Joint stability after total shoulder arthroplasty in a cadaver model

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A cadaver model was used to test the hypothesis that glenohumeral joint stability is independent of articular surface conformity after total shoulder arthroplasty. For the purposes of this study joint stability was defined as the minimum force required for joint dislocation. After arthroplasty components were implanted into fresh-frozen glenohumeral joints, specimens were mounted on a load frame and tested for joint stability. For each specimen the amount of conformity between the articular surfaces was varied from 0 to 5 mm by changing the humeral head radius of curvature. Because the glenoid component was not changed, the wall height, or joint constraint, was maintained constant for a given specimen. Variations in joint conformity changed dislocation forces by an average of only 3%. These small differences are not believed to be clinically relevant, indicating that design changes affecting the joint conformity of a total shoulder arthroplasty system will not significantly affect glenohumeral joint stability, assuming that all other factors remain constant. (J Shoulder Elbow Surg 1997;6:506-11.)

The reported incidence of glenohumeral instability after total shoulder arthroplasty varies, ranging from 0% to 22%.[18] Clinically, joint instability may be related to implant design and surgical techniques including component alignment and soft tissue balancing and repair.[13] Instability may result shortly after implantation or may develop over time as a result of component damage.[12, 18] Short-term complications from instability include loss of shoulder function and pain.[11] Long-term complications may involve consistent superior translation of the humeral head component especially in those patients with damaged rotator cuffs.[2] This translation may be associated with an increased incidence of glenoid component loosening.[3]

Two design parameters of a total shoulder arthroplasty system have been proposed as factors affecting glenohumeral joint stability: joint conformity and joint constraint. Whereas joint conformity relates to the relative radius of curvature between the two articulating surfaces, joint constraint relates to the wall height of the glenoid component. With the birth of modern total shoulder arthroplasty in the 1970s, two distinct designs were proposed. One design replaced the natural anatomy with a very shallow glenoid component (Figure 1, *A*). It was postulated that this so-called unconstrained implant would reproduce the humeral head translations of the natural joint.[8, 14, 19] An alternate design involved creating a deeper socket for the glenoid component not unlike the acetabulum of the hip joint (Figure 1, *B*). This so-called constrained implant was designed to prevent joint dislocation by restricting humeral head translations.[9, 15]

Although the amount of constraint (wall height) of these early implants varied, all were considered conforming designs because the radius of curvature of the glenoid and humeral head components was the same (Figure 1, *A, B*). Some recently introduced unconstrained arthroplasty systems incorporate a glenoid component that has a greater radius of curvature than its mating humeral head component (Figure 1, *C*). The rationale behind

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these so-called nonconforming implants is to reproduce natural glenohumeral kinematics more accurately than traditional conforming implants.

Several investigations have studied the effects of conformity and constraint on glenohumeral joint stability in a cadaver model. These studies measured the minimum force necessary for joint dislocation. Fukuda et al.[4] found that the resistance to subluxation increased with increased component curvature. Although this curvature was not defined, it is presumed to represent the glenoid arc length, or the constraint of the component. With a similar model Severt et al.[16] studied the effects of both joint constraint and joint conformity. The latter term refers to the relative radii of curvature of the glenoid with respect to the humeral head component. Like Fukuda et al., they concluded that higher joint constraint leads to higher subluxation forces. They also concluded, however, that higher joint conformity leads to higher subluxation forces.[16] This result is in contrast to rigid body modeling of the glenohumeral articulation, which predicts that joint stability is independent of joint conformity.[6]

The goal of this study was to test the hypothesis that joint stability, or dislocation force, is independent of joint conformity. Experiments were conducted on an MTS load frame (MTS Systems Corp, Minneapolis, Minn.) with cadaver specimens implanted with arthroplasty components. To eliminate any confounding effects of joint constraint, this parameter was held constant throughout the experiments.

