equinoxe® SHOULDER SYSTEM Reference Papers

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Scapular Notching Radiographic Analysis: Recommendations for Glenoid Plate Positioning and Glenosphere Overhang in Male and Female Patients

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Scope

This memo reports on the results of a radiographic study accepted for presentation at the 2012 Orthopaedic Research Society.[1]

Introduction

Scapular notching is the most common reported complication of reverse shoulder arthroplasty; occurring in 44 to 96% of patients having a reverse shoulder design with a medialized glenoid center of rotation (CoR) *(Table 1)*.[2-9]

Recent work has recommended design modifications for manufacturers and surgical technique modifications for surgeons to improve range of motion and stability and reduce scapular impingement.[10-19] The purpose of this radiographic analysis of 226 patients who received one particular reverse shoulder design (Equinoxe®; Exactech, Inc) and have been followed-up post-operatively for an average of 21.2 months is to correlate the position of the glenoid plate and the amount of glenosphere overhang to the clinical observation of notching. A statistical analysis of theses radiographic measurements will yield recommendations for prosthesis placement for male and female patients in this study population to avoid notching.

Methodology

Immediate post-op and the latest follow-up radiographs (21.2 ± 8.6 months) were collected from 226 patients (age = 72.9 ± 7.1 yrs; females = 155 and males = 71) who received a 38mm (n = 135), 42mm (n = 81) and 46mm (n=10) Equinoxe reverse shoulder by seven different surgeons at seven different institutions. Each patient's radiograph was scored for scapular notching by the implanting physician according to the Sirveaux grading scale *(Figure 1)*.[2] Placement of the glenoid baseplate peg from the inferior glenoid rim (e.g. craniocaudal positioning) and the amount of

Table 1. *Reported Scapular Notching Rate for Medialized CoR Reverse Shoulder Prostheses*

Study	Sample Size; Avg Follow-up	Scapular Notching Rate	Notches > Grade 2
Sirveaux; 2004[2]	n = 80; 44 months	64%	17%
Werner; 2005[3]	n = 48; 38 months	96%	46%
Boileau; 2006[4]	n = 45; 40 months	68%	11%
Simovitch; 2007[5]	n = 77; 44 months	44%	18%
Karelse; 2008[6]	n = 27; 43 months	59%	26%
Levigne; 2010[7]	n = 461; 51 months	68%	23%
Stechel; 2010[8]	n = 59; 48 months	87%	5%
Kempton; 2011[9]	Group 1: n = 43; 30 months (no glenoid tilt) Group 2: n = 28; 24 months (inferior glenoid tilt)	Group 1: 77% Group 2: 61%	Group 1: 23% Group 2: 4%
Weighted Average Scapular Notch Rate	46.0 months	**68.2%**	**20.9%**

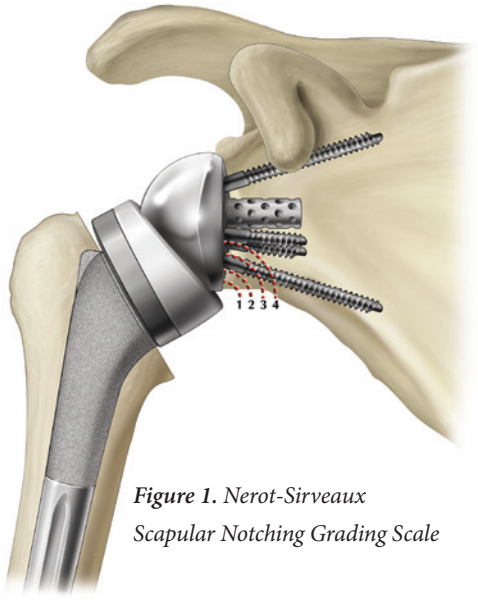
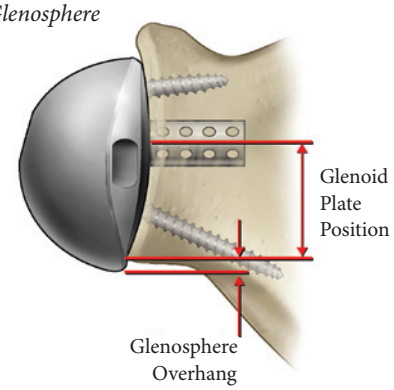


Figure 1. *Nerot-Sirveaux Scapular Notching Grading Scale*

glenosphere overhang was measured from each immediate post-operative AP (Grashey) radiograph using digital calipers *(Figure 2)*. Each radiographic measurement was compared to the surgeon scored notching grade; a student's two-tailed, unpaired t-test was used to identify differences in the radiographic measurements between patients with and without a notch, where $p<0.05$ denoted a significant difference.

Figure 2. *Measurement of Glenoid Plate Position and Glenosphere Overhang*



Results

The radiographic analysis demonstrated that 22 of 226 patients had a scapular notch (16 Grade 1 and 6 Grade 2; no Grade 3 or 4 notches were observed) for a scapular notching rate of 9.7%. Female patients (116 38mm and 39 42mm glenospheres) were observed to have a scapular notching rate of 9.7% (13 Grade 1 and 2 Grade 2) whereas male patients (19 38mm, 42 42mm and 10 46mm glenospheres) were observed to have a scapular notching rate of 9.9% (3 Grade 1 and 4 Grade 2). Patients with 38mm glenospheres were observed to have a scapular notching rate of 13.3% (13 Grade 1 and 5 Grade 2); patients with 42mm glenospheres were observed to have a scapular notching rate of 4.9% (3 Grade 1 and 1 Grade 2); and patients with 46mm glenospheres were observed to have a scapular notching rate of 0%.

The average glenoid plate position for patients without a notch (19.1 ± 2.5mm) was significantly lower on the glenoid ($p = 0.037$) than the average position for patients with a notch (20.1 ± 2.4mm). The average glenosphere overhang for patients without a notch (5.3 ± 2.5mm) was significantly more ($p = 0.002$) than the average overhang for patients with a notch (3.6 ± 2.4mm).

Regarding differences for females, the average glenoid plate position for female patients without a notch (18.5 ± 2.5mm) was significantly lower on the glenoid ($p = 0.026$) than the average position for females with a notch (20.0 ± 1.8mm). The average glenosphere overhang for female patients without a notch (5.4 ± 2.5mm) was significantly more ($p = 0.0025$) than the average overhang for females with a notch (3.4 ± 1.6mm). Regarding differences for males, the average glenoid plate position for male patients without a notch (20.0 ± 4.9mm) was not significantly different than the average position for males with a notch (20.6 ± 3.7mm). The average glenosphere overhang for male patients without a notch (4.9 ± 2.7mm) was not significantly different than the average overhang for males with a notch (3.6 ± 3.9mm).

Comparing males and females, the average glenoid plate position for female patients without a notch was significantly less ($p = 0.00003$) than the average position for males without a notch. For 38mm and 42mm glenospheres, the average glenoid plate position for females was significantly lower on the glenoid ($p = 0.006$ and 0.015, respectively) than the average position for males. Similarly, for 38mm and 42mm glenospheres, the average glenosphere overhang for females was significantly less ($p = 0.001$ and 0.009, respectively) than the average overhang for males.

Using the 95% confidence intervals on the average glenoid plate position and glenosphere overhang for male and female patients in this study population without a notch yields the following recommendations on placement of the glenoid plate and the minimum glenosphere overhang necessary to avoid scapular notching. For females, if the cage peg of the glenoid plate is placed at 18.1mm from the inferior glenoid rim, it would reduce the female scapular notching rate from 9.7% to 1.9%. For males, if the cage peg of the glenoid plate is placed 19.5mm from the inferior glenoid rim, it would reduce the male scapular notching rate from 9.9% to 1.4%. Similarly for females, a minimum glenosphere overhang of 5.9mm would reduce the female scapular notching rate from 9.7% to 0.7%. For males, a minimum glenosphere overhang of 5.6mm would reduce the male scapular notching rate from 9.9% to 2.8%.

Discussion and Conclusion

Scapular notching is the most commonly reported complication of reverse shoulder arthroplasty and generally deemed to be a function of a medialized glenoid CoR. This study reports a 9.7% scapular notching rate with a reverse shoulder prosthesis whose CoR is slightly lateralized (2mm) relative to the glenoid, where only 2.7% have a grade 2 notch and no grade 3 or 4 notches were observed at a mean follow-up of 21.2 months. These radiographic

EQUINOXE REVERSE SHOULDER DESIGN FEATURES

1. 4mm Superiorly Shifted Glenoid Baseplate Cage Peg
2. Glenosphere Distal Offset
   38mm = 2.25mm
   42mm = 4.25mm
   46mm = 6.25mm
3. 145° Humeral Neck Angle
4. Six Polyaxial Locking Screws

results for scapular notching are very favorable (~7x reduction in the overall scapular notching rate) relative to other published complication rates: as described in Table 1, the weighted average scapular notching rate reported for medialized glenoid CoR reverse shoulder designs is 68.2%, where 20.9% have notch > grade 2.[2-9] These results confirm the conclusions of previous work which demonstrated that subtle prosthesis designs changes (i.e. inferiorly shifted glenosphere/superiorly shifted baseplate peg, curved back glenoid plate, 145° humeral neck, 2mm lateralized CoR) can dramatically reduce impingement and improve range of motion.[11,12,15,17]

The results of this study also demonstrate significant differences in both the glenoid plate position and glenosphere overhang between males and females and between patients with and without a notch. Gender differences result from differences in bone size, reflected by the larger percentage of males who received a 42 or 46mm glenosphere (73.2% vs 24.5% of females who received a 42 or 46mm glenosphere). Given this implant size distribution, this study identifies differences and makes recommendations for optimal implant placement in order to reduce notching in males and females. These recommendations are specific to the Equinoxe reverse shoulder; care should be made when extrapolating these results to other reverse shoulder devices due to differences in design parameters. Additionally, there is a functional limit to how much glenosphere overhang is achievable; implant positioning should take a particular patient's soft tissue laxity into account. The primary limitation of this study is the degree that the study population represents the global reverse shoulder patient population; this concern is mitigated by the large sample size (n = 226) and wide distribution of surgery sites (seven institutions: three teaching and four private hospitals; two different countries).

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A Correlation of Five Commonly Used Clinical Metrics to Measure Outcomes in Reverse Shoulder Arthroplasty

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Scope

This memo reports on the results of a clinical outcomes study accepted for presentation at the 2012 Orthopaedic Research Society.[1]

Introduction

Standardizing the tools and methods by which healthcare professionals collect clinical outcomes is a critical component of evidence-based medicine. Numerous outcome measurement scores are available to evaluate the success of treatment of patients with debilitating conditions in the shoulder. The most commonly used scoring systems are the Simple Shoulder Test (SST), the UCLA Score, the American Shoulder and Elbow Surgery Score (ASES), the Constant-Murley (Constant), the Shoulder Pain and Disability Index (SPADI), the Disabilities of the Arm, Shoulder, and Hand (DASH), the Rowe Score, and the Oxford Shoulder Score. While each metric attempts to rate the quality of care, each varies by the method in which it gauges the success of treatment (based upon the restoration of function, motion, and strength and by the reduction of pain). Additionally, each metric varies in how it distributes and weighs subjective patient responses and clinical observations from objective clinician/independent examiner assessment measurements.

In the US, the five most commonly used scoring systems for shoulder arthroplasty are the SST, UCLA, ASES, Constant, and SPADI clinical metrics. The SST score is derived from a series of 12 Yes /No questions that measure the patient's ability to carry out activities of daily living; 12 is the highest/best score. The UCLA score is derived from a series of 5 questions that evaluates pain, satisfaction of treatment, and restoration of function, strength, and motion; 35 is the highest/best score. The ASES score is derived from a series of 11 questions that evaluates pain (50%) and restoration of function (50%); 100 is the highest/best score. The Constant score is derived from a series of 23 questions that evaluates pain (15%), restoration of function (20%), range of motion (40%), and strength/power (25%); 100 is the highest/best score. The SPADI score is derived from a series of 13 questions that evaluates pain and restoration of function; 130 is the highest score and 0 is the best score. The pre-op and post-op outcomes data was collected and scored using the 5 aforementioned metrics on 45 patients who received a primary reverse shoulder for the treatment of cuff tear arthropathy (CTA). The pre-op and post-op scores were normalized, correlated, and compared to gain a better understanding of the relationship between the metrics.

Methodology

45 Patients (age = 73.9 ± 5.9 yrs; 34 females; 34 right shoulders) received a primary Equinoxe reverse shoulder (Exactech, Inc; 30 38mm glenospheres and 15 42mm glenospheres) by the senior author (PHF) through the delto-pectoral approach between May 2007 and June 2010 for treatment of CTA. These patients were evaluated and scored pre-operatively and at latest follow-up using the SST, UCLA, ASES, Constant, and SPADI scoring metrics; the average follow-up for all patients was 25.3 ± 10.3 months. A Student's two-tailed, paired t-test was used to identify differences in pre-operative and post-operative results, where $p<0.05$ denoted a significant difference. In order to compare the results on the same scale, all 5 metrics were normalized on a 100 point scale. The normalized scores were then correlated to one another to gain a better understand the relationship between the metrics.

Results

The average pre-op and post-op outcomes scores are presented in Table 1. The average pre-op and post-op objective clinician assessment for active abduction, active forward flexion, and active external rotation are presented in Table 2. No instances of instability or glenoid loosening were reported; the only

complications observed were 4 instances of scapular notching (3 Grade 1 and 1 Grade 2) for a rate of 8.9%. Table 3 presents the method by which each metric was normalized to a 100 point scale and also presents the normalized pre-op and post-op values for each scoring metric. Table 4 presents the correlation between the scoring systems.

Comparing the normalized pre-op scores, the SST score was significantly different than UCLA (p<<<0.00), ASES (p<<<0.00), Constant (p<<< 0.00), and SPADI (p<<< 0.00). The UCLA score was significantly different that the ASES (p<<<0.00) and Constant (p<<<0.00). The ASES score was significantly different than the Constant score (p=0.027). Comparing the normalized post-op scores, the SST score was significantly different than UCLA (p=0.0127), ASES (p=0.0139), Constant (p<<<0.00), and SPADI (p=0.013). The UCLA score was significantly different than the Constant (p<<<0.00) and SPADI (p=0.022). The ASES score was significantly different than the Constant (p<<<0.00). The Constant score was significantly different than the SPADI (p<<<0.00).

Discussion and Conclusion

The results of this study demonstrate that all 5 clinical metrics demonstrated significant improvements in treatment using the Equinoxe reverse shoulder at a mean follow-up of 25.3 months. These motion and outcome score results are favorable relative to other published motion and outcome scores: Tables 5 and 6 compares the pre- and post-op motion and outcome scores in this study relative to that previously reported for other reverse shoulder designs, respectively. All 5 clinical metrics utilized in this outcomes study to measure the short term results of a reverse shoulder prosthesis are all very highly correlated (>94%). Despite this agreement and high degree of correlation, there was a wide difference in the percent increase and mean values between the normalized pre-op and post-op scores. These differences likely arise from the different weights within the scoring systems (e.g. reduction of pain, restoration of function, motion and strength) and from how subjective measurements and objective measurements are distributed. Additional work is required to better understand how these differences in weights and methods within systems influence the evaluation and scoring of patient treatment.

