

## Posterior Glenohumeral Subluxation: Active and Passive Stabilization in a Biomechanical Model\*

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**ABSTRACT:** We examined the role of the glenohumeral and coracohumeral ligaments as well as the forces provided by the rotator cuff muscles, the long head of the biceps, the anterior and middle deltoids, and the pectoralis major in the stabilization of the glenohumeral joint in the posterior direction. Simulated muscle forces were mechanically applied to eight shoulder specimens. The humeroscapular position for testing simulated the 90-degree forward-flexion (humerothoracic) position used clinically for the so-called jerk test, which is the most clinically important position with regard to posterior instability of the shoulder. Experiments were performed with a variety of configurations of ligamentous and capsular cuts, humeral rotation, and levels of muscle force. Stability was investigated by measuring the force required to subluxate the humeral head a specified amount from its reduced position. Of the muscles and ligaments tested, the subscapularis muscle contributed the most to this subluxation force. The coracohumeral ligament was an effective contributor in neutral humeral rotation, and the inferior glenohumeral ligament was an effective contributor in internal humeral rotation. The long head of the biceps was found to reduce the subluxation force in certain positions.

**CLINICAL RELEVANCE:** It is widely agreed that a complex interaction of passive and active stabilizing structures and forces is necessary for clinical stability of the shoulder. The present study identified the contributions of ligaments and muscles to posterior stability of the shoulder in the position of greatest clinical importance — posterior subluxation with the shoulder in forward flexion.

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The glenohumeral joint has the greatest range of motion of any major joint in the body, but it also is associated with the highest prevalence of dislocation. Posterior instability is believed to be caused and exacerbated by forward flexion and internal rotation of the arm, with a posteriorly directed force aligned along the axis of the humerus. Many investigators<sup>3,9,21</sup> have quantified posterior instability as the counterpart to anterior instability, observing posterior subluxation of the humeral head with the shoulder in abduction. Others<sup>2,4,13</sup> have found posterior migration of the humeral head during flexion or abduction. Although some important questions have been addressed in those studies, we know of only two investigations in which the most clinically relevant condition of compression along the axis of the humerus during forward flexion of the shoulder was examined<sup>3,19</sup>. However, the role of specific active joint-stabilizing factors, such as muscles, was not examined in either study, and specific passive stabilizers, such as ligaments, were studied in only one<sup>3</sup>. We found only one previous study<sup>9</sup> in which such quantification was attempted with use of physiological joint-compression forces, but the role of specific muscle forces or ligamentous structures was not examined in that study.

The purpose of our study was to examine simultaneously the passive and active stabilizers of the shoulder in the most clinically relevant position. With use of a biomechanical model, we measured the contributions to subluxation force, through a range of posterior subluxations, of the superior and middle glenohumeral ligaments combined, the inferior glenohumeral ligament, the posterior aspect of the capsule, the coracohumeral ligament, and the tensile forces in the subscapularis, the supraspinatus, the combined external rotators, the long head of the biceps, the anterior and middle deltoids, and the pectoralis major. In addition to determining the over-all characterization of these factors, we sought to test four hypotheses: (1) all rotator cuff muscles contribute to the subluxation force in neutral or internal humeral rotation, (2) the long head of the biceps increases the subluxation force in neutral humeral rotation and reduces the force in internal humeral rotation, (3) the superior and middle glenohumeral ligaments contribute to the subluxation force in either rotation,