MATERIAL AND METHODS

Seven fresh-frozen human glenohumeral joints (mean age 72 years, range 54 to 88 years) were obtained from a previous study of implant kinematics.[7] Each glenoid was obtained from a separate cadaver and was implanted with a keeled glenoid component with a nominal radius of curvature of 25 mm. All muscles and ligaments were stripped from the specimen, and the humerus was discarded. The scapula was cut at the level of the scapula notch, and the acromion was also removed. The specimen was potted with Bondo car body filler (Dynatron/Bondo Corp., Atlanta, Ga.) in a metal cylinder with a 3-inch outer diameter. To reduce specimen bending the scapula was potted to the level of the glenoid neck.

A 1.5-inch diameter aluminum cylinder was machined with a reverse Morse taper on one of its faces for attachment of humeral head components. This cylinder replaced the humerus in supporting the humeral articular surface. Six modular humeral head components were used. These components had nominal radii of curvature of 25, 24, 23, 22, 21, and 20 mm. In combination with the 25 mm radius of curvature glenoid component, this yielded a total of six articulations for each specimen, with radial mismatches of 0, 1, 2, 3, 4, and 5 mm. Each humeral head component was oriented so that the longitudinal axis of the taper was perpendicular to the face of the glenoid component, which represents a position of approximately 45° of glenohumeral elevation in the scapular plane in neutral rotation.

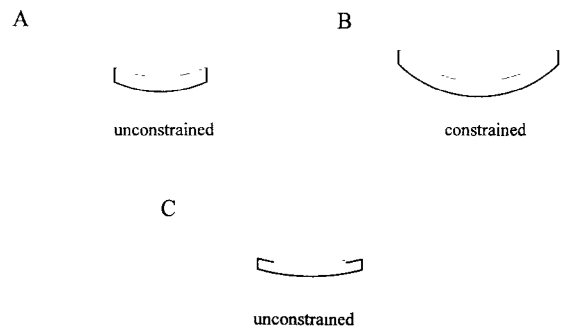
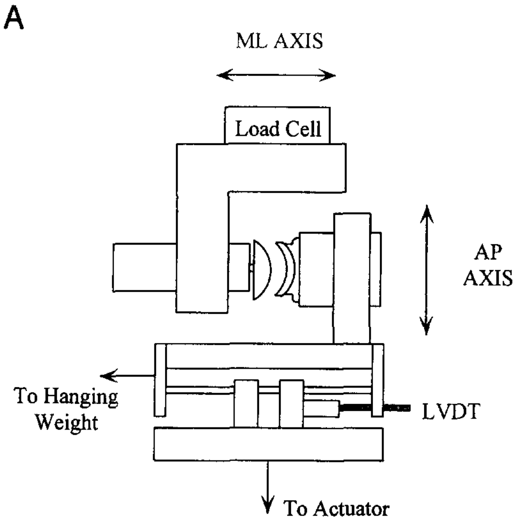


Figure 1 Total shoulder arthroplasty designs. **A** and **B,** Original conforming design. **A,** Standard unconstrained implant with conforming surfaces. **B,** Constrained implant (increased glenoid wall height), also with conforming surfaces. **C,** Unconstrained implant with nonconforming surfaces

The stability testing apparatus consisted of a biaxial translation table mounted to the actuator of an MTS 312 load frame . This table allowed for translations orthogonal to the axis of the MTS actuator. The glenoid was clamped onto the table and aligned so that the medial/lateral and superior/inferior (SI) axes were in-line with the table axes, and the anterior/posterior (AP) axis was in-line with the actuator axis. The 1.5-inch aluminum cylinder was clamped to the MTS load frame so that the humeral head components were in full contact with the glenoid component (Figure 2).