Table 1. *Average Pre- and Post-op Outcome Scores (not normalized)*

	SST	UCLA	ASES	Constant	SPADI
Pre op Avg ± St Dev	2.4 ± 1.3	13.3 ± 3.4	29.3 ± 8.2	32.6 ± 10.4	83.7 ± 14.9
Post op Avg ± St Dev	10.0 ± 2.1	31.2 ± 2.9	87.8 ± 13.3	70.7 ± 11.7	17.0 ± 16.5
P value	**<<0.00**	**<<0.00**	**<<0.00**	**<<0.00**	**<<0.00**

Table 2. *Average Pre- and Post-op Motion Data*

	Active Abduction	Active Forward Flexion	Active External Rotation
Pre op Avg ± St Dev	83.4 ± 27.4	118.1 ± 42.9	11.6 ± 20.5
Post op Avg ± St Dev	101.6 ± 19.0	145.7 ± 20.2	35.8 ± 15.7
P value	0.0001	0.0001	0.00001

Table 3. *Average Pre- and Post-op Outcome Scores (Normalized)*

	SST	UCLA	ASES	Constant	SPADI
Normalizing Factor	Score* 100/12	Score* 100/35	Score* 100/100	Score* 100/100	100-(Score* 100/130)
Normalized Pre-op Score	19.6 ± 10.8	38.0 ± 9.7	29.3 ± 8.2	32.6 ± 10.4	35.6 ± 11.4
Normalized Post op Score	83.3 ± 17.3	89.3 ± 8.2	87.8 ± 13.3	70.7 ± 11.7	86.9 ± 12.7
% Increase	324.5%	135.1%	200.0%	116.8%	143.9%

Table 4. *Correlation of the 5 Clinical Outcome Metrics*

	SST	UCLA	ASES	Constant	SPADI
SST	1				
UCLA	0.943				
ASES	0.962	0.977	1		
Constant	0.960	0.959	0.958	1	
SPADI	0.959	0.950	0.958	0.971	1

Table 5. Comparison of Reverse Shoulder Motion Data Reported in Literature

Study	Pre-op Avg Active Abduction	Post-op Avg Active Abduction	Pre-op Avg Active Forward Flexion	Post-op Avg Active Forward Flexion	Pre-op Avg Active External Rotation (arm at side)	Post-op Avg Active External Rotation (arm at side)
Sirveaux, 2004[2]	*	*	73	121	3.5	11.2
Werner, 2005[3]	43 (0 to 90)	90 (0 to 165)	42 (0 to 90)	100 (0 to 145)	17 (-20 to 70)	12 (-50 to 60)
Frankle, 2005[4]	41.4 (0 to 110)	101.8 (30 to 180)	55.0 (0 to 120)	105.1 (30 to 180)	12.0 (-15 to 45)	41.1 (10 to 65)
Boileau, 2006[5]	*	*	55 (95% CI: 47 to 63)	121 (95% CI: 111 to 131)	7 (95% CI: 1 to 13)	11 (95% CI: 5 to 16)
Levigne, 2008[6]	*	*	70	125	7	9
Stechel, 2010[7]	46	93	47	105	-9	19
Nolan, 2011[8]	*	*	61.2 (0 to 137)	121.3 (52 to 170)	13.8 (-35 to 60)	14.6 (-44 to 60)
Flurin, 2012[1]	83.4 ± 27.4	101.6 ± 19.0	118.1 ± 42.9	145.7 ± 20.2	11.6 ± 20.5	35.8 ± 15.7

**denotes measurement not reported.*

Table 6. Comparison of Reverse Shoulder Outcome Scores Reported in Literature

Study	Sample Size	Avg Follow-up (months)	Pre-op Avg Constant Score	Post-op Avg Constant Score	Pre-op Avg ASES Score	Post-op Avg ASES Score
Sirveaux, 2004[2]	80	44	22.6 (4 to 50)	65.5 (34 to 85)	*	*
Werner, 2005[3]	58	38	29 (3 to 53)	64 (10 to 100)	*	*
Frankle, 2005[4]	60	33	*	*	34.3 (0 to 65)	68.2 (15 to 100)
Boileau, 2006[5]	45	40	17 (95% CI: 14 to 19)	58 (95% CI: 51 to 64)	*	*
Levigne, 2008[6]	337	47	23	58	*	*
Stechel, 2010[7]	59	48	15 (2 to 55)	55 (17 to 96)	*	*
Nolan, 2011[8]	71	24	27.5 (5 to 58)	61.8 (30 to 87)	26 (0 to 63)	76.1 (21 to 100)
Flurin, 2012[1]	45	25	32.6 ± 10.4	70.7 ± 11.7	29.3 ± 8.2	87.8 ± 13.3

**denotes measurement not reported.*

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A Comparison of Glenoid Fixation Using Two Different Reverse Shoulder Designs with an Equivalent Center of Rotation in a Low and High Density Bone Substitute

This memo reports on the results of a computer analysis presented at the 2013 Orthopaedic Research Society.[1]

This paper has been accepted for publication in JSES.

Introduction

Initial fixation of the noncemented reverse shoulder glenoid baseplate is critical to achieve a stable bone/implant interface during the first few months after surgery to potentiate biologic fixation and avoid aseptic glenoid loosening. The Grammont reverse shoulder is notable relative to historical reverse shoulder designs for its low glenoid loosening rate.[2] Short and mid-term clinical outcome studies have reported aseptic glenoid loosening rates between 0 and 12% with modern reverse shoulder arthroplasty designs,[3-6] with an average rate of 5%.[7] These and other concerns led Sirveaux et al. to project a survival curve with failure defined as revision at 5, 7, and 8 years to be 91.3, 74.6, and 29.8%, respectively.[3] This low loosening rate is commonly attributed to its placement of the center of rotation directly on the face of the glenoid; thereby, limiting the torque on the bone/implant interface.[10] Other design features less commonly attributed to this low loosening rate include a circular profile, flat-backed baseplate with a press-fit 8mm central peg, and options for up to 4 poly-axial compression, locking, or compression-locking screws, depending upon manufacturer. Over the past decade, designers have modified the Grammont design in an attempt to improve implant performance and further decrease the types of complications and their associated rates. Design variations include: baseplate profile (circle vs. oval shape), baseplate size (25 to 34mm), backside geometry (flat vs. curved back), center of rotation (0 to 1cm lateral to glenoid surface), surface finish and coatings (grit blasted vs. porous coated vs. Hydroxapatite coated), fixation screw diameters (3.5 to 6.5mm), number of fixation screw options (2 to 6), and type of screw fixation (poly-axial compression vs. locking vs. compression-locking).

A reverse shoulder glenoid loosening method was presented previously to assess initial glenoid fixation for the purpose of analyzing implant performance.[8-9] This study utilizes that methodology to compare the initial fixation associated with two commercially-available reverse shoulder prostheses having an equivalent center of rotation in both low and high density bone-substitute substrates. The goal of this study is to refine our knowledge of how implant geometry and design contributes to fixation and determine if parameters other than center of rotation impact initial fixation. We evaluate the null hypothesis that 2 different reverse shoulder designs having an equivalent center of rotation will exhibit equivalent initial fixation in both low and high density bone-substitute before and after cyclic loading.

Methods

This study evaluated the initial glenoid fixation associated with the 36mm Trabecular metal reverse shoulder (Zimmer, Inc; Warsaw, IN) and the 38mm Equinoxe® reverse shoulder (Exactech, Inc; Gainesville, FL) in two different densities (0.24 and 0.48 g/cm^3) of polyurethane blocks (Pacific Research, Inc.; Vashon, WA), each conforming to ASTM F 1839. Both reverse shoulder designs have been available in the US market for over 6 years, have an equivalent center of rotation that is located approximately 2.5mm lateral to the face of the reamed glenoid surface, and achieve initial fixation using a press-fit peg and poly-axial compression screws with supplemental locking caps. The primary differences between the glenoid baseplate designs is that the Trabecular metal device is 28mm in diameter, is flat backed, fully porous, and utilizes 2 locking-compression screws; whereas, the Equinoxe glenoid baseplate is 34mm long and 25mm wide, is curved back, grit blasted, and utilizes up to 6 locking-compression screws. (Table 1 and Figure 1) All glenoid components were fixed to each density substrate using 4.5x30mm polyaxial compression screws locked with caps: according to the manufacturers recommended technique[11,12] the trabecular metal device was secured using 2 screws and the Equinoxe was secured using 4 screws. The peg of each glenoid baseplate was press-fit using a drill diameters specified in each manufacturer's surgical technique. Each reverse shoulder design was tested in 7 low density and 7 high density bone substitute blocks, for a total of 28 tests.

The reverse shoulder glenoid loosening method consists of two tests: a displacement test and a cyclic test, and is conducted in three phases: phase 1) pre-cyclic displacement test, phase 2) cyclic test, and phase 3) post-cyclic displacement test. In the displacement test, the axial test machine (Instron Corp;

Table 1: *Comparison of reverse shoulder designs used in this glenoid loosening study*

	Trabecular Metal Reverse Shoulder	Exactech Reverse Shoulder
Surface Texture	Porous Trabecular metal	Grit-blasted Titanium
Center of Rotation	2.5mm lateral to bone/implant interface	2.5mm lateral to bone/ implant interface
Glenosphere Diameter	36mm	38mm
Glenoid Plate Profile	Circle: 28 mm diameter	Oval: 34 mm long, 25 mm wide
Glenoid Baseplate Backside Geometry	Flat-back	Spherically-curved back
Press-fit Central Post	Porous peg: 8x15mm (0.5mm press-fit in good bone, 0.5 to 1.5mm press-fit in bad bone)	Tapered cage peg: 8x16mm (0.2 to 0.7mm press-fit)
Poly-Axial Compression Screws with Locking Caps	2, 4.5x30mm self-tapping bone screws	4, 4.5x30mm self-tapping screws

Figure 1: *Trabecular Metal (Zimmer, left) and Equinoxe (Exactech, right) reverse shoulder prostheses*

Norwood, MA. Resolution of 1 micron) and 3 digital indicators (Mitutoyo, Japan. Resolution of 1 micron) measure displacement as a 50 N compressive axial load is applied perpendicular to the glenoid and a 357 N shear load is applied parallel to the face of the glenoid baseplate along its superior/inferior (S/I) axis and then performed a second time turning the component 90° and loading it along its anterior/posterior (A/P) axis. Dial indicators are used to subtract out any compliance of the test construct; displacement was measured in the direction of the applied shear and compression loads to the nearest micron and applied along both the S/I and A/P axes of each prosthesis. It should be noted that the compressive and shear loads were applied directly to the Trabecular metal glenosphere (because the glenosphere was connected with a taper and removing the taper could adversely affect fixation) while the compressive and shear loads were applied directly to the Equinoxe baseplate (because the glenosphere was connected with a screw and was easily removed without affecting fixation). (Figure 2) Additionally, the magnitude of the compressive load was reduced compared to the previously presented methodology; this was done to further challenge each prosthesis as it is believed these devices fail in shear.[8-9]

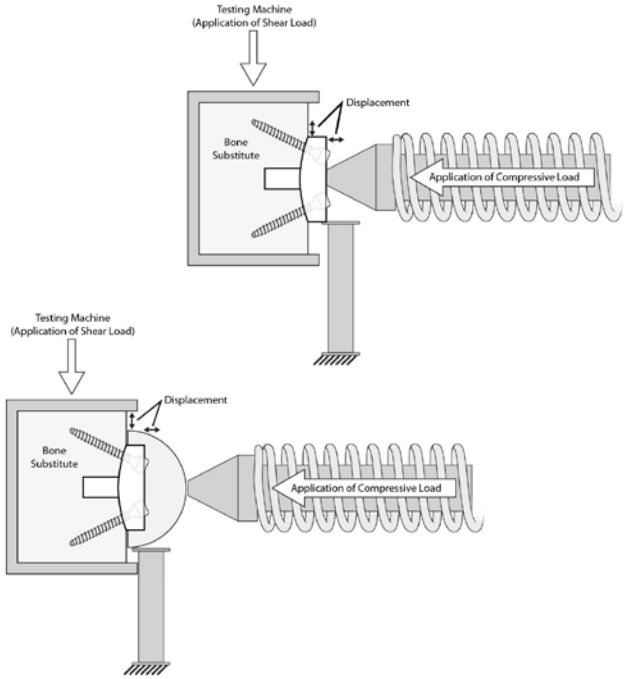

Figure 2: *Depiction of the displacement test in which shear and compressive loads were applied directly to the baseplate of the Equinoxe device (top) while the shear and compressive loads were applied directly to the glenosphere of the Trabecular metal device (bottom), before and after cyclic loading.*

In the cyclic test, a 750N axial load is constantly applied through the center of the humeral liner as the glenosphere/glenoid baseplate/bone-substitute block are rotated about the humeral component with a stepper motor to create a sinusoidal angular displacement profile encompassing an arc of 55° at 0.5 Hz for 10,000 cycles. (Figure 3) This 55° arc of motion was inferiorly biased by 10° relative to the previously presented methodology in order to simulate a lower degree of abduction (e.g. 15 to 70° humeral abduction in the scapular plane relative to a fixed scapula). This loading profile (over the 55° arc) would induce a maximum shear load of 456N (with a corresponding compressive load of 595N) at the lower extreme of rotation and a maximum compressive load of 750N (with no corresponding shear load) when applied perpendicular to the baseplate. It should be noted that a 36mm humeral liner with a 145° neck angle was used to test the Trabecular metal prosthesis while a 38mm humeral liner with a 145° neck angle was used to test the Equinoxe prosthesis. The Trabecular metal device actually has a 147.5° humeral neck angle; however, the neck angle was reduced to 145° to ensure each device was subjected to the same combination of shear and compression loads during the cyclic test. The components were cooled with a continuous jet of air with no lubrication during the cyclic test. Statistical analysis was performed by means of a two-tailed unpaired student's t-test (significance defined as $p < 0.05$) to compare S/I and A/P prosthesis displacements relative to each density block in the direction of the applied shear load before and after cyclic loading.

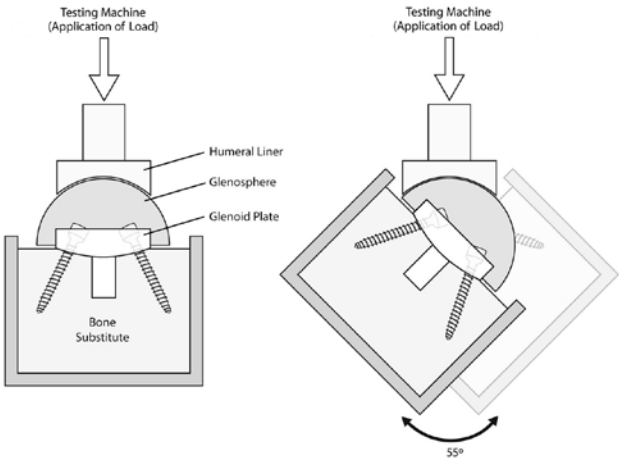

Figure 3: *Depiction of the cyclic test in which a 750N load is applied through the humeral liner as the glenoid component is cycled about an arc of 55° at 0.5 Hz for 10,000 cycles*

Results

The average S/I and A/P pre-cyclic and post-cyclic glenoid baseplate shear displacement for each reverse shoulder design in the low and high density substrates are presented in Tables 2 and 3, respectively. During the cyclic test in the low density bone substitute, 6 of the 7 36mm Trabecular metal reverse shoulders catastrophically loosened after an average of 2603 ± 980 cycles (range: 1144 to 3810); (Figure 4) all of the 38mm Equinoxe remained well-fixed after 10k cycles. In the cyclic test using the high density bone substitute, all of the 36mm Trabecular metal and 38mm Equinoxe reverse shoulders remained well fixed after 10k cycles of loading. As described in Tables 2 and 3, all S/I and A/P shear displacements before and after cyclic loading associated with the 36mm Trabecular metal reverse shoulder were significantly greater than that of the 38mm Equinoxe reverse shoulder in both low and high density bone substitutes.

Table 2: *Comparison of average glenoid plate/glenosphere shear S/I and A/P motion in the low density polyurethane substitutes*

Displacement (microns)	36mm Trabecular Metal	38mm Equinoxe	p-values
S/I pre-cyclic, 0.24 g/cm³	381 ± 59	181 ± 30	<0.001
S/I post-cyclic, 0.24 g/cm³	NA	186 ± 34	NA
A/P pre-cyclic, 0.24 g/cm³	481 ± 73	180 ± 58	<0.001
A/P post-cyclic, 0.24 g/cm³	NA	181 ± 70	NA

Table 3: *Comparison of average glenoid plate/glenosphere shear S/I and A/P motion in the high density polyurethane substitutes*

Displacement (microns)	36mm Trabecular Metal	38mm Equinoxe	p-values
S/I pre-cyclic, 0.48 g/cm³	247 ± 66	102 ± 11	<0.001
S/I post-cyclic, 0.48 g/cm³	207 ± 65	112 ± 28	0.004
A/P pre-cyclic, 0.48 g/cm³	254 ± 73	98 ± 31	<0.001
A/P post-cyclic, 0.48 g/cm³	269 ± 161	96 ± 25	0.016

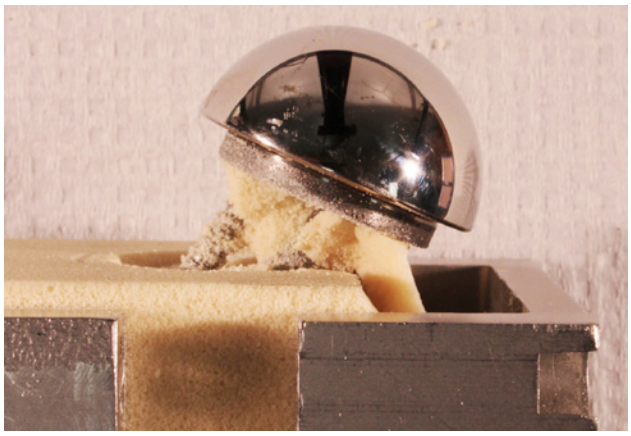

Figure 4: *Representative image of the Trabecular metal reverse shoulder in the low density substitute after disassociation during the cyclic test*

Discussion

The results of this study demonstrate that the 36mm Trabecular metal reverse shoulder is associated with significantly more motion in both the A/P and S/I shear directions following 10k cycles of 750 N loading in both a low and high density bone substitute than the 38mm Equinoxe reverse shoulder. Despite numerous similarities in design (e.g. same center of rotation (2.5mm), same humeral neck angle (145°), same diameter (8mm) press-fit post, same diameter and length (4.5x30mm) self-tapping poly-axial compression screws with locking caps), significant differences in glenoid fixation were observed between these reverse shoulder prostheses.

The oval Equinoxe baseplate is bigger (25x34mm) than the Zimmer Trabecular metal baseplate (28mm diameter); therefore, the Equinoxe has a larger surface area to distribute the applied loads. Additionally, the Equinoxe baseplate has a curved-back that theoretically converts some of the applied shear to compression; the Zimmer baseplate is flat-backed and may be more susceptible to rocking in shear. The Equinoxe baseplate provides up to 6 locations for screw fixation (though only 4 were used in this study) while the Zimmer baseplate provides 2. The distribution of screw position is also different, the Equinoxe has 2 screws along the superior/inferior axis located 19.5mm from one another and 2 anterior and 2 posterior screws located 13.5mm from one another; the Zimmer has 2 screws along the superior/inferior axis located 13.5mm from one another. Finally, the Equinoxe baseplate is a grit-blasted while the Zimmer baseplate is porous; whereas, a porous surface is expected to have better biologic fixation potential it is also possible that a porous surface may abrade the substrate in the presence of significant micromotion and increase the probability of loosening. It is unclear which design parameters were most responsible for the observed differences in stability; however, it is clear that these design differences (e.g. larger baseplate surface

area, curved-backside geometry, more locations for screw fixation, and a wider distribution of screw fixation) contributed to the observed improvements in fixation associated with the Equinoxe.

