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Postural differences in shoulder dynamics during pushing and pulling

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ABSTRACT

Assessments of shoulder dynamics (e.g. the inertial, viscous, and stiffness properties of the joint) can provide important insights into the stability of the joint at rest and during volitional contraction. The purpose of this study was to investigate how arm posture influences shoulder dynamics while generating pushing or pulling torques in the horizontal plane. Sixteen healthy participants were examined in seven postures encompassing a large workspace of the shoulder. At each posture, the participant's shoulder was rapidly perturbed while measuring the resultant change in shoulder torque about the glenohumeral axis. Participants were examined both at rest and while producing horizontal flexion and extension torques scaled to 15% of a maximum voluntary contraction. Shoulder stiffness, viscosity, and damping ratio were estimated using impedance-based matching, and changes in these outcome measures with torque level, elevation angle, and plane of elevation angle were explored with a linear mixed effects model. Shoulder stiffness was found to decrease with increasing elevation angles ($p < 0.001$) without subsequent changes in viscosity, leading to a greater damping ratios at higher elevation angles ($p < 0.001$). Shoulder stiffness, viscosity, and damping ratio (all $p < 0.05$) were all found to significantly increase as the plane of elevation of the arm was increased. The relationship between the viscosity, stiffness and the damping ratio of the shoulder is one that the central nervous system must regulate in order to maintain stability, protect against injury, and control the shoulder joint as the inertial and muscle contributions change across different arm postures.

1. Introduction

The shoulder is a highly mobile complex (including glenohumeral, scapulothoracic, acromioclavicular, and sternoclavicular articulations) that moves the upper extremity through a broad range of motion during activities of daily living. The majority of daily tasks involving the shoulder, including reaching and grasping, take place with the shoulder adducted below eighty degrees (Coley et al., 2008). A shoulder with greater postural stability has greater resistance to rotational movement in response to a postural perturbation (Riemann and Lephart, 2002). At rest, this stability is provided by passive structures including the glenoid fossa, ligaments, and tendons (Bigliani et al., 1996). During dynamic tasks, these passive structures continue to stabilize the shoulder, but postural stability is largely achieved by the coordinated activation of shoulder musculature (Blasier et al., 1997; Wuelker et al., 1998).

Assessments of shoulder dynamics (e.g. the inertial, viscous, and stiffness properties of the joint (Perreault et al., 2004)) can provide important insights into the stability of the joint at rest and during volitional contraction. Clinically, shoulder stiffness is evaluated by comparing the resistance of the affected arm when moved passively through a rotational range of motion to the unaffected arm (Goldberg et al., 1999). However, this clinical evaluation is highly subjective. Quantitative measurements of shoulder stiffness can provide improved insights into the static stability of the shoulder in response to a postural disturbance. Prior studies have shown that shoulder stiffness increases with increased muscle activity (Zhang et al., 2000). However, it is unclear how changes in arm posture affect shoulder stiffness. Previous studies have only directly assessed shoulder stiffness in a single posture, including the arm abducted 90° (Lipps et al., 2015) or elevated 45° in the scapular plane (Zhang et al., 2000).

Postural stability is also provided by viscous damping at the joint, which quantifies the dynamic response of the shoulder to a postural disturbance. Increasing muscle contraction produces greater joint stiffness. For a given arm inertia, viscosity must also increase to allow for fluid shoulder movement as an overly

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compliant joint with little damping would be instable and cause the joint to be jerked around from one position to another (Zhang et al., 2000). Therefore, the shoulder remains a relatively constant underdamped system during volitional contractions (Zhang et al., 2000). However, little is known regarding how the viscous damping properties of the shoulder are influenced by arm posture. Increases in elevation angle produces decreased viscosity under passive conditions (Engin, 1984), but viscosity during muscular contractions was not explored.

Postural changes to the dynamic properties of the shoulder are likely influenced by the muscles that stabilize the joint. Demand from shoulder muscles are influenced both by the task performed (e.g. pushing or pulling) and shoulder elevation angle (McFarland et al., 2018). The moment produced by each shoulder muscle will destabilize the shoulder in a specific magnitude and direction, but the controlled activation of all shoulder muscles allow for joint stability across the entire range of motion (Lee and An, 2002; Van der Helm et al., 1992). Changes in arm posture will affect how much each muscle contributes to joint stability. For example, the resultant muscle force vector produced by the shoulder musculature reduces postural stability as the shoulder is elevated (Labriola et al., 2004). This reduced stability near the end range of motion is primarily caused by destabilizing forces from the pectoralis major and deltoid muscles (Labriola et al., 2005).

