Scapular Notching in Reverse Shoulder Arthroplasty
Validation of a Computer Impingement Model


Abstract

Purpose: The purpose of this study is to validate a reverse shoulder computer impingement model and quantify the impact of implant position on scapular impingement by comparing it to that of a radiographic analysis of 256 patients who received the same prosthesis and were followed postoperatively for an average of 22.2 months.

Methods: A geometric computer analysis quantified anterior and posterior scapular impingement as the humerus was internally and externally rotated at varying levels of abduction and adduction relative to a fixed scapula at defined glenoid implant positions. These impingement results were compared to radiographic study of 256 patients who were analyzed for notching, glenoid baseplate position, and glenosphere overhang.

Results: The computer model predicted no impingement at 0° humeral abduction in the scapular plane for the 38 mm, 42 mm, and 46 mm devices when the glenoid baseplate cage peg is positioned 18.6 mm, 20.4 mm, and 22.7 mm from the inferior glenoid rim (of the reamed glenoid) or when glenosphere overhang of 4.6 mm, 4.7 mm, and 4.5 mm was obtained with each size glenosphere, respectively. When compared to the radiographic analysis, the computer model correctly predicted impingement based upon glenoid baseplate position in 18 of 26 patients with scapular notching and based upon glenosphere overhang in 15 of 26 patients with scapular notching.

Conclusions: Reverse shoulder implant positioning plays an important role in scapular notching. The results of this study demonstrate that the computer impingement model can effectively predict impingement based upon implant positioning in a majority of patients who developed scapular notching clinically. This computer analysis provides guidance to surgeons on implant positions that reduce scapular notching, a well-documented complication of reverse shoulder arthroplasty.

Assessment tools that simulate clinical conditions are essential to the design and development process to ensure that new generations of implant systems have lower complications and failure rates than their predecessors. To this end, a computer impingement model was developed and utilized in 2005 to design a reverse shoulder prosthesis (Equinoxe; Exactech, Inc.) to minimize the well-documented complication of scapular notching.1 This 3-D computer impingement model simulated humeral abduction in the scapular plane and permitted the modification of several prosthesis design parameters (e.g., humeral neck angle, humeral liner constraint, glenosphere diameter, glenosphere thickness, and inferior glenosphere offset) to quantify their effect on several different functional measurements (e.g., range of motion, stability, and impingement). The purpose of this study is to validate the computer impingement model...
and quantify the effect of prosthesis positioning on scapular impingement by comparing it to that of a large scale radiographic analysis of 256 patients who received this prosthesis and have been followed postoperatively for an average of 22.2 months. The results of this comparison will be used to make recommendations for prosthesis positioning to avoid scapular notching.

**Materials and Methods**

The reverse shoulder prostheses were geometrically modeled using 3-D computer-aided design software (Unigraphics; UGS, Inc.). The 38 mm, 42 mm, and 46 mm designs have a 145° neck and liner angle; a humeral liner constraint (i.e., ratio of humeral liner mouth depth to width) of 0.260, 0.250, 0.240; and a glenosphere geometry of 38 mm x 21 mm, 42 mm x 23 mm, and 46 mm x 25 mm, respectively. The center of rotation of each glenosphere is on average 2.0 mm lateral to the spherically reamed glenoid surface. Due to the 4 mm superiorly shifted glenoid baseplate peg, when the inferior rim of the glenoid baseplate is aligned with the inferior rim of the glenoid, the 38 mm, 42 mm, and 46 mm designs provide 2.25 mm, 4.25 mm, and 6.25 mm of glenosphere overhang, respectively. After modeling, each prosthesis was assembled to a 3-D digitized scapula (3-D male scapula; Zygote Media Group, Inc.) to create a functional glenohumeral joint. Prior to assembly, less than 1 mm of bone was removed from the glenoid of the digitized scapula to create a conforming surface. Each size implant was oriented on the normalized digitized scapula with a prosthesis-scapular neck angle (PSNA) of 110°. To quantify the impact of inferiorly shifting the glenoid baseplate on impingement, the glenoid baseplate cage peg was implanted in the center of the glenoid and subsequently inferiorly shifted by 2 mm and 4 mm for each glenosphere diameter. As depicted in Figure 1, a geometric computer analysis was then conducted to quantify anterior and posterior scapular impingement as the humerus was internally and externally rotated at varying levels of abduction and adduction relative to a fixed scapula. The locations of impingement on the scapula were plotted versus motion (i.e., humeral position); the total motion was calculated from the enclosed area of each plot.