It should be noted that this reverse shoulder glenoid loosening test method only evaluates initial fixation and makes no attempt to simulate biologic fixation; as such, it is difficult to extrapolate if 2600 cycles of 750N loading represents sufficient time to achieve biologic fixation in the porous glenoid baseplate. Nevertheless, these results demonstrate that subtle changes in glenoid baseplate design can have a dramatic impact on initial fixation, particularly in a low density bone substitute which is intended to simulate the bone quality of the recipient population for reverse shoulders.

Conclusions

These results are the first to demonstrate that differences in fixation exist between baseplate designs having an equivalent center of rotation in both low and high density polyurethane bone substitutes. For this reason, we reject the null hypothesis and conclude that reverse shoulder glenoid design parameters, other than the position of the center of rotation, significantly impact fixation. Subtle changes in glenoid baseplate design can dramatically impact fixation, particularly in low density bone substitutes which are intended to simulate the bone quality of the recipient population for reverse shoulders. Future work should attempt to isolate differences in design and evaluate which parameters are the most important contributors to achieve glenoid fixation in reverse shoulder arthroplasty.

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Computer Assessment of Scapula Cortical and Cancellous Bone Removal When Correcting a Posterior Defect Using Three Different Glenoid Prosthesis Designs

This memo reports on the results of a computer analysis presented at the 2013 Orthopaedic Research Society.[1]

Introduction

Posterior glenoid wear is common in glenohumeral osteoarthritis. Tightening of the subscapularis and the anterior musculature of the shoulder causes posterior humeral head subluxation and a posterior load concentration on the glenoid. This reduced contact area causes glenoid wear and eventually posterior instability. Farron et al. demonstrated that glenoid prostheses implanted with posterior wear are associated with increased stress in the bone, cement, and implant and are also associated with increased implant micromotion, all of which could lead to one of the most common complications of total shoulder arthroplasty: glenoid loosening.[2] To correct posterior wear, shoulder surgeons typically eccentrically ream the anterior glenoid to re-center the humeral head prior to resurfacing with a prosthesis. Unfortunately, the technique of eccentric reaming undermines prosthesis support by requiring the removal of the stronger (non-worn) anterior cortical glenoid bone. This removal of bone creates a functional limit of 10 to 15° of eccentric correction prior to the glenoid bone being too small to support the implant and/or cement fixation becomes compromised due to peg perforation.[2-5]

As a result of these challenges, orthopaedic device manufacturers have developed posteriorly augmented glenoid implants to minimize the removal of anterior glenoid bone when attempting to restore glenoid retroversion and re-center the humeral head when performing a total shoulder arthroplasty in a patient with a posteriorly worn glenoid. These augmented glenoid implants are provided in two general styles: wedge (Equinoxe® posterior augment glenoid; Exactech, Inc; Gainesville, FL) and step (Global Step-Tech®; Depuy, Inc; Warsaw, IN). (Figure 1) The purpose of this study is to quantify the amount of cortical and cancellous glenoid bone removed to correct three different sizes of posterior glenoid defects using 3 different glenoid prosthesis designs: 1) the traditional method of eccentric reaming and using a standard glenoid (Equinoxe standard nonaugmented pegged), 2) using a 8, 12, and 16° Equinoxe posterior augment glenoid, and 3) using a 3, 5, and 7mm Global Step-Tech posterior augment glenoid.

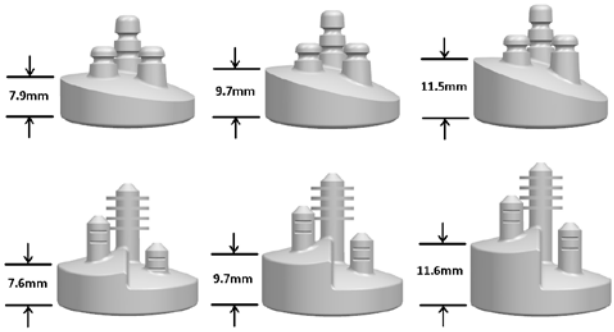

Figure 1: *Representative Images of the Exactech Equinoxe 8, 12, and 16° posterior augment glenoid (top row, left to right, respectively) and the Depuy Global Step-Tech 3, 5, and 7mm posterior augment glenoid (bottom row, left to right, respectively)*

Methods

A computer analysis was conducted to quantify and compare the cortical and cancellous bone removed when correcting three sizes of posterior glenoid defects using 3 different glenoid prostheses. Digital models of all 3 glenoid implants were created in a 3-D computer modeling software (Unigraphics; Siemens PLM; Plano, TX, USA). A cortical/cancellous digital scapula and humerus (Pacific Research, Inc.; Vashon, WA) were assembled and 3 sizes (small, medium, and large) of posterior glenoid defects were created in the digital scapula by posteriorly shifting the humeral head by 5.6mm (until greater tuberosity impingement with the acromion) and then medially translated the humeral head by 7.5, 9, and 10.5mm into the scapula, respectively. (Figure 2) The digital scapula and humerus had a uniform thick 1mm shell to

simulate cortical bone; the size of each defect was determined by volume calculations of each cortical and cancellous bone model. The same size of each posterior augment glenoid implant type was then seated in each corresponding glenoid defect size; cortical and cancellous bone were removed from the bone model to restore the retroversion of the original scapula and permit fully-seating of each glenoid implant. To clarify, the 8° wedge and 3mm step posterior augment glenoids were seated in the small defect scapula, the 12° wedge and 5mm step posterior augment glenoids were seated in the medium defect scapula, and the 16° wedge and 7mm step posterior augment glenoids were seated in the large defect scapula. The standard/nonaugmented pegged glenoid acted as the control in this analysis by quantifying bone removed to simulate the traditional method of eccentrically reaming the anterior glenoid bone and implanting the nonaugmented/standard glenoid. To isolate only differences in reaming between the three glenoid styles, bone was not removed to prepare for each implant's pegs as each device has pegs of different lengths which could confound the results.

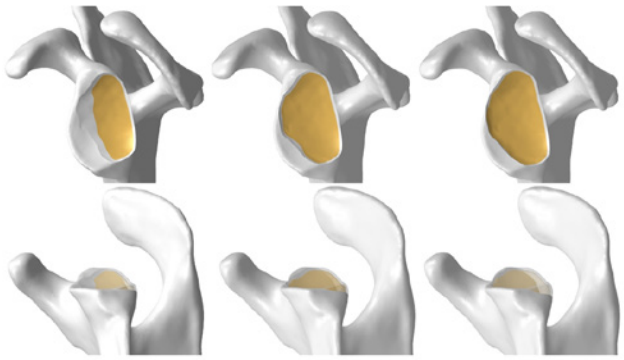

Figure 2: *Lateral and Inferior Views of the Small (left), Medium (middle), and Large (right) Posterior Glenoid Defects Created by Posteriorly Shifting the Humeral Head and Medially Translating it by 7.5, 9.0, and 10.5mm, into the Scapula, respectively. Note that in the each View the Cortical Bone is White and the Cancellous Bone Yellow. Also Note that in the Inferior Views, the Removed Bone is Transparent.*

Results

The cortical and cancellous bone removed to correct a small (Figure 3), medium (Figure 4), and large (Figure 5) posterior glenoid defect using the 3 different prosthesis designs are described in Tables 1-3, respectively. For the small defect, the 8° wedge-style posterior augment glenoid conserves 58% more cancellous bone, 26% more cortical bone, and 50% more bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming. For the medium defect, the 12° wedge-style posterior augment glenoid conserves 72% more cancellous bone, 55% more cortical bone, and 69% more bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming. For the large defect, the 16° wedge-style posterior augment glenoid conserves 48% more cancellous bone, 49% more cortical bone, and 48% more scapula bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming.

For the small defect, the 3mm step-style posterior augment glenoid conserves 25% more cancellous bone, 18% more cortical bone, and 23% more bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming. For the medium defect, the 5mm step-style posterior augment glenoid removes 4% more cancellous bone, conserves 7% more cortical bone, and conserves 2% more bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming. For the large defect, the 7mm step-style posterior augment glenoid removes 39% more cancellous bone, 22% more cortical bone, and 36% more scapula bone overall when correcting glenoid retroversion than using the nonaugmented/standard glenoid and eccentrically reaming.

Comparing posterior augment glenoid designs, for the small defect, the 8° wedge-style posterior augment glenoid conserves 35% more cancellous bone, 8% more cortical bone, and 27% more bone overall than that of the 3mm step-style posterior augment glenoid. For the medium defect, the 12° wedge-style posterior augment glenoid conserves 76% more cancellous bone, 48% more cortical bone, and 71% more bone overall than that of the 5mm step-style posterior augment glenoid. For the large defect, the 16° wedge-style posterior augment glenoid conserves 83% more cancellous bone, 69% more cortical bone, and 80% more bone overall than that of the 7mm step-style posterior augment glenoid.

Table 1: *Comparison of scapula bone removed to correct a small glenoid defect using 3 different prosthesis designs*

Bone Removed (cm3)	Cortical Bone Volume	Cancellous Bone Volume	Total Bone Volume
Small Defect Size	0.639	0.901	1.540
Nonaugmented glenoid and eccentric reaming	0.508	1.639	2.147
8° Wedge	0.393	0.901	1.295
3mm Step	0.426	1.278	1.704
Percent Difference (Wedge vs Step)	8.0%	34.6%	27.3%
Percent Difference (Wedge vs Standard)	25.5%	58.1%	49.5%
Percent Difference (Step vs Standard)	17.5%	24.7%	23.0%

Table 2: *Comparison of scapula bone removed to correct a medium glenoid defect using 3 different prosthesis designs*

Bone Removed (cm3)	Cortical Bone Volume	Cancellous Bone Volume	Total Bone Volume
Medium Defect Size	0.885	1.704	2.589
Nonaugmented glenoid and eccentric reaming	0.459	2.196	2.655
12° Wedge	0.262	1.032	1.295
5mm Step	0.426	2.294	2.720
Percent Difference (Wedge vs Step)	47.6%	75.9%	71.0%
Percent Difference (Wedge vs Standard)	54.5%	72.1%	68.9%
Percent Difference (Step vs Standard)	7.4%	4.4%	2.4%

Table 3: *Comparison of scapula bone removed to correct a large glenoid defect using 3 different prosthesis designs*

Bone Removed (cm3)	Cortical Bone Volume	Cancellous Bone Volume	Total Bone Volume
Large Defect Size	1.016	2.720	3.736
Nonaugmented glenoid and eccentric reaming	0.541	2.474	3.015
16° Wedge	0.328	1.524	1.852
7mm Step	0.672	3.671	4.343
Percent Difference (Wedge vs Step)	68.9%	82.6%	80.4%
Percent Difference (Wedge vs Standard)	49.1%	47.5%	47.8%
Percent Difference (Step vs Standard)	21.6%	38.9%	36.1%

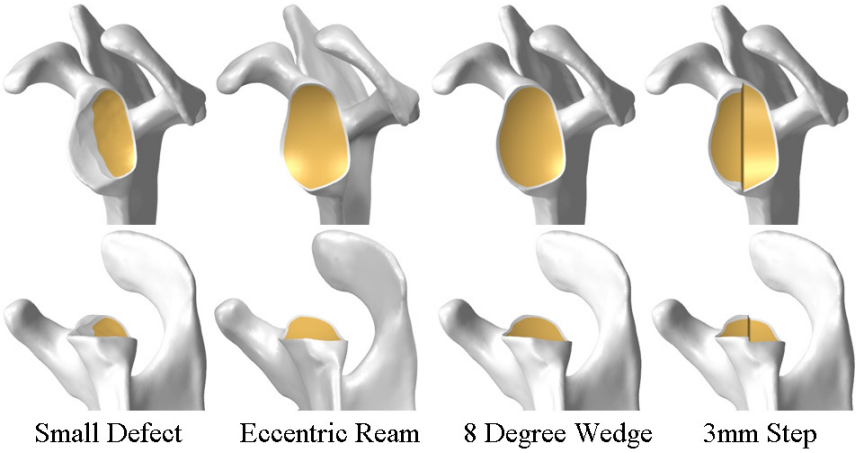

Figure 3: *Lateral and Inferior Views of the Glenoid Bone Remaining After Correction of Small Defect: Small Defect (left), Eccentric Ream (middle left), 8° Wedge (middle right), and 3mm Step (right)*

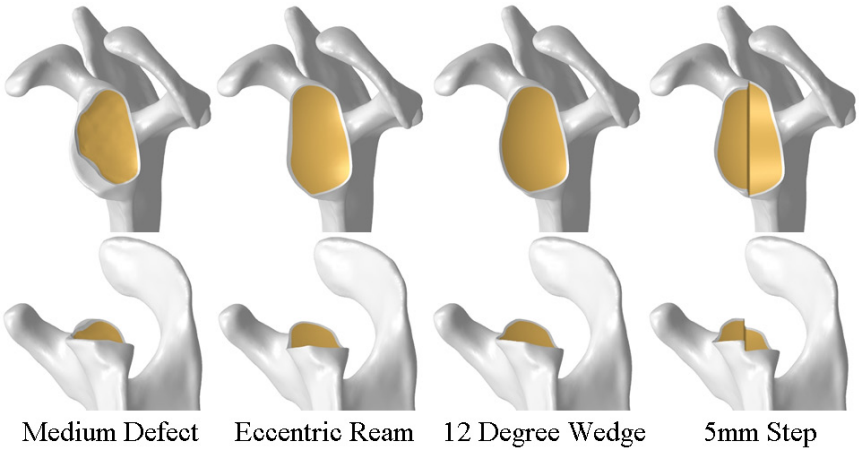

Figure 4: *Lateral and Inferior Views of the Glenoid Bone Remaining After Correction of Medium Defect: Medium Defect (left), Eccentric Ream (middle left), 12° Wedge (middle right), and 5mm Step (right)*

Discussion and Conclusions

The results of this study demonstrate that the wedge posterior augment glenoid design removed less cortical, cancellous, and total bone than the step posterior glenoid design to correct glenoid retroversion in each sized posterior glenoid defect. Both the wedge and step posterior augment glenoid designs can be used to conserve anterior glenoid bone when correcting glenoid retroversion in small and medium sized glenoid defects relative to that of the traditional method of eccentric reaming and using a nonaugmented glenoid implant. However for large glenoid defects, only the wedge posterior augment glenoid design conserved anterior glenoid bone relative to the traditional method, the step posterior augment glenoid design removed more scapula bone in large glenoid defects than the nonaugmented glenoid with eccentric reaming.

This computer analysis only quantified bone removed at the face of the glenoid; the more scapula bone was reamed away, the smaller the size of the glenoid face. No glenoid overhang was observed with the wedge-style posterior augment glenoids after correcting each size defect; whereas, glenoid overhang was observed with the step-style posterior augment glenoids after correcting both the medium and large glenoid defect. It should be noted

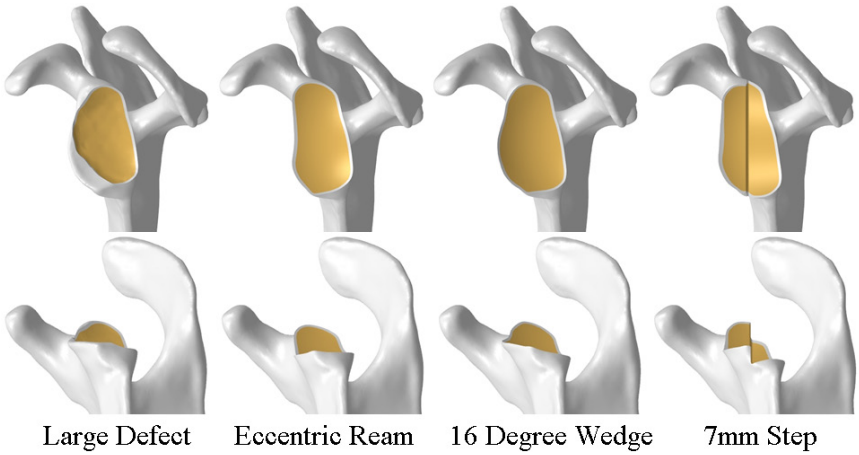

Figure 5: *Lateral and Inferior Views of the Glenoid Bone Remaining After Correction of Large Defect: Large Defect (left), Eccentric Ream (middle left), 16° Wedge (middle right), and 7mm Step (right)*

that bone removed by drilling out the glenoid pegs was not considered, as the Global Step-Tech posterior augment glenoid has longer pegs than Equinoxe posterior augment glenoid, the total bone actually removed by the Step-Tech is expected to be larger. The step posterior augment glenoid perforated the anterior scapula in the medium and large defect and the wedge posterior augment peg glenoid perforated the anterior scapula in the large defect. This study is limited because it only compared bone removal between the 3 glenoid designs; however, these results are clinically relevant because it provides the surgeon guidance on how to treat these patients and conserve the most glenoid bone for three different sizes of posteriorly worn glenoids when performing total shoulder arthroplasty. While the fixation of the 8, 12, and 16° cemented wedge-style posterior augment glenoid has been previously presented,[6] future work should evaluate the fixation differences associated with each design given these differences in bone volume removed during implantation.