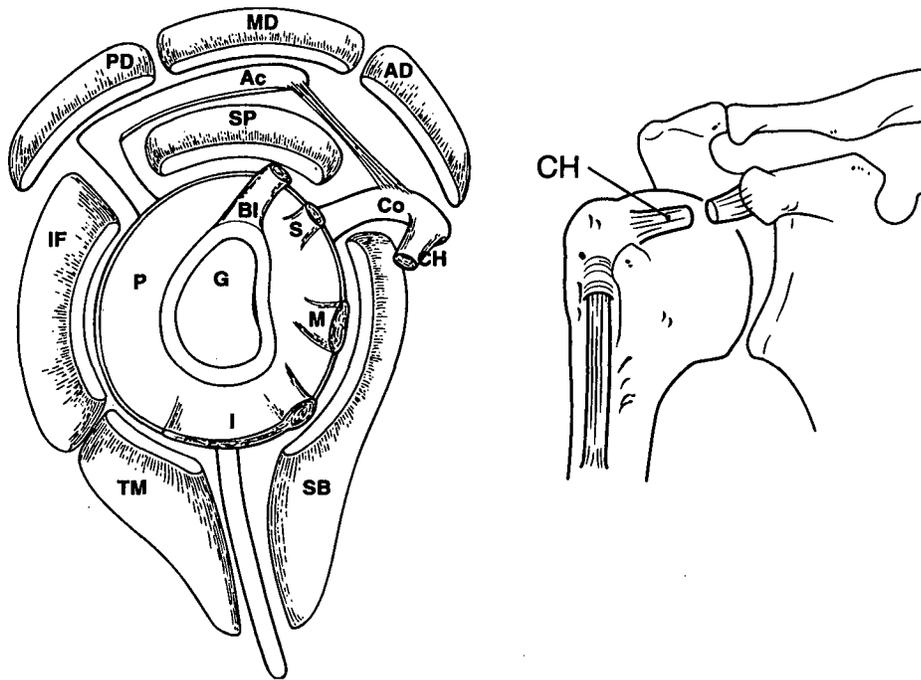


FIG. 1

Diagrams of the shoulder showing muscles, ligaments, and the zones defined for cutting. The diagram on the left is a lateral view of the shoulder socket as seen from the humeral head. The muscles shown are the posterior deltoid (PD), the middle deltoid (MD), the anterior deltoid (AD), the supraspinatus (SP), the infraspinatus (IF), the long head of the biceps (BI), the subscapularis (SB), and the teres minor (TM). The ligaments and capsule shown are the superior glenohumeral ligament (S), the coracohumeral ligament (CH), the middle glenohumeral ligament (M), the inferior glenohumeral ligament (I), and the posterior aspect of the capsule (P). Other structures include the acromion (Ac), the glenoid (G), and the coracoid (Co). The zones defined for cutting paralleled the ligamentous and capsular structures, except that the superior and middle glenohumeral ligaments were grouped into a single anterior zone.

and (4) the coracohumeral ligament increases the force in neutral rotation.

### Materials and Methods

#### *Specimens, Ligamentous Zones, and Muscles*

Eight normal shoulders were obtained from the cadavera of individuals who had been a mean (and standard deviation) of  $68 \pm 11$  years old at the time of death. Each specimen was dissected, leaving intact only the scapula, the humerus, the long head of the biceps tendon, the glenohumeral capsule and ligaments, the coracohumeral ligament, and the rotator cuff tendons at the sites of their humeral insertion. A specimen was considered normal if there was no tear or fraying of the rotator cuff, no defect of the rotator interval, and no degenerative change that could be detected by palpation through the remaining tissues of the rotator cuff and the capsule. Furthermore, we used only specimens that had a normal range of elevation and rotation, and smoothness of subluxation was ensured before testing. Because all of the specimens were to be tested before and after a ligamentous or capsular cut, all specimens were initially vented by a small stab incision in the rotator interval to eliminate the contribution to stability of intracapsular pressure as a confounding factor. The cuts were made in four zones (Fig. 1): the anterior zone (the superior and middle glenohumeral ligaments), the inferior zone (the inferior glenohumeral ligament), the posterior zone (the posterior aspect of the capsule), and the coracohumeral zone (the coracohumeral ligament).

All cuts were made from external to the joint. The coracohumeral zone was defined as the structure between the humerus and the coracoid. Thus, this cut was made from the tip of the coracoid to its base, taking care not to sever tissues near the glenoid rim or the coracohumeral ligament. The cut in the anterior zone was the most demanding technically. To avoid accidental cutting in the coracohumeral zone,

the cut was started anteroinferiorly, approximately twenty millimeters lateral to the glenoid, and was extended superiorly and then adjacent to the glenoid to avoid the confluence of the coracohumeral zone with the superior aspect of the capsule. The cut was continued until the superior glenohumeral ligament was sectioned. To further ensure that the coracohumeral ligament was not lacerated, the humerus was distracted to apply tension to the coracohumeral ligament, which was visualized and palpated throughout the cut in the anterior zone.

With use of servopneumatic cylinders, we modeled the forces of seven muscles (Fig. 1): the supraspinatus, the subscapularis, the infraspinatus and the teres minor as combined external rotators, the long head of the biceps, the anterior and middle deltoids (the posterior deltoid was not necessary for the selected position of forward flexion), and the clavicular portion of the pectoralis major.

#### *Testing Fixture*

Each scapula was reinforced internally by an injection of hot-melt glue into the scapular body to promote rigid fixation to the testing fixture (Fig. 2). A clamp was attached to each rotator cuff and biceps tendon, and each clamp was connected to a 3.2-millimeter braided Dacron cord that was used to apply the simulated muscle force. To simulate the forces of the deltoid and pectoralis major muscles, the Dacron cords were attached to the sites of insertion of these muscles on the humerus. Each cord was aligned through the approximate center of the cross section of the respective muscle with use of pulleys (for the rotator cuff tendons) or Teflon eyelets (for the deltoid muscles and the pectoralis major); all of the cords then were routed to low-friction pneumatic cylinders (CM2Q; SMC, Indianapolis, Indiana). The long head of the biceps was controlled with a similar pneumatic cylinder that was mounted directly on the humerus to avoid the application of artificial external forces. All cylinders were controlled by electropneumatic regulators (VIP; LDI Pneutronics,

