. 4).

The Grammont-style reverse TSA is believed to increase deltoid efficacy by increasing the amount of the anterior and posterior deltoid recruited after medialization of the glenohumeral center of rotation.[1,5] Although not universally statistically significant, the percentage of load contributed by the anterior and posterior deltoid increased after implantation. Because the simulated deltoid lines of action were unchanged between test cases, this was likely a result of the medial/inferior shift of the humeral center of rotation (Table I).

The present study is the first to report data on lateral stability of reverse TSA. Neutral humeral version and ≤10° of glenosphere retroversion were shown to maximize anterior stability in a Delta III (DePuy Inc., Warsaw, IN, USA) reverse TSA,[9] but clinical interpretations are limited because the implants were not performed in a cadaveric shoulder. No other studies have evaluated reverse TSA stability in soft tissue−constrained shoulders.

Lateral dislocation in the present study required ~60% less force than anterior dislocation (Fig. 5). This suggests that reverse TSA dislocations may be initiated by smaller laterally directed forces than higher anteriorly directed forces. Version and deltoid tension did not significantly alter dislocation forces. As a consequence, the perceived benefit of increasing implant thickness to improve stability may be due to limiting the range of motion (increase adduction deficit, cause scapular reorientation), limiting inferior impingement, and lateral dislocation. Surgeons should therefore assess anterior *and* lateral stability during reverse TSA procedures.

As with other shoulder simulators, recognized limitations exist.[3,22,25,34] Active muscle contraction, proprioceptive control, and dynamically changing muscle lines of action could not be simulated by mechanical actuators and static loads. Similarly, a reduced set of muscles spanning the glenohumeral joint included only the rotator cuff and deltoid. Scapular articulation was not modeled, but experimental measures were taken at 60° of glenohumeral abduction to simulate ~90° abduction, assuming 2:1 scapulohumeral rhythm.[23,31] In addition, constant rotator cuff loads were estimated from physiologic models,[37] but this cannot recreate the dynamic in vivo balance of forces. Patients who receive reverse TSA often present with substandard rotator cuff tissue,[5] which further limits the prescribed loading that is possible in a clinical population.

Although these issues limit the ability to describe the in vivo conditions, this study was internally controlled to test how changes in humeral version and deltoid tension affected the outcome variables. The absolute magnitudes of the data should be interpreted with caution, but the relative affects of humeral version and implant thickness on range of motion, forces, and external rotation are considered relevant because boundary conditions were held constant between trials.

Conclusion

Implant thickness in reverse TSA increased resting abduction (adduction deficit) compared with a native shoulder. Reverse TSA significantly reduced the external rotation capability compared with the intact shoulder. Deltoid tension and humeral version had little effect on relative external rotation. Implantation reduced abduction forces by 30%, but humeral version and deltoid tension did not alter abduction efficacy. Finally, the force to cause lateral dislocation was significantly less than for anterior dislocation. Improved stability was associated with increased insert thickness, and deltoid tension may result from an increased adduction deficit and limitation of inferior impingement as opposed to larger forces required for dislocation.

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