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Kinematics and Biomechanics of Reverse Total Shoulder Arthroplasty

This report is a summary of the AAOS Orthopaedic Knowledge Unit Update: Shoulder and Elbow 4 book chapter on Reverse Shoulder Kinematics and Biomechanics.[1]

Glenohumeral Joint Anatomy

The glenohumeral joint is the most mobile joint in the human body and also (due to nonconforming articular curvature) inherently unstable. Its motion has been described as "spinning" (rotation only), "sliding" (translation only), and "rolling" (rotation + translation).[2-3] Glenohumeral joint stability is assisted throughout the range of motion by coordinated action of muscle contractions (and controlled by ligament and capsular tightening), which varies according to joint position and for different types of motion. As such, the muscles of the shoulder are required for both mobility and stability.[4]

The deltoid is the largest and most important muscle in the shoulder girdle. It is the primary mover in the shoulder, and generates forward elevation in the scapular plane. The deltoid consists of three distinct heads: 1) anterior (anterior acromion and clavicle), 2) middle (lateral margin of the acromion), and 3) the posterior deltoid (scapular spine); and accounts for approximately 20% of the mass of the shoulder muscles.[4] At low levels of abduction, the wrapping of the middle deltoid around the greater tuberosity of the humeral head generates a stabilizing compressive force; however, this compressive force is small relative to that generated by the rotator cuff.[6-7]

The rotator cuff muscles generate the torque necessary for rotation of the humerus about the glenoid fossa while also compressing the humeral head into the glenoid concavity.[8] The rotator cuff muscles are aligned around the proximal humerus for effective joint compression at all glenohumeral joint positions, allowing it to dynamically balance the joint; thereby, compensating for the lack of osseous constraint in the glenohumeral joint.[9-11] Specifically, the anatomic arrangement of the anterior (subscapularis) and posterior (infraspinatus and teres minor) rotator cuff muscles creates a transverse force couple that (approximately) centers the humeral head on the glenoid fossa in the anterior and posterior directions for all joint positions.[12-14]

Cuff Tear Arthropathy

Disruption of rotator cuff integrity, most commonly by a tear in the rotator cuff muscle tendons, can have devastating consequences on glenohumeral joint stability. As the rotator cuff fails to achieve concavity compression and balance the forces of the other muscles in the shoulder (primarily the deltoid), the humeral head tends to migrate superiorly and impinge with the undersurface of the acromion. This impingement can lead to further tearing of the rotator cuff and fatty infiltration of the cuff muscles which results in the onset of arthritic changes secondary to increased friction and a lack of nutrients supplied to the cartilage. Continued tearing propagates further compression and results in humeral head collapse, biceps tendon dislocation, and superior glenoid, acromion, and coracoid erosion.[15-16] Dr. Charles Neer coined the term cuff tear arthropathy (CTA) to describe this arthritic, eroded/collapsed condition of the glenohumeral joint following prolonged/progressive subacromial compression resulting from massive, full thickness rotator cuff tears.[15]

Reverse Shoulder Philosophy & Design History

The Reverse shoulder was first conceived in the early 1970's to treat patients suffering from CTA; this device was intended to relieve pain and prevent progressive acromial, coracoid, and glenoid erosion by resisting humeral head superior migration.[17] The reverse shoulder inverts the anatomic concavities making the glenoid articular component convex and the humeral articular component concave, creating a fixed fulcrum that prevents the humerus from migrating superiorly. As these articulations are conforming, the motion of a reverse shoulder is limited to only "spinning". Several historical reverse shoulder designs were developed, including: the Fenlin,[18] Reeves/Leeds Shoulder,[19-20] Kessel,[21-23] and the Neer-Averill.[24] Each prosthesis had a constrained and conforming articulation whose center of rotation was lateral to the glenoid fossa. These combined design features generated excessive torque on the glenoid, which compromised fixation and caused mechanical failure. As a result, these early designs were removed from the US market.[17,24]

In 1987, Dr. Paul Grammont introduced his reverse shoulder design, consisting of a 42mm convex glenoid component whose thickness was approximately 2/3 of a sphere and a concave humeral component whose depth was approximately 1/3 of a sphere. This early design was also associated with glenoid failure and in 1991 was redesigned for noncemented fixation with a fixed central peg to include divergent locking and compression screws.[17,27] Additionally, the thickness of the convex glenoid was decreased to 1/2 the diameter of the sphere so that the joint center of rotation would be placed medially on the glenoid fossa.[17,25-27]

The Grammont reverse shoulder medialized the joint center of rotation which minimized the torque on the glenoid bone-implant interface and increased the moment arm lengths of the abductor muscles. All contemporary reverse shoulder designs share this heritage. While some have suggested lateralizing the center of rotation (relative to the Grammont) to reduce the complication of scapular notching, the preferred mitigation strategy, as first recommended by Nyffeler et al., is to position the glenoid component along the inferior glenoid rim (with or without an inferior tilt).[28] Doing so, has the added benefit of inferiorly shifting the joint center of rotation which elongates the deltoid and improves its resting tone/tension.

Reverse Shoulder Kinematics

Muscles generate straight line forces that are converted to torques in proportion to their perpendicular distance between the joint center of rotation and the muscle's line of action.[4,29] This perpendicular distance is termed the muscle moment arm; thus, a 50% larger moment arm implies a 50% lower force required by a particular muscle to induce a given torque/motion. The location of the moment arm relative to the joint center of rotation determines the type of motion the muscle will create. In the shoulder, these motions are abduction/adduction (in the scapular plane and/or in the transverse plane), internal/external rotation (rotation of long axis of humerus), and flexion/extension (in the sagittal plane). Muscles can only induce force in tension, which can be generated two general ways, by either contraction (i.e. muscle shortening) or by stretching a muscle beyond its resting length (i.e. muscle lengthening). Thus, motion can be associated with muscle shortening, muscle lengthening, or no change in muscle length. No change in muscle length occurs when the line of action is coincident with the joint center of rotation.[29] Said another way, contraction of a muscle can either cause motion or stabilize motion depending upon the muscle's line of action relative to the center of rotation. Some muscles can function as agonists (which cause motion), antagonists (which stabilize motion), or in a biphasic manner functioning as both agonists and antagonists depending upon the specific joint position during a given range of motion.[4,29] The greater the muscle's moment arm, the greater capacity for that muscle to generate the torque required for motion and to support external loads. The trade off for a larger moment arm is that the muscle then requires a greater excursion (i.e. more muscle shortening to generate a given amount of motion). It should be recognized that a muscle's moment arm is only one component of a muscle's ability to generate torque, other factors include the muscle's physiologic cross sectional area, architecture, neural activity, and its length-tension relationship.[4]

The inversion of the anatomic concavities and the inferior-medial shift of the center of rotation associated with contemporary reverse shoulder designs dramatically alters the relationship of each shoulder muscle to its normal physiologic function. Medially shifting the center of rotation increases the length of the anterior, middle, and posterior deltoid abduction moment arms and lengthens the anterior, middle, and posterior deltoid allowing them to contribute more toward abduction.[6,30-33] These larger abductor moment arms enhance the capacity of the deltoid to elevate the arm in the scapular and coronal planes, compensating for the impaired function of the supraspinatus and the superior portions of the subscapularis and infraspinatus rotator cuff muscles which are typically involved in the pathology. Medially shifting the center of rotation also translates the humerus medially which increases the laxity of any remaining rotator cuff muscles and also leads to impingement of the humerus with the scapular neck at low elevation (i.e. scapular notching).[34,35]

Inferiorly shifting the center of rotation with reverse shoulder arthroplasty elongates the deltoid relative to a normal shoulder. Deltoid elongation between 10 and 20% improves its resting tone/tension and has been suggested to increase its strength and improve the overall stability of the joint.[6,30] De Wilde et al. reported that the Grammont and DJO reverse shoulders elongated the deltoid when the arm was at 0° abduction by 16.4% and 13.0%, respectively (relative to the normal shoulder).[6] Similarly, Jobin et al. reported that the average deltoid elongation of three different reverse shoulders was 17.0% when the arm was at 0° abduction.[30] Ladermann et al. taught that optimal tension is based upon humeral lengthening, where arm lengthening greater than 2.5cm was recommended as the surgical objective as these patients were associated with larger active elevation and shortening the humerus was associated with an increased risk of dislocation.[36,37] However, increased deltoid elongation modifies the normal deltoid contour, which decreases its wrapping angle around the greater tuberosity (which reduces stability), and also creates cosmetic concerns.[6,7,34] Increased humeral lengthening could also lead to acromial stress fractures and brachial plexopathy.[38,39] Restoring the lateral position of the humeral tuberosities is important to tension the remaining rotator cuff muscles in a more natural physiologic manner and offers the potential to better restore rotational strength.[33,34]

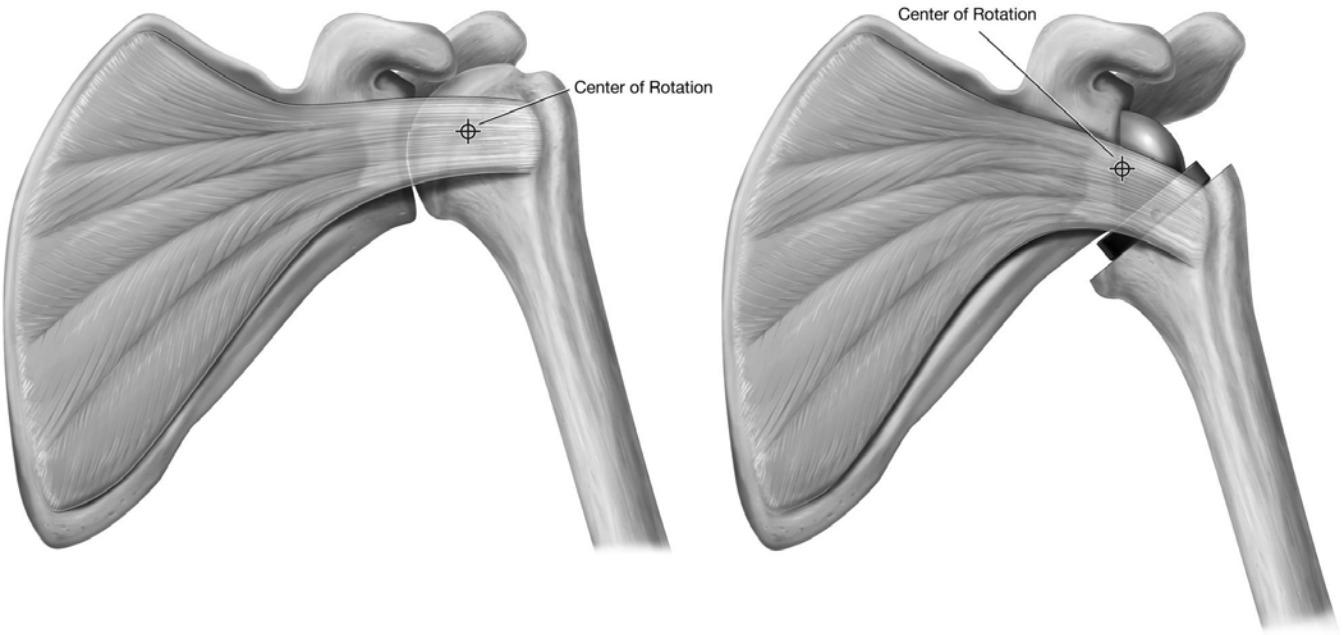

Figure 1: *Impact of Reverse Shoulder Arthroplasty on Subscapularis Tendon Positioning Relative to the Center of Rotation with a Normal Shoulder*

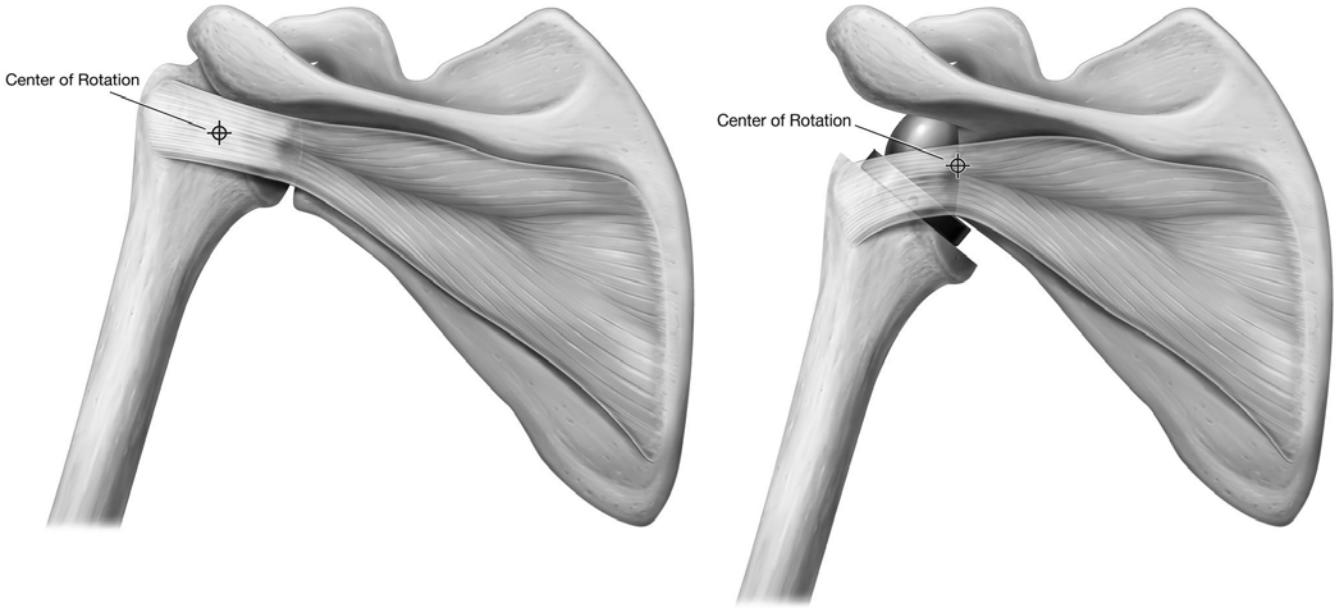

Figure 2: *Impact of Reverse Shoulder Arthroplasty on Infraspinatus Tendon Positioning Relative to the Center of Rotation with a Normal Shoulder*

While over-tensioning these muscles may offer the possibility of improved resting tone/tension, it may also make it more difficult to repair following tenotomy (in the case of the subscapularis). Inferiorly shifting the center of rotation also changes when the subscapularis (Figure 1) and infraspinatus (Figure 2) muscles convert from behaving as adductors at low elevation to behaving as abductors at high elevation; in some situations, these muscles may lose their biphasic function altogether which has important implications on both mobility and stability.[31,34]

The Grammont reverse shoulder is effective at restoring active abduction and forward flexion but is less effective at restoring active internal and external rotation in patients with a deficient rotator cuff and a functioning deltoid.[40-42] Lateralizing the joint center of rotation relative to the Grammont design has been proposed as a method to improve active internal and external rotation, strength, and stability.[34] Lateralizing the joint center of rotation lateralizes the humerus, tensions the remaining rotator cuff muscles, and minimizes impingement of the humeral

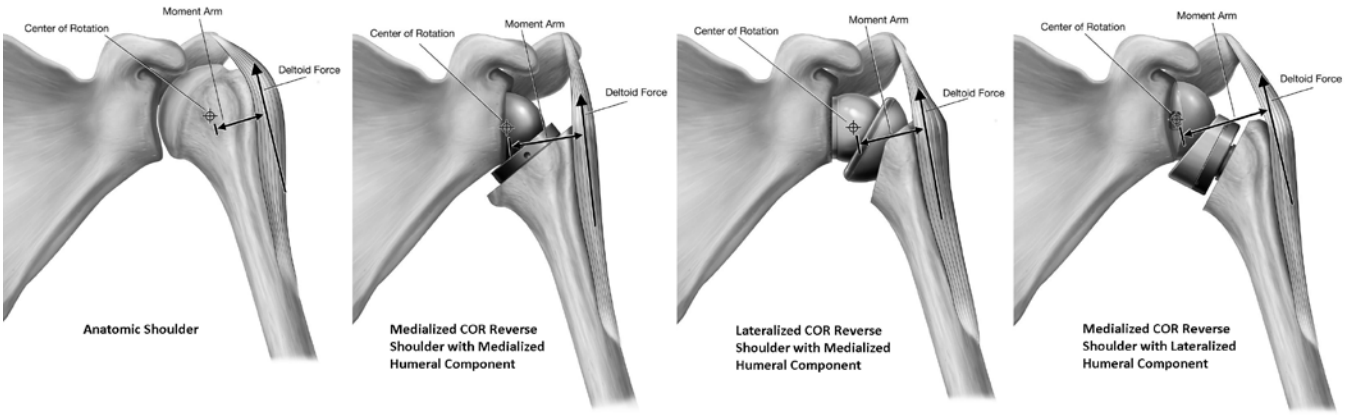

Figure 3: *Illustrations of center of rotation, deltoid moment arm, direction of deltoid elongation, and humeral position with a normal shoulder and three different reverse shoulder prosthesis positioned along the inferior glenoid rim at 0° tilt. All drawings show abduction to 15° in the scapular plane. Left: The normal shoulder. Middle-Left: The 36-mm Grammont prosthesis (Depuy, Inc). Middle-Right: The 32-mm RSP prosthesis (DJO, Inc). Right: The 38-mm Equinoxe® prosthesis (Exactech, Gainesville, FL).*

component along the inferior scapular neck. Lateralizing the joint center of rotation also increases the torque on the glenoid fixation surface and decreases the lengths of the deltoid abductor moment arms.[6,38] Because the deltoid abductor moment arms are decreased as the center of rotation is lateralized, the deltoid becomes less effective as an abductor and requires a greater force to elevate the arm in both the scapular and coronal planes.[38] These elevated loads and torques can have negative implications on patient rehab, muscle fatigue, stress fractures, and prosthesis fixation. It should be recognized that the humerus can be lateralized without lateralizing the joint center of rotation. Doing so has the advantage of restoring the anatomic rotator cuff muscle length/tension while maintaining Grammont's abductor moment arm lengths and minimizing the torque on the glenoid-bone interface. Roche et al. first demonstrated that the humerus can be lateralized to place the tuberosities in a more anatomic position while minimizing humeral liner impingement with the inferior scapular neck.[43,44] This can be accomplished by decreasing the humeral neck angle from the Grammont humeral neck angle of 155°, proportionally increasing the Grammont glenosphere diameter and thickness, decreasing the humeral liner constraint, and/or by increasing the medial offset of the humeral liner/humeral stem. Lemieux et al. demonstrated that increasing the medial offset of the humeral stem in total shoulder arthroplasty increased the middle deltoid moment arm and also increased the middle deltoid wrapping angle about the greater tuberosity which helps to stabilize the joint by compressing the humeral head into the glenoid fossa.[7] To illustrate the impact of differing reverse shoulder design philosophies on humeral positioning, Figure 3 illustrates the impact of reverse shoulder arthroplasty design on humeral positioning by depicting a normal shoulder relative to three commercially available prostheses: 1) Grammont-style (medialized COR and a medialized humeral component; Depuy, Inc), 2) RSP-style (lateralized COR and a medialized humeral component; DJO, Inc), and 3) Equinoxe-style (medialized COR and a lateralized humeral component; Exactech, Inc). To enable a direct visual comparison of center of rotation, deltoid moment arm, direction of deltoid elongation, and humeral position, each prosthesis is shown positioned along the inferior glenoid rim at 0° tilt when abducted in the scapular plane at the same angle.

