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A Stochastic Analysis of Glenoid Inclination Angle and Superior Migration of the Humeral Head

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Abstract

Background—Superior glenoid inclination, which is a relatively upward facing of the glenoid in the plane of the scapula, has been associated with rotator cuff pathology. Increased glenoid inclination may cause superior humeral head migration, which can cause impingement of the supraspinatus tendon. The purpose of this study was to test the hypothesis that inclination angle affects the probability of superior humeral head migration.

Methods—A three-dimensional model of the glenohumeral joint was developed in which muscle forces were modeled as random variables. Monte Carlo simulation was used to compute the probability that the glenohumeral reaction force was directed such that superior humeral head migration should occur. An electromyogram-driven model was used to estimate shoulder muscle forces in healthy volunteers performing arm elevation.

Findings—The model predicted that the probability of superior humeral head migration increased as glenoid inclination angle was increased. This finding was independent of the assumed shape of the muscle force probability distributions.

Interpretation—The results support the theory that glenoid inclination may be a risk factor for rotator cuff pathology.

Keywords

shoulder; glenoid; rotator cuff; stochastic; modeling

1. INTRODUCTION

The rotator cuff muscles provide stabilizing forces that act to prevent the deltoid from pulling the head of the humerus superiorly during the initiation of arm elevation. Superior translation of the humerus may result in rotator cuff pathology because translation of the humerus can compress the supraspinatus tendon between the humerus and acromion. Superior translation occurs when the net force vector acting on the humerus points superiorly out of the glenoid.

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The angle of the glenoid in the plane of the scapula, which we term glenoid inclination, has been shown to be associated with rotator cuff tendon tears (Hughes et al., 2003; Tetreault et al., 2004), and has been hypothesized that an increased glenoid inclination angle affects the tendency for the humeral head to translate superiorly (Wong et al., 2003).

Most models of glenohumeral joint loading have been deterministic (de Luca and Forrest, 1973; Poppen and Walker, 1978; Karlsson and Peterson, 1992; van der Helm, 1994; Nieminen et al., 1995; Hughes and An, 1996), and model parameters have been the mean or median of reported values. A deterministic modeling approach predicts that, for average conditions, the mean net force on the humerus is directed into the glenoid (Poppen and Walker, 1978). This does not explain how increased glenoid inclination may cause cuff pathology.

However, there is significant variability in motor control (Carlton and Newell, 1993), which produces variability in muscle force. We hypothesize that motor control variability can cause superior humeral head migration in certain circumstances even though the average condition would imply that this does not occur. Furthermore, we theorize that an increase in glenoid inclination angle increases the probability of superior humeral head migration. There were two objectives of this study: (1) to develop a stochastic three-dimensional biomechanical model of the glenohumeral joint to assess the effect of motor control variability on the probability of superior humeral head migration during the initiation of arm elevation, and (2) to use the biomechanical model to determine if an increased glenoid inclination angle increases the probability of superior humeral head migration.

2. MATERIALS AND METHODS

The stochastic simulation model uses muscle forces estimated from electromyographic (EMG) signals recorded from human volunteers performing shoulder abduction. The work described below followed the EMG-force relationship calibration procedure described by Gatti et al. (2007b), thus details concerning the subjects, EMG electrode preparation and placement, and EMG processing apply to the current work as well.

2.1 Experimental Procedures

Marker arrays consisting of three infrared emitters each (NDI, Waterloo, Ontario, Canada) were placed on the subject at three locations in order to determine the posture of the humerus: (1) seventh cervical vertebra (C7), (2) scapula (using a skin-mounted fixture developed by Karduna et al. (2001)), and (3) posterior aspect of the distal humerus just proximal to the elbow. In order to minimize the effect of soft-tissue motion, markers were placed on the most bony prominence near the desired location. Anatomic landmarks were then digitized to determine the locations of the subject's anatomy relative to the three markers, which were used to create a computer representation of the subject and determine the arm posture throughout the abduction motion trials.

