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The influence of reverse arthroplasty humeral component design features on scapular spine strain

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Background: Reverse shoulder arthroplasty (RSA) humeral implant parameters have been previously studied with respect to range of motion, deltoid function, and stability. However, limited literature exists on the influence of humeral design features on scapular spine strain. The purpose of this cadaveric biomechanical simulator study was to evaluate the role of humeral component lateralization and neck-shaft angle (NSA) on scapular spine strain.
Methods: Eight fresh-frozen cadaveric shoulders were tested using an in vitro shoulder simulator. A custom-designed modular RSA system was implanted that allowed for the in situ adjustment of humeral lateralization and NSA. Scapular spine strain was measured by strain gauges placed along the acromion and scapular spine in clinically relevant positions representative of the Levy fracture zones. All testing was conducted in both abduction and forward elevation.
Results: In Levy zones 2 and 3, increasing humeral lateralization caused significant incremental decreases in scapular spine strain at 0° and 90° abduction ($P < .042$). Strain decreases as high as 34% were noted with increases in humeral lateralization from –5 to 15 mm ($P = .042$). Changing NSA had no statistically significant effect on scapular spine strain ($P > .14$).
Conclusions: Some humeral implant design features in RSA have effects on scapular spine strain. Humeral component lateralization had significant effects, whereas adjusting NSA resulted in no substantial differences in scapular spine strain. Understanding humeral component variables is important to allow for design optimization of future RSA implants.
Level of evidence: Basic Sciences Study; Biomechanics

Keywords: Reverse shoulder arthroplasty; RSA; humeral lateralization; neck-shaft angle; scapular spine strain; acromial fracture

Reverse shoulder arthroplasty (RSA) is a widely accepted surgical treatment option for rotator cuff tear arthropathy, acute proximal humerus fractures, and revision surgery after failed rotator cuff repair or arthroplasty.[4,9-12,14,26,28,29,31,33,35,39,43] As more surgeons are performing this procedure, postoperative complications are becoming more evident. Scapular spine and acromial fractures have a reported incidence as high as 10% following RSA, although this value may be underestimated as these fractures are challenging to identify and diagnose.[6,7,13,19,20,27,30,36,38] The pathophysiology of scapular spine fractures following RSA is likely multifactorial, and one theory is that they may occur through an insufficiency fracture mechanism.[15,19,30,38,44]

Institutional review board approval was not required for this basic science study.

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Grammont's original design for the reverse implant was semiconstrained, with a medialized and distalized center of rotation. The design also included a nonanatomic neck-shaft angle (NSA) of 155° aimed to lengthen the arm and increase deltoid tension, allowing for theorized optimal deltoid function.[1,3,40] The design has subsequently evolved over time, with commercially available RSA systems today offering a variety of implant configurations including NSA ranging from 127.5°-155° and variably lateralized humeral stems and glenoid baseplates.[42] These changes in humeral stem type have reported implications with respect to range of motion, joint contact stress, deltoid abduction force, and postoperative deltoid tension and arm length.[3,8,17,18,22-24,37,41]

There remains a lack of knowledge regarding the consequence of adjusting humeral stem parameters and the resultant effect on scapular spine strain following RSA. A finite element analysis found that with lateralization of the humerus, small decreases in acromial stress were noted.[44] However, NSA has not been investigated in direct correlation to scapular spine strain in a cadaveric model. As such, the purpose of this in vitro biomechanical cadaveric study was to investigate the role of humeral lateralization and NSA on scapular spine strain following RSA implantation. A custom modular RSA system allowed for testing of 3 levels of humeral lateralization (–5, 5, and 15 mm) and 3 NSAs (135°, 145°, and 155°). This was evaluated through 4 planes of elevation (0° and 90° abduction, 0° and 90° forward elevation) using a shoulder simulator. We hypothesized that increasing humeral lateralization and decreasing NSA would both result in a decrease in scapular spine strain.

