INVESTIGATION OF THE NEUROMUSCULAR CONTROL OF THE SHOULDER

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WHEN PERFORMING CONCURRENT UPPER EXTERMITY TASKS

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ABSTRACT

The purpose of the thesis was to evaluate the neuromuscular control of shoulder muscles when performing shoulder efforts concurrently with hand or elbow efforts in healthy and injured individuals. Of particular interest was the response of the supraspinatus and infraspinatus muscles to performing an additional hand task, such as targeted gripping, while also performing different shoulder actions. Prior to doing so, two studies were undertaken to provide the necessary groundwork for the remaining two studies of this thesis. The first study investigated whether changes to shoulder muscle activity previously seen with gripping where solely the result of the novelty of using feedback to regulate grip force. The results of this study suggested that the alterations in shoulder muscle activity with gripping are not diminished with repetition. The second study provided an improved method of normalizing electromyograms from dynamic contractions and was used in the subsequent studies of this thesis. Studies 3 and 4 of this thesis examined the response of shoulder muscles in healthy individuals during static sub-maximal efforts and maximal dynamic efforts in flexion and scapular planes with neutral and supinated forearm postures. Three conditions were tested in both studies: (i) no additional load, (ii) gripping to 30% of maximum and (iii) contracting the biceps to 30% of maximum. A prevailing theme found during sub-maximal contractions was individuality in neuromuscular recruitment strategies and precluded any significant effects of gripping or biceps contractions. During dynamic contractions, concurrent shoulder efforts with gripping and biceps contractions was found to significantly decrease

deltoid, supraspinatus and infraspinatus muscle forces during flexion with supinated forearm posture.

This thesis provided a thorough examination of shoulder electromyography in healthy individuals, improving our understanding of the neuromuscular control of the shoulder musculature. A common theme of this thesis was the individuality of neuromuscular strategies of the shoulder.

Keywords: Shoulder, Rotator Cuff, Electromyography, Neuromuscular Control.

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THESIS FORMAT AND ORGANIZATION

This thesis contains material from the PhD work of Joanne N. Hodder and has been prepared in a “sandwich” format as outlined in the McMaster University School of Graduate Studies’ Guide for the Preparation of Thesis. The thesis begins with a general introduction to the research area (Chapter 1), followed by 3 studies that have been prepared as 4 manuscripts and individual thesis chapters (Chapters 2-5). The thesis ends with a concluding chapter (Chapter 6) that provides a discussion of the findings and recommendations for future research in the area.

Chapter 2 investigated the effect of repeated exposure to simultaneous targeted gripping during shoulder exertions to determine role of feedback in the alterations in shoulder muscle activity. Chapter 2 is ‘in press’ in the *Journal of Electromyography and Kinesiology*. Corrected proof is online at http://dx.doi.org/10.1016/j.jelekin.2011.11.011.

Chapter 3 of this thesis evaluated the effect of several methods of determining maximal muscle excitation on normalized EMG amplitude. Chapter 3 has been prepared to submit to the *Journal of Electromyography and Kinesiology*, upon completion of the dissertation.

Chapters 4 and 5 of this thesis investigated the effects of concurrent gripping and biceps contractions during static and dynamic shoulder exertions on shoulder and rotator cuff muscle activity. Chapters 4 and 5 have been prepared in manuscript form and will be submitted upon completion of the dissertation.

CONTRIBUTIONS TO PAPERS WITH MULTIPLE AUTHORS

Chapter 2 – In Press

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Chapter 3 - Formatted for Publication

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Chapter 4 - Formatted for Publication

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Contributions

Both authors have contributed substantially to all chapters. The study conception, methodology, data collection, analysis and manuscript preparation was performed by Joanne Hodder with significant input at all stages from Dr. Peter Keir.

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LIST OF ABBREVIATIONS

MVE	Maximum voluntary excitation
ROM	Range of motion
DSEM	Delft Shoulder and Elbow Model
SIMM	Software for Interactive Musculoskeletal Modeling
EMG	Electromyography
AD	Anterior deltoid
MD	Middle deltoid
PD	Posterior deltoid
TR	Trapezius
IN	Infraspinatus
TM	Teres major
LD	Latissimus dorsi
BB	Biceps brachii
MVG	Maximum voluntary grip
ANOVA	Analysis of variance
MVC	Maximum voluntary contraction
MVIC	Maximum voluntary isometric contraction
MVDC	Maximum voluntary dynamic contraction
LT	Lower trapezius
UT	Upper trapezius
PM	Pectoralis Major

MVIE	Maximum voluntary isometric excitation
MVDE	Maximum voluntary dynamic excitation
MEE	Maximum experiment excitation
CCW	Counter clockwise
CW	Clockwise
3D	Three dimensional
BI	Biceps
TRI	Triceps
INF	Infraspinatus
SUP	Supraspinatus
MRI	Magnetic resonance imaging

CHAPTER 1: INTRODUCTION

1.1 Introduction

The shoulder complex is characterized by its unique multi-articular anatomy and large multi-directional range of motion. These characteristics allow for the performance of a wide array of actions, ranging from functions of daily living to more strenuous occupational and athletic activities. To allow for this vast mobility, the glenohumeral joint has little passive stability from boney structures. Accordingly, it relies heavily on precisely coordinated muscle activity to not only produce motion but to also to maintain joint integrity. Unfortunately, this also means the shoulder is highly susceptible to injury.

Shoulder injuries are ranked third in the North America as the most common injury to the general population (Harrast et al, 2004; Mehta et al, 2003). The rotator cuff muscles are most often the site of injury in the shoulder complex. Disorders to the rotator cuff muscles have been reported to be present in 39- 60% of cadaveric specimens, with the supraspinatus being the most frequent site of injury (Krishnan and Hawkins, 2003). Although much is known about the occupational risk factors for shoulder injury, the intrinsic mechanisms of injury to the rotator cuff muscles are still widely debated (Browning & Desai, 2004). Biomechanical models have been developed to more closely examine the mechanics of the shoulder and can provide further insight into potential injury mechanisms (DeLuca & Forrest, 1973; Hogfors et al, 1987; Van der Helm & Veenbaas, 1991, van der Helm, 1994; Dickerson et al, 2008). Being able to accurately describe the neuromuscular control and function of the shoulder in both healthy and pathological populations will allow for greater insight into how the shoulder becomes injured (injury mechanisms) and how it functions once injured (adaptations). Greater knowledge about how injuries develop and manifest themselves will not only help to advance treatment outcomes, but also improve the ability to identify risk factors and prevent future shoulder injuries.

1.2 Anatomy and Function

The shoulder complex consists of four bones, the scapula, sternum, clavicle and humerus. These bones articulate with each other and the thorax to form the shoulder's complex of joints, consisting of the glenohumeral, acromioclavicular, sternoclavicular and scapulothoracic joints. These joints allow the shoulder's wide range of motion in both rotational and translational planes (Lee et al, 2000). To coordinate movement about these joints, many muscles are needed. Inman et al (1944) assembled the muscles into three groups. The scapulohumeral group, which spans the scapula to the humerus and consists of the rotator cuff muscles (infraspinatus, supraspinatus, subscapularis and teres minor) as well as the three heads of the deltoid muscle (Figure 1.1). These muscles act primarily at the glenohumeral joint, producing moments to cause humeral external rotation, abduction (frontal plane) and flexion (sagittal plane). The rotator cuff muscles, however, are also known to be active in many other postures as they are highly involved in providing stability to the glenohumeral joint. The axioscapular group spans from the spine and ribs to the scapula and consists of the trapezius, rhomboids, serratus anterior and levator scapulae. This group of muscles extend over the scapulothoracic joint and function to elevate and rotate the scapula. The third group, the axiohumeral group spans from the sternum, ribs and spine to the humerus and the muscles often cross multiple joints. This group consists of the pectoralis major, pectoralis minor and latissimus dorsi (Figure 1.1). These muscles are primarily responsible for humeral flexion, adduction and internal rotation (Inman, 1944). Table 1.1 summarizes the muscles required to perform basic movements at the shoulder.

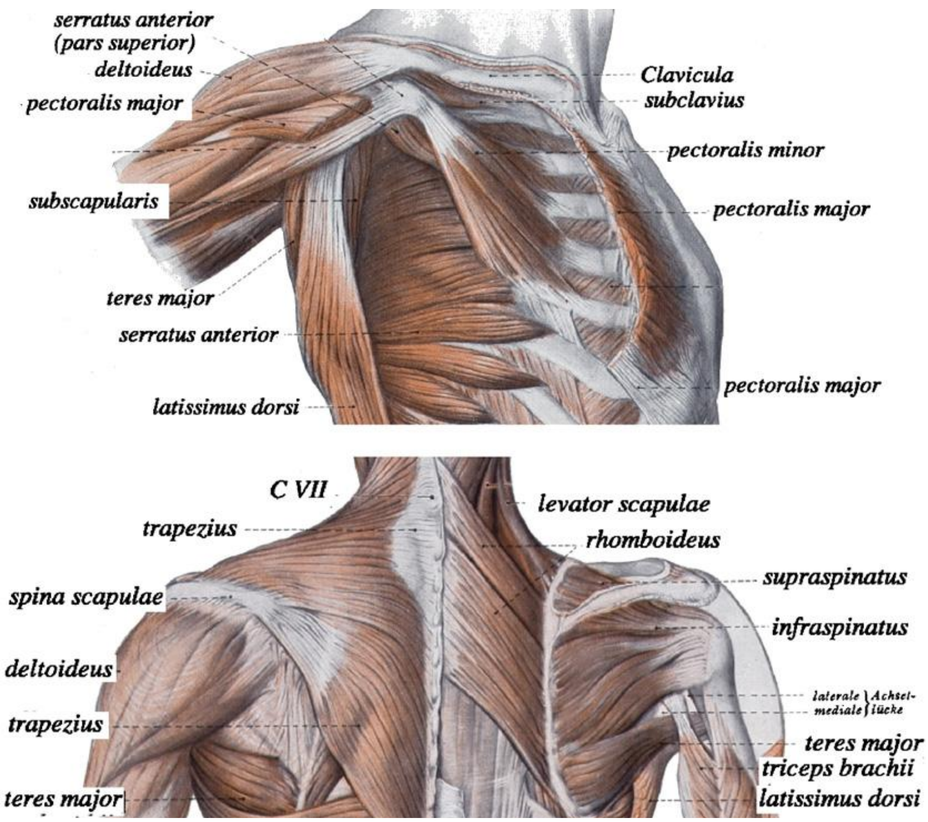

Figure 1.1. Muscles of the shoulder (from Benninghoff-Goertler, 1964, Lehrbuch der Anatomie des Menschen, 9th edition, Urban & Schwarzenberg, Berlin.)

The shoulder is reliant on passive and active tissues to position and sustain the humeral head within the glenoid cavity to maintain joint integrity. The glenoid cavity consists of the glenoid fossa, an indentation in the scapula creating the articular surface, and the labrum, a ring of soft tissue which surrounds the glenoid fossa providing a deeper groove in which the humeral head articulates. The rotator cuff muscles are predominantly responsible for providing structural support to the joint. The rotator cuff muscles surround the joint complex and work both passively and actively to centre the humeral head within the glenoid cavity during shoulder function. Due to the location and the orientation of the rotator cuff muscles, they function passively to mitigate dislocation of the humeral head, especially at end range of motion (Lee et al, 2000). Actively, the individual lines of force for the rotator cuff muscles lend them to providing active redirection of the humeral head within the centre of the glenoid fossa. This occurs in particular during mid range postures and dynamic movements.

Table 1.1: Summary of the primary shoulder muscles active during basic shoulder movements.

Shoulder Motion	Active Muscles
Shoulder Flexion	Anterior deltoid, clavicular head of pectoralis major, coracobrachialis, biceps brachii
Shoulder Extension	Posterior deltoid, latissimus dorsi, sternocostal fibres of pectoralis major, long head biceps, teres major
Shoulder Abduction	Middle deltoid, supraspinatus, serratus anterior (>80°), trapezius (>80°)
Shoulder Adduction	Pectoralis major, latissimus dorsi, teres major, coracobrachialis, long head triceps
Shoulder External Rotation	Infraspinatus, teres minor, posterior deltoid
Shoulder Internal Rotation	Subscapularis, pectoralis major, latissimus dorsi, anterior deltoid, teres major
Scapular Elevation	Upper trapezius, levitator scapulae, rhomboids
Scapular Depression	Pectoralis major and minor, latissimus dorsi, serratus anterior, lower trapezius, subclavius
Scapular Upward Rotation	Trapezius, serratus anterior
Scapular Downward Rotation	Rhomboids, levator scapulae, pectoralis major and minor, latissimus dorsi
Scapular Protraction	Serratus anterior, pectoralis major and minor
Scapular Retraction	Trapezius, rhomboids, latissimus dorsi

Since active contraction is responsible for joint integrity in the middle range of motion, muscle activity must be highly coordinated to avoid the joint from being compromised while performing a multitude of tasks. To understand the neuromuscular control of the shoulder, it is necessary to understand the role and function of each muscle as they work collectively to achieve a desired action.

1.3 Assessing Shoulder Muscle Function

Common methods used to assess muscle function are electromyography (EMG), dynamometry and motion analysis. All movements of the body are produced by the arrival of an electrical stimulus to the muscle. The extent to which a muscle is active during a task can be monitored via electromyography (Kamen, 2004) which may be used to assess the individual muscle contributions to externally produced forces and moments. The net effect of muscle activity can be measured by joint kinetics (forces and moments) and kinematics (displacement, velocity and acceleration). Dynamometry (kinetics) and motion analysis (kinematics) can be used to assess the characteristics of the externally produced forces and moments that result from muscle activity.

Shoulder muscles are thought to work together in functional units (Hagberg, 1981). To determine the role and function of each muscle, early investigations used nerve blocks. The ability of the shoulder muscles to exert force in various directions of effort were examined before and after a nerve block was applied to temporarily paralyze muscles surrounding the muscle of interest. To examine the function and strength of specific rotator cuff muscles in isolation, studies have used either a suprascapular nerve block to induce temporarily paralysis of the supraspinatus and infraspinatus or an axillary nerve block to induce temporary paralysis in the deltoid (Van Linge and Mulder, 1963; Colachis and Strom, 1971; Howell et al, 1986). These studies concluded that the deltoids and supraspinatus were equally involved in producing moments in all functional planes of the shoulder, demonstrating that the supraspinatus provides more than stability and external rotation. The involvement of the supraspinatus in many shoulder actions helps to explain why it is the most commonly overused and injured rotator cuff muscle (Krishnan and Hawkins, 2003).

Many shoulder muscles have demonstrated dual functionality making it challenging to define a solitary function for each muscle during any given task. Electromyography (EMG) has been used to examine shoulder function for decades and much valuable insight has been gained (Bagg and Forest, 1986; Kronberg et al, 1990; Illyes and Kiss, 2005). Perhaps the most comprehensive EMG analysis of healthy shoulder muscle activity was conducted by Kronberg et al (1990). They examined patterns of shoulder muscle activity during flexion, abduction and internal/external rotation movements and found that every muscle, with the exception of the pectoralis major, was activated during every movement analyzed (Kronberg et al, 1990). Kronberg et al (1990) also found that opposing muscle pairs were commonly activated synchronously during movement, potentially as a mechanism to maintain joint integrity while still producing force in the desired direction.

Meskers et al (2004) conducted a thorough examination of the contributions of individual shoulder muscles via EMG to forces produced in multiple directions. With the shoulder flexed to 45° and elbow flexed to 90° (Figure 1.2). Static muscle contractions were performed from flexion to abduction to extension to adduction and back toward flexion in 30° increments about the humerus. This detailed study provided the optimal directions of force production for each muscle and also provided the range of angles at which each muscle contributed force. Many muscles produced force in multiple directions. The latissimus dorsi was found to contribute force equally in two opposing directions, as did subscapularis and teres major. Similar to Kronberg et al (1990), opposing muscle pairs were seen to be co-active during shoulder exertions. For example, during forward flexion force production, the greatest contributors were the anterior and middle deltoid, supraspinatus and teres major. However, latissimus dorsi, infraspinatus, and subscapularis were also active, despite these muscles having lines of action opposing humeral flexion. In this circumstance, they are likely acting to balance forces to maintain joint integrity. This work clearly demonstrates the complexities of the shoulder joint and the activation of functional units of muscles rather than simple agonist-antagonist muscle pairs (Hagberg, 1981; Meskers, 2004).

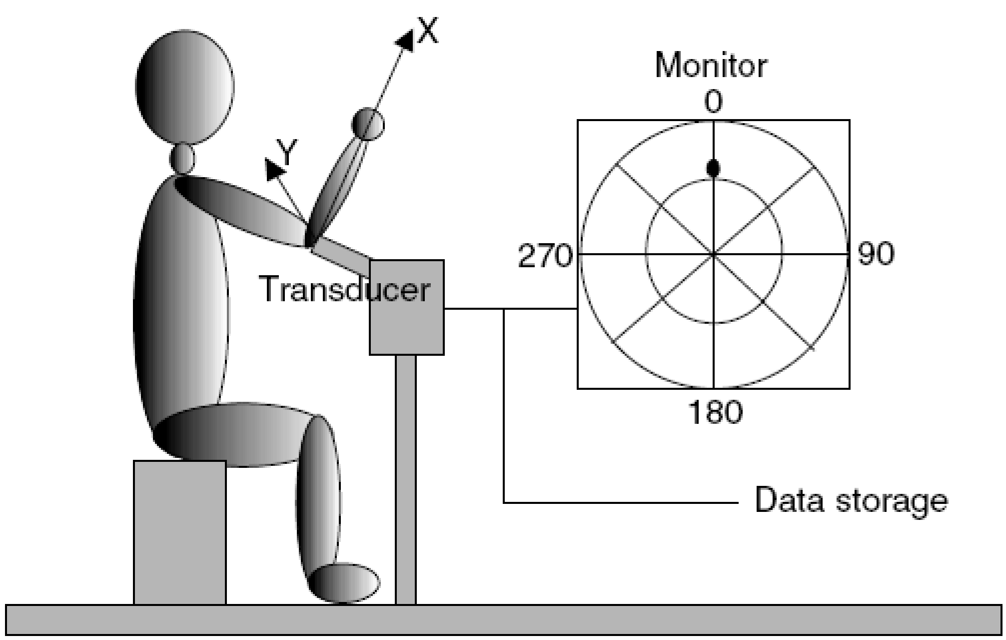

Figure 1.2. From Meskers et al, 2004. Static muscle contractions in flexion (0°) to abduction (90°) to extension (180°) and adduction (270°) in 30° increments.

1.4 Occupational Risk Factors and Mechanisms of Injury

Work related injury to muscles, tendons, ligaments and nerves are prevalent in many labour intensive occupations. Shoulder injuries are second only to back injuries as the most common site of occupational musculoskeletal injury (Walker-Bone and Cooper, 2005). Identified risk factors include: sustained or intermittent forceful exertions using the upper extremity, awkward postures, arm or hand vibrations and repetitive movements (Staal et al, 2007; Leclerc et al, 2004; Frost et al, 2002; Zakaria et al, 2002).

LeClerc et al (2004) examined over 1400 workers who performed highly repetitive upper extremity movements at work. They found that both age and gender were factors contributing to the incidence of shoulder pain. Overall, 45% of the workers experienced shoulder pain over a 6 month period and the prevalence grew as worker age increased. Men reported slightly lower incidence (37%) of shoulder pain than women (39%). Women reported several factors that contributed to the incidence of shoulder pain, including: bending forward, working with arms at shoulder level and the use of vibrating tools. For men, the repetitive use of a tool was the only factor that was highly associated with incidence of shoulder pain. Frost et al (2002) found that in industrial and service workers, the prevalence of shoulder tendinitis was two to three time greater for those performing repetitive tasks than those who did not. Other factors that were identified to impact the prevalence of injury were the amount of force required and the frequency and duration micro-pauses or breaks (Frost et al, 2002).

In order to improve upon risk assessment, rehabilitation and prevention of injury, it is essential to understand how the musculoskeletal system controls joint function and the mechanisms that can influence this operation. During functional tasks using the upper extremity, shoulder actions are often coupled with activities of the elbow and hand. The addition of a secondary task to a shoulder muscle contraction has been shown to alter muscle activity (Sporrong et al, 1996; Sporrong et al, 1995) however the mechanism(s) by which this secondary task influences muscle activity are still unclear. It is possible that the addition of the secondary task increases the cognitive load (Au & Keir, 2007; MacDonell & Keir, 2005), in particular if it is a novel task (Shemmell et al, 2005) or perhaps necessitates the sharing of cognitive resources, as seen in dual task paradigms (Kahneman, 1973). It may also be possible that the addition of a secondary task that uses the hand affects the balance of moments across the joints of the upper extremity traveling from the distal to proximal extremity. Further research is necessary to determine what mechanism is involved in coordinating neuromuscular activity of the shoulder during dual tasks.

1.5 Motor Control, Learning and Dual Task Paradigms

Waersted (2000) proposed three sources of vocational muscle activity: i) muscle activity related to the biomechanical need of force production to produce movements or to generate forces necessary to maintain posture, ii) muscle activity related to the biomechanical need to stabilize body segments to allow for other muscles to produce moments, and iii) muscle activity occurring without any obvious biomechanical purpose. Muscle activity will typically arise from at least two if not all three of these sources. When multiple factors are engaged simultaneously, it is difficult to delimit how much each factor is contributing to the overall muscle activity. Greater knowledge of how complex tasks affect muscle activity will improve our understanding of the mechanisms that contribute to injury development.

It has been suggested that muscle activity arising from psychological, organizational and social factors or other non-biomechanical factors (Jacobson, 1927), may be the source of individual differences to tolerance of occupational tasks (Waersted & Bjorklund, 1987). Included in the factors influencing muscle activity are the needs for precision (Sporrong et al, 1998; Laursen et al 1998), perceived psychosocial stress (Waersted, 2000) and work situation (ie. limited training or time deadlines) (Waersted, 2000). In all cases, these factors were shown to increase muscle activity (Waersted, 2000, Sporrong et al, 1996; Laursen et al 1998).

Occupational tasks involving the shoulder are often coupled with actions of the distal extremity and rely on visual, auditory or proprioceptive feedback to perform the task. For example, when using a hand drill, the shoulder musculature supports the load of the drill and helps apply force in the intended direction while using the hand to squeeze the trigger appropriately. Using three forms of feedback (visual, auditory, proprioceptive), the magnitude of shoulder muscle force and grip force necessary to complete the drilling task is monitored and gauged. If this feedback results in an increase in cognitive load, previous studies suggest that an increase in muscle activity would be expected (Waersted, 2000, Sporrong et al, 1998; Laursen et al 1998). However, in the shoulder complex, this is not necessarily the case. When shoulder contractions are coupled with targeted gripping, the activity of some shoulder muscles increase while others decrease (Smets et al, 2009; Antony & Keir, 2009; Au & Keir, 2007; MacDonell & Keir, 2005; Sporrong et al, 1996, Sporrong et al, 1995). In addition, the overall force generating capacity of the shoulder musculature decreases (Smets et al, 2009; MacDonell & Keir, 2005). The altered muscle activity and motor control patterns associated with concurrent gripping and shoulder actions have warranted further investigation.

MacDonell and Keir (2005) investigated the impact of both combined visual and auditory targeted gripping (30% of maximum voluntary grip) on maximal shoulder strength. The gripping task was performed in conjunction with maximal isometric shoulder flexion and abduction contractions at 30°, 60° and 90° degrees of elevation in each plane. Overall, gripping lowered shoulder maximal moments by 3.4-10 Nm, depending on the angle of elevation. To determine the effect of gripping and a known cognitive load, maximum shoulder moments were measured while concurrently performing the Stroop colour word test (Stroop, 1935). Similar reductions in maximum shoulder moment were found when the Stroop was performed concurrently with shoulder actions, with changes ranging from 3.5 -9.1 Nm. Furthermore, when targeted gripping and the Stroop task were each performed during maximum shoulder flexion and abduction exertions, moments fell as much as 5.8 – 14.2 Nm. Decreased shoulder moments were reflected in the patterns of muscle activity during these tasks, as the anterior deltoid, middle deltoid and infraspinatus activity were lowered by upwards of 6.5% of maximum voluntary excitation (MVE) with gripping. Decreases found with the Stroop were upwards of 10 % MVE in these muscles, and when both gripping and the Stroop were performed, muscle activity was lowered by upwards of 14.6% MVE.

These changes in motor performance may be seen as an example of a dual task paradigm where the addition of the secondary task, either in the form of Stroop or targeted gripping, reduces the ability to perform the maximum shoulder contraction when compared to performing each task alone. It is thought that this paradigm occurs at the central level and is a reflection of our limited capacity for attention or information processing. When two tasks are performed simultaneously, cognitive resources need to be shared and the performance of one or both tasks is reduced (Kahneman, 1973).

The effect of dual task paradigms on muscle activity is not well researched. Primarily, EMG has been used to determine temporal latencies of muscle onset as an indicator of performance and EMG magnitude has not been investigated (Woollacott & Shumway-Cook, 2002). In the few studies that have examined muscle activity during dual tasks, there appear to be conflicting results (Smets et al, 2009, Antony & Keir, 2009; Au & Keir, 2007 Rankin et al, 2000; Caldwell et al, 1992). Rankin et al (2000) found that the presence of a secondary task, in the form of mental math computation, interfered with standing balance and lowered gastrocnemius and tibialis anterior activity. However, Caldwell et al (1992) found that performing forearm supination to a visually targeted moment level while maximally contracting the biceps, increased the overall magnitude of muscle activity from the biceps and brachioradialis than during either forearm supination or biceps contraction alone (Caldwell et al, 1992).

