MODELING THE FUNCTIONAL ROLES OF SCAPULOHUMERAL MUSCLES

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ABSTRACT

A high degree of variability is commonly encountered in biomechanical investigations of the shoulder. Researchers have hypothesized that the variation between individuals explains why only certain workers are injured when performing the same tasks as other individuals. One source for the variability is inter-individual differences in shoulder musculoskeletal geometry. The purpose of this thesis was to use computational modeling to assess the functional roles of the scapulohumeral muscles, compare model-predicted data to the reviewed literature, and quantify the sensitivity of these functional roles to changes in muscle geometry. Muscle moment arms, lines of action, stability ratios, and forces were quantified throughout arm elevation in the scapular plane using a widely investigated upper extremity model – Delft Shoulder and Elbow Model. Monte Carlo simulations were performed to iteratively adjust muscle attachment locations in order to reflect potential inter-individual differences in muscle geometry. Model-predicted muscle moment arms agreed well qualitatively with the reviewed literature; however, several muscle lines of action were inconsistent between the model and previous data collected in cadavers available in the literature. Sensitivity of muscle functional roles to attachment changes was muscle-specific, and depended upon the elevation angle as well as outcome measure. Regressions were developed to identify which attachment locations at the clavicle, scapula, and humerus caused the greatest change in muscle functional roles. In general, muscle moment arms were most sensitive to changes of the muscle attachment closest to the joint centre (humeral attachment for rotator cuff muscles; scapular attachment for deltoids). Lines of action were most affected by perturbations in scapular attachment location. Overall, these findings indicate that inter-individual musculoskeletal geometry differences can substantially alter muscle functional roles, which are expected to require altered muscle activity and kinematic coordination patterns between people. These variations in musculoskeletal geometry may differentially affect risk of work-related shoulder musculoskeletal disorders among individuals.

Keywords: Glenohumeral joint, stability, shoulder, probabilistic modeling, variability

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THESIS FORMAT AND ORGANIZATION

This thesis contains the MSc work completed by Daanish M. Mulla, prepared in "sandwich" format as outlined in the McMaster School of Graduate Studies' Guide for the Preparation of Thesis.

This thesis begins with a general introduction overviewing the development of shoulder musculoskeletal disorders in the workplace, with a focus on challenges encountered by recent research conducted at the McMaster Occupational Biomechanics Laboratory and the potential benefits of computational modeling to address some of these limitations (Chapter 1). Next, an in-depth review of the literature examining shoulder anatomy and function, biomechanical models, and probabilistic modeling is presented in Chapter 2.

Chapters 3 and 4 used a probabilistic modeling approach to evaluate the potential effects of inter-individual differences in musculoskeletal geometry on the functional roles of the scapulohumeral muscles. Chapter 3 examines the distribution of muscle moment arms and lines of action with changes in muscle attachment locations, while qualitatively comparing model-predictions to reviewed literature to verify the anatomical fidelity of the model, and will be submitted to the Journal of Biomechanics. Chapter 4 further evaluates the muscle lines of actions into glenohumeral stability ratios and discusses the clinical relevance of inter-individual differences in dynamic stability, which will be submitted to Clinical Biomechanics.

This thesis ends with a final summary chapter (Chapter 5) discussing the overall findings and future directions.

CONTRIBUTIONS TO PAPERS WITH MULTIPLE AUTHORS

Chapter 3

Mulla, D. M., Hodder, J. N., Maly, M. R., Lyons, J. L., & Keir, P. J. Modeling the effects of musculoskeletal geometry on scapulohumeral muscle moment arms and lines of action. Prepared for submission to the Journal of Biomechanics.

Contributions

This study was conceived by Daanish M. Mulla and Dr. Peter J. Keir. Method and model development, and data analysis were conducted by Daanish Mulla, with input from Dr. Keir. Dr. Joanne N. Hodder, Dr. Monica R. Maly, and Dr. James L. Lyons contributed expertise. Interpretation of results was completed by Daanish Mulla, with input from Dr. Hodder, Dr. Maly, Dr. Lyons, and Dr. Keir. The manuscript was written by Daanish Mulla, with input from Dr. Keir. All co-authors contributed to this manuscript.

Chapter 4

Mulla, D. M., Hodder, J. N., Maly, M. R., Lyons, J. L., & Keir, P. J. Glenohumeral stabilizing roles of the scapulohumeral muscles: Implications of muscle geometry. Prepared for submission to Clinical Biomechanics.

Contributions

This study was conceived by Daanish M. Mulla and Dr. Peter J. Keir. Method and model development, and data analysis were conducted by Daanish Mulla, with input from Dr. Keir. Dr. Joanne N. Hodder, Dr. Monica R. Maly, and Dr. James L. Lyons contributed expertise. Interpretation of results was completed by Daanish Mulla, with input from Dr. Hodder, Dr. Maly, Dr. Lyons, and Dr. Keir. The manuscript was written by Daanish Mulla, with input from Dr. Keir. All co-authors contributed to this manuscript.

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LIST OF ABBREVIATIONS

AC……………………………..……………………………..Acromioclavicular Joint

$C_x/C_y/C_z$………………………………………..Clavicular Attachment Location (XYZ)

DSEM……………………………..………………..Delft Shoulder and Elbow Model

EMG………………………………...………………………..Electromyography

GH……………………………………………………..…………..Glenohumeral Joint

$H_x/H_y/H_z$……………..…………………………..Humeral Attachment Location (XYZ)

ISB……………………………..………………..International Society of Biomechanics

$LOA_{A\text{-}P}$……………………………………………………..Anterior/Posterior Line of Action

$LOA_{S\text{-}I}$…….………………………………..…………..Superior/Inferior Line of Action

MA…………………………………………………..………………..Moment Arm

MSDs…………………………..…………………………..Musculoskeletal Disorders

ROM..……………………………………………………………..Range of Motion

SAI……………………………………….………………..Subacromial Impingement

SAS………………………………………………………………..Subacromial Space

SC…………………………………………..………………..Sternoclavicular Joint

ST……………………………………..…………..Scapulothoracic Joint/Articulation

$ST_{A\text{-}P}$……………………………………………………..Anterior/Posterior Stability Ratio

$ST_{S\text{-}I}$……………………………...…………………..Superior/Inferior Stability Ratio

$S_x/S_y/S_z$…………………………………………..Scapular Attachment Location (XYZ)

LIST OF APPENDICES

CHAPTER ONE: INTRODUCTION

Musculoskeletal disorders (MSDs) affecting the upper extremity are common in the workplace, posing significant health and financial burdens to workers and the health care system. In 2016, the upper extremity was the second most common source for worker injuries (behind back injuries), accounting for almost one quarter of all injury claims in Ontario (WSIB Ontario, 2017). In contrast to back injuries, incidence rates of work-related upper extremity injuries have been consistent over the past 5 years, with lost time injuries costing workplaces over $250 million (WSIB Ontario, 2017). As the shoulder is vital to everyday tasks, a clear understanding of shoulder function is needed for identifying factors that may lead to impaired function and MSDs, to improve prevention strategies.

One risk factor for shoulder MSDs is muscle fatigue accumulated from repetitive work (Sommerich et al., 1993). Recently, we investigated upper extremity muscular and movement adaptations to fatigue and simulated assembly work. The primary finding from a series of investigations was that there is a high degree of inter-individual variability in upper extremity muscular and kinematic adaptation strategies to compensate for fatigue while maintaining task performance (Ebata, 2012; Tse et al., 2016; McDonald et al., 2016, 2018a,b; McDonald, 2017; Mulla et al., 2018). Further complicating matters is that fatigue-related compensatory strategies may be inconsistent within-individuals across time (McDonald, 2017; Mulla et al., 2018). The variability between- and within-individuals is hypothesized to be largely a function of the number muscles and degrees of freedom at the shoulder, enabling individuals to coordinate countless muscular and movement strategies to perform any task. While we might consider that this redundancy within the musculoskeletal system is advantageous to maintain performance during fatiguing activities, changes in upper extremity muscle and movement coordination can cause differences in muscle loading and impingement of the rotator cuff (Michener et al., 2003; Ebaugh et al., 2006ab; Chopp-Hurley et al., 2016a). Differences in muscle loading may be positively (load-sharing) or negatively adaptive (load-localization). In addition, kinematic changes can alter the width of the subacromial space (SAS), with narrowing of this space implicated in a frequently occurring shoulder injury termed subacromial impingement (SAI) (Michener et al., 2003). As such, we presume that the large between-individual variability in muscular and movement strategies (in addition to anthropometric differences, such as bone/muscle morphology and geometry) contributes to the highly differential risk for workplace upper extremity MSDs among workers performing similar tasks (Kilbom & Persson, 1987). Accordingly, differentiating individual strategies may identify fatigued workers at increased risk for developing upper extremity injuries due to potential muscle overload and/or impingement. As a result, greater knowledge of the consequences of fatigue-induced alterations in muscle activity and kinematics can be used to help workers that may be predisposed to increased risk of developing shoulder MSDs.

Cadaveric studies have provided significant insight into the functional roles of shoulder muscles by quantifying moment arms and lines of action; however, there are a number of inconsistencies across experiments. In addition, due to the nature of cadaveric work, only a limited number of postures among small sample sizes that do not necessarily represent broader populations (e.g. older specimens, atrophied muscles) can be evaluated, with certain groups of muscles receiving limited attention (e.g.. axioscapular muscles such as the trapezius and serratus anterior). Alternatively, musculoskeletal models are powerful tools that can build upon experimental studies and advance current understanding of muscle function. These models are simplified representations of the muscular and skeletal systems of the human body, often used to estimate joint loading due to muscle forces (van der Helm, 1994; Erdemir et al., 2007). Although computer modeling is an effective tool, the question of each muscle's functional role across the shoulder complex range of motion (ROM) largely persists. Determining these capacities would allow us to better distinguish how muscular strategies can affect risk of SAI by assessing how muscles can promote changes in glenohumeral joint loading as well as alter SAS due to kinematic changes.

Although biomechanical computer simulations offer the promise of broadening our knowledge of shoulder muscle function across a range of upper extremity postures, there is significant potential for findings from modeling approaches to differ from experimental studies due to a number of assumptions required, such as when representing musculoskeletal geometry mathematically. As a result, current models need to be assessed against existing experimental data. More than that, what is truly needed is a deeper evaluation of the mechanical function of muscles. It is commonplace to judge how muscles may be contributing to a given action based on conventional "textbook" understanding. For instance, among our laboratory studies investigating fatigue-induced kinematic and muscular strategies, if a muscle displays increased electromyography

(EMG) activity, it is presumed to promote a specific kinematic strategy that aids the individual in maintaining task performance. However, whether these apparent muscle functions, based on common knowledge, especially across the large range of shoulder motion, are reflected in current biomechanical models have yet to be thoroughly evaluated.

Furthermore, recent trends strongly suggest an individualized approached to modeling. The inconsistencies across experimental studies and reports of high degree of variability suggest significant differences in the musculoskeletal geometry across individuals, which can meaningfully alter muscle function. This likely contributes to the large variability in fatigue-induced muscular and kinematic strategies observed previously in our laboratory investigations. To gauge the need for subject-specific models, the sensitivity of muscle function (moment arms, lines of action, model-predicted muscle forces) to variability in musculoskeletal geometry should be assessed. In the process, we can determine the muscle-specific inputs that need to be individualized or alternatively represented by a probabilistic modeling approach, to incorporate individuality and variability in model predictions that is more reflective of experimental work. Ultimately, using a probabilistic modeling approach can greatly improve our current understanding of shoulder muscle function, allowing us the ability to associate individual muscular strategies to kinematic changes, and consequently differentiate muscular and kinematic strategies that may be implicated in upper extremity MSDs.

Overall, there is a great need to better understand the capacity of shoulder muscles to move and stabilize the upper extremity with changes in posture and musculoskeletal geometry. Thus, the global aim of this thesis is to assess and use computer modeling to gain a comprehensive understanding of the roles of shoulder muscles across different postures and quantify the effect of individual differences in musculoskeletal geometry to muscle function among healthy men and women. Resolving these research questions will improve our ability to determine the potential risk for shoulder MSDs caused by fatigue-induced changes in muscular and movement strategies.

CHAPTER TWO: REVIEW OF LITERATURE

2.1. Skeletal Anatomy and Motion

The shoulder complex consists of three bones – humerus, scapula, and clavicle – that articulate with each other and the thorax across four joints (Inman et al., 1944). These include the sternoclavicular joint (SC), acromioclavicular joint (AC), scapulothoracic (ST), and glenohumeral (GH) joints. The SC joint is the articulation between the sternum and the medial aspect of the clavicle. The SC joint is assumed to act as a "ball and socket" joint, exhibiting three rotational degrees of freedom. Motion at the SC joint describes clavicular rotations relative to the trunk. In accordance with the International Society of Biomechanics standards (Wu et al., 2005), the three rotations at the SC joint will referred to as clavicular protraction/retraction, clavicular depression/elevation, and clavicular axial rotation (Figure 2.1). The AC joint is the articulation between the acromion of the scapula and the lateral aspect of the clavicle. Motion at the AC joint describes scapular rotations relative to the clavicle. In accordance with the International Society of Biomechanics (ISB) standards (Wu et al., 2005), the three rotations at the AC joint will be referred to as AC protraction/retraction (i.e. internal/external rotation), AC medial/lateral rotation (i.e. inferior/superior or downward/upward rotation), and AC posterior/anterior tilt (Figure 2.1). The ST joint is the articulation between the scapula and the thorax. The ST joint is not considered a true synovial joint and does not have a fixed axis of rotation, but rather describes the relative motion between the scapula and thorax. Motion at the ST joint is afforded through the synchronous movement contributions from the SC and AC joints (Ludewig et al., 1996; Teece et al., 2008). In accordance with the ISB standards (Wu et al., 2005) and consistent with AC motion, the rotations at the ST joint will be referred to as ST protraction/retraction (i.e. internal/external rotation), ST medial/lateral rotation (i.e. inferior/superior or downward/upward rotation), and ST posterior/anterior tilt (Figure 2.1). The scapula also exhibits translation movements relative to the thorax; however, these are not widely reported in the biomechanics literature. The GH joint is the articulation between the glenoid fossa of the scapula and the humerus. The rotations at the GH joint will be referred to as GH plane of elevation (i.e. horizontal adduction/abduction), GH depression/elevation, and GH axial (i.e. internal/external) rotation (Figure 2.1). The GH joint does not strictly adhere to a "ball and socket" joint as the humeral head is observed to slightly translate on the glenoid during glenohumeral motion, primarily along the superior/inferior and anterior/posterior directions (Poppen & Walker, 1976; Wuelker et al., 1994; Graichen et al., 2000).

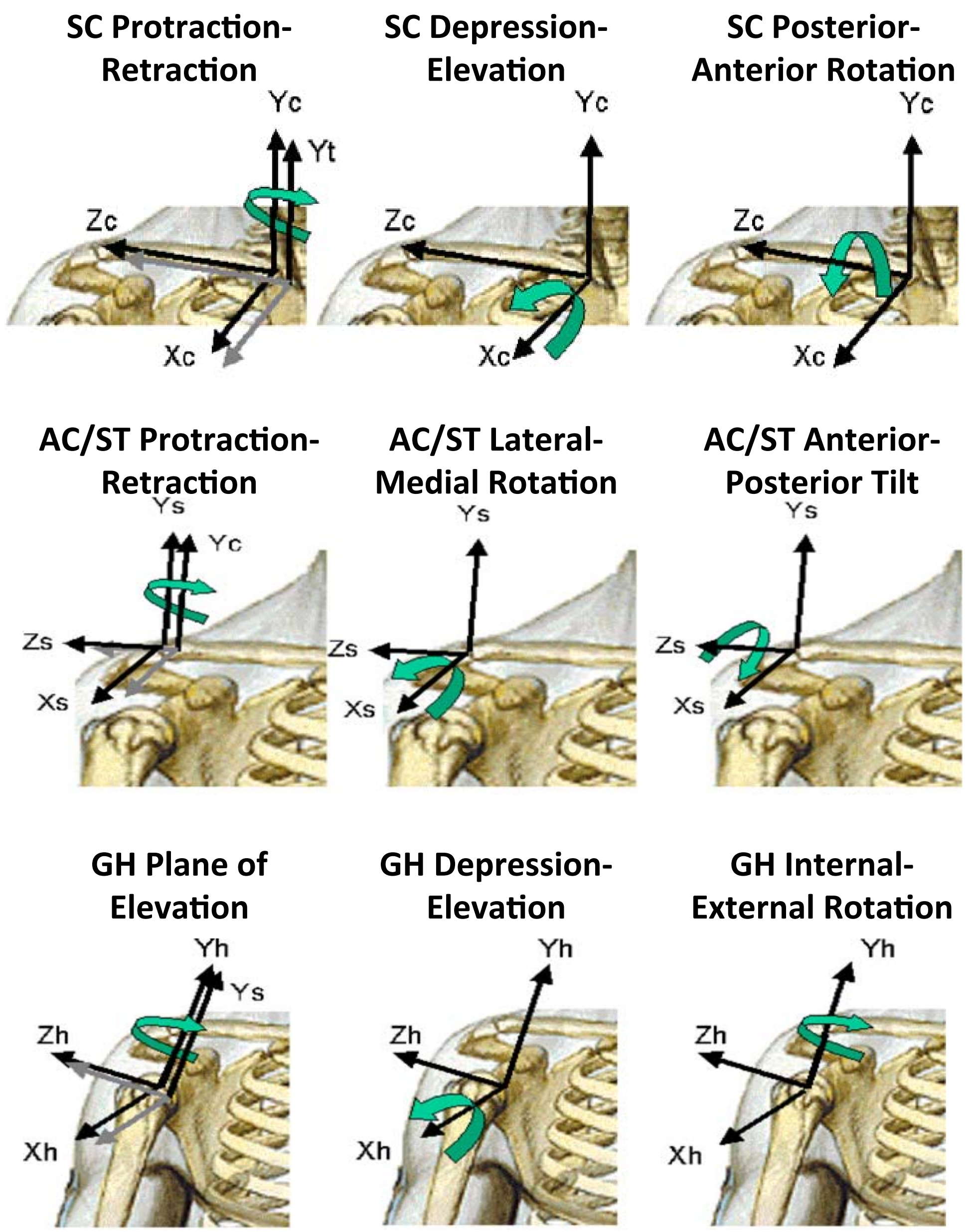


Figure 2.1: Coordinate systems and motion descriptions for each rotational degree of freedom at the sternoclavicular (SC), acromioclavicular (AC), scapulothoracic (ST), and glenohumeral (GH) joints (from Wu et al., 2005).

The shoulder complex is a highly mobile area, allowing placement of the hand across a large spatial volume. The shoulder exhibits a large range of motion (ROM) across numerous degrees of freedom, permitting a countless number of ways to perform unrestricted upper extremity movement (Veeger & van der Helm, 2007). In an effort to standardize motion analysis, investigations examining shoulder movements commonly study simple, planar arm elevations, which can be referred to as humerothoracic (HT) elevation (i.e. humeral motion relative to the thorax). Early investigations into shoulder motion found that HT elevation is primarily produced through the synchronous motion between the scapula and humerus, termed the scapulohumeral rhythm (Inman et al., 1944). The scapulohumeral rhythm can be expressed as the ratio of movement contribution from the humerus (i.e. GH elevation) and scapula (i.e. ST lateral rotation) in the plane of the elevation. Inman et al. (1944) first documented a 2:1 ratio of humerus to scapular motion during arm elevation, such that for every 2° of GH elevation, there was 1° of ST lateral rotation. In that investigation, motion was presumably measured in a single individual, with subsequent studies observing largely varying ratios (Freedman & Munro, 1966; Doody et al., 1970; Poppen & Walker, 1976, 1978; Bagg & Forrest, 1988; Bourne et al., 2007; Crosbie et al., 2008; Braman et al., 2009; Ludewig et al., 2009). Although some of the variations may be due to methodological differences across studies, a relatively high degree of variability in scapulohumeral rhythm is reported between healthy subjects within the same investigations. More importantly, although ST lateral rotation and GH elevation exhibit a large change in ROM during arm elevation, there is a non-linear relationship between the two rotations with considerable motion at the other available degrees of freedom to permit arm elevation (Bagg & Forrest, 1988; McQuade & Smidt, 1998; Bourne et al., 2007; Crosbie et al., 2008). As such, a single scapulohumeral rhythm ratio is inadequate in providing a complete picture of upper extremity movement.

Three-dimensional motion analyses of the upper extremity have revealed that overall movement at the shoulder complex is coordinated through the simultaneous motion contributions at each of the four joints. In general, there is a pattern towards increased ST lateral rotation, ST posterior tilt, GH external rotation, and GH elevation contributing to arm elevation (Tables 2.3 and 2.4). ST protraction/retraction and GH superior/inferior translation have greater inconsistencies across individuals, with discrepancies in directional patterns during arm elevation (Tables 2.3 and 2.4). ST motion is a result of coupling motion at the SC and AC joints (Ludewig et al., 1996; Teece et al., 2008), with SC retraction/elevation/posterior axial rotation and AC protraction/lateral rotation/posterior tilt all contributing towards arm elevation (Tables 2.1 and 2.2). Among all the degrees of freedom that are observed to have consistent directional trends with arm elevation, the relative magnitude of these rotations varies highly across studies (Tables 2.1, 2.2, 2.3, 2.4). It should be acknowledged that scapular kinematics are difficult to measure due to the shape of the scapula and motion occurring primarily underneath the skin surface, potentially resulting in considerable measurement error. Furthermore, differences in study methods make it challenging to compare results across investigations. Important factors include heterogeneity of study participants, movement conditions, arm tested (Crosbie et al., 2008; Matsuki et al., 2011), motion capture devices (Karduna et al., 2001; van Andel et al., 2009), rotation sequences during analysis (Phadke et al., 2011), and coordinate systems and reference frames (Meskers et al., 1998; Ludewig et al., 2010). Nonetheless, substantial between-participant differences in the magnitude and directional trends of upper extremity kinematics is noted within many of the same studies (Freedman & Munro, 1966; Ludewig et al., 1996; Graichen et al., 2000; Price et al., 2000; McClure et al., 2001; Borstad & Ludewig, 2002; Ludewig et al., 2004; Sahara et al., 2007; Ludewig et al., 2009; Matsuki et al., 2012; Picco et al., 2017). For example, the 95% confidence intervals (95% CI) of ST protraction (27-55°), ST upward rotation (-2-11°), and ST anterior tilting (0-28°) with the arm by the side are relatively wide, as measured using bone pins, the "gold standard" for scapular motion capture (Ludewig et al., 2009; Note: 95% CI were calculated from the standard errors and sample size given in the study). Although movements in the research setting are often restricted and may not be representative of everyday actions, large variation in upper extremity motion is also observed during tasks simulating occupational activities (Ebata, 2012; McDonald et al., 2016; Tse et al., 2016; McDonald, 2017).

Table 2.1: Summary of in vivo studies measuring sternoclavicular (SC) motion during dynamic, planar humeral (HT) elevations among healthy individuals. Means and measures of variability (± standard deviation, (range), [95% CI]) are provided where applicable. Positive values indicate SC protraction, depression, and posterior axial rotation (Note: values from studies have been assigned the appropriate signs to meet this convention). Initial represents the mean joint angle at beginning of HT elevation. Range of motion (ROM) represents the mean change in joint angle from initial value at end of HT elevation. *Italicized values* are estimates from graphs. Values denoted with (*) are estimates from equations. Papers denoted with (†) include standard deviations manually calculated from standard error values. Abbreviations include: study sample size (n), males (M), females (F), and age in years (y).

Reference	Sample	Measurement	Plane of Movement	HT Elevation	Protraction/Retraction		Depression/Elevation		Posterior/Anterior Axial Rotation	
					Initial	ROM	Initial	ROM	Initial	ROM
Inman et al. (1944)	n=1	2D X-Ray Bone Pins	0° Frontal 90° Sagittal	0-170° 0-170°	-	-	*-90°* *-90°*	*-30°* *-30°*	*-90°* *-90°*	*-50°* *-50°*
Meskers et al. (1998)	n=15 (7F) 24.3±8.4y	3D Sensors (Static)	0° Frontal 90° Sagittal	0-150° 0-150°	*-20°* *-20°*	*-30°* *-28°*	*-3°* *-7°*	*-12°* *-7°*	*0°* *0°*	*65°* *60°*
McClure et al. (2001)	n=8 (3F) 32.6y (27-37)	3D Sensors; Bone Pins (Dynamic)	40° Scapular 90° Sagittal	11-147° 16-153°	-	-21° -20°	-	-10° -9°	-	-
Ludewig et al. (2004)	n=30 (14F) 26.9±5.2y	3D Sensors (Dynamic)	0° Frontal 40° Scapular 90° Sagittal	0-110° 0-110° 0-110°	-18.2±5.8°	-10.5° *-6.8°* 3.3°	-1.6±3.3°	-10.6° -9.4° -13.4°	0.5±2.5°	14.1° 17.7° 30.8°
Dayanidhi et al. (2005)	n=15 (7F) 28.8±4.3y	3D Sensors (Dynamic)	40° Scapular	25-125°	*-26°*	*-8°*	*-8°*	*-3°*	-	-
Ebaugh et al. (2005)	n=20 (10F) 22.5y (18-30)	3D Sensors (Dynamic)	40° Scapular	20-120°	-21.3±4.9°	-13°	-5.9±5.8°	-12.8°	-	-
Sahara et al. (2007)	n=7 (0F) 23.6y (19-30)	3D open MRI (Static)	0° Frontal	0-180°	-28.6±7.1°	-30.6°	-6.3±5.6°	-7.3°	0°	33.2°

Ludewig et al. (2009)[†]	n=12 (5F) 29.3±6.8y	3D Sensors; Bone Pins (Dynamic)	0° Frontal 40° Scapular 90° Sagittal	0-120° 0-120° 0-120°	-19.2±7°	-24.8° -17.2° -12.5°	-5.9±3°	-13.6° -11.0° -8.6°	0.1±0°	24.7° 24.2° 24.9°
Nagai et al. (2013)	n=12 (0F) 22.8±3.1y	3D Sensors (Dynamic)	90° Sagittal	30-120°	*-23°*	*-3°*	*-12°*	*2°*	-	-

Table 2.2: Summary of in vivo studies measuring acromioclavicular (AC) motion during dynamic, planar humeral (HT) elevations among healthy individuals. Means and measures of variability (± standard deviation, (range), [95% CI]) are provided where applicable. Positive values indicate AC internal rotation, downward rotation, and posterior tilt (Note: values from studies have been assigned the appropriate signs to meet this convention). Initial represents the mean joint angle at beginning of HT elevation. Range of motion (ROM) represents the mean change in joint angle from initial value at end of HT elevation. *Italicized values* are estimates from graphs. Values denoted with (*) are estimates from equations. Papers denoted with (†) include standard deviations manually calculated from standard error values. Abbreviations include: study sample size (n), males (M), females (F), and age in years (y).

