Understanding The Mechanical Trade-offs In Changing Centers Of Rotation For Reverse Shoulder Arthroplasty Design

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Disclosures:

Introduction: Reverse Shoulder Arthroplasty (RSA) is increasingly being used to relieve pain and to restore function and range of motion for patients with shoulder arthritis and a deficient rotator cuff [1]. However, in the wake of favorable post-op results, RSA has exhibited high early to mid-term complication rates. The most common complication is scapular notching, which has been reported to occur after RSA in 31-97% of patients (CITE). Scapular notching is characterized by bone loss just inferior of the glenosphere baseplate and is caused by impingement of the humeral cup on the scapula in terminal adduction. The desire to prevent notching has led to different implant designs and placement to increase range of motion before impingement occurs. Most of these options involve lateralizing the center of rotation, usually by increasing the lateral offset of the gleno sphere or placing a bone graft behind the baseplate. Unfortunately, this lateralization also reduces the effective torque of the deltoid, potentially leading to difficulty elevating the arm or a decrease in strength.

There is clear need for a better understanding of the mechanical tradeoffs involved in RSA, especially those that involve moving the center of rotation. A finite element modeling approach was developed to provide greater insight into the mechanics of RSA.

Methods: Finite element models including an Aequalis RSA design (Tornier, Montbonnot, France) were created. Two common implantations were studied using 7 different variations. The first model involved inferiorly tilting the glenosphere 10°, according to a common implantation technique. Additional models were created of the Bony Increased Offset (BIO) technique, which utilizes bone harvested from the humeral head to lateralize the gleno sphere [2]. Six different bony offsets were modeled: 2.5, 5, 7.5, 10, 15, and 20 mm. The scapular geometry was segmented from a CT scan of the Visible Female. The implant geometries were obtained from 3D laser scans, with surfaces fit to the implant geometries using Geomagic Studio software (Geomagic USA, Morrisville, NC).

Finite element meshes were generated in TrueGrid (v 2.3 XYZ Scientific Application, Inc. Livermore, CA). The bone and polyethylene were modeled with linear elastic hexahedral brick elements, while the glenosphere and humeral cup and stem were modeled as rigid shell elements (Fig. 1), since they are substantially stiffer than the bone and polyethylene.
Four linear spring elements were placed symmetrically around the glenosphere in superior-anterior, superior-posterior, inferior-anterior, and inferior-posterior positions to model the capsule. The spring elements were given spring constants based upon the reported stiffness of the middle glenohumeral ligament [3,4,5]. In addition, a series of near-rigid slipring elements were used to model the deltoid muscle. Slipring elements were used as cable and pulley systems, with wrapping points represented as frictionless pulleys. In this way, the tension of the deltoid muscle can be transmitted through a cable whose line of action and wrapping points are dynamically linked to the humerus. The attachment point of the deltoid on the humerus was kept constant, but its initial position moved laterally away from the glenosphere, as necessitated by the increasing BIO offsets. The other deltoid slipring wrapping points were placed in the same global positions for all models.

All seven models underwent two finite element runs to test for impingement angle and the required isometric deltoid force for humeral abduction. The initial pose of the humerus was 40° of abduction. For the impingement angle test, a load representative of the weight of the arm was placed on the implant, and the humeral component was allowed to fall into adduction. The angle at which impingement occurred was recorded. For the deltoid muscle force test, the humeral component was loaded with the weight of the arm while an adduction moment (taken from clinical values from RSA patients collected with a dynamometer) was applied to the shoulder. The origins of the slipring elements were held fixed in space, enabling the cables to carry tension to counter the adduction torque resulting from the arm load and applied torque. The tensions in the slipring elements, analogous to deltoid force, were recorded. All jobs were run using Abaqus/Explicit 6.11.1 (Dassault Systèmes, Vélizy-Villacoublay, France).

**Results:** As was hypothesized, larger BIO offsets required larger deltoid muscle forces to prevent the humerus from falling into further adduction(Fig 2). On the other hand, larger BIO offsets also led to smaller impingement angles (a larger range of motion). At a BIO offset between 7.5 and 10 mm, the impingement angle recorded became negative, signifying that the humerus must go beyond anatomic neutral to impinge on the scapula.

**Discussion:** The results suggest that between 7.5 and 10 mm is optimum for BIO offset. In this range, impingement is prevented with only a 28% increase in deltoid muscle force as compared to neutral implantation. The effect of this decrease in deltoid muscle force needs to be further examined, along with the clinical trade-off between notching and strength. Ongoing work is aimed at more accurately modeling the capsule and the deltoid.

**Significance:**

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Figure 2. This graph shows the relationship between BIO offset, deltoid force required, and impingement angle. The red line marks 0° adduction or anatomic position. The inferior tilt model was plotted as 0 mm BIO size.