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Effect of the humeral neck-shaft angle and glenosphere lateralization on stability of reverse shoulder arthroplasty: a cadaveric study

Manuel Ferle, MS, PhD, Marc-Frederic Pastor, MD, Jakob Hagenah, Christof Hurschler, PhD, Tomas Smith, MD, PhD*

Department of Orthopaedic Surgery, Medical School Hannover, Hannover, Germany

Background: Lateralizing the glenosphere and decreasing the humeral neck-shaft angles are implant design parameters that reduce the risk of scapular impingement. The effects of these parameters on joint stability remain unclear. This study evaluated the effect of glenosphere lateralization and humeral neck-shaft angle on joint stability by quantifying the anterior dislocation force in different arm positions.
Methods: Reverse shoulder arthroplasty was performed on 19 human shoulder specimens. Anterior dislocation force and maximum external rotation were evaluated using a robot-based shoulder simulator. By varying the neck-shaft angle and magnitudes of glenosphere lateralization, 12 configurations were analyzed with the glenohumeral joint in 30° and 60° of abduction, in neutral, and in 30° of external rotation.
Results: At 30° of abduction, measurements showed significantly higher dislocation forces for the 9-mm and 6-mm lateralized glenosphere than for the 0-mm ($P < .0001$, $P = .007$) nonlateralized glenosphere. At 60° of abduction, measurements showed significantly higher dislocation forces for the 9-mm and 6-mm lateralized glenosphere than for the 0-mm ($P < .0001$, $P = .0007$) and 3-mm ($P = .0003$, $P = .04$) glenosphere. Configurations with a neck-shaft angle of 135° showed significantly higher dislocation forces than configurations with a neck-shaft angle of 145° ($P = .02$) or 155° ($P = .02$) at 30° of abduction in 30° of external rotation. Neck-shaft angle and glenosphere lateralization had no influence on maximum external rotation capability.
Conclusion: Glenosphere lateralization significantly increased anterior stability of the glenohumeral joint without influencing the range of passive external rotation. The humeral neck-shaft angle only had a minor effect on anterior stability.
Level of evidence: Basic Science Study; Biomechanics

Keywords: Reverse shoulder; arthroplasty; shoulder; biomechanics; neck-shaft angle; lateralization; stability

Reverse total shoulder arthroplasty (RSA) has become the favored surgical option to reduce pain, improve function, and achieve stability of the joint in cases of rotator cuff arthropathy, severe proximal humeral fractures, and failed anatomic total shoulder arthroplasty.[8,13,36] The success of this treatment is reflected in the steady increase of RSA procedures in the USA, Europe, and Australia.[2,15,24,26] Despite this seemingly consistent trend, several studies have shown that with the current surgical techniques and implant designs, the procedure is still associated with various problems and complications, such as instability, impingement, infection,

*Reprint requests: Tomas Smith, MD, PhD, Hannover Medical School, Department of Orthopaedic Surgery, Anna-von-Borries-Straße 1-7, 30625 Hannover, Germany.

E-mail address: tomas.smith@diakovere.de (T. Smith).

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component loosening, and periprosthetic fractures.[9,19,26] Thus, total complication rates ranging from 7% to 68% have been reported.[20,32,40]

The most common complication inherent to the implant design is inferior scapula impingement,[5,33,34] with an incidence of between 12% and 96%.[1,6,29,44,45] Among the more serious complications leading to implant failure, instability is the most common cause for revision, with a prevalence of up to 48%,[2,3,10] with most stabilities occurring in the anterior direction.[10,41] Thus, the most prevalent complications in RSA (impingement and instability) are still related to mechanical or design-specific issues, despite the introduction of the Grammont-style shoulder arthroplasty decades ago.

The challenges of the reverse joint design for shoulder arthroplasty thus remain. One explanation is that the interaction of implant design parameters and joint function is particularly complex in RSA. Inappropriate choice of implant size, configuration, and placement are associated with pain, reduced range of motion, risk of scapular impingement, or instability.[4,9,25,33,35] Implant-specific design parameters, such as humeral cup depth, lateralization or neck-shaft angle, glenosphere diameter, eccentricity, or lateralization in some cases, have contradictory effects on stability, impingement, and range of motion.