AP translations were controlled by the MTS and monitored with an internal linear variable differential transformer (LVDT) and medial/lateral translations were unconstrained and monitored with an LVDT (Transicoil, Inc, Valley Forge, Pa.) mounted beneath the translation table. AP forces were measured by a load cell (Lebow, Troy, Mich.) that was fixed to the MTS frame above the humerus. A constant medial joint centering force was maintained during each test

A



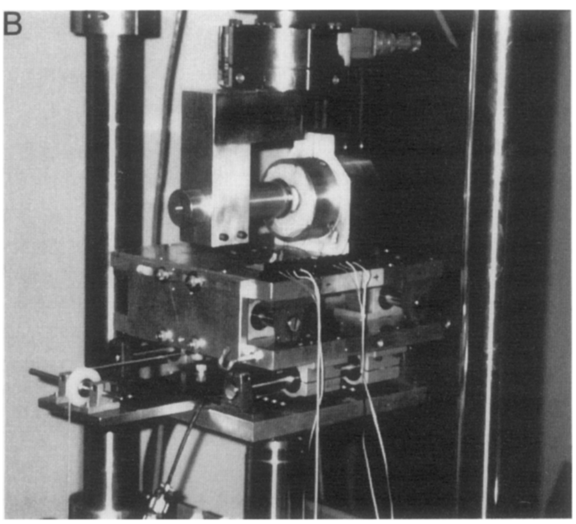


Figure 2 MTS Experimental setup. Anterior/posterior *(AP)* translations were controlled by actuator, medial/lateral *(ML)* translations were unconstrained, and superior/inferior *(SI)* translations were constrained **A,** Schematic representation of apparatus showing actuator axis and only one table axis shown. Other table axis (SI axis) has not been included in drawing for simplicity. **B,** Photograph of apparatus Wires shown in this picture were used for another study

by means of a pulley and hanging weight system. Experiments were conducted with only 2 degrees of translation freedom, because all rotations and SI translations were constrained.

The humeral head was centered in the glenoid cavity by adjusting the AP and SI translations until the head was in its most medial location, which represented the deepest portion of the glenoid socket. The table was locked for translations along the SI axis in this position. The joint was translated posteriorly to its starting position, and a constant medial load (either 10 N or 400 N) was applied with a hanging weight. An anterior translation was applied, followed by a posterior translation. The experimental protocol consisted of testing for both anterior and posterior stability with all six radial mismatches in a random order. For all specimens experiments were first performed with a 400 N medial load, which approximated half body weight, and were then repeated with a nominal load of 10 N.

Although data from a single experiment provide a continuous measurement of the force-displacement relationship, only the maximum recorded transverse (anterior or posterior) force was used for data analysis. These values were used to calculate the stability ratio (maximum transverse force/medial force) for each experiment, which serves as an indication of relative joint stability.[4]

Systat 5.2 for the Macintosh (Systat, Evanston, Ill.) was used for statistical analysis.[1] A repeated measure analysis of variance was performed with two within subject factors: radial mismatch (0 to 5 mm) and medial load (10,400 N). The acceptable rate for a type I error was set at 5% (p = 0.05).

RESULTS

Changes in radial mismatch were found to have a dramatic effect on the transverse force-displacement curve. As the radial mismatch decreased so that a joint became more conforming, the force-displacement relationship became steeper (Figure 3). The maximum force, however, appeared to remain constant. These patterns were observed in all specimens, at both force levels.

The stability ratio at 400 N for all experiments was 0.43 ± 0.02 mm (mean ± SD). This ratio increased to 0.51 ± 0.03 when the medial load was reduced to 10 N ($p < 0.001$). This increase was observed for each joint conformity tested (Figure 4).

When the different joint conformities were accounted for, the mean stability ratio at 400 N ranged from 0.42 to 0.45 anteriorly and from 0.40 to 0.43 posteriorly. At 10 N the range was 0.51 to 0.55 anteriorly and 0.47 to 0.52 posteriorly (Figure 4). For all of these cases joint conformity was found to have a statistically significant influence on the stability ratio ($p < 0.01$, all cases). It can be observed, however, that these differences were small. For all four cases the average group deviance from its experimental group mean was 3%, and in no case did it exceed 7%.

