Evaluation of Shoulder Stability During Forceful Arm Exertions

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Abstract: Shoulder musculoskeletal disorders (MSDs) are a major cause of morbidity and pain in the modern working population. Epidemiological literature suggests that forceful arm exertions pose an increased risk for shoulder MSD development. The majority of shoulder MSDs involve the glenohumeral joint. The glenohumeral joint is characteristically unstable and stabilized by concavity compression mechanism. In this study a biomechanical model of the shoulder complex was used to examine the concavity compression mechanism. Mechanical loading of the glenohumeral joint during forceful arm exertions was analyzed to quantify the angular position of the resultant muscle force vector in 3D space. The resultant muscle force vectors were almost always directed anteriorly, medially, and inferiorly, independent of the magnitude and the direction of the external force application. The knowledge gained in this study could potentially be used to develop a new assessment method to evaluate the risk of injury of the shoulder complex during forceful arm exertions.

Keywords: Shoulder, Concavity Compression Mechanism, Forceful Arm Exertions

1. Introduction

Musculoskeletal disorders (MSDs) place a substantial burden on both the employer and worker in terms of healthcare costs, human suffering, and the resulting socioeconomic impact. In particular, MSDs of the shoulder are a major cause of morbidity and pain in the modern working population. In 2011, shoulder disorders were the second most prevalent type of MSD and were the most severe requiring 21 median days away from work compared to 11 days for all other MSDs combined (Bureau of Labor Statistics, 2012). In addition to lost workdays, shoulder MSDs also generate expensive medical costs. For compensation claims data spanning from 1997 to 2005, the average total direct cost of a work-related shoulder disorder was $16,092 per claim in the state of Washington (Silverstein & Adams, 2007).

Epidemiological investigations have reported evidence that several work-related exposures are positively associated with the development of shoulder disorders. These studies have found that physical exposures such as awkward and prolonged sustained postures of the upper extremities, repetitive and forceful arm exertions can potentially lead to work-related shoulder disorders (Devereux et al., 2002; Larsson et al., 2007; da Costa & Vieira, 2010). Occupations such as nursing, material handling, janitorial work, transportation, and manufacturing have been found to have workers suffer more commonly from shoulder MSDs as they are frequently exposed to these risk factors; in particular, exposure to forceful arm exertions in pushing and pulling directions (Putz-Anderson et al., 1997; Dunning et al., 2010; Bureau of Labor Statistics, 2012).

Previous studies investigating the shoulder MSDs have utilized physiological and biomechanical methodologies. The goal of the physiological approach is to determine whether a task falls within acceptable limits based on the body's response. One common method used to gauge physiological response is muscle fatigue, which is the point where the muscle can no longer maintain the required contraction (De Luca, 1997). Electromyography (EMG) has been widely used to evaluate muscle fatigue as a measure of shoulder loading in a number of studies. In a study by Strasser and Muller (1999), subjects were asked to transfer loads from several remote starting points on a table located along several angles to the frontal body plane to a target location near the subjects' body. The area on the table containing the axial directions from 90° to 160° counterclockwise from the right side of the frontal body plane was found to place the greatest amount of strain on the shoulder muscles with 150° being the most unfavorable and 30° found to be the most optimal (Strasser & Muller, 1999). McDonald et al., (2012) measured the strain of several muscles of the shoulder-arm system during push/pull exertions at locations along X, Y, and Z axes corresponding to the frontal, sagittal, and transverse planes respectively. They found that for pulling exertions along the Y-axis, total muscle activity for the shoulder muscles decreased as the hand location moved forward. For the X-axis, muscle activity was parabolic with higher activity occurring at the extreme left and right positions tested. For the Z-axis, activity increased with an increase in the vertical position. For pushing exertions, the X positions showed a similar parabolic pattern as in the pulling exertions. For the Z-axis, only the highest vertical hand location was significantly different than any of the other hand locations. The Y-axis paralleled the pulling results, but to a less pronounced extent (McDonald et al., 2012).