While increased stability may be achieved by the compensatory action of muscles with smaller moment arms or by the combined force generated by 2 opposing muscle lines of action positioned about a joint, each however, increases the overall joint reaction force.[45] For these reasons, repair of the subscapularis/lesser tuberosity in reverse shoulder arthroplasty is controversial. As the subscapularis functions as an adductor, its line of action opposes the deltoid at low and mid humeral elevation in the scapular and coronal planes; this opposing force requires the deltoid to generate a larger force to achieve a given motion.[31,33,46] These opposing forces increase joint stability and perhaps also counteract the deltoid's shear force on the glenoid component. This question of subscapularis/lesser tuberosity repair is even more relevant to patients with weak external rotators. Using a shoulder controller, Onstott et al. demonstrated that releasing the subscapularis requires less force to be generated by the deltoid and the posterior rotator cuff in abduction.[46] If the subscapularis/lesser tuberosity is repaired, the joint reaction force increased 426%, the required deltoid force increased 132% (at 15°), and the required posterior rotator cuff force increased 460% (at 15°).[46] These increased forces are problematic in this patient population as the posterior rotator cuff is often compromised (either by tear or fatty infiltration) and may not be able to support these elevated loads.[46] Some

have recommended repairing the subscapularis/lesser tuberosity whenever possible as this has been associated with a lower dislocation rate with the Grammont reverse shoulder.[47] All of the aforementioned factors should be accounted for when considering such recommendations as they may not be universally applicable to all reverse shoulder designs and/or to all clinical situations.

Loss of external rotation (and excessive internal rotation) impairs the patient's ability to maintain their arm in neutral rotation as the arm is elevated (e.g. positive horn blower's sign), preventing numerous activities of daily living including: shaking of hands, drinking/eating, and washing of hair.[40-42] Given the natural predominance of internal rotator muscles in a normally functioning shoulder: four internal rotators (subscapularis, teres major, pectoralis major, and latissimus dorsi) vs. two external rotators (infraspinatus and teres minor), external rotation deficiency is more debilitating to a patient's activities of daily living than internal rotation deficiency (particularly when the arm is elevated).[40,41] Muscle transfers are often recommended in reverse shoulder patients with external rotation deficiency because the posterior deltoid alone is insufficient to restore active external rotation, even with lateralized reverse shoulder designs. In general, internal rotation muscles (e.g. muscles that attach to the anterior side of the humerus) are transferred across the joint center of rotation to the posterior side of the humerus where their contraction now causes external rotation. The latissimus dorsi is the most common muscle transferred in reverse shoulder arthroplasty, it is detached from the anterior shaft of the humerus and reattached to the greater tuberosity.[48] Another common muscle transfer is a modification of L'Episcopo method[40,41] in which both the latissimus dorsi and the teres major are transferred to the greater tuberosity. While muscle transfers have been demonstrated to successfully restore active external rotation, they should not be performed if the teres minor is functional.[40,41] Additionally, it should be recognized that such procedures limit active internal rotation and further alter the relationship of each shoulder muscle to its normal physiologic function.

Reverse shoulder kinematics are also altered by scapular morphology. Abnormal glenoid wear patterns and strategies used to prepare the glenoid and position the prosthesis (e.g. 10 to 15° inferior tilt) to minimize the complication of scapular notching can impact the joint center of rotation, muscle moment arms, muscle lengths, and muscle lines of action. As the prosthesis is further medialized and inferiorly tilted, the middle deltoid wrapping around the greater tuberosity is eliminated. Additionally, Norris et al. demonstrated that with a sufficient amount of medial glenoid wear, the deltoid can actually generate a distraction force that can result in instability.[49]

Conclusions

The reverse shoulder inverts the anatomic concavities to restore stability to the unstable shoulder while the inferior-medial shift in the joint center of rotation (relative to a normal shoulder) lengthens the abductor moment arms and elongates the deltoid to facilitate these improvements in function and mobility. Reverse shoulder kinematics can be impacted by prosthesis design parameters, prosthesis positioning on the scapula, and abnormal scapular morphology/glenoid wear patterns. Future work should seek to optimize these kinematic parameters in a method that increases the overall function of the reverse shoulder and facilitates a patient's ability to conduct their activities of daily living.

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Biomechanical Analysis of Three Commercially Available Reverse Shoulder Designs in a Normal and Medially Eroded Scapula

This report is a summary of the results of two different computer analyses presented at the 2013 Orthopaedic Research Society.[1,2]

Introduction

The reverse shoulder inverts the anatomic concavities to restore stability to the unstable shoulder and inferiorly and medially shifts the center of rotation to lengthen the abductor moment arms and elongate the deltoid to facilitate improvements in function and mobility. The magnitude of change in the center of rotation and position of the humerus has important consequences on muscle tensioning, range of motion, and stability.[3-8] Inferiorly shifting the center of rotation elongates the deltoid relative to a normal shoulder which improves its resting tone/tension, increases its strength, and improves the overall stability of the joint.[3] However, increased deltoid elongation modifies the normal deltoid contour, decreases its wrapping angle around the greater tuberosity (which reduces stability), and creates cosmetic concerns.[3-5,9] Increased humeral lengthening may also lead to acromial stress fractures and brachial plexopathy.[10-12] Medially shifting the center of rotation also translates the humerus medially which increases the laxity of any remaining rotator cuff muscles and leads to impingement of the humerus with the scapular neck at low elevation.[3,5,7-8] Shortening the rotator cuff muscles relative to their resting length, has important consequences on function and stability, impairing the rotator cuffs ability to generate the torque necessary for rotation of the humerus about the glenoid fossa while also compressing the humeral head into the glenoid concavity to dynamically balance the joint.[13-15]

Numerous design concepts and surgical implantation methods have been attempted to solve these challenges and have led to the differentiation of reverse shoulder designs available in the marketplace. How each design translates the center of rotation and positions the humeral tuberosities can be characterized according to its specific humeral neck angle, glenoid offset, and humeral offset. The 36x18mm Grammont (Depuy Orthopaedics, Inc; Warsaw, IN, USA) has a humeral neck angle of 155°, a center of rotation 0mm lateral to the glenoid fossa, and a humeral stem/liner medial offset of 9.8mm. The 32x26mm RSP® (DJO Surgical; Austin, TX, USA) has a humeral neck angle of 135°, a center of rotation 10mm lateral to the fossa, and a humeral stem/liner medial offset of 10.9mm. The 38x21mm Equinoxe® (Exactech, Inc; Gainesville, FL, USA) has a humeral neck angle of 145°, a center of rotation 2mm lateral to the fossa, and a humeral stem/liner medial offset of 20.8mm.

Changing prosthesis design to restore the lateral position of the humeral tuberosities could re-tension the remaining rotator cuff muscles to their natural length and offers the potential to better restore rotational strength. The anatomic arrangement of the anterior (subscapularis) and posterior (infraspinatus and teres minor) rotator cuff muscles creates a transverse force couple that centers the humeral head on the glenoid fossa in the anterior and posterior directions for all joint positions. Modifying the surgical implantation method by increasing or decreasing humeral retroversion in reverse shoulder arthroplasty may be able to selectively bias tensioning of the anterior or posterior rotator cuff depending upon patient need. The purpose of this computer analysis is to quantify the biomechanical impact of varying design parameters and varying the Grammont humeral retroversion on muscle elongation and deltoid wrapping as each reverse shoulder is abducted from 0 to 80° in both a normal and medially eroded scapula.

Methods

A 3-D computer model was developed in Unigraphics (Siemens PLM; Plano, TX, USA) to simulate abduction in the scapular plane for the normal shoulder and for three reverse shoulder designs commercially-available in the United States since 2007 (or before). Each reverse shoulder was geometrically modeled and implanted in a 3-D digitized scapula and humerus; a 3-D digital clavicle and ribcage were also assembled (Pacific Research Laboratories, Inc; Vashon Island, WA, USA). The digital humerus and scapula were assembled to simulate a normal shoulder, functioning as the control in this analysis; the humeral head was centered on the glenoid and offset by 4mm from the center of the glenoid to account for the thickness of the cartilage and labrum. The computer model simulated seven muscles as three lines

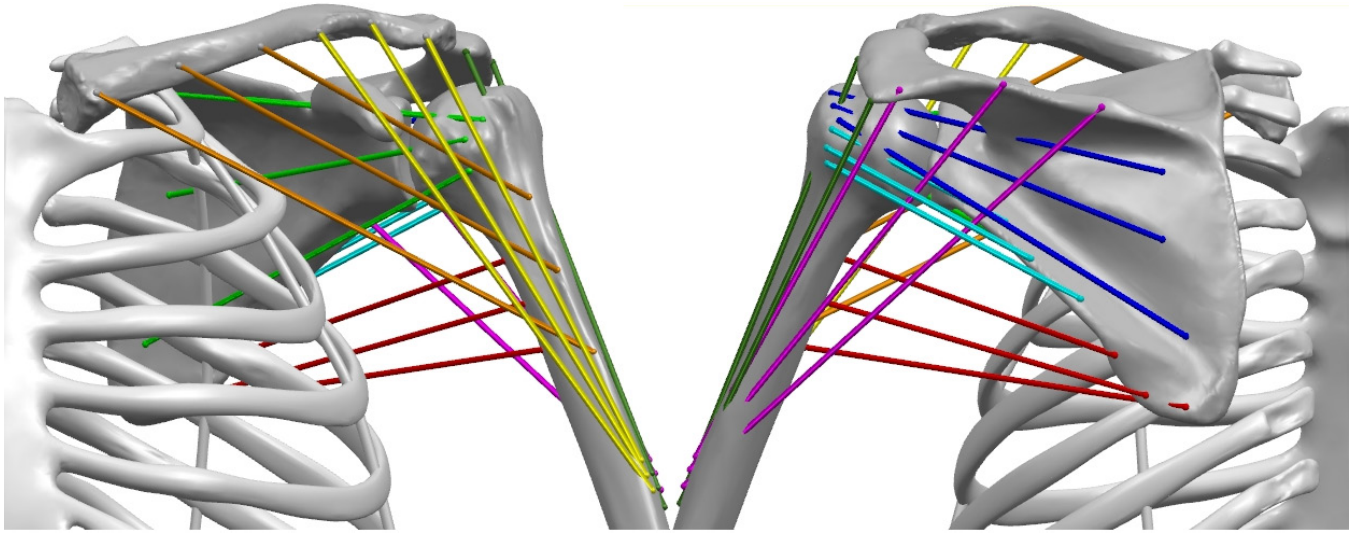
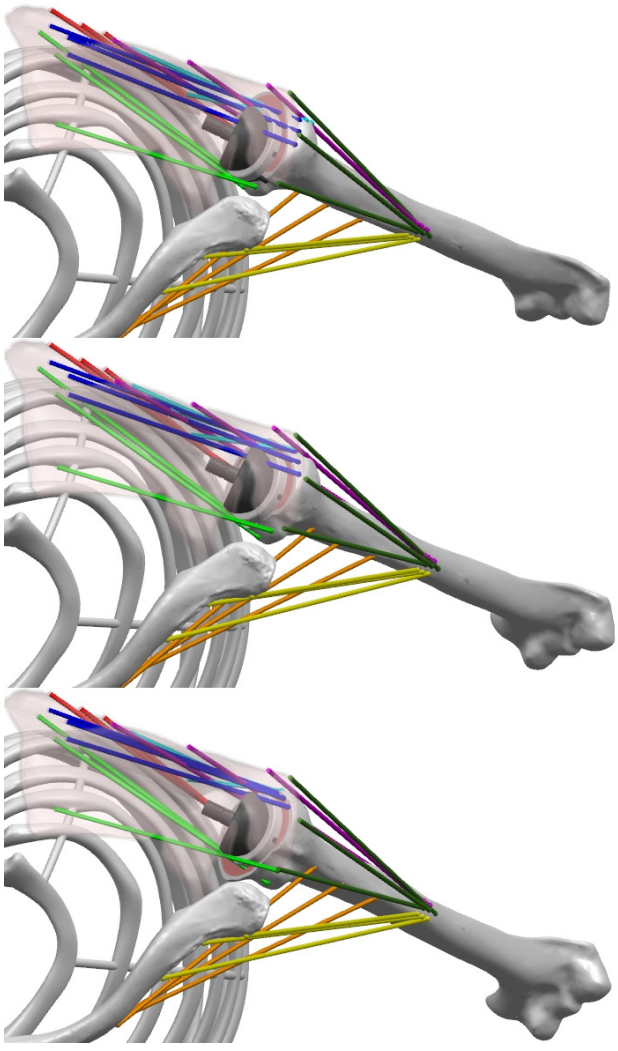


Figure 1: *Computer Model of 7 Muscles Simulated as 3 Lines from Origin to Insertion, Anterior (left) and Posterior (right) Views of the Normal Shoulder at 25° Abduction in the Scapular Plane. Note that the Clavicular Portion of the Pectoralis Major is Depicted (Orange) but was not Used in this Analysis.*



from origin to insertion: anterior deltoid (yellow), middle deltoid (dark green), posterior deltoid (magenta), subscapularis (light green), infraspinatus (dark blue), teres major (red), and teres minor (cyan). (Figure 1) The origin and insertion of each muscle were maintained for each prosthesis and for the normal shoulder.