Reference	Sample	Measurement	Plane of Movement	HT Elevation	Protraction/Retraction		Medial/Lateral Rotation		Posterior/Anterior Tilt	
					Initial	ROM	Initial	ROM	Initial	ROM
Inman et al. (1944)	n=1	2D X-Ray Bone Pins	0° Frontal 90° Sagittal	0-170° 0-170°	-	-	*-90°* *-90°*	*-25°* *-25°*	-	-
Meskers et al. (1998)	n=15 (7F) 24.3±8.4y	3D Sensors (Static)	0° Frontal 90° Sagittal	0-150° 0-150°	*60°* *60°*	*10°* *10°*	*-6°* *-7°*	*2°* *-1°*	*-7°* *-7°*	*11°* *12°*
McClure et al. (2001)	n=8 (3F) 32.6y (27-37)	3D Sensors/Bone Pins (Dynamic)	40° Scapular 90° Sagittal	11-147° 16-153°	-	24° 26°	-	-50° -46°	-	30° 31°
Sahara et al. (2007)	n=7 (0F) 23.6y (19-30)	3D open MRI (Static)	0° Frontal	0-180°	62.8±8.1°	15.6°	11.3±5.0°	-21.5°	-15.8±11.2°	22.2°
Teece et al. (2008)	n=30 (14F) 25.2±3.5y	3D Sensors (Dynamic)	Scapular	0-90°	-	4.3°	-	-14.6°	-	6.7°
Ludewig et al. (2009)[†]	n=12 (5F) 29.3±6.8y	3D Sensors Bone Pins (Dynamic)	0° Frontal 40° Scapular 90° Sagittal	0-120° 0-120° 0-120°	60.0±7°	2.2° 3.5° 6.2°	-2.5±3°	-14.6° -13.0° -12.9°	-8.4±7°	23.0° 19.0° 20.2°

Table 2.3: Summary of in vivo studies measuring scapulothoracic (ST) motion during dynamic, planar humeral (HT) elevations among healthy individuals. Means and measures of variability (± standard deviation, (range), [95% CI]) are provided where applicable. Positive values indicate ST internal rotation, downward rotation, and posterior tilt (Note: values from studies have been assigned the appropriate signs to meet this convention). Initial represents the mean joint angle at beginning of HT elevation. Range of motion (ROM) represents the mean change in joint angle from initial value at end of HT elevation. *Italicized values* are estimates from graphs. Values denoted with (*) are estimates from equations. Papers denoted with (†) include standard deviations manually calculated from standard error values. Abbreviations include: study sample size (n), males (M), females (F), and age in years (y).

Reference	Sample	Measurement	Plane of Movement	HT Elevation	Protraction/ Retraction		Medial/Lateral Rotation		Posterior/ Anterior Tilt	
					Initial	ROM	Initial	ROM	Initial	ROM
Inman et al. (1944)	n=1	2D X-Ray Bone Pins	0° Frontal 90° Sagittal	0-170° 0-170°	-	-	-	-45°* -35°*	-	-
Freedman & Munro (1966)	n=61 (0F) 17-24y	2D X-Ray (Static)	30° Scapular	0-135°	-	-	-5.3±6.8°	-54.7°	-	-
Poppen & Walker (1976)	n=12 22-63y	2D X-Ray (Static)	30° Scapular	2.5-150°	-	-	-4.7° (-11 - (+)10)	-54°*	-	37°*
Bagg & Forrest (1988)	n=20 (0F) “Young”	2D Motion Analysis (Static)	30° Scapular	0-168.1°	-	-	-	-63.8°	-	-
Johnson et al. (1993)	n=15 (0F) 18-60y	3D Sensor Locator (Static)	0° Frontal (1) 0° Frontal (2)	0-120° 0-120°	1.62° -1.67°	-8.94° -3.29°	-0.40° -2.48°	-29.36° -35.41°	-0.34° 1.45°	7.24° 11.18°
Ludewig et al. (1996)	n=25 (14F) 25.9±5.2y	3D Digitizer (Static)	30° Scapular	0-140°	33±9°	-12°	-2±6°	-34°	-8±4°	15°
Meskers et al. (1998)	n=15 (7F) 24.3±8.4y	3D Sensors (Static)	0° Frontal 90° Sagittal	0-150° 0-150°	*27°* *30°*	*3°* *0°*	*-3°* *0°*	*-60°* *-60°*	*-11°* *-12°*	*13°* *25°*
Price et al. (2000)	n=10 (1F) 50y (17-78)	3D Sensors (Static)	0° Frontal	10-50°	-	*-2.5°*	-	*-16°*	-	*2.5°*
Borstad & Ludewig (2002)	n=26 (0F) 39.9±13.3y	3D Sensors (Dynamic)	40° Scapular	40-120°	*40°*	*7°*	-16.7±2°	-24°	*-10°*	*2°*

Dayanidhi et al. (2005)	n=15 (7F) 28.8±4.3y	3D Sensors (Dynamic)	40° Scapular	25-125°	*42°*	*7°*	*-17°*	*-30°*	*-0.5°*	*3.5°*
Ebaugh et al. (2005)	n=20 (10F) 22.5y (18-30)	3D Sensors (Dynamic)	40° Scapular	20-120°	44.5±8.2°	2.3°	-29±5.4°	-38°	0.38±7.8°	0.55°
Crosbie et al. (2008)	n=32 (32F) 38.2y (19-74)	3D Sensors (Dynamic)	0° Frontal 30° Scapular 90° Sagittal	0-ROM 0-ROM 0-ROM	-	-29° [25-33] -29 [24-33] -30° [26-35]	-	-23° [20-26] -20° [17-23] -20° [17-22]	-	-3° [1-6] -2° [1-3] -3° [0-5]
Ludewig et al. (2009)[†]	n=12 (5F) 29.3±6.8y	3D Sensors Bone Pins (Dynamic)	0° Frontal 40° Scapular 90° Sagittal	0-120° 0-120° 0-120°	41.1±7°	-10.5° -3.9° 2.3°	-5.4±3°	-41.0° -38.1° -37.9°	-13.5±7°	17.3° 16.2° 18.9°
van Andel et al. (2009)	n=13 (7F) 22-33y	3D Motion Analysis (Static)	0° Frontal 90° Sagittal	0-120° 0-120°	23.4°	-1° 18°	-11.1°	-27.0° 26.5°	-3.5°	7.7° 3.4°
Matsuki et al. (2011)	n=12 (0F) 32y (27-36)	3D Fluoroscopy (Dynamic)	Scapular	0-135°	*0°*	*-5°*	*-2°*	*-36°*	*0°*	*20°*
Nagai et al. (2013)	n=12 (0F) 22.8±3.1y	3D Sensors (Dynamic)	90° Sagittal	30-120°	*30°*	*3°*	*-8°*	*-37°*	*-5°*	*9°*
Habechain et al. (2014)	n=26 (14F) 35.3±11.7y	3D Sensors (Dynamic)	45° Scapular	30-120°	*32°*	*4°*	*-4°*	*-35°*	*-4°*	*10°*
Schwartz et al. (2016)	n=22 (11F) M: 22.5±2.5y F: 22.2±1.8y	3D Motion Analysis (Dynamic)	0° Frontal 90° Sagittal	0-120° 0-120°	*25°* *35°*	*-2.5°* *5°*	*0°* *0°*	*-30°* *-30°*	*-5°* *-7.5°*	*8°* *4°*
Picco et al. (2017)	n=29 (14F) M: 23.4±1.5y F: 22.8±3.0y	3D Motion Analysis (Dynamic)	0° Frontal 30° Scapular 40° Scapular 60° 90° Sagittal 120°	0-120° 0-120° 0-120° 0-120° 0-120° 0-120°	*25°* *30°* *30°* *35°* *35°* *40°*	*0°* *2°* *2°* *4°* *5°* *15°*	*5°* *5°* *5°* *5°* *5°* *0°*	*-35°* *-35°* *-35°* *-35°* *-35°* *-30°*	*-5°* *-5°* *-5°* *-5°* *-5°* *-5°*	*15°* *15°* *15°* *12°* *12°* *15°*

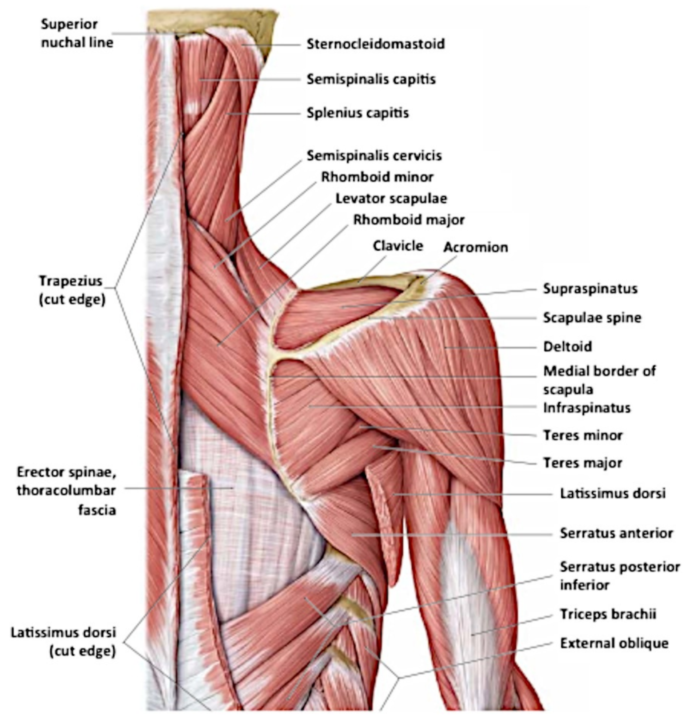
Table 2.4: Summary of in vivo studies measuring glenohumeral (GH) motion during dynamic, planar humeral (HT) elevations among healthy individuals. Means and measures of variability (±standard deviation, (range), [95% CI]) are provided where applicable. Positive values indicate GH elevation, internal rotation, superior and anterior translation (Note: values from studies have been assigned the appropriate signs to meet this convention). Initial represents the mean joint angle at beginning of HT elevation. Range of motion (ROM) represents the mean change in joint angle from initial value at end of HT elevation. *Italicized values* are estimates from graphs. Values denoted with (*) are estimates from equations. Papers denoted with (†) include standard deviations manually calculated from standard error values. Abbreviations include: study sample size (n), males (M), females (F), and age in years (y).

Reference	Sample	Measurement	Plane of Movement	HT Elevation	Elevation/Depression		Internal/External Rotation	
					Initial	ROM	Initial	ROM
Inman et al. (1944)	n=1	2D X-Ray Bone Pins	0° Frontal	0-170°	-	-	-	-
Freedman & Munro (1966)	n=61 (0F) 17-24y	2D X-Ray (Static)	30° Scapular	0-135°	4.46±6.95°	74.06°	-	-
Doody et al. (1970)		2D Motion Analysis (Static)	Scapular	5-176°	-	112.52°	-	-
Poppen & Walker (1976)	n=12 22-63y	2D X-Ray (Static)	30° Scapular	2.5-150°	-	95.1°*	-	-
Bagg & Forrest (1988)	n=20 (0F) "Young"	2D Motion Analysis (Static)	30° Scapular	0-168.1°	-	104.3°	-	-
Meskers et al. (1998)	n=15 (7F) 24.3±8.4y	3D Sensors (Static)	0° Frontal 90° Sagittal	0-150° 0-150°	*-15°* *-20°*	*100°* *100°*	*10°* *30°*	*-70°* *-80°*
Crosbie et al. (2008)	n=32 (32F) 38.2y (19-74)	3D Sensors (Dynamic)	0° Frontal 30° Scapular 90° Sagittal	0-ROM 0-ROM 0-ROM	-	100° [94-106] 105° [98-111] 147° [141-154]	-	-
Ludewig et al. (2009)†	n=12 (5F) 29.3±6.8y	3D Sensors/Bone Pins (Dynamic)	0° Frontal 40° Scapular 90° Sagittal	0-120° 0-120° 0-120°	0.8±3	73.5° 76.0° 76.9°	-14.1±14°	-43.4° -47.9° -49.7°

2.2. Muscle Anatomy and Function

2.2.1. Roles of Shoulder Muscles – Movement and Stability

Movement capabilities across the numerous degrees of freedom at the shoulder complex are enabled by a large group of muscles with overlapping actions (Figure 2.2). The combination of muscles allows individuals to exert force at the hand in any direction over an endless number of postures. The increased mobility at the shoulder however, comes at the expense of intrinsic stability (Veeger & van der Helm, 2007; Dickerson et al., 2011). The shoulder has limited passive constraints across the majority of its mid ROM, requiring muscle forces to dynamically stabilize the shoulder in addition to providing movement (An, 2002; Veeger & van der Helm, 2007). Shoulder stability is typically viewed as the balance of the humeral head on the glenoid fossa of the scapula, with instability resulting in large translations of the humeral head that could cause dysfunction, joint subluxation/displacement, and/or injury. Accordingly, joint loading at the shoulder requires a delicate balance of coordination across muscles to simultaneously promote movement and maintain integrity of the complex (An, 2002).



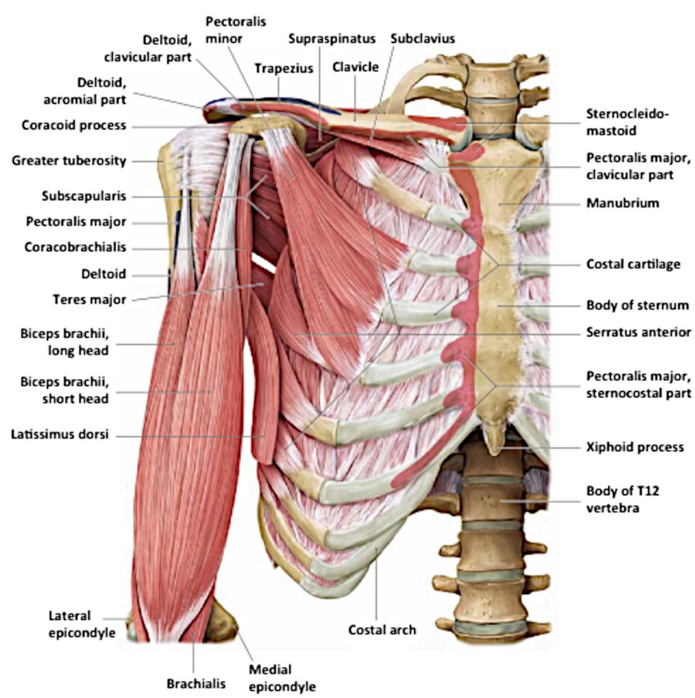


Figure 2.2: Shoulder muscles depicted in the posterior (left) and anterior (right) (from Schuenke et al., 2010).

To correspond with the demands at the shoulder complex, the mechanical function of shoulder muscles can be grouped into two main roles: stabilization and rotational movement (An, 2002). These roles can be quantified using two musculoskeletal geometric parameters: (i) lines of action and (ii) moment arms. Muscle forces are transmitted on a bone along the muscle pathway or line of action. As such, the direction of segment translation is determined by the muscle's line of action, which describes the capacity of a muscle to stabilize or de-stabilize a joint. For example, at the GH joint, muscle lines of action that are directed superiorly/inferiorly and anteriorly/posteriorly will result in a shear force that will de-stabilize and translate the humeral head away from the glenoid fossa in the direction of the force (Lee et al., 2000; An, 2002; Ackland & Pandy, 2009). In contrast, lines of action that are directed medially (i.e. from the humeral head to the glenoid) will cause compression of the humeral head into the scapula and tend to stabilize the GH joint (An, 2002; Yanagawa et al., 2008). Furthermore, muscle forces are transmitted at a distance away from the joint center, enabling muscles to cause rotational movement. Muscle moment arms are the perpendicular distance between the joint center and muscle line of action. The moment arm indicates the direction of segment rotation (e.g. whether a muscle internally or externally rotates the humerus) and along with muscle size, quantifies the potential of a muscle to rotate the segment. It is important to note that each muscle's line of action and moment arm is defined 3-dimensionally. Consequently, a single muscle may have the capacity to translate and rotate segments across 3 orthogonal directions simultaneously. As the shoulder has relatively limited constraints to movement, there needs to be a delicate coordination of many muscles to allow task performance while minimizing undesirable rotations and translations (Veeger & van der Helm, 2007). Accordingly, concurrent activation of two muscles with opposing functions can function to stabilize the GH joint even if either muscle has the individual potential to de-stabilize the humeral head (Ackland & Pandy, 2009).

Table 2.5: The muscles crossing the shoulder complex, grouped by their attachment sites (Inman et al., 1944). Abbreviations for each muscle are provided in brackets. Note: some muscles (pectoralis major, deltoid, and trapezius) have sub-divisions attaching on to the clavicle, but are still grouped based on the entire muscle.

Muscle Group	Muscles
Scapulohumeral: Muscles attaching from the scapula to the humerus	**Deltoids**
	Anterior Deltoid
	Middle Deltoid
	Posterior Deltoid
	Rotator Cuff
	Infraspinatus
	Supraspinatus
	Subscapularis
	Teres Minor
	Teres Major
	Coracobrachialis
Upper Arm: Muscles attaching from the forearm to the scapula	**Biceps Brachii**
	Long Head
	Short Head
	Triceps Brachii
Axiohumeral: Muscles attaching from the trunk to the humerus	**Pectoralis Major**
	Clavicular Head
	Sternal Head
	Latissimus Dorsi
Axioscapular: Muscles attaching from the trunk to the scapula	**Trapezius**
	Upper Trapezius
	Middle Trapezius
	Lower Trapezius
	Levator Scapulae
	Pectoralis Minor
	Rhomboid
	Rhomboid Major
	Rhomboid Minor
	Serratus Anterior

2.2.2 Scapulohumeral Muscles

The muscles crossing the shoulder complex can be divided into four general groups based on their attachment sites (Inman et al., 1944) (Table 2.5). The group most extensively studied are the scapulohumeral muscles, which consist of the deltoid, rotator cuff, teres major, and coracobrachialis. The deltoid can be divided into three sub-regions (anterior, middle, posterior) based on the location and orientation of the muscle fibres. The anterior and middle deltoids are prominent elevators of the GH joint, with greater potential for elevation at higher angles as a result of increasing moment arms (Bassett et al., 1990; Otis et al., 1994; Kuechle et al., 1997; Liu et al., 1997; Ackland et al., 2008) (Figure 2.3). Changing the plane of elevation affects the relative contributions of these muscles to arm elevation, with the anterior deltoid showing greater leverage in the sagittal plane (i.e. forward flexion) than the middle deltoid, and vice versa in the frontal plane (i.e. abduction) (Kuechle et al., 1997). In contrast, the posterior deltoid typically depresses the humerus, but its moment arm vs. GH angle displays a "biphasic" pattern, indicating that it has the capacity to behave as a GH elevator in certain postures (Bassett et al., 1990; Otis et al., 1994; Kuechle et al., 1997; Liu et al., 1997; Ackland et al., 2008). In general, the posterior deltoid shifts from a GH depressor to an elevator at higher elevation angles; however, this pattern is dependent upon the plane of elevation (Kuechle et al., 1997; Ackland et al., 2008). The anterior and posterior deltoids are strong horizontal adductors and horizontal abductors at the GH joint respectively, with the middle deltoid having a small capacity for horizontal abduction (Kuechle et al., 1997). Each of the deltoids is weak at axially rotating the humerus. The anterior and posterior deltoids have small internal and external moment arms respectively, and the middle deltoid is observed to have the capacity to rotate in either direction, but to a negligible effect (Bassett et al., 1990; Kuechle et al., 2000; Ackland & Pandy, 2011). All three deltoid sub-regions have significant lines of action directed superiorly, which will result in an upward shear of the humeral head and potentially de-stabilize in the superior direction (Ackland & Pandy, 2009). Alternatively, this can be viewed as stabilizing against inferior translations of the humeral head and preventing inferior subluxation, which may occur if an external weight is held at the hand (in addition to the weight of the arm itself) (Halder et al., 2001a).

Similarly, all three deltoids can stabilize against anterior translations of the humeral in an abducted and externally rotated position (Kido et al., 2003). The anterior-posterior lines of action for the deltoids are inconsistent across studies. One investigation found the anterior deltoid (in the sagittal plane) and middle deltoid (in the sagittal and frontal planes) to have a significant posterior shear component throughout arm elevation (Ackland & Pandy, 2009), whereas another study found all three deltoids to have an anterior directed shear components during certain static frontal plane elevation angles (Lee & An, 2002). The entire deltoid is a complex muscle comprised of many different fibre directions, and it is possible that dividing it into 3 sub-regions may be an oversimplification that can lead to inconsistent results across studies in addition to individual variability in musculoskeletal geometry. In fact, a biomechanical model of the upper extremity divided the entire deltoid muscle into 14 different elements, each representing a distinct fibre direction (van der Helm, 1994). A meaningful component of the line of action for all sub-regions is directed medially, highlighting some capacity to compress and stabilize the humeral head against the scapula, albeit not to as great of an extent as the rotator cuff muscles (Lee & An, 2002). The posterior deltoid appears to be a functional intermediary between the anterior/middle deltoids and the rotator cuff muscles. Similar to its deltoid counterparts, the posterior deltoid has a significant prime mover role, but with smaller moment arms. However, a greater component of its line of action is directed towards compression than the anterior/middle deltoids, but not to the level of the rotator cuff muscles (Lee & An, 2002).

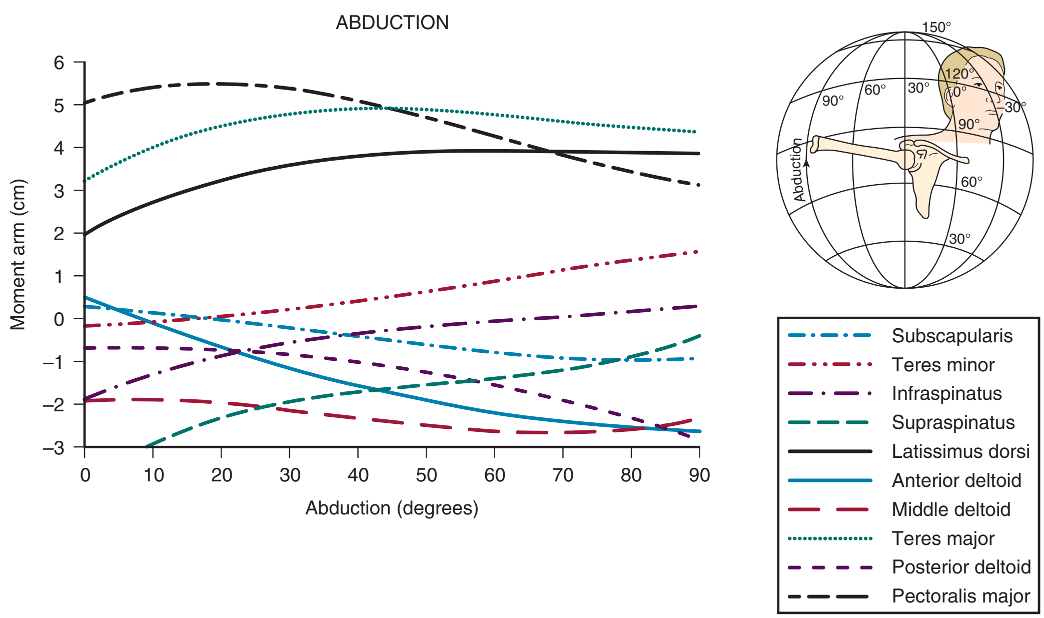


Figure 2.3: Moment arms for a number of scapulohumeral and axiohumeral shoulder muscles in the frontal plane of elevation (from Rockwood et al. 2009; redrawn from Kuechle et al. 1997).