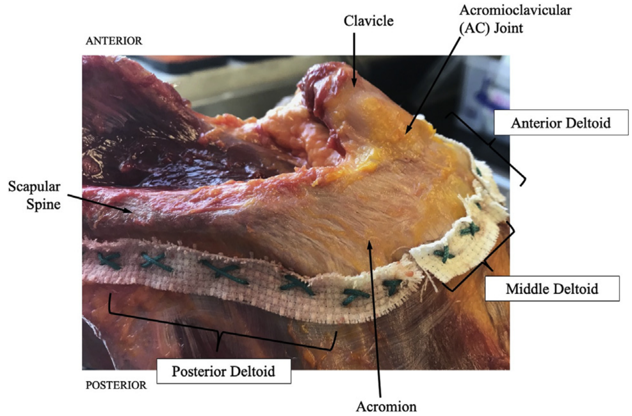

Figure 1 Deltoid preparation. Nylon mesh and no. 5 Ethibond suture was used to secure the musculotendinous origin of the anterior, middle, and posterior deltoid.

Materials and methods

Specimens

Eight fresh-frozen right male cadaveric shoulders (mean age 73 years, range 61-88 years) were thawed for at least 24 hours before testing. The specimens were prepared by creating full-thickness supraspinatus and upper infraspinatus tears in order to simulate a rotator cuff–deficient shoulder. The 3 heads of deltoid were then identified based on the anatomic description by Sakoma et al[32] and tagged along the tendinous origin with nylon mesh and no. 5 Ethibond suture (Ethicon, Johnson & Johnson, New Brunswick, NJ, USA), as seen in Fig. 1. Of note, the coracoacromial ligament was left intact and the clavicle was affixed to the simulator in its anatomic position.

A custom modular RSA system was then implanted using a modified technique from the Wright Medical-Tornier Aequalis surgical technique manual (Wright Medical Technologies, Memphis, TN, USA).[16,25] The glenoid was first prepared and the baseplate was secured with 3 screws in an inverted triangle orientation and placed in neutral inclination. After humeral head resection, the canal was reamed and the humeral component was cemented in anatomic version relative to the transepicondylar axis. In order to affix the specimen to the shoulder simulator, an intramedullary rod was cemented into the distal humeral shaft.

For this study, uniaxial strain gauges were employed to measure strain across the acromion and scapular spine. This required meticulous preparation of the bony surface, removing all soft tissue and periosteum while leaving the deltoid insertion intact. Cyanoacrylate adhesive was then used to place 4 strain gauges in clinically relevant locations along the acromion and scapular spine (Strain gauge model KFH-06-120-C1-11L3M3R; OMEGA Engineering, Quebec, Canada). Strain gauges were placed according to their location within Levy scapular spine fracture zones (Fig. 2).[27] The gauge leads were connected through a data acquisition unit (model NI USB-9237; National Instruments, Austin, TX, USA) to a central computer for data collection.

Shoulder simulator and testing protocol

The specimens were mounted onto a shoulder simulator using a scapular clamp via bolts drilled into the lateral scapular body (Fig. 3). A load cell was mounted to a distal humeral abduction arc, which allowed for the accurate prescribing of load onto the humeral shaft. The shoulder simulator allowed freedom of motion of the humerus in all 3 directions of translation, as well as axial rotation, while constraining plane of elevation angles. Computer-controlled pneumatic actuators were attached by cables to the 3 heads of deltoid as well as the anterior and posterior cuff muscles along their physiologic lines of action. These permitted loading to be applied to each muscle-tendon unit. The anterior and posterior cuff had a cumulative load of 10 N applied to maintain tension, whereas the 3 heads of deltoid were loaded based on muscle loading ratios from a previous biomechanical study.[2]