Numerous studies have evaluated the effect of individual implant parameters on scapular impingement and range of motion.[21,25,27,35,36,38] Only a few studies, however, have focused on investigating the effect of implant design parameters on joint stability.[7,12,17,31] Furthermore, many of the biomechanical studies did not take into account the physiological soft tissue envelope of the shoulder, and thus, the role of passive soft tissue tension was neglected.

The goal in a previous study we performed was to investigate the influence of the glenosphere diameter, polymer insert configuration, and rotator cuff condition on anterior stability, while taking into account the soft tissue envelope of the shoulder.[31] The study showed better stability for a greater glenosphere diameter and a more conforming humeral onlay, that joints were more stable at external rotation compared with neutral and internal rotation, and anterior stability decreased when the subscapularis tendon was absent.[31]

In the current second study, our goal was to address stability issues of the recently discussed changes[11] to the Grammont prosthesis design (humeral neck-shaft angle, glenosphere lateralization), with the intention of reducing the risk of impingement. The purpose of this biomechanical study was therefore to investigate the effect of the humeral neck-shaft angle as well as glenosphere lateralization on anterior stability and maximum external rotation during RSA. The effects of these implant design parameters on anterior stability were evaluated by measuring the anterior dislocation force in 30° and 60° of glenohumeral abduction with the arm in neutral and 30° of external rotation. We hypothesized that the neck-shaft angle as well as glenosphere lateralization would have an effect on joint stability and that the passive external rotation range of motion would be affected by the neck-shaft angle and glenosphere lateralization.

Materials and methods

Sample size estimation

Before the experiments were conducted, a power analysis was performed to determine the minimum number of shoulder specimens required to detect differences in stability between implant configurations and joint positions. The analysis was performed based on results of a pilot and a previous study[31] and showed that a minimum of 12 specimens would be required, assuming a power of 0.80 and a α of 0.05.

Specimen preparation and mounting

The investigation used 19 fresh frozen cadaveric shoulders and elbow joints (13 male, 6 female) without evidence of shoulder injury or prior surgery (Science Care Inc., Phoenix, AZ, USA). The donors were a mean age of 69.7 ± 10.9 years (range, 37-91 years), and the mean body mass index was 23.4 ± 2.9 kg/m^2 (range, 19.8-30.9 kg/m^2). The specimens were stored at −20°C and thawed at room temperature for 24 hours before testing.

The soft tissues on the lower part of the scapula were carefully removed on the anterior and posterior side, leaving the superior and lateral portion of the rotator cuff muscles and the musculotendinous junction intact. The scapula was then embedded with use of a custom-made potting box and cold curing casting resin (Rencast FC 52/53 Isocyanate, Polyol FC 53, Filler DT 982; Huntsman Corp., The Woodlands, TX, USA). Before the humerus was prepared, neutral rotation was defined by holding the elbow joint in 90° of flexion and drilling a Kirschner wire aligned parallel to the axis of the forearm into the humeral shaft. Soft tissue and muscles were then partially resected. The humerus was transected approximately 20 cm distal to the center of the humeral head and potted in a brass cylinder (cold curing casting resin).

After preparation, the potted scapula was secured to a mounting plate using 3 threaded rods with 10° of forward tilt to approximate its physiologic orientation on the thorax. The scapula was then firmly mounted to the mounting tower of the shoulder simulator and further aligned by orienting the medial scapular margin vertically in the coronal plane. The humeral shaft was then attached to the flange of the robot by inserting the brass cylinder into a mounting fixture on the robot.

Shoulder simulator

Testing was performed with a robot-assisted shoulder simulator as used in several previous studies.[25,28,31,35,42] The testing rig consisted of a mounting tower and an industrial robot system (KR16-2; KUKA AG, Augsburg Germany) outfitted with a 6-component force-moment sensor (Delta; ATI Industrial Automation, Apex, NC, USA). While the scapula was rigidly attached to the mounting tower, the flange of the robot arm was attached to the humerus, allowing force- and moment-controlled motions to the glenohumeral joint (Fig. 1).

