A comparison of onlay versus inlay glenoid component loosening in total shoulder arthroplasty

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Background: Glenoid component loosening is common in total shoulder arthroplasty (TSA), often resulting from the mechanical interaction of glenohumeral components. This cadaveric study was performed to evaluate and to compare commercially available onlay and inlay glenoid prosthetic designs with respect to loading characteristics and loosening.

Methods: Sixteen prescreened cadaveric shoulders (8 matched pairs) underwent either onlay or inlay TSA. We created a custom glenohumeral loading model and used cycles of 5 mm anterior-posterior humeral translation to simulate a rocking-horse loosening mechanism for all testing. Articular TekScan measurements were performed with 9.1 kg (88.9 N) of glenohumeral compression before and after TSA. Fatigue testing was performed with 34.0 kg (333.6 N) of glenohumeral compression using high-definition video to document gross glenoid loosening. Testing ended with gross loosening or a maximum of 4000 cycles. Mean contact area, pressure, and joint reaction force were used to compare the 2 glenoid designs.

Results: In both implant types, contact area decreased and pressure increased after TSA \((P < .0001)\). Force increased at the onlay component edge only \((P = .0012)\) compared with native glenoid testing. Force was greater in the onlay vs. the inlay implants \((P < .0001)\). During fatigue testing, all onlay glenoid components exhibited gross loosening at a mean of 1126 cycles (range, 749-1838), whereas none of the inlay glenoid components exhibited gross loosening \((P < .0001)\).

Conclusion: The inlay glenoid implant exhibited biomechanical characteristics favoring stability and decreased loosening compared with the onlay glenoid implant in this cadaveric model.

Level of evidence: Basic Science Study; Biomechanics

Keywords: Glenoid component loosening; total shoulder arthroplasty; inlay glenoid component; onlay glenoid component; glenoid implant design; biomechanical testing

Glenoid loosening remains a common complication of total shoulder arthroplasty (TSA). All types of traditional onlay glenoid prostheses exhibit signs of loosening at a high rate, even with optimally placed components.19,20,21,24 Optimizing
implant survivorship is a fundamental part of all arthroplasty research and development in orthopedics. Glenoid implant design has often attempted to restore native anatomy, with focus on minimizing micromotion at the bone-implant interface. It is generally accepted that implant micromotion <150 μm is necessary for bone ingrowth. Metal-backed and all-polyethylene glenoid designs, with varying backside conformations, have each attempted to optimize fixation to bone. Metal-backed glenoids have unacceptably high failure rates and higher failure rates than all-polyethylene glenoids in multiple studies. All-polyethylene glenoid designs use backside keels or pegs to enhance fixation to bone. Although the literature is contradictory, it appears that pegged glenoids show superior survivorship to keeled glenoids.

Glenoid loosening is especially concerning in younger patients, who have a longer life expectancy and higher demands. The “rocking-horse” mechanism of loosening occurs when the humerus translates on the glenoid in any plane, producing edge loading, which can result in opposite edge liftoff and component loosening. Repetitive eccentric rim loading by the humeral head results in the torque on the implant, applied tensile stress, and opposite edge glenoid implant liftoff at the bone-implant interface. Various recent follow-up studies have reported on glenoid loosening after TSA. Radiolucent lines are a radiographic finding consistent with component loosening. At 2 to 10 years of follow-up, radiolucent lines are present in 30% to >75% of TSAs, whereas gross implant position shift with clinical failure requiring revision is found in 2% to >10% of TSAs. Published rates of loosening emphasize the need for improvements in glenoid component design.

Compared with the traditional onlay glenoid design, a glenoid prosthesis that is inset, or an “inlay design,” may possess better biomechanical characteristics with respect to mechanical loosening. It may exhibit less gross loosening because of implantation in the native glenoid within a bone socket. To our knowledge, there is only one widely commercially available inlay glenoid prosthesis. The purpose of this study was to compare 2 available TSA systems with similar published indications for the treatment of glenohumeral degenerative joint disease and osteonecrosis. We focus on the loading characteristics and resistance to loosening of a traditional onlay glenoid implant compared with an inlay glenoid implant. We hypothesized that in this matched pair cadaveric TSA model, the inlay glenoid implant will exhibit greater resistance to loosening and superior biomechanical characteristics compared with the onlay glenoid implant.