As alluded to earlier, the rotator cuff muscles are considered the primary stabilizers at the GH joint (Figure 2.4). The rotator cuff muscles consist of the subscapularis, supraspinatus, infraspinatus, and teres minor. The lines of action for all 4 muscles are largely directed towards compression of the humeral head into the glenoid fossa (Lee et al., 2000; Lee & An, 2002; Ackland & Pandy, 2009). As a result, all four of these muscles can effectively stabilize the GH joint and minimize the potential for shear forces to translate the humeral head in the anterior-posterior and superior-inferior axes (Itoi et al., 1994a; Sharkey & Marder, 1995; Halder et al., 2001a, 2001b; Mura et al., 2003; Blasier et al., 2007; Soslowsky et al., 2007). The relative stabilizing potential between the rotator cuff muscles varies based on posture and direction, with some discrepancies observed across studies. Generally, the subscapularis, infraspinatus, and teres minor have greater potential to resist superior translations of the humeral head than the supraspinatus, due to lines of action that are directed more inferiorly (Sharkey & Marder, 1995; Lee et al., 2000; Halder et al., 2001b; Mura et al., 2003; Ackland & Pandy, 2009). The stabilizing potential for all the muscles peaks at around mid-elevation angles (Sharkey & Marder, 1995; Halder et al., 2001b), with the relative contribution of the compression component to the resultant force increasing for the infraspinatus and teres minor upon GH external rotation (Lee & An, 2002). The downward directed lines of action can cause the subscapularis, infraspinatus, and teres minor to be potentially de-stabilizing in the inferior direction (Ackland & Pandy, 2009); however, they function antagonistically well with the deltoids, which produce a superiorly directed shear force on the humerus upon elevation at the GH joint. In contrast, the supraspinatus is more effective at stabilizing the humeral head against inferior subluxation (Soslowsky et al., 1997). The subscapularis and supraspinatus have slight posterior directed lines of action, which is in contrast to the infraspinatus and teres minor (Lee et al., 2000; Lee & An, 2002; Ackland & Pandy, 2009); however, these patterns are switched upon external rotation of the humerus (Lee et al., 2000).

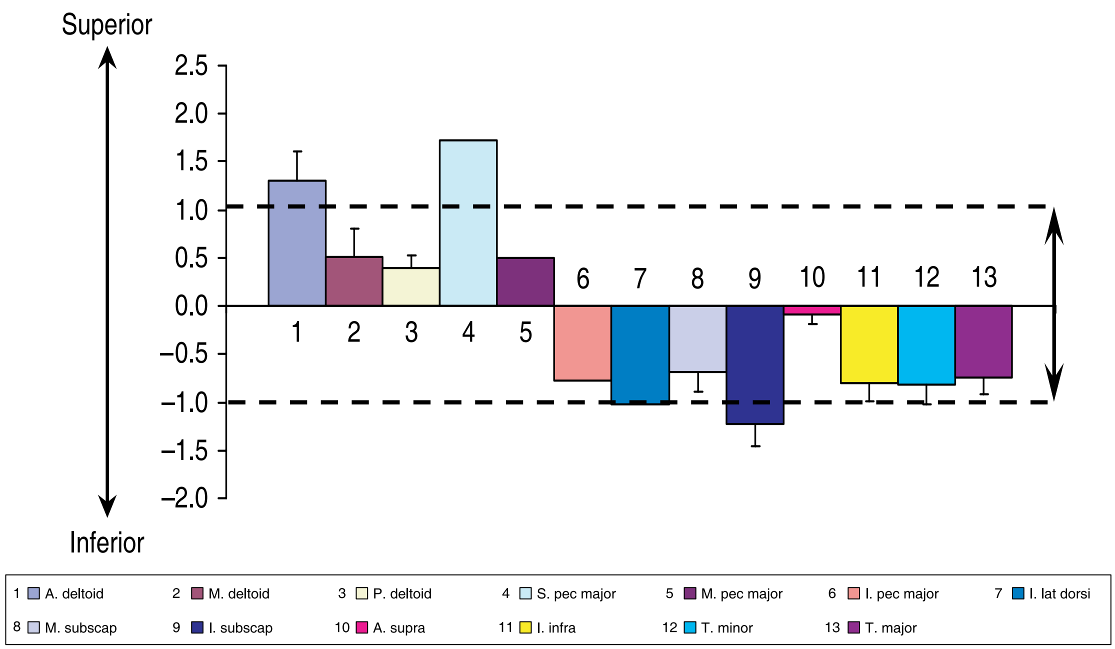


Figure 2.4: Superior/Inferior stability ratios for a number of axiohumeral and scapulohumeral muscles averaged across scapular plane humeral elevation. Stability ratios greater than 1.0 (or less than -1.0) indicate the muscle is de-stabilizing in the superior (or inferior) direction due to a greater superior-inferior shear force component than compression component (from Ackland & Pandy, 2009).

In addition to their stabilizing roles, each of the rotator cuff muscles has the capacity to provide rotational movement at the GH joint. The subscapularis is a strong internal rotator of the humerus, while the infraspinatus and teres minor are strong external rotators across all planes of elevation (Otis et al., 1994; Kuechle et al., 2000; Langenderfer et al., 2006a; Ackland & Pandy, 2011). The supraspinatus is a relatively weak rotator, transitioning from an external rotator in the frontal plane to an internal rotator in the sagittal plane (Otis et al., 1994; Kuechle et al., 2000; Langenderfer et al., 2006a; Ackland & Pandy, 2011). The external/internal rotation moment arms peak at slight external and internal rotated humerus angles for the subscapularis and infraspinatus/teres minor, respectively. The supraspinatus is generally considered to act as a strong elevator at initial GH elevation angles, with decreasing moment arm at greater angles (Kuechle et al., 1997; Liu et al., 1997; Ackland et al., 2008); however, opposing trends can be observed using different techniques to calculate moment arms (Hughes et al., 1998). There are also some discrepancies in the contribution of the remaining rotator cuff muscles to elevate and depress the humerus, with findings of "biphasic" functions (Kuechle et al., 1997). Generally, the subscapularis has a depression moment arm at the GH joint (Otis et al., 1994; Kuechle et al., 1997; Liu et al., 1997; Hughes et al., 1998; Ackland et al., 2008). Some findings have observed it to assist with elevation in the sagittal plane (Kuechle et al., 1997) and transition as an elevator at higher elevation angles (Ackland et al., 2008), but results are not consistent across studies. Similarly, the infraspinatus is found to assist with elevation at the GH joint, with some reports of a depression moment arm based on postures, planes of elevation, or specific to certain sub-regions of the muscle (Otis et al., 1994; Kuechle et al., 1997; Liu et al., 1997; Hughes et al., 1998; Ackland et al., 2008). The opposite is found for the teres minor – small depression moment arms with some cases for assistance an elevation moment arm (Otis et al., 1994; Kuechle et al., 1997; Ackland et al., 2008). Overall, the musculoskeletal geometry of the rotator cuff and deltoid muscles are well designed to function together in elevation of the arm while maintaining balance of the humeral head on the glenoid (Yanagawa et al., 2008).

The last two muscles comprising the scapulohumeral group are the teres major and the coracobrachialis. The teres major acts as a strong depressor at the GH joint across all planes of elevation (i.e. adduction in the frontal plane and extensor in the sagittal plane), with its moment arm reaching its peak magnitude at mid-elevation angles (Kuechle et al., 1997; Ackland et al., 2008). In addition, it has varying capacity to internally rotate the humerus and a limited ability to horizontally abduct at the GH joint (Kuechle et al., 1997, 2000; Ackland & Pandy, 2011). The line of action for the teres major is largely directed inferiorly and medially, causing compression, with a slight posterior component (Ackland & Pandy, 2009). Accordingly, it has a strong ability to inferiorly translate the humeral head that can counteract superior displacements. The coracobrachialis is not as extensively studied as the remaining scapulohumeral muscles, but is involved with elevation of the humerus in anterior planes (i.e. flexion in the sagittal plane) and depression of the humerus in posterior planes (i.e. adduction in the frontal plane) (Bassett et al., 1990). Based on these findings and its musculoskeletal geometry, it is presumed that the coracobrachialis can horizontally adduct the humerus. It also has some capacity to translate the humeral head superiorly and counteract inferior shear forces, although not to the same effect as the deltoids (Halder et al., 2001a).

2.2.3. Upper Arm Muscles

The upper arm muscles, biceps brachii and triceps brachii, are not immediately thought of as part of the shoulder complex, but also cross the GH joint with attachment sites on the scapula and the forearm. Although both muscles have prominent functions at the elbow, only their actions relevant to the shoulder will be discussed. The biceps brachii has been extensively studied to identify its capacity to stabilize the humerus, mostly in clinically unstable shoulder postures (i.e. elevation in the frontal plane with varying humeral rotation angles). The long head of biceps brachii can stabilize and limit humeral displacements when subjected to forces in the anterior-posterior and superior-inferior, presumably due to its compression oriented line of action (Kumar et al., 1989; Itoi et al., 1993, 1994a, 1994b; Rodosky et al., 1994; Pagnani et al., 1996; Blasier et al., 1997; Soslowsky et al., 1997; Halder et al., 2001b). It is more effective at limiting anterior-posterior and inferior translations than superior translations (Itoi et al., 1994b), but is posture-dependent. Specifically, the long head of the biceps brachii is more effective at minimizing anterior translations in neutral and internal rotated humeral postures, and posterior translations in external rotation (Pagnani et al., 1996; Blasier et al., 1997). The stabilizing capacity of the long head of the biceps brachii is comparable to that of the rotator cuff muscles (Itoi et al., 1994a; Soslowsky et al., 1997; Halder et al., 2001b). The short head of the biceps brachii and the triceps brachii are both observed to have limited capacity to stabilize against inferiorly directed shear forces, with the former also counteracting anterior shear forces (Itoi et al., 1993; Halder et al., 2001a).

2.2.4. Axiohumeral Muscles

The axiohumeral group consists of two broad-spanning muscles that attach from the trunk to the humerus that are greatly involved with humeral motion, pectoralis major and latissimus dorsi. Both muscles are strong depressors of the humerus in posterior planes of elevation (i.e. abduction in the frontal plane) (Bassett et al., 1990; Kuechle et al., 1997; Ackland et al., 2008). As the plane of elevation moves toward flexion, the pectoralis major transitions to elevate the humerus (i.e. flexion in the sagittal plane), with its superior fibres most effective in this role (Bassett et al., 1990; Ackland et al., 2008); however, one study found the pectoralis major to continue acting as a depressor in the sagittal plane (Kuechle et al., 1997). This may perhaps be due to representing the broad-spanning muscle with only a single line of action. In contrast, the latissimus dorsi remains a strong depressor of the humerus (i.e. extension in the sagittal plane) in anterior planes of elevation (Bassett et al., 1990; Kuechle et al., 1997; Ackland et al., 2008). Both muscles have a large capacity to internally rotate the humerus, although to a lesser extent than the subscapularis, with their moment arms peaking at low to mid ranges of arm elevation (Bassett et al., 1990; Kuechle et al., 2000; Ackland & Pandy, 2011). In addition, the pectoralis major is a strong horizontal adductor of the humerus, while the latissimus dorsi has limited capacity as a horizontal abductor (Kuechle et al., 1997). Finally, both muscles have significant de-stabilizing components to their lines of action at the GH joint. The resultant force of the pectoralis major has a large component directed anteriorly in the sagittal and frontal planes of elevation, which can promote anterior shear of the humeral head (Labriola et al., 2005; Ackland & Pandy, 2009). The latissimus dorsi has a large component of its line of action directed inferiorly and can promote downward displacement of the humerus; however, this capacity is lessened at increasing arm elevation angles, with the line of action orienting more towards compression (Ackland & Pandy, 2009). Similarly, the superiorly directed line of action for the upper fibres of the pectoralis major is lessened with increasing arm elevation (Ackland & Pandy, 2009). This finding is inconsistent with a prior study that found the pectoralis major to better promote inferior translations of the humerus and counteract superior shear forces at lower elevation angles (Halder et al., 2001b). As before, this may be due to the differences in representing the pectoralis major with a single line of action as opposed to multiple divisions based on the orientation of the fibres across different sub-regions.

2.2.5. Gaps in the Muscle Function Literature

To date, several well-designed investigations have been conducted to elucidate the roles of different shoulder muscles to movement and stability. However, there remain important gaps that limit our current understanding of shoulder muscle functions. The majority of investigations conducted to evaluate muscle function are cadaveric and EMG studies. The work presented in Section 2.2 focused primarily on the former. Although EMG can partially deduce muscle functions using activity patterns (Inman et al., 1944; Kronberg et al., 1990), it provides no information on the precise functional roles as quantified using moment arms and lines of action. Furthermore, notable discrepancies are observed in muscle moment arms and lines of actions across studies. General patterns in posture related changes to muscle moment arms and lines of action were described previously; however, some meaningful differences in directional changes across investigations were also identified (Section 2.2). Furthermore, the differences in magnitude of muscle moment arms can be quite large across experimental studies. For instance, the range of infraspinatus moment arm across 4 studies was 0-2.20 cm at 60° of

GH elevation in the scapular plane (Gatti et al., 2007). As muscle moment arms are small in magnitude, usually around 2 cm, a difference of a couple centimeters can be quite consequential. Methodological and anthropometric differences may explain some of these discrepancies. Two common methods for experimentally collecting muscle moment arms are using tendon excursion versus joint angle curves (i.e. principle of virtual work) and estimating muscle pathways after identifying muscle attachment sites (Hughes et al., 1998). Differences in methods can lead to large differences in both the magnitudes and directional changes muscle moment arms with motion (Hughes et al., 1998; Arnold et al., 2000; Pal et al., 2007). It is also possible that musculoskeletal geometry differences (e.g. muscle attachment sites) across subjects contribute to the variability observed across studies and likely affect muscle function (Inman et al., 1944; DeLuca & Forrest, 1973; Friederich & Brand, 1990; Garner & Pandy, 2001; Kaptein & van der Helm, 2004). Cadaveric studies are often limited to a small sample size (usually fewer than 10 samples). Moreover, many cadaveric investigations quantifying changes in moment arms or lines of action across different postures depict only regression lines or trend-lines fitting the data (e.g. Otis et al., 1994; Kuechle et al., 1997, 2000; Ackland & Pandy, 2009, 2011) (Figure 2.3). This makes it challenging to assess participant differences in muscle moment arms and lines of action; however, substantial variability across samples is reported or observed using standard deviations within some investigations (Bassett et al., 1990; Liu et al., 1997; Murray et al., 2002). Moving forward, musculoskeletal geometry differences should be evaluated for their effect on muscle function and to quantify the degree to which these differences explain the high degree of shoulder kinematic variability between individuals (Veeger & van der Helm, 2007).

2.3. Computer Modeling

Musculoskeletal modeling is a powerful computational tool that offers the potential to broaden our knowledge on the mechanical function of shoulder muscles across a range of upper extremity postures. To date, a number of upper extremity and shoulder models have been developed (Dvir & Berme, 1978; Högfors et al., 1987; Dul, 1988; Karlsson & Peterson, 1992; Raikova, 1992; van der Helm, 1994; Hughes & An, 1997; Garner & Pandy, 2001; Holzbaur et al., 2005; Charlton & Johnson, 2006; Dickerson et al., 2007; Blana et al., 2008; Favre et al., 2012; Nikooyan et al., 2012). These models typically use vectors replicating muscle pathways and three-dimensional bone geometry data to mathematically represent the muscular and skeletal systems of the human body. It is not feasible to directly measure muscle force (Dennerlein et al., 1998, 1999; Bey & Derwin, 2012). Accordingly, musculoskeletal models are an effective alternative to estimate internal joint loads experienced during movement and in response to external forces acting on the body (Erdemir et al,. 2007; Hicks et al., 2015).

One of the most widely visible models in the literature and will be the focus of this thesis is the Delft Shoulder and Elbow Model (DSEM) (van Der Helm, 1994). The DSEM was originally designed as a finite element model in the SPACAR software. As part of the shoulder, the DSEM includes the thorax, modelled as an ellipsoid, and 3D bone geometry representations of the clavicle, scapula, and humerus (Figure 2.5). The model includes the sternoclavicular (SC), acromioclavicular (AC), and glenohumeral

(GH) joints, each of which are modelled as a "ball and socket" with 3 rotational degrees of freedom, but no translational degrees of freedom. Twenty-nine muscles (20 specific to the shoulder) are included in the DSEM, the majority of which are composed of multiple elements with separate origin/insertion attachment sites and lines of action to represent the broad span of each muscle. Muscle origin/insertion locations were digitized based on a single cadaveric specimen and muscle pathways are modelled as the shortest line connecting the attachment sites for each muscle segment. To prevent muscles from passing through bones, wrapping objects are added to direct muscle pathways around bony contours. These consist of spheres around the humeral head and cylinders around the shaft of the humerus, specifically for the muscle pathways of the scapulohumeral (rotator cuff, deltoid) and axiohumeral groups (teres major, latissimus dorsi, pectoralis). In addition, the ellipsoid representing the thorax acts as a wrapping object for the serratus anterior.

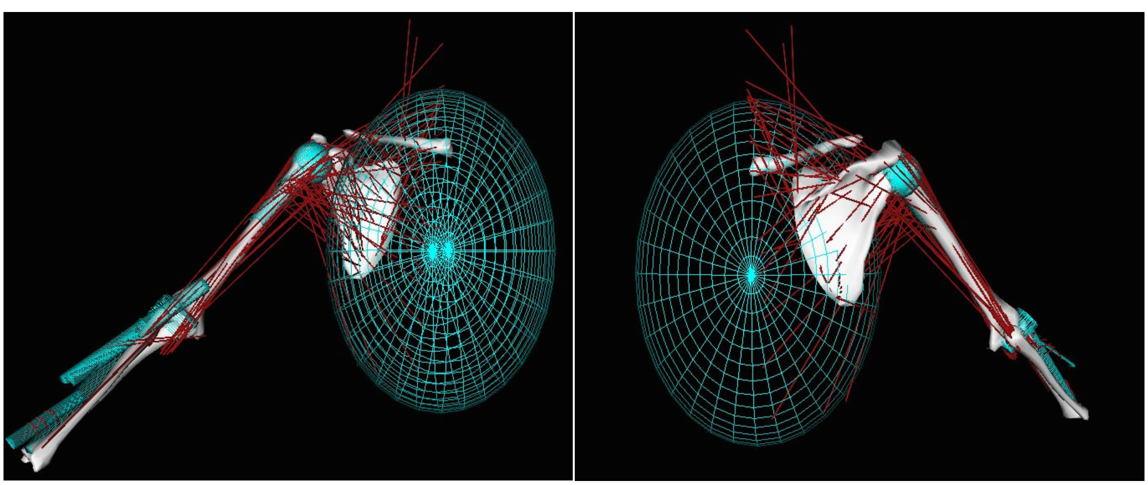


Figure 2.5: Anterior (left) and posterior (right) views of the Delft Shoulder and Elbow Model as currently available in OpenSim 3.3 (Stanford, CA, USA).

A significant advantage of modeling is that individual muscle force contributions to internal joint loads can be quantified, which can help clarify the degree to which different muscles contribute to movement and stability during different tasks (Yanagawa et al., 2008). One of the major obstacles with modeling is the indeterminacy problem, caused by the greater number of unknown variables (usually muscle forces) compared to the equilibrium equations available. There are multiple methods available to resolve this issue, but they are not without their limitations (Erdemir et al., 2007). Nevertheless, the individual muscle forces are not required to identify muscle roles, as muscle function, described using moment arms and lines of action, is independent of the magnitude of the forces (Yanagawa et al., 2008). As such, musculoskeletal models may also be used to predict musculoskeletal geometric parameters across different upper extremity postures and estimate muscle functional changes across the shoulder's ROM (Yanagawa et al., 2008; Favre et al., 2009). These models can expand upon the results from cadaveric work by assessing postures and evaluating conditions that are not formally tested during experimental trials (Gatti et al., 2007; Willemot et al., 2015). However, prior to doing so, the model-predicted muscle moment arm and lines of action need to be validated against experimentally measured data in postures previously examined and/or other models (Gatti et al., 2007; Hicks et al., 2015).

Probabilistic analysis is an effective method applied in conjunction with musculoskeletal modeling to predict outcomes with uncertain parameters among groups of heterogeneous individuals (Laz & Browne, 2009). Standard deterministic models often use input parameters estimated from population averages to predict outcomes. However, these outcomes only represent the “average” individual (Langenderfer et al., 2006b) and are unable to identify which individuals may be at risk for developing MSDs. The labour force consists of a diverse range of individuals that are not accurately represented by standard musculoskeletal models. Although current trends in modeling strongly encourage an individualized approach to account for subject-specificity in order for our models to reflect the varying characteristics of the population, it is challenging and time-consuming to determine input parameters for each individual (Bolsterlee et al., 2013). In contrast to deterministic models, probabilistic models predict a range and distribution of possible outcomes while incorporating uncertainty in input parameters (Laz & Browne, 2009). As such, probabilistic models incorporate variation across individuals and can be more robust when applying results across heterogeneous groups (Laz & Browne, 2009). Monte Carlo simulations are a type of probabilistic analysis where input parameters are randomly sampled from distributions of each input parameter. Previous use of probabilistic modeling has revealed that model-predicted muscle forces can be highly sensitive to muscle origin-insertion attachment sites (Langenderfer et al., 2006b; Bolsterlee & Zadpoor, 2014; Chopp-Hurley et al., 2014), likely as a result of altered moment arms and lines of action. However, the sensitivity of moment arms and lines of action throughout arm elevation to muscle attachment sites has not been formally evaluated. Doing so would allow us to understand the degree to which muscle musculoskeletal geometry can influence muscle function, and as a result, determine the influence of inter-individual musculoskeletal geometry difference on shoulder kinematic variability. In addition, probabilistic modeling can serve as an equally effective tool to provide robustness to study results (Hicks et al., 2015). For instance, there are no experimental data quantifying muscle moment arms and lines of action with changes in scapular orientation, preventing direct validation if quantified computationally. Alternatively, model-predicted moment arms and lines of action can be evaluated using different origin-insertion attachment sites across varying scapular kinematics to reflect the uncertainty in the results. In this capacity, the sensitivity analysis provides rigour to the methodology and increases the robustness of the conclusions.

2.4. Summary

The shoulder has multiple degrees of freedom, enabling individuals the capacity to utilize countless different movement and muscular strategies to perform a single task. Alterations in shoulder kinematics and muscle activities from typical, ideal ranges can be a mechanism for injury. However, only certain individuals are injured while others remain relatively healthy. For these reasons, it is believed that differences in movement and muscular strategies across individuals may have a significant role in placing workers at differential risk for upper extremity MSDs. Modeling is an effective tool at determining the mechanical function of muscles and can allow us to better discriminate muscular strategies based on risk for injury. Prior to doing so, biomechanical models need to be assessed to ensure that modeling efforts are aligned with experimental data available in literature. Musculoskeletal geometry is pivotal at determining the functional role of muscles, as muscle origin and insertion sites guide muscle pathways and ultimately influence moment arms and lines of action. There is potential for findings from modeling and experimental studies to differ due to discrepancies in musculoskeletal geometry. As such, model-predicted moment arms and lines of action need to be evaluated against existing experimental data. Although cadaveric studies have performed thorough investigations identifying the functional role of shoulder muscles across postures, there are a number of inconsistencies between studies, likely due to individual differences in musculoskeletal geometry. Over the past decade, greater emphasis is being placed on individualized models. To determine the need for subject-specific models, we should assess the sensitivity of muscle functional roles and model-estimated forces to variability in musculoskeletal geometry. Thus, the purpose of this thesis is to advance current understanding of shoulder muscle function and evaluate the effects of posture and musculoskeletal geometry on muscle function. To address the above purpose, the specific research objectives/questions of this thesis are (Figure 2.6):

i. Quantify the DSEM predicted functional roles for the scapulohumeral muscles.
ii. How do the DSEM predicted shoulder muscle functional roles, as defined using moment arms, lines of action, stability ratios, and muscle forces compare to literature data?
iii. How sensitive are model-estimated muscle forces to variations in musculoskeletal geometry?

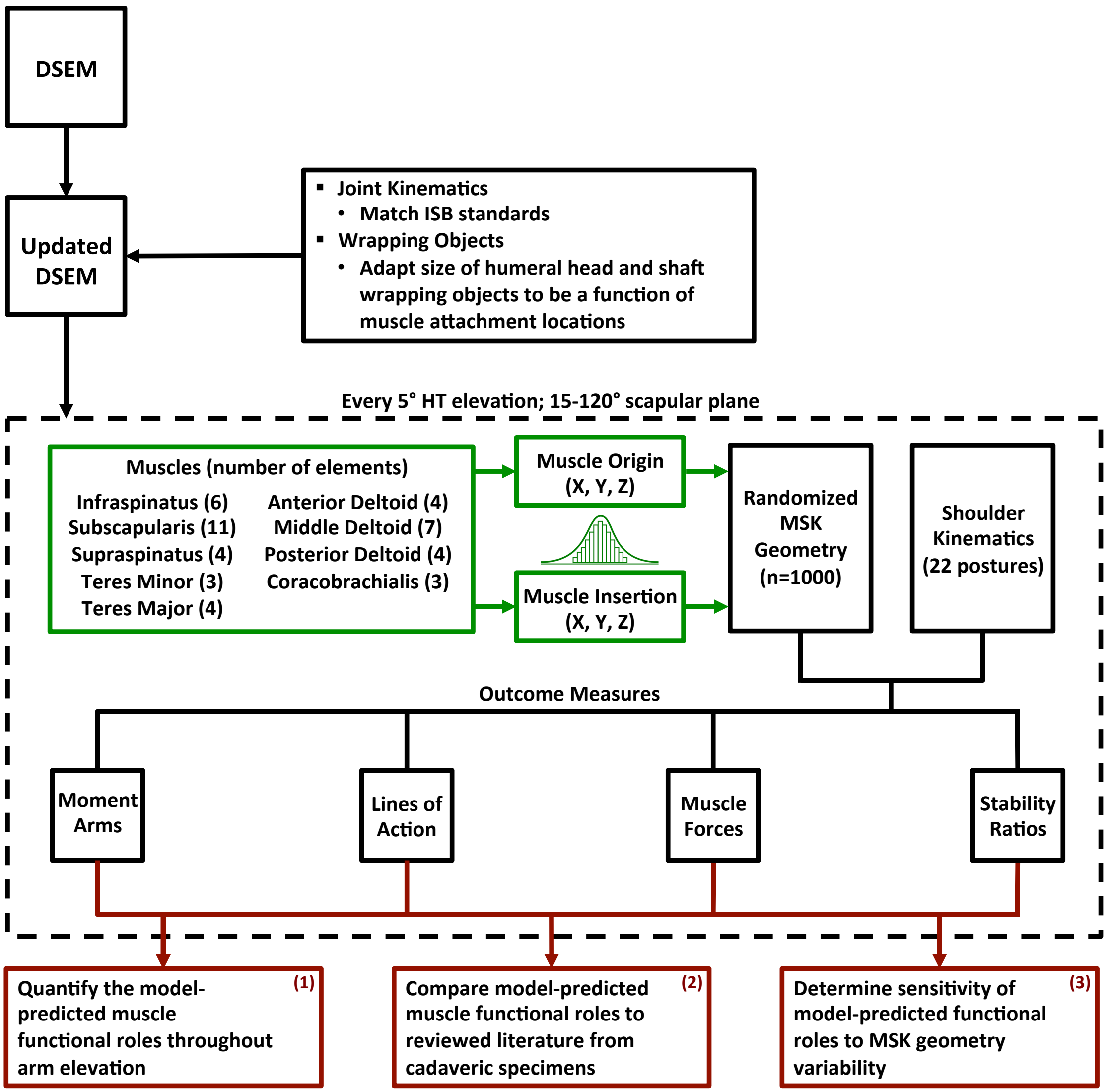


Figure 2.6: Overall plan and direction of thesis. Maroon coloured boxes denote the specific research objectives (numbers in brackets) identified on the previous page. The dotted box indicates the use of Monte Carlo simulations (i.e. probabilistic modeling), with the boxes coloured in green indicating the specific input parameters for which probability distributions were constructed.