When visually targeted gripping was performed with maximal isometric shoulder efforts in the push/pull, cranial/caudal and medial/lateral directions, Smets et al (2009) found consistent decreases in both deltoid muscle activity and shoulder strength, ranging from 15-30% and 18-25%, respectively. During sub-maximal shoulder abductions, Au and Keir (2007) found that visually targeted gripping acted to significantly decrease anterior and middle deltoid activity by approximately 2% MVE in the frontal plane. When the Stroop test was performed during sub-maximal shoulder abduction, similar decreases in anterior deltoid activity were observed. Given that the Stroop test produced similar changes to shoulder muscle activity as targeted gripping, it seems logical that the feedback used to maintain grip force in these studies may have caused an increase the cognitive load in the same manner as the Stroop test (Au & Keir, 2007; MacDonell & Keir, 2005). This supports the notion that the mechanism influencing these changes is cognitive in nature rather than physical. However, this effect may be task dependent.

Further supporting the notion of a cognitive mechanism are studies examining high level of precision tasks. Muscle activity was found to increase when task precision increased, purportedly due to greater cognitive demands during higher precision tasks (Visser et al, 2004; Laursen et al, 1998). However, Au and Keir (2007) found that increasing the level precision in grip force did not affect muscle activity. With such a wide variety of documented responses to dual tasks in the upper extremity it is difficult to determine a single mechanism responsible for the changes seen in muscle activity and shoulder strength.

There is some evidence to suggest that gripping may cause a chain of physical events that result in altered shoulder muscle activity. When forearm and shoulder muscle activities were assessed during concurrent visually targeted gripping and low level push and pull forces, no changes in shoulder muscle activity were found (Di Domizio and

Keir, 2009). Antony and Keir (2009) examined sub-maximal shoulder muscle contractions in the frontal, sagittal, and mid-way between the two planes at multiple angles of elevation, with and without visually targeted gripping. Across all postures, the addition of the gripping task resulted in moderate decreases in anterior deltoid and middle deltoid activity of 2.4% and 2.2% MVE, respectively, while activity increased in the other muscles investigated. On average, infraspinatus activity was 1.7% MVE higher and biceps brachii activity 6.0% MVE higher when gripping. These results provide the basis for the hypothesis that gripping may result in a redistribution of force, which in the some postures can increase the activity of the rotator cuff muscles (Antony and Keir, 2009).

There is a physical link between the muscle activity in the forearm and that of the shoulder. The biceps brachii originates from the scapula, as the short head extends from coracoid process and the long head from the supraglenoid tubercle. The biceps has many functions, one of which is to produce a moment about the elbow. Forces, such as those exerted by the wrist and finger flexors and extensors during gripping, may necessitate increased biceps activity to balance moments at the elbow joint. The biceps also lends itself to having a role in shoulder function and provides a potential physical pathway for gripping to alter shoulder muscle activity. It is not yet clear how the shoulder muscles are controlled and coordinated to meet the demands of the task, while also balancing joint forces. However, the biceps may have a role in this balance during gripping tasks.

Rarely is shoulder moved in isolation of the rest of the upper limb. Rather, it is typically used in combination with movements of the distal limb, often for the purpose of performing tasks with the hands. If the mechanism responsible for altered shoulder muscle activity with gripping is found to be physical in nature, it could provide invaluable insight into how the shoulder controls such precisely orchestrated muscle activity.

The ability to assess risk and prevent shoulder injury is reliant on understanding how stability of the joint is maintained and what postures or tasks make it unstable. From a mechanical perspective, the definition of instability is the inability of an object to return to its initial resting position once being perturbed. However, the clinical definition of instability is defined somewhat differently. A review by Veeger & van der Helm (2007) commented that, "in clinical terms, (instability) is similar to a dislocation of the joint…synonymous with too large (of a) displacement after force exertion on the humerus." Shoulder instability has been well documented to be coupled with the presence of altered patterns of muscle activity, particularly of the posterior and anterior deltoids, latissimus dorsi and infraspinatus (Jaggi et al, 2009; Barden et al, 2005; Labriola et al, 2005). A few studies have investigated the patterns of muscle activity during rehabilitation exercises in people with shoulder disorders, such as Ballantyne et al (1993), who examined individuals with recurrent unilateral pain and weakness, and Clisby et al (2008), who examined individuals with symptomatic subacromial impingement. These studies indicated that the shoulder muscles in persons with injury behave differently than healthy individuals. However, none have examined how alterations with tasks such as gripping might affect the activity of shoulder muscles. Clinically, shoulder injuries are often difficult and complicated to rehabilitate. Understanding how injured shoulder muscles behave and the affect this may have on joint stability would greatly contribute to the improvement of treatment and the prevention of future injuries.

1.6 Biomechanical Models of the Shoulder

Biomechanical models have been used clinically with some success. For instance, computer simulation has been very useful in providing the visualization necessary to evolve orthopaedic surgical strategies (Delp & Loan, 1995). Biomechanical models have been developed to better understand the mechanical effect of muscle forces on joints, such as those experienced at work. There are two stages to the development to assess the mechanics of a complex musculoskeletal system. The first is to develop an accurate representation of the system and the second to use it to estimate muscle forces and motions and how that impacts the joints of the system (DeLuca & Forrest, 1976; van der Helm & Veenbaas, 1991).

The complexity of the shoulder makes it challenging to develop an anatomically and mechanically accurate biomechanical model. DeLuca and Forrest (1976) began by examining the contributions of individual muscles during an isometric abduction effort in a single posture. A challenge to doing this, one that has yet to be resolved, is that the axis of rotation of the shoulder is instantaneous and changes with movement. Thus, DeLuca and Forrest (1976) sought to find the instantaneous centre of rotation in specific arm postures and then used simplified anatomy to estimate the muscle contributions to joint moment. This model only included representation of one rotator cuff muscle, the supraspinatus, and the three heads of the deltoid, thus, further development was necessary.

Hogfors et al (1987) proposed a model that was capable of computing parameters during dynamic motions. This model included definitions of many structural elements of the shoulder. A fixed coordinate system for the bone structure, determination of muscle insertions and geometrical constraints were all included to more accurately represent in vivo anatomical limitations. There were two major assumptions made by this model: (i) muscles were modeled in a single line of force, acting about the centroid of the joint (which was estimated by the authors) (ii) joint contact forces, friction and most ligament forces were assumed to be negligible and ignored as they were thought to act directly into the joint, and thus were not capable of producing a moment. Four coordinate systems were used to describe the shoulder system; the sternal, clavicular, scapular and humeral systems were defined about the fixed system of the thorax. Each system had three Cartesian coordinates, resulting in a rigid body model with 12 degrees of freedom with three sets of Euler angles to describe each systems position from that of the fixed system. External force due to the mass of the arm was dependent on position and any load added to the hand could also be considered when calculating joint moments.

Unfortunately, most shoulder models did not include the biceps brachii, despite it being identified as a contributor to shoulder function. Because the biceps originates from the scapula, its line of action allows it to contribute to shoulder flexion. Additionally, the biceps has been shown to play a role in shoulder adduction and when used eccentrically, in extension (Inman et al, 1944). Thus, the biceps shares many of the same functions as the supraspinatus, infraspinatus, subscapularis and teres minor, as they act to stabilize the glenohumeral joint and thus the biceps could have a large impact on the activity of these muscles (Hogfors et al, 1987). Dickerson et al (2007) further expanded upon the geometric model of Hogfors et al (1987) by including the biceps and triceps muscles, as well as integrating anthropometric and geometric parameters from experimentally collected kinematic data to predict muscle forces. Muscle activity in this model was viewed as either on or off. Using this on/off approach, Dickerson et al (2008) found that model predicted forces were highly correlated with EMG predicted forces when muscle activity greater than 5% MVE. The model, however, did not perform well at predicting muscle activity lower than 5% MVE and predicted forces using the model were often in poor agreement with those calculated from recorded EMG (Van der Helm, 1994). Therefore, this model may not be the best choice when analyzing the control of joint stability when low level forces are expected in the form of small stabilizing muscle forces necessary to maintain joint integrity.

Van der Helm and Veenbass (1991) developed the most widely recognized and used shoulder model. Known as the Delft Shoulder and Elbow Model (DSEM), it has highly detailed description of muscle architecture, and also includes the biceps and the distal extremity. Since many muscles in the shoulder region broaden and fan out from their origin, muscles in the DSEM are subdivided into bundles of 95 lines of action to best represent the mechanical effect of 20 muscles surrounding the shoulder complex. The biceps long head was included as one of these 20 muscles, as well as elbow and wrist angles.

To date, DSEM is one of the most comprehensive shoulder models. However, further research is necessary to understand the neuromuscular complexities of shoulder function in healthy and injured shoulders situ. In order for models to be able to accurately compute muscle forces and joint loading, the parameters of neuromuscular control need to be better defined so the objective functions used to govern these models can best reflect the motor control decisions of humans. Delp & Loan (2000) developed an interactive 3D simulation of the human skeletal structure, known as Software for Interactive Musculoskeletal Modeling (SIMM). An open source version of this software, OpenSim, uses the geometry of the DSEM model along with muscle parameters described by Holzbaur et al (2005) to simulate the upper extremity. The muscle fibre lengths, pennation angles and physiological cross sectional areas used in the model are from Langenderfer (2004). These musculoskeletal elements are representative of the 50th percentile male, at a height of 170 cm (Holzbaur et al, 2005).

1.7 Purpose and Hypotheses

To better understand the mechanisms controlling shoulder function, many questions need to be answered. What governs the coordination of shoulder muscles? How does injury status affect activation strategies? Why does gripping while maintaining a shoulder action alter muscle activity? Does the biceps brachii play a role in the alterations of shoulder muscle activity with gripping? Examination of the muscle activity of individuals with rotator cuff pathology during activities of daily living is limited at best.

The purpose of this thesis was to examine the mechanism(s) responsible for altering patterns of muscle activity when targeted gripping and shoulder isometric and isokinetic muscle contractions are performed concurrently. To date, little is known about the role of the biceps in this paradigm, what effect it has on the rotator cuff muscles during dynamic movements and how this changes when rotator cuff injury is present. This thesis used advanced techniques in the form of fine wire EMG to capture the activity from the supraspinatus and infraspinatus. As well, a musculoskeletal model was used to compute muscle forces to determine if the alterations found in experimentally collected muscle activity effect joint loading. This collection of studies provided greater insight into the motor control and function of the shoulder muscles in both healthy individuals and individuals with supraspinatus pathology during commonly paired tasks of exerting the shoulder and gripping.

The purpose of the first study (chapter 2) was to investigate whether changes in muscle activity found during visually moderated gripping concurrent with isometric shoulder abduction were the result of increased cognitive load from the novelty of the gripping task. A motor learning approach was used to examine the effect of repeated exposure to simultaneous targeted gripping during sub-maximal shoulder abduction on changes to shoulder muscle activity. It was hypothesized that if targeted gripping was perceived as a novel task, repeated exposure would result in diminished effects on muscle activity.

The purpose of the second study (chapter 3) was to establish a more appropriate method of determining a reference value for normalizing muscle activity recorded during dynamic contractions of the shoulder. It was hypothesized that maximal dynamic contractions would generate more optimal reference values than conventional methods of obtaining maximal voluntary excitations from manually resisted muscle specific isometric contractions.

The purpose of the third study (chapter 4) was to investigate the effect of combining targeted gripping and targeted biceps contractions with sub-maximal isometric shoulder flexion and abduction contractions on shoulder muscle activity and forces in healthy individuals. Intramuscular electrodes were used to record activity from the infraspinatus and supraspinatus muscles. It was hypothesized that gripping would increase biceps, supraspinatus and infraspinatus activity, while deltoid activity would decrease. It was also hypothesized that contracting the biceps during shoulder contractions would create a similar response.

The purpose of the fourth study (chapter 5) was to examine EMG based muscle forces recorded during maximal dynamic sagittal and scapular plane shoulder motions while concurrently gripping or contracting the biceps in healthy individuals. It was hypothesized that concurrently gripping and contracting the biceps would lower maximal shoulder moment, decrease deltoid muscle force and increase supraspinatus and infraspinatus muscle force.

CHAPTER 2

Study 1: Targeted gripping reduces shoulder muscle activity and variability

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Targeted gripping reduces shoulder muscle activity and variability

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ABSTRACT

The purpose of this study was to determine if the effect of visually targeted gripping on shoulder muscle activity was maintained with repeated exposures. Eleven healthy males had eight shoulder muscles monitored via surface electromyography while maintaining shoulder elevation at 90° in the scapular plane with and without a 30% grip force. Three non-gripping trials were followed by 15 gripping trials and another 3 non-gripping control trials. Gripping significantly decreased the activity of the anterior deltoid, trapezius, and latissimus dorsi over the exposure of 15 trials. Gripping also reduced variability in all muscles' activity. The changes in shoulder muscle activity are likely in response to forces being transferred through multi-articular muscles spanning from the forearm to the shoulder. Targeted gripping during shoulder elevation resulted in small but significant decreases in muscle activity and reduced variability which supports previous evidence for increased risk of upper extremity disorders in occupational settings.

1. Introduction

As the most versatile joint in the body, the shoulder achieves mobility and stability through muscular action. The inherent lack of passive stability in the shoulder necessitates precise coordination of muscular force, particularly from the rotator cuff muscles. During many activities of daily living, the relatively small rotator cuff muscles are exposed to large forces making them susceptible to injury (Harrast et al., 2004; Lee et al., 2000). In addition to the obvious biomechanical factors of arm posture and hand load, gripping and cognitive effort have also been shown to increase shoulder load and the risk of injury (Smets et al., 2009; Au and Keir, 2007; MacDonell and Keir, 2005; Meskers et al., 2004; Sporrong et al., 1996, 1995; Kronberg et al., 1990).

In both everyday life and work, gripping often occurs with varying levels of shoulder exertion in tasks that are frequently coupled with some form of sensory feedback. For example, when using a hand drill, the amount of force used to grip the trigger and push the drill may be modulated visually, by how far one wishes to drill, tactilely by the resistance of the material or acoustically from the sound of the drill. Sporrong and coworkers (1995, 1996) investigated intermittent and static gripping during static shoulder exertions at 30°, 60°, 90° and 120° of elevation in flexion and abduction. The addition of gripping decreased deltoid activity by 4–14% of maximum voluntary excitation (MVE) as arm elevation angle increased. Although others have found similar decreases in deltoid activity with gripping, the reaction of the rotator cuff to gripping is not as clear (Antony and Keir, 2009; Smets et al., 2009; Au and Keir, 2007; MacDonell and Keir, 2005). Sporrong et al. (1995, 1996) found supraspinatus activity increased by 4–20% MVE while changes to infraspinatus activity were less consistent, in some cases decreasing by as much at 8% MVE while in other cases increasing as much as 21%. MacDonell and Keir (2005) also found infraspinatus activity to decrease by as much as 6.7% MVE with gripping during maximal shoulder exertions. In submaximal dynamic shoulder raises, Antony and Keir (2009) found infraspinatus activity to increase by 2% MVE. While gripping has been shown to alter shoulder muscle activity under a variety of conditions, it is not clear whether the nature of the task plays a role.

A recent study in which participants generated maximal arm efforts in several directions with and without a concurrent visually targeted grip force, suggested that the cognitive effort, or attention paid to the grip force feedback, may be the stimulus for altered shoulder activity (Smets et al., 2009). They found that when grip force was targeted to visual feedback on a computer screen, it always resulted in lower strength and deltoid muscle activity than the unconstrained grip condition, despite the latter sometimes resulting in greater grip force. Increased cognitive effort, in terms of added mental tasks or that associated with visual and/or auditory feedback, has also been implicated in the direction of change in muscle activity (Au and Keir, 2007; MacDonell and Keir, 2005; Visser et al., 2004; Finsen et al., 2001). Finsen et al. (2001) found forearm extensor and flexor activity increased when a task with high memory demands was introduced to a computer mousing activity. Performing a mental task, such as the Stroop word-colour test, produced changes in shoulder muscle activity similar to gripping during both maximal (MacDonell and Keir, 2005) and

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submaximal (Au and Keir, 2007) shoulder exertions. These studies suggest that cognitive effort or interference from visually targeting a grip force may play a role in the redistribution of muscle activity in the shoulder. The need for attentive or cognitive effort may lessen with experience or learning (Galotti et al., 2010) and thus it was thought that the effect on muscle activity may also lessen with repeated exposure.

The purpose of this study was to investigate the effect of repeated exposure to simultaneous targeted gripping during shoulder exertions on muscle activity in an attempt to determine if gripping with feedback plays a role in shoulder muscle activity. Based on motor learning, it was hypothesized that if targeted gripping was perceived as a cognitive load, repeated exposure would reduce its effect on muscle activity due to a learning effect.

2. Methods

2.1. Participants

Eleven healthy males (177.9 ± 5.4 cm; 79.1 ± 10.4 kg; 27.3 ± 4.6 years) with no history of shoulder pain within the last year were recruited from the university population. This study was approved by the Human Research Ethics boards at McMaster University, Hamilton, Ontario. All participants provided written informed consent prior to data collection.

2.2. Electromyography and equipment set-up

The muscle activities of eight right shoulder muscles were collected using bipolar surface electromyography (EMG) (AMT-8, Bortec Biomedical Ltd., AB, Canada). Prior to electrode placement each site was prepared by shaving and scrubbing with isopropyl alcohol. Electrodes were placed over the mid belly of each muscle, parallel to the direction of the muscle fiber with an inter-electrode distance of 2.5 cm. The muscles monitored were the anterior (AD), middle (MD) and posterior (PD) deltoid, trapezius (TR), infraspinatus (IN), teres major (TM), latissimus dorsi (LD) and biceps brachii (BB). Maximum voluntary excitation (MVE) tests were performed for each muscle. Static MVEs for all muscles (except IN) were collected using a cable weight machine with a height adjustable handle, pinned such that the load was securely fixed to provide maximal resistance. The MVE tests were as follows: (i) AD, shoulder flexion exertion at 90° of shoulder flexion; (ii) MD, abduction exertion at 90° abduction; (iii) PD, static pull at 90° abduction with 90° elbow flexion; (iv) TR, static shoulder shrug; (v) LD, static pull, 45° abduction with 90° elbow flexion. For the IN, the MVE was obtained by a manually resisted external rotation at 90° shoulder abduction with 90° elbow flexion. A custom grip strain gauge dynamometer was constructed into the handle of the cable machine (MLT003/D, AD Instruments, CO, USA) and had a total mass of 1.47 kg. Three 5 s trials of maximal voluntary grip (MVG) force were measured with the elbow extended and arm at the side. The largest of the MVG trials was used to normalize the grip force. EMG was differentially amplified and band pass filtered between 10 and 1000 Hz (CMRR > 115 dB at 60 Hz, input impedance ~10 GΩ). EMG and grip force were sampled at 2048 Hz with custom Labview software, A/D converted (12 bit, PCIMIO-16E-4, National Instruments, TX, USA) and saved for processing.

2.3. Experimental protocol

Participants performed a series of static shoulder exertions, 6 with minimal grip and 15 with a 30% grip, for a total of 21 trials, each 10 s in duration. All trials were performed with the elbow extended holding the grip dynamometer with the forearm pronated

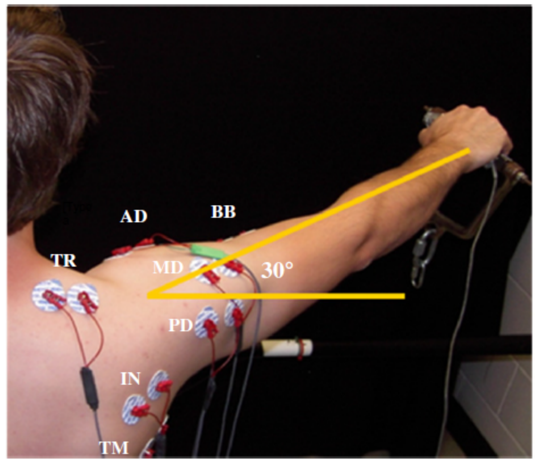

Fig. 1. Arm posture in scapular plane held for each of the 21 trials with custom grip dynamometer in hand. Electrode locations for anterior, middle and posterior deltoid (AD, MD and PD), biceps (BB), trapezius (TR), infraspinatus (INF) and teres major (TM) are shown. Latissimus dorsi (LD) not shown.

(palm down) in the scapular plane of the shoulder (90° abduction and anteriorly rotated 30°; Fig. 1). The first three trials acted as controls as participants simply maintained the experimental posture for 10 s while holding the grip dynamometer. No further instruction was given regarding amount of grip force to use beyond holding the apparatus, nor was feedback provided during what will be referred to as the "no grip" instruction trials. Participants then performed 15 trials in the same posture while simultaneously gripping at 30 ± 1.5% of maximum. Visual feedback was provided on a computer monitor using the real-time grip force with upper and lower boundaries formed by two horizontal lines at 28.5 and 31.5% MVG, respectively. Participants ramped up to the target grip force within the 10-s trial resulting in a minimum of 6 s of constant grip force. Following the 15 grip trials, three additional no grip control trials were completed. Thirty seconds of rest was given between trials.

2.4. Data analysis

Maximum, minimum, mean and within-trial standard deviations were calculated for grip force (% MVG) and muscle activity (% MVE) from a 6 s window of each trial. Within subject variability was also calculated for both grip force and EMG for each muscle for each trial. Due to the nature of the protocol and data, the 21 trials were analyzed using the mean of groups of three trials. Trials 1–3 and 19–21 represented the control or no grip trials, while trials 4–18 were divided into 5 sets of 3 gripping trials. A repeated measures analysis of variance (ANOVA) was performed to ensure that there were no statistically significant differences between the 3 trials in each group. The means, maximums, and standard deviations of seven sets of grip force and activity for all muscles were input into a repeated measures ANOVA to determine the effect of repetition ($\alpha = 0.05$). Significant effects were further examined using least significant difference (LSD) post hoc test.

3. Results

3.1. Grip force

The mean grip force during the 15 visually targeted gripping trials was 28.0 ± 0.1% MVG. The mean grip force during the control trials was 1.8 ± 0.6% (trials 1–3) and 1.5 ± 1.0% MVG (trials 19–21). Significant differences ($F_{6,60}$ = 5645.2, p = 0.00) were found between the no grip control trials (1–3, 19–21) and grip trials (4–18).

3.2. Muscle activity

Mean AD and LD muscle activity was significantly lower during most of the gripping trials than both sets of control trials (Fig. 2). Specifically, AD activity ($F_{6,60}$ = 4.2, p = 0.001) was lowest during 4th and 5th sets with grip (trials 13–18), with set 4 being 2.1% MVE (p = 0.03) lower than the initial no grip control period (trials 1–3) (Table 1). Mean LD activity was significantly lower during all gripping trials than control trials ($F_{6,60}$ = 7.6, p = 0.000), with decreases in LD activity ranging from 0.9 to 1.5% MVE when gripping with force (Fig. 2). Peak EMG followed the same trends as mean EMG, thus provides the same interpretation and will not be discussed.

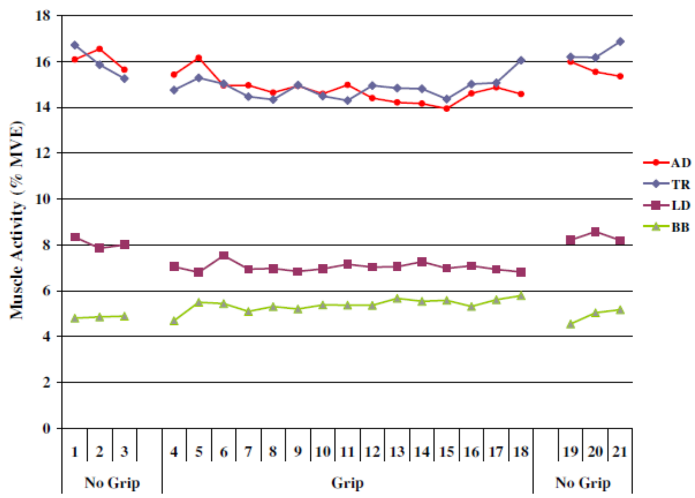

Figure 2. Mean EMG (% MVE) of anterior deltoid, trapezius, latissimus dorsi and biceps brachii muscles for all 21 trials.

Table 1
Mean grip force and AEMG (±standard deviation) as a percentage of maximum for each set of three trials (control, grip, control).

	Control (pre – no grip) (trials 1–3)	Grip 1 (trials 4–6)	Grip 2 (trials 7–9)	Grip 3 (trials 10–12)	Grip 4 (trials 13–15)	Grip 5 (trials 16–18)	Control (post – no grip) (trials 19–21)
Mean (% maximum)							
Grip	1.8 ± 0.6	27.9 ± 0.9[a,b]	28.0 ± 0.9[a,b]	28.0 ± 1.0[a,b]	28.1 ± 0.9[a,b]	28.0 ± 1.0[a,b]	1.5 ± 1.0
AD	17.2 ± 9.7	16.4 ± 10.2	15.9 ± 9.2[a,b]	15.7 ± 9.2[a,b]	15.1 ± 8.3[a,b]	15.8 ± 9.5[a,b]	16.8 ± 8.5
MD	15.7 ± 5.7	15.4 ± 5.6	15.5 ± 6.0	15.7 ± 6.0	15.6 ± 5.8	16.0 ± 6.3	16.1 ± 7.2
PD	10.3 ± 4.7	10.7 ± 4.2	11.1 ± 4.6	10.9 ± 4.6	10.8 ± 4.4	11.2 ± 5.2	10.7 ± 5.8
TR	16.9 ± 7.3	15.8 ± 7.7[b]	15.4 ± 8.2[b]	15.5 ± 8.2[b]	15.5 ± 8.1[b]	16.3 ± 8.6[b]	17.4 ± 9.3
IN	7.8 ± 4.4	7.7 ± 4.9	7.5 ± 4.8	7.3 ± 4.8	7.4 ± 4.6	7.5 ± 4.7	7.7 ± 4.1
TM	4.8 ± 4.5	4.5 ± 3.8	4.6 ± 4.1	4.6 ± 4.1	4.6 ± 4.1	4.5 ± 3.8	4.6 ± 3.8
LD	8.6 ± 4.6	7.7 ± 4.5[a,b]	7.3 ± 4.1[a,b]	7.5 ± 4.3[a,b]	7.5 ± 4.3[a,b]	7.4 ± 4.1[a,b]	8.8 ± 5.1
BB	5.2 ± 4.8	5.5 ± 3.9	5.6 ± 4.2	5.8 ± 4.5	6.0 ± 4.4	6.0 ± 4.6	5.3 ± 4.8

[a] Significantly different than "pre" control trials.
[b] Significantly different than "post" control trials (p < 0.05).