The dynamics of the shoulder are important for maintaining stability throughout the full range of motion, but there is a knowledge gap regarding how joint dynamics change in different arm postures. Therefore, the purpose of this study was to investigate how arm posture influences shoulder dynamics as participants produced pushing or pulling torques. Robot-assisted measures of joint stiffness and viscosity were acquired as participants were examined across a large range of postures encompassing the mobility of the shoulder joint. We hypothesized that decreasing shoulder elevation would increase both stiffness and viscosity as these shoulder positions are more often used for daily tasks, thus requiring greater stability. This study aimed to better identify shoulder postures with lower postural stability and increased risk of injury, which could be important in a clinic setting.

2. Materials and methods

2.1. Participants

Eight male and eight female healthy adults (mean ± SD age 21.6 ± 0.9 years, weight 63.2 ± 13.2 kg, and height of 1.68 ± 0.11 m) with no history of neuromuscular pathologies or shoulder injury were recruited for this study. All participants provided written consent to the testing procedures approved by the University of Michigan Institutional Review Board (HUM00140520). The dominant arm of each participant was examined, resulting in fifteen right shoulders and one left shoulder being examined.

2.2. Experimental setup

Participants sat in a modifiable chair (Biodex Medical Systems, Shirley, New York) with torso movement restricted by placing straps across the torso and positioning cushioned plates along the lower back and sides. Motion of the scapula was not restricted. The participant's upper extremity was placed into a plastic removable cast that fixed the wrist in a neutral posture and the elbow in 90 degrees of flexion and supported it against gravity. The plastic cast contained embedded aluminum plates that attached the cast to the crank arm of a computer-controlled brushless servomotor (Baldor Electric Company, Fort Smith, AR). The participants chair was moved to ensure the center of rotation of the glenohumeral joint, identified and palpated as the medial portion of the acromion, was aligned with a laser pointer that projected from the axis of rotation of the high-precision rotary motor. The motor was position-controlled in real-time using the Simulink Real Time toolbox in MATLAB (v2016a, Mathworks Inc, Natick, MA). Forces and torques were measured using a six degrees-of-freedom load cell between the cast and crank arm (45E15A4, JR3 Inc., Woodland CA). Our measurement coordinate system at the glenohumeral joint was defined using established guidelines by the International Society of Biomechanics, where the plane of elevation is the first Euler angle (θ), elevation is the second Euler angle (ϕ), and rotation is the final Euler angle (ψ) (Wu et al., 2005).

Surface electromyography (EMG) electrodes (Bagnoli system, Delsys, Natick, MA) were placed on nine key muscles surrounding the shoulder to measure changes in muscle activity across various shoulder postures and tasks. The skin was prepared with exfoliant and alcohol wipes, and electrodes were attached to the skin with double-sided tape. Electrodes were placed over the muscle bellies of the sternocostal (PM SC) and clavicular regions (PM CL) of pectoralis major, the three heads of the deltoids (anterior deltoid: AD, middle deltoid: MD, posterior deltoid: PD), the upper (UT) and lower trapezius (LT), latissimus dorsi (LD), and serratus anterior (SA). A ground electrode (Dermatrode, Irvine, CA) was placed on the acromion process of the opposite shoulder.

2.3. Experimental design

All testing was performed in a single, two-and-a-half-hour session. Participants were examined with their shoulder positioned at seven different postures that encompassed a wide variety of arm postures, replicable to postures produced during daily activity. These seven postures were randomized in order to control for the effects of fatigue and exercise on joint stiffness. Four of these postures included testing at 90° shoulder elevation, with plane of elevation angles at 0°, 30°, 60°, and 90°. Participants were also examined in three other arm postures with plane of elevation angles and elevation angles, respectively, of 0° and 45°, 30° and 60°, and 60° and 75° (Fig. 1). These seven different shoulder postures were achieved by a combination of rotating the position of the motor, the motor crank arm and the base of the Biodex chair. Special care was given to ensure the rotational axis of the glenohumeral joint aligned with the axis of the crank arm in all postures. At the beginning of each arm posture, participants were asked to generate maximum voluntary contractions (MVC) in horizontal flexion and extension in order to normalize the prescribed torque during the experimental trials to each participant's MVC. The experimenters provided a visual demonstration of the flexion/extension prior to the collection of MVCs, as certain postures had flexion/extension directions that were not intuitive.