Using the custom modular RSA implant, each specimen underwent testing of the variable implant configurations in a randomized order. For evaluation of humeral lateralization, implant configurations were tested based on altering humeral component position from –5, 5, to 15 mm lateralization (Fig. 4). All other parameters were held constant, including NSA of 145°, glenoid lateral offset of 5 mm, and glenosphere and humeral cup size of 42 mm. For assessing NSA, 3 implant configurations were

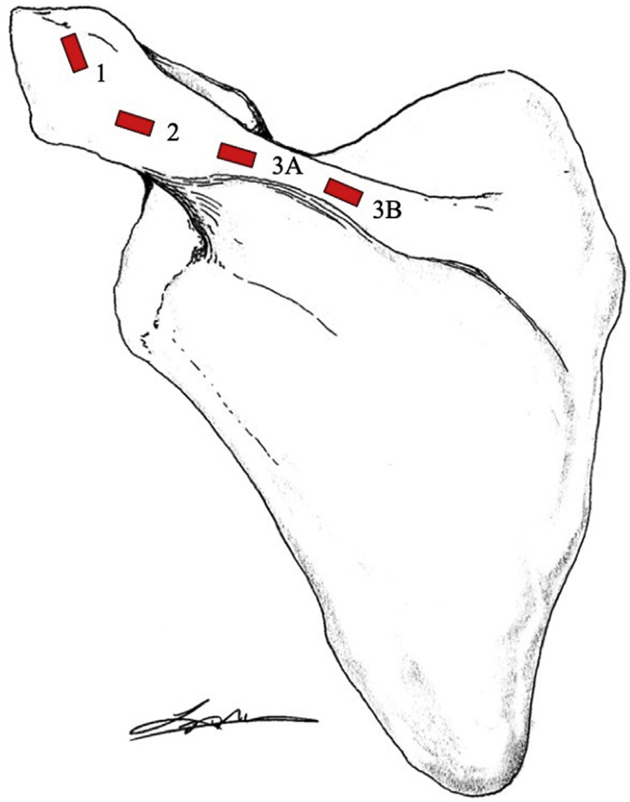

Figure 2 Illustration of a left scapula depicting the placement of strain gauges within the Levy zones. The strain gauge in Levy zone 1 was placed midway between the anterior and posterior edge of the acromioclavicular (AC) joint, and midway between the medial and lateral edge of the acromial tip. The strain gauge in Levy zone 2 was placed halfway between the leading edge of strain gauge 3A and the lateral edge of the acromion. The strain gauge in Levy zone 3A was placed directly above the spinoglenoid notch. The strain gauge in Levy zone 3B was placed 2 cm medial to strain gauge 3A, measured from the leading edge (lateral edge) of both gauges.

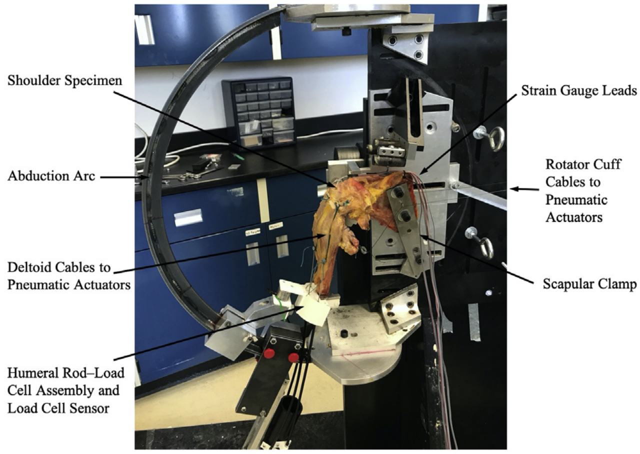

Figure 3 Shoulder simulator. A right cadaveric specimen mounted onto the shoulder simulator using a scapular clamp. The load cell is seen mounted to a distal humeral abduction arc. Computer-controlled pneumatic actuators (not pictured) are attached to cables running from tagged tendons (deltoid, rotator cuff).

evaluated, including NSA of 135°, 145°, and 155° (Fig. 5). The remainder of the implant parameters remained unchanged, with 5 mm glenoid lateral offset, 5 mm humeral lateralization, and 42 mm glenosphere and humeral cup size.