Table 2
Mean within-trial variability (standard deviation of each trial) for each set of three trials (control, grip, control).

	Control (pre – no grip) (trials 1–3)	Grip 1 (trials 4–6)	Grip 2 (trials 7–9)	Grip 3 (trials 10–12)	Grip 4 (trials 13–15)	Grip 5 (trials 16–18)	Control (post – no grip) (trials 19–21)
Within-trial variability (% maximum)							
Grip	0.14	0.96[a,b]	0.88[a,b]	0.75[a,b]	0.80[a,b]	0.81[a,b]	0.18
AD	2.69	2.09[a,b]	2.20[a,b]	2.11[a,b]	2.00[a,b]	2.24[a,b]	2.69
MD	2.17	1.86	1.89	1.84	1.88	1.92	2.46
PD	1.59	1.43[a,b]	1.55[a,b]	1.45[a,b]	1.48[a,b]	1.55[a,b]	1.77
TR	2.66	2.12[a,b]	2.15[a,b]	2.06[a,b]	2.09[a,b]	2.22	2.75
IN	0.95	0.77[a,b]	0.78[a,b]	0.74[a,b]	0.76[a,b]	0.81[a,b]	0.90
TM	0.58	0.47[a,b]	0.49[a,b]	0.49[a,b]	0.52[a,b]	0.47[a,b]	0.57
LD	1.44	1.10[a,b]	1.11[a,b]	1.17[a,b]	1.10[a,b]	1.15[a,b]	1.61
BB	0.95	0.88	0.97	0.93	1.05	1.08	1.05

[a] Significantly different than "pre" control trials.
[b] Significantly different than "post" control trials (p < 0.05).

Mean TR activity was also lower during the gripping trials ($F_{6,60} = 2.7$, $p = 0.021$) but post hoc evaluation revealed this was only significant when compared to the post-gripping control trials (Table 1). Biceps brachii activity appeared to increase marginally with successive gripping trials, but did not attain statistical significance (Fig. 2).

3.3. Within-trial variability of grip force and EMG

The within-trial variability of subject's grip force was significantly greater during the gripping trials ($F_{6,60} = 27.4$, $p = 0.000$), however the increase was relative to the greater grip force. Although not significant, grip force variability was lower in the last three sets of gripping (trials 10–18) than the first two sets of gripping trials (trials 4–9) (Table 2). The variability of AD, PD, TR, IN, TM and LD activity were all significantly lower with gripping than no grip trials (before or after grip trials) in all muscles except MD and BB (Table 2).

4. Discussion

In this study, the activity of many muscles changed when gripping was combined with maintaining shoulder posture, in spite of there being no additional external mechanical load at the shoulder. It was hypothesized that if the effect was due to mental effort associated with the visual grip force feedback it should decrease with exposure due to habituation or a learning effect. Gripping acted to decrease the activity of several muscles of the shoulder complex, significantly for the anterior deltoid (AD), trapezius (TR), and latissimus dorsi (LD). The response was not diminished over the exposure of 15 trials, suggesting a constant effect, most likely through muscular forces caused by gripping, or possibly due to processing feedback from the screen. The task in the current study presented physical and sensory demands similar to those experienced using a hand tool. The changes in muscle activity found in this study, although small, resulted from muscular or cognitive loads which have not been traditionally considered when assessing muscular demand at the shoulder and may be overlooked in assessing shoulder loading and risk of injury.

Anterior deltoid (AD), trapezius (TR), and latissimus dorsi (LD) activity were significantly lower than the control trials (pre- or post-gripping) in at least 4 of the 5 groups of gripping trials depending on the muscle. On average, the AD was 1.3% MVE lower with gripping, TR was 1.0% MVE lower and latissimus dorsi was 1.1% MVE lower during gripping. These findings are similar to previous studies that found that anterior and middle deltoids (in abduction) were 2% MVE lower with a 30% grip while maintaining a mid-abduction arm posture (Antony and Keir, 2009) or with slightly greater (40% maximum) shoulder effort (Au and Keir, 2007). The slightly lower effect and lack of change in middle deltoid activity in the current study is likely due in part to the use of the scapular plane rather than the pure abduction plane and perhaps a function of the small level of shoulder exertion. In previous studies that have shown similar effects of feedback modulated gripping, the degree to which muscle activity changes appears also to be dependent on the level of force that muscle would contribute to said posture. Both Sporrong et al. (1996) and Antony and Keir (2009) found greater changes to deltoid activity with gripping as arm angle increased from 30° to 90° elevation, which would typically increase muscle activity without gripping. One study found gripping to decrease anterior deltoid activity by 3–5% during low force pushes but it did not attain significance, due in part to the constraints of the task (Di Domizio and Keir, 2010).

Part of the impetus for the current study was an attempt to tease out the effects of additional muscular effort, in the form of a concurrent grip, from the potential cognitive effect of processing grip feedback. Cognitive load theory suggests that there is an increase in cognitive load while using working memory to learn a new task (Sweller, 1988). Many studies have demonstrated that the addition of cognitive tasks (in various forms) to an existing task increases the activity of specific muscles (Au and Keir, 2007; MacDonell and Keir, 2005; Visser et al., 2004; Finsen et al., 2001). These studies have increased memory demands (Finsen et al., 2001), added the Stroop colour-word test (MacDonell and Keir, 2005; Au and Keir, 2007), or used other forms of simple and complex processing (Davis et al., 2002). A recent study suggested that visual feedback may be acting in a similar manner, and it rather than grip force, may play a role in the changes in muscle activity. Smets et al. (2009) found that targeting a 30% grip force through visual feedback resulted in reduced strength and muscle activity even when the unconstrained grip force was higher than the constrained grip force in one condition. The current study used a somewhat indirect method to test whether gripping with visual feedback had an associated cognitive load by testing for a learning effect. By using this approach, we examined the variability that may be associated with training subjects to grip at 30% without feedback. Our results indicate that the effect of gripping with feedback on muscle activity was not lessened with repeated exposure (Table 1). Although it is not possible to dismiss potential interference of feedback for grip force on muscle activity, the trend toward lower within-trial variability during the later gripping trials provides some indication of improved performance with repetition, thus the lack of habituation may be due to other processes.

Given previous findings, gripping during shoulder efforts has been shown to cause a redistribution of muscular load from the deltoids to the rotator cuff (Sporrong et al., 1995, 1996; Antony and Keir, 2009; Au and Keir, 2007). Gripping has been shown to increase infraspinatus activity by 2% MVE under similar conditions to the current study (i.e. maintaining shoulder abduction) (Antony and Keir, 2009) and up to 6.7% MVE during maximal shoulder exertions (MacDonell and Keir, 2005). While the current study found no changes in either infraspinatus or teres major activity, there was a trend to slightly elevated biceps brachii activity during the gripping trials (up to 0.8% MVE). Interestingly, the small increase in biceps activity was similar (marginally smaller) than the significant decrease in anterior deltoid activity suggesting some form of offset or redistribution. Antony and Keir (2009) found gripping to increase biceps activity by up to 6% MVE in the shoulder flexion and mid-abduction planes but found much smaller changes in biceps activity in the abduction plane which used the same pronated forearm posture as the current study. The biceps crosses both the elbow and the glenohumeral joints, allowing it to flex and stabilize the shoulder while offsetting extrinsic forearm muscle forces at the elbow required for gripping. Thus the role of the biceps is important but was likely lessened by the use of the pronated forearm posture in the current protocol rather than in neutral or supinated forearm postures.

Co-activation of proximal and distal upper extremity muscles has been shown to occur during gripping. There are two plausible explanations for this co-activation. First, in primates, a single corticomotoneuronal cell has been shown to elicit a response from both proximal and distal muscles of upper limb (McKiernan et al., 1998; Fetz and Cheney, 1980). Thus, the motor command to generate a grip may elicit activity in the proximal shoulder muscles. Second, our previous work suggests that the multi-articular muscles of the arm play a role in transferring forces at the elbow and shoulder (Antony and Keir, 2009; Au and Keir, 2007; MacDonell and Keir, 2005). For example, generating a grip force activates the extrinsic flexors and extensors of the wrist and fingers which also cross the elbow creating forces and moments in three directions. These forces and moments are balanced (partially) by the biceps brachii which also acts at the shoulder, thus a completing a chain of musculoskeletal forces.

Our analysis also examined the variability within each trial as it has been recently shown that injury risk increased with a decrease in muscle activity variability in occupational tasks (Madeleine et al., 2008). We found small decreases, or no change, in muscle activity when gripping during shoulder exertions with lower within-trial variability (standard deviation during each 6 s window) for all muscles (significant for most)(Table 2). While traditional thinking suggests that, over time, small increases in muscle activity may accumulate and lead to chronic disorders in the workplace, recent research has associated decreased motor variability with upper extremity work-related disorders and pain (Madeleine et al., 2008). The current study found that muscles such as the anterior deltoid, latissimus dorsi and, likely most importantly, trapezius, may be susceptible to both increased activity and decreased variability. In the workplace, an additional task (gripping in this case) may further constrain the task, as seen by reduced variability in muscle activity, potentially increasing the risk of musculoskeletal disorder development.

There are several limitations to the current study. The number of participants was relatively small ($n = 11$) and the task was quite limited, thus generalizing the results to all populations and tasks should be done with caution. With a repeated trials protocol, fatigue can be a concern, however, we found no evidence of fatigue in pilot studies or in the current data (there were no significant differences in EMG amplitude between pre- and post-gripping control trials which is required, along with lower frequency content to indicate fatigue; De Luca, 1984). While an original goal of the study was to parse out the effects of gripping versus feedback through repeated exposure, the number of repetitions (15 trials) may not have provided enough of practice for habituation to occur. Redistribution of force from AD, TR and LD muscles to others was not apparent in the current study. However, we were limited to using surface EMG, thus only the infraspinatus rotator cuff was accessible with a reasonable certainty. In previous studies, the greatest reaction has been in the supraspinatus which has not been reliably accessed with surface EMG. Indwelling EMG may have revealed distribution to the other rotator cuff muscles.

5. Conclusion

This study found that changes in muscle activity during shoulder exertions concurrent with gripping persisted despite repeated exposures, indicating that there was not a learning effect on visually targeted gripping. More interestingly, the gripping task reduced variability in all muscles potentially forcing a change in control strategy for the shoulder complex and perhaps limiting options in control. Further research is needed to examine reduced variability as a control mechanism and separate the effect of attention to visual targeting versus a secondary task such as gripping. Understanding shoulder muscle coordination through a range of functional tasks is integral to the comprehension of injury mechanisms.

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CHAPTER 3

Study 2: Normalization of shoulder electromyography for use in dynamic contractions

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3.1 Abstract

Muscle specific maximal voluntary isometric contractions are commonly used to provide reference amplitudes to normalize electromyographic signals (EMG). However, it has been questioned whether this process is appropriate for normalizing EMG activity obtained during dynamic contractions. The purpose of this study was to investigate four methods of determining maximal muscle excitation and their impact on normalized upper extremity EMG during dynamic contractions of the shoulder muscles. The muscles monitored were the anterior, middle and posterior deltoid, upper and lower trapezius, pectoralis major, latissimus dorsi and infraspinatus. The four methods used to identify maximal muscle excitations were: i) from a muscle specific manually resisted maximal voluntary isometric contraction; ii) from any manually resisted contraction in (i); iii) from maximal dynamic isokinetic contractions during flexion-extension, abduction-adduction and internal-external rotation movements; and iv) from any contraction from (i-iii), as well as maximal performance of the task of turning an isokinetic wheel in both directions. When EMG from the wheel trials were normalized to muscle specific isometric contractions (method i), mean muscle activity was 45% greater than when normalized to maximal experimentally collected excitations (method iv). In seventy-five percent of cases, maximal excitation was identified during maximal dynamic contractions (methods iii and iv). These data strongly suggest that EMG obtained during maximal dynamic effort should be considered when normalizing EMG from dynamic conditions.

Keywords: EMG, normalization, shoulder, dynamic contractions

3.2 Introduction

Electromyographic (EMG) signals are influenced by many technical, anatomical and physiological factors. To account for many of these factors, reference contractions are typically used to normalize the EMG signal and allow comparison, between individuals, testing sessions or at any point when electrodes have been reapplied. Reference contractions are typically maximal efforts and thus the recorded muscle activity is expected to reflect 100% of the muscle's capacity (DeLuca et al, 1997; Kumar et al, 1996).

Paramount to the proper interpretation of EMG signal is that a true maximum activation be found for each muscle and this value is used for normalization. Without a true maximum, normalized EMG will be inflated and will thus misrepresent the level of muscular effort being used during the experimental condition (Lehman and McGill, 1999). The issue of proper normalization is of particular concern when investigating the shoulder musculature. In healthy individuals, low level rotator cuff activity is expected during many tasks as these muscles are used to maintain joint integrity (Inman et al, 1994). However, inconsistencies in determining maximum activity of shoulder muscles have resulted in concerns about the interpretation of shoulder muscle activity and coordination.

Methods of obtaining maximal muscle activation are described by a number of similar terms, which can hinder interpretation and comparison of data. The predominant method of obtaining a normalization factor is to perform a muscle specific maximal voluntary isometric contraction, referred to as either a MVC (Anders et al, 2005; Morris et al, 1998) or more specifically, a MVIC (Sparkes and Behm, 2010; Burnett et al, 2006; McLean et al, 2003). MVICs are static contractions typically performed at a specified mid-range joint angle to optimize the muscle's force length relationship (Ball and Scurr, 2010; Chapman et al, 2010; Rouffet and Hautier, 2008; Netto and Burnett, 2006; Hunter et al, 2002; Burden & Bartlett, 1999; Morris et al, 1998). MVICs are commonly performed against manual resistance provided by an experimenter or clinician. Other methods of providing resistance include cable systems and isokinetic dynamometers in static mode (Hodder and Keir, *in press* Netto and Burnett, 2006; Hunter et al, 2002). EMG signals normalized to these contractions are often expressed in % MVIC. However, this can be misinterpreted as a percent of maximum force or torque and thus EMG has also been expressed in terms of % "MVE" to avoid confusion. Some define MVE as maximal voluntary effort (Shirasawa et al, 2009) or exertion (Fischer et al, 2010; Granata and Gottipati, 2008; Thelan et al, 1994), while others refer more specifically to maximal voluntary electrical activity (Madill and McLean, 2006; Mogk and Keir, 2006; Balogh et al, 1999) or excitation (Hodder & Keir, submitted; Chopp et al, 2010; Meyers and Keir, 2003). Thus, MVE is used to distinguish the greatest activity for a desired muscle from a contraction producing maximal force or torque. In the current communication, we will differentiate between the contraction used to elicit maximum excitation (e.g. MVIC) and the maximum voluntary excitation itself (MVE).

Despite the wide use of isometric contractions to elicit MVE for normalizing muscle activity, many studies have found that normalizing EMG from dynamic contractions to an isometric MVE yields values greater than 100% (e.g. Decker et al, 1999; Morris et al, 1998; McGill and Sharratt, 1990; Jobe et al 1984; Clarys et al, 1983). Normalized activity in excess of 160% MVIC has been reported for of the triceps brachii during swimming (Clarys et al., 1983). Serratus anterior activity in excess of 226% MVIC was reported during baseball pitching (Jobe et al., 1984), and rotator cuff muscle activity exceeding 300% MVIC has been reported during common rehabilitation exercises (Morris et al., 1998). In these studies, the absence of a true maximum reference value is clear, as reported EMG exceeded 100%. However, there is cause for concern that others may have over-reported EMG due to inaccurate reference values, and yet it has gone undetected since values exceeding 100% were not reported.

To minimize normalization error, dynamic reference contractions have been used to improve the possibility of obtaining the greatest muscle excitation. Some studies have selected to use a muscle specific dynamic movement to obtain their reference excitation (Ball and Scurr, 2010; Rouffet and Hautier, 2007; Kyrolainen et al., 2005), while others have used the excitation level obtained during maximal performance of the actual experimental task (Ball and Scurr, 2010; Rouffet and Hautier, 2007; Kyrolainen et al., 2005; Arampatzis et al., 2001; Morris et al, 1998). It is important to determine which reference contractions are best suited to obtain maximal muscle excitation as factors such as muscle fatigue, study duration and resource availability will limit the ability to perform a wide range of maximal contractions.

The purpose of this study was to evaluate the effects of several methods of obtaining maximal muscle excitation and determine which contraction was best at achieving the highest excitation level. Normalizing muscle activity to its maximum excitation is an important factor in correctly interpreting the capacity of the muscle being used. In this study we compare maximal excitations obtained from conventional manually resisted maximal voluntary isometric contractions (MVIC), to those obtained during maximum voluntary dynamic concentric contractions (MVDC) using an isokinetic dynamometer, as well as those elicited during any contraction performed during the experiment, including maximal effort during the experimental task itself.

3.3 Methods

3.3.1 Participants

Twelve healthy males (176.2 ± 9.1 cm; 76.3 ± 11.9 kg; 22.0 ± 1.5 years) who were free of shoulder pain in the last year were recruited from the university population. This study was approved by the Human Research Ethics board at McMaster University. All participants provided written informed consent prior to participating in the study.

3.3.2 Surface EMG

Bipolar surface EMG was collected from eight muscles of the shoulder and trunk on the right side. The muscles monitored were the anterior (AD), middle (MD) and posterior (PD) deltoid, upper trapezius (UT), lower trapezius (LT), pectoralis major (PM), latissimus dorsi (LD) and infraspinatus (INF). Prior to electrode placement, each site was prepared by shaving and scrubbing with isopropyl alcohol. Disposable Ag-AgCl electrodes were placed over the belly of each muscle, parallel to the fibre direction with a centre-to-centre electrode distance of 3 cm (Kendall MediTrace 130, Mansfield, MA, USA). Electrode placements were confirmed with muscle specific contractions, including manually resisted shoulder flexion (AD), abduction (MD) and extension (PD), shoulder shrug (UT), scapular retraction (LT), cross flexion (PM) and adduction (LD). EMG was differentially amplified and band pass filtered between 10 and 1000 Hz (CMRR > 115 dB at 60 Hz, input impedance ~ 10GΩ; AMT-8, Bortec Biomedical Ltd., AB, Canada). EMG was sampled at 2048 Hz (12 bit, USB-6259, National Instruments, TX, USA). All EMG data were full wave rectified and filtered with a critically damped dual low pass filter with a 3 Hz cutoff.

3.3.3 Reference Contractions and Experimental Design

Participants performed two types of reference contractions: 1) manually resisted muscle specific maximum voluntary isometric contractions (MVIC) and 2) maximum voluntary dynamic concentric contractions (MVDC) on an isokinetic dynamometer (Biodex 4, Biodex Medical Systems, New York, USA). Participants then performed the experimental task, a bidirectional concentric wheel rotation with maximal effort.

1) *Muscle Specific Maximum Voluntary Isometric Contractions (MVIC).* For each muscle, an MVIC was performed against resistance provided by the investigator to determine the maximal voluntary isometric excitation (MVIE). The tests for each muscle were as follows: (i) AD, straight arm shoulder flexion at 45° of elevation, resistance at the wrist, (ii) MD, straight arm shoulder abduction at 45° of elevation, resistance at wrist, (iii) PD, bent arm extension with arm abducted to 90° and elbow bent 90° and forearm rotated anteriorly, resistance applied proximal to the elbow; (iv) UT, shoulder shrug with resistance provided by grasping participants hand and wrist and downward resistance, (v) LT, scapular retraction with arm abducted 90° and elbow bent 90°, resistance applied

against the arm just distal to the shoulder as the participant exerted force posteriorly, squeezing their scapula together, (vi) PM, horizontal adduction (also referred to as cross flexion) with arm flexed to 90° and elbow bent 90°, resistance applied proximal to the elbow, (vii) LD, arm adduction exertion with arm elevated to 90° with elbow flexed to 90° and rotated externally so the hand was superior to the upper arm, resistance from below applied just proximal to the elbow as participant pushed downwards; (viii) INF, external humeral rotation exertion with arm at side and 90° elbow flexion, resistance was provided proximal to the wrist. Participants stood while performing all maximal exertions. They were instructed to use a broad stance and to brace themselves against a solid table top using their left hand. In this posture they were most capable of counterbalancing the contractions without hesitating to exert maximal effort. Exertions were held for 5 seconds each and repeated twice. One minute of rest was given between each contraction. The highest activity found during the two repetitions was defined as the muscle specific maximum voluntary isometric excitation (MVIE). The highest excitation for each muscle from any MVIC test was defined as $MVIE_{ALL}$.

2) Maximal Voluntary Dynamic Concentric Contractions (MVDC). Participants were positioned in the dynamometer using the upper extremity attachment following manufacturer guidelines. A previous study found little difference between muscle activity collected during concentric contractions at 30°/s, 90°/s and 120°/s (Kellis & Baltzopoulos, 1996). For the current study, it was decided that maximal concentric isokinetic contractions would be performed at the more controlled velocity of 30°/s. The contractions were performed as follows: (i) Flexion-Extension: straight arm flexion from

0 - 90° (forward arm raise from arm at the side to 90° horizontal) followed immediately by extension down to 0°, intended to target AD and PD, respectively; (ii) Abduction-Adduction: straight arm raise (abduction) from 0 - 90°, immediately followed by adduction to 0°, intended to target MD and LD, respectively; (iii) External-Internal Rotation: external (humeral) rotation at 45° of abduction and 90° elbow flexion from 0° to maximum comfortable external rotation (approximately 75°) intended to target the INF, immediately followed by internal rotation back to centre (0°). Muscle specific isokinetic contractions were performed for the AD, MD, PD, LD and INF. Each paired movement (flexion/extension, abduction/adduction and external/internal rotation) was repeated three times. The greatest excitation for each muscle from all concentric contractions was defined as the maximum voluntary dynamic excitation (MVDE). Optimal motions for the UT, LT and PM were not readily available on the dynamometer.

3) Experimental Task: A bi-directional wheel rotation task was performed under concentric conditions to represent a non-specific task that would activate most muscles tested (Figure 3.1). The upper extremity wheel attachment from the work simulation toolkit of the Biodex 4 System was affixed, tilted to 45° from horizontal and centered in front of the participant. The participant abducted their arms to 45° and flexed their elbows to 45°, as measured by a manual goniometer. While maintaining this posture, the height of the wheel and seat were adjusted such that the participant's hands were in the 10 and 2 o'clock positions (Figure 3.1b). The start position of the wheel turn was determined by placing the hands at the 10 and 2 o'clock positions and rotating the wheel counter-clockwise (CCW) until the right hand reached the 12 o'clock position (Figure

3.1a). Participants then rotated the wheel clockwise (CW) approximately 230° until the right hand reached just past 6 o'clock (Figure 3.1c) and returned by rotating CCW to the starting position (Figure 3.1a). The range of velocities of occupational wheel use has been reported to be between 12 and 1200°/s, depending on the type of wheel (Helsen, 1949). Thus, three cycles were performed with maximal effort at 60°/s and 120°/s (as constrained by the dynamometer) as they fell within both the operating speed of the dynamometer and the range of occupational wheel use.

a) b) c)

Figure 3.1: Wheel work simulation tool in the a) start position, b) middle or 10 and 2 o'clock position, and c) end position. The movement was performed continuously, rotating clockwise and returning counter-clockwise.

3.3.4 Data Analysis

EMG data from the wheel rotation task were separated into clockwise (CW) and counter clockwise (CCW) directions and clipped from 5 to 230° of rotation in each direction. EMG values from each muscle collected during the wheel rotations were then normalized using 4 reference values: 1) muscle specific MVIE, 2) $MVIE_{ALL}$, 3) MVDE, and 4) the maximum experimental excitation (MEE), which was the maximum excitation when MVIE, $MVIE_{ALL}$, MVDE and wheel rotation contractions were considered. For each participant, the contraction that elicited the greatest overall excitation (MEE) was identified for each muscle. If this occurred during a MVDC, the angle at which peak activity occurred was also documented.

For the purposes of this communication, only data from the 120°/s wheel trials were analyzed. The data from the 60°/s wheel trials yielded the same interpretation, but the data from the 120°/s trials were larger in magnitude and better served the purpose of this communication. Muscle activity from the wheel rotations were normalized to each of the four reference values. Peak muscle activity was found for both the wheel rotation data during CW and CCW rotations and normalized using each method. Mean peak muscle activity for each method of normalization was calculated. Between-subject variability was calculated as standard deviation between peak activations for each subject under each condition. An analysis of variance (ANOVA) with a 95% confidence interval was used to compare the differences in peak muscle activity from the wheel rotations when normalized to each of the four reference values. Main effects were further evaluated with a Bonferroni corrected post hoc t-test. Additionally, muscle activity normalized to each reference value was re-sampled to every 0.5° and ensemble averaged across subjects. Ensemble averages were graphed to qualitatively assess the impact that each of the four reference values had on the shape and magnitude of muscle activity over the entire range of motion of the wheel rotations.

3.4 Results

3.4.1 Contraction type and maximal excitation

Maximal experimental excitations (MEE) were most often found in the MVDC trials and not in the muscle specific MVIC trials. The abduction MVDC was most consistent at eliciting peak excitation from MD, with 83% (10/12) of participants achieving maximal experimental excitation (MEE) for the MD during this trial (Table 3.1). The mean angle (± standard deviation) at which the MEE occurred was 71.0° ± 20.4º (Figure 3.2). The abduction MVDC also elicited the largest responses from UT and LT for 58% and 50% of the participants, respectively. The mean angle at which MEE was obtained for the UT was 63.2° ± 21.5°, however ranged from 29.0° to 86.2°. For LT, the mean angle at which MEE was found was 68.3° ± 25.2° and ranged from 26.9° to 90.0° (Figure 3.2).