The shoulder simulator allowed motion control with a repeatability of 0.04 mm and measurement of forces and moments with a resolution of less than 0.25 N and 7.5 Nmm, respectively. The

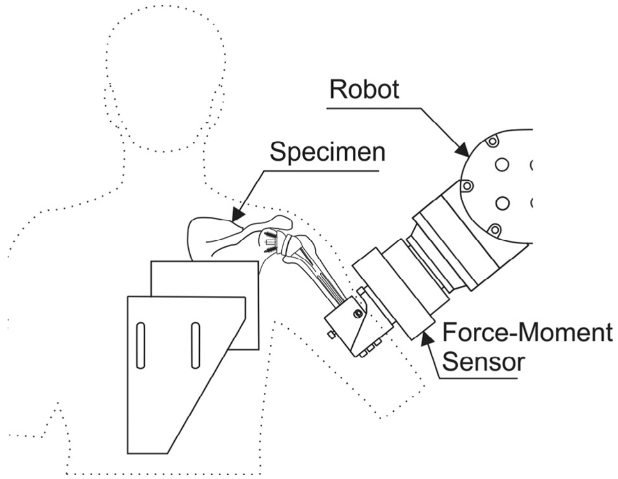

Figure 1 Schematic representation of the experimental setup shows the mounted specimen with the humerus attached to the flange of the robot, which is equipped with a force-moment sensor.

following global coordinate system was defined for the robot control: the *x*-axis was directed medially parallel to the previously defined scapular plane. The *y*-axis was defined as being perpendicular to the scapular plane and directed posteriorly. Finally, an axis orthogonal to the *x*- and *y*-axis was defined as the *z*-axis and directed superiorly.

Second, a specimen-specific humeral coordinate system at the geometric center of the humeral head was defined to describe the motion of the humerus with respect to the scapula as follows: before testing, passive elevation, flexion-extension, and rotation of the joint centered in the glenoid were performed using the force-moment-controlled robot. The geometric center was then defined as the point of the humerus that moved the least. The humeral coordinate system was oriented codirectional to the global coordinate system after the glenohumeral joint was centered with the arm hanging under its own weight in 0° abduction and neutral rotation. All moments were resolved relative to the humeral coordinate system.

Surgical procedure

The reverse shoulder prosthesis (Delta Xtend; DePuy Synthes, Raynham, MA, USA) was implanted through a deltopectoral approach, and surgery was performed as described in previous studies.[25,31] Before the humerus was prepared, the supraspinatus tendon was resected and tenotomy of the long head of the biceps was performed. Resection of the humeral head was undertaken with an inclination angle of 155° and 10° retroversion. Therefore, the cutting guide was aligned to a Kirschner wire in 10° retroversion position of the humeral implant driver, which was placed in line with the forearm axis. The humeral shaft was then reamed to properly fit an appropriately sized press-fit stem into the humerus.

The glenoid component was prepared by circumferentially releasing the labrum and dissecting the remaining joint capsule to achieve a clear view of the entire glenoid. The hole for the central peg of the metaglene was drilled near the center of the inferior glenoid circle so that the border of the metaglene followed the inferior edge of the glenoid to minimize risk of impingement. The metaglene was then fixed to the glenoid with four 4.5-mm screws. Centric glenospheres with a diameter of 38 mm and varying degrees of lateralization were implanted. Regarding the final soft tissue status after surgery, the deltoid muscle, infraspinatus, teres minor and conjoint tendon were intact, the supraspinatus, long head of the biceps, and the subscapularis tendon were resected , and the capsule was incised around the glenoid rim.

