SUPRASPINATUS CONTRIBUTION AND PROPRIOCEPTIVE BEHAVIOR AT THE SHOULDER

by

DAVID ALAN PHILLIPS

A DISSERTATION

Presented to the Department of Human Physiology
and the Graduate School of the University of Oregon
in partial fulfillment of the requirements
for the degree of
Doctor of Philosophy

September 2017

DISSERTATION APPROVAL PAGE

Student: David Alan Phillips

Title: Supraspinatus Contribution and Proprioceptive Behavior at the Shoulder

This dissertation has been accepted and approved in partial fulfillment of the requirements for the Doctor of Philosophy degree in the Department of Human Physiology by:

Andrew Karduna	Chairperson
Michael Hahn	Core Member
Brain Dalton	Core Member
Roland Good	Institutional Representative

and

Sara D. Hodges	Interim Vice Provost and Dean of the Graduate School

Original approval signatures are on file with the University of Oregon Graduate School.

Degree awarded September 2017

DISSERTATION ABSTRACT

David Alan Phillips

Doctor of Philosophy

Department of Human Physiology

September 2017

Title: Supraspinatus Contribution and Proprioceptive Behavior at the Shoulder

Shoulder complaints constitute a significant portion of primary care visits each year in the US, costing $7 billion in annual medical treatment. Shoulder complaints arise from some type of trauma caused by muscle imbalances, proprioception, overuse, anatomical or a combination of these factors. More than two thirds of complaints involve the rotator cuff. Literature regarding shoulder mechanics and proprioception is mixed and with contradictory results. This may be the reason for the high incidence and low success rate in treating shoulder complaints. Here the contributions of the supraspinatus muscle to humeral elevation, and shoulder proprioception are investigated. The results of this dissertation are applicable to developing shoulder injury treatment and preventative strategies, computational shoulder models, and understanding proprioception at the shoulder.

This dissertation includes previously published and unpublished co-authored material.

CURRICULUM VITAE

NAME OF AUTHOR: David Alan Phillips

GRADUATE AND UNDERGRADUATE SCHOOLS ATTENDED:

University of Oregon, Eugene
Barry University, Miami Shores
University of Pretoria, Pretoria

DEGREES AWARDED:

Doctor of Philosophy, 2017, University of Oregon
Master of Science, 2013, Barry University
Bachelor of Arts (Honors), 2010, University of Pretoria

AREAS OF INTEREST

Injury Prevention
Neuromechanics of the Shoulder

PROFESSIONAL EXPERIENCE

Graduate Employee, Department of Human Physiology, University of Oregon, Eugene, 2013-2017

Graduate Research Assistant, Department of Human Performance and Leisure Sciences, Barry University, Miami Shores, 2012-2013

PUBLICATIONS

Phillips D, Karduna A (*In Press*). Deltoid EMG is reliable during submaximal isometric ramp contractions. *Journal of Applied Biomechanics*

Phillips D, Karduna A (*In Press)*. No relationship between joint position sense and force sense at the shoulder. *Journal of Motor Behavior*

ACKNOWLEDGEMENTS

I have really enjoyed being a part of Andy's lab. I learnt far more of what I should do than what I should not do here. I have become more focused and enthralled by science than ever before. This very well demonstrates what a great mentor Andy has been. The best advice he said was not do what others have done but do what I want to do. This comment will likely drive my research philosophy and agenda for the rest of my career. Thank you Andy.

I was also fortunate to enter the lab with two well settled senior graduate students – Jacqi and Yin-Liang. Both spent a lot of time with me. Going through complex statistics, teaching and research problems. But it was the little things, like graph editing, class registration, finding the best place to eat, that I will remember. Having good peers as mentors was a huge factor in gaining more confidence as a scientist. I also want to acknowledge the cohort of people that started at UO with me. Their support, friendship and drinks out made graduate school far less daunting.

I want to specifically mention Dr Peter Kosek who is the first clinician I have worked with on any project. Without him donating time and resources, half this dissertation would not have been possible. I want to thank the rest of my dissertation committee – Dr Brian Dalton, Dr Michael Hahn and Dr Roland Good – for advising me throughout the project.

Lastly, I would like to thank my family. My parents and sister for supporting my path through graduate school – it would not have been possible without you. Most importantly, I want to thank my wife who took a gamble and moved from Florida to Oregon with me. I love you and could not have done this without you.