To evaluate the validity of the computer impingement model, immediate postoperative and the latest follow-up scapular anteroposterior (AP) radiographs were collected from 256 patients (72.5 ± 7.7 years; range: 42 to 93 years) who received a 38 mm (N = 155), 42 mm (N = 91), or 46 mm (N = 10) primary reverse shoulder by nine different surgeons using a deltopectoral approach. Each surgeon involved in the study reamed the glenoid at neutral, making no attempt to eccentrically ream the glenoid to inferiorly tilt the glenosphere. The average follow-up was 22.2 ± 8.7 months (range: 10 to 44 months).

All nine reviewers assessed all radiographs for notching in a blinded-fashion according to the Nerot-Sirveaux grading scale. The location of the glenoid baseplate cage peg from the inferior glenoid rim and the amount of glenosphere overhang was measured on each immediate postoperative AP radiograph using digital calipers. The PSNA was measured from each immediate postoperative AP radiograph using a goniometer. Each radiographic measurement was sorted and compared according to the 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.

**Results**

The motion profile defining the locations of humeral impingement on the scapula for the 38 mm, 42 mm, and 46 mm reverse shoulders when the glenoid baseplate cage peg is positioned in the center of the glenoid and inferiorly shifted by 2 mm and 4 mm (as the humerus was internally/externally rotated at varying levels of abduction/adduction relative to a fixed scapula) is depicted in Figures 2, 3, and 4, respectively. When the glenoid baseplate cage peg is positioned in the center of the glenoid for the 38 mm, 42 mm, and 46 mm glenospheres, the center of the peg is positioned 21.0 mm, 19.1 mm, and 17.3 mm from the inferior glenoid rim (of the reamed glenoid) resulting in a glenosphere overhang of 2.2 mm, 6.1 mm, and 9.8 mm, respectively. The computer model predicted no impingement at 0° humeral abduction in the scapular plane for the 38 mm, 42 mm, and 46 mm devices when the glenoid baseplate cage peg is positioned 18.6 mm, 20.4 mm, and 22.7 mm from the inferior glenoid rim (of the reamed glenoid), respectively. Similarly, the computer model predicted no impingement at 0° humeral abduction in the scapular plane for the 38 mm, 42 mm, and 46 mm devices when glenosphere overhang of 4.6 mm, 4.7 mm, and 4.5

![Figure 1](image-url) Representative image of the 38 mm reverse shoulder model: humeral liner impingement with the scapula at 11° and 68° abduction when the humeral component is externally rotated at 45° relative to the plane of the scapula.
mm was obtained with each size glenosphere, respectively. The total motion of the 2 mm and 4 mm inferiorly shifted 38 mm device is 13% and 21% greater than when positioned in the center of the glenoid. The total motion of the 2 and 4 mm inferiorly shifted 42 mm device is 11% and 20% greater than when positioned in the center of the glenoid. The total motion of the 2 mm and 4 mm inferiorly shifted 46 mm device is 11% and 22% greater than when positioned in the center of the glenoid.

At low-levels of humeral abduction, the humeral liner was observed to impinge on the inferior scapular neck during both internal and external rotation within the physiologic range of motion for the 38 mm and 42 mm designs when the glenoid baseplate cage peg is positioned in the center of the glenoid. When the baseplate peg is positioned in the center of the glenoid for the 46 mm glenosphere, inferiorly shifted by 2 mm for the 42 mm glenosphere and inferiorly shifted by 4 mm for the 38 mm glenosphere, no such impingement

Figure 2 38 mm reverse ROM profile: impact of glenoid baseplate superior-inferior position on impingement. Color graph available online at www.nyuhjdbulletin.org.

Figure 3 42 mm reverse ROM profile: impact of glenoid baseplate superior-inferior position on impingement. Color graph available online at www.nyuhjdbulletin.org.

