QUANTIFICATION OF THE DYNAMIC GLENOHUMERAL STABILITY PROVIDED BY THE ROTATOR CUFF MUSCLES IN THE LATE COCKING PHASE OF THROWING

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The purpose of this study was to investigate quantitatively the dynamic glenohumeral stability provided by the rotator cuff muscles in the late-cocking phase of throwing. Using ten cadaveric shoulders, a new biomechanical parameter (dynamic stability index) was calculated considering compressive and shear forces to the glenoid provided by each cuff muscle. The rotator cuff muscles provided the joint with significant dynamic stability in the late cocking. However, the dynamic stability provided by the cuff muscles with the arm in coronal plane significantly decreased when compared with that in the scapular plane. Rehabilitation of the throwing athletes should emphasize strengthening of the external and internal rotators. Moving the arm into the coronal plane in the late cocking phase should be avoided by a specific strengthening program to prevent further anatomic damage to the static stabilizers.

KEY WORDS: dynamic stability, rotator cuff, late cocking, dynamic stability index

INTRODUCTION: During the late cocking phase of throwing, the humerus maintains its level of abduction and moves into the scapular plane while externally rotating from 46° to 170°. In this position the head of the humerus is angled so that it can stretch the anterior structures. This creates the potential for anterior instability. Baseball pitchers with unstable shoulders demonstrated several significant differences from pitchers with normal shoulders during the late cocking phase. Jobe et al. (1983) reported that the player with unstable shoulder might begin with some compensatory mechanics, such as moving the humerus into the coronal plane. When the humerus moves into the coronal plane, the head of humerus angles even further anteriorly.
Both static (capsule and ligaments) and dynamic factors (muscle contraction) are responsible for glenohumeral joint stability. Many authors have noted that muscle activity across a joint leads to increased stability through concavity-compression mechanism (Matsen, 1991; Lippitt, 1993). The rotator cuff muscles are primarily dynamic stabilizer of the glenohumeral joint.
The purpose of this study was to twofold. First, we wanted to define a new biomechanical parameter which considered actual compressive and shear force vectors generated by each rotator cuff muscle and the concavity-compression mechanism, and thus could estimate quantitatively the dynamic glenohumeral stability provided by the rotator cuff muscles. Second, by calculating quantitatively the dynamic stability, we wanted to compare the dynamic stability provided by the rotator cuff muscles in the late cocking phase of throwing with the arm in the scapular plane with that in the coronal plane.

METHOD: Ten fresh-frozen shoulders (6 right and 4 left) from human cadavers (age range 48 to 74 years) were prepared. The glenohumeral joint was disarticulated by resecting the gleno-humeral joint capsule after four rotator cuff muscles (subscapularis, supraspinatus, infraspinatus, and teres minor) were released from the scapula origin. The glenoid labrum was preserved. The glenoid neck was osteotomized 2.5 cm medial to its articular surface. A specially designed Plexiglas frame was constructed to permit placement of the humerus in the desired glenohumeral position (Figure 1). The distal shaft of the humerus was fixed to the frame on which the load-cell (AMTI Model FS 160A-600; Barry Wright, Watertown, Massachusetts) was mounted. An anatomical neutral position was the position in which the humerus is unelevated

(parallel to the medial border of the scapula) and in 0 degrees of rotation (the elbow was flexed and the forearm was perpendicular to the coronal plane). The osteotomized glenoid was
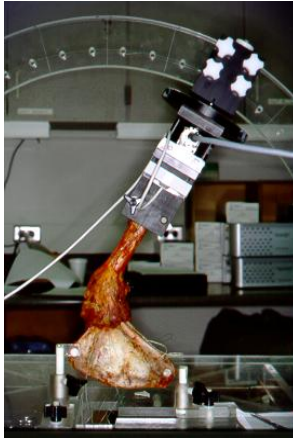

replaced back to the neck of the scapula and fixed temporarily for exact positioning of the humerus on the scapula. The scapula was rigidly fixed to the specially designed mounting device that permitted excursion of lines connected to each rotator cuff muscle around the scapula. The mounting device enabled the scapula and its glenoid to be anatomically aligned to the humeral head in any glenohumeral position. The glenoid piece was then taken out of the place before data collection to avoid bony contact between the humerus and scapula that could affect the measurement of force. Then, the position of the humeral head was maintained by secure fixation of the whole frame.

Muscle contraction was simulated by application of a constant force of 20N to each muscle individually by means of a hanging weight system. A six-degree-of-freedom electromagnetic tracking device (Fastrak, Polhemus Navigational Sciences Division, Colchester, VT) was used to measure position and orientation of the glenohumeral joint. The load-cell permitted accurate resolution of the forces that were applied to the humeral head by the rotator cuff muscle force across the joint. Anatomical axes for the measurement of force vectors were defined to be in line with the anterior/posterior and superior/inferior axes of the glenoid. The medial/lateral axis was defined as perpendicular to both axes.