TABLE OF CONTENTS

Chapter Page

LIST OF TABLES

LIST OF FIGURES

CHAPTER I

INTRODUCTION

Shoulder complaints account for an estimated 1% of all primary care visits each year (Linsell et al., 2006). Up to 81% of all shoulder injuries are soft tissues lesions with 65% involving the rotator cuff (Vecchio, Kavanagh, Hazleman, & King, 1995). Of elderly individuals that develop shoulder pain or an injury at the shoulder, only 1 in 2 are pain free 6 months later, with only a slight improvement at one year (Chard, Hazleman, Hazleman, King, & Reiss, 1991) . As many as one third of individuals where shoulder pain is present have difficulty performing tasks of daily living like cleaning or feeding (Hasvold & Johnsen, 1993). Annual direct medical expenses in the USA are estimated at $7 billion (Meislin, Sperling, & Stitik, 2005). Risk identification, prevention and effective treatment of shoulder pain and injuries may help reduce the economic cost and the loss of individual independence.

The rotator cuff consists of four muscles: the supraspinatus, infrspinatus, teres minor and subscapularis (Moore, Dalley, & Agur, 2014). The primary function of the rotator cuff muscles is to stabilize the humeral head in the glenoid cavity of the scapula (Oatis, 2009). These muscles also assist with motion of the upper arm. Specifically the supraspinatus contributes to shoulder elevation (de Witte et al., 2014). The supraspinatus is of specific interest in shoulder pain and injury, as more that half of all rotator cuff tears involve the supraspinatus (Zbojniewicz, Maeder, Emery, & Salisbury, 2014). Overuse, dysfunction, muscle imbalances, trauma, anatomy, poor proprioception or a combination of these factors can be responsible for the development of rotator cuff tears and other shoulder pathologies (Michener, McClure, & Karduna, 2003; Parsons et al., 2007). Proprioception, or awareness of the position of a limb in space, may initially be poor but can be further compromised by injury (Blasier, Carpenter, & Huston, 1994). This creates a vicious positive feedback cycle of recurring injury and ultimately leading to loss of function (Borsa & Lephart, 1994). This in turn is a burden to the economy through lost work days and the cost of treatment (Meislin et al., 2005). Proprioception deficits may therefore be an indicator of possible shoulder injury risk and, if corrected, could prevent an injury from occurring. Thus, the purpose of this dissertation is to investigate proprioception, neuromuscular control and supraspinatus mechanics in the shoulder. The goal of this research is to quantify supraspinatus function and proprioception behavior at the shoulder. This information will help clinicians develop more effective treatment protocols and identify individuals at risk of shoulder injuries sooner.

This dissertation consists of two parts: supraspinatus biomechanics and proprioception both discussed further in this introduction. Each part has two experiments, with four experiments in total. The experiments for supraspinatus biomechanics are presented in Chapter II and Chapter III. The experiments on proprioception are presented in Chapter IV and Chapter V. Concluding remarks are presented in Chapter VI.

Part I: Supraspinatus Biomechanics

The skeletal anatomy and muscles crossing the GH joint allow three degrees of freedom of movement and the greatest range of motion of any joint in the body. This large range of motion is achieved through the low level of skeletal constraints and from the multifuctionality of each muscle. For instance, the deltoid is divided up into the anterior, middle and posterior deltoid. The anterior deltoid assists with internal rotation, the posterior deltoid assists with external rotation and the all three portions contribute to elevation. Likewise the supraspinatus has a role in both elevation and stabilization at the shoulder (de Witte et al., 2014). While the multifunction aspect of each muscle is beneficial to joint function, it makes analysis of the joints difficult. A muscle's function depends on the position of the limb, due to changes in mechanical advantage (moment arm), muscle fiber length and fiber orientation.

At 0° elevation the deltoid has a small moment arm, so high forces in the muscle are needed to generate joint torque (Ackland, Pak, Richardson, & Pandy, 2008; Yanagawa et al., 2008) compared to higher angles, where the moment arm is greater (Ackland et al., 2008). At low elevations angles, the insertion of the deltoid creates a more vertical force applied to the humerus in comparison to the medially directed supraspinatus force (de Witte et al., 2014). This can result in superiorly directed shear forces due to the insertion angle of the deltoid (Yanagawa et al., 2008).