Interestingly, the greatest activity for AD was split between the flexion (50% at 48.1 ± 26.1º) and abduction (30% at 75.9 ± 6.3°) MVDC trials, with the remaining 20% being achieved during the MVIC trials. The flexion angles at which the greatest AD activity was found spanned almost 60°, ranging from 13.8° to 73.0°. For PD and PM, the majority of participants achieved the greatest excitations during the bi-directional wheel task (Table 3.1), while the contraction that elicited the greatest INF activation was dispersed across the MVIC and MVDC trials (Table 3.1).

Table 3.1: Summary of the contractions in which the maximal experimental excitations (MEE) were found. The numbers represent how many participants were able elicit the MEE during said contraction. Muscles investigated were the anterior (AD), middle (MD) and posterior (PD) deltoids, the upper (UT) and lower (LT) trapezius, pectoralis major (PM), latissimus dorsi (LD) and infraspinatus (INF).

Maximal Experimental Excitation Contraction (# of participants)

	Specific MVIC	*Other MVIC*	*MVDC Flexion*	*MVDC Extension*	*MVDC Abduction*	*MVDC Adduction*	*MVDC External*	*MVDC Internal*	*MVDC Wheel*	*Total (n=12)*
AD	2		6		4					12
MD	1		1		10					12
PD	1	1	1		1		3		5	12
UT	2	2	1		7					12
LT		2			6		4			12
PM	2				1			2	7	12
LD	3	5		4						12
INF	2	1	4		3		2			12
Total	13	11	13	4	32	0	9	2	12	*96*

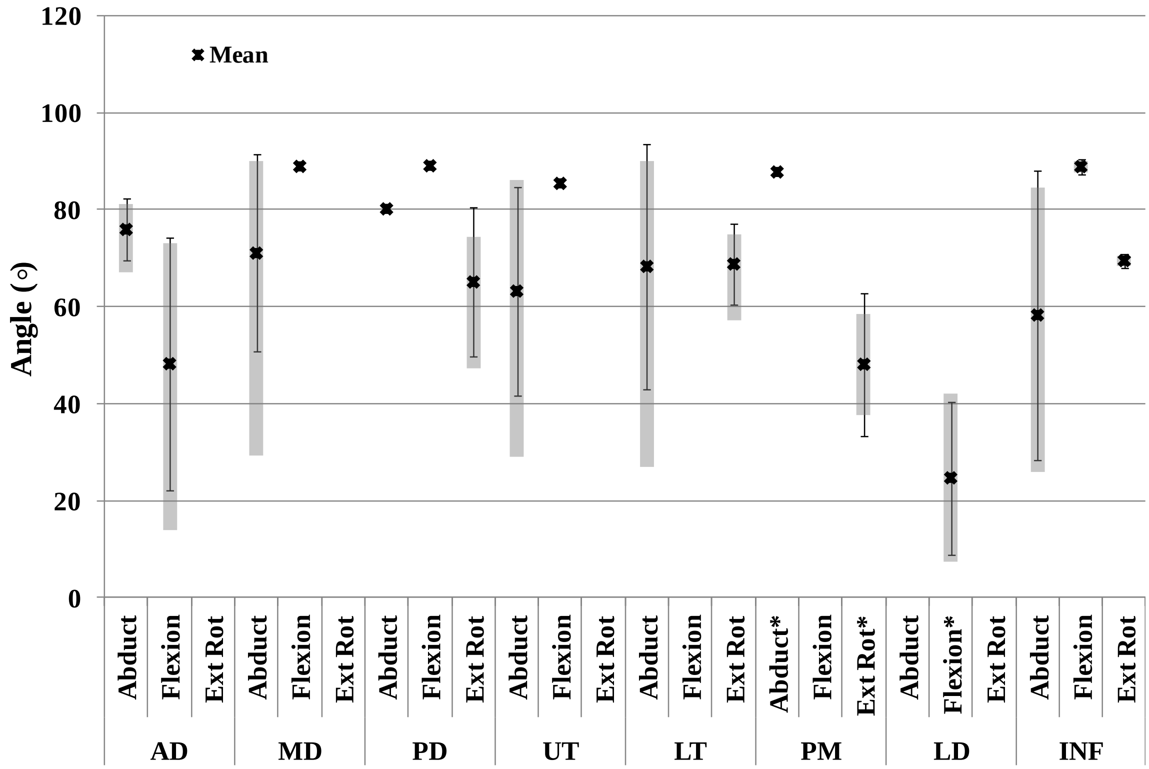

Figure 3.2: Mean angle for each muscle (as indicated by the "x") in which the maximum experimental excitation (MEE) was found during dynamic concentric contractions. Grey bars represent the range of angles from largest to smallest in which MEE values were found across all subjects. The black lines with capped ends represent the calculated standard deviation in the mean angle between subjects. The absence of grey bars and capped black lines signifies that only one subject had an MEE identified during that contraction. Note that * in the x axis labels indicate that direction of movement changes to extension for LD and adduction and internal rotation for PM.

3.4.2 Effect of reference value on normalized EMG

Peak muscle activity and between subject variability (SD) of EMG data normalized to MVIE, MVIE ALL, MVDE and MEE are summarized in Table 3.2. Overall, the greatest differences were found between MEE and MVIE. The standard deviations of MVIE normalized data were 1.4 to 1.8 times larger than those found when data were normalized to MEE (Table 3.2). For both CW and CCW wheel rotations, the effect of the normalization value on peak EMG was largest for the PM and MD. During CW rotation, peak PM activity was 102.7 ± 55.2% MVIE but only 62.9 ± 18.3% when normalized to MEE ($F_{(3,30)} = 3.5$, $p = 0.03$; Table 3.2). During CCW rotations, peak MD activity was 34.5 ± 13.3% MEE which was significantly lower than 81.7 ± 85.3% MVIE ($F_{(3,33)} = 19.2$, $p < 0.0000$). Additionally, during CCW rotations, PM activity was only 49.2 ± 22.1% MEE, yet was 80.1 ± 42.7% MVIE ($F_{(3,33)} = 4.8$, $p < 0.007$). Differences between the MVDE and MEE normalized data were small and often not significantly different (Table 3.2).

Ensemble averages of muscle activity normalized to each of the four reference values are profiled in Figure 3.3. In all participants, the AD specific MVIC was better than any other muscle's MVIC at targeting AD activity, hence the absence of an $MVIE_{ALL}$ line in Figure 3.3. MD was the only muscle in which the MEE was always found during the MVDC trials and hence the absence of a MVDE line in Figure 3.3. Notably, in all muscles the activity was always lowest when normalized to MEE, followed by MVDE and $MVIE_{ALL,}$ and always highest when normalized to MVIE. Additionally, the ensemble average curve for the MVIE normalized muscle activity appears to have exaggerated peaks compared to MVDE or MEE normalized activity. Difference in MVIE and MEE curves were as large as 40% (MD and INF in CCW).

Table 3.2: Mean (and SD) of peak EMG collected during wheel simulation trials as normalized to MVIE, $MVIE_{ALL}$, MVDE and MEE. Main effects were further examined with a Bonferroni corrected post hoc t-test and significant differences between activity when normalized to each reference value have been indicated as follows: [a] indicates a difference between MVIE and $MVIE_{ALL}$, [b] between MVIE and MVDE, [c] between MVIE and MEE, [d] difference between $MVIE_{ALL}$ and MVDE, [e] between $MVIE_{ALL}$ and MEE, and [f] between MVDE and MEE.

Normalized Muscle Activity

	Wheel									
	Counter Clockwise (CCW)					Clockwise (CW)				
	% MVIE	*%MVIE ALL*	*% MVDE*	*%MEE*	*SIG DIFF*	*% MVIE*	*%MVIE ALL*	*% MVDE*	*%MEE*	*SIG DIFF*
AD	54.5 (20.2)	51.9 (19.4)	38.2 (13.3)	37.3 (13.5)	b, c	49.1 (20.8)	47.7 (21.0)	35.0 (14.4)	34.0 (14.2)	c, e
MD	57.9 (23.0)	53.1 (18.9)	34.7 (13.4)	34.5 (13.3)	b, c, d, e	49.4 (29.1)	46.6 (29.9)	32.1 (24.1)	31.7 (23.5)	b, c, d, e
PD	40.8 (20.5)	34.4 (20.1)	32.5 (21.6)	25.7 (12.7)		65.9 (24.4)	61.9 (28.0)	58.7 (29.8)	46.6 (19.7)	c, e
UT	45.0 (37.2)	37.0 (33.2)	29.9 (16.4)	26.4 (15.4)		38.4 (44.9)	32.9 (45.7)	25.1 (20.0)	22.2 (19.7)	
LT	38.1 (15.8)	31.9 (16.8)	23.0 (10.8)	22.3 (10.9)	a, b, c, e	42.0 (22.3)	35.7 (23.8)	26.8 (15.9)	25.6 (15.6)	b, c
PM	80.1 (42.7)	77.0 (41.2)	80.5 (57.0)	49.2 (22.1)	c, e	102.7 (55.2)	100.2 (52.7)	111.21 (101.3)	62.9 (18.3)	c, e
LD	14.9 (8.9)	12.3 (8.8)	14.0 (8.6)	11.2 (6.5)		43.4 (31.3)	38.1 (23.4)	40.6 (20.3)	35.1 (20.1)	
INF	75.2 (61.3)	52.1 (33.2)	43.5 (25.8)	41.5 (24.1)		65.0 (43.9)	45.9 (22.5)	39.7 (21.5)	37.4 (18.3)	

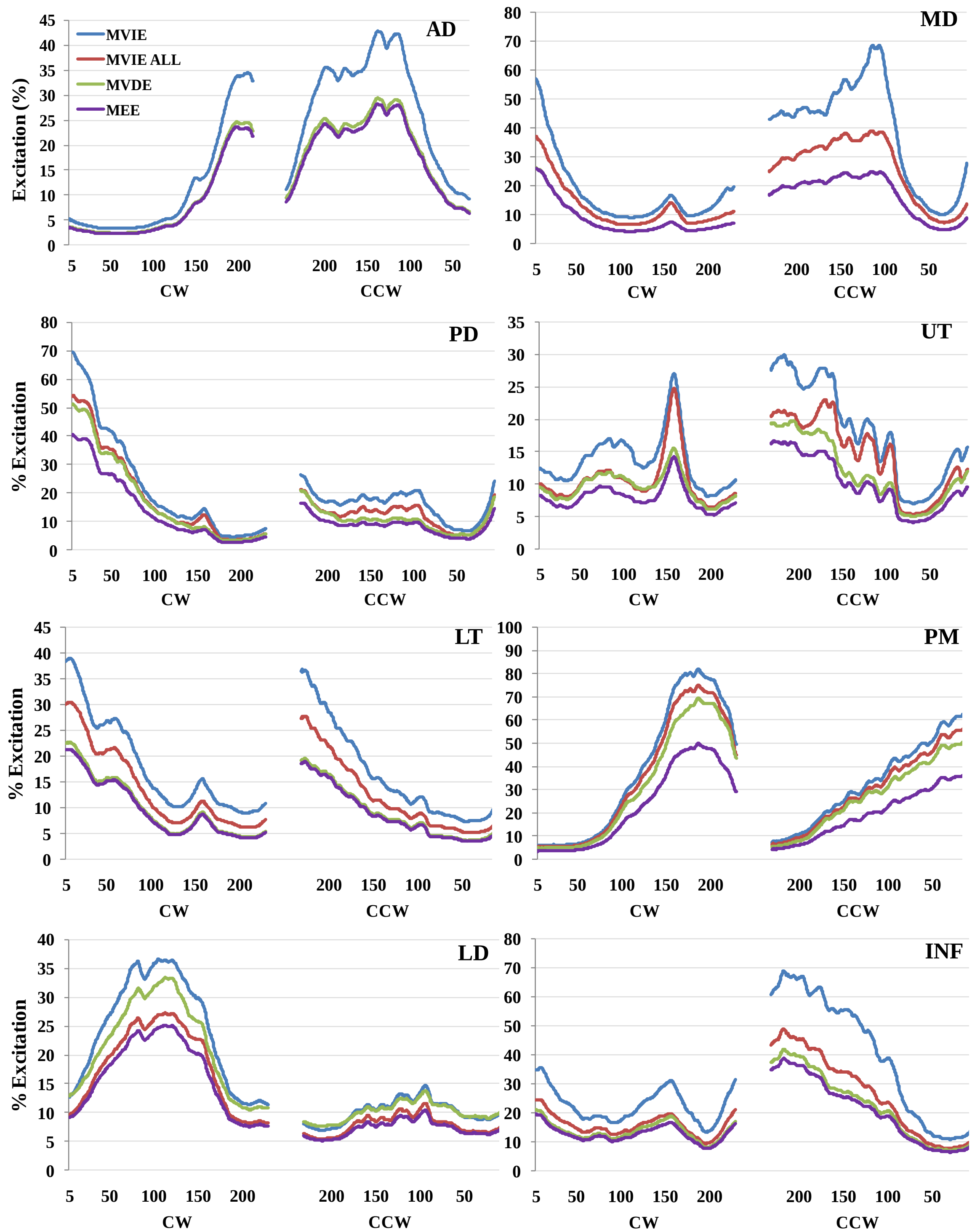

Figure 3.3: Ensemble average of muscle activity normalized to each reference value during the wheel rotation task in both clockwise (CW) and counter clockwise (CCW) directions.

3.5 Discussion

The purpose of this study was to improve estimates of maximal muscle activity used for normalizing EMG during dynamic contractions of the shoulder muscles. When EMG collected during a wheel task was normalized to a conventional MVIE, muscle activity was overestimated by as much as 74% when compared to the MEE method outlined in this study (Table 3.2). Another important finding was that the variability between subjects was more than halved when peak EMG was normalized to MEE (24.1 %) versus MVIE (61.3%). In most muscles, MEE normalized EMG was significantly lower than when normalized to either $MVIE_{ALL}$ or muscle specific MVIE. The MEE reference values were most often elicited (in 75% of participants) during dynamic contractions of either the experimental task or, most frequently, during the MVDC trials.

For AD, MD, UT, LT and INF, the angle in which peak excitation was achieved during the MVDC trials varied greatly across subjects, spanning over 60° of the range of motion (Figure 3.2). In part, this may explain why there are difficulties in achieving consistent maximal excitations with MVICs at a pre-selected angle. While the impact of lower reference signals, such as those from MVIE and $MVIE_{ALL}$, may not be obvious at low activity levels, the difference was more noticeable when muscle activity was greater or fluctuated rapidly. Visually, the impact of inadequate normalization values can be seen in Figure 3.3. The changes in gain when MVIE and $MVIE_{ALL}$ are used rather than MVDE or MEE make the slopes appear to be steeper and peaks higher. This is particularly noticeable in MD activity when it increases at 100° of CCW wheel rotation (Figure 3.3). The by-products of normalizing to a smaller denominator are the overestimation of the magnitude and exaggeration of the rate of change in muscle activity (Figure 3.3). Results from the current study strongly suggest that maximal voluntary dynamic contractions are a more appropriate means of obtaining true maximal activity when normalizing dynamic EMG.

This study was prompted by pilot data where EMG collected during a dynamic contraction was over 100% when normalized to conventional MVIE, as seen for the PM during CW wheel trial (Table 3.2). The literature is wrought with reports of EMG surpassing 100% of “maximal” activity, at times reaching 150-300% (Decker et al, 1999; Morris et al, 1998; Jobe et al, 1984; Clarys et al, 1983). When examining of intra-abdominal pressure and trunk EMG, McGill and Sharratt (1990) reported peak rectus abdominus activity as high as 127% during a sit up exercise, but simply commented that values over 100% of MVIC are commonplace during dynamic contractions. Theoretically, these muscle activities should be capped at 100% if a true maximum was obtained and used for normalization. Expressing muscle activity as a percentage of a true maximum is imperative to correctly interpreting the capacity at which each muscle is being used during dynamic contractions. As well, when assessing muscle coordination, one must be confident that 100% activity has the same meaning for each muscle. If some muscles are normalized to a true maximum, while others are normalized to sub-maximal excitation levels, whether due to a poor effort or a poor test, interpretation of the relative activity of each muscle will likely be misunderstood.

EMG normalized to excitations found during dynamic contractions has seen previous success. A recent review of EMG normalization techniques identified two methods of obtaining maximal excitation for normalizing dynamic contractions (Burden, 2010). One was to obtain the peak excitation from within the experimental trial itself (Ball and Scurr, 2010; Rouffet and Hautier, 2007; Kyrolainen et al., 2005; Arampatzis et al., 2001; Morris et al, 1998), and the other was to use the peak found during maximal isokinetic contractions targeted at each muscle being investigated (Ball and Scurr, 2010; Rouffet and Hautier, 2007; Kyrolainen et al., 2005). In a gait study, Burden et al. (2003) found greater inter-individual reliability when quadriceps and hamstring EMG was normalized to peak isokinetic concentric flexion-extension activity than isometric contractions. At the shoulder, Morris et al. (1998) found supraspinatus activity to be significantly lower when normalized to peak excitation from their experimental internal-external rotation trial than traditional MVIC methods. In addition, Morris et al (1998) also reported the coefficient of variation to be lower when normalizing to the experimental task, indicating a more repeatable signal.

Further support for normalization to excitation values obtained during dynamic contractions was presented by Rouffet and Hautier (2007). Much like the current study, they found that the tasks most similar to their sub-maximal cycling experimental condition, a maximal effort cycle sprint, elicited greater amplitude EMG. When used to identify peak reference muscle activity for normalization, lower variability was found than when the MVIC method was used. Ball and Scurr (2010) used a repeated measures design to determine the type of contraction that best produced maximum excitation values in lower limb muscles over three days of testing. They found that a squat jump was the most reliable (coefficient of variation < 5%) method of eliciting maximal excitation from the gastrocnemii and the soleus across all three days, and thus recommended its use over more traditional MVIC methods.

It has been proposed that a reference contraction in which to normalize EMG may be considered valid if it meets the following criteria: i) produces repeatable and reliable EMG measurements, ii) is relevant to the experimental task, iii) is feasible to collect, iv) allows for better estimations of the proportion of muscle capacity being used, and, v) accurately reflects the true variability of the non-normalized EMG pattern (Yang and Winter, 1984; Burden et al 2003). While the current study cannot make claim to all of the above conditions, the maximal experimental excitation (MEE) approach appears to be an improvement over typical MVIC methods. Specifically, EMG normalized to MEE provided better estimates of the proportion of muscle capacity used, thus did not inflate changes in activity seen when EMG was normalized to conventional MVIE methods, as demonstrated in Figure 3.3. The maximal experimental excitation (MEE) determined in this paper was unique compared to previous dynamic methods as it included the maximal voluntary dynamic (isokinetic) contractions and maximum voluntary isometric contractions, both aimed at the muscles being investigated, as well as the peak EMG from within the experimental task itself. Thus, MEE values identified in the current study are the most accurate representation of peak activity from each muscle.

There are a few limitations to the current study. It is recognized that the MVDC used in the current study targeted the predominant shoulder flexors, abductors and external rotators (AD, MD, PD and INF), while other supporting musculature, such as the PM and LD, were not as well activated by the selected dynamic contractions. So although the isokinetic dynamometer provided superior maximal values for many muscles targeted, especially for AD, MD and PD, it was limited in its ability to test all muscles in their primary mode of function. However, using the same dynamometer, it was found that the maximal bi-directional wheel was more effective at eliciting peak activity from the PM and LD muscles than the other testing modes. MVIC were also limited by being performed in a single posture versus the full range of motion for the dynamic contractions. Performing MVICs in numerous postures (as afforded by dynamic contractions) would not be feasible while also avoiding fatigue. Thus while maximal excitation may be obtained using an MVIC, it appears to be more effective to with MVDCs.

3.6 Conclusions

This study found that maximal voluntary dynamic contractions, either in the form of muscle specific tasks or maximal performance of the experimental task, were more effective at eliciting maximal excitations than maximal voluntary isometric contractions. During maximal dynamic contractions, peak activity was achieved over a wide range of angles and likely explains why maximal isometric contractions performed at a single angle were poor at eliciting maximal activity across all participants. Without proper maximal reference values, inconsistencies in the reported normalized muscle activity can occur, resulting in misinterpreting the capacity of muscle’s that are being used. While it is recognized that the manner in which the maximal dynamic contractions were performed in this paper are not always feasible, when investigating dynamic tasks their use is recommended whenever possible. Otherwise, maximal performance of the experimental task or performance of the muscle specific MVIC's at multiple angles in the midrange, will improve the probability of obtaining maximum excitations from the muscles being investigated.

3.7 Acknowledgements

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CHAPTER 4

Study 3: The effects of additional constraints on muscle activity and force distribution during static shoulder efforts

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4.1 Abstract

Shoulder muscle activity has been shown to change when additional hand tasks such as gripping are performed during shoulder efforts. The response of the rotator cuff muscles has been varied and recent evidence suggests that the biceps brachii may play a role. The purpose of this study was to examine the neuromuscular response of the shoulder to gripping and biceps contractions during static shoulder efforts in the sagittal and scapular planes. The activity of 8 shoulder muscles (including infraspinatus and supraspinatus activity via intramuscular electrodes) was recorded from 10 healthy males performing submaximal static shoulder efforts at 30°, 60° and 90° elevation in both sagittal and scapular planes. Shoulder moments at 40% of maximum were performed in neutral and supinated forearm postures, alone and in combination with a 30% of maximum grip and a 30% of maximum isometric biceps contraction. To assess the effects of these tasks on the shoulder, muscle activity was also used to compute three dimensional muscle forces in each posture. Significant differences in shoulder muscle activity were found with concurrent gripping and biceps contraction in the scapular plane with neutral forearm posture. However, the two load conditions did not elicit the same reaction. The lack of significance in many conditions may have been the result of large inter-individual variability, likely due to the numerous degrees of freedom in the upper extremity. While it was hypothesized that gripping and biceps contractions would change rotator cuff activity, this was not found due to the large variance in muscle forces between individuals. Each individual generated consistent patterns across conditions.

Keywords: Neuromuscular Control, Rotator Cuff, Fine Wire EMG, Hand Load

4.2 Introduction

Many tasks of daily living require the use of the hand while exerting effort at the shoulder. In particular, gripping is frequently paired with sub-maximal actions of the shoulder, such as lifting a pot of coffee, carrying groceries, or using a hand tool. Studies have found that when gripping is performed in elevated arm postures, or while flexing or abducting the arm against resistance, the pattern of shoulder muscle activity is significantly altered (Hodder & Keir, in press; Antony & Keir, 2009; Smets et al, 2009; Sporrong et al, 1996; Sporrong et al, 1995). Commonly, a decrease in deltoid activity is seen, often accompanied by increased activity of at least one of the rotator cuff muscles (Hodder & Keir, in press; Antony & Keir, 2009; Smets et al, 2009; Sporrong et al, 1996; Sporrong et al, 1995). This finding is important due to the pervasiveness of rotator cuff injuries in today's society (Harrast et al, 2004). Thus it is important to understand the neuromuscular mechanisms that may act to overload the rotator cuff muscles and influence the control of shoulder muscle activity.

Assessing shoulder muscle activity requires more than standard surface electromyography (EMG) as only a small portion of the infraspinatus is accessible with surface EMG with any reliability. The remaining three muscles of the rotator cuff (supraspinatus, subscapularis and teres minor) require the use of intramuscular electrodes to monitor their activity (Waite et al, 2010). Due to these limitations, only two studies have investigated the response of the rotator cuff muscles (supraspinatus and infraspinatus) during concurrent shoulder elevation efforts and gripping (Sporrong et al, 1996; Sporrong et al, 1995). They found supraspinatus activity to increase by up to

17.1% with gripping during sub-maximal contractions in both sagittal and frontal planes, while infraspinatus activity was varied, increasing up to 2.7 % in flexion and decreasing by as much as 8.4% in abduction (Sporrong et al, 1995). Sporrong et al (1995) noted that there were large differences in the neuromuscular response to concurrent gripping between individuals. They proposed that these differences were the result of naturally occurring biological variation between people, however, they were uncertain why changes occurred with gripping and what mechanisms were responsible.

Neuromuscular control of the shoulder is not well understood. Continuous feedback is essential to allow for the precise coordination of muscle activity necessary to maintain shoulder joint integrity. Many have suggested that this feedback is provided by proprioceptors located either within the glenoid fossa or the surrounding labrum (Veeger & van der Helm, 2007; Gohkle et al, 1998; Vangsness et al, 1995). Thus, if the pattern of shoulder muscle activity results in the humeral head being off centre within the glenoid cavity, these proprioceptors would send signals to correct the pattern of activity. With gripping, patterns of shoulder muscle activity are changed, however, it is not yet understood what precipitates this change.

A recent study found that biceps activity increased significantly while gripping during sub-maximal arm elevations (Antony & Keir, 2009) and suggested that biceps activity increased in response to decreased deltoid activity during the gripping trials. The long head of biceps tendon originates from the supraglenoid tubercle, which allows the biceps to have a number of secondary roles ranging from assisting with shoulder flexion, abduction and medial rotation, as well as providing anterior-posterior shoulder stability and resisting superior humeral translation (Warner & McMahon, 1995; Itoi et al, 1993). It is also possible that biceps activity increased to stabilize the elbow as forearm flexors and extensors are engaged during gripping and thus would be the impetus for decreased deltoid activity. Biceps activity increased by 6% of maximal voluntary excitation (MVE) when gripping during sub-maximal arm elevations in the sagittal plane and mid-way between the sagittal and frontal planes. This change was accompanied by a small but significant increase in infraspinatus activity of 1.7% MVE, as measured with surface EMG (Antony & Keir, 2010).