Shoulder dynamics were measured by applying small, stochastic perturbations in the horizontal flexion/extension plane with a pseudo-random binary sequence (0.06 rad (3.44°) amplitude and 150 ms switching interval) while measuring the resultant change in shoulder torque in this plane (Lipps et al., 2015). The perturbations were small in magnitude and centered about the axis of rotation of the glenohumeral joint. Each perturbation trial lasted 60 s, during which the participant was asked to either relax or produce torque at 15% MVC in horizontal flexion or extension. Participants were provided visual feedback regarding their real-time 3D shoulder torque production on an LCD monitor to assist with the task. This insured participants only generated torques in horizontal flexion/extension and not in any off-axis directions. Two perturbation trials were performed at each of the three torque conditions (at rest, 15% MVC in the flexion/extension plane) in the seven different shoulder postures, yielding 42 total trials. To control for the effects

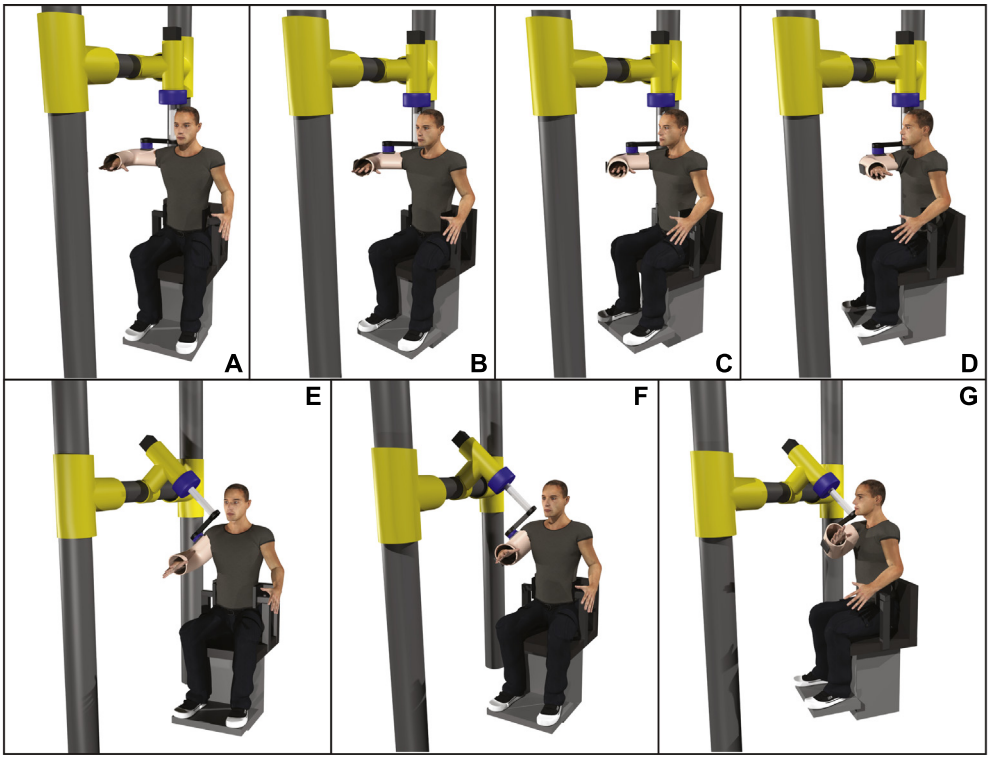

Fig. 1. The experimental setup attached each participant's dominant arm to the crank arm of a single axis rotary motor using a plastic cast. A six degrees-of-freedom load cell measured forces and torques about the glenohumeral joint. Participants were examined in seven arm postures with plane of elevation and elevation angles, respectively, of (A) 0° and 90°, (B) 30° and 90°, (C) 60° and 90°, (D) 90° and 90°, (E) 0° and 45°, (F) 60° and 30°, and (G) 75° and 60°.

of fatigue, participants were given breaks before and after each MVC collection, and as needed between trials.