These forces are counteracted by the rotator cuff muscles with relatively large moment arms at low angles of shoulder elevation (Ackland et al., 2008; Liu, Hughes, Smutz, Niebur, & Nan-An, 1997; Otis et al., 1994). The rotator cuff muscles generate compressive forces, which keep the humeral head in the glenoid fossa and prevent superior translation, and are primarily responsible for dynamic stabilization (Apreleva, Parsons, Warner, Fu, & Woo, 2000; Halder, Zhao, O'Driscoll, Morrey, & An, 2001; Yanagawa et al., 2008). The humeral head may translate superiorly if the deltoid and rotator cuff coordination pattern is compromised (J. S. Juan, Kosek, & Karduna, 2013). This superior translation has been implicated in the development of shoulder disorders such as shoulder impingement syndrome (SIS) (Jørgensen & Andersen, 2010; Michener et al., 2003).

The deltoid and the supraspinatus muscles both contribute to shoulder elevation. Other minor contributors include the pectoralis major, subscapularis and infraspinatus, depending on plane and angle of elevation (Ackland et al., 2008; Gatti, Dickerson, Chadwick, Mell, & Hughes, 2007). Both the deltoid and supraspinatus muscles are able to elevate the arm if the other is unavailable (Howell, Imobersteg, Seger, & Marone, 1986), but normal movement is brought about through synergistic action between these muscles. Based on modeling studies, the majority of the torque generated during elevation comes from the deltoid muscle. However, under a suprascapular or axillary nerve block, where the supraspinatus or deltoid is paralyzed respectively, there was a ~50% drop in maximal torque (Howell et al., 1986). This has been interpreted as both muscles contributing equally to shoulder elevation. Howell et al., (1986) utilized a maximal 60°/s isokinetic contraction to determine shoulder torque through shoulder elevation, while Yanagawa et al. (2008) modeled the muscles on an unloaded elevation movement. The inconsistent results may be due to the differing methods of measurement. The muscles may have behaved differently due their inherent multifunctional nature or due to the nature of the imposed demands.

In circumstances where both the suprascapular nerve and axillary nerve are blocked in the same arm, paralyzing the supraspinatus and deltoid, subjects are unable to lift their arm in the scapular plane (Howell et al., 1986). The elevation torque generated by muscles other than the supraspinatus or deltoid is therefore less than the torque due to the weight of the arm. Other potential muscles which may contribute to elevation (based on moment arms) are the upper subscapularis and infraspinatus at low elevation angles and the superior pectoralis major at high angles (>60°) (Ackland et al., 2008). While it has been demonstrated that a loss of either the supraspinatus or deltoid results in a ~50% reduction in maximal shoulder elevation torque

(Howell et al., 1986), this relationship has not been quantified at submaximal levels, where activities of daily living occur. Further justification that maximal contraction levels may not represent submaximal function of the supraspinatus comes from electromyography. While deltoid activation increases as higher, submaximal loads are applied, supraspinatus activation remained the same (de Witte et al., 2014). Since increasing EMG correlates with increased for within a muscle (Suzuki, Conwit, Stashuk, Santarsiero, & Metter, 2002), supraspinatus for would not have increased. A final note on the mechanical behavior of the supraspinatus is that many anatomical texts list its function as an initiator of shoulder abduction (Moore et al., 2014). Evidence to support this is lacking and recent EMG data show that the deltoid and supraspinatus activate simultaneously (Reed, Cathers, Halaki, & Ginn, 2013).

It has been consistently demonstrated that as force or torque increases, so does EMG activity (Milner-Brown & Stein, 1975; Moritani & DeVries, 1978; Suzuki et al., 2002). Muscles are activated in a consistent pattern leading to a repeatable relationship between external load and an individual muscle's EMG signal during submaximal isometric contractions (Yang & Winter, 1983). Interestingly, in a suprascapular nerve block model, compensatory deltoid activation has been observed when generating the same level of torque before the block (McCully, Suprak, Kosek, & Karduna, 2007). The deltoid compensates for the loss of the torque that would have been generated by the supraspinatus. When the deltoid is activated to the same amplitude after a suprascapular nerve block the amount of elevation torque measured externally, with a biodex or load cell, will have decreased. This is due to the loss of the supraspinatus muscle torque contribution. The difference between the pre and post block external torque measures at a specific deltoid EMG amplitude would give a quantitative measure of the contribution of the supraspinatus muscle.

However, it is not possible to predict the load needed to elicit a specific EMG amplitude after the suprascapular nerve block. This makes using the proven, reliable method of isometric contractions based on maximal voluntary contraction (MVC) difficult (Yang & Winter, 1983). A method that allows continuous external load recording and EMG activity would not require that a specific target be predicted post nerve block. An isometric ramp contraction would therefore be best to measure EMG and external torque. As this is a new method, reliability between deltoid EMG and external torque during an isometric ramp contraction will need to be established.