Figure 4 46 mm reverse ROM profile: impact of glenoid baseplate superior-inferior position on impingement. Color graph available online at www.nyuhjdbulletin.org.
occurs during internal and external rotation. At mid-levels of humeral abduction for the 38 mm, 42 mm, and 46 mm glenospheres, no prosthesis impingement occurs within the physiologic range of motion regardless of the position of the baseplate on the glenoid. At high-levels of humeral abduction for the 38 mm, 42 mm, and 46 mm glenospheres, the humeral liner impinges on the base of the coracoid in internal rotation and the base of the acromion in external rotation within the physiologic range of motion regardless of the position of the baseplate on the glenoid.

The radiographic analysis demonstrated that 26 of 256 patients (10.2%) had a scapular notch (20 Grade 1 and 6 Grade 2; no Grade 3 or 4 notches were observed). Patients with 38 mm, 42 mm, and 46 mm glenospheres had a scapular notching rate of 14.2% (16 Grade 1 and 6 Grade 2), 4.4% (4 Grade 1), and 0%, respectively. The average position of the glenoid baseplate peg from the inferior glenoid rim was 19.1 ± 2.4 mm. The average glenoid plate position of 38 mm patients (18.7 ± 2.3 mm) was significantly lower on the glenoid (p < 0.001) than that of 42 mm patients (19.0 ± 2.9 mm) but not statistically different than that of 46 mm patients (19.0 ± 2.9 mm). The average glenosphere overhang was 5.0 ± 2.4 mm. The average glenosphere overhang of 46 mm patients (8.1 ± 2.9 mm) was significantly more (p < 0.001) than that of 38 mm patients (4.6 ± 2.2 mm) and 42 mm patients (5.3 ± 2.5 mm). Additionally, the average glenosphere overhang of 42 mm patients was significantly more (p = 0.036) than that of 38 mm patients. The average PSNA was 98.0 ± 12.0°.

As described in Table 1, the average glenoid plate position of patients without a notch (19.0 ± 2.4 mm) was significantly lower on the glenoid (p = 0.03) than that of patients with a notch (20.1 ± 1.9 mm). The average glenoid plate position of 38 mm patients without a notch (18.5 ± 2.3 mm) was significantly lower on the glenoid (p = 0.002) than that of 38 mm patients with a notch (19.9 ± 1.8 mm). The average glenosphere overhang of patients without a notch (5.1 ± 2.5 mm) was significantly greater (p = 0.006) than that of patients with a notch (3.8 ± 1.8 mm). The average glenosphere overhang of 38 mm patients without a notch (4.8 ± 2.3 mm) was significantly more (p = 0.042) than that of 38 mm patients with a notch (3.7 ± 1.9 mm). The average PSNA of patients without a notch (97.4 ± 12.0°) was significantly less (p = 0.017) than that of patients with a notch (103.4 ± 10.7°). The average PSNA of 38 mm patients without a notch (96.9 ± 11.8°) was significantly less (p = 0.009) than that of 38 mm patients with a notch (104.3 ± 11.2°).

Comparing the results of the computer model with that of the radiographic analysis for glenoid plate position (i.e., the computer model predicted no impingement for the 38 mm, 42 mm, and 46 mm devices when the glenoid plate peg was positioned 18.6 mm, 20.4 mm, and 22.7 mm from the inferior glenoid rim, respectively) demonstrates that 18 of 26 patients (15 of 22 38 mm glenospheres and 3 of 4 42 mm glenospheres) who notched clinically were correctly predicted to impinge based upon the placement of the glenoid plate. Similarly, using glenosphere overhang (i.e., the computer model predicted no impingement for the 38 mm, 42 mm, and 46 mm devices when glenosphere overhang of 4.6 mm, 4.7 mm, and 4.5 mm was achieved relative to the inferior glenoid rim, respectively) demonstrates that 15 of 26 patients (12 of 22 38 mm glenospheres and 3 of 4 42 mm glenospheres) who notched clinically were correctly predicted to impinge based upon the amount of glenosphere overhang.