In another study, Chopp et al., (2010), measured EMG from the shoulder muscles during seated overhead exertions. The subjects were asked to exert a specified force level on a force transducer located above their heads while pulling backwards, pushing forwards, downwards, sideways, and upwards in four hand position. Pulling backwards showed the highest total muscular demand. Sideways pushing produced higher muscle activity than other directions (Chopp et al., 2010). Overall results suggested that, if possible, positioning overhead work as close to the worker as possible will result in lower upper extremity muscle demand. Another study conducted by Anton et al., (2001) to investigate the effect of overhead drilling positions on shoulder joint moment and electromyography yielded a similar conclusion. The authors found that moving the task closer to the worker decreased anterior deltoid and biceps activity and shoulder moment.

Several previous studies have also investigated the biomechanical loading of the shoulder taking into consideration the joint reaction forces present at the glenohumeral joint. Hoozemans et al., (2004) evaluated the mechanical load on the low back and shoulders during cart pushing and pulling using one or two hands, three different cart weights, and two handle heights. They found that the exerted force and handle height both had a considerable effect on the mechanical loading of the shoulder. Their recommendation was that cart weight should remain as low as possible and to push or pull at shoulder height with the general idea that the net shoulder moment is kept lower by keeping the shoulder joint close to the line of action of the exerted force (Hoozemans et al., 2004). Nimbarte et al., (2013) also evaluated the effects of a dynamic cart pushing task on the biomechanical loading of the shoulder and low back. In this study, subjects performed dynamic cart pushing tasks on a walkway of varying gradient (0°, 5°, and 10°) using three different cart weights (20, 30, and 40 kg). Peak reaction forces at the acromioclavicular and glenohumeral joints were found to be comparable to one another and were higher than the peak forces at the sternoclavicular joint. Variation on the cart weight was found to significantly affect the reaction forces at the shoulder complex joints. The peak reaction forces for all three shoulder joints increased with an increase in cart weight suggesting that higher exertion forces required to push the cart resulted in a higher joint loading. For the glenohumeral joint, reaction forces in the distraction (medial-lateral) direction were found to be substantially higher than the reaction forces in the anterior-posterior and inferior-superior directions. Another study by de Looze et al., (2000) investigated the changes in force direction of pushing and pulling as result of changes in handle height and force level. An increase in the push/pull force exertion level was reflected in an increased net shoulder torque.

In summary, a significant stress-strain relationship for the shoulder complex during physically demanding exertions has been reported in the previous physiological and biomechanical investigations. However, a clear assessment method or criteria to evaluate the risk of injury to the shoulder complex during forceful arm exertions currently does not exist. The complexity of the shoulder contributes to this lack of research. The shoulder complex is the most mobile part of the human body due to its high functional degrees of freedom. It comprised of 1) the glenohumeral (GH) joint, where the head of the humerus articulates with the glenoid fossa of the shoulder, 2) the sternoclavicular (SC) joint, where the superior aspect of the sternum articulates with the medial end of the clavicle, 3) the acromioclavicular (AC) joint, where the acromion process of the scapula articulates with the lateral end of the clavicle, and 4) the scapula-thoracic joint, where the scapula slides against the rib cage. The majority of shoulder MSDs involve the glenohumeral joint (Culham & Peat, 1993).

The glenohumeral joint, provides much of the shoulder's mobility functioning as a ball-and-socket type joint. The glenohumeral joint is characteristically unstable in that the humeral head is not fully encapsulated by the glenoid with only around 30% of the humeral head in contact with the glenoid in various shoulder postures (McCluskey & Getz, 2000). Because of this, the glenohumeral joint is typically stabilized by the forces produced by the shoulder muscles that press the humeral head into the glenoid cavity through a mechanism called "concavity compression" (Figure 1) (Konrad et al., 2006).

Currently it is not well understood as to how the concavity compression mechanism works to stabilize the GH joints during different types of physical exertions. In this study a biomechanical model of shoulder complex was used to evaluate mechanical loading of the GH joint during forceful arm exertions. The exertions were performed in six anatomical directions using different force levels. Based on the concept of concavity compression, where the shoulder muscle exerts forces to compress humeral head into the glenoid cavity, it was hypothesized that the angular position of the resultant muscle force vector at the GH joint would not be affected by the direction and the magnitude of the force exertion.