Figure 1 - A specially designed Plexiglas frame.

Data acquisition and analysis. To simulate the late cocking phase of throwing, testing was performed with the humerus (1) in the scapular plane and (2) in the 45° of extension (coronal plane), while glenohumeral joint was maintained in 60° of abduction, 90° of external rotation. The glenohumeral joint in 60° of abduction corresponded to the shoulder joint in 90° of abduction. The force components in the medial/lateral (compression), anterior/posterior (shear), and superior/ inferior (shear) directions generated by each rotator cuff muscle/tendon unit were measured at both glenohumeral positions before and after application of the constant force of 20N to each muscle. Data measured before force application were used as the control. Raw load cell data were then transformed three dimensionally according to the defined anatomical axes. To verify the accuracy of measurement, vector summation of the three measured force components for each muscle was compared to the applied force in every glenohumeral position. The following calculations/derivations were made on the force vector data;

$$\text{Percent compressive force} = F_{comp} \Big/ \sqrt{(F_{comp}^2 + F_{A-P\,shear}^2 + F_{S-I\,shear}^2)} \times 100$$

$$\text{Percent anterior shear force} = F_{A-P\,shear} \Big/ \sqrt{(F_{comp}^2 + F_{A-P\,shear}^2 + F_{S-I\,shear}^2)} \times 100$$

$$\text{Percent superior shear force} = F_{S-I\,shear} \Big/ \sqrt{(F_{comp}^2 + F_{A-P\,shear}^2 + F_{S-I\,shear}^2)} \times 100$$

, where F_{comp}, $F_{A-P\,shear}$, and $F_{S-I\,shear}$ denoted measured compressive, anterior shear, and superior shear forces, respectively. The denominator of the above equations represented a vector sum of the measured force vector.

A new biomechanical parameter, the dynamic stability index (DSI), was defined to represent the combined stabilizing effects of the rotator cuff muscle force vectors and the concavity-compression mechanism of the glenohumeral joint. The dynamic stability index in the anterior direction can be derived as;

Dynamic stability index (DSI)

= (percent compressive force x stability ratio* in anterior direction) - percent anterior shear force

The stability ratio[*] is a concept describing the maximum dislocating force in a given direction that can be stabilized by the compressive muscle load on the glenoid fossa, assuming that frictional effect is minimal. From the published data regarding the concavity-compression mechanism, the stability ratio in the anterior direction was determined as 0.35 for glenohumeral joint with the labrum intact (Lippitt, 1993). The unit of DSI is a percent magnitude of the minimal external anterior shear force that can dislocate the joint to the magnitude of contraction force of each cuff muscle. Thus the higher the DSI of a rotator cuff muscle, the greater the dynamic glenohumeral stability. All the percent force vectors and the dynamic stability index with the humerus in scapular and coronal plane were compared by analysis of variance (two-way layout with equal numbers of observations in the cells), for the force range that occurred in all specimens.

RESULTS: The rotator cuff muscles were primarily compressors as the compressive forces were far greater than the shear forces regardless of humeral position. The shear force vector generated by each rotator cuff muscle could either stabilize or destabilize the joint depending on its direction, directly affecting the dynamic stability.

Dynamic Stability Index (DSI). DSI of the teres minor, infraspinatus, supraspinatus, and subscapularis in the late cocking phase with the arm in the scapular plane was 58 ± 9, 34 ± 14, 18 ± 10, and 43 ± 14, respectively. DSI with the arm in the coronal plane was 40 ± 13, 39 ± 13, 6 ± 16 and 28 ± 9, respectively for each muscle (Figure 2). The supraspinatus was the least effective stabilizer in both positions ($p<.05$). Dynamic anterior stability provided by the teres minor and subscapularis decreased significantly when the arm moves from the scapular plane to the coronal plane in terms of DSI ($p<.05$)
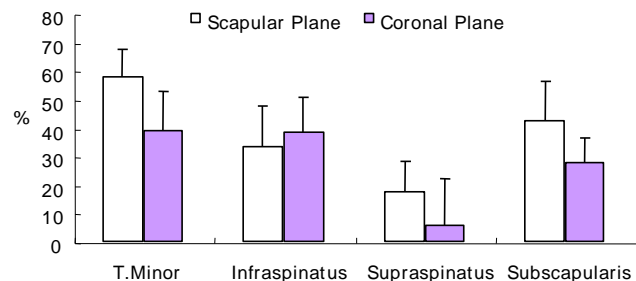

Figure 2 - DSI with the arm in the coronal plane

DISCUSSION: All joint force vectors can be resolved into compressive and shear components. The compressive force component by the muscle stabilizes the glenohumeral joint by the mechanism referred to as concavity- compression, where the stability is related to the depth of concavity and the magnitude of the compressive force (Matsen et al., 1991). We defined the dynamic stability index (DSI) to more realistically represent the biomechanical role of the force vectors providing dynamic glenohumeral stability. The DSI of a muscle comprised both the effects due to the concavity-compression mechanism and the shear force generated by a muscle itself. The subscapularis can stabilize the glenohumeral joint sufficiently to resist a dislocating anterior shear force of up to 43N, if DSI of the subscapularis in a specific humeral position were 43 per cent and the muscle force were 100N for example. The higher the DSI, the greater the dynamic glenohumeral stability.