Testing protocol

We tested 12 implant configurations by varying 2 parameters: glenosphere lateralization was varied from 0 mm to 9 mm in 3-mm increments at 3 different neck-shaft angles (135°, 145°, and 155°). Glenosphere lateralization was achieved with different magnitudes of spherical segment thickness, with 0-mm lateralization based on the glenosphere thickness of the DePuy Delta Xtend device (Fig. 2). For the adaption of the neck-shaft angle, the manufacturer provided metaphyseal components of 155° and 145° shaft angles, together with humeral onlays in 2 designs. The standard onlay for an overall neck-shaft angle was 155°, or 145° using an additional onlay with a –10° tilt, for an overall neck shaft-angle of 135° when combined with the 145° epiphyseal component (Fig. 2).

It should be noted that the change in the neck-shaft angle also involves a construction-related change in the position of the humerus relative to the center of rotation of the prosthesis. When the neck-shaft angle is changed from 155° to 145°, in neutral position the humerus position shifts relative to the center of rotation by about 2 mm in superior direction (*z* direction; Fig. 2) and about 3 mm in lateral position (–*x* direction; Fig. 2). A change from 155° to 135° results in a change of the humerus position relative to the scapula of about 3 mm in superior direction (*z* direction) and about 2 mm in lateral position (–*x* direction).

The testing order of the 12 implant configurations was randomized and performed at different arm positions that were approached by rotating the arm in force-moment control, directing the forces to 0 and allowing maximum moments of 2 Nm to prevent soft-tissue injuries. Translation stability was evaluated by loading the glenohumeral joint in anterior direction at 30° and 60° of abduction with the arm in neutral and 30° of external rotation. During stability testing, the joint was allowed to translate freely in anterior-posterior, superior-inferior, and medial-lateral directions while all rotations were held constant. For stability evaluation, an anterior force of 10 N was applied and gradually increased while simultaneously centering the humerus in the glenoid with a 30 N medially oriented force. Anterior force was increased until the reverse shoulder joint dislocated. Dislocation was defined as anterior humeral translation of 19 mm (radius of the glenosphere).

Maximum external rotation was evaluated with the arm in 30° and 60° of abduction, defined as the maximum angle reached by applying a 2-Nm external-rotation moment and a medially oriented centering force of 30 N. During application of the moment, the joint was allowed to translate freely in anterior-posterior, superior-inferior, and medial-lateral directions, while the abduction and flexion angle were held constant. Each of these tests was performed once for each configuration in a randomized order.

Statistical analysis

Statistical analysis of the data was performed using R 3.3.2 software (R Foundation for Statistical Computing; https://www.r-project.org/foundation/). Differences between groups were analyzed by using 3-way repeated-measures analysis of variance, with the

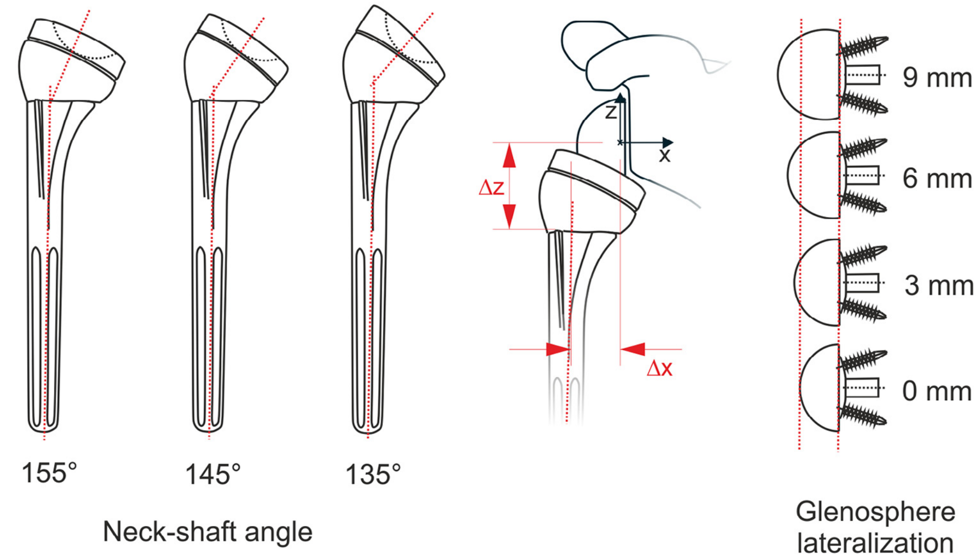

Figure 2 Overview of the implant configurations shows a schematic representation of the humeral components with 135°, 145°, and 155° neck-shaft angles. The 4 investigated glenoid components with 0 mm, 3 mm, 6 mm, and 9 mm glenosphere lateralization are also shown.