To characterize the biomechanical differences between reverse shoulder designs and their impact on each muscle, each prosthesis was implanted according to the manufacturer's recommendations so the glenoid plate aligns with the inferior glenoid rim; each humeral component was oriented at 20° retroversion. To characterize the biomechanical impact of changing humeral retroversion on each muscle, the Grammont reverse shoulder humeral component was successively implanted at 0, 20, and 40° retroversion. (Figure 2) To characterize the biomechanical impact on each muscle of using each devices in an abnormal scapula, each reverse shoulder was implanted along the inferior glenoid rim of a normal scapula and a scapula with a 10mm medially eroded glenoid. After assembling each in the normal and medially eroded scapula, the humeral component was abducted from 0 to 80° in the scapular plane relative to a fixed scapula. (Figure 3) Muscle lengths were measured as the average length of the three

Figure 2: *Representative Computer Model Image of Scapular View of the 36mm Grammont 0° humeral retroversion (top), 36mm Grammont 20° humeral retroversion (middle), and 36mm Grammont 40° humeral retroversion (bottom) Reverse Shoulders at 30° Abduction, Scapula made Transparent to Permit Visualization of the Glenoid Component*

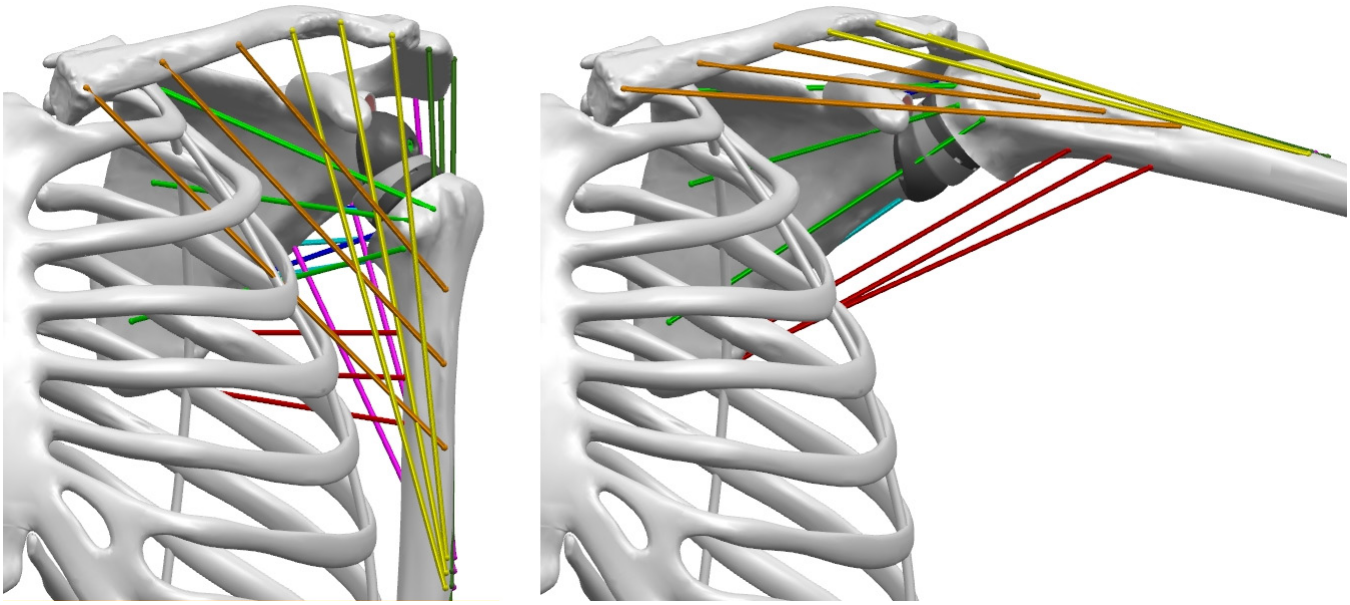


Figure 3: ***Simulated Abduction from 0 (left) to 80° (right) in the Scapular Plane relative to Fixed Scapula; representative image of the 38mm Equinoxe***

lines simulating each muscle at each degree of abduction; each average muscle length, for each design, at each angle of abduction was compared at the corresponding arm position for the normal shoulder. The angle of abduction in which the middle deltoid stops wrapping around the greater tuberosity was also quantified for the normal shoulder and each reverse shoulder design in both the normal and medially eroded scapulas.

Results

Each reverse shoulder design was associated with an inferior and medial shift in the center of rotation, this change in center of rotation resulted in a medial shift in humeral tuberosity position which decreased the middle deltoid wrapping angle relative to that of the normal shoulder. The angle of abduction in which the middle deltoid stops wrapping greater tuberosity for the normal shoulder and each reverse shoulder design in both the normal and medially eroded scapula is presented in Table 1. As depicted in Figure 4, the Grammont design shifted the humerus the most medial and resulted in the largest decrease the deltoid wrapping angle, the RSP design shifted the humerus the second most medial and resulted in the second largest decrease, and the Equinoxe design kept the humerus the most lateral and resulted in the smallest decrease in the deltoid wrapping angle. When each design was implanted in a medially eroded scapula, the joint line was shifted medially by the corresponding amount and the deltoid wrapping angle was further decreased for each design according the same trends.

Table 1: *Medial/Lateral Position of the Humerus and its Impact on Deltoid Wrapping*

	Lateral Distance from Coracoid to Greater Tuberosity with Humerus Abducted at 0°	Angle of Abduction when Middle Deltoid Stops Wrapping Greater Tuberosity
Normal Shoulder	56.2 mm	48°
36mm Grammont, 0° retro	36.3 mm	14°
36mm Grammont, 20° retro	34.7 mm	8°
36mm Grammont, 40° retro	32.6 mm	7°
36mm Grammont, 20° retro, 10mm wear	25.6 mm	-1°
32mm RSP, 20° retro	44.5 mm	28°
32mm RSP, 20° retro, 10mm wear	35.4 mm	12°
38mm Equinoxe, 20° retro	47.1 mm	40°
38mm Equinoxe, 20° retro, 10mm wear	38.0 mm	18°

Each reverse shoulder design elongated the three heads of the deltoid, shortened the internal rotators (subscapularis and teres major), and shortened the external rotators (infraspinatus and teres minor). The average length of each muscle (relative to the normal shoulder) for each reverse shoulder design when implanted in a normal and medially eroded scapula and abducted in the scapular plane from 0 to 80° and from 0 to 20° is presented in Tables 2 and 3, respectively. The Equinoxe reverse shoulder elongates the anterior, middle, and posterior deltoid the most, followed by the Grammont design, and the RSP design elongated each head of

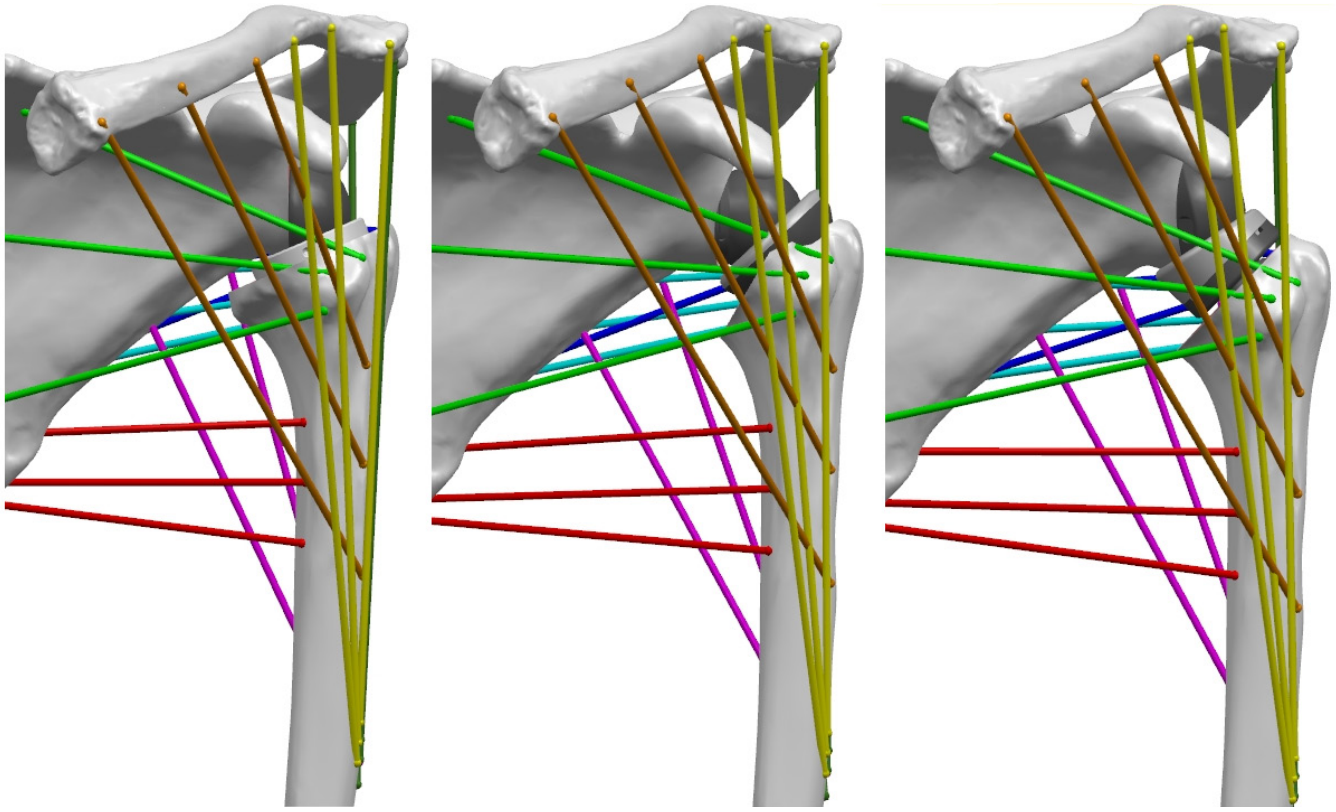


Figure 4: *Scapular View of the Grammont (left), RSP (middle), and Equinoxe (right) Reverse Shoulders at 0° Abduction, Rib Cage Removed to Permit Visualization of Humeral Position*

the deltoid the least, though all elongated relative to the normal shoulder by more than 10% and all less than 20%. The Grammont reverse shoulder medialized the humerus the most and was associated with the most shortening, the RSP design medialized the humerus the second most and was associated with the second most shortening, the Exactech design kept the humerus the most lateral and was associated with the least shortening of both the internal and external rotators. When each design was implanted in a medially eroded scapula, the joint line was shifted medially by the corresponding amount and the internal and external rotators were further shortened for each design according the same trends. Comparing Table 2 and 3 demonstrates the internal and external rotators were shortened most at low levels of abduction for all designs.

Decreasing humeral retroversion from 20 to 0° with the Grammont reverse shoulder changed the position of the humeral tuberosities, increased the laxity of the anterior shoulder muscles (anterior deltoid, subscapularis, and teres major), and better tensioned the posterior shoulder muscles (posterior deltoid, infraspinatus, and teres minor), though both the anterior and posterior muscles were shortened relative to their normal length. Conversely, increasing humeral retroversion from 20 to 40° with the Grammont reverse shoulder changed the position of the humeral tuberosities, better tensioned the anterior shoulder muscles (anterior deltoid, subscapularis, and teres major), and increased the laxity of the posterior shoulder muscles (posterior deltoid, infraspinatus, and teres minor), though both the anterior and posterior muscles were shortened relative to their normal length. Comparing Table 2 and 3 demonstrates that this observed asymmetric tensioning with varying humeral retroversion was more pronounced at low levels of abduction.

Discussion

The results of this study demonstrate that depending upon the specific combination of glenoid offset, humeral offset, and humeral neck angle, each reverse shoulder design medialized the humerus, decreased the deltoid wrapping angle, elongated the three heads of the deltoid, and shortened the internal and external rotator muscles. The Grammont design elongated the deltoid in the inferior direction and positioned the humerus the most medial, resulting in the largest decrease in the wrapping angle and the most shortening of the internal and external rotators due to its most medialized center of rotation, most vertical humeral neck angle, and smallest humeral stem/liner offset. The RSP design elongated the deltoid in the inferior/lateral direction and

positioned the humerus the second most medial, resulting in the second largest decrease in the wrapping angle and the second most shortening of the inferior and external rotators due to its most lateralized center of rotation, smallest humeral neck angle, and second smallest humeral stem/liner offset. Finally, the Equinoxe design elongated the deltoid in the inferior/lateral direction and positioned the humerus the most lateral, resulting in the smallest decrease in the wrapping angle and the least shortening of the internal and external rotators due to its second most medialized glenoid center of rotation, second most vertical humeral neck angle, and largest humeral stem/liner offset.

Changing humeral retroversion with the Grammont reverse shoulder asymmetrically tensioned the anterior (anterior deltoid, subscapularis, and teres major) and posterior (posterior deltoid, infraspinatus, and teres minor) shoulder muscles; however, regardless of humeral retroversion, the anterior and posterior shoulder muscles were shortened relative to their normal length. Surgeons using the Grammont reverse shoulder should implant the humeral component in less humeral retroversion if the patient requires better tensioning of the posterior shoulder muscles and should implant the humeral component in more humeral retroversion if the patient requires better tensioning of the anterior shoulder muscles.

Using reverse shoulders in a medially eroded scapula further shortens the anterior and posterior rotator cuff muscles and significantly decreases the deltoid wrapping angle by 9 to 22°, depending upon reverse shoulder design. In these situations, surgeons should use thicker glenospheres, augmented glenoid baseplates, or bone graft the glenoid to lateralize the joint line and improve the muscle tensioning to achieve stability and potentially restore rotational strength. This study was limited by its evaluation of only one plane of motion; future work should evaluate the impact of these parameters for multiple types of motion.

Conclusions

The reverse shoulder is biomechanically different than the normal shoulder. Minor differences in prosthesis design (<10mm of glenoid and humeral offset and 10° of humeral neck angle) dramatically impacted humeral position, deltoid wrapping angle, deltoid elongation, and shortening of the anterior and posterior rotator cuff muscles relative to the normal shoulder. These observations related to decreased deltoid wrapping and rotator cuff shortening with reverse shoulder arthroplasty may explain the mechanism for instability and lack of internal and external rotation associated with some prosthesis designs having a medialized humerus.

Table 2: *Average Muscle Length Relative to Normal Shoulder as Each Reverse Shoulder is Abducted in the Scapular Plane from 0 to 80°. Color coding denotes muscle shortening >10% (Yellow), > 20% (Orange), and >30% (Red)*

	Ant. Deltoid	Mid Deltoid	Post. Deltoid	Subscapularis	Infraspinatus	Teres Major	Teres Minor
36mm Grammont, 0° retro	4.5%	4.9%	1.9%	-14.8%	-9.5%	-10.1%	-13.5%
36mm Grammont, 20° retro	4.7%	4.8%	1.7%	-11.2%	-12.8%	-11.0%	-20.5%
36mm Grammont, 40° retro	5.1%	4.8%	1.5%	-7.6%	-16.6%	-9.0%	-27.7%
36mm Grammont, 20° retro, 10mm wear	2.3%	2.0%	-1.4%	-17.7%	-19.2%	-17.2%	-29.8%
32mm RSP, 20° retro	6.2%	7.0%	4.6%	-3.9%	-5.6%	-4.5%	-9.7%
32mm RSP, 20° retro, 10mm wear	3.6%	3.9%	1.3%	-10.5%	-12.0%	-10.1%	-19.1%
38mm Equinoxe, 20° retro	7.3%	8.2%	6.3%	0.0%	-1.6%	-1.1%	-3.5%
38mm Equinoxe, 20° retro, 10mm wear	4.6%	5.1%	2.9%	-6.6%	-8.0%	-6.7%	-12.9%

Table 3: *Average Muscle Length Relative to Normal Shoulder as Each Reverse Shoulder is Abducted in the Scapular Plane from 0 to 20°. Color coding denotes muscle shortening >10% (Yellow), >20% (Orange), and >30% (Red)*

	Ant. Deltoid	Mid Deltoid	Post. Deltoid	Subscapularis	Infraspinatus	Teres Major	Teres Minor
36mm Grammont, 0° retro	10.9%	12.8%	7.5%	-20.1%	-14.2%	-19.5%	-21.9%
36mm Grammont, 20° retro	11.1%	12.8%	7.3%	-15.5%	-17.8%	-17.5%	-29.7%
36mm Grammont, 40° retro	11.5%	12.7%	7.0%	-11.1%	-22.1%	-14.8%	-37.8%
36mm Grammont, 20° retro, 10mm wear	10.6%	12.3%	5.5%	-21.9%	-24.0%	-24.2%	-39.4%
32mm RSP, 20° retro	10.9%	12.6%	8.9%	-8.0%	-10.6%	-9.4%	-18.2%
32mm RSP, 20° retro, 10mm wear	10.0%	11.8%	6.8%	-14.5%	-16.8%	-16.1%	-28.1%
38mm Equinoxe, 20° retro	13.5%	15.5%	11.9%	-4.9%	-8.3%	-6.5%	-14.5%
38mm Equinoxe, 20° retro, 10mm wear	12.6%	14.6%	9.7%	-11.4%	-14.6%	-13.2%	-24.5%

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Impact of Scapular Notching on Reverse Shoulder Glenoid Fixation

This memo reports on the results of a computer analysis presented at the 2013 Orthopaedic Research Society.[1]

Introduction

Scapular notching is commonly reported in ~70% of patients with reverse shoulder designs having a medialized center of rotation (>20% of which are large notches of grade 3 or 4). [2-9] Scapular notching has also been demonstrated to be progressive,[5,10] correlate with the presence of radiolucent lines,[2,5,10] and negatively affect clinical outcomes.[6-8] The impact of scapular notching on glenoid fixation is unknown and particularly worrisome given that glenoid loosening was the primary failure mode of historical (e.g. Grammont predecessor) reverse shoulder prostheses whose center of rotation was laterally offset relative to the glenoid fossa.[11] Short and mid-term clinical outcome studies have reported aseptic glenoid loosening rates between 0 and 12% with modern reverse shoulder arthroplasty designs,[7,9,12,13] with an average rate of 5%.[14] A reverse shoulder test method was previously presented to assess initial glenoid fixation for the purpose of analyzing performance.[15] The purpose of this study is to quantify glenoid fixation in a composite scapulae with and without a scapular notch and evaluate the null hypothesis that a grade 4 scapular notch has no impact on the fixation of a reverse shoulder glenoid implant.