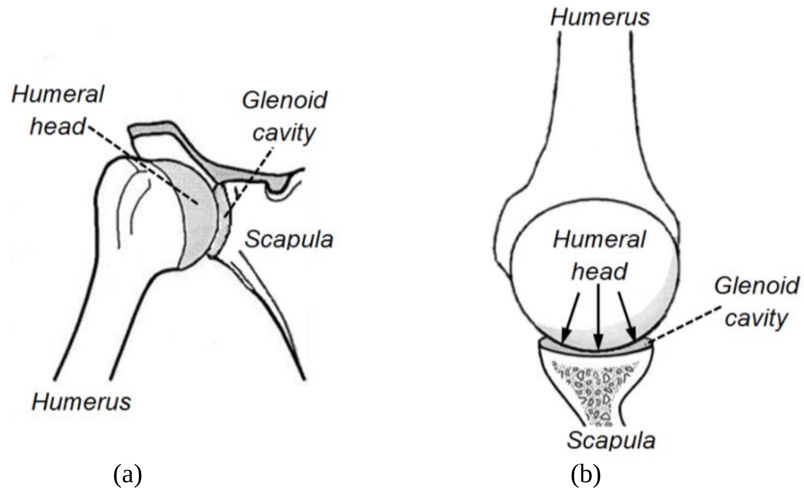

Figure 1. (a) Position of the humeral head with respect to the glenoid cavity; (b) Concavity compression mechanism

2. Method

2.1 Approach

Biomechanical modeling analysis was performed by simulating forceful arm exertions in six orthogonal directions using two force levels. A well-established and previously tested biomechanical model of the shoulder complex (Nimbarte et al., 2013) was used to quantify the muscle forces placed on the glenohumeral joint due to the exertions. The angular position of the resultant muscle force vector in 3D space was estimated using planer angle.

2.2 Biomechanical Model

The resultant reaction forces acting at the glenohumeral joint during the forceful arm exertions were estimated using the AnyBody Modeling System™ (version 5.0, AnyBody Technology, Aalborg, Denmark). This is a full-body biomechanical modeling system where models are formulated using AnyScript modeling language, which is similar in syntax to the C++ computer language. It is an object-oriented language developed specifically to define and model bones, joints, and muscle-tendon units based on real physiological properties. Joints in the AnyBody modeling system can be driven by experimentally obtained kinematic and kinetic data. Muscle and joint forces are computed using inverse dynamics analysis. AnyBody musculoskeletal models have been used previously to estimate musculoskeletal shoulder loading during cart pushing and pulling (Nimbarte et al., 2013) and wheelchair propulsion (Dubowsky et al., 2008) tasks.

2.3 Experimental Design

A two-factor design was used in this research. Factor 1, direction of force exertion, was treated at six levels: 1) anterior, 2) superior, 3) lateral, 4) posterior, 5) inferior, and 6) medial. Factor 2, force exertion level, was treated at two levels: 1) 40 N, and 2) 60 N. The force exertion levels used were obtained based on the findings of our preliminary study where the average maximum force that individuals were able to exert in anterior, superior and lateral directions were 75 (21) N, 111 (32) N, and 220 (54) N, respectively (Cutlip et al., 2013). The force exertion values used in the modeling analysis were kept below the maximum strengths to ensure realistic representation of the human participants' physical strength.

2.4 Modeling Analysis

2.4.1 Subject

The modelling analysis was performed for five male and five female participants. The mean (SD) age, weight and height of the participants were 26.5 (4.6) years, 65.3 (10.9) Kg, and 169.9(9.1) cm, respectively. These participants were originally recruited for a different study on shoulder strength measurement (Cutlip et al., 2013).