For the dynamic abduction motion trials, subjects held a 1.3 kg weight in the right hand and were instructed to smoothly abduct to 90° in the plane of the scapula at a comfortable speed. Fifteen abduction motion trials were performed with sufficient rest between each trial in order to avoid subject fatigue. EMG and kinematic data were recorded for each motion trial. EMG data were measured from 12 muscles that cross the shoulder: sternocostal (StPec) and clavicular (ClPec) heads of pectoralis major; long (LHB) and short (SHB) heads of biceps; anterior (ADelt), middle (MDelt), and posterior (PDelt) deltoids; infraspinatus (Infra); long head of triceps (Tri); latissimus dorsi (Lat); suprapinatus (Supra), and subscapularis (Subsc). EMG signals were recorded at 1000 Hz with a Noraxon Myosystem 2000 EMG system (Noraxon Inc., Scottsdale, AZ, USA) and were filtered using a 4th order Butterworth filter (4 Hz low-pass cutoff) and subsequently post-processed using methods presented by Lloyd and Besier

(2003), which synchronize muscle activation and muscle force production by accounting for electromechanical delay. Kinematic data were collected using Optotrak Certus software (NDI, Waterloo, Ontario, Canada). Data from the three marker arrays were converted to Euler angles according to the recommendations by the International Society of Biomechanics (Wu and Cavanagh, 1995). Arm posture was described by the plane of arm elevation, humeral rotation, and arm elevation with respect to the torso.

2.2 Muscle Force Estimation

The force generated by the i^{th} muscle (F_i^{muscle}) during a motion trial was estimated using

$$F_i^{\text{muscle}} = E_i \cdot \psi_i \cdot \phi_i \cdot G_i \quad (1)$$

where E_i is the filtered and processed EMG data, ψ_i accounts for the muscle force-velocity relationship (Kaufman et al., 1991), and ϕ_i accounts for the muscle length-tension relationship, and G_i is a constant muscle parameter which defines a linear relationship between EMG and muscle force. The values of ϕ_i and the parameters necessary for computing ψ_i were obtained using the computational shoulder model developed by Holzbaur et al. (2005), implemented with Software for Interactive Musculoskeletal Modeling (SIMM) (Musculographics, Inc., Evanston, IL, USA). Subject variability in the speed of the motion trials is reflected in the force-velocity relationship, and this parameter was found to have only a small effect on the predicted muscle force. Muscle parameters (G_i) were determined using sequential principal-components regression with data from EMG-muscle force calibration experiments; see Gatti et al. (2007b) for a comprehensive description of the calibration experimental procedures and parameter calculations. The variable parameters in Eq. 1 (E_i , ψ_i , and ϕ_i), as well as the muscle lines of action, were calculated at the instant when the subject's arm was positioned at 30° of arm elevation. This arm angle was chosen to most closely match the arm posture used in the EMG-force calibration. The means and standard deviations of the muscle force estimates for all 15 trials, for each subject, were inputs to the computational model and are shown in Table 1.

2.3 Probabilistic Computational Model

The computational model accounts for the effects of variations in muscle forces on the probability of superior migration, and incorporates gravitational forces, inertial forces and moments, anthropometry, and muscle lines of action as independent random variables. The gravitational and inertial force were estimated using a dynamic analysis, assuming the arm was rigid and was moving in a circular arc centered at the glenohumeral joint. Subject height, weight, and arm length measurements were individually entered into the model. Locations of the centers of mass of the arm segments were calculated from published anthropometric values (Chaffin and Andersson, 1984). Muscle lines of action were taken to be from origin to insertion, and were obtained using the Holzbaur shoulder model; this model accounts for scapular movement based on shoulder elevation angle. As this study analyzes a posture in the mid-range of motion, muscle forces provide the majority of glenohumeral stability (Veeger and van der Helm, 2007), therefore other structures have not been included in the current model.

The probability of humeral head migration was estimated from the EMG-derived muscle forces and arm kinetics using the Monte Carlo simulation method. At each iteration, individual muscle forces were randomly selected from independent probability distributions, with all other inputs kept constant, and the net glenohumeral force was used to determine if conditions were sufficient for superior migration (Figure 1). For each subject, and for each simulation, consisting of 10,000 iterations, the probability of superior migration was the proportion of the iterations resulting in superior migration. The reported probabilities (and standard deviations) were determined by averaging the probability of superior migration for each simulation, across

all 11 subjects, of the same muscle force distribution. Initial simulations were run multiple times to ensure that this number of iterations was sufficient to determine the steady-state output probability; similar biomechanical analyses have used far fewer iterations per simulation (Hughes and An, 1997; Valero-Cuevas et al., 2003; Langenderfer et al., 2006).