Discussion
This computer impingement study quantified the 3-D motion profile and the locations of prosthesis impingement for three glenospheres sizes when the glenoid baseplate cage peg is positioned in the center of the glenoid and subsequently inferiorly shifted by 2 mm and 4 mm. These results demonstrate that inferiorly shifting the glenoid baseplate is an effective method to increase humeral motion, where each

<table>
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<tr>
<th>Table 1</th>
<th>Radiographic Measurements, a Comparison of Patients with and without a Notch</th>
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<tbody>
<tr>
<td>Glenoid Plate Position, all Patients</td>
<td>19.0 ± 2.4 mm</td>
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<tr>
<td>Glenoid Plate Position, 38 mm</td>
<td>18.5 ± 2.3 mm</td>
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<tr>
<td>Glenoid Plate Position, 42 mm</td>
<td>19.8 ± 2.4 mm</td>
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<tr>
<td>Glenoid Plate Position, 46 mm</td>
<td>19.0 ± 2.9 mm</td>
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<tr>
<td>Glenosphere Overhang, all Patients</td>
<td>5.1 ± 2.5 mm</td>
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<tr>
<td>Glenosphere Overhang, 38 mm</td>
<td>4.8 ± 2.3 mm</td>
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<tr>
<td>Glenosphere Overhang, 42 mm</td>
<td>5.3 ± 2.5 mm</td>
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<tr>
<td>Glenosphere Overhang, 46 mm</td>
<td>8.1 ± 2.9 mm</td>
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<tr>
<td>PSNA, all Patients</td>
<td>97.4 ± 12.0°</td>
</tr>
<tr>
<td>PSNA, 38 mm</td>
<td>96.9 ± 11.8°</td>
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<tr>
<td>PSNA, 42 mm</td>
<td>98.8 ± 12.3°</td>
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<tr>
<td>PSNA, 46 mm</td>
<td>92.0 ± 9.8°</td>
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2 mm inferior shift of the glenoid baseplate on the glenoid bone provided on average of 11% more motion. Furthermore, no impingement was observed at 0° humeral abduction in the scapular plane for the 38 mm, 42 mm, and 46 mm devices when the glenoid plate cage peg was positioned 18.6 mm, 20.4 mm, and 22.7 mm from the inferior glenoid rim of the reamed glenoid, respectively. Additionally, no impingement was observed at 0° humeral abduction in the scapular plane for the 38 mm, 42 mm, and 46 mm devices when glenosphere overhang of 4.6 mm, 4.7 mm, and 4.5 mm was obtained with each size glenosphere, respectively. The computer model correctly predicted impingement based upon glenoid baseplate position in 18 of 26 patients and based upon glenosphere overhang in 15 of 26 patients who notched clinically.

Numerous studies have quantified scapular impingement using various methods and models. Nyfeller and coworkers conducted a cadaveric ROM study in three planes (scapular, frontal, and sagittal planes), quantifying the effect of prosthesis placement on impingement. For four different glenoid implant positions, Nyfeller reported impingement in all three planes; motion in the scapular plane was significantly greater than motion in the frontal and sagittal planes. Additionally, Nyfeller reported that 2 mm to 4 mm inferiorly shifted glenoid baseplate position resulted in the least “adduction deficit” (i.e., impingement) in all three planes of motion. Similarly, Kontaxis and colleagues utilized a 3-D computer model to simulate humeral abduction, forward flexion, and elevation in order to quantify the effect of various reverse shoulder design parameters on impingement. Kontaxis reported impingement in all three motions for every design parameter configuration. Inferiorly shifting the glenoid baseplate provided the largest improvement in motion (1 mm inferior shift resulted in approximately 5° more motion). Kontaxis also reported that when the glenoid baseplate was inferiorly shifted 6 mm, no inferior impingement was observed for any motion; however, when the glenoid baseplate was inferiorly shifted by 4 mm or greater, all glenoid plate compression screws were located outside of the glenoid border. This observation of screw perforation contrasts with that of our study in which neither the glenoid baseplate cage peg nor the compression screws were observed to perforate the scapula in either the computer model or in the evaluation of the scapular AP radiographs.