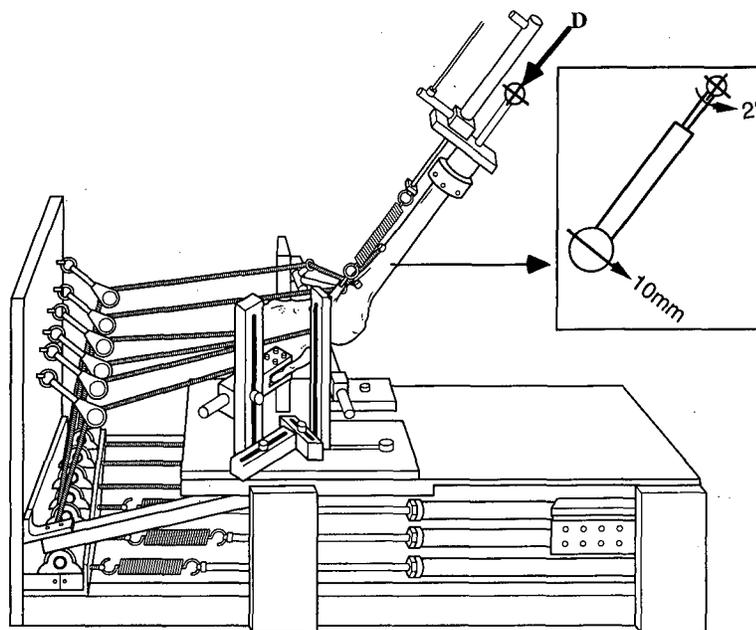


FIG. 2

Illustration of the experimental fixture, showing the scapular mounting (center), the direction of the humeral subluxation (D) and the biceps cylinder (top), the lines used to simulate the muscle forces (left), and the pneumatic cylinders (bottom). Subluxations were applied through a ball joint at the distal end of the humerus. The inset depicts the relationship between a rotational degree of freedom about the ball joint and the subluxation at the center of the humeral head. A similar rotational degree of freedom exists in a plane out of the paper (not shown for clarity). Ten millimeters of humeral subluxation produced only 2 degrees of rotation, which is considered to be clinically unimportant.

Hollis, New Hampshire), which facilitated fine control of the muscle forces. Friction was minimized throughout the muscle-force application system.

### Humeral Positioning and Definition of Standard Muscle Loads

The humerus was elevated to simulate the clinically important position of 90 degrees of forward flexion (humerothoracic), according to the published guidelines of a technique of active joint-positioning, which were based on electromyographic data, data on the cross-sectional area of muscles, and information regarding lever arms<sup>5,6,17</sup>. The humeroscapular position for the current study was described with use of the reference frame of Pearl et al. According to that frame, the plane of elevation is defined by the plane of the non-elevated and elevated humerus, relative to the plane of the scapula (positive values reflect anterior positions), and the angle of elevation is defined as the angle between the medial border of the scapula and the elevated humerus<sup>16</sup>. In the present study, the plane of elevation was  $39 \pm 5$  degrees and the angle of elevation was  $61 \pm 5$  degrees; we referred to this position as the 39,61 position. Initially, the humerus was in a position that simulated the arm at the side, near the position described by Pearl et al. as +.15, with ten newtons of nominal muscle forces. (The plane of elevation is not well defined for small angles of elevation; thus, according to the frame of Pearl et al., it is simply referred to as positive [+].) Muscle forces were progressively increased against a simulated weight of the arm to bring the humerus into the 39,61 position, with a minimum of forty newtons specified for each rotator cuff muscle. The biceps load was applied after flexion, as the long head of the biceps is not believed to be strongly involved in forward flexion of the shoulder. To simulate the weight of the arm, a twenty-five-newton load was applied to the humerus at the center of gravity of the upper limb<sup>20</sup>. This force was directed such that the vector for the weight of the arm was reproduced accurately at both the +.15 and the 39,61 position; however, intermediate positions may have been simulated less accurately, as little is known about scapular rhythm during forward flexion. The results of pilot studies involving the

shoulders of two subjects and from two cadavera, as well as anatomical data, were used to determine a final orientation of the humerus and scapula in which posterior instability was most reproducible and that represented the desired position of forward flexion. This 39,61 position is a less extreme version of the 51,75 so-called cross-body adduction position in the study by Pearl et al., and the 61-degree angle of elevation corresponds to the 60 degrees of forward flexion chosen by Harryman et al.<sup>2</sup>. The technique of active positioning of the joint resulted in the definition of a kinematically determined mean standard load (and standard deviation) for each active muscle force that was modeled. The standard load was  $62.0 \pm 8.1$  newtons for the supraspinatus,  $45.0 \pm 2.5$  newtons for the subscapularis,  $61.0 \pm 14.0$  newtons for the external rotators,  $65.0 \pm 12.0$  newtons for the anterior deltoid,  $72.0 \pm 9.7$  newtons for the middle deltoid, and  $36.0 \pm 8.0$  newtons for the pectoralis major. The standard load for the long head of the biceps was not defined by elevation and was set at 31.0 newtons on the basis of its relative cross-sectional area.