specimen as the repeated measure and neck-shaft angle as well as glenosphere lateralization and arm position as the relevant factors. Significant findings were further analyzed with the Tukey post hoc test. The significance level was set to $\alpha = 0.05$. For relevant data, means with standard deviation are presented.

Results

Dislocation force in dependence on joint abduction

Disregarding the effects of cofactors (glenosphere lateralization and neck-shaft angle), dislocation forces differed significantly with joint abduction and the external rotation angle. Thus, the anterior dislocation force at 60° of abduction and neutral external rotation was significantly higher than at 30° of abduction ($P < .0001$). The dislocation force at 60° of abduction and 30° of external rotation was the largest of the joint positions investigated. This was greater than at 30° of abduction and neutral rotation ($P < .0001$), at 30° of abduction and 30° external rotation ($P < .0001$), and at 60° abduction and neutral rotation ($P = .007$). Rotating the arm from neutral to external rotation at 30° of abduction but not at 60° of abduction resulted in an increase in the dislocation force ($P = .04$).

Dislocation force in dependence on glenosphere lateralization and neck-shaft angle

At 30° of abduction and neutral external rotation, the dislocation forces increased with the degree of glenosphere lateralization. Thus, dislocation forces increased from 0 mm (f_{avg} [force average across neck shaft angles] = 38.0 ± 23.5 N) to 6 mm (f_{avg} = 73.0 ± 49.5 N, $P = .0002$) and 9 mm (f_{avg} = 90.5 ± 52.5 N, $P < .0001$) of lateralization. Furthermore, 3 mm of lateralization resulted in significantly lower dislocation force (f_{avg} = 49.0 ± 33 N, $P = .0004$) compared with that with 9 mm of glenosphere lateralization. There was no difference in stability among the different neck-shaft angles in this joint position (Fig. 3, *A*).

At 30° of abduction and 30° of external rotation, dislocation forces were significantly lower for the 0-mm lateralized glenosphere (f_{avg} = 58.0 ± 24.5 N) than for the 6-mm (f_{avg} = 75.0 ± 41.5 N, $P = .03$) and 9-mm (f_{avg} = 84.5 ± 46 N, $P = .002$) lateralized glenosphere. The dislocation force for the 3-mm lateralized glenosphere (f_{avg} = 61.0 ± 34.0 N) was significantly less than for the 9-mm lateralized glenosphere ($P = .04$). At this joint position, a neck-shaft angle of 135° showed significantly higher dislocation force compared with that with a neck-shaft angle of 145° ($P = .02$) or 155° ($P = .02$; Fig. 3, *B*).

At 60° of abduction and neutral rotation, dislocation forces increased from the 0-mm lateralized glenosphere (f_{avg} = 54.5 ± 26.5 N) to the 6-mm (f_{avg} = 81.5 ± 47.5 N, $P < .0007$) and 9-mm (88.5 ± 41.5 N, $P < .0001$) lateralized glenosphere. Furthermore, the dislocation force for the 3-mm lateralized glenosphere (f_{avg} = 60.5 ± 37.0 N) was significantly lower than that of the 6-mm ($P = .04$) and 9-mm ($P = .0003$) lateralized glenosphere. There was no difference in dislocation force between the different neck-shaft angles (Fig. 4, *A*).