The supraspinatus portion of this dissertation has two aims. Aim 1 is to establish the reliability of deltoid EMG during isometric ramp contractions (Chapter II). Aim 2 is to determine the relative contribution of the supraspinatus to shoulder elevation (Chapter III).

Part II: Shoulder Proprioception

Proprioception is the awareness of a limb's position, posture or status in space without visual confirmation. Afferent information from peripheral receptors to the CNS regarding joint position, kinesthesia and sense of resistance or force are considered proprioceptive sub-modalities. The information from these three sub-modalities of proprioception is integrated in the CNS, which helps to produce efferent output commands to muscles resulting in appropriate and coordinated limb movements that accomplish a task. The afferent information arises from Golgi tendon organs and muscle spindles in the muscle, mechanoreceptors in the joint capsule, ligaments and soft tissue surrounding the joint (Riemann & Lephart, 2002).

The afferent proprioceptive information travels to three different structural levels of the CNS control. The first are reflex arcs at the spinal cord level. These are thought to contribute to rapid joint stabilization for movements too fast for the information to reach the brain (Kandel, Schwartz, & Jessell, 2000). The second is the brain stem, which integrates the proprioceptive information with visual and vestibular information to control automatic movements in posture and balance. Lastly the proprioceptive information terminates in the somatosensory cortex and is involved with motor planning, strategy (open loop control) and movement adjustments in the premotor and primary motor cotices (Kandel et al., 2000).

Proprioception can be assessed through force reproduction (Docherty & Arnold, 2008) and joint position sense (JPS) (Lephart, Myers, Bradley, & Fu, 2002) methods. Sensory information during proprioception tasks comes from receptors in the muscle, musculotendinous junction, the skin and the joint capsule (Riemann & Lephart, 2002). Each detects a specific type of stimuli and relays information to the CNS. The CNS is thought to rely primarily on the afferent information from muscle spindles for proprioception. This has been determined from studies in joint replacement surgery, dorsal column lesions, muscle vibration and muscle thixotrophy (U. Proske & Gandevia, 2012). The muscle spindle afferents can be disrupted using tendon vibration (Cordo, Gurfinkel, Bevan, & Kerr, 1995). Although this reduces the accuracy and control during a movement tasks, subjects are still able to complete the tasks (Cordo et al., 1995; Vuillerme, Teasdale, & Nougier, 2001). Although muscle spindles may detect the most important stimuli (U. Proske & Gandevia, 2012), other mechanoreceptors can contribute afferent information during a proprioceptive test. Tension in the muscle will activate GTOs, compression in the joint will activate joint receptors and stretch and compression of the skin about the joint will activate cutaneous receptors (Kandel et al., 2000). The specific contribution of each receptor to a proprioceptive task is currently still under investigation.

Proprioceptive ability is not simply determined by the detection of stimuli, but also by integration of the detected signals in the CNS (Han, Waddington, Adams, Anson, & Liu, 2015).

In both JPS and force reproduction tasks, the same proprioreceptors are active, but the afferent signal composition may differ (Fallon & Macefield, 2007). JPS testing stimulates GTO’s, muscle spindles and receptors in the skin and joint capsule. Force sense matching tasks are mainly conducted using an isometric contraction. While no change of length occurs during an isometric contraction, muscle spindle discharge may increase and become irregular due to gamma motor neuron activation (Kakuda & Nagaoka, 1998; Vallbo, 1974; Wilson, Gandevia, & Burke, 1997). Force sense tasks will also have afferent information incoming from the skin where pressure is applied by the restraints to keep the limb in position. As force increases a greater contribution of the proprioceptive afferent will be from Golgi tendon organs (Fallon & Macefield, 2007).

The procedure for each proprioceptive measurement method is specific to the purpose of the experiment and the involved joint. Although these methods aim to measure force sense and JPS, a relationship between them has yet to be established. In a subject pool comprised of healthy individuals and individuals with functional ankle instability or a history of ankle sprains, no statistically significant correlations between force sense and JPS were found (Docherty, Arnold, Zinder, Granata, & Gansneder, 2004; Kim, Choi, & Kim, 2014). The inclusion of injured subjects and significant torque difference requirements between the force sense task and JPS task may have confounded the results.