### Humeral Fixation and Degrees of Freedom

Once the position of 39,61 forward flexion had been achieved and the standard muscle force had been defined, the distal part of the humerus was fixed to a low-friction ball joint, which was later used to apply subluxation displacement. Humeral rotation was fixed in one of two defined positions by a tie-rod attached perpendicularly to the humeral shaft. Neutral rotation of the humerus was defined as the mid-point between full passive internal and external rotation<sup>18</sup>, to allow maximum laxity of the ligaments. Internal rotation of the humerus was defined as 10 degrees less than full passive internal rotation, to prevent damage to the ligaments from the application of subluxations that could cause excessive tightening of the ligaments. With four degrees of freedom fixed or defined (axial rotation and the three translations of the distal part of the humerus), the humerus had two remaining degrees of freedom. A greater number of degrees of freedom, while desirable from a physiological viewpoint, would have led to confounding results. Fewer than two degrees of freedom would have led to overestimation of the subluxation force and possibly dam-

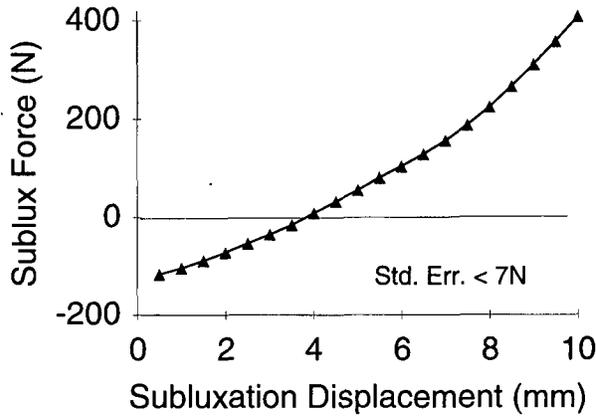


FIG. 3

Total-force-deflection curve for the glenohumeral joint in the standard configuration, in neutral rotation. The curve represents the average of all specimens. The curve for internal rotation was similar (not shown). The ordinate is the subluxation force that was measured when the posterior subluxation, shown on the abscissa, was applied. The standard error represents the largest value over all of the displacements.

age to the glenoid<sup>19</sup>. The large distance from the ball joint to the glenoid compared with the small amount of subluxation allowed substantial subluxation of the humeral head with only slight changes in the rotational position of the humerus (Fig. 2, inset). For example, ten millimeters of humeral subluxation (a clinically large amount) resulted in only 2 degrees of rotation (a clinically small amount). This 2-degree shift in position was not considered large enough to affect the results substantially.

*Measurements of Subluxation Forces*

Posterior stability of the shoulder was investigated by measuring the force necessary to subluxate the joint to a specified displacement<sup>1</sup>. A servohydraulic machine (MTS, Minneapolis, Minnesota) was used to apply the subluxation displacement through the ball joint to the distal end of the humerus, directed along the humeral shaft. The subluxation force was measured for displacements of zero to 10.0 millimeters. A single test cycle started at zero millimeters, subluxated posteriorly to 10.0 millimeters, reversed to -4.0 millimeters, and finally returned to zero millimeters. The rate of displacement throughout the cycle was 0.5 millimeter per second. This slow rate of subluxation was chosen to minimize viscoelastic effects. The displacement to -4.0 millimeters was performed to allow all data between zero and 10.0 millimeters to be collected with consistent hysteresis effects.

Each shoulder was tested with the capsule intact and then again after only one randomly chosen cut (in the anterior, inferior, posterior, or coracohumeral zone). To avoid the difficulties in analysis that are associated with the serial cutting of structures, no shoulder was tested after multiple cuts. Testing was performed in both neutral and internal humeral rotation and was done first with standard muscle forces and then with single, randomly ordered variations in muscle force of 0, 50, and 150 per cent of the standard load. Each test was performed twice. The combination of the aforementioned conditions required fifty-two tests, and each was repeated once, yielding a total of 104 tests. Each shoulder was preconditioned for ten cycles in each rotation. A pilot study had indicated that, after five to ten cycles of preconditioning, the test could be repeated more than 150 times without causing meaningful tissue fatigue. Preconditioning also was performed after each change in rotation and after cutting in each zone. Before any preconditioning, the standard muscle load was applied to identify the zero-displacement subluxation position.