2.4.2 Model Description

The shoulder model used in this study consists of 118 muscle fascicles and defines the three main shoulder complex joints: the glenohumeral joint, the acromioclavicular joint, and the sternoclavicular joint. This model is part of the public-domain model repository provided by AnyBody Technology as part of their AnyBody Modeling System™.

In AnyBody Modeling System, the muscle force required to generate motion or sustain body posture is computed using inverse-dynamic methods by solving a multi-body dynamics problem. The muscle recruitment in the inverse dynamics process is solved using a linear optimization procedure (Rasmussen et al., 2001), within which the objective function is to minimize the maximal normalized muscle force. The objective function of the optimization procedure is:

$$G\left(f^{(M)}\right) = max\left(\frac{f_i^{(M)}}{N_i}\right) \qquad (1)$$

Subjected to the following constraints:

$$Cf = d \qquad (2)$$

$$\left(f_i^{(M)}\right) \geq 0; i \in \left\{1, \dots, n^{(M)}\right\} \qquad (3)$$

Where, **f** is the vector of $n^{(M)}$ unknown muscle forces, $\mathbf{f}^{(M)}$, and joint reactions, $\mathbf{f}^{(R)}$. N_i is the momentary strength of muscle *i*. *C* is the coefficient matrix for the ‘‘unknown’’ forces/moments in the system while *d* is a vector of the ‘‘known’’ (applied or inertia) forces.

The lower bound ($f_i^{(M)} \geq 0$) simply states that muscles can only pull (not push) and that the upper bound for the force in each muscle *i* is the corresponding muscle strength, N_i;

2.4.3 Model Input and Scaling

The input to the biomechanical model included the external force (magnitude and direction) and participant anthropometric and characteristic data such as height, weight, and gender. In order to scale the model for individual participant (using AnyBody Modeling System™ files with extension .any) ScalingUniform.any and AnyManUniform.any within the model were used to make the appropriate participant specific changes to the model. The ScalingUniform.any file takes the individual participant total body mass and height data, defined by the user in the AnyManUniform.any file, and uses these two properties to proportionally scale the model’s segments in all three directions. This allows the model to be able to simulate the unique anthropometric characteristics of each participant, for example whether they are tall and lean or short and stout. Additionally, gender specific changes were made to the model such as changes to the model’s defined segment lengths and masses and changes to the model’s body fat composition in the AnyManUniform.any file. Without changing the default model values for segment length and mass as well as body fat percentage, which are based on male anthropometry measures, the model would not be able to differentiate between male and female participants and would simply use the default male values for both. Using data available from Chaffin et al. (2006), individual model segments such as the upper arm, forearm, hand, etc. were adjusted for segment length and segment mass for both males and females which allowed the model to properly scale the individual segments depending on the gender of the participant. For body fat percentage, equations by Frankenfield et al. (2001) were used to define the model’s body fat percentage for both males and females. To run the simulation, a 90 degree flexed elbow posture was used (Figure 2). The force and posture data were implemented into the model where it then runs the appropriate calibration and calculations to resolve the model constraints. As mentioned in the experimental design section, the force input to the model was marinated at two levels (40 and 60 N) and in six anatomical directions (anterior, superior, lateral, posterior, inferior, and medial).

2.4.4 Model Output

The outcomes of the biomechanical analysis included the reaction forces acting on the right glenohumeral joint of the shoulder complex in the following anatomical directions: distraction (medial-lateral (Z)), inferior-superior (Y), and anterior-posterior (X) (Figure 2).