Range of motion was conducted in 2 planes of elevation at 2 angles: 0° and 90° abduction in the scapular plane (scaption), and 0° and 90° forward elevation. Using the distal humeral load cell for feedback, the pneumatic actuator loads were adjusted until target deltoid abduction moment of 1.5 Nm was reached. Strain gauge data was then captured. This cycle was repeated 3 times for each plane of elevation, and averaged strain values were recorded.

Outcome variables and statistical analysis

This study employed strain as the main outcome variable, as measured by the 4 strain gauges placed within the Levy fracture zones. Based on the testing protocol, each implant configuration was tested in 4 planes of elevation, and scapular strain was recorded.

Strain is defined as a change in length of a material over the original length and is reported as a ratio or a percentage. For the purpose of this study, strain values were analyzed in microstrain units (μStrain, strain $\times 10^6$).

Each strain gauge was independently analyzed using repeated measures analysis of variance. For both the analysis of humeral lateralization and NSA, a 3-way (humeral lateralization, plane of elevation, angle of elevation; NSA, plane of elevation, angle of elevation) repeated measures analysis of variance was used (SPSS, version 25.0; IBM, Armonk, NY, USA). Pairwise comparisons and analyses of interactions were performed to assess for significance, using Fisher least significant difference test. Statistical significance was defined as P <.05. A sample size of 8 specimens was chosen based on previous similar biomechanical cadaver studies.[2,5,34] All statistically significant differences detected in the outcome variables were found to have a power greater than 0.8.

Results

The effects of humeral lateralization on scapular spine strain for each strain gauge at each plane of elevation are shown in Fig. 6. Among all strain gauges, the highest strain values were measured in Levy zone 2 (P < .045). Furthermore, significantly higher strain values were measured in forward elevation than abduction (P < .001), and at the initiation of motion (0°) than at 90° terminal motion (P = .002).

In Levy zone 1, lateralization of the humeral component from –5 to 15 mm was shown to cause a 10% statistically significant decrease in scapular spine strain at 90° forward

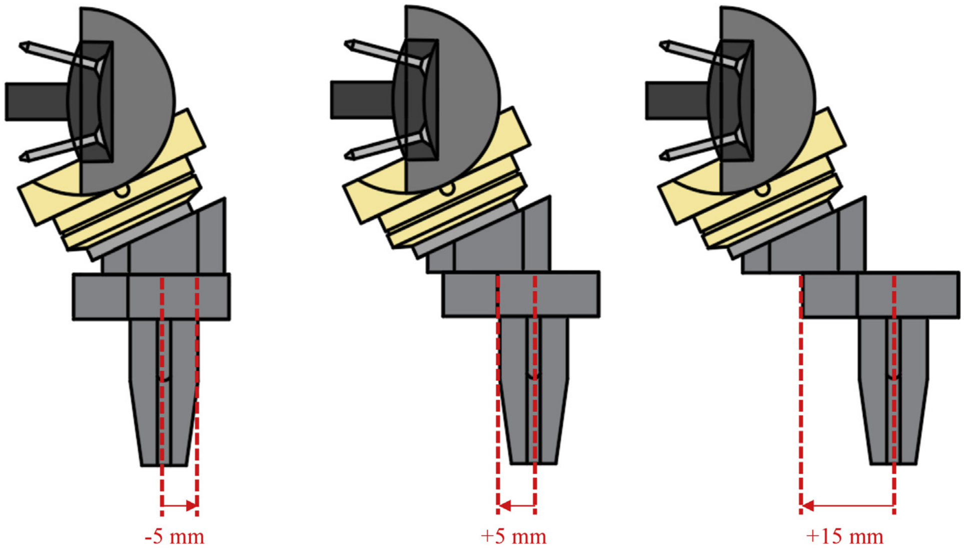

Figure 4 Humeral lateralization. Illustration depicting increased humeral lateralization using a custom modular implant.

elevation (P = .005). No other statistically significant changes were seen with the arm moving through abduction.