2.4. Data analysis

A single-input, single-output system was used to measure shoulder impedance, the dynamic relationship between the torque response to an imposed change in joint position. Impedance was estimated using a frequency response function (FRF) from 0 to 10 Hz (Perreault et al., 2001). The dynamic properties of the shoulder were approximated by fitting the FRF with a 2nd order linear system with inertial (*I*), viscous (*B*) and stiffness parameters (*K*) (Perreault et al., 2001). The damping ratio ζ was then calculated using Eq. (1) as:

$$\zeta = \frac{B}{2\sqrt{IK}} \tag{1}$$

The accuracy of this parametrization was measured using the variance accounted for (%VAF) between the measured torque and the torque predicted by the parametric model. Each experimental trial was reviewed using a custom-written MATLAB algorithm to insure a minimum of 30 s of data was used to accurately measure joint dynamics. This algorithm insured accuracy of our parametric fitting was not influenced by transient shifts in torque production at the beginning or end of each trial. The EMG data were detrended, rectified, low-pass filtered at 5 Hz, and normalized to the maximal EMG for each muscle recorded during the MVC trials. Muscle activity was measured for all nine muscles as the average muscle activation across each trial.

All statistical procedures were performed in SPSS (v24, IBM Corporation, Chicago, IL, USA). We examined our hypothesis that decreasing shoulder elevation would increase stiffness, viscosity and damping ratio using a linear mixed model. Shoulder stiffness, viscosity, damping ratio, EMG activity (by each muscle), and parametric %VAF were treated as outcome measures. Muscle activation level (-15%, 0%, 15% MVC), plane of elevation (45°, 60°, 75°, 90°) and elevation angle (0°, 30°, 60°, 90°), were treated as within-subjects factors, and subjects were treated as a random intercept. Bonferroni-corrected pairwise comparisons were used to analyze significant main effects at a significance level of $p < 0.05$.

3. Results

3.1. Identification and parameter fitting

The overall mean %VAF ± SE for all subjects and trials was 86.4 ± 3.3%, indicating our parametric fits to estimate glenohumeral stiffness were robust. There was a significant main effect of elevation angle on %VAF ($F_{3\ 225} = 24.01$, $p < 0.001$). The elevation angle of 90° (74.7 ± 2.9%) produced statistically significant lower horizontal %VAFs than the elevation angles of 75° (92.6 ± 3.8%), 60° (87.1 ± 3.7%) and 45° (90.9 ± 3.8%) (all $p < 0.001$), which could indicate participants had increased difficulty maintaining steady torques at higher elevation angles.

3.2. Shoudler dynamics

The measured shoulder stiffness, viscosity, and damping ratio across the seven experimental postures and three torque levels are shown in Table 1.

Shoulder stiffness was found to be significantly affected by changes in torque level ($F_{2\ 100} = 149.8$, $p < 0.001$), plane of elevation ($F_{3\ 257} = 2.964$, $p = 0.033$) and elevation angle ($F_{3\ 216} = 35.5$, $p < 0.001$) (Fig. 2). Stiffness was statistically greater when participants generated flexion or extension torque when compared to at rest (both $p < 0.001$). At an elevation angle of 90°, shoulder stiffness was statistically lower when compared to the elevation angles of 75°, 60° and 45° (all $p < 0.001$). Additionally, shoulder

Table 1
Stiffness, viscosity, and damping ratios across the seven shoulder postures and three prescribed torques. Values are reported as mean (standard error).

		−15% MVC		
Elevation Angle	Plane of Elevation Angle	Stiffness (Nm/rad)	Viscosity (Nm-s/rad)	Damping Ratio
90	0	38.8 (3.8)	2.3 (0.4)	0.23 (0.04)
	30	40.8 (4.0)	2.6 (0.5)	0.24 (0.04)
	60	42.4 (4.8)	3.0 (0.4)	0.29 (0.03)
	90	45.2 (5.7)	2.6 (0.4)	0.28 (0.04)
75	60	55.3 (5.2)	3.2 (0.3)	0.21 (0.02)
60	30	60.9 (4.8)	3.0 (0.3)	0.19 (0.02)
45	0	61.8 (4.1)	3.2 (0.4)	0.21 (0.04)
		0% MVC		
90	0	10.2 (1.4)	1.3 (0.3)	0.24 (0.04)
	30	10 (1.2)	1.7 (0.3)	0.30 (0.04)
	60	9.1 (0.8)	1.5 (0.3)	0.27 (0.05)
	90	10.7 (1.6)	1.9 (0.3)	0.39 (0.07)
75	60	20.5 (1.8)	1.4 (0.2)	0.18 (0.05)
60	30	31.8 (2.0)	1.2 (0.2)	0.10 (0.02)
45	0	31.2 (1.4)	1.3 (0.2)	0.11 (0.01)
		15% MVC		
90	0	45.2 (4.8)	2.7 (0.4)	0.24 (0.02)
	30	38.5 (5.4)	3.9 (0.5)	0.45 (0.06)
	60	47.7 (5.0)	3.4 (0.4)	0.31 (0.03)
	90	55.9 (5.8)	3.8 (0.5)	0.32 (0.03)
75	60	60.7 (7.3)	4.2 (0.4)	0.28 (0.03)
60	30	55.4 (5.7)	4.1 (0.4)	0.30 (0.04)
45	0	65.2 (6.2)	3.6 (0.4)	0.22 (0.02)