Materials and methods

Specimens

Eight fresh frozen, male, cadaveric shoulder matched pairs were thawed and prescreened grossly and with high-resolution computed tomography to rule out pre-existing bone disease that could potentiate prosthetic failure. The mean age of the donors was 59.86 years (range, 54-65 years). Each shoulder was dissected free of all soft tissues, leaving the glenoid labrum and bone stock intact. All scapulae were prepared identically, with removal of the coracoid process, acromion, and inferior angle to allow accommodation within resin cement fixation blocks. Humeral shafts were potted in metal fixation cylinders, such that the superior extent of resin cement was approximately 6 cm below the inferior articular margin. The specimens were frozen and stored in a standard freezer at an average temperature of −18°C.

Arthroplasty procedures

All component implantations were performed by a single, experienced orthopedic surgeon, exactly according to manufacturers’ guidelines and published techniques. All backside pegs were cemented: all cement was pressurized and allowed to fully mature before any testing. The right-sided shoulders were implanted with all-polyethylene pegged onlay glenoids (size, 46 mm) with backside cement fixation and humeral (46 × 16-mm neutral head) component (Turon system; DJO Global, Vista, CA, USA). The left-sided shoulders were implanted with all-polyethylene round, pegged inlay glenoids (size, 20 mm) with backside cement fixation and uncemented humeral (head size range, 48 × 44 to 56 × 52 mm) components (Ovo system; Arthrosurface, Franklin, MA, USA), with head sizes determined by humeral head bone anatomy per the company’s recommended sizing methods.

Testing procedure

An experimental glenohumeral loading model and 2 testing protocols were created with a materials testing machine (Instron Model 8874; Instron, Norwood, MA, USA) using a 1 kN load cell (Dynergy; Instron) and cycling software (WaveMatrix; Instron) (Fig. 1). Our testing methods and TekScan protocol were adopted and tailored from prior similar studies. Glenohumeral contact area (A) and pressure (P) were measured using a digital pressure sensor (Model 5051; TekScan, Inc., South Boston, MA, USA) placed on the glenoid articular surface (Fig. 2). Sixty degrees of glenohumeral abduction in the scapular plane and neutral rotation was used for all testing as this position has been shown to be able to successfully quantify glenohumeral joint kinematics. A 2-point method was performed, using 4.5 and 13.6 kg of compressive force, to calibrate the TekScan sensors for each specimen. The center starting point for each glenoid was set using TekScan output mapping, ensuring that the starting point was equidistant from each opposing glenoid edge. For all 3 protocols, the humeral component was cycled on a stationary glenoid component, translating 5 mm in an anterior and posterior direction from the centered starting point along the y-axis (Fig. 3), yielding raw data. Table I provides a summary of our testing protocols. Joint compression forces were based on prior studies evaluating glenohumeral loads occurring with activities in wheelchair-bound patients. During testing protocol 1, preimplantation and postimplantation glenohumeral contact area and pressure were measured using 9.1 kg (88.9 N) of compressive force and shear at the glenohumeral articulation, within our defined zones of interest. Postimplantation testing was performed before fatigue testing. Anterior and posterior glenoid rim regions of interest were designated zones A through D (Fig. 4), where the glenohumeral articulation resulted in rim loading. Fatigue testing protocol 2 was used to assess for glenoid component gross loosening with 34 kg (333.6 N) of joint force.
Biomechanical testing

Contact area and pressure data were recorded at the glenohumeral articulation using the TekScan pressure sensor (Fig. 3). These measurements were performed on all specimens during 2 testing phases: (1) before implantation and (2) after implantation, before fatigue testing. TekScan pressure and area measurements were made at the anterior and posterior edges of both onlay and inlay intact shoulders (zones A1 and A2, respectively), the implanted onlay (zone B), and the inlay glenoids (zones C and D). Because of inlay implant architecture, simultaneous contact occurs at the peripheral glenoid native tissues (zone D) adjacent to the edge of the inlay glenoid implant (zone C). Finally, each specimen was fatigue tested for gross loosening, determined by video analysis, with a maximum set at 4000 cycles.

Data analysis

Mean pressure (P) and area (A) were determined for each cycle at the zones of interest, and mean values were further calculated from all cycles to achieve a final mean P and A for each specimen. Resultant force (F) was then mathematically calculated from the measured P and A for each specimen. Mean F values were then calculated for the inlay and onlay groups.

Mean resultant forces were evaluated and compared in 3 ways (Figs. 4 and 5):

1. Onlay preimplanted (intact) shoulder compared with implanted (prosthetic) shoulder—comparison of zones A1 and B.
2. Inlay preimplanted (intact) shoulder compared with implanted (prosthetic) shoulder—comparison of zones A2 to C and A2 to D.
3. Comparison of implanted inlay and onlay shoulders—comparison of zones B, C, and D.