We calculated the efficiency, or stability ratio, of each muscle (a normalization of the effect of a muscle on the subluxation force) by dividing the change in subluxation force by the change in muscle

force<sup>11</sup>. Changes in muscle force were calculated by the difference in its applied standard load, and changes in subluxation force were found with use of the statistical model (to be described). For example, with an increase in muscle force of 1.0 newton, an efficiency of 0.4 means that the subluxation force will be raised by 0.4 newton. A negative efficiency reflects a reduction in the subluxation force.

*Analysis of the Data*

The data were studied with use of multivariable regression analysis at a 5 per cent level. Partial F tests applied to the fitted regression model were used to evaluate the individual effects of the various independent factors on the dependent variable (the subluxation force). The regression model included variations among specimens, muscle forces, and ligamentous or capsular cuts. The results of testing in neutral and internal humeral rotation were analyzed separately. The data were analyzed at 0.5-millimeter increments for displacements of 0.5 to 10.0 millimeters.

**Results**

*Over-All Model*

The squared correlation coefficient for twenty regression models ranged from 0.83 to 0.98. Several significant and clinically interesting relationships were found.

*Total Force Deflection*

The over-all average subluxation force for the standard test case (an intact capsule and a standard muscle load) reached a peak of 407 newtons in neutral humeral rotation (Fig. 3) and 418 newtons in internal humeral rotation. The trend was similar for testing in each rotation, with a mild exponential increase as greater displacements were reached.

*Effects of the Muscles*

In neutral humeral rotation (Fig. 4-A), elimination of the simulated supraspinatus force resulted in a significant decrease ( $p < 0.007$ ) in the subluxation force at all displacements greater than 2.5 millimeters, elimination of the simulated subscapularis force decreased the subluxation force at all displacements ( $p < 0.018$ ), and elimination of the simulated external-rotator and biceps forces decreased the subluxation force at displacements greater than 4.0 and 3.5 millimeters, respectively ( $p < 0.020$  and  $p < 0.028$ ).

In internal humeral rotation (Fig. 4-B), elimination of the simulated supraspinatus force resulted in a significant decrease ( $p < 0.032$ ) in the subluxation force at all displacements greater than 4.0 millimeters, elimination of the simulated subscapularis force caused a decrease ( $p < 0.036$ ) at displacements greater than 1.5 millimeters, and elimination of the simulated external-rotator force resulted in a decrease ( $p < 0.030$ ) at displacements greater than 3.0 millimeters. When the simulated biceps force was eliminated, the subluxation force increased at displacements less than 7.0 millimeters ( $p < 0.017$ ), with a maximum subluxation force of twenty-two newtons at a displacement of 0.5 millimeter during testing in internal rotation. This reversal in the effect on the subluxa-

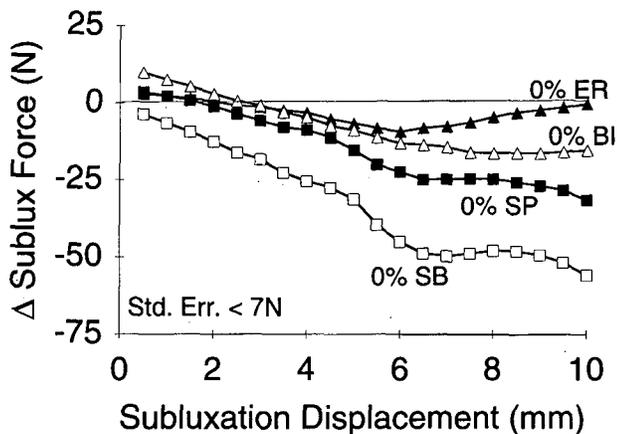


FIG. 4-A

Figs. 4-A and 4-B: Change-in-force-deflection curves after elimination of a muscle force. The ordinate is the difference in subluxation force between the standard case (100 per cent of all standard muscle forces) and the case after elimination of an individual muscle force. The horizontal line passing through zero newtons represents the results for the standard case. The standard error represents the largest value over all of the displacements. ER = external rotators, BI = long head of the biceps, SP = supraspinatus, and SB = subscapularis.

Fig. 4-A: In neutral rotation.