These observations of multi-plane impingement by Nyfeller and Kontaxis are corroborated clinically by Simovitch and associates who reported inferior (44%), posterior (33%), and anterior (8%) scapular notching in a radiographic evaluation of 77 reverse shoulders with an average follow-up of 44 months. The likely mechanism for posterior notching is described by Middernacht and coworkers, who conducted a cadaveric study of 200 scapulae to characterize the geometry of the lateral border of the scapula and reported that its posterior portion is offset in the glenoid plane. Middernacht stated that this offset could create humeral impingement on the scapular during external rotation. The ROM assessment presented in our study is in agreement with these reports of anterior and posterior impingement. The ROM profile depicted in Figures 2 through 4 demonstrate that there is a wide range of internal and external rotation that could result in scapular impingement while the arm is at the patient’s side, particularly when the glenoid plate is not inferiorly shifted when using the smaller diameter glenospheres.

In the aforementioned cadaveric study, Middernacht noted that due to the infraglenoid tubercle, glenosphere overhang could not always be achieved: 75% of 36 mm glenospheres (150 scapulae) and 27% of 42 mm glenospheres (55 scapulae) when assembled to the glenoid baseplate as inferiorly as possible had no overhang. This reported majority of implants unable to achieve glenosphere overhang contrasts with that of our study which reported an average glenosphere overhang of 5.0 ± 2.4 mm, where only 2.6% of 38 mm glenospheres (4 of 155), 1.1% of 42 mm glenospheres (1 of 91), and 0% of 46 mm glenospheres (0 of 10) were observed to have no glenosphere overhang.

De Wilde and colleagues also developed a 3-D computer model to quantify the impact of varying reverse shoulder design parameters on scapular impingement as the humerus was abducted and adducted in the scapular plane. De Wilde reported that an inferiorly shifted glenoid baseplate is the most effective way to minimize scapular notching. Inferiorly shifting the glenoid baseplate to obtain a glenosphere overhang from 0 mm to 5 mm resulted in an increase of 39° of humeral motion in the average scapula and an increase of 16° in the worst-case scapula, where inferior impingement was completely eliminated with 3 mm of overhang in the average scapular. Based upon these results, De Wilde recommended a lower placement of the glenoid baseplate in order to increase the surgeon’s ability to achieve glenosphere overhang and minimize scapular notching. Chou and coworkers conducted a similar ROM study in sawbones to quantify scapular impingement using glenoid implants of varying inferior and lateral offsets as the humerus was abducted and adducted in the scapular plane. Chou reported that approximately 1 mm of inferior glenosphere offset resulted in 3.25° to 5° more humeral motion, depending upon the size of the glenosphere. Based upon these results, Chou recommended using inferiorly offset glenosphere designs to avoid scapular notching. The recommendations of each of these studies are in agreement with our radiographic analysis and our computer model, which demonstrate that inferiorly shifting the glenoid component is an effective method to increase humeral motion and minimize scapular impingement.

There are several limitations associated with our study methodology. First, the measurements of impingement obtained from the computer analysis do not consider the effect of soft tissue constraint, active motion, or the presence of scapular neck osteophytes. As a result, care should be taken
in extrapolating these analytical ROM results to clinical motion because clinical measurements include scapular motion, while this computer model does not. Second, the computer model utilizes a normalized scapula morphology; we did not evaluate the effect of scapula size or anatomic variation on impingement as did De Wilde. Third, we measured inferior scapular notching from immediate postoperative and latest follow-up scapular AP radiographs. Simovitch and associates used axillary radiographs to report both anterior and posterior scapular notching; since we did not collect axillary radiographs, we were unable to report our anterior or posterior scapular notching rate. And finally, the computer model did not evaluate the effect of different glenoid wear patterns on impingement. Sirveaux and colleagues states that glenoid baseplate positioning in types E2 or E3 worn glenoids can increase impingement, and Levigne and co-workers reported that notching was correlated to the type of preoperative glenoid erosion. As a result, our recommendations on glenoid plate position and glenosphere overhang may not apply to severely worn glenoids.

In conclusion, the results of this study demonstrate that inferiorly shifting the glenoid baseplate or glenosphere is an effective method to reduce scapular impingement. A comparison of our computer model results with that of the radiographic evaluation demonstrates that the utilized model can effectively predict impingement in a majority of patients who developed scapular notching clinically based upon both glenoid plate position and the amount of glenosphere overhang. The computer impingement model has the potential to be a cost effective method to simulate scapular impingement, and as such, it is useful to the design and development process. Future work should permit the model to accommodate anatomic variations in scapular size and morphology and address the presence of deformities.

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Disclosure Statement

References