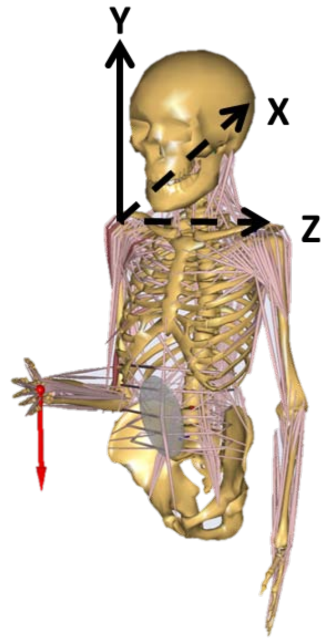

Figure 2. Graphical representation of AnyBody biomechanical model

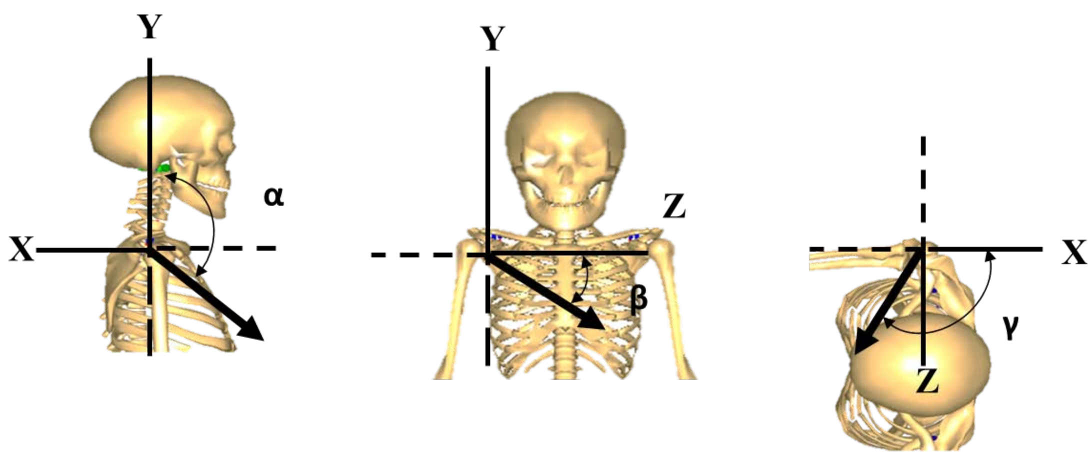

Figure 3. Two-dimensional angular deviations for the resultant muscle force vector

2.5 Data Analysis

The resultant muscle force vector was mirrored onto a two dimensional plane to calculate the angle the vector makes with each of the three axes X, Y, and Z (Figure 3). Angles α, β, and γ correspond to the angles the resultant force makes with the Y, Z, and X axes, respectively, and were calculated using the following equations:

$$\alpha = 90 + tan^{-1}\left(\frac{Fy}{Fx}\right) \tag{4}$$

$$\beta = tan^{-1}\left(\frac{Fy}{Fz}\right) \tag{5}$$

$$\gamma = 90 + tan^{-1}\left(\frac{Fx}{Fz}\right) \tag{6}$$

3. Results

The mean resultant muscle force vector was found to be located at an angle of 160 degrees with respect to the positive Y-axis (α), 40 degrees with respect to the positive Z-axis (β), and 107 degrees with respect to the positive X-axis (γ) (Figure 4).

The position of the resultant muscle force vector with respect to the positive Y-axis (α) ranged between 155 to 168 degrees for the exertions tested in this study (Table 1). With the increase in the weight from 40 to 60 N the angular position changed by about 9 degrees. Change in the direction of external force application produce a change of about 13 degrees in the direction of the resultant muscle force vector.

Comparatively a much smaller change in the position of the resultant muscle force vector with respect to the positive Z-axis (β) was observed (Range 36 to 42 degrees). The increase in the force from 40 to 60 N changed the position of the resultant muscle force vector by 1 degree; and change in the direction of external force application produce a change of about 6 degrees.

Table 1: Mean angular position of the resultant muscle force vector
(Numbers in parenthesis represents one standard deviation)

		A	**β**	**γ**
Force	**40 N**	155.0(9.55)	39.5(2.30)	110.9(7.58)
	60 N	164.3(3.17)	40.2(2.67)	103.5(3.52)
Direction	**Anterior**	155.3(6.26)	40.0(2.12)	110.9(4.51)
	Superior	154.2(9.06)	40.3(1.89)	112.0(6.82)
	Lateral	155.7(13.2)	38.5(1.59)	109.8(10.7)
	Posterior	161.7(0.67)	41.8(0.62)	106.4(0.33)
	Inferior	163.3(0.74)	42.1(0.16)	105.1(0.76)
	Medial	167.7(1.31)	36.5(2.24)	99.1(1.08)

The position of the resultant muscle force vector with respect to the positive X-axis (γ) changed between 99 to 112 degrees. The increase in the force from 40 to 60 N changed the position of the resultant muscle force vector by about 7 degrees; and change in the direction of external force application produce a change of about 13 degrees.