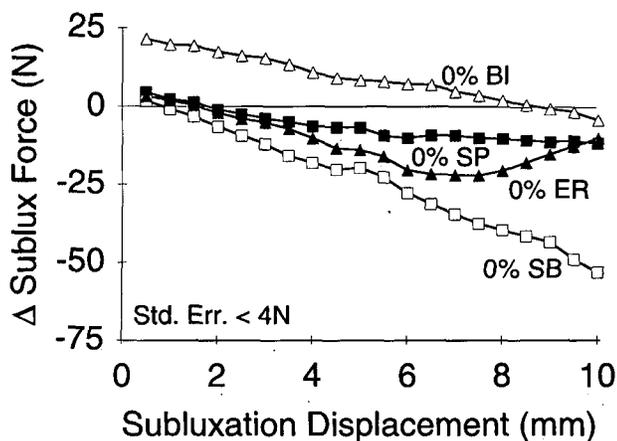


FIG. 4-B

In internal rotation.

tion force was found, to a lesser degree, with elimination of the simulated supraspinatus and subscapularis forces, at small displacements in neutral humeral rotation. Despite the low standard load for the subscapularis (forty-five newtons), that muscle contributed a maximum of fifty-six newtons to the subluxation force during testing in neutral rotation and fifty-three newtons during testing in internal rotation.

*Efficiency of the Muscles*

The relationship between the muscle force and the subluxation force was close to linear over the range of muscle forces tested, with squared correlation coefficients averaging 0.91 (Figs. 5-A and 5-B). The maximum efficiencies, in neutral and internal rotation, were 0.40 and 0.15 for the supraspinatus, 0.78 and 1.08 for the

subscapularis, 0.15 and 0.33 for the external rotators, and 0.44 and 0.10 for the biceps. However, the minimum efficiencies of the biceps were -0.33 and -0.73. The minimum efficiency was not less than -0.07 for any rotator cuff muscle.

*Effects of the Ligaments*

Cuts in the anterior zone (the superior and middle glenohumeral ligaments) resulted in decreases in the subluxation force of as much as thirty-seven newtons in neutral humeral rotation and forty newtons in internal humeral rotation (Figs. 6-A and 6-B). These decreases were significant ( $p < 0.001$ ) at all displacements during testing in both humeral rotations. After sectioning of the coracohumeral ligament, the subluxation force decreased by a maximum of fifty-seven newtons during testing in neutral rotation and forty-six newtons during testing in internal rotation ( $p < 0.001$ ). The cut made in the inferior zone (the inferior glenohumeral

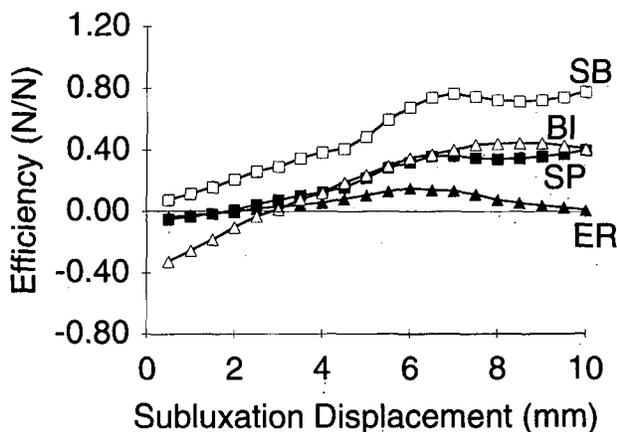


FIG. 5-A

Figs. 5-A and 5-B: Graphs of the muscle efficiencies at a range of subluxations. Efficiency was defined as the change in subluxation force divided by the change in muscle force. SB = subscapularis, BI = long head of the biceps, ER = external rotators, and SP = supraspinatus.

Fig. 5-A: In neutral rotation.

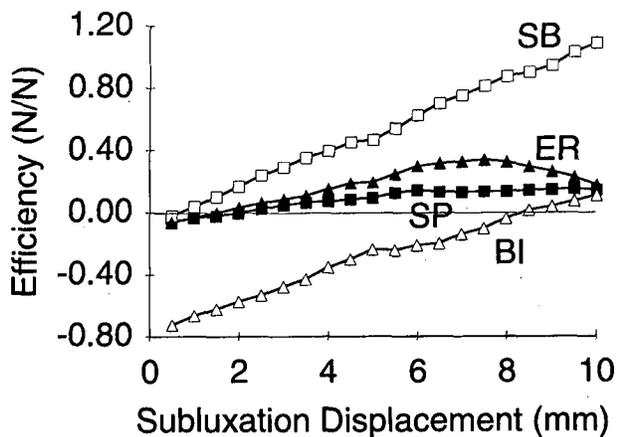


FIG. 5-B

In internal rotation.