The purpose of this study was to investigate the neuromuscular response to performing gripping and biceps contractions simultaneously with static sub-maximal shoulder flexion and abduction efforts. To investigate this objective, muscle activity during the tasks was recorded and analyzed, including the activity of the infraspinatus and supraspinatus using intramuscular electrodes. Furthermore, a musculoskeletal model was used to compute the magnitude and direction of each muscle's force contribution. It was hypothesized that gripping would increase biceps, supraspinatus and infraspinatus activity, while decreasing deltoid activity. It was thought that contracting the biceps during shoulder exertions would create a similar response. Given the number of degrees of freedom of the upper extremity, inter-individual variability was identified as a potential confounding factor; thus a secondary objective was to examine individual differences in muscular response.

4.3 Methods

4.3.1 Participants

Ten healthy males (177.0 ± 5.2 cm; 79.5 ± 10.0 kg; 27.3 ± 4.4 years) with no history of shoulder pain within the last year were recruited from the university population. This study was approved by the Human Research Ethics boards at McMaster University. All participants provided written informed consent prior to participating in the experiment.

4.3.2 Experimental Protocol and EMG Measurement

The experimental protocol consisted of 36 sub-maximal static shoulder efforts at 40% of each participant's maximal shoulder moment. Sub-maximal contractions were performed at three angles of elevation (30°, 60° and 90°) in both sagittal and scapular (30° anterior to adduction) planes using two forearm postures, neutral and supinated. In each combination of postures, the three load conditions were tested: (i) no additional load, (ii) 30% of maximum grip force and (iii) 30% maximum isometric biceps contraction. All exertions were 10 seconds in duration. The order in which the contractions were performed was randomized. First the plane was randomly chosen then forearm posture and order of load conditions. Elevation angle was always tested in the order of 30°- 60°- 90°. To avoid fatigue, one minute of rest was given between each angle of elevation and a minimum of 2 minutes of rest was given before changing forearm posture or plane of elevation. During the trials, visual feedback was provided via computer monitor to maintain the 40% shoulder moment, 30% grip force and 30% biceps contraction.

To obtain the specific values for feedback, maximal grip, maximal isometric biceps contractions and maximal static shoulder moments at 30°, 60° and 90° were recorded in both sagittal and scapular planes. While seated in the isokinetic dynamometer, maximal grip efforts were performed with the elbow extended and arm at the side, using a custom strain gauge grip dynamometer (MLT003/D, AD Instruments, CO, USA). The grip dynamometer was then affixed to the upper extremity attachment of the Biodex 4 system (Biodex Medical Systems, New York, USA). Maximal biceps contractions were performed against manual resistance with a slightly flexed elbow (<5°), as this was the arm posture used during the protocol. Maximal isometric shoulder moments at 30°, 60° and 90° of elevation in sagittal and scapular planes were obtained using the isokinetic dynamometer and with the elbow slightly flexed. Two repetitions of the maximal grip effort, biceps contraction and shoulder moment were performed, all 5 seconds in duration with 2 minutes of rest given between repetitions. The peak grip force, shoulder moment and biceps activation from either of the two repetitions was used to determine the respective feedback levels. The shoulder moment target was 40 ± 2% of the angle and plane specific maximum. Grip force and biceps contraction targets were both 30 ± 1.5% of their respective maximums.

The activity of eight shoulder muscles from the right arm and trunk were monitored. Six muscles were monitored using bipolar surface electromyography (EMG), while activity of the two rotator cuff muscles was monitored via fine wire electrodes. The muscles monitored with Ag-AgCl surface electrodes were the anterior (AD), middle (MD) and posterior (PD) deltoid, upper trapezius (UT), biceps long head (BI) and triceps long head (TRI). Prior to surface electrode placement, each site was shaved and scrubbed with isopropyl alcohol. Electrodes were placed over each muscle belly, parallel to the fibre direction.

The supraspinatus (SUP) and infraspinatus (INF) muscles were monitored with fine wire EMG. Paired hooked fine wire electrodes were inserted with a 25 gauge hypodermic needle (Chalgren Enterprises Inc., CA, USA). The 50 μm nickel chromium alloy wires had 2 mm of exposed wire at each tip. Due to the large anatomical variation in scapular geometry (Bigliani et al, 1986), the medial and inferior boarders of the scapular spine were carefully palpated and outlined on the skin. With the participant seated, the needle for the supraspinatus wire electrode was inserted into the muscle vertically to an approximate depth of 3-4 cm (depending on muscle and adipose thickness). For the infraspinatus wire electrode, the needle was inserted with the tip angled upward towards the scapular spine at approximately 30°. The emerging wires were taped securely to the skin and the exposed ends were attached to spring adapters (IPE500, Bortec Biomedical Ltd., AB, Canada).

Once the electrodes were in place, participants performed manually resisted maximal voluntary isometric contractions using muscle specific tasks as described in an earlier communication (*Chapter 3*). Participants were then seated in an isokinetic dynamometer and secured with chest and waist restraints to constrain their trunk posture. Participants then performed maximal isokinetic voluntary contractions at 30°/s in both scapular and sagittal planes with both forearm postures.

EMG was differentially amplified and band pass filtered from 10 to 2000 Hz (fine wire and surface CMRR > 115 dB at 60 Hz, input impedance ~ 10GΩ;AMT-8, Bortec Biomedical Ltd., AB, Canada). EMG was sampled at 4000 Hz (16 bit, USB-6259, National Instruments, TX, USA) with custom Labview software and saved for processing.

4.3.3 Data Processing

After rectification of EMG, all data (including torque and position) were filtered using a critically damped dual low pass filter with a 3 Hz cut-off. EMG was normalized to the maximum experimental excitation (MEE), the peak activity found in any trial during the experiment including the manually resisted muscle specific isometric contraction, the maximal isokinetic contractions, or all experimental contractions (*Chapter 3*). Trials were then clipped to a 3 second window starting from when the target conditions became stable. During the no additional load, or "no" load condition, the window would start when the shoulder moment met the target level of 40%. During the gripping and biceps contraction conditions, the window started when both the shoulder moment and the grip force or biceps contraction stabilized at their respective target levels.

The upper extremity musculoskeletal model in OpenSim was used to compute 3D muscle forces, the parameters of which are described in Holzbaur et al (2005). This musculoskeletal model uses literature based estimates of muscle origin and insertion, muscle fibre length, pennation angle, and physiological cross sectional area (Langenderfer 2004), incorporated in a Hill-based muscle model to calculate individual muscle forces (Holzbaur et al, 2005). The skeletal elements of the model are representative of the 50^{th} percentile male with a height of 170 cm (Holzbaur et al, 2005). The OpenSim force reporter analysis tool was used to calculate muscle forces with respect to the centre of the humeral head in each condition. Muscle forces from AD, MD, PD, SUP, INF and BI were calculated from muscle activity collected for each participant. UT and TRI were not included in the model. The UT was not included as it does not span the glenohumeral joint. Only the long head of the TRI spans the glenohumeral joint and thus to not overestimate its contribution to the generation of shoulder moment, it was not included in the current model (Anglin et al, 2000). The force reporter tool uses the optimal force, which is defined by maximum isometric force, optimal fibre length, tendon slack length and pennation angle (Schutte et al, 1993), and normalized muscle activity, represented as a proportion from 0 to 1 (0 to 100% of maximum excitation), to compute muscle force.

Muscle forces were modeled to allow for a better understanding of how each muscle acts to direct the humeral head into the glenoid. The direction of each muscle's force was determined with respect to a local axis system established at the centre of rotation of the humeral head (Bassett et al, 1990; Poppen & Walker, 1976). The Y axis was defined along the long axis of the humerus, positive forces acting proximally and negative forces acting distally (Figure 4.1a). In anatomical position, the Y axis was oriented in the cranial (+Y) and caudal (-Y) direction, the X axis was oriented in the anterior (+X) and posterior (-X) direction and the Z axis, the medial (+Z) and lateral (-Z) direction (Figure 4.1a). Muscles were modeled as a series of straight line segments connected by nodes to represent the curved pathway of muscle (Holzbaur et al, 2005). The direction of muscle force was described as the pathway starting from the muscle insertion point, to each node and ending the point of origin for each muscle. Muscle force was multiplied by each segment length as a proportion of that muscle's total length. The components (X, Y and Z) of muscle force for each segment of its length were then summed and individual muscle forces calculated. For each participant, the effect of all muscles or resultant force was calculated for every posture. The direction of the muscle forces and resultant force was defined by angles relative to the right horizontal between X and Z axes (Figure 4.1b), between the Y and Z axes (Figure 4.1c) and between the X and Y axes (Figure 4.1d).

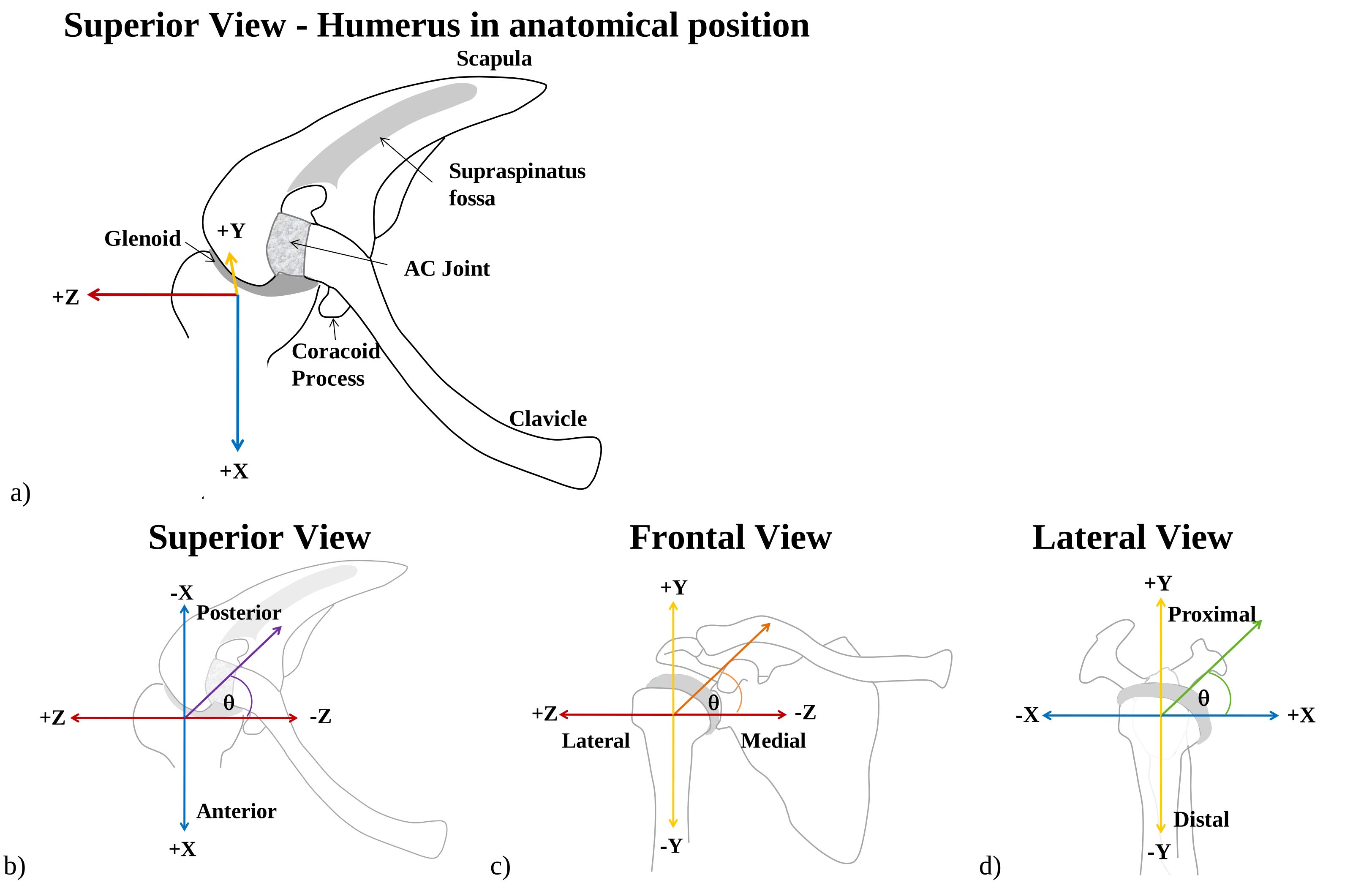

Figure 4.1: Definition of the local axis system at the joint center in the anatomical position. The X axis (anterior - posterior), Y axis (proximal - distal), and Z axis (lateral) are shown in a), and the angles between b) X and Z axes, c) Y and Z axes and d) X and Y axes.

4.3.5 Statistical Analysis

One subject had mean grip forces that were three standard deviations outside the group mean, thus that subject's data were removed from further analysis. For the remaining nine participants, means and standard deviations for relative shoulder moment, grip force and normalized muscle activity were calculated. The effect of load condition on mean muscle activity and resultant force was assessed in each posture using a 3-way repeated measures analysis of variance (ANOVA) with a 95% confidence interval. Significant effects were followed up with a least significant difference (LSD) post hoc test. To examine inter-individual variability in muscle activity, coefficients of variation were calculated for each muscle by dividing the standard deviation (between subjects) by the mean as well as polar plots of each individual's axial muscle forces in a selected posture.

4.4 Results

4.4.1 Effect of Load on Muscle Activity

Across all trials, the mean relative shoulder moment was 40.6 ± 2.2% of maximum. Mean grip force across all gripping trials was 27.3 ± 1.8% of maximum voluntary grip (MVG), which was higher than grip forces recorded during the no load conditions (21.1 ± 4.6% MVG) and biceps contraction conditions (22.3 ± 5.2% MVG).

A main of effect of load was found for biceps activity ($F_{(2,16)}$= 5.0, p=0.02). Mean biceps activity was 29.1 ± 4.2% MEE during the biceps contraction condition, which was significantly greater than activity with gripping (23.0 ± 9.8% MEE, p = 0.03) and shoulder contraction alone (24.4 ± 10.7% MEE, $p = 0.05$). Load (gripping or biceps contraction) affected shoulder muscle activity only during scapular plane shoulder moments with neutral forearm posture (Figure 4.2). At 30° of elevation, UT activity was 9.4% MEE higher ($F_{(2,16)} = 6.3$, $p = 0.02$) during the biceps contraction condition than in the gripping condition (Figure 4.2a). At 60° of elevation, middle and posterior deltoid activity was significantly lower during the gripping condition than the no load condition, as MD activity decreased by 6.6% MEE ($F_{(2,16)} = 3.5$, $p = 0.05$) and PD activity ($F_{(2,16)} = 4.7$, $p = 0.02$) by 8.3% MEE (Figure 4.2b). Similar differences were found at 30° of elevation.

The response of the rotator cuff muscles was variable (Figure 4.2). INF activity was noticeably lower when contracting the biceps at 60° (9.2% MEE lower than no load) and with gripping at 90° (8.5% MEE lower than no load), however this was not significant. SUP activity was relatively low in this posture, less than 10% MEE, and subtle differences with load condition were considered negligible. There appeared to be a trend for AD activity to increase when gripping or biceps contractions were added, however the variability between individuals was too large for it to be considered significant. There was no effect of load on resultant force in any posture (Appendix A) due to large inter-individual variability.

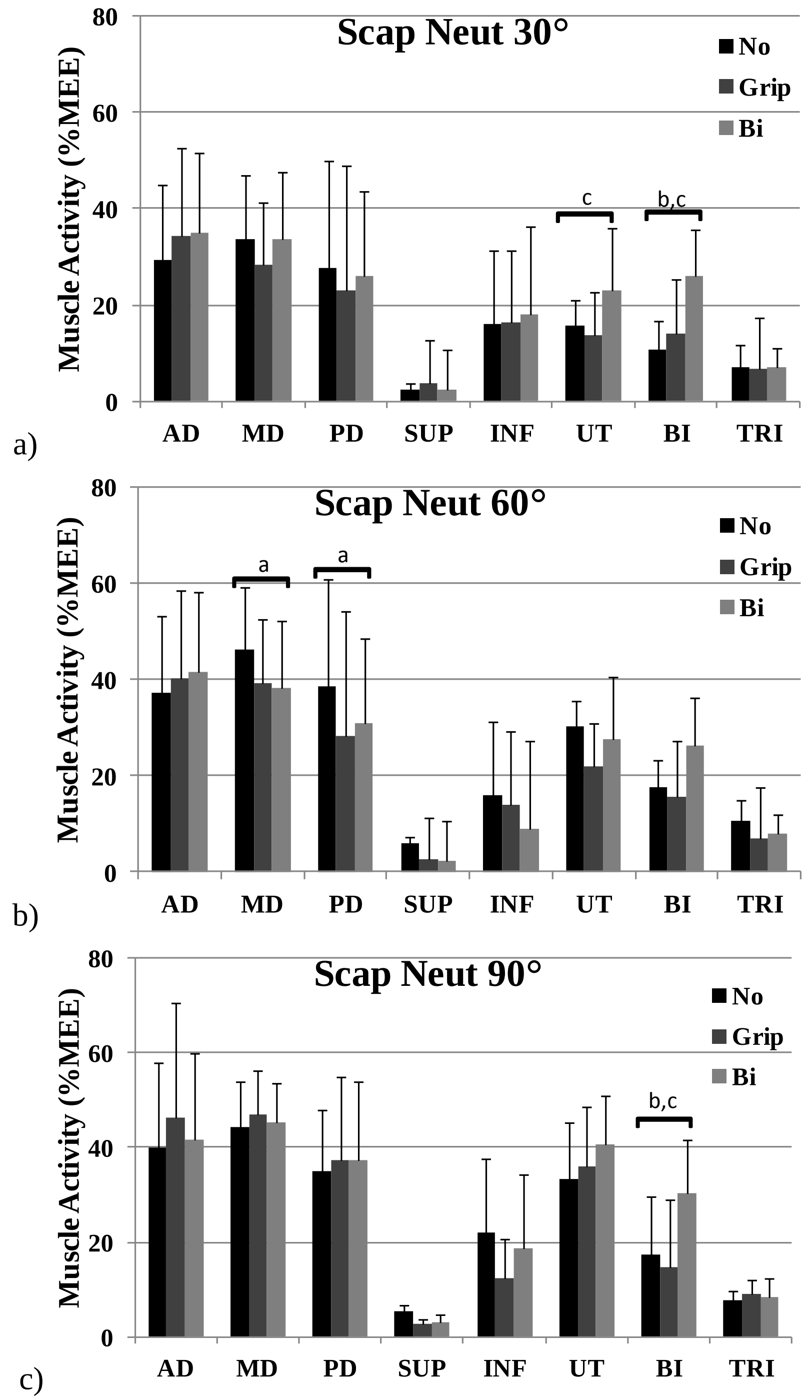

Figure 4.2: Mean (and SD) of muscle activity (% MEE). Significant differences ($p < 0.05$) in mean muscle activity with loading task are indicated as follows: a - between the no load trial and gripping trial, b - between no load and biceps contraction trial, and c - between gripping and biceps contraction trials.

4.4.2 Individual Variability

Across all postures and conditions, the coefficients of variation in muscle activity were large. AD had the lowest inter-individual variability, with a coefficient of variation of 43.7% while SUP was the highest with a coefficient of variation of 208.7 %. MD and INF activity levels were also highly variable between participants, with overall coefficients of variation of 181.5% and 128.3 %, respectively.

Muscle forces were determined using EMG, thus individual and resultant muscle forces were highly variable as well. Muscle forces and resultant force from X and Z components, representative of the axial forces, were plotted for each participant during scapular plane abduction at 90° with the forearm supinated (Figure 4.3). Of the deltoids heads, the AD was used most predominantly by all participants in this scapular plane posture (Figure 4.3). In five participants, INF forces were larger than those from any of the deltoid heads (Figure 4.3d-g,i). The direction of the resultant force of these five participants differed from the other participants by 60° to 210°, changing from anterior-cranial to posterior-caudal in direction (Figure 4.3). Participants S03 (Figure 4.3g) and S04 (Figure 4.3h) had distinct strategies, using proportionately higher MD, PD and BI force than the other participants.

Although the effect of load condition on resultant muscle force was not statistically significant, it appears dependent on each individual's pattern of force distribution (Figure 4.3). In participants S03, S04 and S09 the direction of resultant force was altered by 10° to 20° during the gripping condition, and was accompanied by changes in resultant force magnitude in two, as it increased with gripping in S04 and decreased with gripping in S09 (Figure 4.3g-i).

Despite the large variability in force between individuals, the relative distribution of muscle force within each participant appeared to be consistent. Figure 4.4 depicts the forces from participant S04 during exertions at 60° of elevation in both shoulder planes and forearm postures. The proportions of AD, MD, PD, INF and SUP appeared to be similar in all postures. Plane, elevation angle and forearm posture only acted to gain up or down the magnitude of force from these shoulder muscles (Figure 4.4).

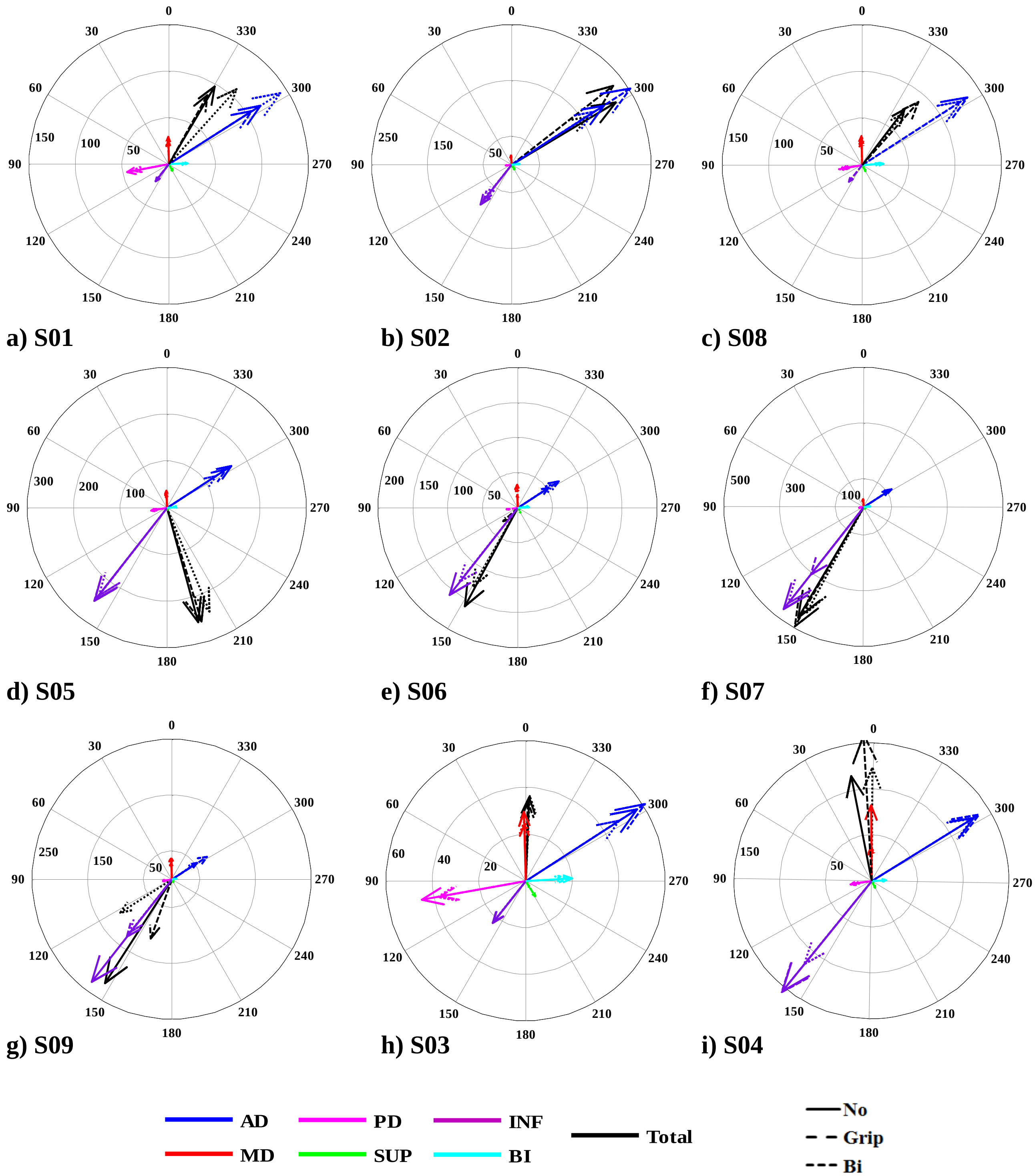

Figure 4.3. Muscle forces and resultant force (N) of X and Z component for all participants in the scapular plane at 90° of elevation with supinated forearm posture. The vantage point of the plot is looking up the long axis of the humerus into the glenoid (lateral view). The shoulder muscle contraction alone ("No") is represented by a solid line, concurrent gripping ("Grip") with a dashed line and concurrent biceps contraction ("Bi") by the dotted line. The resultant force from these muscles is depicted with the black line.