Standard error and 95% confidence intervals were calculated, when possible, for A, P, and F. Mean number of cycles to gross loosening for the inlay and onlay shoulders during fatigue testing were calculated and compared. Statistical analysis was based on a 2-factor analysis of variance with terms for zone (A1, A2, B, C, D) and donor. The donor term was included for correction of observations within a donor. If the zone term was significant, Fisher least significant difference Student t-test was used to make specific zone comparisons. Statistical significance was set at \( P < .05 \).
Results

Figure 5 and Table II summarize our results. One onlay shoulder was removed from the study after native glenoid testing and before any implant testing because of damage during handling and gross implant loosening. After implantation, glenohumeral implant contact area decreased ($P < .0001$) and contact pressure increased ($P < .0001$) compared with the matched, intact shoulder for inlay and onlay study arms (zones A1 vs. B and A2 vs. C). The mean contact area at the onlay implant edge (39.81 mm$^2$) zone B was not statistically different ($P = .2240$) from the inlay implant edge (27.47 mm$^2$) zone C. The pressures were greater at the inlay implant edge zone C (2.682 MPa) and at the onlay implant edge zone B (3.286 MPa) than the pressure (0.641 MPa) at the intact peripheral glenoid rim zone D ($P < .0001$). This pressure (0.641 MPa) at the inlay zone D was not statistically different ($P = .9236$) from the pressure (0.673 MPa) at the intact shoulder zone A$_2$.

Mean force at the intact onlay shoulder zone A$_1$ increased from 85.7 N to 124.8 N at the onlay implant edge zone B ($P = .0012$). Force at the intact inlay shoulder zone A$_2$, 91.8 N, was not significantly different from force at the inlay implant edge zone C, 73.3 N ($P = .0836$). Force at the intact inlay shoulder zone A$_2$ significantly decreased from 91.8 N to 20.9 N at the inlay peripheral glenoid rim zone D ($P < .0001$).

Force of 124.8 N found at the onlay implant edge zone B was greater than the force of 73.3 N at the inlay implant edge zone C ($P < .0001$). There was a greater force found in comparing either implant edge (zones B and C) to the peripheral glenoid rim surrounding the inlay implants (zone D); zone C (73.3 N) was greater than zone D (20.9 N; $P < .0001$), and zone B (124.8 N) was greater than zone D (20.9 N; $P < .0001$).

During fatigue testing, all onlay glenoid components exhibited gross loosening at a mean number of cycles (1126; range, 749-1838). None of the inlay glenoid components exhibited gross loosening at the predetermined maximum of 4000 cycles. This difference in cycles to gross loosening between inlay and onlay glenoid implants was statistically significant ($P < .0001$).

Discussion

Glenohumeral component articulation plays a key role in mechanical loosening of an implant.$^{1,13,18,22,27}$ Humeral translation and eccentric glenoid loading occur with normal shoulder range of motion.$^{10,12}$ The rocking-horse mechanism of loosening is a frequently cited cause of glenoid implant loosening.$^{1,7,18,21}$ In this study, we chose anterior to posterior humeral translation as it represents glenoid implant edge load that could occur with high-demand activities, such as bench press, and could reproduce a rocking-horse phenomenon in the laboratory. In this study, we observed a dramatic increased resistance to loosening in the inlay implants compared with the onlay implants under the same fatigue-testing conditions. Our results suggest that an inlay glenoid design resists loosening better than an onlay design, with statistical significance ($P < .0001$). In addition, our biomechanical analysis revealed differ-
ences in the area, pressure, and force that occur at the implant articulations, supporting our observations.

Increased component conformity results in increased load transfer to the glenoid component and its backside fixation to bone.\textsuperscript{1,13,22} Periprosthetic radiolucent lines can be a sign of loosening and are more common if the radius of curvature (ROC) of the 2 components is more similar,\textsuperscript{27} also known as a smaller ROC mismatch. Significantly fewer radiolucent lines occur if the ROC mismatch is \textgreater 6 mm. In all cases, onlay and inlay prosthetic combinations were implanted within published and accepted manufacturer guidelines. To limit cost, we used a single humeral component for the onlay side of the study. This was possible because after the humeral head resection, any size humeral head component could be

\begin{figure}[h]
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\includegraphics[width=\textwidth]{figure5.png}
\caption{Mean glenohumeral resultant force and comparisons of zones of interest, with \(P\) values included for the various comparisons. The error bars represent standard error. Zone comparisons (change in force, \(\Delta F\)): onlay zone A\textsubscript{1} – B: 85.7 N – 124.8 N, \(\Delta F\) (\(P = .0012\)); inlay zone A\textsubscript{2} – C: 91.8 N – 73.3 N, \(\Delta F\) (\(P = .0836\)); inlay zone A\textsubscript{2} – D: 91.8 N – 20.9 N, \(\Delta F\) (\(P < .0001\)); inlay zone C – D: 73.3 N – 20.9 N, \(\Delta F\) (\(P < .0001\)); onlay zone B – inlay zone C: 124.8 N – 73.3 N, \(\Delta F\) (\(P < .0001\)).}
\end{figure}