CHAPTER THREE

Modeling the effects of musculoskeletal geometry on scapulohumeral muscle moment arms and lines of action

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3.1. Abstract

Biomechanical investigations examining shoulder function commonly observe a high degree of inter-individual variability in muscular and kinematic patterns during a range of tasks that incorporate static and/or dynamic upper extremity exertions. Substantial differences in musculoskeletal geometry between individuals can alter muscle moment arms and lines of action that would theoretically alter muscular/kinematic patterns. The purposes of this research were to quantify model-predicted functional roles (moment arms, lines of action) of the scapulohumeral muscles, compare model predictions to experimental data in the literature, and use probabilistic modeling to evaluate sensitivity of muscle functional due to changes in muscle attachment locations. Monte Carlo simulations were performed to iteratively adjust muscle attachment locations at the clavicle, scapula, and humerus of the Delft Shoulder and Elbow Model. Muscle moment arms, lines of action, and estimates of muscle force were quantified throughout arm elevation in the scapular plane. In general, model-predicted moment arms agreed well with the reviewed literature; however, notable inconsistencies were observed when comparing lines of action. Variability in moment arms and lines of action were muscle-specific, with 2 standard deviations in moment arm and line of actions as high as 25.8 mm and 30.0° for some muscles. Moment arms were particularly sensitive to changes in attachment site closest to the joint centre. The variations in muscle functional roles due to changes in musculoskeletal geometry are expected to require different muscle activity and movement patterns for upper extremity exertions.

3.2. Introduction

A recurrent theme across many biomechanical investigations of the shoulder is the high degree of variability in outcomes. Large discrepancies in kinematics are exhibited between participants during tasks ranging from planar arm motions (Graichen et al., 2000; McClure et al., 2001; Ludewig et al., 2004, 2009; Picco et al., 2017) to more complex, simulated work activities (Ebata, 2012; Tse et al., 2016; McDonald, 2017; Sandlund et al., 2017). Similarly, variability is noted in muscle coordination during static and dynamic tasks (Hammarskjöld et al., 1990; Mathiassen et al., 2003; Ebata, 2012; Hodder, 2012; Tse et al., 2016; McDonald, 2017). Further complicating matters, the high variability is observed within individuals (Mathiassen et al., 2003; Ebata, 2012; Samani et al., 2015; McDonald, 2017; Sandlund et al., 2017; Mulla et al., 2018), and in activity patterns across sub-regions of the same muscle (McCann et al., 1994; Holtermann et al., 2009; Kim et al., 2017). Recent research has found increasing evidence on individual-specific upper extremity muscular and movement adaptations to repetitive, fatiguing work (Gates & Dingwell, 2011; Chopp-Hurley et al., 2016a; Tse et al., 2016; McDonald et al., 2016; McDonald, 2017). In conjunction with evidence implicating neuromuscular and kinematic factors to shoulder injuries (Michener et al., 2003), researchers have hypothesized that high inter-individual variability may lead to differential risk for musculoskeletal disorders between workers. The shoulder is one of the most frequently injured parts of the body in the workplace (US Department of Labor, 2015; WSIB, 2016), thus, there is a need to determine causative factors of variation to better understand the development of work-related shoulder injuries such that they can be prevented.

Sources for the variation in neuromuscular and movement patterns include the numerous degrees of freedom of the shoulder complex controlled by a large group of muscles with overlapping functions. Accordingly, individuals are afforded the opportunity to perform a task using many muscular and kinematic strategies (McDonald et al., 2016); however, this does not directly explain why or how individuals use a particular strategy. Interestingly, movement differences between individuals may be partially explained by the high degree of variation observed in muscle moment arms (Bassett et al., 1990; Hughes et al., 1998; Langenderfer et al., 2006a) (Figure 3.1), and bone geometry (Boileau & Walch, 1997; Hertel et al., 2002; Chopp-Hurley et al., 2016b). Bone morphology is identified as a potential risk factor for shoulder injuries (Hughes et al., 2003; Moor et al., 2016) partly due to their impact on tissue compression within the subacromial space (Michener et al., 2003; Chopp-Hurley et al., 2016c). However, little research to date has been conducted on the modulation of muscular/kinematic strategies due to inter-individual variations in shoulder musculoskeletal geometry. Geometric differences in anatomy alter muscle functional roles by changing muscle moment arms and lines of action (Duda et al., 1996; Hughes et al., 2003) that can result in differences in muscle activity and movements between individuals.

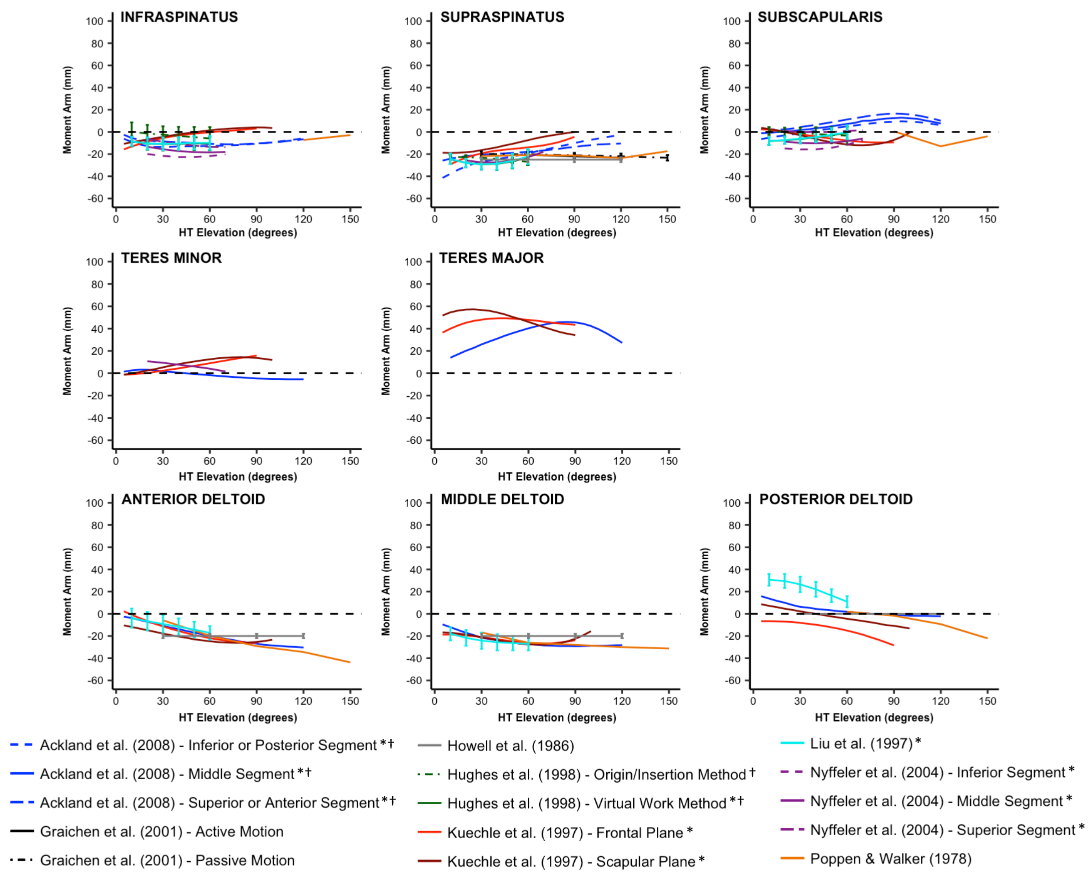

Figure 3.1: Compilation of experimentally measured glenohumeral elevation moment arms from the literature. Positive (negative) values indicate humeral depression (elevation). Lines represent mean values within each study across humeral elevation. Error bars indicating standard deviations are depicted as available. An asterisk (*) indicates the tendon excursion method was used for calculating moment arm. A dagger (†) indicates moment arms were measured in the frontal plane; otherwise all studies examined elevation in the scapular plane.

Computational modeling is a powerful tool to examine muscle function. Biomechanical models are developed mathematically by vector representation of muscle pathways and 3-D bone geometry data, providing insight into the mechanical function of muscles that may be limited by studies using electromyography (unable to quantify moment arms and lines of action) or cadavers (limited postures; change in tissue quality)

(Mansour & Pereira, 1987; Gatti et al., 2007). Many shoulder models have been developed to date (e.g. Högfors et al., 1987, 1991; Karlsson & Peterson, 1992; van der Helm, 1994a,b; Garner & Pandy, 2001; Holzbaur et al., 2005; Charlton & Johnson, 2006; Dickerson et al., 2007). One of the most extensively used models is the Delft Shoulder and Elbow Model (DSEM) (van der Helm, 1994a,b). Originally created from the shoulder complex musculoskeletal geometry of the median of 7 cadavers (van der Helm, 1992), the model was advanced with the inclusion of elbow musculoskeletal data (Veeger et al., 1997), muscle architecture parameters based upon a single cadaver (Minekus, 1997; Klein Breteler et al., 1999), and combined inverse-forward dynamics functionality (Nikooyan et al., 2011, 2012). A distinguishing feature of the DSEM is the high degree of anatomical fidelity. A total of 31 muscles are represented by 139 muscle elements to reflect the wide spanning, complex shoulder musculature that is insufficiently represented by a single line of action (van der Helm & Veenbaas, 1991; Ackland & Pandy, 2009; Webb et al., 2014). A modified version of the DSEM, incorporating the same musculoskeletal geometry dataset, but different algorithms for computing muscle pathways, was implemented into an open-source software, OpenSim (Blana et al., 2008). To ensure valid understanding of muscle functional roles through modeling, congruency of musculoskeletal geometry between biomechanical models and experimental data should be ensured (Hicks et al., 2015). Although good agreement was previously found between the DSEM predicted muscle moment arms and experimental data from cadavers (Gatti et al., 2007), the comparisons were limited to the rotator cuff muscles at two postures, thus more robust verification is needed.

Recent approaches to biomechanical modeling have emphasized inter-individual differences using probabilistic methods (Laz & Browne, 2010). By representing uncertainty in inputs using a probability distribution function, probabilistic modeling can predict the distribution of possible model-predicted outcomes (Laz & Browne, 2010). In the process, probabilistic approaches can serve to assess the model robustness to uncertainties in inputs (Hicks et al., 2015), and perhaps more importantly, may be used to differentiate individuals at the population level based on the sensitivity of an outcome of interest (e.g. injury risk) to certain inputs (Hughes & An, 1997; Langenderfer et al., 2006b; Flieg et al., 2008; Chopp-Hurley et al., 2016c). Using probabilistic modeling, the sensitivity of model-predicted outcomes to changes in a variety of inputs has been investigated, including anatomical landmarks, body segment parameters, muscle model parameters, and musculoskeletal geometry (e.g. Hoy et al., 1990; Scovil & Ronsky, 2006; Pal et al., 2007; Langenderfer et al., 2008; De Groote et al., 2010; Ackland et al., 2012; Chopp-Hurley et al., 2014, 2016c). Among the studies ranking the relative influence of the different inputs, model predictions at the lower extremity are consistently most sensitive to perturbations in musculoskeletal geometry (i.e. muscle attachment locations) and muscle model parameters (e.g. tendon slack length) (Carbone et al., 2012, 2016; Valente et al., 2014; Myers et al., 2015; Navacchia et al., 2016). At the shoulder, only two studies have quantified the sensitivity of muscle force predictions to perturbations in musculoskeletal geometry to date, with both reporting substantial effects of varying muscle attachment sites (Bolsterlee & Zadpoor, 2014; Chopp-Hurley et al., 2014). Although these investigations provide significant evidence into the large role musculoskeletal geometry can have on model outcomes, it remains uncertain how the exact shoulder muscle functional roles, as quantified by moment arms and lines of action, are affected by geometry. Determining these sensitivities could improve understanding of inter-individual anatomy on muscle function, and consequently, offer insight into the high variability in shoulder muscular and kinematic coordination.

The purposes of this study were to: (1) analyze the model-predicted functional roles of the scapulohumeral muscles throughout scapular plane humeral elevation; (2) compare model-predictions against reviewed literature; and (3) quantify the sensitivity of model-predicted functional roles to variations in muscle attachment locations using probabilistic modeling.

3.3. Methods

The modified DSEM (van der Helm, 1994a,b; Blana et al., 2008) was used in the current study. The model includes 3 rotational degrees of freedom at the sternoclavicular, acromioclavicular, and glenohumeral (GH) joints. Coordinate systems were transformed to match standardized conventions (Wu et al., 2005). The scapulohumeral muscles were of focus in the current study (Table 3.1). Each muscle is composed of a number of elements, representing distinct groups of muscle fascicles. Pathways are modelled as the shortest distance connecting attachment sites, with wrapping objects prevent pathways from passing through bones.

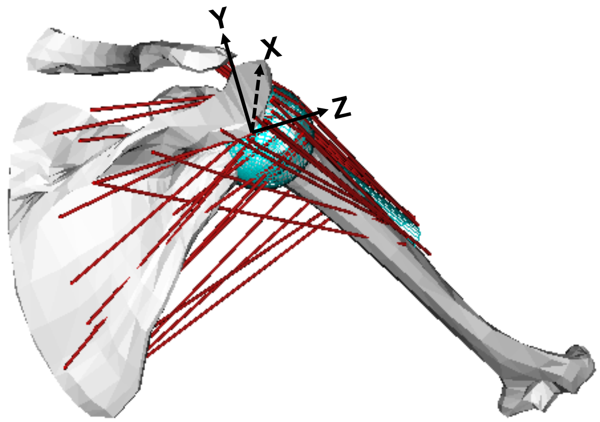

Figure 3.2: The Delft Shoulder and Elbow Model (DSEM) with the scapulohumeral muscles and humeral head/shaft wrapping objects visible (van der Helm, 1994a,b; Blana et al., 2008). Scapula coordinate system is shown, with the positive X-, Y-, and Z-axis referring to the anterior, superior, and lateral directions. Muscle lines of action are expressed as the clockwise angle from the Z-axis for the YZ plane (superior-inferior line of action) and XZ plane (anterior-posterior line of action).

Table 3.1: Means with 1 standard deviations in parentheses of the attachment locations at the clavicle* (C_x, C_y, C_z – anterior deltoid only), scapula (S_x, S_y, S_z), and humerus (H_x, H_y, H_z) for each muscle element as identified in the original DSEM (Högfors et al., 1987; van der Helm, 1994a,b; Blana et al., 2008). Attachment locations are defined with respect to the local coordinate system at the sternoclavicular (for clavicle muscle attachments), acromioclavicular (for scapular muscle attachments), and glenohumeral joints (for humeral muscle attachments).

Muscles and muscle sub-regions	Element	S_x or *C_x (m)	S_y or *C_y (m)	S_z or *C_z (m)	H_x (m)	H_y (m)	H_z (m)
Infraspinatus							
I. Infraspinatus	1	0.0190 (0.0055)	-0.0418 (0.0018)	-0.0609 (0.0069)	-0.0126 (0.0038)	0.0098 (0.0035)	0.0199 (0.0038)
(Inferior region)	2	-0.0012 (0.0055)	-0.0923 (0.0018)	-0.1261 (0.0069)	-0.0069 (0.0038)	0.0160 (0.0035)	0.0184 (0.0038)
	3	-0.0006 (0.0055)	-0.0726 (0.0018)	-0.1077 (0.0069)	-0.0069 (0.0038)	0.0160 (0.0035)	0.0184 (0.0038)
S. Infraspinatus	4	0.0012 (0.0025)	-0.0495 (0.0082)	-0.1091 (0.0078)	-0.0070 (0.0038)	0.0191 (0.0035)	0.0161 (0.0038)
(Superior region)	5	0.0107 (0.0025)	-0.0171 (0.0082)	-0.1044 (0.0078)	-0.0070 (0.0038)	0.0191 (0.0035)	0.0161 (0.0038)
	6	-0.0004 (0.0025)	-0.0004 (0.0082)	-0.0807 (0.0078)	-0.0084 (0.0038)	0.0127 (0.0035)	0.0192 (0.0038)
Supraspinatus							
P. Supraspinatus	1	0.0186 (0.0021)	0.0032 (0.0061)	-0.0603 (0.0099)	-0.0027 (0.0032)	0.0126 (0.0012)	0.0232 (0.0023)
(Posterior region)	2	0.0069 (0.0021)	0.0110 (0.0061)	-0.0596 (0.0099)	-0.0027 (0.0032)	0.0126 (0.0012)	0.0232 (0.0023)
A. Supraspinatus	3	0.0250 (0.0021)	0.0210 (0.0061)	-0.0944 (0.0099)	0.0137 (0.0032)	0.0192 (0.0012)	0.0123 (0.0023)
(Anterior region)	4	0.0403 (0.0021)	0.0212 (0.0061)	-0.0739 (0.0099)	0.0137 (0.0032)	0.0192 (0.0012)	0.0123 (0.0023)
Subscapularis							
S. Subscapularis	1	0.0229 (0.0020)	-0.0132 (0.0063)	-0.0619 (0.0076)	0.0192 (0.0023)	0.0165 (0.0009)	-0.0029 (0.0012)
(Superior region)	2	0.0281 (0.0020)	0.0120 (0.0063)	-0.0837 (0.0076)	0.0216 (0.0023)	0.0129 (0.0009)	-0.0005 (0.0012)
	3	0.0142 (0.0020)	-0.0145 (0.0063)	-0.1049 (0.0076)	0.0192 (0.0023)	0.0165 (0.0009)	-0.0029 (0.0012)
M. Subscapularis	4	0.0080 (0.0016)	-0.0436 (0.0030)	-0.0938 (0.0106)	0.0249 (0.0023)	0.0064 (0.0009)	0.0004 (0.0012)
(Middle region)	5	0.0122 (0.0016)	-0.0571 (0.0030)	-0.0874 (0.0106)	0.0246 (0.0023)	0.0086 (0.0009)	0.0043 (0.0012)
	6	0.0036 (0.0016)	-0.0791 (0.0030)	-0.1058 (0.0106)	0.0280 (0.0023)	0.0023 (0.0009)	0.0018 (0.0012)
I. Subscapularis (L)	7	0.0101 (0.0050)	-0.0947 (0.0030)	-0.0999 (0.0068)	0.0266 (0.0023)	0.0009 (0.0009)	-0.0024 (0.0012)
(Inferior region – long fibres)	8	0.0199 (0.0050)	-0.0655 (0.0030)	-0.0685 (0.0068)	0.0266 (0.0023)	0.0009 (0.0009)	-0.0024 (0.0012)
	9	0.0123 (0.0050)	-0.0921 (0.0030)	-0.0798 (0.0068)	0.0233 (0.0023)	-0.0037 (0.0009)	-0.0066 (0.0012)
I. Subscapularis (S)	10	0.0211 (0.0050)	-0.0543 (0.0030)	-0.0464 (0.0068)	0.0155 (0.0023)	-0.0170 (0.0009)	-0.0074 (0.0012)
(Inferior region – short fibres)	11	0.0288 (0.0050)	-0.0457 (0.0030)	-0.0361 (0.0068)	0.0103 (0.0023)	-0.0229 (0.0009)	-0.0097 (0.0012)

Teres Minor	1	0.0141 (0.0019)	-0.0665 (0.0047)	-0.0557 (0.0074)	-0.0088 (0.0023)	-0.0070 (0.0017)	0.0258 (0.0055)
	2	0.0065 (0.0019)	-0.0691 (0.0047)	-0.0659 (0.0074)	-0.0080 (0.0023)	-0.0012 (0.0017)	0.0281 (0.0055)
	3	0.0170 (0.0019)	-0.0421 (0.0047)	-0.0381 (0.0074)	-0.0088 (0.0023)	-0.0070 (0.0017)	0.0258 (0.0055)
Teres Major							
	1	0.0131 (0.0009)	-0.1049 (0.0033)	-0.0874 (0.0052)	0.0094 (0.0020)	-0.0656 (0.0069)	-0.0060 (0.0009)
	2	0.0105 (0.0009)	-0.1125 (0.0033)	-0.1027 (0.0052)	0.0095 (0.0020)	-0.0843 (0.0069)	-0.0029 (0.0009)
	3	-0.0001 (0.0009)	-0.0901 (0.0033)	-0.0912 (0.0052)	0.0090 (0.0020)	-0.0784 (0.0069)	-0.0038 (0.0009)
	4	0.0111 (0.0009)	-0.0955 (0.0033)	-0.0797 (0.0052)	0.0107 (0.0020)	-0.0498 (0.0069)	-0.0065 (0.0009)
Deltoid (Clavicular)*							
Anterior (A.) Deltoid (1)*	1	-0.0013 (0.0024)	0.0207 (0.0075)	0.1509 (0.0022)	0.0086 (0.0012)	-0.0923 (0.0066)	0.0181 (0.0023)
Anterior (A.) Deltoid (2-4)*	2	-0.0050 (0.0024)	0.0258 (0.0075)	0.1291 (0.0022)	0.0116 (0.0012)	-0.1188 (0.0066)	0.0154 (0.0023)
	3	-0.0007 (0.0024)	0.0223 (0.0075)	0.1343 (0.0022)	0.0097 (0.0012)	-0.0994 (0.0066)	0.0167 (0.0023)
	4	-0.0009 (0.0024)	0.0164 (0.0075)	0.1320 (0.0022)	0.0116 (0.0012)	-0.1188 (0.0066)	0.0154 (0.0023)
Deltoid (Scapular)							
Posterior (P.) Deltoid	1	0.0009 (0.0049)	-0.0008 (0.0009)	-0.0828 (0.0038)	-0.0092 (0.0012)	-0.1113 (0.0066)	0.0135 (0.0023)
	2	-0.0011 (0.0049)	0.0063 (0.0009)	-0.0500 (0.0038)	-0.0053 (0.0012)	-0.0965 (0.0066)	0.0156 (0.0023)
	3	0.0014 (0.0049)	0.0057 (0.0009)	-0.0144 (0.0038)	-0.0053 (0.0012)	-0.0965 (0.0066)	0.0156 (0.0023)
	4	0.0048 (0.0049)	-0.0045 (0.0009)	-0.0030 (0.0038)	-0.0040 (0.0012)	-0.0847 (0.0066)	0.0157 (0.0023)
Middle (M.) Deltoid (1-4)	5	0.0173 (0.0004)	0.0008 (0.0009)	0.0130 (0.0008)	-0.0007 (0.0012)	-0.0778 (0.0066)	0.0168 (0.0023)
	6	0.0050 (0.0004)	0.0053 (0.0009)	0.0036 (0.0008)	-0.0066 (0.0012)	-0.1089 (0.0066)	0.0165 (0.0023)
	7	0.0092 (0.0004)	0.0003 (0.0009)	0.0085 (0.0008)	-0.0052 (0.0012)	-0.1316 (0.0066)	0.0158 (0.0023)
	8	0.0181 (0.0004)	0.0062 (0.0009)	0.0141 (0.0008)	-0.0066 (0.0012)	-0.1089 (0.0066)	0.0165 (0.0023)
Middle (M.) Deltoid (5-7)	9	0.0241 (0.0004)	0.0048 (0.0009)	0.0157 (0.0008)	0.0049 (0.0012)	-0.1174 (0.0066)	0.0185 (0.0023)
	10	0.0331 (0.0004)	0.0051 (0.0009)	0.0177 (0.0008)	0.0061 (0.0012)	-0.1059 (0.0066)	0.0183 (0.0023)
	11	0.0451 (0.0004)	0.0082 (0.0009)	0.0152 (0.0008)	0.0086 (0.0012)	-0.0923 (0.0066)	0.0181 (0.0023)
Coracobrachialis							
	1	0.0766 (0.0013)	-0.0086 (0.0011)	-0.0081 (0.0104)	0.0077 (0.0020)	-0.1463 (0.0040)	-0.0027 (0.0012)
	2	0.0786 (0.0013)	-0.0149 (0.0011)	-0.0135 (0.0104)	0.0018 (0.0020)	-0.1649 (0.0040)	-0.0076 (0.0012)
	3	0.0826 (0.0013)	-0.0082 (0.0011)	-0.0093 (0.0104)	0.0056 (0.0020)	-0.1739 (0.0040)	-0.0053 (0.0012)

Model-predicted moment arms, lines of action, and muscle forces were quantified throughout arm elevation in the scapular plane. The upper extremity was postured in a series of static thoracohumeral (HT) elevation angles from 15-120° at 5° intervals based on available in vivo data measured using bone pins (Ludewig et al., 2009). Moment arms were quantified using the tendon excursion method at each axis of the GH joint: elevation/*depression*, *horizontal adduction*/abduction, and *internal*/external rotation (*positive directions italicized*). Lines of action were computed as the unit vector direction cosine of the muscle pathway from humeral to scapula attachment (or clavicle for the anterior deltoid) defined in the scapula coordinate system (Figure 3.2). For muscle pathways directed around wrapping object, the sum of the resultant vectors at each path-point along the humerus was computed. The lines of action in the superior-inferior (LOA_{S-I}) and anterior-posterior (LOA_{A-P}) directions were calculated as the angle clockwise to the Z-axis in the YZ and XZ planes, respectively (Figure 3.2). The Hill-type musculotendon model incorporating the force-length-velocity properties of a musculotendon actuator was updated (Thelen, 2003). As changes in muscle attachments can alter the musculotendon length, it is expected to affect the operating range on the force-length curve and muscle force. As such, muscle forces were computed at a muscle activation of 1.0 (scale: 0-1.0) to quantify sensitivity of muscle forces to changes in muscle attachment locations.