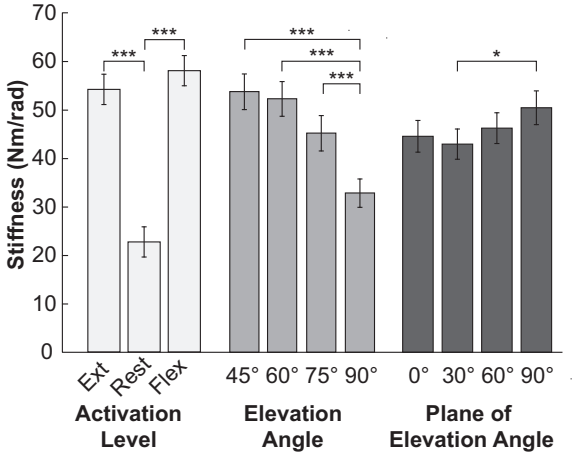

Fig. 2. Shoulder stiffness was sensitive to changes in activation level, elevation angle, and plane of elevation angle. Stiffness was tested at three different activation levels (15% max extension, at rest, and 15% max flexion) at four plane of elevation angles (0°, 30°, 60° and 90°) and four elevation angles (45°, 60°, 75° and 90°). Bars represent mean stiffness ± standard error (* = significance at $p < 0.05$, ** = significance at $p < 0.01$, and *** = significance at $p < 0.001$).

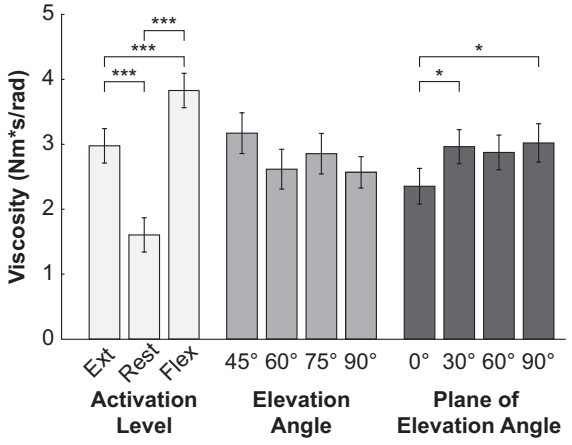

Fig. 3. Shoulder viscosity was influenced by changes in activation level and plane of elevation angle, but not elevation angle. Viscosity was tested at three different activation levels (15% max extension, at rest, and 15% max flexion) at four plane of elevation angles (0°, 30°, 60° and 90°) and four elevation angles (45°, 60°, 75° and 90°). Bars represent mean viscosity ± standard error (* = significance at $p < 0.05$, ** = significance at $p < 0.01$, and *** = significance at $p < 0.001$).

stiffness was statistically lower at a plane of elevation angle of 30° compared to the maximum plane of elevation angle of 90° ($p = 0.021$).

Viscosity was also found to be significantly affected by changes in torque level ($F_{2\ 100} = 67.07$, $p < 0.001$), plane of elevation ($F_{3\ 251} = 3.77$, $p = 0.011$) and elevation angle ($F_{3\ 202} = 3.02$, $p = 0.031$) (Fig. 3). Viscosity was statistically greater when participants generated flexion torque when compared to extension torque ($p < 0.001$) and at rest ($p < 0.001$), and also greater when participants generated extension torques compared to when relaxed ($p < 0.001$). Viscosity was statistically lower at a plane of elevation angle of 0° when compared to plane of elevation angles of 30° ($p = 0.020$) and 90° ($p = 0.034$).

The shoulder behaved as an underdamped system across all postures and activation levels. There was a significant main effect of torque level ($F_{2\ 104} = 6.11$, $p = 0.003$), plane of elevation ($F_{3\ 241} = 4.54$, $p = 0.004$) and elevation angle ($F_{3\ 174} = 7.60$, $p < 0.001$) on damping ratio (Fig. 4). The damping ratio was statistically greater when participants generated flexion torque when compared to extension torque ($p = 0.014$) and at rest ($p = 0.007$). The damping ratio was significantly different between the 60° and 90° elevation angles ($p < 0.001$), with the lowest damping ratios at 60° and the largest damping ratio at 90°. The damping ratio was lowest at a plane of elevation angle of 0°, which was statically different than when the plane of elevation was 30° ($p = 0.01$) and 90° ($p = 0.016$).