Using the DSI as the most comprehensive way to analyze muscle effects on stability, the subscapularis (internal rotator) and infraspinatus and teres minor (external rotator) was found to provide greater dynamic stability in anterior direction than the supraspinatus in the late cocking phase. Gowan et al. (1987) analyzed electromyographic signals during pitching and reported that, during late cocking, the subscapularis had the highest activity, followed by the infraspinatus and teres minor. The supraspinatus had the least activity in this position. The present study exactly shows the economy of the shoulder muscle activity in a physiological circumstance.

Glousman et al. (1988) reported the patients who had chronic anterior instability of the shoulder had markedly lower electromyographic activities in the infraspinatus and subscapularis during late cocking. The levels of activity in the supraspinatus increased throughout the cocking phase. The authors, on the basis of our result, would be able to suggest that the patterns of altered electromyographic activity in anterior shoulder instability contributed to the instability.

Notably, in the present study, dynamic glenohumeral stability provided by the teres minor and subscapuris decreased significantly when the arm moves from the scapular plane to coronal plane ($p<.05$). The maximum muscle force may be estimated from the physiological cross-sectional area of each muscle (Ikai & Fukunaga, 1968). The subscapularis has the largest proportional physiological cross-sectional area among the cuff muscles and the decrease in the DSI of the subscapularis in the coronal plane would be able to have a more pronounced influence on the dynamic stability.

There are limitations in this study. First, simulation of muscle contraction by a single line of action might be different from that in vivo. However, we reproduced the line of action of a muscle that was determined previously by the anatomical and magnetic resonance imaging studies of shoulder (Blasier et al., 1992). Second, the muscle length-tension relationships will differ according to shoulder position. To account for this, the force components and the DSI in this study were defined as percentage of the force applied to each muscle.

CONCLUSION: The rotator cuff muscle provided the glenohumeral joint with significant dynamic stability in the late cocking phase of throwing motion. But, force vectors generated by the cuff muscles with the arm in the coronal plane was less efficient to prevent anterior translation of the humeral head than those in the scapular plane.

Rehabilitation of the throwing athletes should emphasize strengthening of the external and internal rotators, which are efficient stabilizer in the late cocking phase, to enhance the dynamic stability. This study also shows that if a pitcher begins to move the arm into the coronal plane in the late cocking phase, then he can be pulled off the field and put on a specific strengthening program of the scapular muscles to prevent further anatomic damage to the static stabilizers. Dynamic glenohumeral stability was successfully quantified by a new biomechanical parameter, dynamic stability index (DSI), defined in this study.

REFERENCES:

Blasier, R.B., Guldberg, R.E., & Rothman, E.D. (1992). Anterior shoulder stability: contributions of rotator cuff forces and the capsular ligaments in a cadaver model. *J. Shoulder Elbow Surg*, **1**, 140-150.

Glousman, R., Jobe, F., Tibone, J., Moynes, D., Antonelli, D., & Perry, J. (1988). Dynamic electromyographic analysis of the throwing shoulder with glenohumeral instability. *J. Bone Joint Surg*, **70-A**(2), 220-224.

Gowan, I.D., Jobe, F.W., Tibone, J.E., Perry, J., & Moynes, D.R. (1987). A comparative electromyographic analysis of the shoulder during pitching. *Am. J. Sports Med*, **15**, 586-90.

Ikai, M., & Fukunaga, T. (1968). Calculation of muscle strength per unit cross-sectional area of human muscle by means of ultrasonic measurement. *Int. J. Angew. Physiol. Einschl. Arbeitsphysiol*, **26**, 26-28.

Jobe, F.W., Tibone, J.E., Perry, J., & Moynes, D. (1983). An EMG analysis of the shoulder in throwing and pitching: *Am. J. Sports Med,* **11**, 3-5.

Lippitt, S.B., Vanderhooft, J.E., Harris, S.L., Sidles, J.A., Harryman, D.T., II, & Matsen, F.A., III. (1993). Glenohumeral stability from concavity-compression: a quantitative analysis. *J. Shoulder Elbow Surg*, **2**(1), 27-35.

Matsen, F.A., III, Harryman, D.T., II, & Sidles, J.A. (1991). Mechanics of Glenohumeral instability. *Clin. Sports Med*, **10**, 783-788.