Monte Carlo simulations were performed to evaluate sensitivity of model-predicted moment arms, lines of action, and muscle forces to muscle attachment location alterations. Univariate normal distributions were generated for each muscle's attachment locations (Table 3.1). To construct these normal distributions, current model muscle attachment sites were used as the mean origin and insertion locations. Standard deviations were quantified using previous experimental data collected in cadavers, and were linearly scaled to the model's anthropometrics (Högfors et al., 1987). Thus, 6 input parameters were represented by normal distributions for each muscle's attachments: XYZ location at either the clavicle (C_x, C_y, C_z – anterior deltoid only) or scapula (S_x, S_y, S_z), and XYZ location at the humerus (H_x, H_y, H_z). Muscle wrapping objects were adjusted with each simulation as a function of humeral attachment location. Specifically, the humeral head was represented by a spherical wrapping object located at the center of the humeral head, with its radius set as the distance between the humeral head center and the humeral attachment location of the rotator cuff muscles. A separate spherical wrapping object was applied to each element of all 4 rotator cuff muscles. A cylindrical wrapping object centered along the humeral shaft axis represented the proximal humeral shaft, with its radius set as the distance between the axis and the humeral attachment of each element of the teres major. The deltoid wrapping object was unaltered from the original model. Each muscle's optimal fibre length and tendon slack length were linearly scaled to maintain relative percentage to the total musculotendon length (Delp et al., 2007). Based on the normal distributions, each muscle's attachment locations were randomly sampled and the muscle moment arms, lines of action, and forces were quantified. Initial simulations indicated a few hundred simulations were needed for the solutions to converge to a steady state (Valero-Cuevas et al., 2003), thus all Monte Carlo simulations were run to 1000 iterations. The probabilistic modeling process and model outcomes were programmed and computed using the Application Programmer's Interface between OpenSim 3.3 (Stanford, CA, USA) and Matlab 2017b (Mathworks, MA, USA). Model-predicted moment arms and lines of action were assessed against reviewed literature data from experimental studies (Figure 3.1).

Data across some elements were grouped into muscle sub-regions based on either scapula attachment location (Ackland et al., 2008) or displaying differing lines of action changes with elevation (Table 3.1). As a result, 46 muscle elements representing the 8 scapulohumeral muscles were analyzed as 16 muscle sub-regions. Means ± 2 standard deviations (representing 95% of the distribution) were calculated from 15-120° of humeral elevation at 5° intervals, and averaged across all angles for each scapulohumeral muscle sub-region for the following outcomes: glenohumeral moment arm, lines of action, and estimated muscle forces.

Multiple regressions were used to quantify the sensitivity of moment arms and lines of action to alterations in muscle attachment location. To aid with interpretation of the unstandardized beta coefficients and for comparisons across muscles, the attachment locations were centered with respect to the mean and divided by the standard deviations (i.e. a coefficient of 2 is interpreted as a 1 standard deviation change in the predictor variable causes a 2 unit increase in the outcome measure). All 6 attachment locations for each muscle were entered into the regressions. The dependent variables for the regression analyses were moment arms and lines of action. As moment arms and lines of action change non-linearly as a function of posture, elevation angle was added as a covariate to each model. For models predicting moment arms, humeral elevation relative to thorax

(HT) and its quadratic (HT^2) and cubic (HT^3) terms were included. Only the linear (HT) and quadratic terms (HT^2) were included for models predicting lines of action. Assumptions of normality and homogeneity of variance were assessed and verified upon visual inspection of the data. Model fit was assessed using R^2. All statistical analyses were performed using STATA 14.2 (StataCorp LLC, TX, USA).

3.4. Results

3.4.1. Glenohumeral Elevation/Depression Moment Arms

Mean model-predicted elevation/depression moment arms ranged from -29.2 to 54.9 mm depending on the muscle and elevation angle (Tables 3.2, 3.3). The teres major and posterior deltoid displayed the largest depression moment arms. The anterior and middle deltoid, infraspinatus, supraspinatus, and coracobrachialis all exhibited substantial capacity for humeral elevation. The subscapularis, teres minor, and posterior deltoid displayed capacity to elevate or depress the arm depending on humeral elevation angle, muscle sub-region, and attachment location. There was general agreement in magnitude and angle-dependent changes between the model-predicted moment arms with the reviewed literature. Largest deviations between the model-predictions and literature data were observed for the subscapularis, posterior deltoid, and teres major. The model-predicted subscapularis moment arms, specifically the inferior sub-region – short fibres, display a considerable glenohumeral depression moment arm throughout scapular plane elevation, which is not observed until higher elevation angles in the reviewed literature. The model-predicted posterior deltoid and teres major moment arms exhibited greater capacity for glenohumeral depression compared to the reviewed literature. Figures/tables for horizontal adduction/abduction and internal/external rotation moment arms are provided in Appendix A.

Table 3.2: Mean (± 2 standard deviations) values for model-predicted moment arms, lines of action, and force averaged across scapular plane elevation and grouped by scapulohumeral muscle sub-regions. Glenohumeral moment arms (MA) refer to elevation/*depression* (GH_x), *horizontal adduction*/horizontal abduction (GH_y), and *internal*/external rotation (GH_{yy}). Positive directions are italicized. Lines of action are in the superior/inferior direction (LOA_{S-I}) and anterior/posterior directions (LOA_{A-P}). Muscle forces were estimated at an activation of 1.0.

Muscle	GH_x MA (mm)	GH_y MA (mm)	GH_{yy} MA (mm)	LOA_{S-I} (°)	LOA_{A-P} (°)	Force (N)
S. Infraspinatus	-13.4 (9.5)	-19.9 (7.7)	-11.4 (7.1)	185.8 (24.4)	191.3 (8.6)	617.4 (22.3)
I. Infraspinatus	-12.5 (10.1)	-20.6 (8.2)	-17.6 (7.5)	206.9 (11.8)	186.9 (13.5)	574.1 (33.6)
A. Supraspinatus	-20.7 (5.8)	-1.7 (5.4)	8.2 (4.5)	167.0 (9.6)	185.7 (9.9)	362.8 (11.8)
P. Supraspinatus	-13.8 (5.4)	-19.2 (5.1)	-6.6 (4.3)	169.0 (12.5)	190.2 (11.5)	226.1 (5.4)
S. Subscapularis	-10.8 (10.6)	18.7 (7.8)	13.3 (7.5)	174.9 (26.6)	207.5 (30.0)	341.0 (4.9)
M. Subscapularis	-13.7 (8.0)	17.8 (6.1)	20.1 (6.4)	211.2 (12.5)	211.5 (7.7)	487.0 (6.4)
I. Subscapularis (L)	-1.8 (7.3)	19.5 (4.6)	23.0 (5.1)	221.9 (7.1)	215.5 (10.3)	262.1 (4.4)
I. Subscapularis (S)	18.1 (6.4)	15.9 (4.9)	14.3 (4.6)	191.9 (22.2)	224.1 (12.1)	60.7 (0.8)
Teres Minor	4.7 (6.7)	-23.4 (9.7)	-24.3 (9.7)	201.4 (16.1)	185.4 (12.8)	485.2 (7.2)
Teres Major	54.9 (20.2)	13.6 (4.4)	8.4 (3.8)	194.0 (13.2)	209.2 (5.1)	569.2 (6.8)
A. Deltoid (1)	-29.2 (8.0)	11.6 (3.2)	5.6 (1.9)	126.8 (7.5)	199.2 (10.7)	98.3 (6.7)
A. Deltoid (2-4)	-28.3 (9.3)	21.6 (3.6)	6.3 (2.0)	127.7 (5.4)	181.3 (7.5)	301.0 (10.4)
M. Deltoid (1-4)	-6.9 (8.5)	-17.1 (2.7)	-0.2 (1.8)	122.2 (4.2)	224.0 (8.2)	760.4 (1.8)
M. Deltoid (5-7)	-23.0 (10.1)	-9.9 (9.3)	0.3 (2.3)	129.1 (7.4)	208.5 (9.1)	875.8 (11.2)
P. Deltoid	16.4 (25.8)	-13.0 (13.2)	-3.1 (3.0)	135.3 (13.6)	212.2 (18.8)	1045.4 (3.1)
Coracobrachialis	-14.7 (14.1)	20.5 (6.8)	-0.4 (1.4)	125.8 (5.7)	188.5 (8.3)	439.9 (23.2)

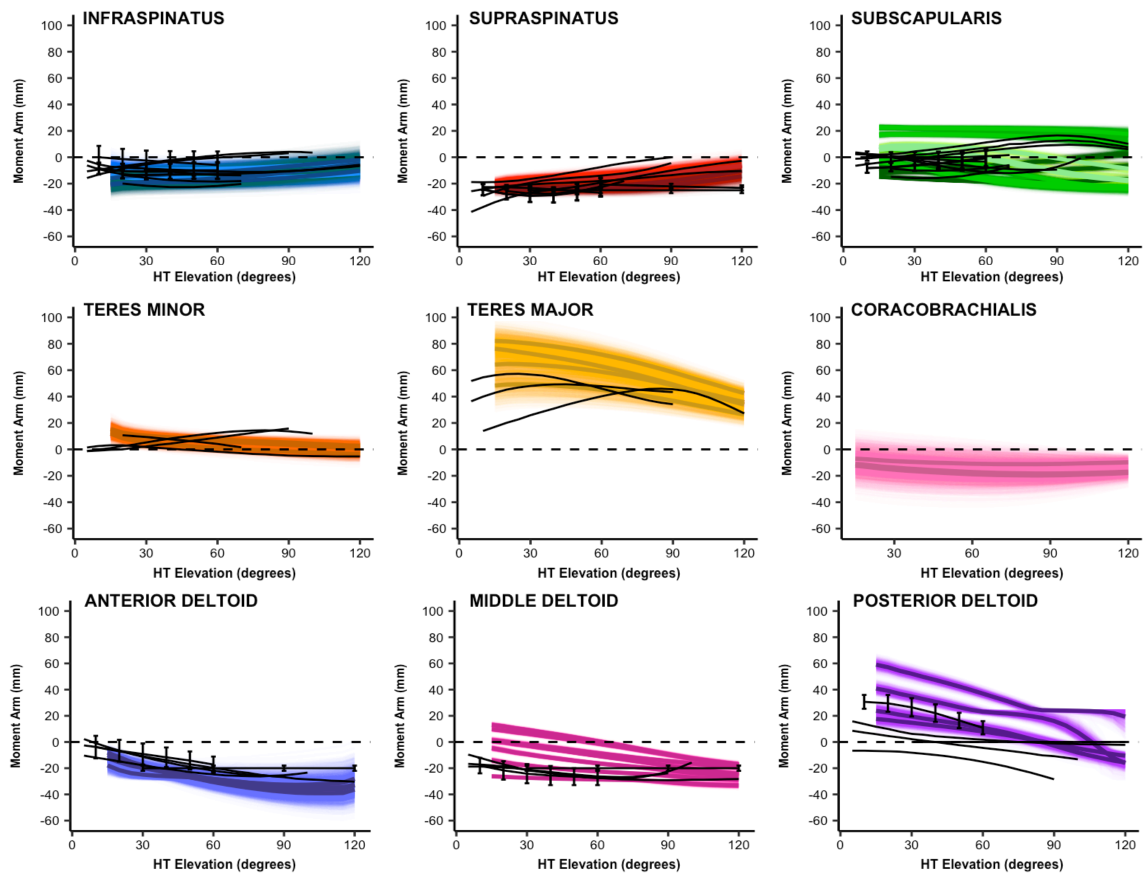

Figure 3.3: Model-predicted humeral elevation moment arms with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Positive (negative) values indicate humeral depression (elevation). Black lines represent experimental data available in the literature as shown in Figure 3.2.

There was high degree variability in model-predicted elevation/depression moment arm values due to changes in muscle attachment locations (Figure 3.3). Variation in moment arms, quantified absolutely as 2 standard deviations around the mean, ranged from 5.4-25.8 mm depending on the muscle (Table 3.2). The coefficients of variation for the mean model-predicted moment were, on average, 46.2% (range: 13.7-202.8%) of the mean magnitudes across the muscles. In general, the moment arms for the rotator cuff muscles and teres major were most sensitive to changes in humeral attachments, especially in the superior/inferior axis (H_y) (Table 3.3). The teres major (H_y = -9.9 mm; note: negative value indicates a decreased elevation or increased depression moment arm), infraspinatus (H_y = -3.2 and -4.2 mm), and teres minor (H_y = -3.0 mm) were the most sensitive to vertical changes in the humeral attachment location, with a 1 standard deviation change in the humeral attachment superiorly predicting change in moment arm of at least 3 mm for each muscle. In contrast, the deltoids and coracobrachialis muscles displayed greater sensitivity to scapular/clavicular attachment changes (Table 3.3). The scapular attachments along the medial/lateral axis had a strong influence on the moment arms for the posterior deltoid (S_z = -11.4 mm) and coracobrachialis (S_z = -6.4 mm).

Table 3.3: Regression models predicting humeral elevation moment arms (mm) for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) and cubic (HT^3) terms were added as covariates. Values represent unstandardized beta coefficients.

Muscle	B_0	S_x or *C_x	S_y or *C_y	S_z or *C_z	H_x	H_y	H_z	HT	HT^2	HT^3	R^2
S. Infraspinatus	-19.9	0.2	1.5	-0.1	-2.1	-3.2	-1.3	9.2E-02	-5.9E-04	6.9E-06	0.93
I. Infraspinatus	-10.0	0.1	0.1	-0.1	-1.5	-4.2	-0.6	-1.7E-01	2.0E-03	-4.9E-06	0.94
A. Supraspinatus	-23.2	1.1	0.4	0.1	-2.3	-0.5	-0.9	-3.2E-02	1.8E-04	7.0E-06	0.93
P. Supraspinatus	-23.2	0.3	0.5	0.1	-2.4	-0.5	-1.0	2.3E-01	-2.3E-03	1.3E-05	0.97
S. Subscapularis	-10.8	0.8	2.6	2.6	0.2	-1.4	0.4	-7.1E-02	8.1E-04	5.2E-07	0.83
M. Subscapularis	4.2	0.3	-2.0	0.2	-0.8	-2.5	-0.8	-2.2E-01	-2.2E-03	1.8E-05	0.92
I. Subscapularis (L)	3.5	0.5	-2.1	1.2	-0.3	-2.3	-0.1	3.3E-01	-8.1E-03	3.4E-05	0.96
I. Subscapularis (S)	20.1	-0.5	-0.5	0.9	-0.3	-3.1	0.2	-2.3E-02	6.4E-04	-7.8E-06	0.91
Teres Minor	17.9	0.3	-1.2	0.4	-0.2	-3.0	-0.1	-4.2E-01	4.4E-03	-1.8E-05	0.88
Teres Major	67.1	0.1	-2.2	0.6	-0.5	-9.9	-0.4	1.1E-01	-4.8E-03	1.3E-05	0.97
A. Deltoid (1)*	-16.6	-0.8	-3.8	-0.4	-0.1	0.1	-0.2	-2.4E-01	8.7E-05	5.6E-06	0.88
A. Deltoid (2-4)*	-2.5	-1.4	-4.7	0.0	-0.2	0.4	-0.1	-5.7E-01	1.3E-03	1.1E-05	0.93
M. Deltoid (1-4)	9.2	-2.1	-1.4	-1.9	-0.4	0.0	-0.3	-1.8E-01	-1.4E-03	7.2E-06	0.98
M. Deltoid (5-7)	-11.5	-4.5	-0.7	-0.5	-0.2	0.0	0.0	-2.7E-01	1.9E-03	-7.5E-06	0.93
P. Deltoid	41.7	-1.0	-2.1	-11.4	-0.5	-0.7	-0.5	-4.3E-01	2.1E-03	-1.5E-05	0.95
Coracobrachialis	-7.0	-1.2	-1.7	-6.4	-0.1	0.0	0.0	-2.3E-01	1.5E-03	-1.2E-06	0.95

3.4.4. Lines of Action

Model-predicted lines of action for the rotator cuff and teres major were primarily directed inferiorly and posteriorly throughout elevation, with a high degree of variability in some muscles due to attachment changes (Figures 3.4, 3.5). Variability (± 2 standard deviations) in the superior-inferior and anterior/posterior lines of action ranged from 7.1-26.6° and 5.1-30.0° respectively, with greater inconsistencies observed for the superior sub-regions of the infraspinatus and subscapularis (Table 3.2). As a result, these sub-regions could have a superiorly directed line of action depending on attachment location. The model-predicted supraspinatus line of action was directed superiorly, contrasting with literature. Model lines of action for the teres major and teres minor muscles under-predicted in the inferior direction and over-predicted in the posterior direction.

Model-predicted lines of action for the coracobrachialis and deltoids were directed superiorly at low elevation angles and reduced with increasing elevation (Figures 3.4, 3.5). Changes in the superior-inferior lines of action for these muscles were not as sensitive as the rotator cuff muscles, with the greatest variation observed for the posterior deltoid (13.6°) but less than 8° for the other deltoids and coracobrachialis (Table 3.2). Each deltoid and coracobrachialis had a posteriorly-directed line of action at elevations over 60°, with the coracobrachialis and few muscle elements of the anterior/middle deltoids displaying an anteriorly-directed line of action at lower elevation. In general, the model predicted deltoid muscles lines of action that were more superiorly directed throughout arm elevation compared to the reviewed literature data. In addition, model predictions displayed a greater posteriorly directed line of action for the deltoids compared to literature data. Model-predicted lines of action sensitivity to individual attachment changes are reported in Tables 3.4 and 3.5.

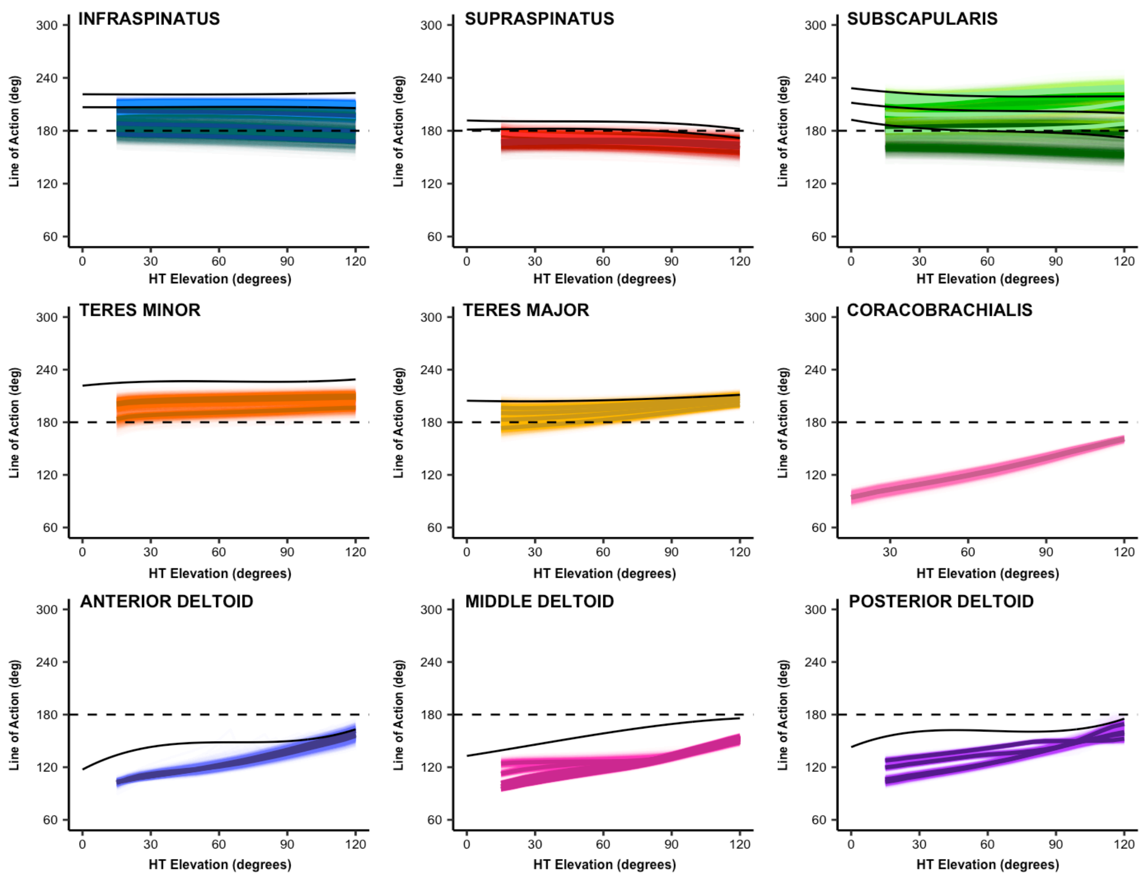

Figure 3.4: Model-predicted glenohumeral superior-inferior lines of action with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Angles less than (or greater than) 180° represented superiorly (or inferiorly) directed muscle lines of action. Black lines represent experimental data from Ackland & Pandy, 2009.

Table 3.4: Regression models predicting superior-inferior lines of action (°) for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) term were added as covariates. Values represent unstandardized beta coefficients. Angles between 0-180° indicate a superiorly-directed line of action, with values greater than 180° indicating an inferiorly-directed line of action.

Muscle	B_0	S_x or *C_x	S_y or *C_y	S_z or *C_z	H_x	H_y	H_z	HT	HT^2	R^2
S. Infraspinatus	188.8	0.70	-11.60	0.30	1.20	1.80	0.70	-5.6E-03	-4.6E-04	0.99
I. Infraspinatus	206.3	-0.40	-8.00	5.50	1.30	2.50	-0.10	6.0E-02	-6.2E-04	0.95
A. Supraspinatus	165.0	0.00	-4.00	-2.20	1.20	0.40	0.90	1.3E-01	-1.2E-03	0.96
P. Supraspinatus	173.5	0.30	-5.60	-1.60	1.60	0.50	1.10	-4.1E-02	-3.2E-04	0.98
S. Subscapularis	175.2	-0.10	-12.60	-1.60	-0.10	0.60	-0.10	5.4E-02	-7.1E-04	0.96
M. Subscapularis	199.5	0.00	-6.90	4.00	0.90	1.60	0.50	2.7E-01	-1.2E-03	0.97
I. Subscapularis (L)	214.0	-0.30	-5.30	5.50	0.80	1.30	0.00	9.2E-02	3.2E-04	0.91
I. Subscapularis (S)	191.2	1.20	-7.40	0.10	0.70	4.50	-0.30	-1.1E-01	1.5E-03	0.92
Teres Minor	194.6	-0.10	-9.00	3.20	0.40	2.00	-1.20	1.4E-01	-4.3E-04	0.95
Teres Major	184.3	0.10	-4.00	1.00	0.40	6.10	0.20	5.8E-02	1.0E-03	0.96
A. Deltoid (1)*	103.3	-0.70	-3.00	0.20	0.20	0.10	1.10	1.7E-01	2.1E-03	0.97
A. Deltoid (2-4)*	100.1	-0.60	-2.60	-0.40	0.30	0.10	0.80	2.4E-01	2.0E-03	0.99
M. Deltoid (1-4)	92.1	1.80	-0.80	-2.80	0.90	-0.20	0.80	3.4E-01	1.3E-03	0.99
M. Deltoid (5-7)	116.8	3.90	-2.50	-1.00	0.20	-0.20	0.40	-6.9E-02	3.0E-03	0.89
P. Deltoid	110.3	0.40	0.00	-7.20	0.90	0.90	0.90	2.7E-01	1.2E-03	0.91
Coracobrachialis	88.6	0.00	-0.60	-2.90	0.50	0.00	0.70	4.3E-01	1.4E-03	0.99

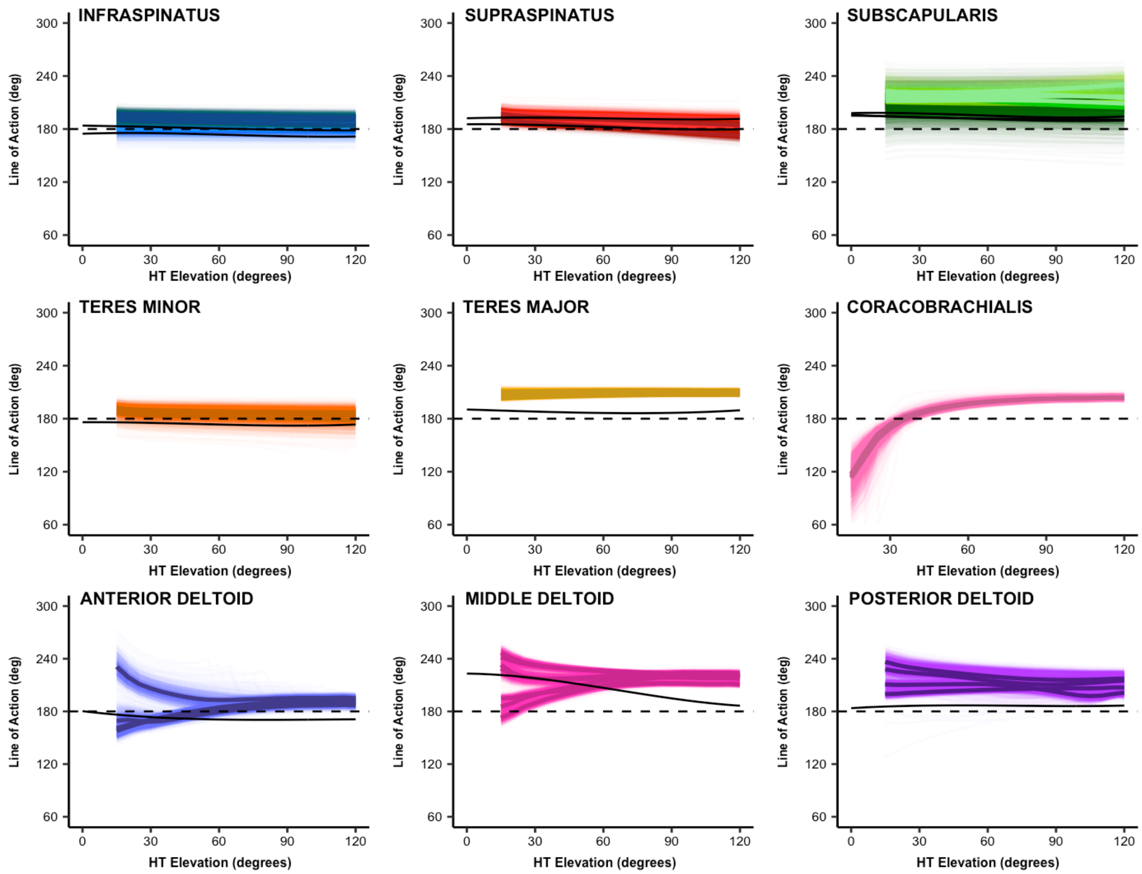

Figure 3.5: Model-predicted glenohumeral anterior-posterior lines of action with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Angles less than (or greater than) 180° represent anteriorly (or posteriorly) directed muscle lines of action. Black lines represent experimental data from Ackland & Pandy, 2009.