3.3. Muscle activity

As expected, all muscles increased activation as torque increased compared to when at rest (all $p < 0.05$) (Fig. 5). Flexion torques, compared to extension torques and at rest, produced

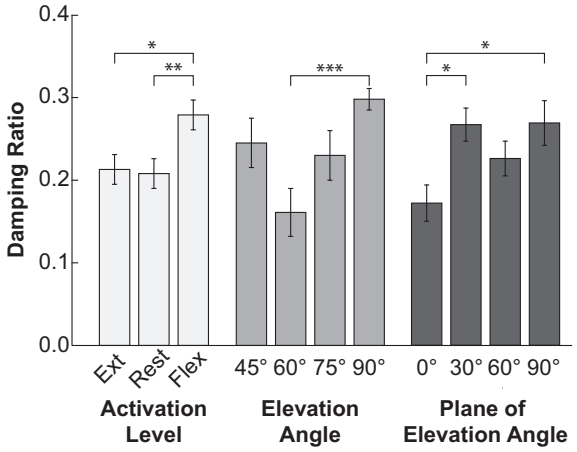

Fig. 4. The shoulder behaved as an underdamped system across all examined conditions. There were significant effects of activation level, elevation angle, and plane of elevation angle on the damping ratio. Damping ratio was tested at three different activation levels (15% max extension, at rest, and 15% max flexion) at four plane of elevation angles (0°, 30°, 60° and 90°) and four elevation angles (45°, 60°, 75° and 90°). Bars represent mean damping ratio ± standard error (* = significance at $p < 0.05$, ** = significance at $p < 0.01$, and *** = significance at $p < 0.001$).

significantly greater activations for the PM SC, PM CL, AD, and SA (all $p < 0.05$). The MD, PD, UT and LT all produced significantly greater activations for extension torques when compared to flexion torques and at rest (all $p \leq 0.001$). The PM SC, PD, and LD were found to be significantly affected by changes in elevation angle [($F_{3\ 168} = 6.06$, $p = 0.001$), ($F_{3\ 210} = 4.68$, $p = 0.003$), ($F_{3\ 198} = 3.47$, $p = 0.017$), respectively]. The PM SC showed significant activation differences between the elevation angles of 60° and 90° ($p = 0.034$), with the lowest activation at 60° and the largest activation at 90°. The LD activation was significantly different between elevation angles of 45° and 60°, and 45° and 75° (all $p < 0.05$) with the lowest activation at 45° and the greatest at 75°. Additionally, the PM SC, AD, and UT were found to be significantly affected by changes in plane of elevation angle [($F_{3\ 248} = 7.60$, $p < 0.001$), ($F_{3\ 266} = 3.72$, $p = 0.012$), ($F_{3\ 251} = 13.27$, $p < 0.001$), respectively]. The PM SC activation was significantly different between plane of elevation angles of 30° and 60° ($p = 0.002$), and 30° and 90° ($p < 0.001$), with the greatest activation at 30° and the lowest at 90°. The UT activation was the only muscle to follow a significant linear decrease as plane of elevation increased (all $p < 0.05$).

4. Discussion

This study provides the most direct evidence to date that changes in arm posture influence the dynamic mechanical properties of the shoulder. Robot-assisted biomechanical measures of joint stiffness and viscosity were found to change across the seven examined experimental postures. We hypothesized that increasing shoulder elevation would decrease stiffness and viscosity. In agreement with our hypothesis, shoulder stiffness decreased with increasing elevation angles. However, we rejected our hypothesis in regards to viscosity, as this decrease in shoulder stiffness happened without concurrent changes in viscosity. This lead to greater damping ratios at higher elevation angles. Stiffness, viscosity, and damping ratio were all found to significantly increase as the plane of elevation of the arm increased, bringing the arm from the side to in front of the chest. These results indicate that the shoulder is less stable, and allows more movement, at postures with greater elevation angles and smaller plane of elevation angles, and therefore may be more prone to injury in these positions.