For Levy zone 2, with the arm in 0° abduction, scapular spine strain decreased by 18% as the humerus was lateralized from 5-15 mm (P = .036), with an overall 34% decrease from –5 to 15 mm (P = .042). Similarly, at 90° abduction, scapular spine strain decreased by 23% as the humerus was lateralized from 5-15 mm (P = .029), with an overall 28% decrease from –5 to 15 mm (P = .012). In forward elevation, as the humerus was lateralized from –5 to 15 mm, no significant differences in scapular spine strain were observed ($P > .06$).

Similar trends were observed in Levy zone 3A, with a significant decrease in scapular spine strain with increasing humeral lateralization at both 0° and 90° abduction. With the arm in 0° abduction, scapular spine strain decreased by 18% and 16% as the humerus was lateralized from –5 to 5 mm (P = .04) and 5-15 mm (P = .036), respectively, with an overall 31% decrease from –5 to 15 mm (P = .034). Similarly, at 90° abduction, scapular spine strain decreased by 21% as the humerus was lateralized from 5-15 mm (P = .023), with an overall 26% decrease from –5 to 15 mm (P = .007). At 0° forward elevation, no significant differences in scapular spine strains were realized with the various humeral lateralizations tested. At 90° forward elevation, however, there was a significant reduction in scapular spine strain by 14% as humeral lateralization increased from –5 to 15 mm (P = .007).

Strains measured in Levy zone 3B demonstrated incremental decrease with increasing humeral lateralization

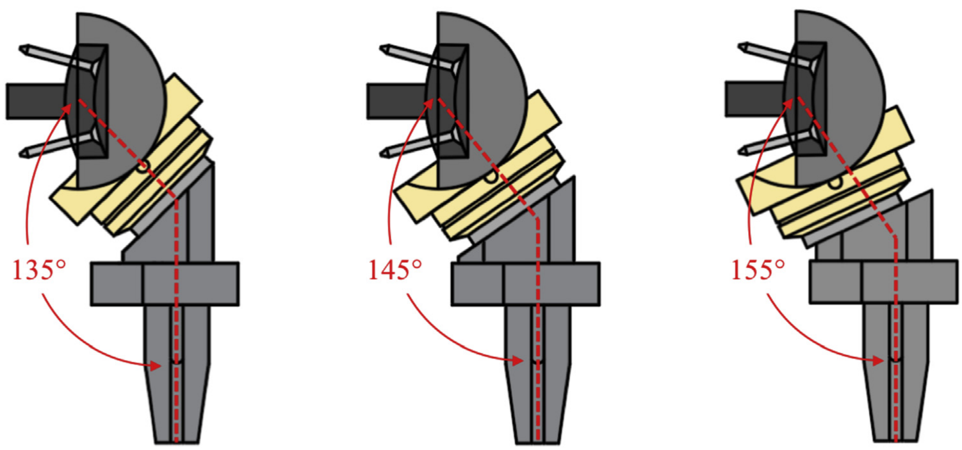

Figure 5 Neck-shaft angle. Illustration depicting increased neck-shaft angle using a custom modular implant.