Table 3.5: Regression models predicting anterior-posterior lines of action (°) for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) term were added as covariates. Values represent unstandardized beta coefficients. Angles between 0-180° indicate an anteriorly-directed line of action, with values greater than 180° indicating a posteriorly-directed line of action.

Muscle	B_0	S_x or $*C_x$	S_y or $*C_y$	S_z or $*C_z$	H_x	H_y	H_z	HT	HT^2	R^2
S. Infraspinatus	188.8	-3.00	0.30	1.70	1.50	-0.60	-1.60	-5.2E-02	9.0E-05	0.98
I. Infraspinatus	206.3	-7.00	0.80	1.10	1.60	-0.70	-2.00	9.2E-03	-1.5E-04	0.94
A. Supraspinatus	165.0	-5.00	0.00	0.80	1.70	-0.30	-0.80	-1.6E-01	1.7E-04	0.96
P. Supraspinatus	173.5	-4.90	0.30	1.50	1.80	-0.20	-1.40	-8.1E-02	7.1E-05	0.97
S. Subscapularis	175.2	-13.50	-0.20	7.80	0.00	0.10	0.10	-1.5E-04	-1.0E-04	0.93
M. Subscapularis	199.5	-1.80	-0.50	4.20	1.40	-0.50	-0.40	-3.1E-02	-4.4E-04	0.94
I. Subscapularis (L)	214.0	-3.50	-0.10	5.90	1.90	0.00	-0.30	8.4E-02	-7.1E-04	0.94
I. Subscapularis (S)	191.2	-4.50	1.10	5.60	3.20	-0.10	-1.30	-9.0E-02	1.3E-03	0.96
Teres Minor	194.6	-4.00	-2.10	1.60	0.80	-0.70	-5.10	-1.0E-01	3.9E-04	0.90
Teres Major	184.3	-1.80	0.00	2.00	0.60	-0.30	-0.40	6.5E-02	-3.4E-04	0.96
A. Deltoid (1)*	103.3	-1.90	-3.00	2.50	0.40	-1.30	-1.80	-1.1E+00	6.7E-03	0.80
A. Deltoid (2-4)*	100.1	-1.40	0.10	2.50	0.80	-2.10	-0.50	6.7E-01	-3.0E-03	0.91
M. Deltoid (1-4)	92.1	-9.30	0.20	6.40	0.30	0.20	-1.00	-3.6E-01	2.1E-03	0.56
M. Deltoid (5-7)	116.8	-4.90	2.20	0.10	1.50	0.10	-0.60	9.6E-01	-5.0E-03	0.86
P. Deltoid	110.3	-4.40	-3.00	9.10	-0.60	-0.90	-1.10	-2.2E-01	9.0E-04	0.73
Coracobrachialis	88.6	-2.00	0.00	-0.60	2.00	-1.60	-0.40	2.4E+00	-1.4E-02	0.85

3.4.5. Muscle Force Estimation

Muscle forces were largely robust to changes in muscle attachment locations (Table 3.2; Figure 3.6). Variation in muscle forces ranged from 0.8-33.6 N, or 0.2-6.8 % relative to the mean total muscle force. The superior (46.8 N; 8.6%) and inferior (63.0 N; 13.6%) infraspinatus, coracobrachialis (58.7 N; 14.8%), and anterior supraspinatus (14.6 N; 6.8%) displayed greater variation at high elevation angles, peaking at 120° of elevation (± 2 standard deviation expressed in absolute and relative terms at 120° in parentheses). The remaining muscles displayed relatively consistent variation throughout elevation.

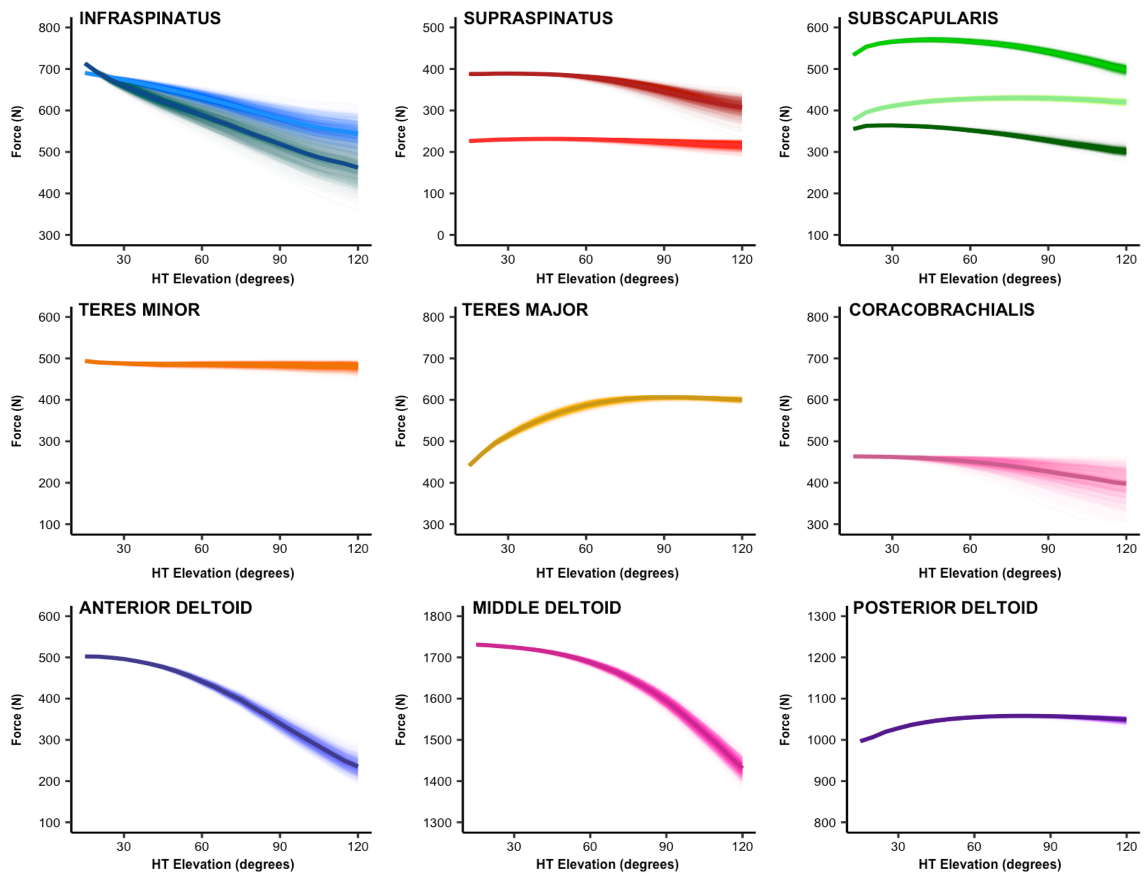

Figure 3.6: Model-predicted muscle forces estimated at 100% activation with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively.

3.5. Discussion

Overall, model-predicted functional roles agreed well with reviewed literature, with some discrepancies that were muscle-specific and more pronounced with lines of action than moment arms. Large variance in muscle functions were identified as a result of perturbations in muscle attachment locations, highlighting the potential for inter-individual musculoskeletal geometry differences among healthy men and women to influence shoulder function.

Defining realistic musculoskeletal geometry within a model is foundational for valid interpretations of muscle function. Model-predicted and reviewed literature moment arms revealed overall agreement, with some exceptions (teres major, posterior deltoid). Greater discrepancies were observed in the muscle lines of action. Simulations under/over predicted lines of action in several cases, being more prevalent for the teres major, supraspinatus and deltoids, especially in the superior-inferior direction. In addition, 'bowstringing' of muscles (Murray et al., 1995; Webb et al., 2014) around the spherical humeral head wrap object was often encountered with increasing elevation. This resulted in discontinuities in the moment arms and lines of action that was especially troublesome for the anterior and posterior deltoids. Traditionally, validation of shoulder model musculoskeletal geometry is conducted solely with experimental moment arms, due in large part to available data in the literature (Garner & Pandy 2001; Holzbaur et al., 2005; Gatti et al., 2007); however, these findings suggest that appropriately defined moment arms do not concurrently guarantee anatomically consistent representation of muscle pathways. As internal tissue loads and joint reaction forces are often the primary outcomes of interest when using models (Erdemir et al., 2007), muscle lines of action can have a substantial impact on modeling results (van der Helm & Veenbaas, 1991).

Variation in musculoskeletal geometry, reflected by muscle attachment changes, had substantial effects on muscle function. The mean (peak) absolute variation, defined as 2 standard deviations per modeling practices by the OpenSim research team (Hicks et al., 2015), was 10 mm (25.8 mm) and 12.0° (30.0°) for elevation/depression moment arm and lines of action, respectively. In general, high sensitivity in rotator cuff functional roles, especially the subscapularis, was observed across moment arms and lines of action. These results are consistent with a recent study employing similar probabilistic changes to attachment sites of the rotator cuff muscles, reporting the greatest sensitivity in predicted subscapularis forces during static internal/external rotation exertions (Chopp-Hurley et al., 2014). Moment arms and lines of action were chosen as the primary outcomes as they geometrically quantify a muscle's functional role as a segment mover and joint stabilizer (An, 2002; Correa et al., 2011), without being affected by assumptions for optimization-derived muscle forces from inverse solutions (Erdemir et al., 2007). It should be noted that the muscle forces in the current study were computed independently for each muscle and were not an inverse solution. To provide context to the moment arm changes, a prior investigation using Monte Carlo simulations that perturbed shoulder muscle moment arms to similar magnitudes as our results (standard deviations ranged from 5.6-13.8 mm), observed coefficient of variations in muscle forces exceeding 200% (Hughes & An, 1997). Similarly, anatomically feasible alterations in muscle lines of action (5-15°), frequently caused moderate to large (>100 N) changes in shear and compressive spinal loads (Nussbaum et al., 1995). These results are consistent with investigations at the lower extremity (Duda et al., 1996; Pal et al., 2007; Scheys et al., 2008; Correa et al., 2011; Carbone et al., 2012; Valente et al., 2014; Bosmans et al., 2015; Navacchia et al., 2016) and shoulder (Bolsterlee & Zadpoor, 2014; Chopp-Hurley et al., 2014) that found high sensitivity of model-predicted outcomes to alterations in musculoskeletal geometry.

Results from this study have important implications for subject-specific modeling and shoulder mechanics at the population level. A primary motivation in conducting this investigation was the high variability in shoulder kinematics and muscular strategies. It was hypothesized that inter-individual differences in musculoskeletal geometry, as observed in bone morphology (Boileau & Walch, 1997; Hertel et al., 2002; Chopp-Hurley et al., 2016b) and moment arms (Figure 3.1), may substantially affect shoulder mechanics. Consistent with the hypothesis, changes in muscle attachment locations caused large variations in muscle function that could theoretically alter muscle coordination patterns and kinematics. Interestingly, the model-predicted variations in glenohumeral elevation/depression moment arm due to attachment changes coincided with the range of magnitudes recorded in the literature for some muscles (infraspinatus, supraspinatus). Model-predicted moment arms appear to be most sensitive to attachment changes closest to the joint centre (i.e. scapular and humeral attachment for the deltoid and rotator cuff, respectively), corresponding with previous observations (Murray et al., 1995, 2002; Pal et al., 2007; Chopp-Hurley et al., 2014). Recently, there has been increasing emphasis on subject-specific models; however, developing these models is a time-consuming process due to the challenges at identifying personalized input parameters. By quantifying the sensitivity of muscle functions to differences in geometry, findings from the current study can inform the level of musculoskeletal geometry individualization needed by subject-specific models (Carbone et al., 2012, 2016). These results also substantiate prior investigations emphasizing incorporating input variability when estimating model-predicted outcomes. Doing so would identify the distribution of possible shoulder kinematic/muscular strategies for a given task at the population level and identify potentially important parameters differentiating injury risk in the workplace (Langenderfer et al., 2006b; Flieg et al., 2008; Chopp-Hurley et al., 2014, 2016c).

A few limitations need to be considered. Univariate normal distributions were used to define the probability distribution functions for muscle attachment locations. Although outcomes distributions can be sensitive to input distributions, especially at tail regions (Hughes & An, 1997; Chopp-Hurley et al., 2014), normal distributions demonstrate good agreement when predicting mean/median and standard deviations compared to other distribution types (e.g. lognormal, gamma) (Flieg et al., 2008; Chopp-Hurley et al., 2014). Multivariate distributions are recommended due to correlations between input parameters (Hughes & An, 1997; Langenderfer et al., 2006b,c). The lack of covariance data across inputs limited use of multivariate distributions, resulting in a larger theoretical set of input parameters. Although perturbations in muscle attachment locations were interpreted to reflect inter-individual anatomical differences, other sources, such as anthropometric scaling and measurement issues with location centroid of muscle origin/insertion sites, can also influence variations in muscle attachment locations (Brand et al., 1982; Duda et al., 1996; Scheys et al., 2008; Bolsterlee & Zadpoor, 2014). Finally, caution must be advised when comparing the model-predicted and experimental data. With the exception of two investigations (Ackland et al., 2008; Ackland & Pandy, 2009), experimental moment arms were collected using a fixed scapula. Model-predictions were based on kinematics previously collected using bone pins that accounted for scapular motion and glenohumeral rotation. Accordingly, the model-predicted versus experimental assessments reflect qualitative comparisons; however, similar results were found using the same kinematics as Ackland and colleagues.

In conclusion, the current study is among the first to quantify sensitivity of model-predicted scapulohumeral muscle functional roles to inter-individual differences in shoulder muscle geometry. Future efforts are directed towards integrating the functional changes across muscles together to assess their impact on muscle coordination needed to perform various tasks. Understanding anatomical factors affecting the mechanical function of muscles may help distinguish inter-individual variation in muscular/kinematic patterns and are a step towards the global aim of understanding the development of workplace shoulder injuries.

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CHAPTER FOUR

Glenohumeral stabilizing roles of the scapulohumeral muscles: Implications of muscle geometry

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4.1. Abstract

Dynamic stability provided by muscles is integral for function and integrity of the glenohumeral joint. Although it is well known that a high degree of inter-individual variation exists in musculoskeletal geometry that is associated with shoulder injuries, there is limited research associating the effects of muscle geometry on the potential stabilizing capacities of muscles. The purpose of this investigation was to evaluate the stabilizing functions of the scapulohumeral muscles using computer modeling and to quantify the sensitivity of muscle stabilizing roles to changes in muscle geometry. Muscle stability ratios in the superior/inferior and anterior/posterior directions were computed as the ratio between the muscle's shear components relative to compression throughout arm elevation in the scapular plane. Muscle attachment locations on the clavicle, scapula, and humerus were iteratively adjusted using Monte Carlo simulations. Consistent with previous experimental studies, the rotator cuff muscles were identified as the primary stabilizers of the glenohumeral joint; whereas the deltoids and coracobrachialis have a strong potential for superiorly translating the humerus at low elevation angles. Variations in the stability ratios due to altered muscle geometry were muscle- and angle-specific. In general, the highest variation was observed for the subscapularis and deltoids (at low elevation angles), while the remaining rotator cuff muscles largely maintained their capacity to provide compressive stabilizing forces at the glenohumeral joint. Changes in muscle stability ratios may affect dynamic stability of the humerus that could differentially predispose individuals to greater risk for injury.

4.2. Introduction

The shoulder complex consists of a large group of muscles that enable movement across a number of degrees of freedom unlike any other part of the body. Enhanced shoulder mobility comes at the expense of instability (Veeger & van der Helm, 2007), which refers to the balance between the glenoid fossa of the scapula and the humeral head at the glenohumeral (GH) joint (Lippitt & Matsen, 1993). Clinically, shoulder instability is defined as a humeral displacement deemed to be too large in response to the resulting force acting on the segment (Veeger & van der Helm, 2007). Although humeral translation is routinely observed in vivo during active shoulder motions (Poppen & Walker, 1976; Graichen et al., 2000; Bey et al., 2008; Chopp et al., 2010), the magnitude is in the range of a couple millimeters within healthy individuals, with larger translations measured among clinical, symptomatic populations (Poppen & Walker, 1976; Howell et al., 1988; Deutsch et al., 1996; Yamaguchi et al., 2000). Abnormal humeral motion due to joint instability can alter muscle moment arms and lines of action, reduce shoulder functionality, and if displacements are large enough, can result in subluxation or dislocation (Michener et al., 2003). In particular, excessive superior humeral translations can cause mechanical compression of the humeral head against surrounding soft tissues (e.g. bursa, supraspinatus and long head biceps brachii tendons), considered to be one possible pathomechanism for the most commonly occurring shoulder injury – subacromial impingement (van der Windt et al., 1996; Michener et al., 2003). A number of structures contribute to GH joint stability, which can be grouped into passive (joint capsule, ligaments, glenoid labrum, intra-articular pressure) and dynamic (muscle) components (Wilk et al., 1997; Veeger & van der Helm, 2007). Although the relative stabilizing contributions from the different structures is debated, it is generally accepted that dynamic stability exerted by muscle forces is the primary mechanism for GH stability at mid range of motion due to limited contributions from passive stabilizers (Veeger & van der Helm, 2007). Accordingly, because muscles can cause translational and rotational accelerations in three dimensions, coordination between muscles requires a delicate balance of forces and moments that can simultaneously elicit movement/strength to perform a task, while maintaining overall integrity of the shoulder complex.

Dynamic stabilizing roles are specific to each muscle and vary largely with upper extremity posture. The rotator cuff muscles (infraspinatus, subscapularis, supraspinatus, teres minor) are considered to be the primary stabilizers of the GH joint. All four rotator cuff muscles resist translations of the humerus in the superior, inferior, anterior, and posterior directions, as simulated experimentally using cadavers (Blasier et al., 1992; Itoi et al., 1994a; Sharkey & Marder, 1995; Deutsch et al., 1996; Karduna et al., 1996; Thompson et al., 1996; Blasier et al., 1997; Soslowsky et al., 1997; Lee et al. 2000; Halder et al., 2001a,b; Mura et al., 2003; Labriola et al., 2005). The strong stabilizing function of the rotator cuff is due to their lines of action, which are directed primarily towards compression with relatively small shear components (Yanagawa et al., 2008; Ackland & Pandy, 2009). As a result, loading across these muscles results in compression between the surfaces of the convex-shaped humeral head and concave-shaped glenoid. The congruent fit between the two articulating surfaces is essential for stability at the GH joint and is the predominant mechanism for dynamic stability, referred to as the concavity compression (Lippitt & Matsen, 1993). As long as the net humeral joint reaction force is directed within the glenoid arc, the area on the glenoid surface in contact with the humeral head, stability is maintained. Other muscles, such as the deltoids, have a greater capacity to exert shear forces on the humerus and can have either a stabilizing or destabilizing effect on the humerus depending on the direction and posture (Poppen & Walker, 1978; Halder et al., 2001a; Lee & An, 2002; Kido et al., 2003).

In general, the stabilizing potential for each muscle at the GH joint can be quantified by a stability ratio, computed as the ratio between the shear vector component (superior-inferior; anterior-posterior) and the compression component (i.e. muscles with larger ratios have greater de-stabilizing role due to a large shear component) (Lippitt & Matsen, 1993; Yanagawa et al., 2008). Quantifying the stability ratios for the shoulder muscles throughout a range of motion can have several important applications: facilitating the interpretation of shoulder neuromuscular strategies using electromyography, determining potential performance and injury consequences of muscle-specific fatigue in the workplace or sports, exploring possibilities for tendon transfer surgeries, and informing rehabilitation programs. To date, a few experimental and modeling studies have computed the muscle stability ratios across functional ranges of motions (Yanagawa et al., 2008; Ackland & Pandy 2009; Ameln et al., 2018). One area that has remained relatively unexplored is inter-individual differences in muscle stability ratios. Several anatomical differences at the shoulder can exist between individuals, as exhibited in the wide population-level distributions of bone geometry parameters for the humerus and scapula (Boileau & Walch, 1997; Hertel et al., 2002;

Chopp-Hurley et al., 2016). Bone geometry differences are thought to substantially alter muscle lines of action (Hughes et al., 2003; Tétreault et al., 2004; Nyffeler et al., 2006), resulting in inter-individual variance in shoulder muscle stabilizing functions and consequently, muscle coordination patterns required for joint stability. Significant correlations between bone morphology and both impingement and prevalence/severity of rotator cuff tears have been found previously (Banas et al., 1995; Hughes et al., 2003; Nyffeler et al., 2006; Balke et al., 2013; Pandey et al., 2016); however, little research has been conducted quantifying the specific alterations in the dynamic stability provided by shoulder muscles to inter-individual muscle geometry differences that may be caused by variations in bone morphology.

The purposes of this study were to: (1) use computational modeling to determine the stabilizing roles of the scapulohumeral muscles throughout humeral elevation in the scapular plane; and (2) quantify the variation in the stabilization roles due to changes in musculoskeletal geometry.

4.3. Methods

The current investigation is an extended analysis from a prior study examining model-predicted muscle moment arms and lines of action, with the methods described in full detail previously (Chapter 3). In brief, the modified Delft Shoulder and Elbow Model (DSEM) (van der Helm, 1994a,b; Blana et al., 2008), as available in OpenSim 3.3 (Stanford, CA, USA), was used to investigate the potential stabilizing roles of 9 scapulohumeral muscles: anterior, middle and posterior deltoids, coracobrachialis, infraspinatus, subscapularis, supraspinatus, teres minor, and teres major (Table 4.1). The modified DSEM is based on the same musculoskeletal geometry data as the original model (van der Helm, 1994a; Nikooyan et al., 2011). Muscle pathways in OpenSim are computed as the shortest distance between origin and insertion attachment locations that may be constrained to wrap around structures representing bone contours (e.g. sphere/cylinder for the humeral head/shaft). Coordinate systems of the modified DSEM were transformed to match the conventions recommended by the International Society of Biomechanics (Wu et al., 2005).

Table 4.1: List of scapulohumeral muscles sub-regions and number of elements representing each muscle included in the model.

Muscle sub-region	Number of Elements
I. Infraspinatus (Inferior)	3
S. Infraspinatus (Superior region)	3
A. Supraspinatus (Anterior region)	2
P. Supraspinatus (Posterior region)	2
S. Subscapularis (Superior region)	3
M. Subscapularis (Middle region)	3
I. Subscapularis (L) (Inferior region – long fibres)	3
I. Subscapularis (S) (Inferior region – short fibres)	2
Teres Minor	3
Teres Major	4
Anterior (A.) Deltoid (1)	1
Anterior (A.) Deltoid (2)	3
Middle (M.) Deltoid (1)	4
Middle (M.) Deltoid (2)	3
Posterior (P.) Deltoid	4
Coracobrachialis	3

The model was postured in a series of static postures from 15-120° of arm elevation in the scapular plane at 5° intervals. Kinematics were based on available data in the literature collected in vivo using bone pins, and accounted for clavicular and scapular rotations in all 3 axes, as well as glenohumeral internal/external rotation (Ludewig et al., 2009). Muscle lines of action acting on the humerus (in the scapula coordinate system) were computed at each posture by computing the unit vector direction cosine of the muscle pathway. For muscles constrained by wrapping objects, the unit vector was computed as the sum of the resultant vectors acting at each path-point at the humerus along the wrapped path. Based on the unit vector direction cosine, stability ratios (Lippitt & Matsen, 1993; Yanagawa et al., 2008) were computed to quantify each muscle's shear contribution in the superior/inferior (f_y) and anterior/posterior (f_x) directions relative to its compression component (f_z):

$$\mathrm{ST_{S-I}} = \frac{\mathrm{f_y}}{|\mathrm{f_z}|} \qquad (1)$$

$$\mathrm{ST_{A-P}} = \frac{\mathrm{f_x}}{|\mathrm{f_z}|} \qquad (2)$$

Where, $ST_{S\text{-}I}$ is the superior/inferior stability ratio, $ST_{A\text{-}P}$ is the anterior/posterior stability ratio, and f_x, f_y, f_z refer to the unit vector direction cosine in each direction. Positive values for $ST_{S\text{-}I}$ and $ST_{A\text{-}P}$ indicate a muscle line of action directed superiorly and anteriorly, with higher values representing greater potential to translate the humerus relative to the scapula towards that direction. It should be emphasized that the stability ratios describe the muscle's shear line of action relative to compression and is computed based on muscle geometry, which is independent of the muscle force magnitude.