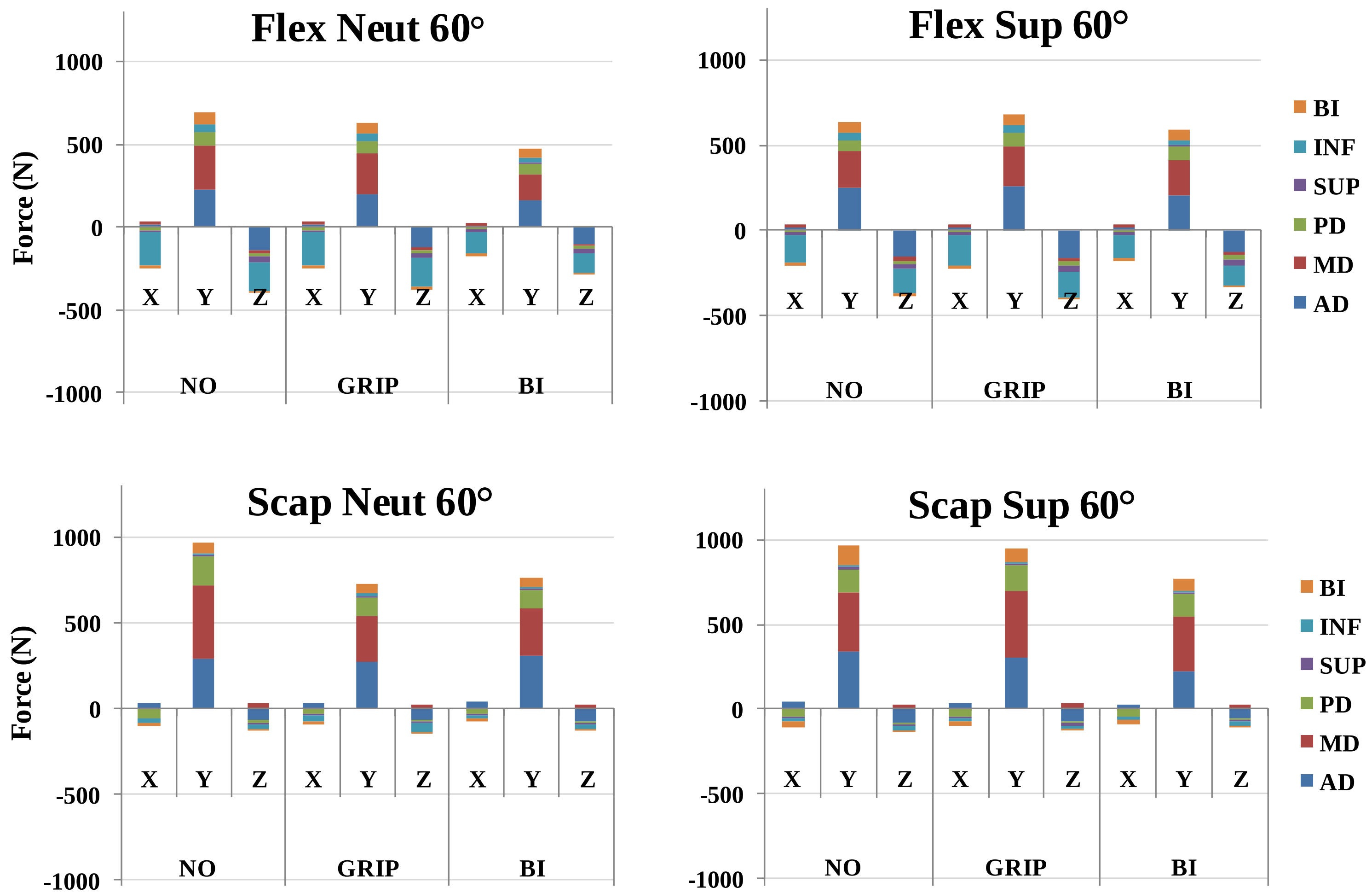

Figure 4.4: Muscle forces (N) observed for one participant (S04) at 60° of elevation across all planes, forearm postures and load conditions investigated. Contributions of each deltoid head appear proportional across conditions. AD (blue) was responsible for about 40% of deltoid force, MD (red) approximately 44% and PD (green) approximately 16%.

4.5 Discussion

Neuromuscular control of the shoulder is intrinsically complex. This study examined neuromuscular control strategies of the shoulder when grip force or biceps contractions were combined with sub-maximal shoulder moments. Large differences were found in control strategies seen between individuals, precluding statistically significant differences in muscle activity and muscle force with load condition. Although suspected prior to the study, the variability between individuals was greater than expected with coefficients of variation exceeding 100% for most muscles. This reflects the number of degrees of freedom of the shoulder muscles which allows for numerous strategies to achieve the same outcome. This resulted in widely variable patterns of muscle force distribution as each individual used a different strategy to perform the same task (Figure 4.3). Sporrong et al (1995) commented that the natural occurring biological variability between individuals is the cause of these large inter-individual differences in the neuromuscular response to concurrent gripping and sub-maximal shoulder moments.

Individual differences in patterns of muscle force were observed and presented graphically for scapular plane isometric moments at 90° of elevation (Figure 4.3). It should be noted that the use of the model was limited to determining the muscle lines of action to provide direction for the muscle activity and determine an estimate of the resultant muscle force. The AD was consistently the primary head of the deltoid used by all participants to generate shoulder moments in the scapular plane, but it was not always the greatest contributor as three patterns of muscle force distribution emerged (Figure 4.3). In three participants (S01, S02, S08), AD force was dominant and little force was exerted from any of the other muscles investigated. In four participants (S05, S06, S07, S09), INF was dominant, but paired with larger forces from the AD. This pattern of force distribution drastically changed the direction of the resultant force. Interestingly, Meskers et al (2004) found that despite their drastically different anatomical positions, the directions of principle action for the AD and INF greatly overlap, thus appearing to work together as a functional unit. Two participants had a very distinct distribution pattern with AD, MD, PD, SUP and INF recruited to a similar extent (S03 & S04). This pattern generally resulted in lower magnitude resultant forces.

Co-activating multiple shoulder muscles is a paradox found in individuals with shoulder pathologies (Myers et al, 2009; Meskers et al, 2004). It acts to increase joint stability and protect the shoulder from further injury, however may also result in overloading muscles. Responses from teres minor, subscapularis and pectoral muscles would allow further evaluation of the force distribution and its effect on humeral head seating. Muscle dysfunction has long been associated with shoulder injury (Itoi et al, 1993), however it is not yet known whether the dysfunction precipitated the injury or is the result of injury. All participants in the current study were healthy and absent of any shoulder injury or pain, thus without longitudinal evaluation, it is unknown which, if any, of these patterns may be indicative of future problems.

Although different strategies appeared between participants in this study, the patterns of muscle force distribution within each individual were relatively consistent. As can be seen in the participant shown in Figure 4.4, while the absolute magnitude of force changes with plane, elevation angle, forearm posture and load condition due to mechanical demands, the relative contribution of each muscle is consistent suggesting that there is a consistent neuromuscular strategy for the individual. Like the current study, low intra-individual variance was also found by Meskers et al (2004), when investigating principle actions of shoulder muscles. While motor variability is reduced with learning and training, is not necessarily beneficial in all circumstances. Madeline et al (2008) found that low variability in muscle activity was linked to an increased rate of workplace injury. The apparent invariance in motor programming within individuals may provide some insight into the injury progression in some individuals.

This study examined the effects of gripping and contracting the biceps on shoulder muscle activity that was inclusive of the supraspinatus, infraspinatus and the biceps. The current study was precipitated by the previous finding that biceps activity increased when gripping was performed concurrently with shoulder raises (Antony & Keir, 2009). The current study found no significant increase in biceps activity with gripping compared to the 40% shoulder moment alone (Figure 4.2). Only the upper trapezius was affected by the biceps contraction condition, and only significantly so at an elevation of 30° in the scapular plane with neutral forearm posture. Thus, biceps activity did not appear to influence the neuromuscular control of the shoulder as hypothesized.

Previous interpretations of shoulder function were limited without the directional force component provided by the OpenSim analysis in the current study (Meskers et al, 2004; Michiels & Bodem, 1992). By using EMG in conjunction with the OpenSim model, this study allowed new insights to the neuromuscular response of the shoulder complex, including the often neglected supraspinatus and infraspinatus muscles via fine wire electrodes. Using this approach we were able to interpret muscle activity as an input for muscle force, thus applying a pathway of force and the effects of muscle size in the analysis, while still preserving muscular responses from each individual. Many believe that proprioceptors within and around the glenoid cavity are responsible for the neuromuscular control of the shoulder and therefore the position of the humeral head and direction of joint contact forces are pertinent to the neuromuscular control of the shoulder (Veeger and van der Helm, 2007; Gohkle et al, 1998; Vangsness et al, 1995). The approach used in this study allowed for a better understanding of how patterns of muscle activity might affect the direction and magnitude of muscle force on the humeral head and thus its influence on the proprioceptive feedback contributing to the neuromuscular control of the shoulder.

There are several limitations to this study. Grip force during the no additional load trials did not differ significantly from those during gripping trials. This was likely a function of the grip dynamometer attachment to the Biodex and being the point of force application. Uncoupling the grip dynamometer from the shoulder dynamometer may have lowered grip forces during the non-gripping trials, but this configuration was selected to maintain the typical use of the dynamometer. Movement of the scapula was not considered and a number of shoulder muscles were not investigated. Examination of more shoulder muscles, in particular teres minor and subscapularis muscles may have provided more insight to the strategies being used, however, it is likely that they also would have shown similar high inter-individual variability.

4.6 Conclusions

In this study, alterations to shoulder muscle activity with concurrent gripping and sub-maximal shoulder isometric contractions were only found in one posture and were not the same as differences identified with concurrent biceps contraction in that posture. Large differences were found between individual patterns of muscle force, yet the relative muscle contributions were consistent across all conditions within each individual. This study highlights the need for further investigation of the neuromuscular response of the shoulder musculature to multiple task performance of the upper extremity but also demonstrated that individual motor programming may explain why certain individuals appear to be predisposed to musculoskeletal injuries in the workplace.

4.7 Acknowledgements

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CHAPTER 5

Study 4: Shoulder muscle control during maximal dynamic shoulder moments with additional upper extremity tasks

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5.1 Abstract

Hand tasks concurrent with shoulder moment generation has been shown to alter deltoid and rotator cuff muscle activity; recent evidence suggests that the biceps may play a role. The purpose of this study was to examine the muscular response of the shoulder to concurrent biceps contraction and gripping during maximal isokinetic shoulder elevations. Muscle activity of ten healthy males was recorded from all three heads of the deltoid, biceps, supraspinatus and infraspinatus during maximal shoulder elevations in the flexion and scapular planes with both neutral and supinated forearm postures. Three conditions were tested, (i) no additional load, (ii) 30% maximum grip effort, and (iii) 30% maximum biceps contraction. Muscle forces were calculated from muscle activity at 0.1° intervals of arm elevation, averaged and peak forces obtained. Across all conditions, both grip and biceps contraction tasks reduced maximum shoulder moments by 26.7-33.0% and 26.1-36.7%, respectively. When gripping, muscle forces appeared to be lower throughout the range of motion; however, analysis of peak muscle forces revealed that only the anterior deltoid was significantly lower when gripping in flexion with a supinated forearm. Contracting the biceps significantly lowered peak force from all three heads of the deltoid as well as supraspinatus and infraspinatus. Isolated biceps contractions had greater influence on the neuromuscular control of the shoulder than gripping itself, thus does not appear responsible for muscle activity changes due to concurrent gripping.

Keywords: Dynamic, Shoulder, Rotator Cuff, EMG, Muscle Force

5.2 Introduction

The rotator cuff muscles are vital to maintaining shoulder joint integrity, especially during dynamic movements (Veeger and van der Helm, 2007). Changes to the coordination of shoulder muscle activity can have a negative effect on the shoulder and may result in dysfunction, pain and injury (Itoi et al, 1993). It has been shown that gripping in conjunction with isometric or dynamic shoulder moments alters the pattern of shoulder muscle activity in a manner that may increase risk of injury to the rotator cuff muscles (Hodder and Keir, in press, Sporrong et al, 1996, Sporrong et al, 1995). Concurrent gripping and shoulder actions have been shown to decrease deltoid activity from 4% to 28% of maximal voluntary excitation (Hodder and Keir, in press; Antony and Keir, 2009; Smets et al, 2009; Au and Keir, 2007; MacDonell and Keir, 2005; Sporrong et al, 1996, Sporrong et al, 1995). More important than the decrease in deltoid activity was that concurrent gripping has been shown to increase supraspinatus activity up to 24% of maximum activity, depending on arm posture (Sporrong et al, 1996, Sporrong et al, 1995). Infraspinatus activity has also been shown to increase with concurrent gripping up to 20% of maximum (Sporrong et al, 1996), however, a study using surface EMG found it to decrease by 7% of maximum (MacDonell and Keir, 2005). Thus, gripping seems to influence shoulder muscle activity but is dependent on the test conditions and specific methods used (e.g. contraction level, grip force level and arm posture). Due to the susceptibility of the rotator cuff muscles to injury (Harrast et al, 2004), it is important to identify the specific factors that influence neuromuscular control of the shoulder.

In their study of arm raises in the sagittal, frontal and mid-way between sagittal and frontal planes, Antony and Keir (2009) found that concurrent gripping increased biceps activity by 6% maximum voluntary excitation. Although the primary function of the biceps is to flex the elbow, it can also contribute to shoulder flexion and abduction (Inman et al, 1944). Given these potential actions, Antony and Keir (2009) suggested that the increase in biceps activity was predicated by or simply off-set the lower deltoid activity found during gripping. However, these potential correlations do not speak to the mechanism that initiated the decrease in deltoid activity with gripping. Examining the pattern of muscle force in the three-dimensional upper extremity musculoskeletal system would provide insight into the relationship between these muscles. Given that the biceps is a strong supinator of the forearm, it is likely that biceps activity increased in order to balance moments and forces generated by both the flexors and extensors of the wrist and fingers with gripping (Mogk and Keir, 2003 Snijders et al, 1987). A consequence of this counteractive biceps action is that it would also generate a shoulder moment and thus decrease the need for deltoid action. By examining the impact of biceps contractions and gripping concurrent to shoulder isometric contractions or dynamic movements, we can test this hypothesis and determine the role of the biceps during gripping. Since changes in rotator cuff muscle activity have also been reported in conjunction with the deltoid and biceps during gripping, examination of the response of the supraspinatus and infraspinatus muscles is also warranted for a more comprehensive evaluation of the neuromuscular control of the shoulder.

The purpose of this study was to conduct a comprehensive examination of the neuromuscular response of the shoulder muscles to the addition of gripping or biceps contraction while performing maximal dynamic shoulder elevations. To investigate this objective, muscle activities were recorded during each task and used as inputs into a musculoskeletal model to provide continuous three-dimensional muscle forces throughout each motion. It was hypothesized that the musculoskeletal reaction to gripping and biceps contractions would be similar and both would cause a decrease in deltoid muscle force and an increase in supraspinatus and infraspinatus muscle force which would result in lower maximal shoulder moments.

5.3 Methods

5.3.1 Participants

Ten healthy males (177.0 ± 5.2 cm; 79.5 ± 10.0 kg; 27.3 ± 4.4 years) with no history of shoulder pain within the last year were recruited from the university population. This study was approved by the Human Research Ethics board at McMaster University. All participants provided written informed consent prior to data collection.

5.3.2 Experimental Protocol

The protocol consisted of 12 maximal isokinetic shoulder elevations. Subjects were seated in an isokinetic dynamometer (Biodex Medical Systems, New York, USA) and the chest and waist restraints fastened to constrain trunk posture. Participants performed straight arm concentric shoulder exertions in the sagittal and scapular (30° anterior to abduction) planes from 0° to 90° of elevation at 30°/s. Movements were performed with two forearm postures, neutral and supinated (palm up) and with three load conditions: (i) no additional load, (ii) 30% maximum grip force and (iii) 30% maximum biceps contraction. Three repetitions of each condition were performed continuously and two minutes of rest was given between conditions.

Muscle activity of the anterior (AD), middle (MD) and posterior (PD) deltoids and biceps brachii (BI) were monitored via surface electromyography (EMG), while the activity of two rotator cuff muscles, the supraspinatus (SUP) and infraspinatus (INF) were monitored with fine wire EMG. Prior to surface electrode placement, each site was shaved and scrubbed with isopropyl alcohol. Surface electrodes were placed over the mid-belly of each muscle parallel to fibre direction with a 2.5 cm inter-electrode distance.

The INF and SUP were instrumented with paired hooked fine wire electrodes, which were inserted with a 25 gauge hypodermic needle (Chalgren Enterprises Inc., CA, USA). The nickel chromium alloy 50 μm wires had 2 mm of exposed wire at each tip and an inter-electrode distance of approximately 5 mm. Due to the large anatomical variation in scapula size (Bigliani et al, 1986), the scapular spine, medial and inferior borders of the scapula were palpated and marked on the skin to guide electrode insertion. The skin over each insertion site was thoroughly cleansed with isopropyl alcohol. For SUP, the needle was inserted into the muscle caudally to an approximate depth of 3-4 cm (depending on the participant). For INF, the needle was inserted into the muscle with the tip angled cranially towards the scapular spine at approximately angle of 30°. The emerging wires were taped in place and attached to spring adapters mounted on the pre-amplifier (IPE500, Bortec Biomedical Ltd., AB, Canada). Fine wire and surface EMG were differentially amplified and band pass filtered from 10 to 2000 Hz (CMRR > 115 dB at 60 Hz, input impedance ~ 10GΩ; AMT-8, Bortec Biomedical Ltd., AB, Canada). EMG was sampled at 4000 Hz with custom Labview software (16 bit, USB-6259, National Instruments, TX, USA).

Prior to commencing the experimental trials, muscle specific manually resisted maximal voluntary isometric contractions were performed. The muscle specific contractions have been described previously (*Chapter 3*). Additionally, two manually resisted maximal voluntary isometric contractions of the biceps were performed with minimal elbow flexion (<5°). Contractions were held for 5 seconds and repeated twice. The greatest activity achieved in any of the biceps contractions was used to normalize biceps activity and provide feedback during the experiment. Three, 5 second maximal voluntary grip force (MVG) trials were collected using a custom strain gauge grip dynamometer (MLT003/D, AD Instruments, CO, USA) with the elbow extended and arm at the side. Maximum grip force (MVG) recorded in these trials was used to normalize grip force and provide feedback during the experiment. The grip dynamometer was then affixed to the upper extremity attachment of the isokinetic dynamometer (Biodex 4, Biodex Medical Systems, NY, USA). Grip force, as well as angle of elevation and shoulder moments (from the Biodex), were sampled at 4000 Hz. Visual feedback to regulate grip force and biceps activity was presented to participants on a computer monitor (*Chapter 4*; Hodder and Keir, in press; Au and Keir, 2007). The targets for both grip force and biceps activity were 30 ± 1.5 % of maximum.

5.3.3 Data Processing

EMG was full wave rectified and filtered with a critically damped dual low pass filter with a 3 Hz cut-off. EMG data were then normalized to maximal experimental excitations (MEE), which was the peak activity found in any trial during the experiment including the manually resisted muscle specific isometric contractions or any of the experimental maximal isokinetic contractions (*Chapter 3*). All trials that met a velocity criterion of 30 ± 3°/s were re-sampled at 0.1° increments.

EMG, grip force and external shoulder moment data from the dynamic contractions were clipped from 10° to 70° of elevation and the EMG recordings used as inputs into a musculoskeletal model to calculate muscle force. The upper extremity model (Holzbaur et al, 2005) and force reporter tool in OpenSim were used to calculate muscle forces from muscle activities of AD, MD, PD, SUP, INF and BI at each 0.1° increment of arm elevation. OpenSim uses a Hill-type model to compute muscle forces. Muscle parameters used in the model were fibre length, pennation angle and physiological cross-sectional area and are representative of the 50th percentile male (Holzbaur et al, 2005).

Muscle forces were determined with respect to a local axis system established at the centre of rotation of the humeral head (Bassett et al, 1990; Poppen & Walker, 1976). Proprioceptors located in and around the glenoid cavity are thought to be involved in the neuromuscular control of the shoulder (Veeger and van der Helm, 2007), thus muscle forces were modeled to determine how each muscle acts to direct the humeral head with respect to the glenoid cavity. With the arm in the anatomical position, the local axis system was defined with the positive X axis directed anteriorly, the positive Y axis directed proximally along the long axis of the humerus, and the positive Z axis directed laterally (Figure 5.1).

In OpenSim, muscles are modeled as a series of straight line segments connected by nodes to represent the curved pathway of muscle (Holzbaur et al, 2005). The direction of muscle force was described as the pathway starting from the muscle insertion point, to each node and ending the point of origin for each muscle. Muscle force was multiplied by each segment length as a proportion of that muscle's total length. Thus the direction of each muscle force was weighted by the direction of each segment. For each condition, the resultant of all muscle forces was determined.

5.3.4 Data Analysis

Shoulder moments collected during gripping and biceps contraction conditions were normalized to the peak shoulder moment collected during the "no load" condition in each movement. Three subjects were not able to maintain velocity within 3°/s of the 30°/s target for a majority of trials and their data were excluded from the analysis. In the seven remaining participants, the mean grip force, mean biceps activity and peak normalized shoulder moments were found from 10° to 70° along with the standard deviation of each measure across this range. Means and standard deviations were then averaged across the three repetitions of each trial. Muscle forces and resultant forces from the three repetitions of every trial were ensemble averaged at 0.1° increments of elevation for each individual and plotted for visual inspection.

The effects of load condition (no load, gripping and biceps contraction) on peak normalized shoulder moment were examined with a repeated measures analysis of variance (ANOVA) with an alpha level of 0.05. To examine the effect of load condition on muscle forces and resultant force, peak forces were found from the ensemble average curves and analyzed in separate repeated measures ANOVA (alpha of 0.05). Significant effects were followed up using the least significant difference post hoc test.

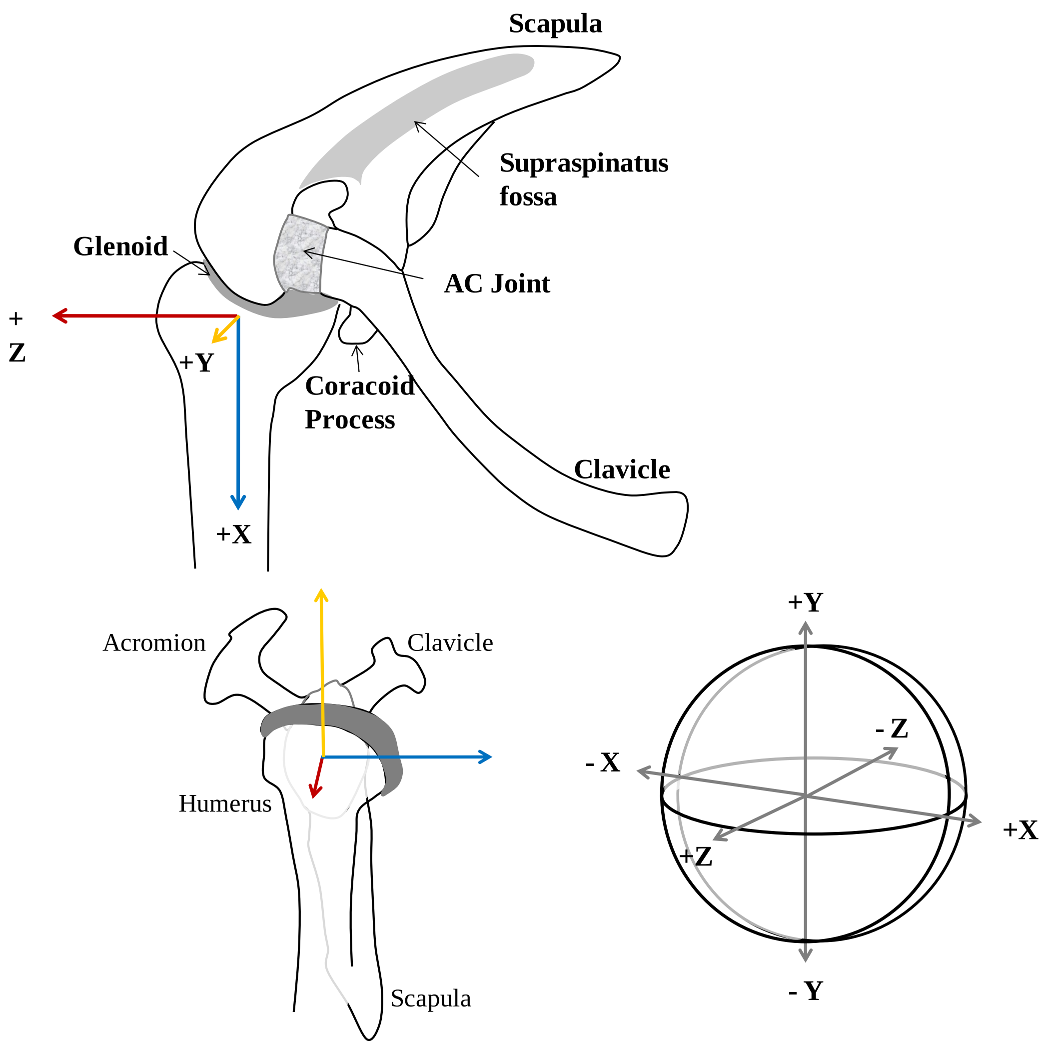


Figure 5.1. Definition of the local axis system at the centre of the humeral head. The positive X direction (anterior) is represented in blue, the positive Y direction (upward)

has been represented in yellow, and the positive Z direction (lateral) has been represented in red.

5.4 Results

Participants were not successful at maintaining grip force within the target range of 30 ± 1.5% MVG across the whole range of motion as mean grip force was 24.4 ± 8.3% MVG. In the other two conditions, mean grip force was higher and much more variable at 30.3 ± 22.3% MVG (no load) and 32.6 ± 18.8% MVG (biceps contraction) (Table 5.1). Participants were successful at maintaining the 30% biceps contraction with a mean activity of 31.7 ± 7.3% MEE during the biceps contraction condition. Biceps activity during gripping (18.0 ± 7.6% MEE, $p < 0.05$) and no load conditions (25.3 ± 10.9% MEE) were lower than the biceps contraction condition (Table 5.1).