At 60° of abduction and 30° of external rotation, a 0-mm (f_{avg} = 75.0 ± 23.5 N, $P = .05$) and 3-mm (f_{avg} = 73.0 ± 26.5 N, $P = .04$) lateralized glenosphere showed significantly lower dislocation force compared with that with a 9-mm (f_{avg} = 84.5 ± 26.0 N) lateralized glenosphere. Again, the neck-shaft angle did not influence dislocation force in this arm position (Fig. 4, *B*).

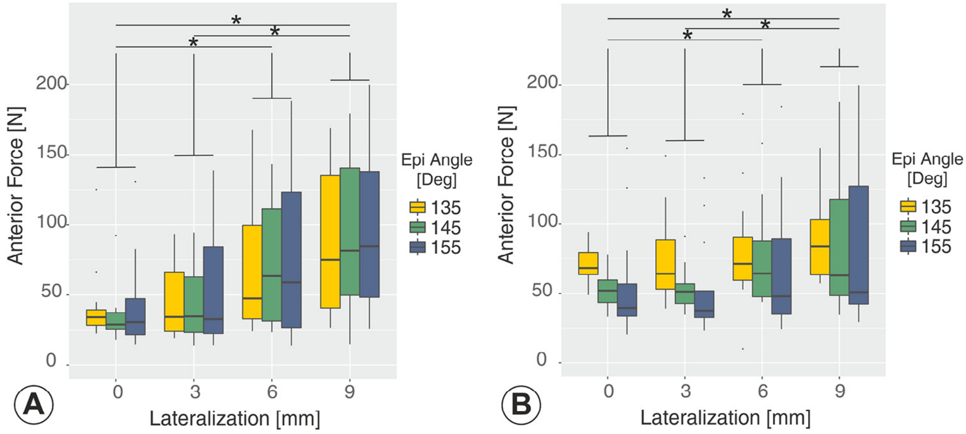

Figure 3 Anterior dislocation forces with the glenohumeral joint in (**A**) 30° of abduction and neutral rotation and (**B**) 30° of external rotation, in accordance with neck-shaft angle and glenosphere lateralization. The *horizontal line* in the middle of each box indicates the median, the *top and bottom borders* of the box mark the 75th and 25th percentiles, respectively, and the *vertical lines* mark the minimum and maximum of all of the data within the 1.5 times interquartile range of 25th and 75th percentile, respectively. Dots represent outliers, defined as values beyond the whiskers. *Significant differences between the configurations, with differing glenosphere lateralization ($\alpha = 0.05$).

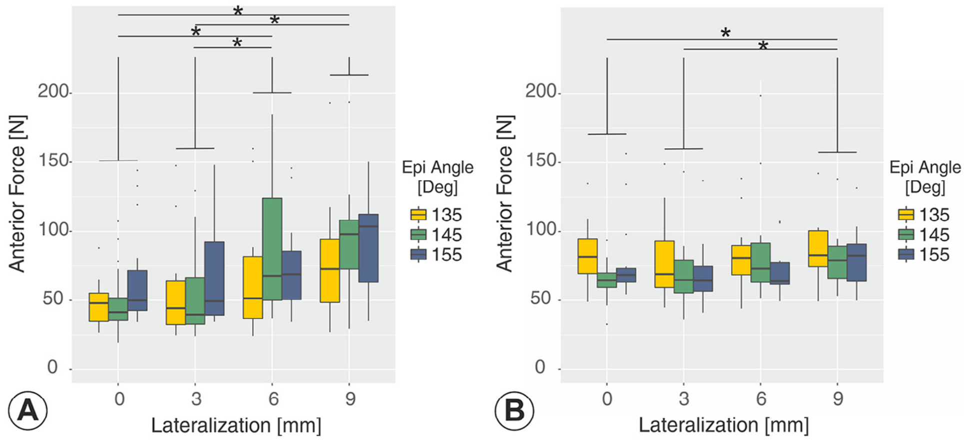

Figure 4 Anterior dislocation forces with the glenohumeral joint in (**A**) 60° of abduction and neutral rotation and (**B**) 30° of external rotation, in accordance with neck-shaft angle and glenosphere lateralization. The *horizontal line* in the middle of each box indicates the median, the *top and bottom borders* of the box mark the 75th and 25th percentiles, respectively, and the *vertical lines* mark the minimum and maximum of all of the data within the 1.5 times interquartile range of 25th and 75th percentile, respectively. Dots represent outliers, defined as values beyond the whiskers. *Significant differences between the configurations, with differing glenosphere lateralization ($\alpha = 0.05$).