Probabilistic modeling was used to quantify the sensitivity of muscle stabilizing roles to alterations in muscle geometry. Univariate normal distributions were generated for each muscle's 3D spatial coordinates (XYZ) defining the attachments site at both the scapula (or clavicle for the anterior deltoid) and humerus. Mean values were based on the DSEM's current attachment sites, with standard deviations computed from literature data linearly scaled to match the model's bone lengths (Högfors et al., 1987). Normal distributions were assumed, as they are considered physiologically relevant and generate comparable summary data to other distributions (Flieg et al., 2008; Chopp-Hurley et al., 2014). Monte Carlo simulations (1000 iterations) were used to randomly sample from the univariate normal distributions and quantify the distribution of model-predicted muscle stability ratios based on perturbations in muscle attachment locations.

As each muscle is comprised of a number of elements to represent the broad-spanning attachments, muscle elements were grouped into sub-regions (Chapter 3) for data reduction purposes during statistical analyses (Table 4.1). Summary data (means ± 2 standard deviations) were computed for each muscle sub-region's stability ratio at 5° increments from 15-120° humeral elevation in the scapular plane, and averaged across the entire range of motion. Multiple regressions were used to quantify the sensitivity of muscle stability ratios to perturbations in muscle attachment locations. The 6 spatial coordinates (XYZ) defining the muscle's attachment at the scapula (or clavicle for the anterior deltoid) and humerus were entered into the regression models, along with the humeral elevation angle (HT) and its quadratic term (HT^2) as covariates. The attachment locations were centered to the mean and divided by the standard deviations (i.e. a beta coefficient of 10 is interpreted as a 1 standard deviation change in the independent variable predicts a 10 unit change in the dependent variable). Model fit was assessed using R^2. Statistical analyses were performed using STATA 14.2 (StataCorp LLC, TX, USA).

4.4. Results

4.4.1. Superior/Inferior Stability Ratios

Model-predicted superior/inferior stability ratios are plotted in Figure 4.1 and summarized in Table 4.2. The rotator cuff muscles and teres major exhibited relatively small stability ratios in the superior/inferior direction, with magnitudes less than 1.0, indicating a larger compression vector component than superior/inferior shear. In general, these muscles were inferior stabilizers at the glenohumeral joint (negative superior/inferior stability ratios), with the exception of the supraspinatus. The superior sub-regions of the infraspinatus and subscapularis also exhibited some capacity to either superiorly or inferiorly stabilize the humerus, depending on the particular muscle element and varying with attachment locations. The stability ratios for the rotator cuff muscles and teres major were relatively consistent with changes in elevation angle, with some divergences at higher arm elevations as modelled by the regressions (Table 4.3). In contrast, the deltoids and coracobrachialis all exhibited large superior stabilizing roles that were especially pronounced at low elevation angles but decreased non-linearly with increasing arm elevation. The stabilizing roles of all the deltoids were highly sensitive to changes in attachment sites at low elevation angles, with standard deviations in $ST_{S\text{-}I}$

ranging from 1.07 to 2.29 across these muscles at 15°. Scapular attachment in the y-axis promoted the greatest changes in the superior-inferior stability ratios, with a 1 standard deviation change at this coordinate predicting a change in stability ratio of at least 0.1 units across all muscles except the posterior deltoid (Table 4.3). In contrast, a 1 standard deviation perturbation at the other coordinates predicted less than a 0.05 unit change in the stability ratio for the majority of cases (73% of the cases).

Table 4.2: Mean with 2 standard deviations in parentheses for model-predicted muscle stability ratios averaged across scapular plane elevation and grouped by scapulohumeral muscle sub-region due to changes in muscle attachment location. Stability ratios are in the superior/inferior direction (ST_{S-I}) and anterior/posterior directions (ST_{A-P}). Positive values for ST_{S-I} (ST_{A-P}) indicate the muscle would impose a superior (anterior) shear force at the humerus.

Muscle sub-region	ST_{S-I}	ST_{A-P}
S. Infraspinatus	-0.11 (0.45)	-0.20 (0.16)
I. Infraspinatus	-0.52 (0.25)	-0.12 (0.24)
A. Supraspinatus	0.23 (0.18)	-0.10 (0.18)
P. Supraspinatus	0.20 (0.23)	-0.18 (0.21)
S. Subscapularis	0.10 (0.49)	-0.58 (0.78)
M. Subscapularis	-0.62 (0.31)	-0.62 (0.19)
I. Subscapularis (L)	-0.91 (0.24)	-0.72 (0.29)
I. Subscapularis (S)	-0.22 (0.41)	-1.00 (0.44)
Teres Minor	-0.40 (0.32)	-0.10 (0.23)
Teres Major	-0.26 (0.25)	-0.56 (0.12)
A. Deltoid (1)	1.69 (0.66)	-0.38 (0.32)
A. Deltoid (2-4)	1.68 (0.35)	-0.02 (0.13)
M. Deltoid (1-4)	2.44 (0.76)	-1.00 (0.31)
M. Deltoid (5-7)	1.41 (0.56)	-0.57 (0.21)
P. Deltoid	1.21 (0.77)	-0.67 (0.52)

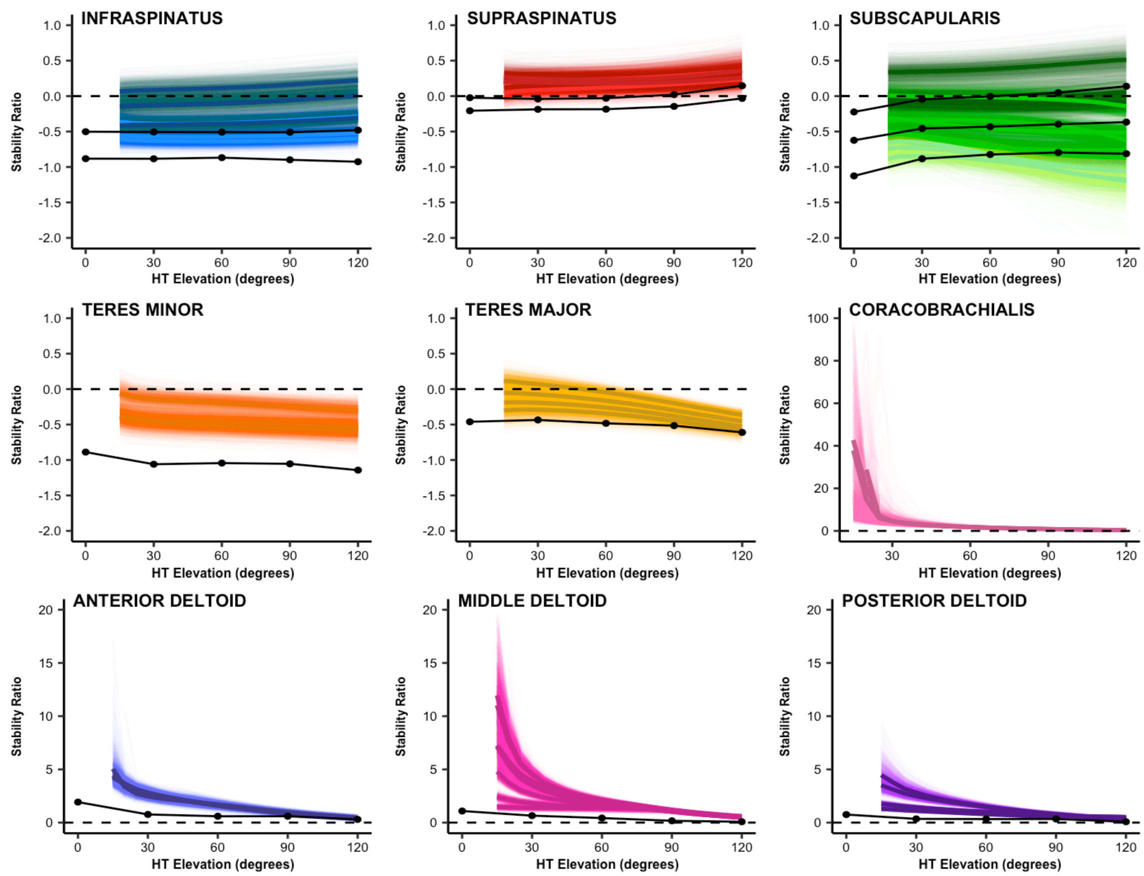

Figure 4.1: Model-predicted muscle superior/inferior stability ratios (ST_{S-I}) acting at the humerus with varying attachment locations. All 1000 simulations are plotted as thin lines. Thick, dark bands represent mean values for each muscle element. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Black lines represent cadaveric experimental data reported by reported by Ackland & Pandy (2009). Positive (negative) values indicate a muscle line of action exhibiting a superior (inferior) shear component. A value of 1.0 (unity) reflects a muscle with an equal vector component in the superior-inferior direction as compression.

Table 4.3: Regression models predicting muscle superior/inferior stability ratios (ST_{S-I}) at the glenohumeral joint for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) terms were added as covariates. Values represent unstandardized beta coefficients. Positive superior/inferior stability ratio indicates the muscle would impose a superior shear force at the humerus.

Muscle	B_0	S_x or *C_x	S_y or *C_y	S_z or *C_z	H_x	H_y	H_z	HT	HT^2	R^2
S. Infraspinatus	-0.16	-0.01	0.21	-0.01	-0.02	-0.03	-0.01	9.33E-05	8.50E-06	0.99
I. Infraspinatus	-0.50	0.01	0.18	-0.12	-0.03	-0.05	0.00	-1.30E-03	1.32E-05	0.96
A. Supraspinatus	0.27	0.00	0.07	0.04	-0.02	-0.01	-0.02	-2.47E-03	2.31E-05	0.95
P. Supraspinatus	0.12	0.00	0.10	0.03	-0.03	-0.01	-0.02	6.76E-04	6.55E-06	0.97
S. Subscapularis	0.09	0.00	0.23	0.03	0.00	-0.01	0.00	-9.96E-04	1.34E-05	0.96
M. Subscapularis	-0.34	0.00	0.16	-0.10	-0.02	-0.04	-0.01	-6.23E-03	2.60E-05	0.94
I. Subscapularis (L)	-0.69	0.01	0.17	-0.19	-0.03	-0.04	0.00	-1.80E-03	-1.91E-05	0.89
I. Subscapularis (S)	-0.21	-0.02	0.14	0.00	-0.01	-0.09	0.01	2.37E-03	-3.05E-05	0.92
Teres Minor	-0.27	0.00	0.18	-0.07	-0.01	-0.04	0.02	-2.65E-03	7.99E-06	0.96
Teres Major	-0.08	0.00	0.08	-0.02	-0.01	-0.11	0.00	-6.86E-04	-2.25E-05	0.97
A. Deltoid (1)*	4.84	0.02	0.11	-0.03	0.00	-0.04	-0.13	-7.30E-02	3.20E-04	0.68
A. Deltoid (2)*	4.94	0.06	0.15	0.05	-0.02	-0.03	-0.07	-7.44E-02	3.15E-04	0.97
M. Deltoid (1)	9.92	-0.35	0.07	0.55	-0.13	-0.10	-0.17	-1.95E-01	1.02E-03	0.83
M. Deltoid (2)	2.90	-0.32	0.16	0.03	-0.01	-0.01	-0.02	-2.83E-02	7.56E-05	0.70
P. Deltoid	3.18	-0.01	-0.04	0.44	-0.04	-0.06	-0.06	-4.31E-02	1.69E-04	0.78

4.4.2. Anterior/Posterior Stability Ratios

Model-predicted anterior/posterior stability ratios are plotted in Figure 4.2 and summarized in Table 4.2. With the exception of the coracobrachialis, anterior deltoid (at low elevation), and supraspinatus (at high elevation), the remaining muscles were predominantly posterior stabilizers of the humerus (negative anterior/posterior stability ratios). The infraspinatus, supraspinatus, and teres minor had small stability ratios, with magnitudes close to 0 (mean ratios ranging from -0.10 to -0.20), indicating a small anterior/posterior shear component relative to compression. In contrast, the teres major and subscapularis displayed greater posterior stabilizing role (mean ratios ranging from -0.56 to -1.00), with the subscapularis in particular displaying a wide range of values across its sub-regions and exhibiting large sensitivity due to alterations in muscle attachment. Similarly, the anterior/posterior stabilizing roles of the deltoids were highly sensitive to changes in muscle attachment at low elevation angles. The stability ratios across the muscles were most sensitive to changes to the scapular attachment along the x- and z-axes (i.e. a 1 standard deviation in either of these axes were predicted to change the stability ratio by at least 0.1 units in 70% of the muscles) (Table 4.4). In contrast, the muscle anterior/posterior stability ratios were fairly robust to changes at the other attachment sites.

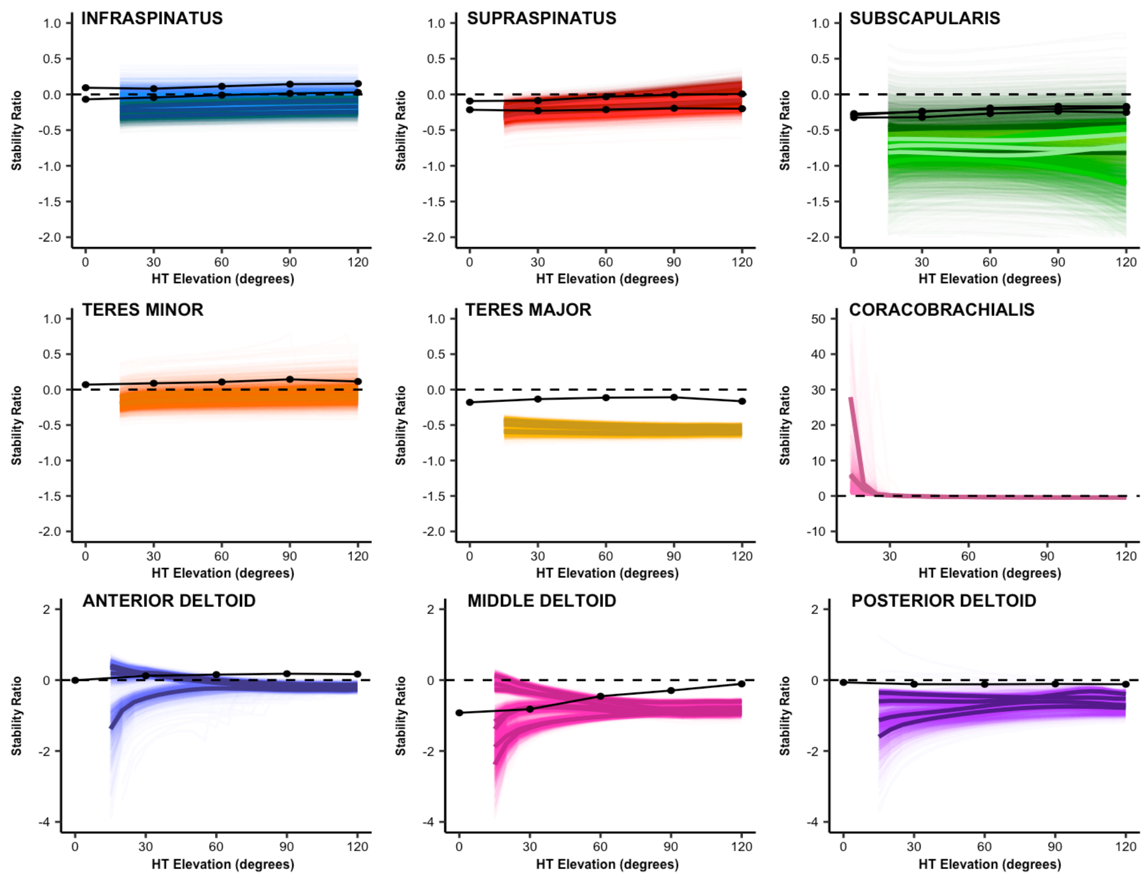

Figure 4.2: Model-predicted muscle anterior/posterior stability ratios (ST_{A-P}) acting at the humerus with varying attachment locations. All 1000 simulations are plotted as thin lines. Thick, dark bands represent mean values for each muscle element. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Black lines represent cadaveric experimental data reported by reported by Ackland & Pandy (2009). Positive (negative) values indicate a muscle line of action exhibiting an anterior (posterior) shear component. A value of 1.0 (unity) reflects a muscle with an equal vector component in the anterior-posterior direction as compression.

Table 4.4: Regression models predicting muscle anterior/posterior stability ratios (ST_{A-P}) at the glenohumeral joint for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) terms were added as covariates. Values represent unstandardized beta coefficients. Positive anterior/posterior stability ratio indicates the muscle would impose an anterior shear force at the humerus.

Muscle	B_0	S_x or $*C_x$	S_y or $*C_y$	S_z or $*C_z$	H_x	H_y	H_z	HT	HT^2	R^2
S. Infraspinatus	-0.26	0.05	-0.01	-0.03	-0.03	0.01	0.03	9.57E-04	-1.74E-06	0.97
I. Infraspinatus	-0.13	0.12	-0.01	-0.02	-0.03	0.01	0.04	-1.67E-04	2.69E-06	0.94
A. Supraspinatus	-0.28	0.09	0.00	-0.01	-0.03	0.01	0.01	2.94E-03	-3.57E-06	0.96
P. Supraspinatus	-0.28	0.09	-0.01	-0.03	-0.03	0.00	0.03	1.53E-03	-1.64E-06	0.97
S. Subscapularis	-0.58	0.34	0.01	-0.22	0.00	0.00	0.00	6.08E-05	1.04E-06	0.90
M. Subscapularis	-0.73	0.04	0.01	-0.10	-0.04	0.01	0.01	9.50E-04	9.14E-06	0.91
I. Subscapularis (L)	-0.67	0.10	0.01	-0.16	-0.05	0.00	0.01	-2.32E-03	1.91E-05	0.90
I. Subscapularis (S)	-0.99	0.15	-0.04	-0.20	-0.12	0.00	0.05	4.29E-03	-5.26E-05	0.92
Teres Minor	-0.18	0.07	0.04	-0.03	-0.01	0.01	0.09	1.85E-03	-6.95E-06	0.89
Teres Major	-0.51	0.04	0.00	-0.05	-0.01	0.01	0.01	-1.43E-03	7.46E-06	0.96
A. Deltoid (1)*	-1.37	0.05	0.08	-0.06	-0.01	0.03	0.05	2.96E-02	-1.80E-04	0.48
A. Deltoid (2)*	0.49	0.03	0.00	-0.05	-0.01	0.04	0.01	-1.22E-02	5.52E-05	0.90
M. Deltoid (1)	-1.63	0.33	0.00	-0.23	-0.01	0.00	0.05	1.83E-02	-1.08E-04	0.53
M. Deltoid (2)	0.21	0.12	-0.05	-0.01	-0.03	0.00	0.02	-1.96E-02	9.86E-05	0.87
P. Deltoid	-1.02	0.12	0.09	-0.25	0.01	0.02	0.03	8.47E-03	-4.02E-05	0.71

4.5. Discussion

The current investigation aimed at determining the stabilizing roles of the scapulohumeral muscles, and quantifying the sensitivity of muscle stability ratios to changes in muscle attachment locations to reflect inter-individual musculoskeletal geometry differences among healthy men and women. The model predicted the rotator cuff muscles to act as the primary stabilizers of the GH joint, while the deltoids and coracobrachialis exhibited substantial superior shear components at low elevation angles. All muscles were predicted to have posterior stabilizing effects. Overall, perturbations in muscle attachment sites predicted small to large effects on muscle stability ratios that were dependent on the muscle of interest and elevation angle.

The dynamic stabilizing function of muscles is critical for maintaining stability at the shoulder complex. Model-predicted stability ratios for the rotator cuff muscles were relatively small in the superior-inferior and anterior-posterior directions, reinforcing previous literature regarding their dominant stabilizing functions due to compression at the GH joint (Poppen & Walker, 1978; Blasier et al., 1992; Itoi et al., 1994a; Sharkey & Marder, 1995; Deutsch et al., 1996; Karduna et al., 1996; Thompson et al., 1996; Blasier et al., 1997; Soslowsky et al., 1997; Lee et a. 2000; Halder et al., 2001a,b; Mura et al., 2003; Labriola et al., 2005). Although the supraspinatus was predicted to have a superior stabilizing effect on the humerus (i.e. positive $ST_{S\text{-}I}$) that was consistent with previous modeling studies (Yanagawa et al., 2008; Ameln et al. 2018), it contradicted an inferior line of action measured in cadavers (Poppen & Walker, 1978; Graichen et al., 2001; Ackland & Pandy, 2009). These differences are small in magnitude, with all studies reporting the supraspinatus to exhibit a predominantly horizontally directed line of action (i.e. superior/inferior stability ratio close to 0; methodological differences may also account for inconsistencies across studies). A significant contribution of these findings that extends the cadaveric work by Ackland & Pandy (2009) is analyzing the muscles as sub-regions to capture the entire breadth of the muscles. Previous studies have commonly grouped muscles into one line of action through the centroid of the muscle. However, shoulder muscles are broad spanning tissues that can have multiple innervation branches and functional sub-regions (McCann et al., 1994; Ward et al. 2006; Kim et al. 2017), hence, a single line of action insufficiently represents these muscles (van der Helm & Veenbaas, 1991; Ackland & Pandy, 2009; Webb et al. 2014). In contrast to the stabilizing function of the rotator cuff muscles and consistent with the literature, the deltoids and coracobrachialis have a large superiorly directed shear component (Poppen & Walker, 1978; Halder et al., 2001a; Lee & An, 2002; Kido et al., 2003). The superior pull is especially pronounced at low elevation angles, which would corroborate with observations of superior translation of the humeral head during the early phase of elevation motions (Poppen & Walker, 1976; Graichen et al., 2000; Bey et al., 2008; Chopp et al., 2010). To visualize the stabilizing functions across the scapulohumeral muscles, the muscle stability ratios in both shear directions (superior/inferior; anterior/posterior) are plotted at 4 discrete elevation angles in Figure 4.3. Included in this figure are stability ratio thresholds determined by Lippitt & Matsen (1993), which can be used to constraint joint reaction force solutions in biomechanical models (Dickerson et al., 2007). Thresholds are based on the magnitude of net humeral joint reaction forces directed tangentially that overcome the concavity compression mechanism for stability and dislocate the humeral head in each direction (thresholds calculated with a 100 N compression force at the GH joint).

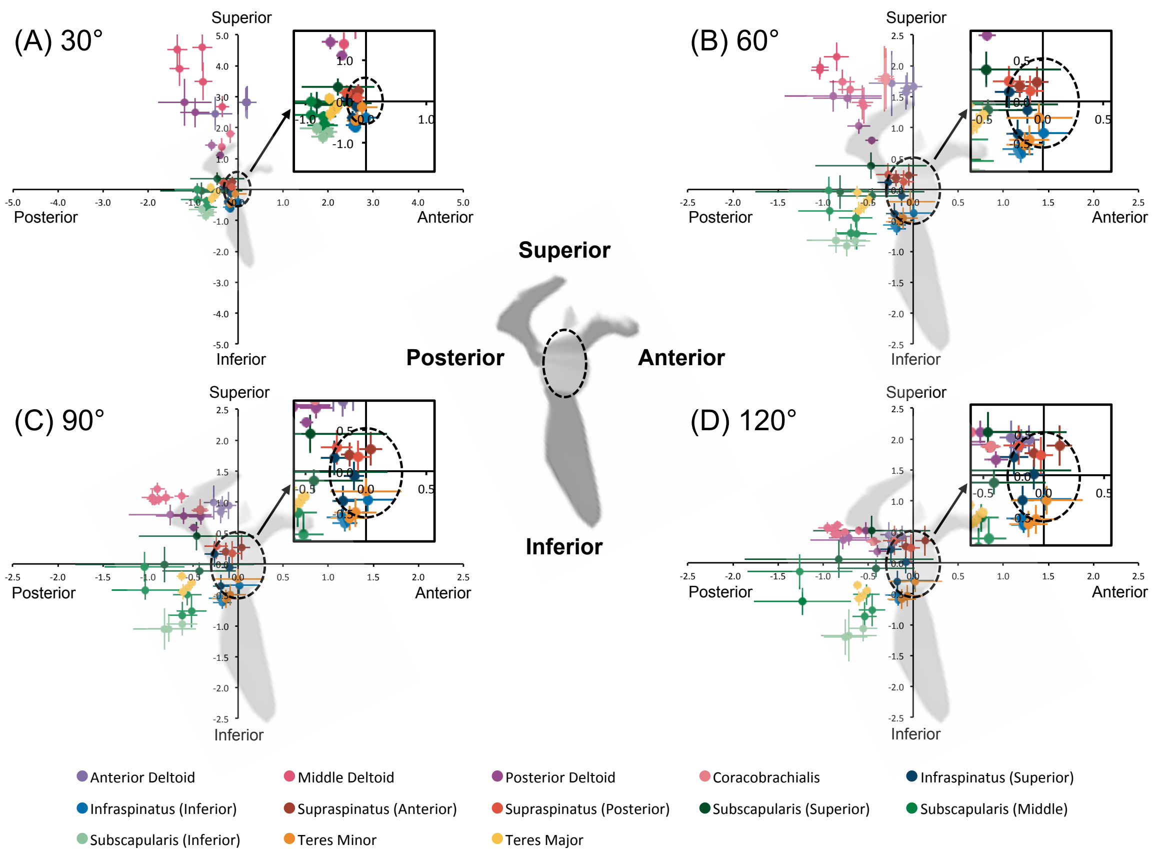

Figure 4.3: Scatterplot between model-predicted muscle anterior/posterior (ST_{A-P}) and superior/inferior (ST_{S-I}) stability ratios for each scapulohumeral muscle at four humeral elevation angles: (A) 30°, (B) 60°, (C) 90°, and (D) 120°. Symbols represent means with 2 standard deviation error bars in either direction. The dotted black asymmetrical ellipse represents experimental thresholds collected by Lippitt & Matsen (1993) (see text for details). Note: the axes for subplot (A) are extended and do not show data for the coracobrachialis as it extended outside the range.