Separate simulations were performed using muscle forces sampled from four different distributions: normal, Gamma, uniform, and exponential. Multiple distributions were simulated in order to test the robustness of the results with respect to the assumed distribution. The normal and Gamma distributions are the most physiologically relevant for three reasons: (1) their values are always positive, (2) the distributions are unimodal, and (3) their parameters can be calculated from the experimental data. The shape and scale parameters of the normal and Gamma distributions were calculated using the mean and standard deviation of the muscle force data from each muscle. The uniform distribution, ranging from zero to twice the mean, was also used as it has been used in other biomechanical studies (Valero-Cuevas et al., 2003). Exponential distributions require a single parameter, equal to both the mean and standard deviation. However, the mean and standard deviation of the experimental muscle force estimates were not similar, so two different exponential distributions were used; one exponential distribution had the parameter equal to the mean and the other had the parameter equal to the standard deviation. All muscle force probability distributions were specific to each subject. Following random number generation, muscle forces were slightly modified to ensure moment equilibrium at the glenohumeral joint. This was done by finding the nearest set of muscle forces satisfying moment equilibrium to the vector of muscle forces generated by the random number generation process. A quadratic programming problem was solved to minimize the Euclidean distance.

In order to analyze the reaction forces of the humeral head in the glenoid, a new coordinate system, local to the glenoid, was developed based on the Holzbaur shoulder model. The origin was at the center of the glenoid and the three orthogonal axes were defined as superior-inferior, anterior-posterior, and medial-lateral with respect to the glenoid. Due to the elliptical shape of the glenoid, the superior-inferior axis was defined as the line connecting the two points on the edge of the glenoid that were furthest from the origin. The anterior-posterior axis was defined to be perpendicular to the superior-inferior axis and parallel to the surface of the glenoid. The medial-lateral axis was defined to be the cross-product of the superior-inferior and anterior-posterior axes and was directed into the glenoid. All glenohumeral kinetic calculations were performed in this coordinate system.

Whether or not the glenohumeral joint experienced kinetics that produce superior humeral head migration (“cause superior migration?” box in Figure 1) was decided by determining if the net force acting on the glenoid by the humerus was sufficient to cause glenohumeral dislocation. This was done by first computing the net force vector, $\mathbf{F}$, and comparing it to the data reported by Lippitt et al. (1993), which describes the necessary translation force required to dislocate the glenohumeral joint in eight different directions. Those data were derived from an experiment in which the humeral head was pulled parallel to the glenoid until it dislocated. The experiment was conducted with a 50 N compression force applied to the humeral head. When viewing the right glenoid face, the eight directions were: 0° (superior), 45° , 90° (anterior), 135° , 180° (inferior), 225° , 270° (posterior), and 315° . Twenty-nine Newtons of force were required to dislocate the humeral head superiorly and 32 N were required inferiorly when there was a 50 N compression force applied. The three-dimensional vectors corresponding to these measurements are $\mathbf{n}^{superior} = (29 \ 0 \ 50)^T$ and $\mathbf{n}^{inferior} = (-32 \ 0 \ 50)^T$, respectively. In a two-dimensional model, the joint was considered to be stable if and only if $\mathbf{F}^{glenohumeral}$, the net force acting on the glenoid by the humerus, was contained in a cone formed by $\mathbf{n}^{superior}$ and $\mathbf{n}^{inferior}$ (Figure 2). This concept was generalized to three-dimensions by considering all eight measurements reported by Lippitt et al. (1993); Table 2 presents the

three-dimensional vectors containing dislocation forces $\mathbf{n}_1, \dots, \mathbf{n}_8$. Similarly, the kinetics were considered sufficient for dislocation if $\mathbf{F}^{\text{glenohumeral}}$ pointed outside the cone formed by $\mathbf{n}_1, \dots, \mathbf{n}_8$. Mathematically, this is when $\mathbf{F}^{\text{glenohumeral}}$ cannot be represented as a non-negative linear combination of $\mathbf{n}_1, \dots, \mathbf{n}_8$. This condition was tested by solving the linear programming problem (Murty, 1983):

$$\text{Minimize } z = [1 \quad 1 \quad 1] \mathbf{u} + [1 \quad 1 \quad 1] \mathbf{v} \quad (2)$$

$$\text{such that } \mathbf{u} - \mathbf{v} + \sum_{i=1}^8 \alpha_i \mathbf{n}_i = \mathbf{F}^{\text{glenohumeral}} \quad (3)$$