Our findings agree with the limited studies that have measured shoulder dynamics. At rest, the shoulder was previously shown to have a stiffness of 30 Nm/rad when the arm was elevated 45 degrees in the scapular plane (Zhang et al., 2000) and 10 Nm/rad

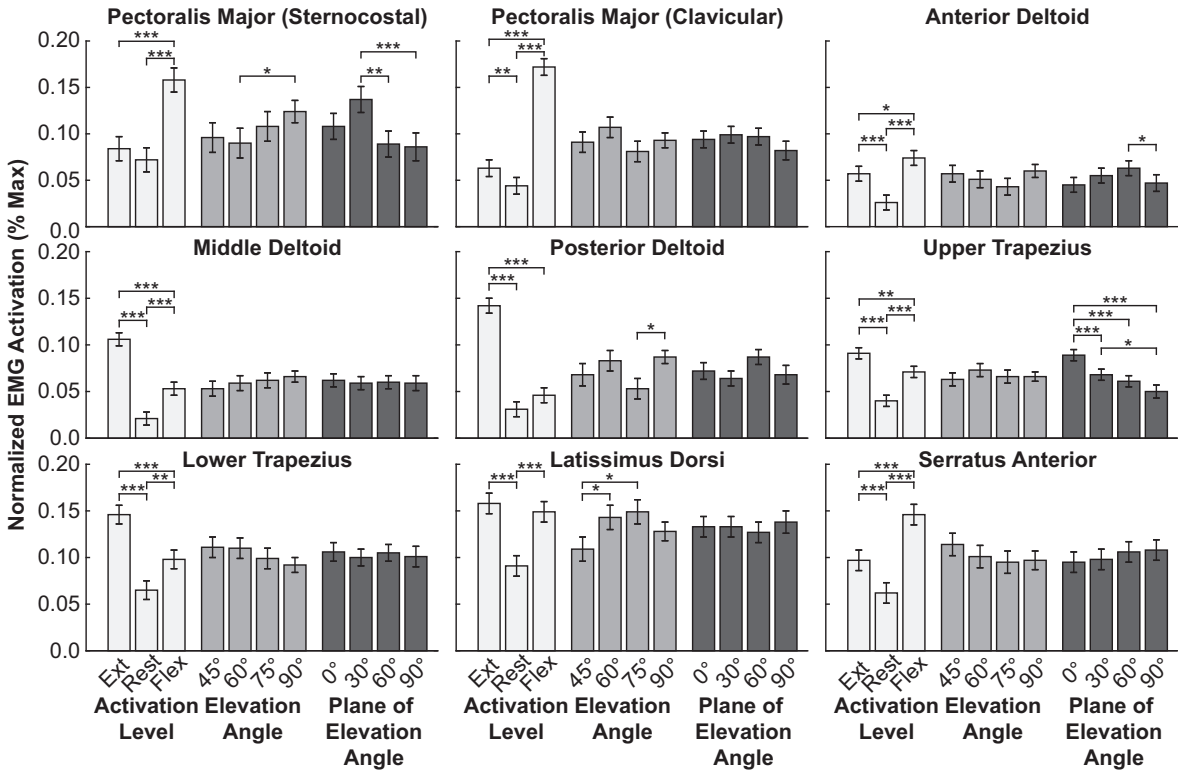

Fig. 5. Normalized muscle activity was recorded from nine shoulder muscles using surface EMG across all conditions. Bars represent mean percent of maximum muscle activation ± standard error (* = significance at $p < 0.05$, ** = significance at $p < 0.01$, and *** = significance at $p < 0.001$).

when the arm was elevated 90 degrees with the arm by the side (Davidson and Charles, 2017; Lipps et al., 2015). We found similar stiffness values of 31 Nm/rad with the arm elevated 45 degrees and 10 Nm/rad with the arm elevated 90 degrees. There are differences in the perturbation direction between these studies, but passive coupling between horizontal flexion and vertical abduction torques produces a similar stiffness response in both directions (Lipps et al., 2015). Joint viscosity was also similar at rest, with previous studies observing 1.6 Nm-sec/rad with the arm elevated 45 degrees in the scapular plane (Zhang et al., 2000) and our studying reporting 1.3 Nm-sec/rad. Finally, the average damping ratio reported here across all trials was 0.25 agrees with previous findings (Zhang et al., 2000).