Maximum external rotation

No effects of glenosphere lateralization or neck-shaft angle on maximum external rotation were found at 30° or 60° of abduction. Maximum external rotation reached 94.9° ± 19.0° and 103.4° ± 19.8° at 30° and 60° of abduction, respectively (Fig. 5).

Movement restrictions in abduction

The 60° abduction position could not be reached in every case. In 21 of 57 cases (37%) with a 9-mm glenosphere lateralization, the abduction angle of 60° could not be reached. This was also observed in 12 cases (21%) with 6-mm lateralization, 11 (19%) with 3-mm lateralization, and 4 (7%) with 0-mm lateralization. Among all cases in which 60° of abduction could not be reached, the neck-shaft angle was 155° in 31%, 145° in 27%, and 135° in 42%.

Discussion

Lateralizing the glenosphere and decreasing the neck-shaft angles are 2 implant design parameters that can reduce the risk of scapular impingement.[22,23,43] This study evaluated the effect of these design parameters on joint stability by quantifying the dislocation force using different implant configurations and different arm abduction and external rotation positions. Thus, we found maximum differences in dislocation forces, ranging from $f_{avg} = 38.0 \pm 23.5$ N to 90.5 ± 52.5 N for different degrees of lateralization of the glenosphere.

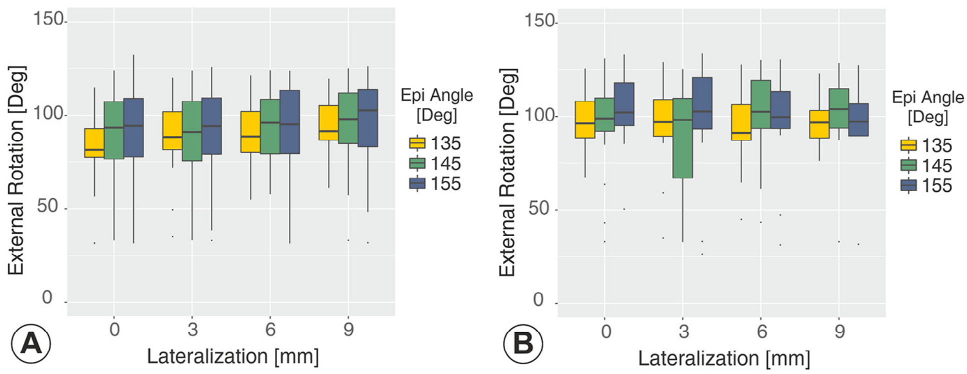

Figure 5 Maximum external rotation at (**A**) 30° and (**B**) 60° of abduction, in accordance with neck-shaft angle and glenosphere lateralization. The *horizontal line* in the middle of each box indicates the median, the *top and bottom borders* of the box mark the 75th and 25th percentiles, respectively, and the *vertical lines* mark the minimum and maximum of all of the data within the 1.5 times interquartile range of 25th and 75th percentile, respectively. Dots represent outliers, defined as values beyond the whiskers.

As hypothesized, increased glenosphere lateralization produced larger dislocation forces than configurations without additional lateralization. These results are in accordance with the findings of Costantini et al,[13] who found increased compressive joint contact forces for configurations with lateralization of the glenosphere. The improved stability can be explained by the increase in soft tissue tension.