$$\alpha_i \geq 0, i=1, \dots, 8 \quad (4)$$

where $\mathbf{u}$, $\mathbf{v}$, and α_i are decision variables; $\mathbf{u}$ and $\mathbf{v}$ are three-dimensional column vectors and α_i are scalars. If the algorithm reached zero, then $\mathbf{F}^{\text{glenohumeral}}$ was contained in the cone formed by $\mathbf{n}_1, \dots, \mathbf{n}_8$ otherwise, $\mathbf{F}^{\text{glenohumeral}}$ was not contained in the cone. If it was determined that dislocation would occur, the direction of dislocation was determined by computing the dot product of $\mathbf{F}^{\text{glenohumeral}}$ and a superiorly-directed vector. Superior migration was considered to occur if this quantity was positive.

A normal glenoid inclination angle (0°) was defined as that represented in the Holzbaur shoulder model. The glenoid inclination angle was varied by rotating the stability vectors ($\mathbf{n}_1, \dots, \mathbf{n}_8$) about the anterior-posterior axis of the local glenoid coordinate system. Superior inclination of the glenoid was assumed not to affect the muscle lines of action, the muscle strength, or the movement of the shoulder during abduction.

2.4 Statistical Analysis

The association between superior migration probability and glenoid angle was assessed using Spearman rank correlation coefficients. SPSS (SPSS Inc., Chicago, IL, USA) was used for the analysis.

3. RESULTS

The stochastic model demonstrated that a greater inclination of the glenoid increased the probability of superior migration of the humeral head (Figure 3 and Figure 4a). The increased probability of superior migration was significantly associated with glenoid inclination angle for simulations using the normal ($P=0.003$), Gamma ($P=0.005$), uniform ($P<0.001$), and exponential ($P<0.001$) distributions. Thus, our results are independent of the assumed statistical distribution of muscle forces.

For all subjects and all probability distributions, an increased glenoid inclination angle resulted in an increasingly superior direction of migration, an increased translation force (parallel to the glenoid surface), and a decreased compression force (normal to the glenoid surface); these effects are shown in Figure 4b, 4c, and 4d, respectively, for muscle forces sampled from the normal and Gamma distributions. For example, the mean translation force increased from 377 to 647 N, and the mean compression force decreased from 1133 to 998 N for simulations using the Gamma distribution.

4. DISCUSSION

The purpose of this study was to employ a stochastic model of the glenohumeral joint to evaluate the effect of motor control variability on the probability of superior humeral head migration during abduction and to establish if an increased glenoid inclination angle increases

this probability. Stochastic simulations demonstrated that random variability in muscle forces can produce superior humeral head migration events. Moreover, this study demonstrated that an increased glenoid inclination angle increased the probability of superior humeral head migration, even when a wide range of assumed muscle force distributions are used in the simulation. The glenoid inclination angles simulated in the current study fall within that of the population (Churchill et al., 2001), thus it is plausible that the results of this probabilistic model could occur *in vivo*.

When using normal and Gamma distributions, the model predicted that the probability of superior humeral head migration increased approximately 27% when the glenoid inclination angle increased 10° from its normal orientation (Figures 3a, 3b). This supports the theory that increased glenoid inclination will increase the probability of superior humeral head migration, which may be associated with rotator cuff disorders (Hughes et al., 2003). The superior tilting of the glenoid causes the magnitude of the translation and compression components of the net force vector to change (Figures 4c, 4d). The translation force increases and the compression force decreases which results in an increased probability of superior movement.

There are four limitations of this investigation. First, we used an idealized representation of the shoulder musculature. The Holzbaur shoulder model was used to compute muscle lengths, moment arms, and lines of action, and although it is sophisticated by current standards, muscles and sub-muscles are modeled using, either single or multiple, idealized straight muscle segments. Nevertheless, computed muscle moment arms agree with empirical measurements (Gatti et al., 2007a). Additionally, the Holzbaur shoulder model is representative of a 50th percentile male, and thus muscle parameters are not subject-specific. Second, it should be noted that the data used in the current study to determine the occurrence of superior migration was based on experimental data which determined the forces required for dislocation, and not merely humeral head migration. It is possible that impingement of the supraspinatus tendon could occur for net glenohumeral forces which are insufficient for dislocation, yet are sufficient for superior migration. Thus, the occurrence of superior migration is likely to be underestimated by the model. The third limitation is that the muscle forces were modeled as independent random variables. This is a common practice in stochastic simulations (Mirka and Marras, 1993; Hughes and An, 1997; Chang et al., 2000; McLean et al., 2003; Valero-Cuevas et al., 2003; Santos and Valero-Cuevas, 2004; Langenderfer et al., 2005), but it does not account for correlations between muscles. More work is needed to develop multivariate simulation methods in biomechanics. Lastly, it should be noted that the results of the model are specific to the abduction motion performed in this study, and it is not known if these results can be extrapolated to other situations or tasks.