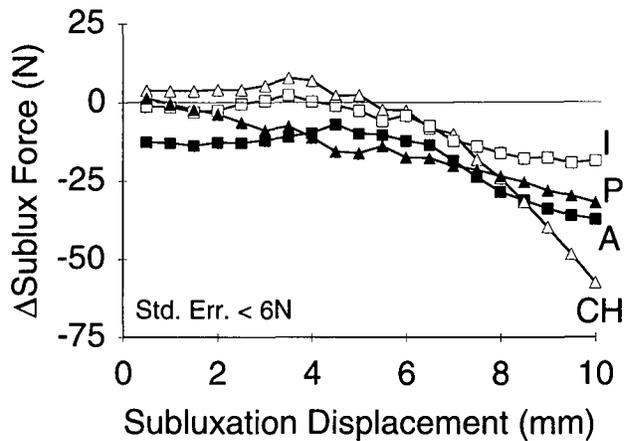


FIG. 6-A

Figs. 6-A and 6-B: Change-in-force-deflection curves for the effect of a cut in a ligamentous zone. The horizontal line passing through zero newtons represents the standard case (no cuts). The standard error represents the largest value over all of the displacements. A = anterior zone (superior and middle glenohumeral ligaments), P = posterior zone (posterior aspect of the capsule), I = inferior zone (inferior glenohumeral ligament), and CH = coracohumeral zone (coracohumeral ligament).

Fig. 6-A: In neutral rotation.

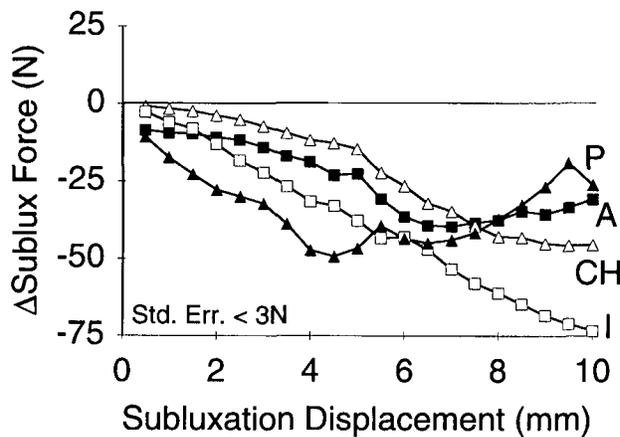


FIG. 6-B

In internal rotation.

ligament) resulted in a decrease in the subluxation force of as much as nineteen newtons during testing in neutral rotation and seventy-three newtons during testing in internal rotation ( $p < 0.001$ ). The sectioning of the posterior zone (the posterior aspect of the capsule) decreased the subluxation force by a maximum of thirty-two newtons in neutral rotation and by forty-nine newtons in internal rotation ( $p < 0.001$ ). It should be noted that, during testing in neutral rotation, no ligamentous zone contributed more than seventeen newtons to the stability of the shoulder at displacements of less than 6.0 millimeters.

### Discussion

#### Over-All Model

Previous comprehensive research addressing anterior instability of the shoulder has demonstrated a com-

plex stabilizing mechanism that includes muscular and ligamentous components<sup>10</sup>. To our knowledge, the most clinically important position of forward flexion has been used in only two previous studies of posterior instability<sup>3,19</sup>. In the current investigation, we tested posterior instability with the shoulder in forward flexion, the position in which posterior instability is detected clinically<sup>12</sup>, in which instability is most often symptomatic, and in which such destabilizing injuries usually occur. Our study also involved the use of a muscle-loading environment that was representative of physiological conditions. Resistance to subluxation is influenced by a multitude of complex factors, such as capsular and ligamentous attachments between the humerus and the scapula, muscle forces acting across the shoulder joint, the amount of subluxation, and the angular position and rotation of the arm. As such, a better understanding of the relative contributions of these factors can be gained only if they are investigated under the same study configurations.

Subluxations of zero to 10.0 millimeters were chosen to reflect a clinically relevant range<sup>3</sup>. Small subluxations were examined because displacements of only a few millimeters might be clinically important. Ten millimeters was chosen as the maximum subluxation because pilot studies demonstrated that substantial impingement against the acromion may occur at larger displacements. Also, much larger displacements would be expected to exceed the capabilities of the active stabilizers and to damage any effective passive stabilizers.

In this investigation, stability was investigated by measuring the force required to subluxate the humeral head a specified amount from its reduced position. This approach, while remaining clinically relevant, allows muscular and ligamentous factors to be analyzed independently as components of the over-all subluxation force<sup>1,10</sup>. Some clinicians believe that displacements of more than a few millimeters represent laxity or instability. We reported the subluxation forces for a large range of displacements so that readers could interpret the results within the context of their own definitions of laxity and instability. In addition, the evaluation of force contributions through a range of subluxations allows characterization of the progressive effects of each muscular or ligamentous factor<sup>10</sup>.