Postural changes in stiffness and viscosity are likely driven by changes in gravitational forces at the joint, changes in joint morphology, and changes in muscle activation as the shoulder moves throughout its range of motion. Gravitational torques on the shoulder will be greatest when the arm is elevated 90 degrees, and consequently result in lower contributions of gravity to joint stiffness in this posture. As elevation angle increases, the humeral head translates inferiorly in the glenoid fossa (Bey et al., 2008) and as plane of elevation increases, the humeral head translates inferiorly (Graichen et al., 2000). Due to the translations of the humeral head throughout different arm postures, it is likely that the location and amount of contact between the glenoid fossa and the humeral head would cause changes in the stiffness and viscosity of the joint. The coordinated activation of muscles will influence the corresponding dynamic response of the shoulder to a postural disturbance. We observed certain muscles (PM CL, MD, LT and SA) produced constant levels of maximal activation across all postures, while other shoulder muscles had optimal positions for maximal activation that likely contribute to the postural changes in viscosity and/or stiffness. Changes in stiffness or viscosity as elevation angle or plane of elevation angle increased were likely caused by significantly greater activation for LD (peaking at 75°), and PM SC (peaking at 90°) at greater elevation angles, and significantly greater activation for AD (peaking at 60°) at greater plane of elevation angles.

Adjusting the damping ratio allows the central nervous system to control the stability of the joint, or the joint's resistance to movement. Zhang et al. (2000) found the damping ratio stays constant over different torque levels as a way for the central nervous system to make control tasks easier by adjusting stiffness and viscosity to hold the joint an underdamped system. These findings were based on examining the shoulder in a single posture while producing torques in vertical ab/adduction. Our results showed that horizontal flexion torques produced greater damping ratios than both horizontal extension torques and at rest. This suggests the central nervous system may need to adjust the dynamic properties of the shoulder more frequently to compensate for this changing damping ratio in the horizontal plane. Joint viscosity is important for managing the damping ratio. At rest, viscosity could increase when the elevation angle was decreased and/or when the plane of elevation was increased (Engin, 1984). Our findings partially agree with this study, as we found viscosity was not affected by elevation angle but did increase with greater elevation angles. We attribute these differences are largely due to our ability to acquire viscosity measurements both at rest and during muscular contraction, which is more physiologically relevant.

Measuring the dynamic properties of the shoulder provides important insight into one's ability to execute everyday tasks, as many such tasks place the upper extremity in postures that destabilize the shoulder (Hu et al., 2011; Rancourt and Hogan, 2001). The current study measured shoulder dynamics as participants produced a pushing or pulling force in the horizontal plane, which could increase the risk of shoulder injury in the workplace (Hoozemans et al., 2014). Decreased postural stability, or a lower resistance to movement, allows for greater range of motion at the joint. However, this decrease in stability can negatively impact quality of living as increased range of motion in response to a postural disturbance may place the joint in postures that are more prone to injury (Edwards et al., 2010). Our finding that stiffness decreases with greater elevation angles and smaller plane of elevation angles could indicate that the shoulder is more prone to injury in these postures, which is consistent with injury patterns observed in athletes who participate in repetitive overhead sports (Clarsen et al., 2015; Matsuura et al., 2016). A greater amount of stability may also negatively impact quality of living by limiting the amount of movement at the shoulder joint. For example, adhesive capsulitis produces decreased range of motion, increased pain, and reduced quality of life (Griggs et al., 2000).

The current study had certain limitations, including that our results only generalize to the seven examined postures. The current experimental paradigm did not allow for more postures to be examined given the time associated with carefully moving the motor, chair, and participant to the desired postures. Future work should consider using robot-assisted measures that can easily perturb and move the joint in 3D rather than the 1D rotary motor used here. We related our results to muscle activity in nine shoulder muscles using surface EMG but cannot conclude how change in rotator cuff activation influence our results as these muscles require intramuscular EMG. The shoulder is an indeterminate system with many possible muscle activation patterns to produce a desired joint torque. We attempted to control for this by limiting the production of off-axis torques by providing participants using visual feedback regarding the 3-D torques produced at the glenohumeral joint. Finally, while the arm is supported against gravity by the motor, there is stiffness component of gravity that differs as the arm is placed in different postures. The central nervous system needs to constantly compensate for this gravitational stiffness when stabilizing the shoulder during everyday activities.

In conclusion, changes in arm posture altered the static and dynamic properties of the shoulder. The relationship between the viscosity, stiffness and the damping ratio of the shoulder is one that the central nervous system must regulate in order to maintain stability and control of the shoulder as the inertia and muscle contributions change across different arm postures. While certain postures have greater stiffness, and thus greater stability, other arm postures which have lower stiffness, and thus lower stability, may be more prone to injury.

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Conflict of interest statement

The authors do not have any financial or personal relationships to disclose that could have inappropriately biased this work.

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