There are several novel aspects of this study. First, we propose a method of analysis that explains how variability in motor control can affect glenohumeral mechanics. The method allows for the analysis of infrequent events in place of conducting an analysis of the average condition, which may be important in understanding injury risk. Second, a data-driven method of determining glenohumeral stability is introduced as an alternative to the idealized ellipsoidal representation of glenoid stability used in previous models (van der Helm, 1994). Although mathematically convenient, the ellipsoid model imposes an anterior-posterior symmetry that is not an accurate anatomical representation. The empirical data reported by Lippitt et al. (1993), which was derived from a study of cadaver shoulders, allows for the modeling of the asymmetric forces required to dislocate the joint.

In summary, this study used a mechanical explanation to demonstrate how a change in glenoid inclination angle can influence the occurrence of superior humeral head migration. The underlying concept of this study was that variability in muscle force production might lead to random superior migration events. Superior migration can lead to impingement of the

supraspinatus tendon under the coracoacromial arch, and such extrinsic injury has been found to be a factor in rotator cuff pathogenesis (Mehta et al., 2003).

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CONFLICT OF INTEREST STATEMENT

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Figure 1.

A flowchart description of the Monte Carlo simulation technique. The inertial forces and moments and gravitational forces, in addition to the average and standard deviation of muscle forces from the 15 trials, are entered into the Monte Carlo simulation computational model. The computational model, shown in the figure as a box with a dotted line, generates a probability distribution based on the muscle force mean and standard deviation. The program randomly samples muscle forces from distributions, combines them with the inertial forces and moments and gravitational forces to calculate the net force vector, and then determines if superior migration occurs. Over the 10,000 iterations, the proportion of times superior migration occurs is the reported probability.

Figure 2.

Two-dimensional example of model used to determine if superior humeral head migration occurs. $\mathbf{F}^{\text{glenohumeral}}$ is the net force acting on the glenoid by the humerus. (a) Inferior and superior limits of stability are defined by vectors $\mathbf{n}^{\text{superior}}$ and $\mathbf{n}^{\text{inferior}}$, which are taken from the experiment performed by Lippitt et al. (1993) on glenohumeral stability. (b) The condition for the humerus to remain centered in the glenoid is that $\mathbf{F}^{\text{glenohumeral}}$ lies in the cone formed by $\mathbf{n}^{\text{superior}}$ and $\mathbf{n}^{\text{inferior}}$. The cone is the set of all non-negative linear combinations of $\mathbf{n}^{\text{superior}}$ and $\mathbf{n}^{\text{inferior}}$. If $\mathbf{F}^{\text{glenohumeral}}$ lies outside the cone, then the kinetics are consistent with humeral movement out of the glenoid; if $\mathbf{F}^{\text{glenohumeral}}$ points superiorly then there would be

superior humeral head movement. Determination of whether $\mathbf{F}^{glenohumeral}$ lies within the cone is done using linear programming.

Figure 3. Probability of superior humeral head migration at glenoid inclinations of -5° to 20° in 5° increments for all five distributions used to sample muscle forces: normal (a), Gamma (b), uniform (c), exponential (mean) (d), and exponential (standard deviation) (e). Probability values are averaged across all subjects; error bars represent ± 1 standard deviation. The normal glenoid inclination angle is defined as 0° .

Figure 4. Predicted probability of superior migration (a), translation angle (b), compression forces (c), and translation forces (d) for simulations using normal and Gamma muscle force distributions. Reported values are averaged across all subjects; error bars represent ± 1 standard deviation. The normal glenoid inclination angle is defined as 0° . *The superiorly-directed glenoid axis is defined as 0° , and a clockwise rotation is defined as positive.

Table 1

Mean ($\pm$ 1 standard deviation) experimental muscle force (N) predictions for all abduction trials for each subject. Muscle name abbreviations correspond to those listed in the Methods section, 2.1. Experimental Procedures.