Table 1: *Comparison of scapula bone removed to correct a small glenoid defect using 3 different prosthesis designs*

Study	Sample Size; Avg Follow-up	Scapular Notching Rate	Notches > Grade 2
Boileau et al.[2]	n = 45; 40 months	68%	11%
Karelse et al.[3]	n = 27; 43 months	59%	26%
Kempton et al.[4]	Group 1: n =43; 30 months (no glenoid tilt) Group 2: n = 28; 24 months (inferior glenoid tilt)	Group 1: 77% Group 2: 61%	Group 1: 23% Group 2: 4%
Levigne et al.[5]	n = 461; 51 months	68%	23%
Simovitch et al.[6]	n = 77; 44 months	44%	18%
Sirveaux et al.[7]	n = 80; 44 months	64%	17%
Stechel et al.[8]	n = 59; 48 months	87%	5%
Werner et al.[9]	n = 48; 38 months	96%	46%
Weighted Avg. Scapular Notch Rate	**46.0 months**	**68.2%**	**20.9%**

Methods

This reverse shoulder glenoid loosening test was conducted in two phases.[15] The first phase is the displacement test. It measures the fixation of the reverse shoulder glenoid baseplate in the composite scapulae before and after the application of 10K cycles of dynamic loading for 55° at 0.5 Hz. In the displacement test, the axial test machine (Instron Corp. Norwood, Mass. Resolution of 1 micron) and 3 digital indicators (Mitutoyo, Japan. Resolution of 1 micron) quantified the glenoid baseplate displacement relative to the composite scapula as a compressive (433 N) and shear (357 N) load is applied. The compressive axial load is applied perpendicular to the reverse glenoid baseplate and the shear load is applied parallel to the face of the glenoid baseplate along its superior/inferior axis. (Figure 1) The second phase is the cyclic test. The cyclic test simulates the primary motion of reverse shoulder arthroplasty; that is, the abduction motion generated by the deltoid.[16-18] The humeral liner and glenosphere/glenoid baseplate/composite scapulae are positioned in the biaxial testing apparatus and aligned along the superior/inferior axis of the glenoid baseplate. A 750N axial load is constantly applied through the center of the humeral liner as the glenosphere/glenoid baseplate/composite scapulae are rotated about the humeral component with a stepper motor to create a sinusoidal angular displacement profile encompassing an arc of 55° at 0.5 Hz for 10,000 cycles. (Figure 2) The components are cooled with a continuous jet of air with no lubrication.

The test method assessed the initial glenoid fixation associated with the 38mm Equinoxe® reverse shoulder (Exactech, Inc; Gainesville, FL) in a fourth generation composite/dual density scapula (Pacific Research, Inc; Vashon WA) with a 1.63g/cm^3 "cortical" shell and a 0.27 g/cm^3 "cancellous" interior structure. A custom cutting jig was designed to create a Nerot-Sirveaux Grade 4 scapular notch (n = 7) in the dual density composite scapulae. The scapular notch was cut to approximate the profile of the humeral liner as it articulates around the center of rotation of the 38mm glenosphere. (Figures 3 and 4) The fixation of these notched scapulae were compared to that of composite scapulae without a scapular notch (i.e. Nerot-Sirveaux Grade 0; n = 7)

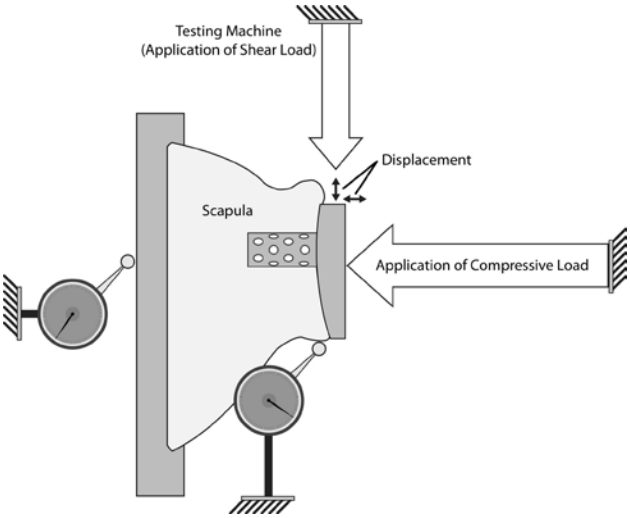

Figure 1: *Displacement test, performed to measure initial glenoid baseplate fixation before and after cyclic loading as a 357N shear load and a 433N compressive load are applied to the glenoid baseplate.*

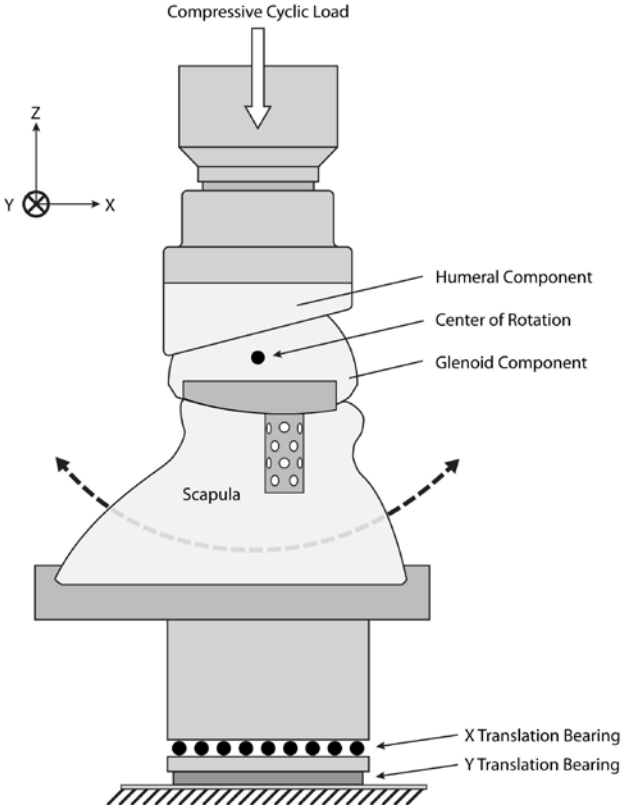

Figure 2: *Cyclic test, performed to simulate 55° abduction in the scapular plane for 10,000 cycles at 0.5 Hz as a 750N load is applied*

before and after cyclic loading. For both the notched and non-notched scapulae, initial fixation of the glenoid baseplate were achieved using four (one superior, three inferior), 4.5x30mm diameter poly-axial locking compression screws and a pressfit tapered cage peg; because the inferior screw is often fractured in a grade 4 scapular notching condition,[7] a shorter inferior screw (18mm) was utilized in the notched scapulae (the other 3 screws were 30mm in length). After assembly, each composite scapulae were cut and potted with bone cement. Statistical analysis was performed by means of a two-tailed unpaired student's t-test (significance defined as $p < 0.05$) to compare prosthesis displacements relative to each scapulae (notched and non-notched) in the directions of the shear and compressive loads before and after cyclic loading.

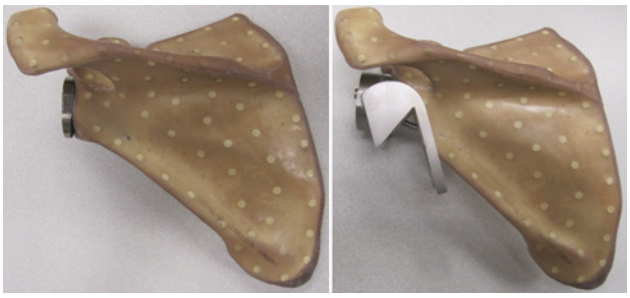

Figure 3: *Posterior views of the non-notched (Grade 0) composite scapula (left) and the Grade 4 scapular notch composite scapula with cut guide (right).*

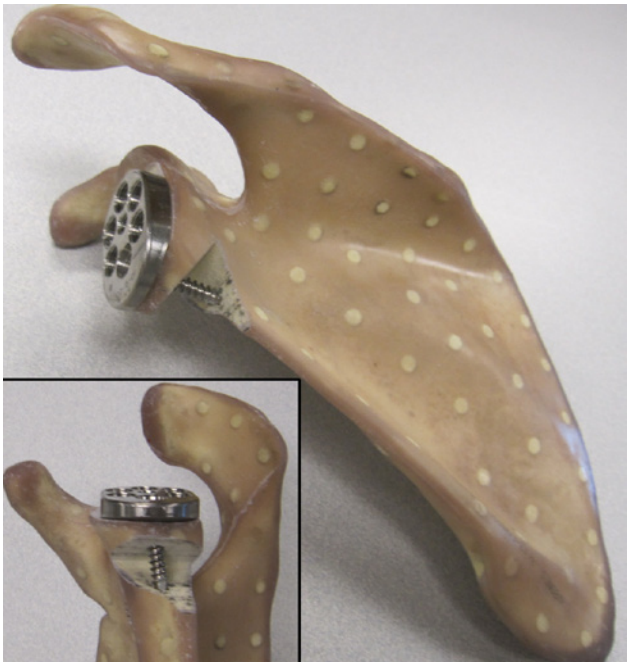

Figure 4: *Image of Grade 4 scapular notch composite scapula*

Results

For the composite scapulae without a scapular notch (e.g. Grade 0), glenoid baseplate displacement did not exceed the generally-accepted 150 micron threshold[21-23] for osseous integration before or after cyclic loading in any component tested. For the composite scapulae with a scapular notch (e.g. Grade 4), glenoid baseplate displacement exceeded 150 microns in 2 of the 7 samples before

cyclic loading and in 3 of the 7 samples after cyclic loading. As described in Table 2, the average pre-cyclic glenoid baseplate displacement in the direction of the shear load was significantly greater in the composite scapulae with a scapular notch than that of composite scapula without a scapular notch both before ($p = 0.030$) and after ($p = 0.023$) cyclic loading.

Table 2: *Glenoid baseplate displacement before and after cyclic loading in Grade 0 and Grade 4 Notched Composite Scapulae*

Glenoid Baseplate Displacement (microns)	Composite Scapulae, Grade 0 Notch	Composite Scapulae, Grade 4 Notch	p Value
Compression, Pre-cyclic	78 ± 27	99 ± 42	NS
Compression, Post-cyclic	83 ± 22	105 ± 46	NS
Shear, Pre-cyclic	66 ± 19	114 ± 28	0.030
Shear, Post-cyclic	58 ± 42	134 ± 65	0.023

Discussion

Our study demonstrated that in a majority of cases, the generally-accepted 150 micron displacement threshold[21-23] for osseous integration was met both before (5/7 scapulae) and after (4/7 scapulae) cyclic loading in the scapulae bone model with a Grade 4 scapular notch. In the scapula bone models without a scapular notch, glenoid baseplate displacement did not exceed the generally-accepted 150 micron threshold before or after cyclic loading in any component tested. Additionally, the glenoid baseplate motion associated with the non-notched scapula in the direction of the applied shear load was significantly less both before ($p=0.030$) and after ($p=0.023$) cyclic loading than that of the scapulae with a scapular notch. Glenoid baseplate fixation was achievable in most cases in scapulae with a severe scapular notch; however, the fact that this micro-motion threshold was not met in all scapulae with a notch is concerning and implies that severe notching may play a role in initial glenoid baseplate fixation. Therefore, the results of our study lead us to reject the null hypothesis and conclude that a large scapular notch does impact glenoid fixation in reverse shoulder arthroplasty.

While the exact usage of reverse shoulder arthroplasty is difficult to know, it is clear that there has been a significant increase in procedures performed in the United States since the device was cleared by the FDA in March 2004. Joshi et al reported that 2,652 reverse shoulder arthroplasty procedures were performed in the United States in 2005 and 15,200 in 2008; they projected 31,584 reverse shoulders will be performed in the United States in 2012.[19] Given this increase in usage and that the long-term survivorship of these implants is unknown, the number of reverse shoulder arthroplasty revision surgeries is expected to increase in the coming years. In many of these revision surgeries the shoulder surgeon can expect that the scapular bone will be compromised due to the high prevalence of scapular notching as a complication.[2-9] Severe scapular notching may potentiate screw loosening and loss of glenoid baseplate fixation in the initial post-operative revision period. In this case, the shoulder surgeon will be faced with a choice between conversion of the failed reverse shoulder arthroplasty to a hemiarthroplasty or attempting to revise to a new reverse shoulder. The inferior scapula is a difficult area in which to bone graft and little information is available to guide shoulder surgeons in this clinical scenario. This study attempts to help clarify whether initial fixation/stability is possible in a representative bone model of a severely notched scapula.

The testing scenario in this study utilized 4 screws to achieve fixation. The glenoid baseplate utilized in this study permits a total of 6 poly-axial locking compression screw positions. It is unclear whether adding more screws to the construct could potentially mitigate the increased micromotion observed in a severely notched scapula scenario. Future work should evaluate this potential strategy given that a previous study demonstrated significant differences in glenoid baseplate displacement between different numbers of screws and different screw configurations in a 0.24 g/cm^3 polyurethane bone substitute model.[20] These fixation results may not be applicable to all reverse shoulder glenoid baseplates since not all offer 4 poly-axial locking compression screws.

Conclusions

Severe scapular notching can be a challenging clinical scenario that does seem to have an impact on glenoid baseplate fixation in a minority of severely notched composite scapulae tested. The use of reverse shoulder arthroplasty has increased exponentially over the past decade, and the Grammont reverse shoulder has been the device implanted most often and is associated with a scapular notching rate >70%. It is expected that in the coming years many of these implants will need to be revised, this study provides the surgeon performing the revision surgery guidance to achieve glenoid fixation in patients with a large scapular notch. Further study will help clarify whether additional strategies exist to mitigate this issue.

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The Effect of Reverse Shoulder Design Philosophy on Muscle Moment Arms

This report is a summary of the results of two different computer analyses related to moment arms. These data were presented at the 2013 Orthopaedic Research Society[1]

Background

Patients with cuff tear arthropathy (CTA) are characterized by an unstable shoulder due to deficient muscles surrounding the shoulder joint. In a healthy shoulder, the rotator cuff provides the force to stabilize the joint on the glenoid face and counteracts the vertical load of the deltoid. This force couple creates the moment that raises the arm. The absence of these muscles prevents the patient from abducting their arm because the loss of the stabilizing force from the cuff allows the humeral head to migrate surperiorly and escape the joint as the deltoid contracts. The design of the reverse shoulder is based on the concept that constraining the center of rotation of the joint to the scapula will allow the deltoid to raise the arm despite the absence of rotator cuff muscles. Early attempts at this concept suffered elevated risks of component loosening.[2,3] However, the modern reverse shoulder prosthesis (i.e. Grammont's reverse shoulder design) minimized this risk by moving the center of rotation of the joint to the face of the scapula to reduce loading at the fixation surface and increase the moment arm of the deltoid.[4] The results with this style of reverse shoulder design were promising and allowed patients that previously had no solution for their shoulder the opportunity to again perform activities of daily living (ADLs). However, complications such as infection and dislocations were more prevalent than with total shoulders.[5–7] Since the launch of modern reverse shoulders in the late 1990s, the designs have evolved to minimize the complications seen in the earlier version (e.g. instability, glenoid loosening, scapular notching, and impingement).[8,9,7] Another outcome that is common among reverse shoulder patients with severe deficiency in the rotator cuff is a lack of external rotation, particularly with the arm abducted. The result of this deficiency is the hand falls internally and stays in front of the face in a motion resembling person holding a horn to their face, hence the term 'horn-blowers'.[10] This has significant clinical impact, as patients who have a horn blower's sign after reverse shoulder are typically unable to perform simple activities with the operative arm, despite the restoration of elevation that one can reliably obtain from RSA. As reverse shoulder designs continue to change, understanding how the remaining active muscles (i.e. deltoid, pectoralis, posterior cuff) function after the device is implanted is critical to improving post-operative motions, and avoiding the horn-blower's sign and its associated problems. The focus of this study is to analyze the impact that three different reverse shoulder design philosophies have on the musculature surrounding the shoulder joint with regard to abduction and external rotation.

Bone and Implant Models:

The analysis of the muscle moment arms in abduction and external rotation began using 3D models of a "healthy" scapula, humerus, clavicle, and ribcage (Pacific Research Laboratories, Vashon, WA). The center of rotation of the humeral head was fixed relative to the face of the scapula including a space to represent the healthy cartilage. Muscle origin and insertion points for the three heads of the deltoid, subscapularis, infraspinatus, teres minor, teres major, and pectoralis major were estimated using a combination of bony landmarks on the anatomy and illustrations from Gray's anatomy.[5] Each muscle was segmented into three sections; either superior, middle, and inferior or anterior, middle, and posterior depending on the muscle. An example of the segmentation of the subscapularis is illustrated Figure 1.

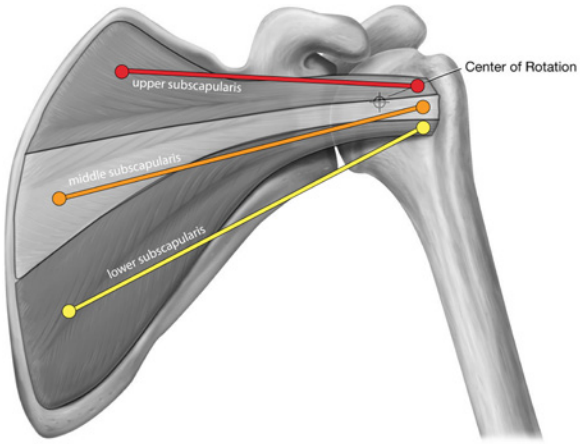

Figure 1: *Illustration of the segmentation of the subscapularis muscle into three sections.*

This model, including all the muscle origin and insertion points, was used for all of the subsequent assemblies to eliminate anatomic variation from the analysis. Three of the most common reverse shoulder design philosophies were chosen. The first is a traditional Grammont style prosthesis, which has a center of rotation on the glenoid face and a humeral cup that is placed into the proximal humeral bone. Countersinking the humeral cup inside the bone results in a small medial offset between the center of rotation of the humeral cup and the axis of the stem in the intramedullary canal. This design will be referred to as a medial glenosphere center of rotation with a medialized humerus (MGMH) meaning that the location of the humerus is the closest to the scapula of all three designs. The second concept is a prosthesis with a lateralized center of rotation on the glenosphere and a humeral component that rests inside the proximal humerus. This results in a lateralized glenosphere and a medialized humerus (LGMH), so the position of the humerus is more lateral than the Grammont-style design. The third design has a medialized center of rotation of the glenosphere with a humeral component that rests on top of the humerus as opposed to inside like the other two designs. The result of resting atop the humeral cut is the liner ends up much more medial relative to the IM axis of the humerus. This concept has a medialized center of rotation of the glenosphere and a lateralized humerus (MGLH), so the humerus is positioned further lateral than the previous two designs. For each design, the most commonly utilized commercially available implant was modeled based on published specifications (i.e. 36mm glenosphere for MGMH, 32mm glenosphere for LGMH, and 38mm glenosphere for MGLH). Each design was implanted into the aforementioned bone models following the manufacturer's recommended surgical technique.