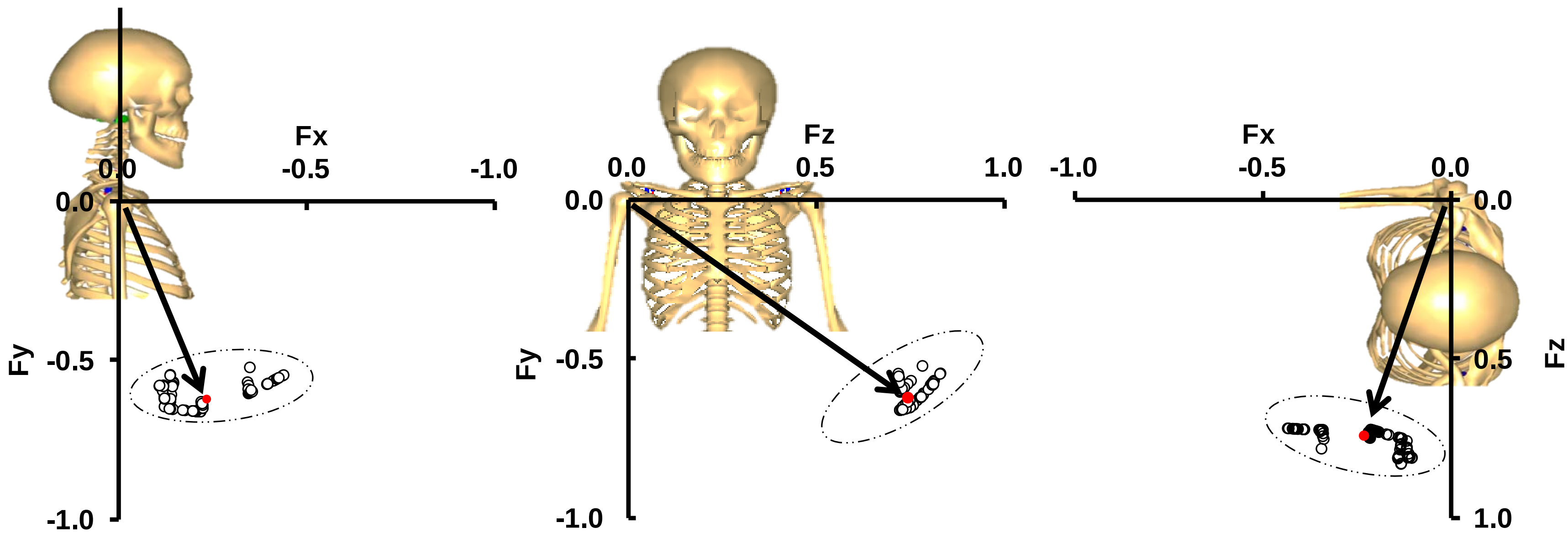

Figure 4. The angular positions of the unit vectors for the resultant muscle force in the sagittal, frontal and transverse planes. In each figure, circles represent the positions of the unit vectors for the exertions performed under different conditions of direction and magnitude and red circle represents the mean position.

4. Discussion

In this study the role played by the concavity compression mechanism in stabilizing the GH joint was evaluated by using angular positions of the resultant muscle force vectors. The resultant force vectors generated by the shoulder muscles were almost always directed anteriorly, medially, and inferiorly, independent of the magnitude and the direction of the external force application. The data supports our hypothesis and corroborates that the concavity compression mechanism is used by the shoulder muscles to improve its stability.