DISCUSSION

Theoretic predictions indicate that the minimum force required for joint dislocation is independent of joint conformity, depending instead on the glenoid radius of curvature and wall height and on the coefficient of friction between the articulations.[6, 17] Altering joint conformity therefore should simply change the slope of the force-displacement curve without changing the maximum force experienced. A less conforming joint is predicted to experience greater translations for a given force. These calculations assume that both the humeral head and glenoid components are rigid bodies.

By varying head sizes for the same glenoid component, this study eliminated wall height and glenoid radius of curvature as variables. As predicted, experimental results indicate that as joint conformity increases, the slope of the force-displacement curve becomes steeper (Figure 3). However, component conformity had a very small (3%) but statistically significant effect on the mean stability ratio, or relative forces required for dislocation. The small observed difference may have been due to slight changes in the glenoid wall height resulting from the head not being perfectly centered in the glenoid. Also, the mean stability ratio was less when the 400 N medial load was tested. This effect is presumably due to deformation of the polyethylene glenoid components.

Fukuda et al.[4] measured subluxation forces in components from various manufacturers. They restricted their analysis and discussion to the issue of joint constraint, which is controlled by wall height and not by joint conformity. They also reported that the minimum normalized forces required for dislocation were decreased for high medial loads, which is in agreement with this study. Severt et al.[16] performed a similar study to that of Fukuda et al. but measured both joint conformity and constraint in the components they were studying. Because they used only one head size for each glenoid, they were not able to independently control for changes between these two parameters. Subsequently, their analysis often combined joint conformity and constraint, and their conclusions made no distinction between the two. In contrast, this study varied only joint conformity and kept joint constraint a constant by using the same glenoid component for multiple head sizes.

Although the stability ratio is relatively independent of radial mismatch, less conforming joints experienced these forces at greater translations, as noted by the curves in Figure 3. Other definitions of joint stability such as joint stiffness or energy for dislocation therefore might indicate a difference between joints with different conformities. However, these definitions would probably be more of an indication of how joints behave for smaller translations not leading to joint dislocation. For example, because the contributions of glenohumeral ligaments to joint stability tend to be greater at larger translations,[10] by allowing more translation nonconforming implants may help to distribute potentially harmful forces more evenly between component and ligaments. This may serve as a mechanism to help reduce component forces without sacrificing stability.

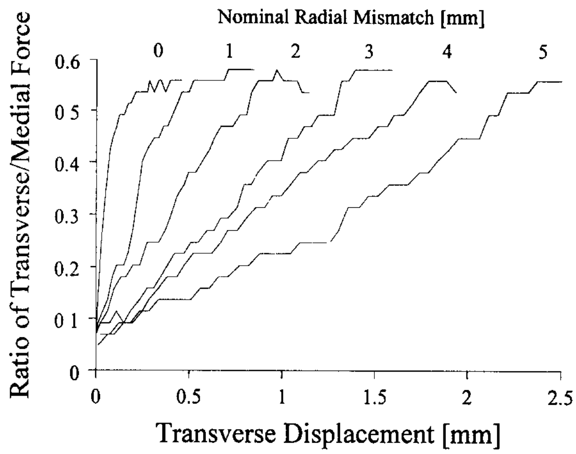
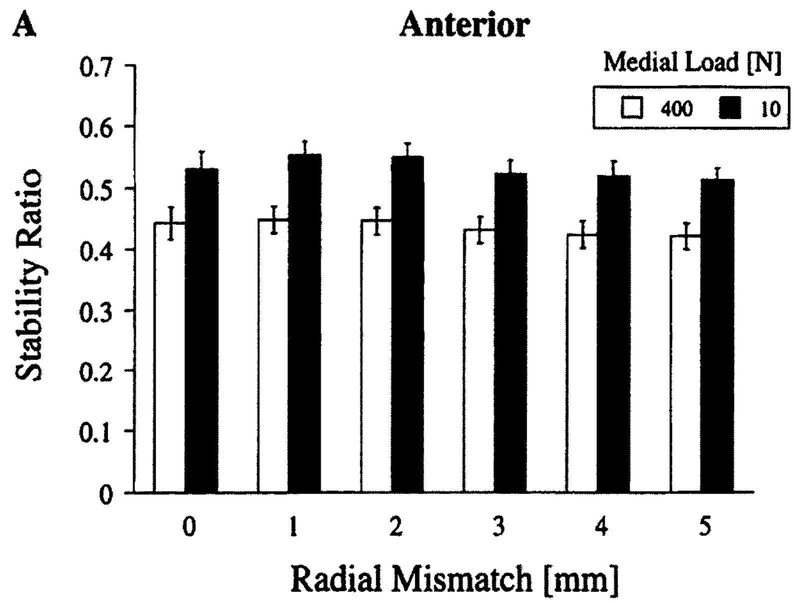