Changes in muscle attachment sites alter the lines of action and have varying effects on the stabilizing function of muscles. For some groups of muscles, such as the supraspinatus, teres minor, and infraspinatus, changes in muscle attachment sites maintained the capacity to stabilize the GH joint, as visualized by stability ratios remaining within the threshold ellipse. In contrast, the subscapularis and deltoids (at low elevation angles) are observed to have large deviations in stability ratios that can alter their capacity to stabilize/de-stabilize the humerus at the GH joint. It is important to note that the net humeral joint reaction force combines components beyond muscle forces, including gravity, externally applied forces, and other tissue forces (e.g. ligaments). Overall, the net effect across these forces must be directed within a small area on the glenoid face for the GH joint to be stable (Lippitt & Matsen, 1993). As muscles are an essential component of the net joint reaction force that are required for dynamic stability, muscle coordination patterns need to concurrently optimize stability and movement. It is widely acknowledged that inter-individual anatomical differences can affect the functional roles and stabilizing effects of shoulder muscles but has not been explicitly investigated (Veeger & van der Helm, 2007; Yanagawa et al., 2008). Variations in bone geometry are thought to alter muscle lines of action (Hughes et al., 2003; Tétreault et al., 2004; Nyffeler et al., 2006), and can significantly affect GH joint stability and risk of impingement due to superior translation of the humeral head (Flieg et al., 2008; Moor et al., 2016). As observed here, inter-individual differences in muscle geometry affect muscle stabilizing roles, which would theoretically require varying muscle coordination patterns needed to maintain shoulder stability across people. In fact, injured individuals often display alterations in muscle activity patterns during upper extremity tasks (Ludewig & Cook, 2000; Phadke et al., 2009). Whether these altered muscle activity patterns are a cause or consequence of injury is unknown. Nevertheless, understanding individual differences in musculoskeletal geometry and its effect on shoulder function can provide several insights into muscle activity patterns that can help evaluate how individuals optimize between shoulder movement and stability, and consequently who may be at greater risk for injury due to reduced glenohumeral stability. In addition, knowing the association between individual anatomy and function can better inform clinical decisions at the patient level (Veeger & van der Helm, 2007), such as guiding tendon transfer surgeries specific to the individual and developing neuromuscular rehabilitation protocols aimed at reducing risk for future injury.

A few limitations need to be considered when interpreting the results. Only the scapulohumeral muscles were analyzed in the current study, and other muscles acting on the humerus were not included. The biceps and triceps brachii, pectoralis major, and latissimus dorsi have attachment sites on the humerus, with previous investigations identifying these muscles to have substantial effects on GH joint stability (Kumar et al., 1989; Itoi et al., 1993, 1994a,b; Rodosky et al., 1994; Blasier et al., 1997; Halder et al., 2001a,b; McMahon & Lee, 2002; Labriola et al., 2005). Stability ratios (Ackland & Pandy, 2009; Ameln et al., 2018) and sensitivity of model-predicted outcomes to alterations in input parameters are task-specific (Scovil & Ronsky, 2006; Ackland et al., 2012), thus the results derived here are primarily applicable for arm elevation in the scapular plane. It should be highlighted that the model kinematics were different (scapular motion in all 3 axes; glenohumeral internal/external rotation included) than the methods used in the comparisons made to available experimental data in cadavers (Ackland & Pandy, 2009). Thus, comparisons to experimental data should be done qualitatively; however, it should be noted that similar results were found when the model was prescribed the same kinematics as used in the aforementioned experimental study.

In conclusion, this study determined the stability ratios of the scapulohumeral muscles and found varying effects of muscle geometry on these stabilizing functions. It is expected that inter-individual differences in anatomy would require altered muscular coordination. Future efforts should be aimed towards associating musculoskeletal geometry differences with in vivo muscle activity patterns throughout the shoulder range of motion, while accounting for the influence of altered scapular kinematics, in order to develop an integrated anatomical- and mechanics-based model determining the pathomechanism and risk for shoulder injuries.

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CHAPTER FIVE: DISCUSSION

The shoulder complex is a remarkable structure, allowing movement unparalleled by any other region in the human body. A diverse group of shoulder muscles with distinct functional roles enables the enhanced mobility. Although shoulder muscles actions are fundamental for upper extremity movement and force production, they are equally important for providing dynamic stability, especially at the glenohumeral joint. As a result, muscle coordination at the shoulder represents a delicate balance between movement, force production and stability, such that an individual can have the strength perform a task while maintaining integrity and stability of the complex. The global aims of this thesis were to determine model-predicted functional roles (moment arms, lines of action, muscle forces, stability ratios) for the scapulohumeral muscles, compare the model predictions to reviewed literature, and quantify the sensitivity of muscle functional roles to inter-individual muscle geometry differences.

The current study was largely motivated by challenges encountered with recent research conducted at the McMaster Occupational Biomechanics Laboratory primarily lead by Dr. Alison C. McDonald under the supervision of Dr. Peter J. Keir. A major research focus over the past 5 years has been evaluating upper extremity muscular and kinematic adaptations to repetitive, fatiguing work such that we can better understand the development of work-related shoulder musculoskeletal disorders. However, the consistent theme across a series of studies was the substantial levels of between- and within-subject variation in fatigue-induced compensations to repetitive work (Tse et al., 2016; McDonald et al., 2016, 2018a,b; McDonald, 2017; Mulla et al., 2018). It was consequently challenging to arrive at strong conclusions that could differentiate specific strategies that may be more inducive for fatigue and injury development. One of the factors that could explain this variability is inter-individual anatomical differences, particularly in musculoskeletal geometry. However, little research has been conducted to quantify the influence of musculoskeletal geometry on shoulder function. Enhances in computational modeling have allowed some researchers to more robustly investigate the potential for musculoskeletal geometry differences to affect shoulder function (Hughes & An, 1997; Flieg et al., 2008; Chopp-Hurley et al., 2014, 2016b; Bolsterlee & Zadpoor, 2014). Nevertheless, these studies have not explicitly quantified the extent to which inter-individual anatomical differences can affect muscle functional roles, as fundamentally determined by moment arms, lines of action, and stability ratios. In addition, whether these models accurately capture muscle functions and are consistent with the cadaveric literature has not been robustly verified.

To verify whether interpretations of model-predicted muscle functions are valid, qualitative comparisons between the model data and reviewed literature were made. Some inconsistencies were noticed, with more differences observed for muscle lines of action than moment arms. The shoulder complex consists of a large group of muscles of varying sizes, shapes, and functional sub-regions that are intricately arranged spatially (superficial to deep) and exhibit substantial physical interactions between each other and around bony contours (Ward et al., 2006; Webb et al., 2014). For example, the deltoids were predicted to have the most inconsistent lines of action when compared to the reviewed literature, which is thought to be due to the simplified spatial arrangement of these muscles in the model and in the literature. The deltoids are superficial muscles that likely have significant physical interactions with the surrounding muscles that can affect their pathways, but are not incorporated into the model. Accordingly, model-predicted lines of action for the deltoids are directed straight from origin to insertion while only being constrained by wrapping objects representing bone. The interactions between elements of the same muscle are also not considered. As a result, muscle elements behave independently, which can cause some groups of fascicles within a muscle to have diverging pathways that are non-physiological and violate the overall volume of the muscle. This was particularly observed for the anterior and posterior deltoids in Chapter 3, where certain elements would experience 'bowstringing' patterns around the humeral head wrapping object, resulting in diverging element pathways within the same muscle sub-region (Figures 3.3 and 3.4). Overall, the simplifications made to muscle pathways by representing them using single lines of action are insufficient compared to models with greater muscle divisions (van der Helm & Veenbaas, 1991; Cleather & Bull, 2010) and finite element models using 3D fibre mapping (Blemker & Delp, 2006; Webb et al., 2014). These complex 3D finite element models are, however, computationally burdensome. Recently, a mesh model offered some promise by constraining muscle elements such that it respected the changes in the overall volume of the muscle and did not exhibit diverging muscle pathways (Hoffmann et al., 2017). Based on the findings from this thesis in conjunction with the literature, it is clear that musculoskeletal models should include several elements to represent each muscle that need to consider the physical interactions between and within muscles. In the process, it is important for both muscle moment arms and lines of actions to be considered and validated throughout a range of postures.

The current work found the sensitivity of functional roles to muscle geometry varied across muscles and elevation angle. It is important to note that the muscle attachment changes used in this thesis were based on anatomical datasets rather than perturbations to input parameters based on a user-defined percent (i.e. 50% of the input value), as often done in previous probabilistic modeling approaches. Although not modeled, it is expected that these changes in muscle function would consequently require altered muscle coordination patterns for shoulder movement, force production, and maintaining stability. As such, it is hypothesized that the inter-individual anatomical differences may partially explain the high degree of variability in the laboratory studies investigating shoulder mechanics. This study provides a framework for future studies to further investigate the associations between musculoskeletal geometry on shoulder mechanics. By quantifying the mechanical function of muscles using moment arms, lines of action, and stability ratios, data from this work can help facilitate interpretations of upper extremity muscle activity strategies that individuals use when performing simulated workplace tasks in the laboratory.

Bone morphology is found to correlate with severity and prevalence of rotator cuff tears (Banas et al., 1995; Nyffeler et al., 2006; Balke et al., 2013; Pandey et al., 2016). The proposed pathomechanisms involves enclosing of the subacromial space and altered muscle lines of action that reduce glenohumeral joint stability (Michener et al., 2003; Hughes et al., 2003; Nyffeler et al., 2006). An immediate next step of this thesis is to use this probabilistic modeling approach to associate specific muscle attachment changes with individual bone morphology to better understand the interaction between musculoskeletal geometry and shoulder musculoskeletal disorders. For example, an increase in glenoid inclination angle by 10° could lead to a predictable amount of change in a particular muscle's scapular attachment location. The regression equations developed and reported in this thesis can then be used to quantify the effects of the individual muscle attachment location on moment arms and dynamic stability at the GH joint. An intermediate research direction is to quantify the theoretical solution set of possible muscle strategies that can be used to perform a task while maintaining joint stability, and estimate the potential effects of fatigue or injury on the balance between movement, force production, and stability. Long-term, these research goals aim to help develop predictive models to estimate muscle coordination and kinematic strategies based on musculoskeletal geometry. The ergonomic application of these models is to aid in the design of workplace tasks that accommodate for individual-specific variability in shoulder mechanics and account for effects of muscle fatigue.

Lastly, scapular kinematics are an area of research that has traditionally received limited attention, but are of growing importance due to the link between abnormal scapular kinematics (i.e. scapular dyskinesis) and shoulder injuries (Paine & Voight, 1993; Kibler, 1998; Ludewig & Reynolds, 2009; Kibler et al., 2013). The vast majority of the shoulder muscles have scapular attachments, with their functions presumably altered by variations in scapular orientation (Bagg & Forrest, 1988). This is evidenced by changes in scapular orientation affecting shoulder strength and muscle activity patterns

(Smith et al., 2002, 2006; Kibler et al., 2006; Picco et al., 2010). Accordingly, axioscapular muscles controlling the scapula can indirectly affect movement and stability of the humerus. Further research is required to quantify the role of axioscapular muscles on scapular movement, determining changes in scapulohumeral muscle moment arms and lines of action with scapular orientation, and the ensuing consequences to humeral movement and stability. A comprehensive, integrated understanding of the roles of all shoulder muscle to movement and stability of the shoulder complex, including altered scapular kinematics and inter-individual differences in musculoskeletal geometry, would enable us to better identify individuals at risk for work-related MSDs based on their muscle and movement coordination.

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APPENDICES

Appendix A: Horizontal Adduction/Abduction and Internal/External Rotation Moment Arm Results

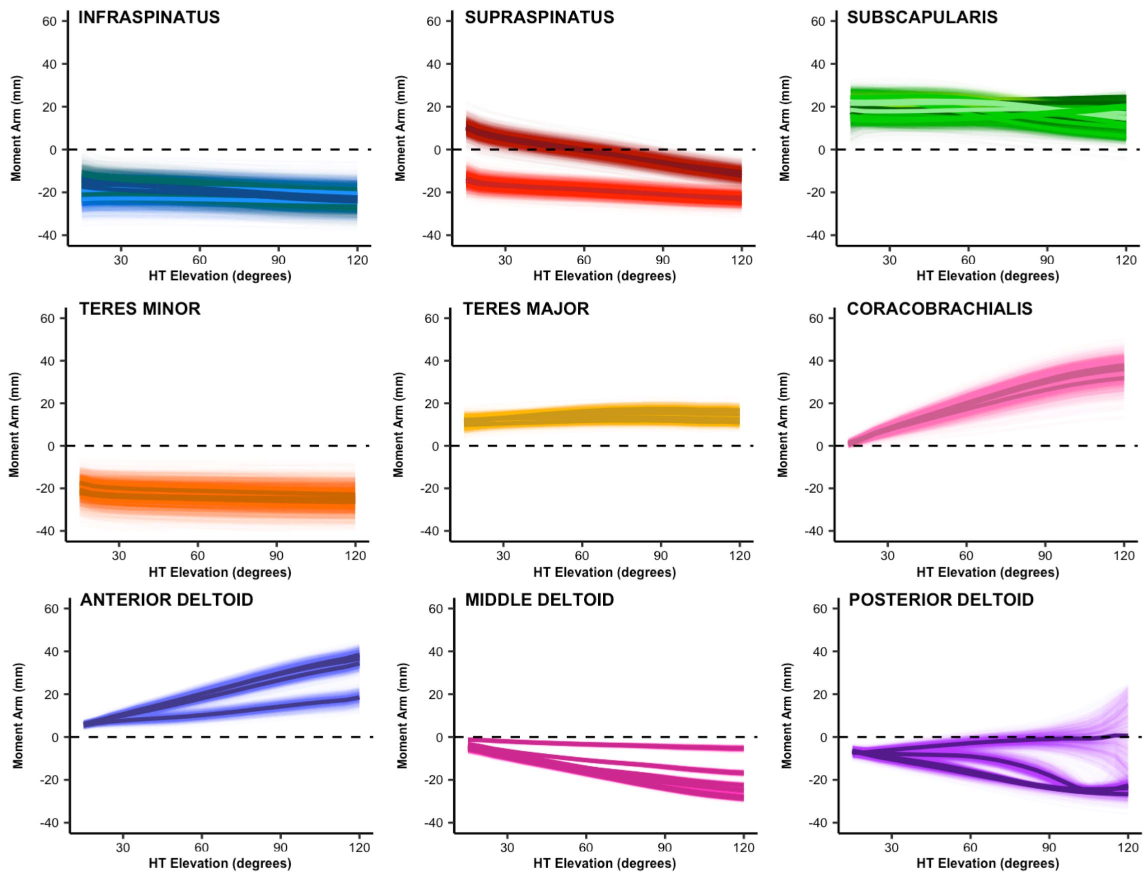

Figure A.1: Model-predicted humeral horizontal adduction/abduction moment arms with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Positive (negative) values indicate humeral horizontal adduction (horizontal abduction).

Table A.1: Regression models predicting humeral horizontal adduction/abduction moment arms (mm) for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) and cubic (HT^3) terms were added as covariates. Values represent unstandardized beta coefficients.

Muscle	B_0	S_x or *C_x	S_y or *C_y	S_z or *C_z	H_x	H_y	H_z	HT	HT^2	HT^3	R^2
S. Infraspinatus	-14.9	-0.2	-0.1	0.1	2.6	-0.9	-2.8	-6.0E-02	-4.0E-04	2.8E-06	0.94
I. Infraspinatus	-21.3	-0.3	-1.0	0.8	2.5	-0.6	-2.9	8.4E-02	-1.3E-03	4.6E-06	0.98
A. Supraspinatus	14.3	0.2	0.0	-0.1	2.3	-0.4	-1.1	-3.5E-01	2.2E-03	-9.4E-06	0.98
P. Supraspinatus	-12.5	0.5	0.7	0.0	1.9	-0.2	-1.6	-1.7E-01	1.4E-03	-5.9E-06	0.97
S. Subscapularis	18.9	-2.2	-0.1	2.6	0.0	0.0	0.0	6.7E-02	-1.4E-03	6.4E-06	0.67
M. Subscapularis	23.0	0.0	0.1	-0.1	2.1	-0.9	-0.7	1.4E-01	-4.5E-03	2.0E-05	0.95
I. Subscapularis (L)	19.9	-0.2	1.2	-0.8	2.1	-0.5	-0.5	3.5E-02	-8.1E-05	-4.6E-06	0.83
I. Subscapularis (S)	15.9	0.9	0.9	-1.7	2.4	-0.4	-0.9	-1.1E-01	1.7E-03	-4.5E-06	0.93
Teres Minor	-18.4	0.5	-2.7	0.9	0.7	-0.1	-4.5	-1.7E-01	1.9E-03	-7.6E-06	0.98
Teres Major	9.3	1.4	0.1	-1.4	0.9	-0.8	-0.9	1.3E-01	-8.5E-04	6.8E-07	0.88
A. Deltoid (1)*	4.5	0.0	-0.7	-1.2	0.0	0.1	0.3	9.3E-02	1.7E-05	1.5E-06	0.96
A. Deltoid (2-4)*	2.1	0.3	-0.8	-1.6	0.2	-0.1	0.4	2.4E-01	1.3E-03	-7.4E-06	0.99
M. Deltoid (1-4)	-0.7	1.9	0.3	-0.8	-0.3	0.0	-0.4	-2.8E-01	3.0E-04	2.1E-06	0.99
M. Deltoid (5-7)	0.2	4.6	0.2	-0.5	-0.2	0.0	0.0	-2.4E-01	1.6E-03	-5.4E-06	0.92
P. Deltoid	-8.8	1.5	-0.8	-5.0	-0.4	-1.0	-0.9	1.4E-01	-4.4E-03	2.1E-05	0.74
Coracobrachialis	-4.8	1.4	-0.3	-3.2	0.3	-0.1	0.3	4.0E-01	3.8E-04	-7.9E-06	0.98

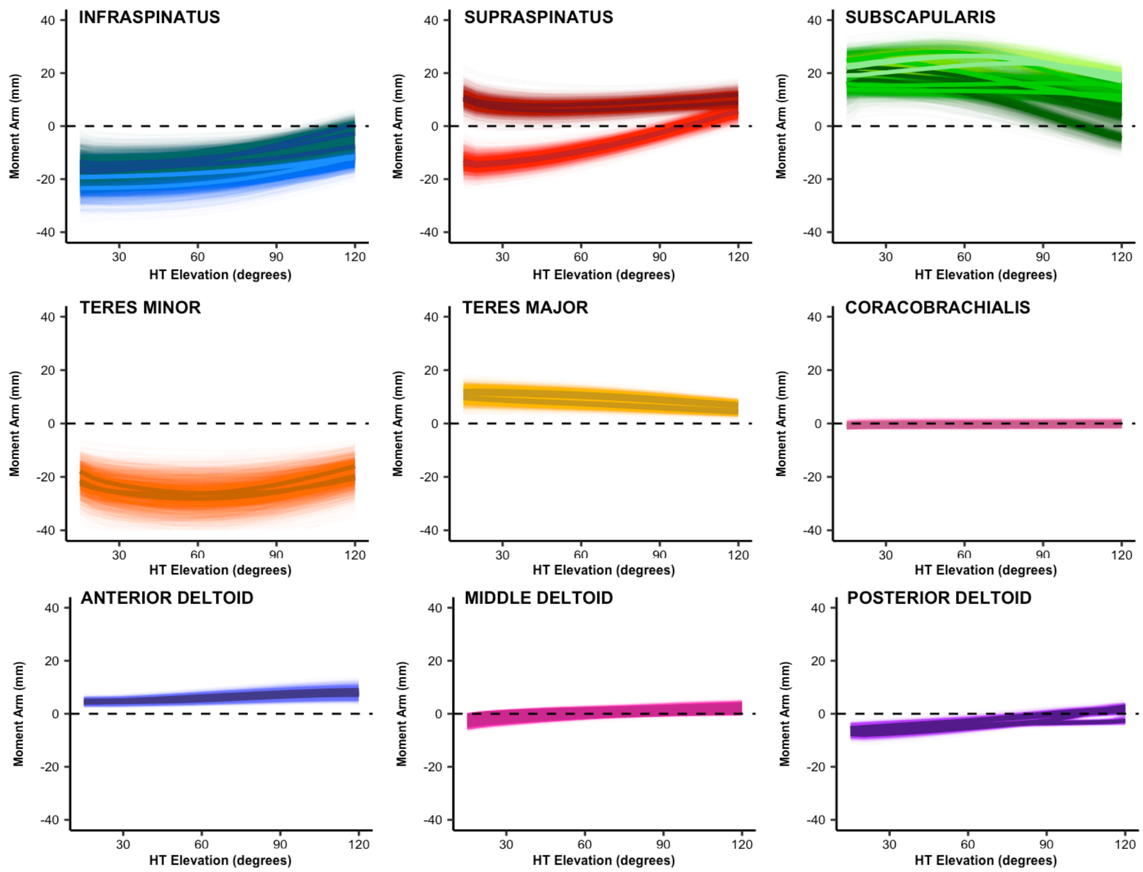

Figure A.2: Model-predicted humeral rotational moment arms with varying attachment locations. All simulations (1000 iterations for each muscle element) are plotted as thin lines. Thick, dark bands represent mean values for each muscle. Darker shades for the infraspinatus/subscapularis and supraspinatus represent superior and anterior regions, respectively. Positive (negative) values indicate humeral internal (external) rotation.

Table A.2: Regression models predicting humeral rotational (i.e. internal/external rotation) moment arms (mm) for each scapulohumeral muscle. Independent variables include attachment changes at the scapula (S_x, S_y, S_z) and humerus (H_x, H_y, H_z) in each axis. For the anterior deltoid, the attachment locations were changed at the clavicle (C_x, C_y, C_z) (denoted with an asterisk*). All attachment changes were centered relative to the mean and divided by the standard deviation. Humeral elevation angle relative to the thorax (HT) and its quadratic (HT^2) and cubic (HT^3) terms were added as covariates. Values represent unstandardized beta coefficients.

Muscle	B_0	S_x or *C_x	S_y or *C_y	S_z or *C_z	H_x	H_y	H_z	HT	HT^2	HT^3	R^2
S. Infraspinatus	-14.9	0.1	2.2	-0.1	2.2	-0.1	-1.7	-6.7E-02	1.7E-03	-2.7E-06	0.89
I. Infraspinatus	-20.5	0.2	1.2	-0.8	2.6	0.0	-2.7	-1.1E-02	4.7E-04	2.0E-06	0.95
A. Supraspinatus	13.9	1.4	0.0	-0.2	1.8	0.1	0.0	-3.2E-01	4.4E-03	-1.7E-05	0.88
P. Supraspinatus	-13.0	1.2	1.6	0.1	1.2	0.0	-0.6	-1.2E-01	3.7E-03	-1.2E-05	0.97
S. Subscapularis	18.3	-0.1	-2.7	0.1	-0.1	0.5	-0.1	1.3E-01	-3.3E-03	9.5E-06	0.89
M. Subscapularis	21.7	0.5	-1.0	-0.2	2.3	-0.5	-0.4	2.8E-01	-5.7E-03	2.3E-05	0.95
I. Subscapularis (L)	18.5	0.3	-0.1	-0.2	2.5	-0.1	-0.5	1.9E-01	-9.3E-04	-5.8E-06	0.85
I. Subscapularis (S)	15.6	0.2	-0.1	-0.2	2.6	0.0	-1.0	-6.6E-02	1.2E-03	-6.4E-06	0.94
Teres Minor	-15.9	0.0	0.5	-0.2	0.8	0.1	-4.6	-4.1E-01	4.4E-03	-9.3E-06	0.92
Teres Major	10.9	0.2	-0.2	-0.1	1.2	0.3	-1.0	-1.4E-02	-3.4E-04	7.2E-07	0.98
A. Deltoid (1)*	5.0	0.1	0.2	-0.3	0.0	0.3	0.8	-5.5E-02	1.3E-03	-5.6E-06	0.92
A. Deltoid (2-4)*	4.3	0.1	0.1	-0.3	0.1	0.5	0.8	-6.3E-04	7.0E-04	-3.7E-06	0.94
M. Deltoid (1-4)	-4.8	0.5	0.2	0.2	-0.6	-0.1	-0.2	1.0E-01	-6.2E-04	2.2E-06	0.97
M. Deltoid (5-7)	-3.1	1.3	0.1	-0.1	-0.4	0.0	0.2	9.5E-02	-8.2E-04	3.0E-06	0.96
P. Deltoid	-7.0	0.3	0.5	1.4	0.1	-0.2	-0.5	4.2E-02	2.2E-04	-1.8E-07	0.93
Coracobrachialis	-1.0	0.0	-0.1	-0.3	0.4	0.0	0.5	2.3E-02	-3.0E-04	1.4E-06	0.95