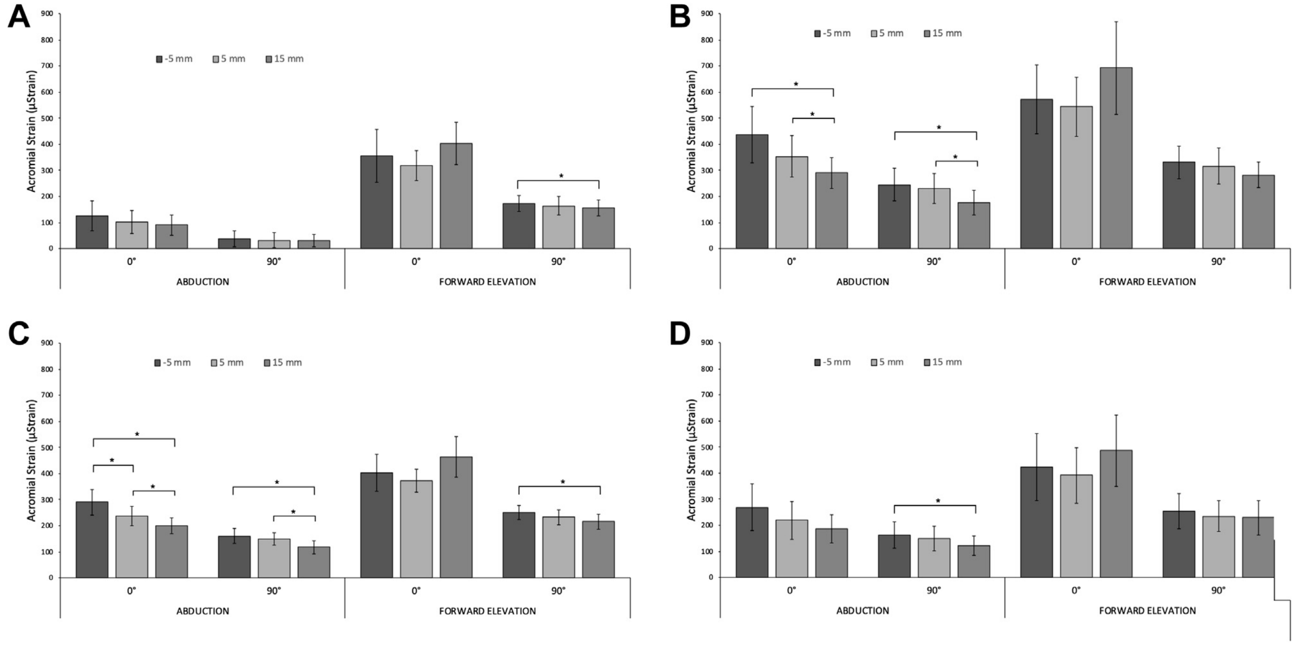

Figure 6 The effect of humeral lateralization on scapular spine strain. Mean (±1 standard deviation) acromial and scapular spine strain measured in (A) Levy zone 1, (B) Levy zone 2, (C) Levy zone 3A, and (D) Levy zone 3B for increasing humeral lateralization (0, 5, and 10 mm) during all 4 planes of elevation. *Significance ($P < .05$).

from –5 to 15 mm, reaching significance with a 24% decrease at 90° abduction ($P = .032$) and trending at 0° abduction with a 30% decrease ($P = .07$). When testing in forward elevation, scapular spine strains were not significantly different with increasing humeral component lateralization.

Figure 7 shows the results of NSA on scapular spine strain for each strain gauge at each plane of elevation. With