Figure 3 Representative plots of transverse/medial loads versus transverse displacement. Data are from one specimen for all radial mismatches. Note that although slopes of curves are different, maximum values (stability ratio) are the same.

Previous kinematic studies performed in our laboratory have demonstrated that as prosthetic joints become less conforming, larger translations are observed during active joint positioning.[7] The current study supports the concept that more translation may be expected for less conforming articulations in cases where the translations do not cause joint subluxation such as active motions. However, despite offering less initial resistance to translation, less conforming joints offer the same resistance to joint dislocation as more conforming joints. This may explain why the kinematic study of Harryman et al.[5] found no significant influence of joint conformity on translations during passive motions, because the translations in that study were large enough to cause joint subluxation.

To isolate the effects of joint conformity, no soft



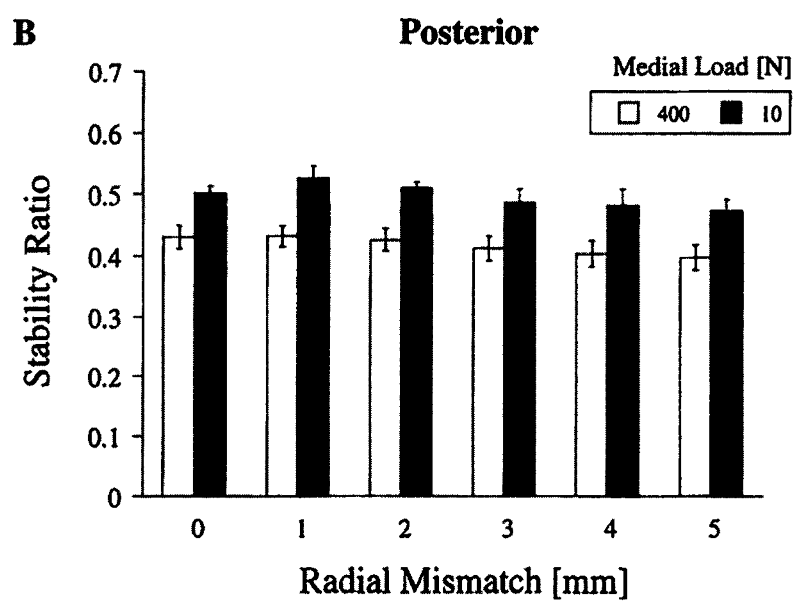


Figure 4 Stability ratio as a function of radial mismatch. Data are for medial loads of 400 N and 10 N. Data are shown as mean ± SD for **A,** anterior translations and **B,** posterior translations.

tissues were included in this study. It was assumed that the contributions of the ligaments, muscles, and intraarticular pressure would not change with changing curvature. It is possible, however, that at the extremes of motion, ligament loads, muscle loads, and the intraarticular pressure will change depending on component geometry. This effect was not tested in this study.

The experimental protocol in this study tested only one loading regimen representing two-dimensional translations only. It was designed in this manner for simplicity and could easily be extended to off-axis loading in three dimensions in the future. Experiments with a medial load of 400 N were always performed first in this study. Because the stability ratio always increased when this load was reduced to 10 N, it is clear that at least some of the polyethylene deformation caused by the 400 N load was not permanent.