A total of 1200 modeling simulations were used in the data analysis. The force exertion direction was controlled at six mutually orthogonal anatomical directions. However, the directions of the resultant muscle force vectors were condensed in a relatively narrow region. This suggests that shoulder muscles always work toward pulling the humeral head into glenoid cavity.

Therefore, for a physical exertion with the resultant forces due to the external loading (i.e., without considering the muscle contribution) directed laterally outward, the shoulder muscles have to work under duress to re-direct the resultant force vector towards the shoulder socket to improve joint stability. On the other hand if the resultant forces due to the external loading promotes the concavity compression i.e., directed anteriorly, medially, and inferiorly then that could reduce the work demand on the shoulder muscles The concept of angular deviation used in this study to evaluate shoulder stabilizing concavity mechanism could be used to measure strain imposed on the shoulder during physically demanding exertions. The resultant of the reaction forces obtained under the without-muscle condition takes into account only the forces that are caused by external loading and is always opposite to the direction of the external force application. On the other hand, the resultant forces generated by the shoulder muscles must instead be redirected towards the shoulder in order to improve the joint's stability by enhancing the concavity compression. Thus, the change in the angular deviation between the resultant force vectors of these two conditions may provide a direct assessment of the strain experienced by the shoulder complex during forceful arm exertions (Figure 5).

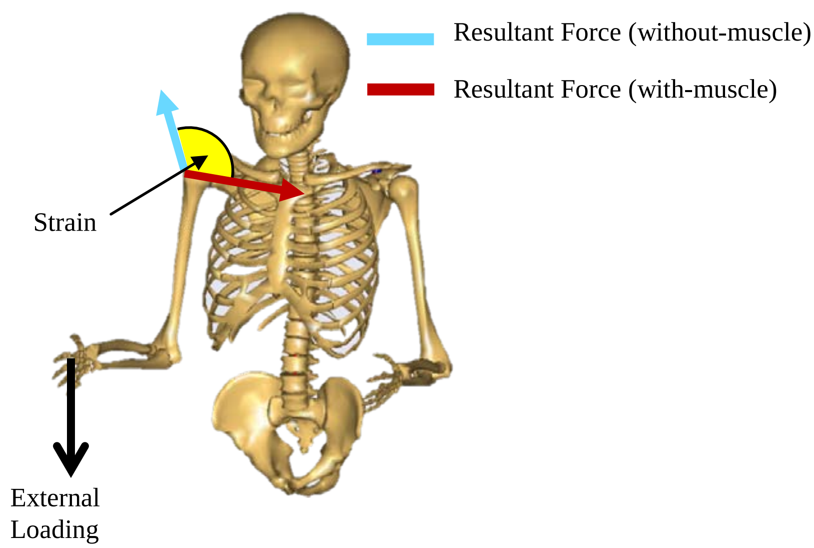

Figure 5. Resultant force vectors as result of external force (blue) and muscles force (red). The yellow highlighted area could be a possible measure of the amount of strain experienced by the shoulder caused by a forceful arm exertion.

The result of this study indicates that the compression of the humeral head against the glenoid is desirable and most of the shoulder muscles strive to achieve it. However there seems to be a lack of a mechanism that could prevent the passive compression (pinching) of the musculature. This is especially critical and strenuous if the reaction force due to the external force application pinches the passive structures (joint capsule, ligaments, etc.) which can potentially lead to tendinitis, rotator cuff tears, nerve impingement, etc. (Staal et al., 2007; van Rijn et al., 2010).

In summary, computer simulations and biomechanical model of shoulder complex were used to study the concavity compression mechanism. Study findings indicate that during forceful arm exertions, the shoulder stabilizer muscles compress or pull the upper arm (head of the humerus) into the shoulder socket (glenoid) by counteracting the forces generated by external loading. In this process the resultant vector of the summed muscle forces always attempts to center the humeral head in the glenoid. The knowledge gained in this study could possibly be used to quantify strain imposed on the shoulder muscle during forceful arm exertions. Future studies should evaluate the model predicted angular position data using objective (electromyography) and subjective (perceived exertions ratings) measurements.

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