5.4.1 Effect of Load Condition on Shoulder Moment

A main effect of load condition was found for maximal shoulder moments as both grip and biceps contraction tasks significantly lowered moments in flexion with neutral ($F_{2,12}$=21.7, $p<0.00$) and supinated ($F_{2,12}$=14.3, $p<0.00$) forearm postures as well as scapular plane moments with neutral forearm posture ($F_{2,12}$=12.4, $p<0.00$) (Table 5.1). Maximal shoulder flexion moment was significantly lower in the grip task by approximately 27% for both neutral and supinated forearm postures ($p<0.00$ and $p<0.02$), respectively) (Table 5.1). In the scapular plane, the grip task resulted in significantly lower maximal shoulder moment in neutral forearm posture (33.0%, $p<0.02$), but not in supination. Concurrent biceps contraction also resulted in significantly lower shoulder moment in flexion with both forearm postures (35-37%, $p<0.01$), but only with a neutral forearm in the scapular plane (26%, $p< 0.01$) (Table 5.1).

Table 5.1. Mean (standard deviation) peak external shoulder moment as a percentage of the no load condition, mean grip force (% MVG) and biceps contraction (% MEE) in each condition. Note: [a] significantly different ($p < 0.05$) than shoulder movement alone ("No"), [b] significant difference ($p < 0.05$) between "Grip" and "Bi".

	Sagittal Plane						**Scapular Plane**					
	Neutral			**Supinated**			**Neutral**			**Supinated**		
	No	**Grip**	**Bi**	**No**	**Grip**	**Bi**	**No**	**Grip**	**Bi**	**No**	**Grip**	**Bi**
Shoulder Moment (%)	100	73.3[a] (15.7)	63.3[a] (16.8)	100	73.1[a] (18.3)	65.2[a] (12.1)	100	67.0[a] (19.9)	73.9[a] (18.3)	100	71.4 (43.2)	88.0 (21.8)
Grip Force (%)	27.9 (22.1)	24.0 (10.3)	29.9 (19.3)	26.7 (24.6)	22.8 (7.6)	24.6 (19.1)	33.1 (24.4)	26.7 (8.5)	45.5 (26.6)	33.8 (22.4)	23.9 (9.4)	28.3 (29.6)
Biceps Activity (%)	24.1 (14.1)	15.7[a] (10.7)	30.0[b] (6.7)	33.5 (12.4)	26.4 (9.9)	34.7 (11.4)	15.2 (9.7)	9.9 (5.5)	28.3[a,b] (10.9)	28.5 (13.3)	19.8 (10.1)	33.9[b] (11.4)

5.4.2 Effect of Load on Muscle Forces

There was a main effect of load found on muscle force during flexion with supinated forearm posture, as forces from the AD, MD, PD, SUP and INF (all $F_{2,12} > 4.9$, $p < 0.03$) where all significantly lower in the biceps contraction condition than the no load condition. In the same motion, AD force was 123.1 N lower in the grip condition than the no load condition. A main effect of load was also found for resultant muscle force ($F_{2,12}$=5.3, p =0.02) during flexion with a supinated forearm, where the grip task lowered resultant force by 499.5 N (p=0.05) and by 337.3 N (p=0.03) with biceps contraction (Table 5.2).

Muscle forces were examined from 10° to 70° of the movement using the group ensemble averages (Figure 5.2). In flexion, load condition appeared to affect individual muscle forces in both forearm postures (Figure 5.2a & b). With neutral forearm posture in flexion, concurrent gripping and biceps contractions appeared to attenuate AD, MD, PD, and SUP forces as well as the resultant (Figure 5.2a). These attenuated forces were also seen with the supinated forearm posture, along with lower INF force (Figure 5.2b).

During scapular plane elevations with neutral forearm posture, neither gripping nor contracting the biceps appeared to alter the magnitude of muscle forces or the resultant force curve. However, the pattern of the force curves did change, as seen in the divergence of the curves at the end range of motion as forces during the biceps condition were reduced (Figure 5.3c). With the forearm supinated in the scapular plane, the grip task again appeared to reduce muscle force and contracting the biceps lowered all muscle forces except BI at the end of shoulder elevation (70°) (top of curve; Figure 5.3d).

Table 5.2: Mean peak muscle forces and resultant force (± standard deviation) for all conditions. A main effect of load was found in flexion with supinated forearm posture. Note: * significant difference ($p<0.05$) from shoulder elevation alone condition ("No").

		Muscle Force (N)					
		Sagittal Plane			Scapular Plane		
		No	Grip	Bi	No	Grip	Bi
Neutral	AD	433.1 (161.7)	346.0 (159.4)	359.8 (102.1)	599.5 (120.2)	437.5 (216.7)	555.8 (230.8)
	MD	289.7 (112.0)	232.5 (113.7)	239.6 (76.7)	397.5 (79.1)	301.4 (140.2)	382.7 (154.9)
	PD	390.2 (147.8)	312.0 (149.1)	328.4 (99.0)	555.9 (110.6)	413.8 (198.6)	526.9 (216.1)
	SUP	487.1 (182.4)	389.7 (182.9)	405.3 (119.4)	598.2 (119.6)	447.8 (213.3)	567.9 (233.3)
	INF	51.3 (19.3)	39.7 (21.4)	42.4 (12.5)	63.5 (11.6)	52.9 (24.0)	62.1 (25.2)
	BI	136.9 (51.3)	105.1 (58.2)	114.0 (32.9)	175.2 (59.8)	137.4 (68.3)	164.3 (74.8)
	Res	**1434.0 (540.2)**	**1116.3 (412.8)**	**1149.5 (338.4)**	**1991.7 (414.3)**	**1487.6 (714.9)**	**1881.4 (725.6)**
Supinated	AD	517.9 (147.4)	394.8 (148.9)*	425.8 (160.8)*	658.3 (150.5)	533.3 (206.5)	609.8 (159.5)
	MD	352.3 (115.4)	262.0 (107.9)	272.3 (107.3)*	437.8 (99.1)	356.1 (130.7)	414.1 (100.9)
	PD	472.8 (148.2)	355.0 (140.1)	369.5 (132.4)*	613.1 (141.7)	495.2 (186.9)	510.0 (80.5)
	SUP	585.4 (173.1)	445.1 (171.0)	477.1 (181.8)*	659.2 (149.9)	533.9 (200.5)	614.5 (155.6)
	INF	61.7 (18.0)	46.7 (18.3)	48.5 (20.8)*	71.6 (15.8)	58.2 (21.7)	67.2 (16.8)
	BI	153.4 (57.9)	110.8 (34.0)	136.4 (51.6)	205.7 (52.4)	169.5 (65.3)	182.0 (56.0)
	Res	**1719.1 (523.7)**	**1291.6 (479.7)***	**1381.8 (532.3)***	**2204.9 (509.1)**	**1789.1 (678.2)**	**1986.4 (419.4)**

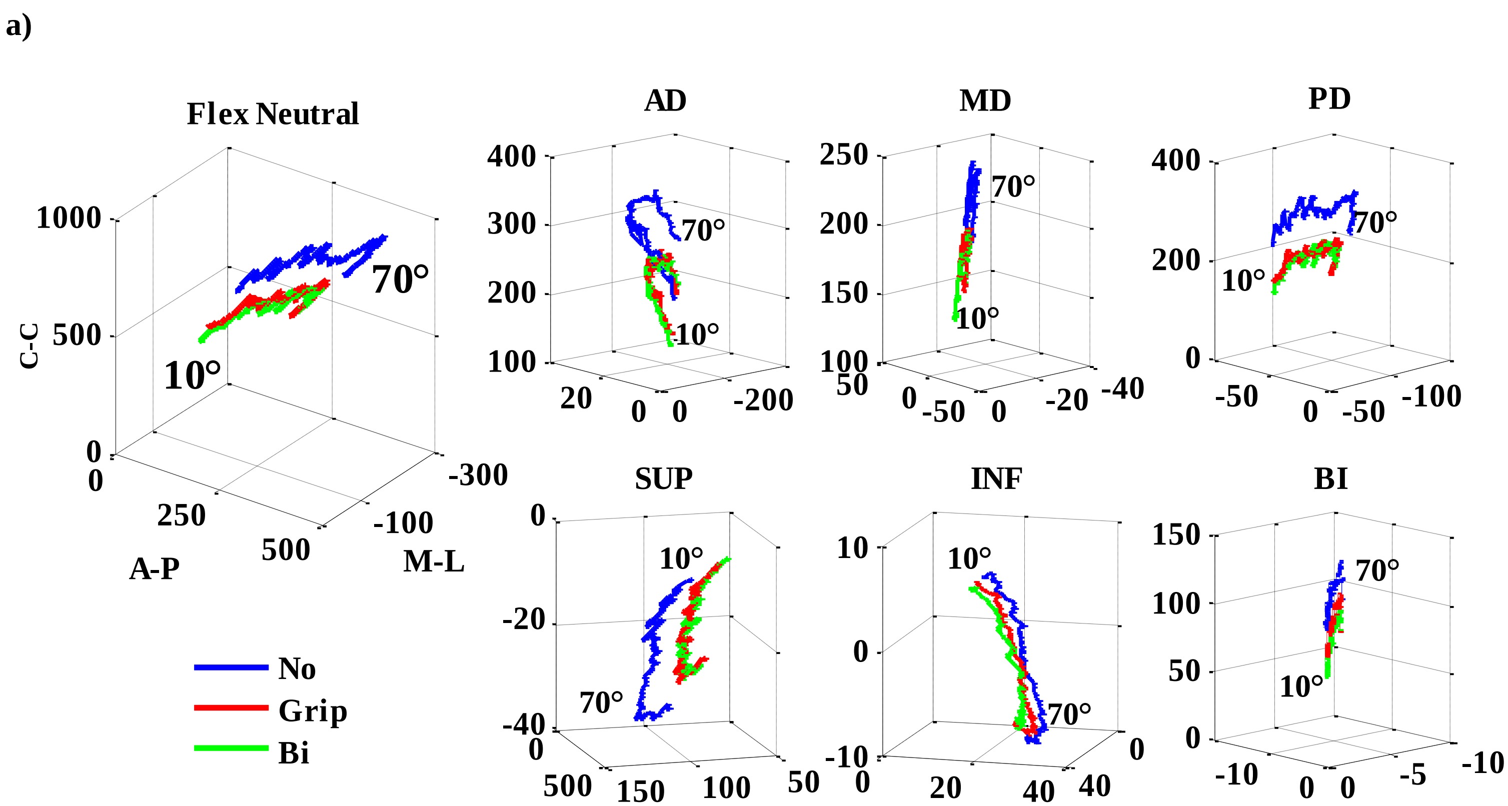
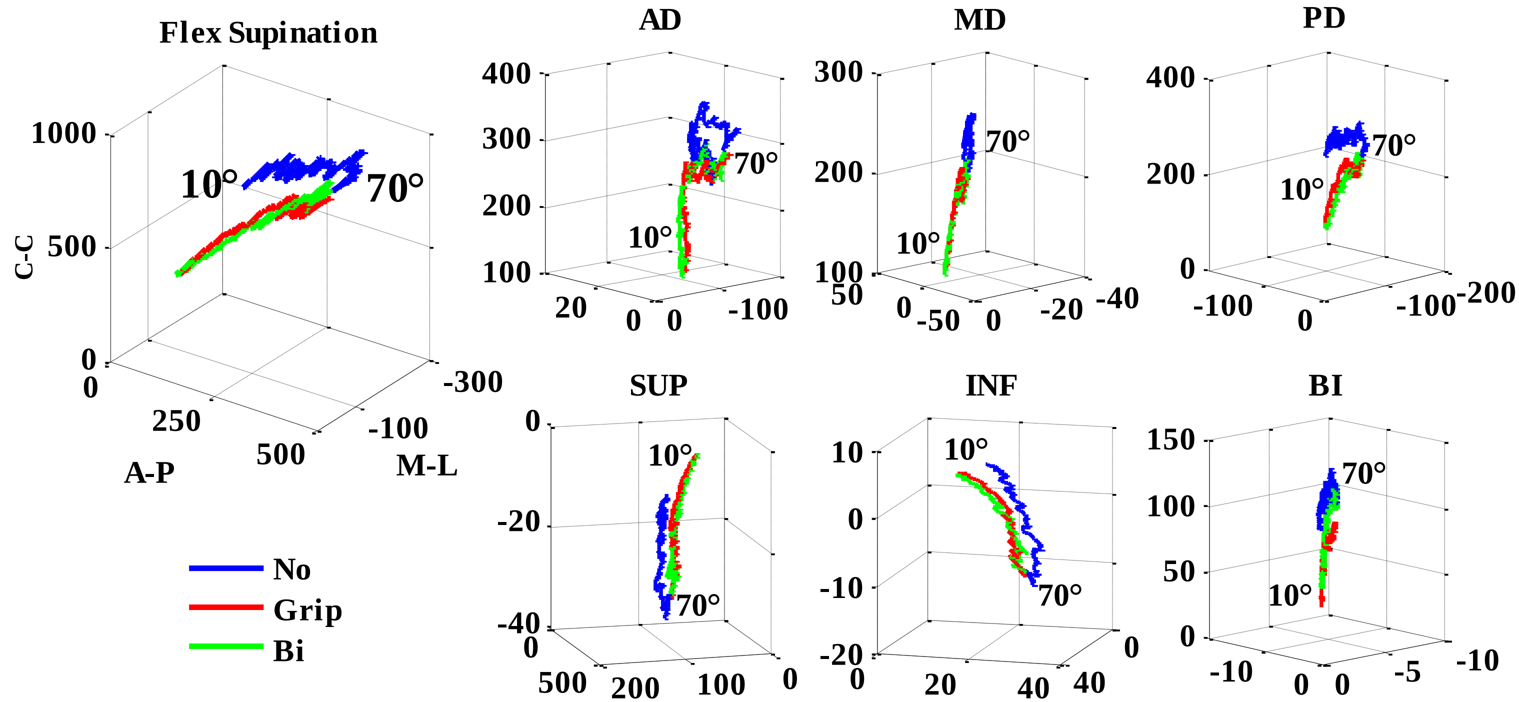
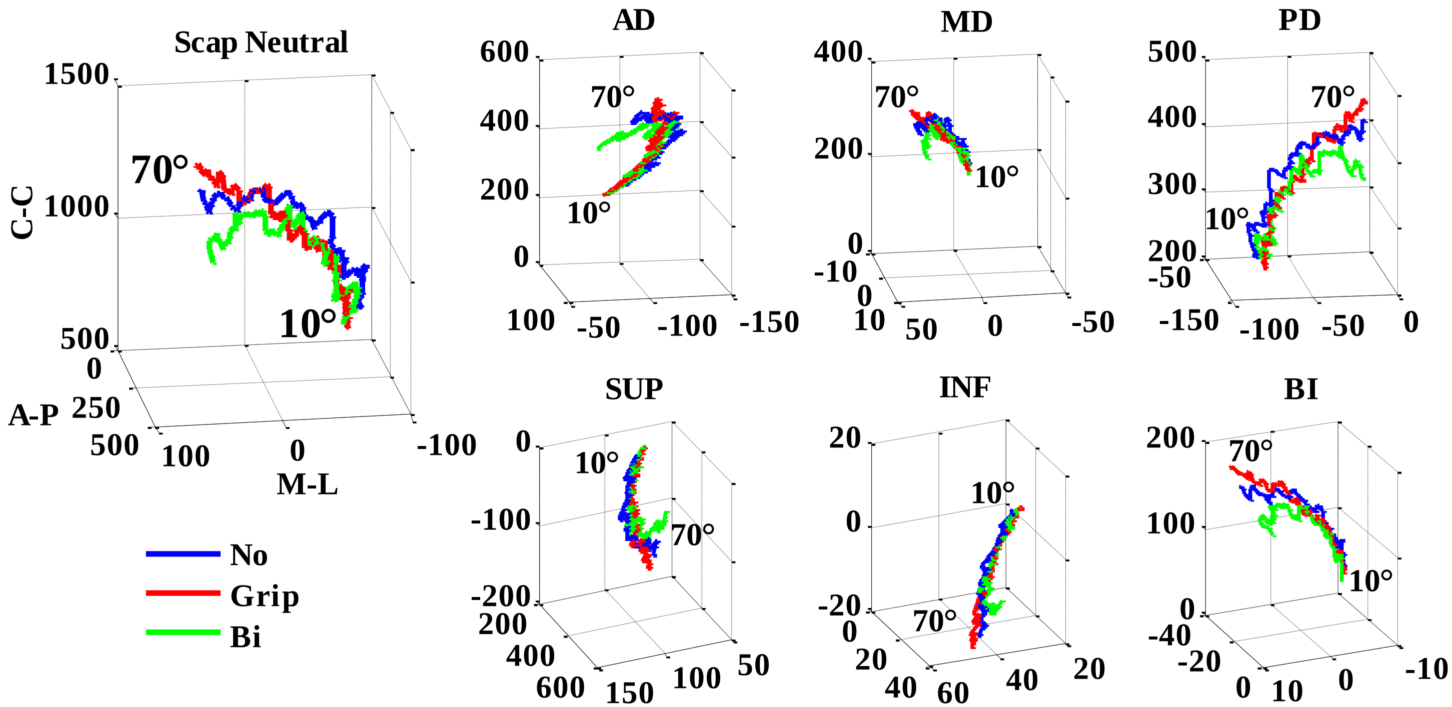


Figure 5.2: Ensemble average of resultant 3D muscle force (larger graph) and individual muscle forces (smaller graphs) across all subjects for: a) flexion with neutral forearm posture; b) flexion with supinated forearm posture; c) scapular plane with neutral forearm posture; d) scapular plane with supinated forearm posture. Cranial-caudal (C-C) force is on the vertical axis (Y), anterior-posterior force on the left horizontal axis (X), medial-lateral (M-L) force is on the right horizontal axis (Z). Exertions performed concurrently with gripping are in red ("Grip"), biceps contractions in green ("Bi") and no load in blue ("No"). The start (10°) and end (70°) of the motion is indicated.

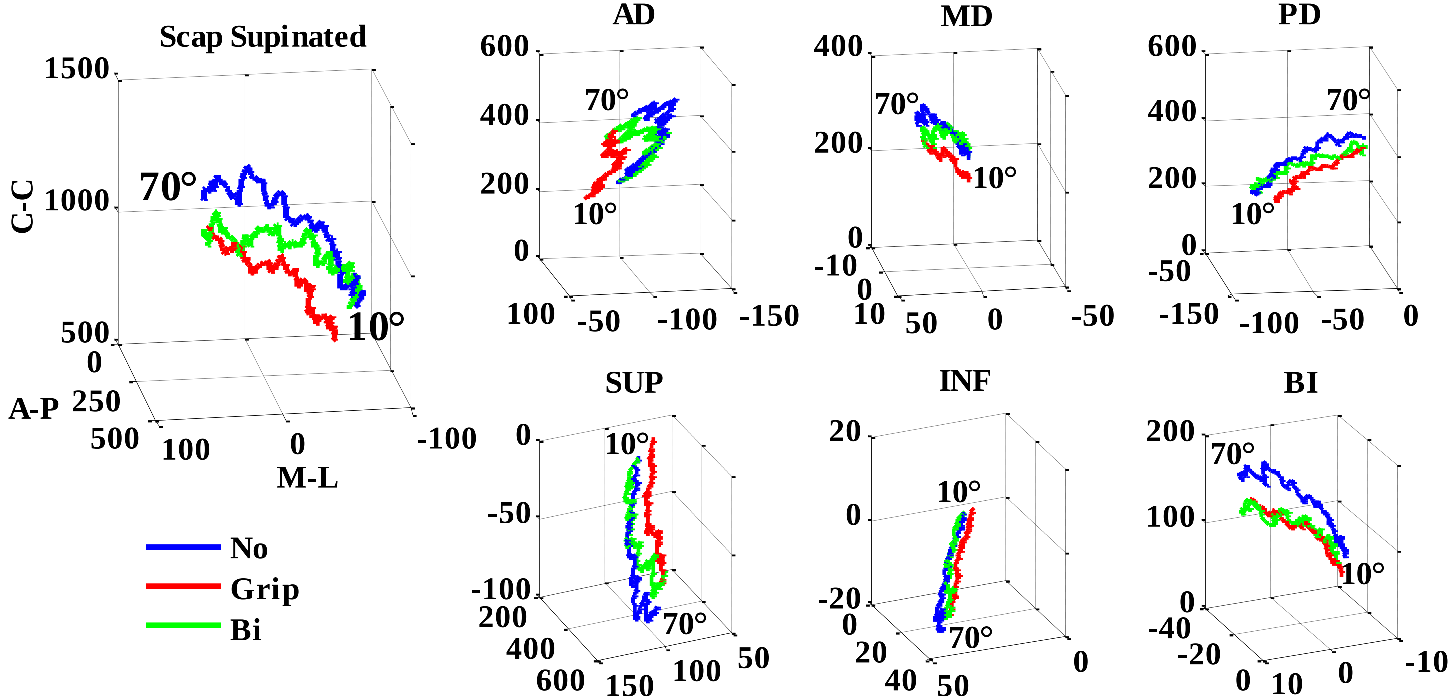
b)



c)



d)


Scap Supinated
C-C
1500
1000
500
70°
10°
A-P
0
250
500
100
0
-100
M-L
No
Grip
Bi
AD
MD
PD
SUP
INF
BI

5.5 Discussion

Both the grip and biceps contraction tasks significantly reduced participants' ability to generate maximal shoulder moments in flexion with both forearm postures and in the scapular plane with neutral forearm posture. The grip task reduced maximal dynamic shoulder moments by 27% to 33% across all postures investigated and support the results of MacDonell and Keir (2005), who also found grip tasks decreased the ability to produce maximal shoulder moments. The biceps contraction task reduced shoulder moments by 27% to 37% in flexion and 12% to 26% in the scapular plane with supinated and neutral forearm postures, respectively. This study also examined the supraspinatus and infraspinatus muscles with fine wire electrodes in response to the gripping and biceps contraction tasks during maximal dynamic elevations of the arm. The addition of a secondary task, either in the form of targeted gripping or biceps contraction appeared to lower muscle forces from supraspinatus, infraspinatus and deltoids throughout the entire motion in flexion exertions, even when grip force was to be higher during the no load condition (Figure 5.2). This differs from the increase in supraspinatus and infraspinatus activity previously found with grip task with sub-maximal shoulder exertions (Sporrong et al, 1996, Sporrong et al, 1995). The neuromuscular response during the grip task differed from that during biceps contractions.

Given previous shoulder muscle activity research, we hypothesized that the bi-articular biceps brachii would play a role in the muscle activity changes seen with gripping. Thus we tested both gripping and biceps contractions concurrent with dynamic shoulder moment. In the current study, gripping to a target force lowered biceps activity by 5.3% to 8.7% MEE when compared to the no load condition (unlike Antony and Keir, 2009). Additionally, participants had difficulty maintaining the 30 ± 1.5% MVG target during the maximum dynamic shoulder efforts which has not been previously found, as mean grip force ranged from 22.8 ± 7.6 % to 26.7 ± 8.5% MVG. This was not anticipated as a similar study investigating maximum shoulder exertions did not report any issues with maintaining grip force, although the grip forces themselves were not reported (MacDonell and Keir, 2005). Despite these unexpected findings, the grip task had consistently resulted in lower maximal shoulder flexion moments, and appeared to lower all muscle forces throughout the motion. With the forearm supinated, AD and resultant forces were significantly lower than the no load condition. Considering that grip force itself was often higher during the no load conditions, these results suggest that the mechanism responsible for changes with targeted gripping may be cognitive in nature, likely due to the attention required for continual online processing of feedback. Many studies have demonstrated that the addition of a cognitive task in various forms to an existing task, alters the activity of specific muscles (Au and Keir, 2007; MacDonell and Keir, 2005; Visser et al, 2002; Finsen et al, 2001) and the ability to generate maximal shoulder moments (MacDonell and Keir, 2005).

The effect of the grip task in the scapular plane was not as consistent. Significant decreases in shoulder moment were found only with a neutral forearm posture as the moments with the forearm supinated were highly variable (71.4 ± 43.2). Furthermore, when assessing the forces throughout the motion, muscle forces in the no load and gripping conditions did not appear to differ with the neutral forearm posture. When supinated, muscle forces during the gripping condition were lower than the no load condition, similar to the pattern seen in flexion. Thus, although not tested statistically, plane and forearm posture seemed to affect the muscular response to targeted gripping. Since the same feedback was used in all conditions, cognitive interference does not fully explain the differences in neuromuscular response to targeted gripping between the two planes and between neutral and supinated forearm posture within the scapular plane. Therefore there must also be a physical component to the mechanism involved.

Biceps activity during the biceps contraction condition was 1.2% to 13.1% MEE greater than the no load condition. Biceps contraction also significantly lowered maximal shoulder moments in the same postures as the grip condition. In flexion with supinated forearm posture, concurrent biceps contraction resulted in significantly lower peak forces from all muscles except the biceps, and also significantly lowered resultant muscle force (Table 5.2). Thus, contracting the biceps had a greater impact on muscle force than gripping, including lowering rotator cuff muscle forces. In the scapular plane, changes with biceps contraction were not as distinct as in flexion. With the supinated forearm posture, biceps contraction did not significantly affect shoulder moment or muscle force despite biceps activity being 5.4% MEE greater than the no load condition. Furthermore, biceps contraction did not appear to affect the muscle force curves in the same manner as in flexion. In the scapular plane, muscle force curves start with forces similar to those in the no load condition, but then deviate and become lower at the end range of the motion. Similar to the grip condition, these differences in muscular response with posture would indicate a physical component to the complex mechanism that is responsible for changes with concurrent biceps contraction and maximal shoulder exertions.

There were a few limitations to the current study. The grip dynamometer was instrumented into a handle for the upper extremity attachment of the Biodex which was the point of contact between the isokinetic dynamometer and the participant. Although the attachment could still be moved without gripping and was intended to replicate typical equipment use, this arrangement likely contributed to the variability seen in grip force. De-coupling the grip force apparatus from the isokinetic dynamometer may have allowed for more precise grip force. Additionally, there were several muscles that cross the glenohumeral joint that were not included in this model. Of particular interest is the subscapularis, however difficulties and risks associated with measuring EMG from this muscle with indwelling electrodes precluded its collection. It is recommended that future analysis include investigation of the subscapularis muscle and other possible contributors to glenohumeral joint loading such as the teres minor and pectoralis muscles.