This study only deals with the effects of joint conformity on joint stability. Decreasing component conformity also results in decreased contact area and increased contact stresses. However, forces across the glenohumeral joint are much smaller than those acting at the hip and knee, and polyethylene wear has not been a clinical problem in total shoulder arthroplasty. Additional work and long-term data are required to fully define the specific effects of wear and joint stability on implant failure.

CONCLUSIONS

For a given translation the more conforming the joint was, the higher the forces observed were. The minimum forces necessary for dislocation, however, were found to be relatively independent of joint conformity. On the basis of these results and the definition of stability in this study, nonconforming joints would not be at a higher risk of joint dislocation than conforming joints for the simplistic loading conditions tested.

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REFERENCES

1 Anonymous SYSTAT Statistics, Version 5 2 Edition Evanston, IL SYSTAT, Inc , 1992

2 Boyd AD, Aliabadi P, Thornhill TS Postoperative proximal migration in total shoulder arthroplasty J Arthroplasty 1991, 6 31-7

3 Franklin JL, Barrett WP, Jackins SE, Matsen FA, III Glenoid loosening in total shoulder arthroplasty J Arthroplasty 1988, 3 39-46

4 Fukuda K, Chen C-M, Cofield RH, Chao EYS Biomechanical analysis of stability and fixation strength of total shoulder prostheses Orthopedics 1988,11:141-9

5 Harryman DT, H, Sidles JA, Harris SL, Lippitt SB, Matsen FA, HI. The effect of articular conformity and the size of the humeral head component on laxity and motion after glenohumeral arthroplasty J Bone Joint Surg Am 1995,77A:555-63

6 Karduna AR, Williams GR, Iannotti JP, Williams JL. The effects of component conformity in total shoulder arthroplasty theoretical and experimental observations Bioengineering Conference, Beever Creek, CO, 1995.439-40

7 Karduna AR, Williams GR, Iannotti JP, Williams JL Kinematics of the glenohumeral joint after total shoulder arthroplasty: effects of component conformity. Advances in Bioengineering, San Francisco, 1995 151-2

8 Kenmore PI, MacCartee C, Vitek B A simple shoulder replacement J Biomed Mat Res Sympos 1974;5 329-30.

9 Lettin AWF, Scales JT Total replacement of the shoulder Joint Proc R Soc Med 1972,65 373-6

10 Malicky DM, Soslowsky LJ, Blasier RB Anterior glenohumeral stabilization factors progressive effects in a biomechanical model J Orthop Res 1996,14.282-8

11. Matsen FA III, Thomas SC, Rockwood CA, Jr Glenohumeral instability In Rockwood CA, Jr, Matsen FA, III. The shoulder. Philadelphia WB Saunders Company, 1990 526-622

12 Miller SR, Bigliani LU Complications of total shoulder replacement In Bigliani LU Complications of shoulder surgery, Baltimore Williams and Wilkins, 1993 59-92.

13 Moeckel BH, Altchek D, Warren RF, Wickiewicz TL, Dines DM Instability of the shoulder after arthroplasty J Bone Joint Surg Am 1993,75A 492-7

14 Neer CS II Replacement arthroplasty for glenohumeral osteoarthritis J Bone Joint Surg Am 1974,56A 1-13

15 Reeves B, Jobbins B, Dowson D, Wright V. A total shoulder endo-prosthesis Eng Med 1972,1·64-8

16 Severt R, Thomas BJ, Tsenter MJ, Amstutz HC, Kabo JM. The influence of conformity and constraint on translational forces and frictional torque in total shoulder arthroplasty Clin Orthop 1993,292 151-8

17 Walker PS, Wolf B The control of anterior-posterior movemere in condylar replacement knee prosthesis Trans Orthop Res Soc 1978,3 153.

18 Wirth MA, Rockwood CA, Jr Complications of shoulder arthroplasty Clin Orthop 1994,307 47-69

19 Zippel J Vollstandiger Schullergelenkersatz ans Kunstoff und Metall Biomed Technik 1972,17 87