\begin{table}[h]
\centering
\caption{Prefatigue testing results}
\begin{tabular}{|l|c|c|c|c|c|c|}
\hline
\multicolumn{2}{|c|}{Onlay} & & & & & \\
\hline
Preimplantation glenoid (zone A\textsubscript{1}) & Mean contact area (mm\textsuperscript{2}) & 123.09 & 9.54 (SE) & 0.716 & 0.063 (SE) & 85.7 & 7.32 (SE) \\
Postimplantation implant edge (zone B) & 39.81 & 6.11, 3.12 & 3.286 & 0.675, 0.34 & 124.8 & 17.19, 8.77 \\
\hline
Inlay & & & & & & \\
Preimplantation glenoid (zone A\textsubscript{2}) & Mean contact area (mm\textsuperscript{2}) & 138.36 & 6.82 (SE) & 0.673 & 0.056 (SE) & 91.8 & 6.61 (SE) \\
Postimplantation implant edge (zone C) & 27.47 & 4.21, 2.15 & 2.682 & 0.662, 0.34 & 73.3 & 17.52, 8.94 \\
Postimplantation glenoid rim (zone D) & 34.79 & 11.27, 5.75 & 0.641 & 0.391, 0.20 & 20.9 & 10.31, 5.26 \\
\hline
\end{tabular}
\end{table}

\textit{CI}, confidence interval; \textit{SE}, standard error.

Note: Figure 5 and the text provide specific zone comparisons and statistical significance.
implanted on the humeral stem. On the inlay side, the humeral cap was dependent on the native humeral head because it is a resurfacing component. Therefore, it was necessary to implant the appropriate-sized humeral cap to achieve adequate bone support of the prosthesis. We did ensure that ROC mismatch was always >6 mm in all situations and combinations. As a result, the inlay ROC mismatch ranged from 8 to 12 mm and the onlay ROC mismatch was 12 mm for all specimens. Therefore, the ROC mismatch for all inlay and onlay replaced shoulder specimens was >6 mm, thereby limiting any known risk of implant loosening simply due to ROC mismatch.

We produced an artificial loading scenario representing potential in vivo conditions to compare stability of the onlay and inlay designs. Gross loosening was defined by video-documented gross motion at the implant-bone interface and measured as cycles to failure. Gross loosening was exhibited in all onlay shoulders with a mean of 1126 cycles (range, 749-1838), whereas none of the inlay shoulders exhibited gross loosening out to 4000 cycles. This type of fatigue loading and failure is not representative of actual daily in vivo activity as these studies rarely represent regular in vivo conditions. It is intended to model a possible load condition and then to repeat this condition until fatigue occurs, allowing accelerated comparison of the 2 implants. These widely differing results may be partially explained by increased bone support and improved cement fixation around the inlay implant. Because of its inset position, cement fixation and bone support exist at the backside of the inlay implant, similar to the onlay design, but also occur circumferentially at the peripheral margin of the implant. The additional bone support and cement fixation present with the inlay components may explain some of the improved resistance to loosening. Finally, because this is a cadaveric study, any contribution to stability and survivorship due to bone ingrowth or remodeling could not be assessed.

Implant size may further explain the differences in loosening. An inlay implant must be smaller than the native glenoid articular surface to be inset, whereas the onlay implant is designed to cover the entire glenoid articular surface. The anterior-posterior diameter of the inlay implant is 20 mm compared with the onlay’s 26.9 mm. Therefore, as the humeral component is translated to implant edge, greater moment arm and torque exist for an onlay vs. an inlay implant. Furthermore, load sharing occurs with the inlay prostheses as the humeral head component articulates simultaneously with the glenoid component and the native tissues surrounding the glenoid component, when it is translated to the periphery. Therefore, only a fraction of the joint reaction force is applied to the inlay implants, whereas all of the force is applied to the onlay implants. These inherent design characteristics may further explain some of the loosening differences observed in our study.