Methodology

Point data from all the models described above were imported into a custom-written analysis software in Matlab (Mathworks, Natick, MA) and put through an abduction range of motion with the humerus in the scapular plane and the forearm in the neutral position. The range of motion for the humerus relative to the ground was 0° to 140°. A portion of this motion is generated at the glenohumeral joint and the remaining motion occurred at the scapulothoracic joint. The ratio of these two motions is referred to as the scapular rhythm. For the purposes of this analysis, the scapular rhythm for abduction in a reverse shoulder was based on data from Alta et al. who estimated the motion at 1.8:1 for glenohumeral to scapulothoracic.[11]

At four points in the range of abduction motion (0°, 30°, 60°, 90°), the humerus was held in the abduction plane and externally rotated from 30° of internal rotation to 60° of external rotation. All moment arm values were calculated relative to the axes of rotation by calculating the vector perpendicular to the axis of rotation starting at the center of rotation of the joint and ending at the line action of the muscle (e.g. the external rotation axis was defined as a line from the center of the epicondylar axis of the elbow through the center of the intramedullary axis of the humerus).

Model Validation

As this is a purely analytical model, it is important to compare to both previously published analytical models and cadaveric models to validate the results and verify the data are reasonable. The results of several papers were chosen for comparison.[12–17] Each of these papers analyzed the anatomic shoulder, the reverse shoulder, or both. The abduction moment arm of the deltoid was the early focus since it is directly affected by design in the reverse shoulder.[18] Not all studies used the same range of motion, so a range that was reported in all studies was chosen. For this range, the minimum and maximum moment values are reported in Table 1.

Table 1: *Comparison of models at 0° and 60° of glenohumeral abduction (not including scapulothoracic motion) for the anatomic shoulder. (AD = Anterior Deltoid, MD = Middle Deltoid, PD = Posterior Deltoid)*

	AD(mm)		MD (mm)		PD (mm)	
	Min	Max	Min	Max	Min	Max
Current Study	-8.2	27.9	-3.3	22.5	-49.3	-4.3
Ackland[14]	2	28	8	30	-18	2
Kontaxis[13]	-30	22	33	33	-26	6
Otis[17]	-7	22	14	32	-55	-23
Terrier[12]	0	15	32	27	-27	1

Table 2: *Comparison of models at 0° and 60° of glenohumeral abduction (not including scapulothoracic motion) for the MGMH reverse shoulder design.*

	AD(mm)		MD (mm)		PD (mm)	
	Min	Max	Min	Max	Min	Max
Current Study	17.2	50.8	25.1	47.1	-24.5	12.8
Ackland[14]	16	36	30	46	0	12
Kontaxis[13]	-7	36	40	55	-10	17
Otis[17]	-	-	-	-	-	-
Terrier[12]	15	34	37	49	-15	15

While the absolute values of the moment arms are highly dependent on the geometry of the specimens and models, the trends and relative changes are consistent across the models. The trends for the anatomic and reverse shoulders, particularly for the anterior and posterior deltoid are very similar. The middle deltoid in the anatomic model has a significant range of motion where the approximated line of action passes through the bony anatomy. This model does not account for muscle wrapping (i.e. where the bony anatomy would physically prevent the muscle line of action

getting that close to the center of rotation) which artificially reduces the moment arm of the middle deltoid through the low abduction range of motion. However, the moment arm values at higher elevations agree well with reported results since the data are not confounded by muscle wrapping.

Results and Discussion

The designs are grouped based on different aspects that are under review (e.g. design philosophy, surgical technique, or bony deformity). All plots include the "normal" anatomy for comparison. As mentioned by Ackland et al. the absolute value of the size of the moment arm for each muscle is not as important as the relative comparison between the assemblies because the absolute value is a function of the bony anatomy used in this study.[14]

The first comparison of the middle deltoid highlights the relative differences between the designs. The primary result in the deltoid is a large increase in efficiency created by moving the center of rotation to the glenoid face. This was originally described by Grammont in the rationale for the Grammont shoulder, and has been confirmed by other biomechanical assessments of the reverse shoulder. While all three designs increase the deltoid moment arm, the MGLH and MGMH designs are 50% more efficient than the LGMH design due to the lateralization of the center of rotation off the face of the glenoid. This has been previously reported in the literature, but serves as an example of how the grouping of designs can help to distinguish certain characteristics.[19]

Next the abduction moment arms of both the infraspinatus and subscapularis muscles were analyzed through the range of motion. In a healthy shoulder these muscles counteract each other to provide stabilization and joint compression. It is apparent that the distal shift inherent in all reverse shoulder designs has a profound effect on the potential function of these muscles. These two muscles' primarily function to compress the humerus against the glenoid with a secondarily function as abductors in the anatomic condition. After the reverse shoulder procedure, these muscles create an adduction moment for a majority of the range of motion, regardless of design.

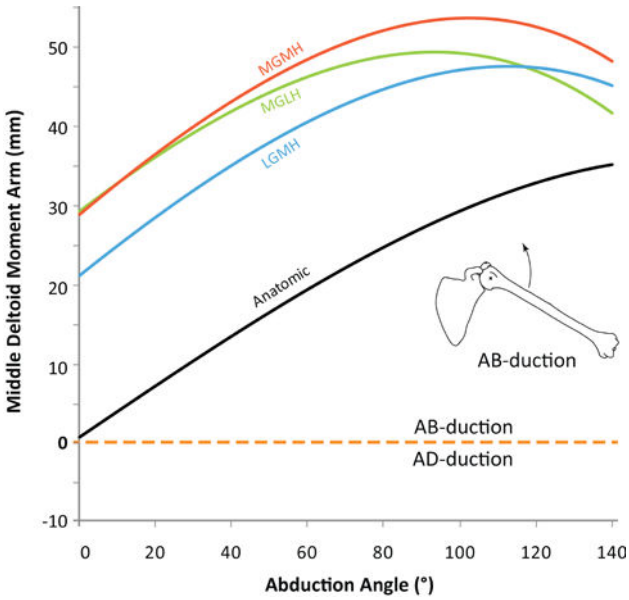

Figure 2: *Abduction moment arm for the middle deltoid through 140° of abduction.*

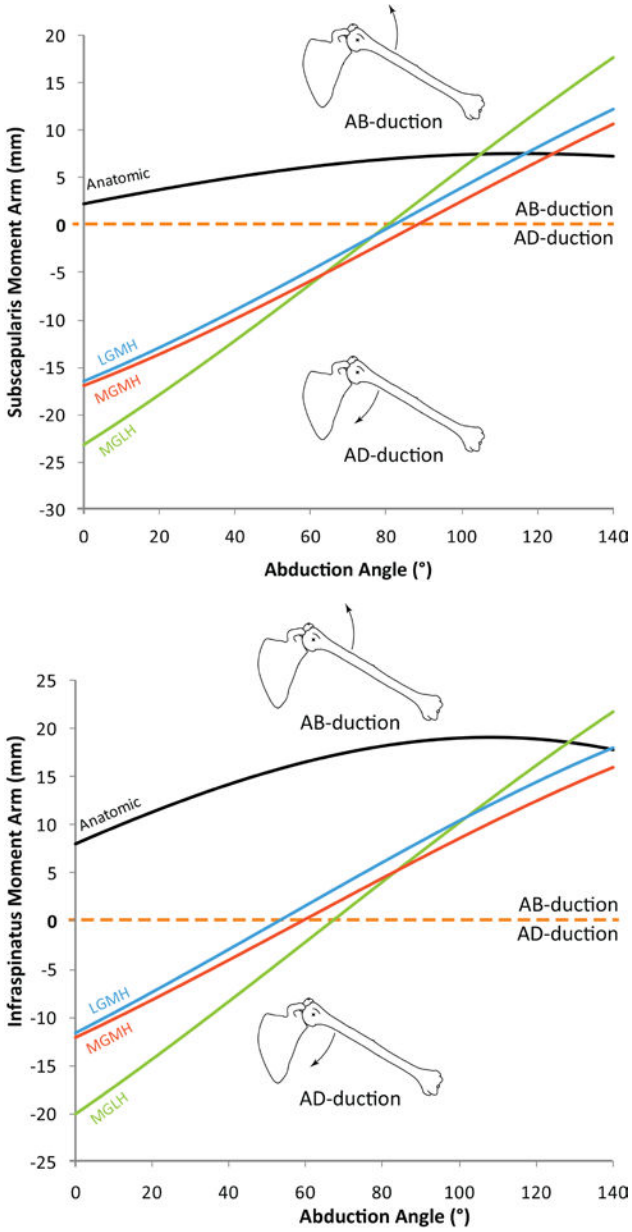

Figure 3: *(top) Abduction moment arms of subscapularis through 140° of abduction; (bottom) Abduction moment arms of infraspinatus through 140° of abduction.*

The external rotation moment arm was analyzed for the

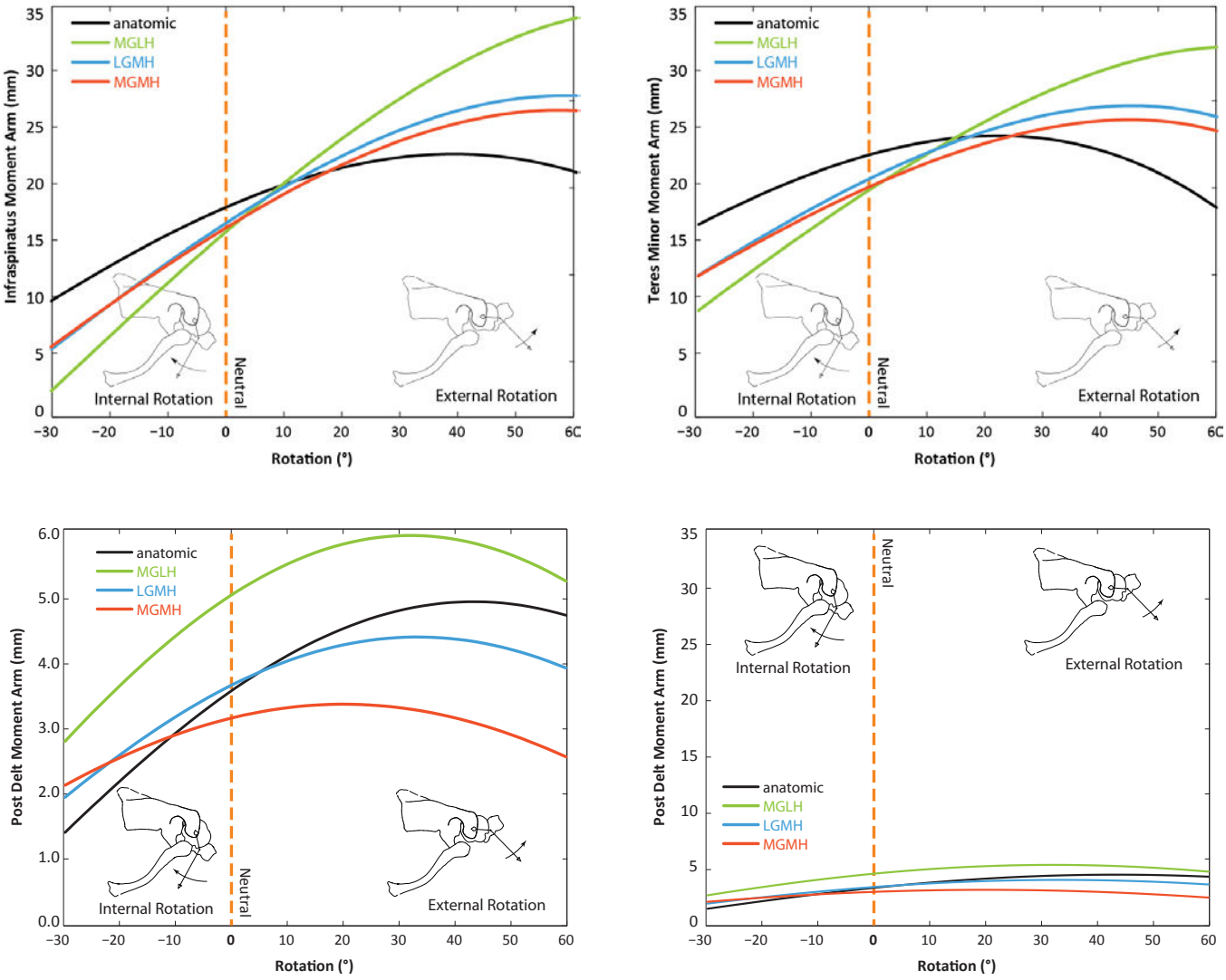

Figure 4. *(upper left) External rotation moment arm of infraspinatus for rotation from 30° internal rotation to 60° external rotation at 30° of abduction. (upper right) External rotation moment arm of teres minor for rotation from 30° internal rotation to 60° external rotation at 30° of abduction. (lower left) External rotation moment arm of posterior deltoid for rotation from 30° internal rotation to 60° external rotation at 30° of abduction. (lower right) same plot of posterior deltoid scaled the same as IS and TM plots for comparison.*

primary external rotators in the shoulder (i.e. posterior deltoid, infraspinatus, and teres minor). Each of these muscles were analyzed at 30°, 60°, and 90° of abduction to determine how the moment arms change through the range of motion.

The neutral position in the figures refers to the position with the forearm pointing anterior. The moment arms for both the infraspinatus and teres minor are decreased relative to anatomic in internal rotation and at neutral for all reverse shoulder designs. Internal rotation is not typically a deficiency in reverse shoulder patients, because of the other internal rotators (e.g. pectoralis major and latissimus dorsi). The efficiency of the external rotators quickly increases beyond anatomic levels as the arm is externally rotated. This trend is continued at larger abduction angles. Unfortunately, patients with severe CTA frequently have either atrophic and intact external rotators or tearing of these external rotators leaving only the posterior deltoid to produce the external rotation moment. The lower left figure demonstrates that the lateralization of the humerus has a minor impact on the moment arm of the posterior deltoid. The MGLH design increases the efficiency of the posterior deltoid relative to the other two designs. While the increase as a function of the anatomic moment arm is substantial (60% increase in efficiency relative to anatomic), the plot on the lower right illustrates the large disparity between the magnitude of the moment arm of the posterior deltoid and the primary external rotators. The maximum moment arm for the PD

is 20% of the IS or TMI muscles. This is a potential explanation for the poor external rotation achieved by some reverse shoulder recipients. This is exacerbated by the designs with medial humeral stems because the posterior deltoid efficiency is decreased relative to anatomic.

Conclusion

Due to the distal shift and medialization of the humerus relative to the scapula, the lines of action of all the muscles are affected by the reverse shoulder. The results confirmed previous studies showing that medializing the center of rotation greatly improves deltoid efficiency, and that this increase in efficiency is decreased as the center of rotation is lateralized off the glenoid face.[19] It appears that the increased distance between the center of rotation of the joint and the muscle attachment sites influence the moment arm for the posterior rotators as evidenced in the plot of the IS muscle. The grouping of the MGMH and LGMH indicates that the overall distance between the attachment site and center of rotation is the most important variable since these two designs change the center of rotation in different ways, but the external rotation moment arms are very similar. More specifically, one places the COR at the glenoid face while the other is 10mm lateral of the glenoid face meaning these moment arms are less affected by the location of the COR in contrast to the abduction moment arm for the deltoid which is highly influenced by changes in COR.

This study demonstrates that the combination of variables making up each reverse shoulder design can impact the moment arms of muscles in different ways. This is particularly important in cases of severe rotator cuff arthropathy where the primary external rotators (infraspinatus and teres minor) may be absent. Different design philosophies can lead to different orientations of the musculature and may behave differently clinically. Longer, focused clinical follow-up must be performed to confirm this theory.

Future Work

The current model does not include muscle wrapping in its analysis. This prevents the model from detecting when the line of action penetrates the bony anatomy. Future work for this model includes adding muscle wrapping following techniques published by Kontaxis and Johnson.[13] All rotations in this model were around a fixed center of rotation. More realistic motions for the anatomic shoulder could change the results in small ways. Future versions will attempt to include this motion. This model was performed using bone models from Pacific Research, which represents a single bony anatomy. Incorporation of CT reconstructions will help determine which types of anatomy are most affected by reverse shoulder design philosophy. Beyond the model, a retrospective analysis of reverse shoulder outcomes will be performed to see if any of the findings in this model have had a clinical impact on post-operative motion.

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