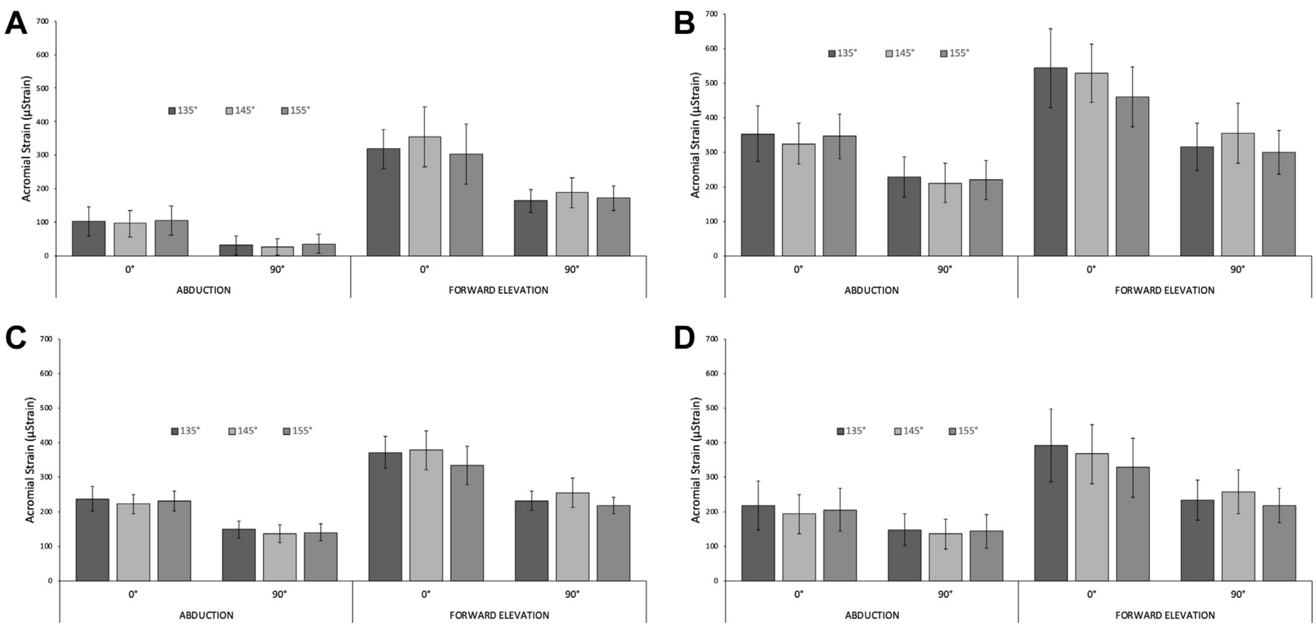

Figure 7 The effect of neck-shaft angle on scapular spine strain. Mean (±1 standard deviation) acromial and scapular spine strain measured in (A) Levy zone 1, (B) Levy zone 2, (C) Levy zone 3A, and (D) Levy zone 3B for increasing neck-shaft angle (0, 5, and 10 mm) during all 4 planes of elevation. *Significance ($P < .05$).

a change in NSA, no statistically significant effect was measured on scapular spine strain ($P > .14$).

Discussion

This biomechanical cadaveric study demonstrated that increased humeral lateralization caused an apparent decrease in scapular spine strain, whereas changes in NSA measured no significant effect. To date, there exists limited biomechanical literature on the effects of RSA implant parameters on scapular spine strain. This study was unique in its use of a custom modular RSA system, which allowed multiple permutations of implant configurations to be tested in each specimen. Furthermore, the shoulder simulator allowed for 4 planes of elevation to be tested in each protocol cycle, yielding important data for the clinical application of these results.

The results of humeral lateralization were associated with several overarching patterns. In Levy zone 1, which is the lateral extent of the scapular spine and the acromion, variations in humeral component lateralization had no substantial effect on scapular spine bone strain. In the remaining Levy zones, an overall trend toward decreased scapular spine strain with increased humeral lateralization was seen in abduction. Our results indicate that with active abduction, the scapular spine is more sensitive to changes in humeral lateralization. This finding seems logical, as increasing the humeral component lateralization results in a more favorable moment arm for the deltoid in abduction, and also changes the line of action of the deltoid as it wraps around the lateralized humerus. These results agree with previous work where Giles et al[16] demonstrated in a cadaveric model that increasing humeral lateralization decreased the deltoid force required for active abduction because of increased muscle moment arm. Additionally, Wong et al[44] demonstrated in a computational study that lateralizing the humeral shaft by 5-mm increments resulted in small decreases in acromial stress. Overall, decreases in scapular spine strain in abduction may be due to changes in the deltoid moment arm as well as changes in the line of action of the deltoid as it wraps around the lateralized humerus.