Muscle	Subject										
	1	2	3	4	5	6	7	8	9	10	11
StPec	98.62 (3.52)	13.42 (1.14)	50.02 (1.55)	31.34 (14.71)	23.28 (2.06)	16.86 (0.30)	47.71 (0.85)	58.59 (1.30)	57.46 (2.35)	146.89 (3.48)	60.99 (4.09)
ClPec	88.34 (2.97)	12.63 (10.16)	51.58 (0.99)	21.00 (5.23)	42.34 (10.68)	17.82 (0.97)	99.39 (46.86)	63.11 (1.06)	71.67 (19.79)	49.19 (14.18)	68.97 (14.43)
LHB	7.14 (1.24)	19.17 (2.73)	18.33 (3.81)	10.32 (6.38)	3.49 (0.75)	24.02 (2.39)	5.10 (0.65)	9.39 (0.96)	8.47 (1.73)	6.40 (0.22)	7.17 (0.90)
SHB	5.63 (0.57)	5.12 (1.16)	18.06 (4.49)	2.88 (1.08)	6.80 (1.36)	8.46 (0.73)	5.70 (0.12)	8.66 (0.22)	37.18 (7.12)	8.24 (0.13)	6.10 (0.39)
ADelt	487.51 (119.03)	699.04 (76.56)	202.98 (69.78)	99.72 (65.33)	169.09 (41.54)	355.21 (45.86)	272.08 (36.70)	119.89 (1.38)	302.91 (60.48)	253.27 (30.26)	200.45 (11.24)
MDelt	204.18 (44.82)	150.49 (13.93)	278.87 (52.05)	64.01 (35.47)	64.02 (10.24)	146.22 (10.86)	64.42 (16.8)	294.85 (46.99)	150.43 (31.12)	102.20 (24.29)	107.37 (21.04)
PDelt	140.69 (38.97)	191.05 (33.74)	369.90 (83.73)	235.89 (109.85)	39.81 (2.82)	259.38 (17.61)	149.45 (25.11)	727.2 (77.63)	160.56 (38.79)	55.97 (0.76)	213.83 (43.96)
Infra	277.52 (25.70)	97.34 (23.31)	154.90 (14.75)	532.85 (183.24)	222.24 (20.73)	47.55 (3.23)	128.29 (12.57)	471.60 (55.62)	186.00 (25.28)	286.76 (9.84)	281.93 (18.92)
Tri	68.32 (3.37)	25.22 (3.29)	115.36 (27.46)	91.88 (97.44)	33.00 (0.52)	43.78 (3.21)	72.21 (6.38)	119.61 (7.31)	151.09 (29.91)	136.14 (3.03)	46.74 (0.79)
Lat	116.41 (6.11)	18.53 (1.28)	61.29 (2.80)	26.45 (2.77)	29.45 (1.18)	68.55 (3.85)	55.41 (1.12)	60.11 (2.60)	54.58 (2.00)	300.99 (5.43)	67.39 (2.33)
Supra	71.83 (5.19)	342.46 (44.48)	104.66 (19.83)	15.76 (13.47)	31.07 (6.97)	23.44 (4.63)	56.18 (5.10)	121.68 (29.34)	76.98 (13.66)	201.23 (37.29)	64.66 (12.37)
Subsc	64.87 (7.19)	177.14 (17.45)	51.65 (15.45)	18.02 (15.06)	334.80 (41.47)	132.00 (14.84)	125.47 (25.31)	19.84 (0.56)	780.22 (63.00)	451.48 (7.50)	247.30 (2.01)

Table 2

Three-dimensional stability vectors (Lippitt et al., 1993) used to determine if the net force vector points into the glenoid. A local glenoid coordinate system is used when performing glenohumeral stability calculations. For the right shoulder, **i** (superior) points from the inferior to the superior aspect of the glenoid, **j** (anterior) is perpendicular to **i** and parallel to the glenoid surface, and **k** = **i** × **j** (medial).

Stability vector (i j k)	
n₁	(29.0 0 50.0)
n₂	(13.4 13.4 50.0)
n₃	(0 17.0 50.0)
n₄	(-16.3 16.3 50.0)
n₅	(-32.0 0 50.0)
n₆	(-17.7 -17.7 50.0)
n₇	(0 -17.0 50.0)
n₈	(14.1 -14.1 50.0)