Generally, an increase in joint stability as a result of lateralization seems to essentially depend on the constitution and integrity of the soft tissue envelope.[31] This is reflected in the significant increase in the range of measured dislocation forces with increasing amount of lateralization. Apart from increasing stability, lateralization and the associated increased soft tissue tension showed no negative effects on the passive external rotation range of motion in our study. Thus, the increased soft tissue tension does not seem to restrict this degree of freedom. This is supported by clinical studies that found improved external rotation after lateralized RSA.[4,14,37]

However, increasing stability with glenosphere lateralization also had drawbacks. In our study, abduction of the glenohumeral joint to 60° was not possible in some configurations and specimens, depending on the magnitude of glenosphere lateralization, due to excessive soft tissue tension. In a clinical context, this might result in increased compensatory demand and resulting stress on the scapulothoracic joint.[39] It is possible that glenosphere lateralization might have some negative effects on deltoid muscle loading[18,22] and might, for example, result in higher loading of the bone-implant interface.[16] In this context, we note that a simulation study by Denard et al[16] showed that the load on the bone-implant interface may be lower with implant-based lateralization of the glenosphere (as in our study) compared with that with bone graft-based lateralization directly after surgery.

The second major finding was the significant effect of the neck-shaft angle on joint stability, with higher dislocation forces for the 135° neck-shaft angle at 30° of glenohumeral abduction and 30° of external rotation compared with that for the 145° or 155° configuration. However, we observed no influence of the 3 neck-shaft angles in any of the other positions. These results support the observations made by Oh et al,[30] who also found similar results in a muscle-driven shoulder model.

This observation of the superiority of the 135° neck-shaft angle can be explained by the geometrically induced, stronger superolateral overlap of the humeral component compared with that with a greater neck-shaft angle. Accordingly, the opposite effect on joint stability has been observed for internally rotated joint positions, with higher dislocation forces for greater neck-shaft angles.[30] Furthermore, we found that the neck-shaft angle has no effect on maximum soft tissue-restrained external rotation.

Abduction significantly increased anterior stability of the glenohumeral joint. Furthermore, we observed significantly higher anterior stability with the arm in the externally rotated position compared with that with neutral external rotation. These results are also in agreement with our previous findings as well as those of Oh et al.[30] Although the increase in dislocation force with the arm in external rotation can be explained by the orientation of the bearing surface of the implant, the increase in stability when abducting the arm is caused by the increase in soft tissue tension.

The present study has some limitations that should be mentioned. Our biomechanical investigations focused only on the glenohumeral joint because the scapula was placed in a fixed position. Therefore, the influence of natural scapulothoracic motion could not be investigated.

Second, we only evaluated joint stability based on passive restraints of the natural soft tissue envelope. The active loaded rotator cuff muscles and the deltoid might, however, also influence the function of the joint. The anterior forces necessary to dislocate the joint are probably higher with active muscles. However, it seems unlikely that a simulation of the active musculature would have fundamentally changed our findings. Furthermore, experimentally simulating the active muscles while also retaining the intact physiologic soft tissue envelope is difficult to implement.

Third, we evaluated only anterior stability, which is, however, the most common clinical situation.

Fourth, the 60° abduction position could not be reached in some cases due to high soft tissue tension. The mean values of the dislocation forces in these positions might somewhat be higher than reported because the higher soft tissue tension in the missing specimens could have restrained motion.

Fifth, this was a cadaveric study at time-zero after implantation; thus, potential biological effects, such as soft tissue stretching or adaption, could not be investigated.

In summary, based on our observations, we suggest moderate implant-based glenosphere lateralization as the favored option in order to simultaneously reduce scapular impingement and to increase joint stability. Furthermore, glenosphere lateralization has a positive effect on recruitment of the rotator cuff muscles and on shoulder aesthetics, pushing the deltoid muscle more laterally and resulting in a more natural shoulder contour.[4]

Conclusion

We found that glenosphere lateralization significantly increased anterior stability of the glenohumeral joint without influencing the range of passive external rotation. However, excessive glenosphere lateralization might negatively affect the glenohumeral abduction range of motion. The 135° neck-shaft angle showed greater stability with the arm in external rotation compared with that with 145° and 155° configurations. Moderate glenosphere joint lateralization might be the favored option for increasing overall joint stability.

Disclaimer

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