Overall, bone strains in the scapular spine were substantially higher with the simulated motion of forward elevation compared with abduction. We suspect this occurs as the anterior deltoid is more active during forward elevation, and the attachment of the anterior deltoid on the acromion is furthest away from the bony connection to the scapular body. As such, the anterior projection of the acromion is analogous to a diving board, where the stresses concentrate at the attachment of the diving board to its stable base, which represents Levy zone II. Interestingly, although forward elevation produces substantially greater scapular spine bone strains, it is more resistant to changes in humeral component lateralization. This also makes sense, as humeral component lateralization increases the lever arm in the coronal plane (abduction) but does not substantially change the lever arm in the sagittal plane (flexion). Generally, our hypothesis of increased humeral component lateralization resulting in decreased scapular spine bone stresses was accepted for more medial portions of the scapular spine in abduction, as the acromion was unaffected.

Changing humeral component NSA was not found to have a significant effect on scapular spine strain in any of the 4 planes of elevation. The effect of altering NSA on shoulder range of motion and scapular impingement has been extensively investigated.[8,17,18,23,37,41] A trade-off in range of motion has been observed, as a more varus neck allows for improved adduction by decreasing impingement, but limits abduction. This occurs as a lower humeral inclination causes a small increase in humeral offset, which may lead to impingement of the greater tuberosity on the acromion.[23,41] Lädermann et al[23] demonstrated that a decrease in humeral inclination from 155° to 135° resulted in only a 2-mm increase in humeral offset. The effect of this change in offset on scapular spine strain is likely negligible. The custom modular implant used in this study was designed so that a changing NSA did not alter offset[16,25] and supports the finding that NSA has no significant effect on scapular spine strain.

Although not statistically significant, the strain in Levy zone 2 and 3B exhibited a general decreasing trend with increased NSA with the arm in 0° forward elevation. A similar pattern was found in a finite element analysis by Langohr et al who concluded that increasing the NSA lead to a decrease in maximum articular contact stress of the implant.[24] Although implant contact stress and scapular spine strain are 2 separate entities, they exist within the same mechanical construct and thus may be interrelated.

Inherent limitations exist in using cadaveric models for biomechanical studies. Computer-controlled simulators cannot truly replicate in vivo mechanics and physiology, and cadaveric specimens themselves may not accurately demonstrate the pathology occurring in patients who undergo RSA. Although the shoulder simulator was able to replicate the static range of motion in 4 planes of elevation, this does not truly reproduce the multiaxial nature of the shoulder joint. Although these limitations may affect the prediction of the absolute magnitudes of scapular spine strain, the repeated measures experimental design using controlled loading in a consistent fashion permits a valid comparison among the variables of humeral component position as assessed herein. Additionally, testing of scapular spine stresses in the native shoulder was not done because of the complexity of physiologically modeling the rotator cuff at various glenohumeral rotations and planes of elevation. Additional parameters that were not measured but would be of future interest include deltoid wrapping and changes in deltoid length during implant testing. In regard to scapular position, in an effort to replicate a 2:1 scapulohumeral rhythm during humeral elevation, 60° of glenohumeral elevation was used to simulate 90° of humerothoracic elevation.[21]

Conclusion

It is well documented that patients with postoperative scapular spine and acromial fractures have inferior outcomes.[24] Despite this, there exists a lack of knowledge surrounding the etiology and factors that contribute toward increased scapular spine strain following RSA. This study provides further insight into the effect of humeral component design features on scapular spine strain following RSA. Humeral component lateralization results in decreased scapular spine strain, specifically during loaded abduction of the arm. In forward elevation, although scapular spine strains are significantly higher than in abduction, the spine is more resistant to changes in humeral component lateralization. With all other parameters held constant, a change in NSA did not have a significant effect on scapular spine strain during abduction or forward elevation. The results of this study have implications in the future design and manufacturing of RSA implants, and help further our understanding of the etiology of scapular spine fractures following RSA.

Disclaimer

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