Our biomechanical results support the gross loosening differences observed between these 2 commercially available glenoid designs. Consistent with prior studies and as expected, pressure increased and area decreased at the glenohumeral articulation after prosthetic replacement, as the metal-plastic articulation results in load over a smaller area than in the native shoulder. No statistical difference was found between inlay and onlay implant edge contact pressures. Pressure in the intact shoulder was not different from pressure at the peripheral glenoid rim of native tissues in the implanted inlay shoulders. Therefore, it appears that the native tissue surrounding the inlay implant experiences pressure similar to the intact state. Pressure was greater at the implant edge of both inlay and onlay components compared with the glenoid rim surrounding the inlay component. This concentration of forces at the implant edge of both designs illustrates eccentric loading and the potential for rocking-horse loosening. However, in this study, the inlay component was able to withstand such pressure without gross loosening, whereas the onlay component loosened.

The measured pressure (P) and area (A) were used to calculate reaction force (F) at the prosthetic articulation (F = P × A). Reaction force provides a comprehensive assessment of potential displacement of the glenoid prosthesis. Because of implant articular surface concavity, it is important to understand that the biomechanics associated with eccentric glenoid loading will be governed by an inclined plane physics model. Therefore, force in addition to the applied 88.9 N is imparted to the system as the humeral head translates from center to edge of implant and up the inclined plane. In the inlay shoulders, a mean force of 73.3 N was found at the implant edge zone C and 20.9 N at the peripheral glenoid rim zone D, illustrating load sharing between implant and surrounding native tissue. The sum of these 2 related forces results in 94.2 N, which is statistically similar to the 91.8 N found in the intact shoulder. In contrast, the force for the onlay intact shoulder significantly increased from 85.7 N to 124.8 N at the onlay implant edge zone B (P = .0012). In addition, onlay implant edge force of 124.8 N was significantly greater than the inlay implant edge force of 73.3 N (P < .0001). This increased force found at the onlay implant edge as a result of inclined plane phenomena may further explain the increased loosening found in the onlay glenoids.

Shoulder hemiarthroplasty has resulted in a higher risk of pain than with TSA, most often attributed to metal articulating with native glenoid tissues. The inlay TSA results in some metal-head articulation with native glenoid and labral tissues and therefore has the potential to represent a source of in vivo pain in this implant type. In our laboratory setting, the force experienced at the peripheral glenoid rim of the inlay prosthesis zone D was less than the force found in the intact shoulder at the same position zone A2, and the pressure was unchanged at zones A1 and D before and after inlay implantation. These findings may have important clinical implications for the inlay design because it inherently has some metal articulation with native tissues. The decreased force and unchanged pressure observed in the inlay specimens may suggest a mitigated risk for development of this hemiarthroplasty effect of pain at the native tissues surrounding an inlay glenoid component.
Limitations of our study include the cadaveric nature of the study and the lack of shoulder soft tissue structures that could alter the biomechanics of in vivo TSA. Because of the high costs of implants and the need for multiple implants to achieve statistical power, we chose to test only 1 onlay glenoid that we believe represents a standard and commonly used onlay pegged glenoid design. We compared this onlay glenoid implant to the only commercially available inlay component in the United States. This study could be strengthened and future studies could be improved by testing multiple onlay glenoid designs. Per the published techniques for the 2 implants and as a result of the inherent design differences, the inlay glenoid was surrounded by native labrum, whereas the onlay was not. In addition, because of inherent differences between the 2 humeral implants, we were able to achieve significant cost savings by using a single onlay humeral head component but were forced to use varying inlay head sizes. The presence of the labrum and variable head sizes on the inlay side may have contributed minor force variations or other variations to the testing conditions; however, given the great difference found between the 2 implant types in biomechanical testing and cycles to failure, we believe there would be no change in final results and conclusions. In addition, it must be emphasized that all implantation techniques were performed within the respective guidelines for each implant.

Prior related studies have used a more specific and defined digitized method to determine glenohumeral center point than our method, using TekScan mapping data. In our case, we believe our centering method is sufficiently accurate for this comparative testing. We attempted to limit any internal error and variation by using matched pairs (ie, left and right shoulders from the same cadaver). We did not randomize inlay and onlay to right or left cadaveric shoulder. We did place great emphasis on careful prescreening of each shoulder pair with computed tomography scans and gross anatomic inspection before any testing to rule out any bone defects or structural differences that could contribute to implant instability. In addition, hand dominance information was unavailable. We therefore chose a convention of assigning the inlay to the left-side and the onlay to the right-side scapulae to simplify our technique. This study and future studies could be further optimized by randomization.

**Conclusion**

In this cadaveric shoulder arthroplasty study, the inlay glenoid design showed results superior to the onlay glenoid design with respect to biomechanical characteristics consistent with loosening. The inlay also exhibited greater resistance to gross loosening during fatigue testing. Given the results of this study, we suggest that future research and development be directed toward inlay glenoid prosthetic design, matching native anatomy and size variation and potentially improving component survivorship.

**References**