The total-force-deflection behavior exhibited negative force when subluxation was less than approximately 4.0 millimeters (Fig. 3). Although a small portion of this 4.0-millimeter offset might have been due to small shifts in the bone-fixture interface, the main causes were likely hysteresis-relaxation (irreversibilities such as friction in the testing apparatus and within the glenohumeral joint or viscoelasticity of the contacting cartilage). Our test protocol ensured that, at the initial position of zero subluxation, the humeral head was fully reduced, but the main emphasis of this study did not depend directly on a zero offset, as we examined only

changes in the force-displacement curve that were due to changes in muscle loads or to cutting of a ligament. These changes were relatively independent of the hysteresis-relaxation effects on the over-all force-displacement curve.

Multivariable regression analysis was selected to allow independent characterization of the force contribution from each factor tested. The number of independent variables studied was limited in the regression analysis to prevent the problem of overestimating the significance. The model included an indicator variable for specimen, which prevented variability among specimens from influencing the estimates for the effects of the muscle forces and the ligamentous cuts.

#### *Effects of the Muscles*

The findings supported the first hypothesis of this study: all of the rotator cuff muscles contributed to the subluxation force at all but the smallest displacements. Of all of the muscles investigated, the subscapularis provided the most resistance to posterior subluxation of the humerus. This may be explained by the redirection of the subscapularis tendon as it turns around the glenoid rim, which results in a more anteriorly directed force, resisting the posterior subluxation. In contrast, the forces in the remainder of the rotator cuff act indirectly by concavity-compression stabilization<sup>8</sup>. Concavity-compression stabilization typically provides efficiencies of approximately 0.3 to 0.4, whereas the subscapularis was found to have efficiencies greater than 1.0. This increase probably is due to the augmentation of concavity-compression by a force directly opposing subluxation, which suggests that, in the clinical situation, strengthening of the subscapularis may augment posterior stability of the shoulder.

The second hypothesis of our study was supported by the finding that the long head of the biceps contributed to the subluxation force at the medium and large displacements in neutral rotation and reduced the subluxation force at all but the largest displacements in internal rotation. With the arm in neutral rotation, the intra-articular portion of the long head of the biceps, from the intertubercular groove to the supraglenoid tubercle, runs beneath and approximately parallel to the supraspinatus and typically acts as a joint stabilizer. With the arm in internal rotation, however, the intra-articular portion of the long head of the biceps places a posteriorly directed force on the humerus, thus aggravating posterior subluxations. The negative efficiencies of the biceps in this rotation (Fig. 5-B) reflect a destabilizing tendency. This is consistent with findings that the biceps prevents anterior subluxation with the arm in internal rotation<sup>14</sup>. As subluxation progresses, the direction of the intra-articular portion of the biceps comes to lie more perpendicular to the glenoid face, and this posterior force is eliminated. Although this destabilizing effect of the biceps is not likely to be

found in many clinical situations, one striking example is consistent with these findings. In an electromyographic study of individuals who were voluntarily dislocating the shoulder posteriorly<sup>15</sup>, one subject used 440 per cent of the normal level of biceps activity to create posterior subluxation in the classic position of forward flexion and internal rotation. Undoubtedly, numerous muscle forces (for example, from the pectoralis, the anterior deltoid, and the short head of the biceps) may aggravate posterior instability. In contrast to previous studies<sup>7,10</sup>, the present study showed that the long head of the biceps can both increase and decrease the resistance to posterior subluxations.

#### *Effects of the Ligaments*

The third hypothesis of this study, regarding the role of the anterior zone, was supported for all displacements in either rotation. The ligaments in this zone had a consistently larger effect on the subluxation force in internal rotation than in neutral rotation. This can be explained by the initial tightness of these structures in internal rotation and the relative laxity in neutral rotation. With the humerus in neutral rotation, the effects of the cutting of the ligaments were important only at large displacements. This finding probably is due to the initial laxity and the later tightening of these structures with subluxation.

A surprising result of this investigation was the dramatic effect of the coracohumeral ligament in neutral humeral rotation at displacements between 5.5 and 10.0 millimeters, where it began as an insignificant contributor to the subluxation force and ended up as the most important contributor (Fig. 6-A). Thus, our final hypothesis was supported, but only at the larger displacements. In internal rotation, the contribution of the posterior aspect of the capsule to the subluxation force first increased but then, surprisingly, decreased (with larger displacements) (Fig. 6-B). This finding may be explained by the fact that the posterior aspect of the capsule first tends to tighten as the humerus displaces posterolaterally, due to the depth of the glenoid. After the humerus is on the glenoid rim, however, the lateral component to the subluxation lessens. Then, additional posterior displacement loosens the posterior aspect of the capsule, and its contribution to the subluxation force tends to decrease. The single most important ligamentous or muscular factor in this study was the inferior glenohumeral ligament at large displacements in internal rotation. The combination of internal rotation and forward flexion tended to position this structure in a more anterior-posterior orientation. This function, combined with the reduction in laxity due to internal rotation, is well suited to resist posterior subluxation.

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