5.6 Conclusions

There were similarities between the effect of targeted gripping and biceps contraction when performed concurrently with resisted shoulder elevations. Both gripping and biceps contraction significantly lowered maximal shoulder moments and muscle activity dependent forces in the same postures, however, the biceps contraction had a greater impact on muscle forces. This may have been due to greater grip force and biceps activity being found during the biceps contraction condition than the gripping condition. Furthermore, decreased shoulder moments were present during the gripping condition, despite grip force being greater during the no load condition. This may indicate that there is another component that influences the changes in muscle activity and thus force with gripping. Differences were apparent between planes and forearm postures, indicating that there must also be a physical component as well. Overall, while the hypothesized link between gripping and biceps involvement was not supported, the biceps had a greater effect on the neuromuscular control of the shoulder.

5.7 Acknowledgements

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CHAPTER 6: THESIS SUMMARY AND DISCUSSION

6.1 Thesis Summary

The shoulder depends on precise neuromuscular control to balance moments across the joint and maintain joint integrity while still allowing for a broad range of arm movements. The musculature of the shoulder is organized in a manner that is redundant, with multiple muscles being capable of performing the same function. Thus there are many possible coordination strategies that can be engaged which would result in the same outcome. Due to the prevalence of shoulder injuries in today's society and the difficulties in rehabilitating the shoulder, this thesis aimed to examine factors that alter the neuromuscular control of the shoulder muscles.

Arm movements are often coupled with use of the elbow and hand, for example, as we grip and/or contract the biceps while carrying groceries or using a hand tool. The addition of gripping while performing a shoulder moment has been shown to decrease deltoid activity and increase rotator cuff muscle activity, potentially placing the rotator cuff muscles at an increased risk of overload and injury. This thesis provided an examination of the neuromuscular response, via electromyography (EMG), of the deltoid and rotator cuff muscles to additional hand and elbow tasks while exerting the shoulder in multiple arm and forearm postures. It also presented a biomechanical evaluation of the impact of altered patterns of muscle activity on the balance of muscular forces with respect to glenohumeral joint. The first two studies of this thesis laid the groundwork necessary to accurately examine and interpret neuromuscular control strategies investigated in the latter two studies.

As we began to investigate the effect of dual tasks on shoulder muscle activity, two issues surfaced and needed to be addressed. First, it was necessary to determine that the changes in muscle activity seen with concurrent gripping and shoulder exertions were not solely the product of increased cognitive processing associated with the novelty of using feedback to regulate grip force. Studies have shown that performing a cognitive processing task while exerting the shoulder produced similar changes to muscle activity previously seen with concurrent targeted gripping (Au & Keir, 2007; MacDonell & Keir, 2005). Yet, in another study using the same gripping task with pushing and pulling exertions, no alterations to shoulder muscle activity were found (DiDomizio & Keir, 2010). Thus, there was reason to believe that the mechanism responsible for altering muscle activity with gripping is not solely the result of cognitive processing used to regulate grip force, but actually signalled that a physical component may be involved. Using a motor learning approach, it was hypothesized that if the response of the shoulder muscles to targeted gripping were the result of task novelty then an acute bout of practice (in the form of repetition) would thereby lessening the effect on muscle activity overtime. The effect found did not diminish with repetition and lower anterior deltoid, latissimus dorsi and trapezius activity persisted across all repetitions of concurrent targeted gripping and shoulder exertions. These results were encouraging and it was felt that alterations in muscle activity found with targeted gripping were at least in part a manifestation of a complex physical interaction. Evidence from a later study of the thesis (Chapter 5) would support this theory, as gripping affected muscular response in some postures and not others, despite the same feedback being used in all conditions.

Once it was determined that changes to patterns of shoulder muscle activity with gripping could be, at least in part, physical in nature, the question still remained as to how. This set the stage for the progression of the latter two studies of this thesis (Chapters 4 &5). It was hypothesized that the biceps would allow for moments generated from the forearm during gripping to be transmitted across the elbow and to the shoulder. Chapters 4 and 5 were designed to investigate the role of the biceps in strategies of shoulder muscle activation during static and dynamic shoulder exertions. While processing the data presented in Chapter 5 of this thesis, a second issue arose. When EMG from the dynamic contractions were normalized to conventional, muscle specific manually resisted isometric contractions, values greater than 100% frequently occurred. Since the computation of the muscle forces used in chapters 4 and 5 were dependent on muscle activity to be expressed in terms of 0%-100% activation, it was paramount to have an accurate and consistent normalization process. Thus, although this issue was addressed in Chapter 3 of this thesis, the study was actually conducted in response to the need for improved methods of normalization due to its importance in the interpretation of the data presented in Chapters 4 and 5.

Poor normalization procedures for dynamic contractions have been an issue for decades and the literature is wrought with reports of EMG greater than 100% of ‘maximal’ excitation (Decker et al, 1999; Morris et al, 1998; McGill & Sharrat, 1990; Jobe et al, 1984; Clarys et al, 1983). Accurate representation of muscle activity with respect to its maximal voluntary capacity is vital to the correct interpretation of neuromuscular strategies and the impact on joint loading. There were a number of important findings from the second study (Chapter 3). First, maximal dynamic contractions in flexion-extension, abduction-adduction and internal-external rotations were significantly better than manually resisted muscle specific tests at eliciting the greatest excitations from the deltoid, infraspinatus, supraspinatus and trapezius muscles. Second, the maximal dynamic contraction that elicited the greatest excitation for each muscle was not consistent between individuals. For example, in some individuals the highest excitation for the anterior deltoid was achieved during abduction not flexion. Furthermore, the angle at which maximal excitation was achieved varied greatly between individuals and may explain why conventional isometric methods are inadequate when seeking maximal excitation for the purpose of normalizing muscle activity during dynamic exertions. Variability in individuals' motor strategies became a prevalent theme in this thesis.

It is widely recognized that precise coordination of the shoulder musculature is responsible for maintaining shoulder joint integrity and much of the literature has been focused on defining the function and role of individual muscles surrounding the shoulder (Veeger & van der Helm, 2007; Meskers et al, 2004; Inman et al 1944). In other joints of the body, muscles can be easily identified into agonist and antagonist pairs and used to assess neuromuscular control strategies; however, it is not as clearly defined in the shoulder (Hawkes et al, 2011; Madeleine et al, 1999; Kronberg et al, 1990). The organization of the shoulder musculature is complex and examining muscular contributions to glenohumeral joint stability requires more extensive investigation (Hawkes et al, 2011). The approach used to quantify neuromuscular control strategies of the shoulder was to compute activity dependent muscle forces, capturing information regarding the muscle's physiological size and line of action and ultimately direction of force, as defined in the literature (Chapters 4 and 5).

Chapter 4 examined the effects of additional hand and elbow loads on static shoulder exertions in flexion and scapular planes with neutral and supinated forearm posture on muscle activity. Neuromuscular control strategies were examined using a muscle force approach. It was hypothesized that concurrent biceps contraction would produce similar results as seen with gripping, increasing rotator cuff activity and decreasing deltoid muscle activity. However, the distribution of muscle and activity dependent muscle forces varied greatly between individuals and precluded this finding. This study then further examined patterns of muscle force in each individual and it was found that neuromuscular strategies within each participant were consistent across all conditions. Individuality of motor strategies may explain why certain people are predisposed to workplace musculoskeletal injury.

In a continuation of this study, Chapter 5 examined the effects of additional tasks during dynamic shoulder exertions on motor strategies using the muscle force approach in Chapter 4. There were similarities between the effect of targeted gripping and biceps contraction. Both significantly lowered maximal shoulder moments and muscle activity dependent forces in the same postures, however, the biceps contraction had a greater impact on muscle forces. In part, the greater impact of the biceps contraction may have been due to greater grip force and biceps activity being found during the biceps contraction condition than the gripping condition. Interestingly, decreases in shoulder moment were still present during the gripping condition despite grip force being greater during the no load condition, supporting the notion that a cognitive component exists in the mechanism responsible for the decreased neuromuscular response. However, differences were apparent between planes and forearm postures and, since the same feedback and thus cognitive load were used in every condition, it indicated that there must also be a physical component to the mechanism. Although it could not be concluded that the biceps was responsible for the changes previously seen with gripping, it did establish a relationship between the neuromuscular control of the shoulder and the biceps.

In summary, this thesis has made a number of significant contributions to the literature. It improved upon the methods used to normalize EMG from forceful dynamic contractions. The collection of studies in this thesis also provided an alternative method of examining neuromuscular control strategies of the shoulder that were inclusionary of the size and direction of muscle force.

6.2 Main Research Contributions

6.2.1 Neuromuscular control of shoulder muscles in healthy individuals

Chapter 2, which is now in press, was a simple study that confirmed that the neuromuscular response to gripping was not from the novelty of the gripping task. Chapter 3 provided a superior method for normalizing EMG from dynamic contractions, which accurately represents the capacity of the muscle being engaged. Without proper normalization, the implementation of the musculoskeletal model and furthermore the interpretation of the results would not truly reflect what was actually occurring. Chapters 4 and 5 provide a comprehensive examination of neuromuscular control strategies in response to multiple concurrent upper extremity tasks in healthy individuals. Previous research had shown that deltoid activity was lowered when gripping concurrently with shoulder exertions yet, the response of the rotator cuff muscles to gripping was still not clear (Antony & Keir, 2009; Smets et al, 2009; Au & Keir, 2007; MacDonell & Keir, 2005; Sporrong et al, 1996, Sporrong et al, 1995). Due to the anatomical location of the rotator cuff muscles, the infraspinatus is the only muscle that is reliably accessible with surface EMG. Thus, to investigate the other muscles of the rotator cuff, intra-muscular electrodes were necessary. Due to the invasiveness of intramuscular electrodes and the skill set needed to insert them, studies inclusionary of rotator cuff muscle activity via intramuscular electrodes, in particular during dual tasks, has been limited (Hawke et al, 2011; Sporrong et al, 1996, Sporrong et al, 1995). Thus, this thesis provided a more comprehensive evaluation of the shoulder while performing multiple upper extremity tasks.

An innovative approach was used to investigate neuromuscular control of the shoulder which described both the magnitude and direction of shoulder muscle forces and provided a more in depth analysis than investigating the muscle activities alone (Chapter 4). Until this collection of studies, the neuromuscular control of shoulder during dual tasks had only ever been examined via muscles activities (Hawke et al, 2001; Antony & Keir, 2009; Smets et al, 2009; Au & Keir, 2007; MacDonell & Keir, 2005; Sporrong et al, 1996, Sporrong et al, 1995). Chapter 4, in particular, revealed that the range and individuality of neuromuscular strategies elicited with concurrent gripping and biceps contraction. Furthermore, it provided insight into the mechanism responsible for the alterations in muscular response to the dual tasks and was found to have a component that is cognitive in nature, but is also driven by physical factors.

Inter-individual variability in muscle recruitment patterns of a healthy population had only been documented in the lumbar muscles (Nussbaum & Chaffin, 1997) and thus Chapter 4 of this thesis is the first to document it in the shoulder. Shoulder muscle recruitment patterns varied between individuals (Chapters 4 and 5) and when more closely examined in the scapular plane, three patterns of deltoid force were identified. Some individuals predominantly activated the anterior deltoid, while others predominantly recruited the infraspinatus. Some used all three heads of deltoid, the infraspinatus and supraspinatus to an equal extent. It is possible that the magnitude of the effect of gripping on muscle activity is dependent on the individual's recruitment strategy.

In the interest of remaining focused on the effects of gripping and biceps contraction in Chapters 4 and 5, the statistical analysis of the effect of plane and forearm posture were not presented. However, it offered some valuable findings as supinated forearm posture was found to increase muscle activity (Chapter 4) and activity dependent force (Chapters 4 and 5), regardless of plane of movement or load condition. At the shoulder, forearm supination would most greatly alter the line of action of the biceps (DeLuca & Forrest, 1973; Inman et al, 1944), thus providing more reason to believe that the biceps greatly affects the coordination of shoulder muscles, as was seen in Chapter 5.

The role of the biceps and its relationship with the shoulder muscles needs to be taken into consideration when assessing risk of injury to the shoulder.

6.3 Future Directions

It is not yet fully understood how proprioceptive feedback regarding the direction of forces in glenohumeral joint is provided to the nervous system. Some have suggested that proprioceptors must exist within the joint and that the most suitable location would be in or around the glenoid labrum (Veeger and van der Helm, 2007). Others suggest that proprioceptors exist in glenohumeral ligaments and/or the labrum (Steinbeck et al, 2003Gohkle et al, 1998; Vangsness et al, 1995). Perhaps most plausible is that the muscles themselves assume the responsibility of preserving joint integrity, in particular during mid-range postures (Lippitt et al, 1993). Having faster response times than proprioceptors, mechanoreceptors such as muscle spindles and Golgi tendon organs are a viable option for providing the required feedback necessary to maintain joint integrity. The number of multi-pennated muscles that surround the shoulder joint provides ample opportunity for feedback thus allows for precise control of movement. The muscle force approach used in this thesis should be expanded upon to include more muscles and be scaled to each individual. Further examination of how the neuromuscular system controls the shoulder is necessary in order to fully understand how injury may interrupt this communication.

A significant finding of this collection of studies was the individuality of shoulder neuromuscular control strategies. Differences found in the plane of elevation that elicited maximal excitation from the anterior deltoids (Chapter 3) and the different recruitment strategies seen in Chapter 4 prompts further examination into the recruitment strategies of deltoids in a larger group of individuals. Knowledge of these different recruitment strategies, how many there are, and what proportion of the population uses each strategy, would provide greater insight into injury risk and the necessity of individualized treatment of injury.

Optimally, one might investigate the pattern of shoulder muscle activation in a large number of workers with labour intensive upper extremity jobs and follow them over their careers, tracking who developed shoulder injuries and who did not. It may then be possible to classify whether specific motor control strategies result in certain individuals being more susceptible to shoulder injury than others. Logistically, such a study would be difficult to conduct and, thus, using a fatigue or induced pain model may serve as a surrogate to determine if compensatory patterns of muscle activity are found and replicate those found in the injured individuals.

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APPENDICES

APPENDIX A: Addendum to line of research for ethics approval of Study 1:

McMaster University Research Ethics Board (MREB)
c/o Office of Research Services, MREB Secretariat, GH-305/H, e-mail: ethicsoffice@mcmaster.ca
CERTIFICATE OF ETHICS CLEARANCE TO INVOLVE HUMAN PARTICIPANTS IN RESEARCH

Application Status: New ☑ **Addendum** ☐ **Renewal** ☐ **Project Number** 2007 148

TITLE OF RESEARCH PROJECT:

Work-related Upper Extremity Loading (Line of Research)

Name(s)	Dept./Address	Phone	E-Mail
Faculty Investigator(s)/ Supervisor(s)			
P. Keir	Kinesiology	23543	pjkeir@mcmaster.ca
Student Investigator(s)			

The application in support of the above research project has been reviewed by the MREB to ensure compliance with the Tri-Council Policy Statement and the McMaster University Policies and Guidelines for Research Involving Human Participants. The following ethics certification is provided by the MREB:

☐ The application protocol is approved as presented without questions or requests for modification.
☐ ~~The application protocol is approved as revised without questions or requests for modification.~~
ΝΜ ☑ The application protocol is approved subject to clarification and/or modification as appended or identified below:

COMMENTS AND CONDITIONS: Ongoing approval is contingent on completing the annual completed/status report. A "Change Request" or amendment must be made and approved before any alterations are made to the research.

See attached.

Reporting Frequency:	Annual:	Other:

Date: Jan. 7, 2008 **Dr. D. Maurer, Chair/ Dr. D. Pawluch, Vice-chair:** D Maurer

APPENDIX B: Ethics approval for Study 2, 3 & 4 (Chapters 3, 4 & 5):

Zoom out (Ctrl+Minus)

McMaster University

Hamilton Health Sciences/McMaster FEB 18 2010 Research Ethics Board

PARTICIPANT INFORMATION SHEET

Title of Study: The effect of hand and arm actions on muscle activity and load distribution in the shoulder complex

Joanne Hodder MSc, PhD Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Andrew Longbottom, BSC Kin Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Kristina Calder, PhD, School of Rehabilitation Science, Faculty of Health Sciences, McMaster University

Peter J Keir, PhD, Associate Professor, Department of Kinesiology, Faculty of Science, McMaster University

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You are being invited to participate in *one of two research related studies* conducted by Joanne Hodder and Andrew Longbottom because you are a healthy male from the university population. The study will help us to learn more about the coordination of shoulder muscles during static shoulder contractions when combined with gripping and biceps contractions.

In order to decide whether or not you want to be a part of this research study, you should understand what is involved and the potential risks and benefits. This form gives detailed information about the research study, which will be discussed with you. Once you understand the study, you will be asked to sign this form if you wish to participate. Please take your time to make your decision. Feel free to discuss it with your friends and family.

WHY IS THIS RESEARCH BEING DONE?

Shoulder injuries are a common workplace injury. We need to better understand these injuries in order to prevent and rehabilitate them. Combining shoulder and hand grip efforts are used frequently in the workplace, for example using a hand tool. However, this combination of tasks have been shown to change how muscles of the shoulder work, such that the larger muscles were not working as hard and the smaller muscles having to work even harder. The small muscles of the shoulder are the most often injured and increased effort may result in greater risk of injury for these muscles.

WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to measure the effort of all the muscles surrounding the shoulder during combined shoulder and gripping tasks to better understand how the change in muscle activity is occurring.

WHAT WILL MY RESPONSIBILITIES BE IF I TAKE PART IN THE STUDY?

If you volunteer to participate in this study, we will ask you to do the following things:

STUDY PROTOCOL

In these studies, we are interested in measuring muscle activity during various tasks using the arm and hand. To measure muscle activity we use two types of electrodes. Surface electrodes will be used to measure the superficial muscles of the shoulder and upper arm, and fine wire electrodes will be used to measure the deep muscles of the shoulder in one study. These electrodes will only monitor the electrical activity of the muscle of interest and will not transmit an electrical signal to the body. A needle will be used to place the fine wire electrodes; however, nothing else will be injected except the fine wire electrode with the needle.

Consent Form Date: January 28, 2010 Page 1 of 4 Protocol # and version date: FW1-A, Feb 2, 2010
Subject Initials:__________

Surface electrodes are small circular self adhesive pads with a conductive gel in the middle. The skin over each muscle of interest will be cleaned with alcohol and two surface electrodes will be placed. For this study, the following muscles of the right side of the participant will be investigated with surface electrodes: three heads of deltoid, trapezius, latissimus dorsi, pectoralis major, biceps and triceps. Fine wire electrodes will be used to investigate three rotator cuff muscles, the supraspinatus, infraspinatus and teres minor muscles. These wire electrodes will be inserted into the muscle being measured with a very fine gauge sterile, single use needle. Once the fine wire electrode is in place and then needle is removed and disposed of and the emerging wire affixed to the skin using medical tape to prevent unwanted movement of the wire electrode. The fine wire will remain in the muscle until the end of collection at which point it will be removed by gently pulling on the wire.

Once electrodes are placed, the participant will complete maximum efforts for each muscle being investigated. These will be static efforts against resistance to prevent movement of the fine wire electrodes. Next a maximal grip effort will be measured using a hand dynamometer and maximum shoulder moment measured on an isokinetic dynamometer (Biodex). An isokinetic dynomometer is a piece of equipment that will provide resistance to a participants movement, in the case of this study, it will resist against arm elevation.

Study A. Hand and Muscle actions on Shoulder Activity.
The study protocol will consist of three tasks in eight arm postures for a total of 24 static tasks and four slow dynamic movements. The postures include 30°, 60°, 90° and 120° degrees of arm elevation to the front of the body (frontal plane) and 30° to the side of the body (scapular plane). In each posture the participant will perform a static shoulder effort at 40% of their maximum shoulder moment. Combined with this shoulder effort will be a 30% grip effort, a 25% biceps static contraction effort or both for a total of 24 static tasks.

After the static trials are completed, four dynamic trials will be conducted. For these trials, you will raise and lower your arm at 20° per second for 4 seconds from 0° to 120° and back to 0° of elevation. These will be performed in the frontal plane and scapular planes. During these trials a 40% shoulder moment will be maintained. In each plane, one trial will be performed with the 30% grip effort and one without any grip effort.

Study B. Work Simulation Tools and Shoulder Activity
The work group will follow a similar protocol for MVE collection and collect maximal efforts at 30°/s in abduction/adduction (shoulder raise/lower to side), internal and external rotation, and flexion/extension (shoulder raise/lower to front) on the Biodex 4 isokinetic dynamometer. Follow this, trials will be conducted using the "work simulation" attachments for the Biodex 4 system. The actions will include turning a steering wheel, turning a screwdriver while seated and raising the lift attachment while standing. Each task will be completed at slow (30°/s) and fast (180°/s) in a continuous 15-20 second trial in which 5 repetitions will be completed

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

It is not possible to predict all possible risks or discomforts that participants may experience in any research study. The present investigator anticipates no major risks or discomforts will occur in the current study. It is important however to recognize the following potential risks and discomforts that may be incurred.

1. You may experience mild discomfort or skin irritation from being shaved and cleansed in preparation for electrode placement. This is usually very mild and clear within 24 hours.
2. There may be discomfort related to the delayed onset of muscle soreness associated with maximal and isometric contractions of the arm muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.
3. Infection & Bruising. Placement of a needle electrode into the muscle can be somewhat uncomfortable. If pain is experienced, the needle will be moved to another area. Like any foreign body, the needle electrode poses a risk of infection. The needle electrodes to be used in this investigation will be sterile and used only once. This is a common procedure used without incident other than the chance of slight bruising that sometimes occurs with a needle insertion.
4. Maximal effort isometric contractions are associated with an increase in blood pressure. If you have received medical clearance and/or are already physically active, the risks are minimal. The researchers' first priority as an investigator is to maintain the emotional, psychological, and physical health of those participating in the study.

HOW MANY PEOPLE WILL BE IN THIS STUDY?

Thirty males from the University population will be recruited for this study.

WHAT ARE THE POSSIBLE BENEFITS FOR ME AND/OR FOR SOCIETY?

Participants will receive no direct benefits from participating in this study. However, participants should know that their willingness to serve as a subject for this experiment will help develop knowledge of mechanisms of injury to the rotator cuff which may benefit individuals in the future.

IF I DO NOT WANT TO TAKE PART IN THE STUDY, ARE THERE OTHER CHOICES?

Participation in this study is voluntary. Refusal to participate will not result in loss of access to any services or programs at McMaster University to which you are entitled. You will inform the investigator, Joanne Hodder, MSc, of your intention to withdrawal at any point during this study.

WHAT INFORMATION WILL BE KEPT PRIVATE?

Your data will not be shared with anyone except with your consent or as required by law. All personal information such as your name, address, phone number or email will be removed from the data and will be replaced with a number. A list linking the number with your name will be kept in a secure place, separate from your file. The data, with identifying information removed will be securely stored in a locked office in the research office and on an encrypted hard drive. The data for this research study will be retained for ten years.

For the purposes of ensuring the proper monitoring of the research study, it is possible that a member of the Hamilton Health Sciences/FHS McMaster University Research Ethics Board, a Health Canada representative may consult your research data. However, no records which identify you by name or initials will be allowed to leave the institution/university/hospital. By signing this consent form, you authorize such access.

If the results of the study are published, your name will not be used and no information that discloses your identity will be released or published without your specific consent to the disclosure.

CAN PARTICIPATION IN THE STUDY END EARLY?

If you volunteer to be in this study, you may withdraw at any time. You have the option of removing your data from the study. You may also refuse to answer any questions you don't want to answer and still remain in the study. The investigator may withdraw you from this research if circumstances arise which warrant doing so.

WILL I BE PAID TO PARTICIPATE IN THIS STUDY?

If you agree to take part, we will reimburse you $ 30 for your time *for Study A and $20 for Study B.*

WILL THERE BE ANY COSTS?

Your participation in this research project may involve additional costs of parking for the duration of the study collection (approximately 2 hours).

WHAT HAPPENS IF I HAVE A RESEARCH-RELATED INJURY?

If you are injured as a direct result of taking part in this study, all necessary medical treatment will be made available to you at no cost. Financial compensation for such things as lost wages, disability or discomfort due to this type of injury is not routinely available.

However, if you sign this consent form it does not mean that you waive any legal rights you may have under the law, nor does it mean that you are releasing the investigator(s), institution(s) and/or sponsor(s) from their legal and professional responsibilities.

IF I HAVE ANY QUESTIONS OR PROBLEMS, WHOM CAN I CALL?

If you have any questions about the research now or later, please contact Joanne Hodder at 905-525-9140, ext. 21334 or Dr. Peter Keir at 905-525-9140, ext.23543.

Subject Initials:___________

If you have any questions regarding your rights as a research participant, you may contact the Office of the Chair of the Hamilton Health Sciences/Faculty of Health Sciences Research Ethics Board at 905-521-2100, ext. 42013.

CONSENT STATEMENT

SIGNATURE OF RESEARCH PARTICIPANT/LEGALLY-AUTHORIZED REPRESENTATIVE*

I have read the preceding information thoroughly. I have had the opportunity to ask questions, and all of my questions have been answered to my satisfaction. I agree to participate in this study. I understand that I will receive a signed copy of this form.

____
Name of Participant

____
Name of Legally Authorized Representative

______________________________ ____________
Signature of Participant (**or Legally Authorized Representative**) Date

Consent form administered and explained in person by:

____
Name and title

______________________________ ____________
Signature Date

SIGNATURE OF INVESTIGATOR:

In my judgement, the participant is voluntarily and knowingly giving informed consent and possesses the legal capacity to give informed consent to participate in this research study.

______________________________ ____________
Signature of Investigator Date

Hamilton Health Sciences/McMaster
FEB 18 2010
Research Ethics Board