Joint position sense protocols require that the limb be moved to a target angle, while some sort of feedback on the angle is provided. The limb is held there while the position is memorized by the subject. The subject then relaxes for a short period and is then required to move the limb to the memorized angle without feedback. The difference between the target angle and reproduced angle is recorded (Dover & Powers, 2003; Kim et al., 2014). The protocol for force sense is similar except the limb angle is kept constant and the target is a level of force. In previously utilized force sense protocols, the target was set as a percentage of maximal voluntary contraction (eg. 10%, 20% and 30%) (Docherty & Arnold, 2008; Dover & Powers, 2003; Kim et al., 2014). Holding the arm at a specific angle requires the torque due to the muscles be equal to the torque due to the weight of the arm. In force sense tasks muscles must first overcome the torque due to the weight of the arm before a positive force will be detected by the load cell. This means that in all cases, even at 10% MVC, the force sense task measures includes the torque due to the weight of the arm and the target force while JPS measures have only the torque due to the weight of the arm at any position. The lack of correlation between JPS and force sense tasks may be due the difference in torque requirements between the two tasks. However, Docherty et al., 2004 found that individuals with functional ankle instability had force sense deficits but not joint position sense deficits. This may also be a reason for a correlation between JPS and force sense not being found in previous studies which have only been performed at the ankle (Docherty, Arnold, Zinder, et al., 2004; Kim et al., 2014).

During proprioceptive tasks requiring a specific joint angle or force level, an efferent command is sent to the muscles of the limb. Closed loop control is used to determine when a limb is at the desired position. This is done through visual feedback or the comparison of incoming afferent sensory (proprioceptive) information to the efferent copy (Kandel et al., 2000). The efferent copy contains expected afferent information resulting from a movement. Information that is incongruent with the efferent copy is passed up to higher processing centers. This is a form of sensory gating that helps prevent wasting or overloading of cognitive resources (Pynn & DeSouza, 2013). At the shoulder, the efferent command will contain directions for both the deltoid and supraspinatus muscles during shoulder elevation. The efferent copy will therefore contain expected afferent information from these muscles as well as from the joint and skin. A nerve block results in changes in incoming afferent information.

The suprascapular nerve block prevents afferent feedback from the supraspinatus, infraspinatus and the majority of the joint capsule. It will also prevent any efferent commands from reaching these muscles. This results in changes in the motor command to the shoulder demonstrated by an increased in deltoid activation (McCully et al., 2007). This procedure creates a disassociation between the two limbs of the upper extremity. The supraspinatus, infraspinatus and glenoid capsule of the contralateral limb's afferent information path remains intact and the muscles are still able to receive motor commands. While proprioception is commonly thought of as only afferent information arising from peripheral receptors, in force sense an alternative hypothesis proposes that conscious sense of effort is coded into the efferent command to the muscle before the command is sent from the motor cortex (U. Proske & Gandevia, 2012). This is primarily investigated in a concurrent contralateral force matching task where one limb has been fatigue (Carson, Riek, & Shahbazpour, 2002) or between the same muscle on opposite limbs but at different muscle lengths (Cafarelli & Bigland-Ritchie, 1979). In fatigue studies, the non-fatigue muscle would overestimate the fatigued muscle while the muscle at optimum length would overestimate the torque of the muscle positioned less optimally. However evidence does exist that individuals may be able to focus their attention on either force sensations in the muscle or the central sense of effort (D. I. McCloskey, Ebeling, & Goodwin, 1974).

The disassociation between the limbs caused by a suprascapular nerve block may help determine if the command to the contralateral limb originates from the sense of effort from the efferent command to the reference limb, the afferent information of the reference limb or a combination of both. This is due to the unique neuroanatomical configuration at the shoulder.

While both the supraspinatus and deltoid perform abduction, they are innervated by different nerves (Moore et al., 2014). Blocking one will leave the other unaffected. The origin of the controlling command can be determined by using the measured force output and deltoid EMG. The measured force accuracy is dependent on afferent feedback and would indicate the role of afferent feedback. Deltoid EMG is representative of the efferent command to the deltoid muscle. Since the deltoid EMG is altered under a nerve block condition, if during a concurrent force matching task the deltoid of both arms exhibit the same relative EMG activation the command can be considered as under the influence of the efferent copy and sense of effort. Since the efferent command still contains a supraspinatus portion, the matching limb will overestimate the force of the reference limb (when matching non-blocked to blocked). If afferent feedback is used the limbs will match relative output force but with different deltoid EMG. However, it may also be that a combination of afferent feedback and central sense of effort may be used for the matching arm.

The proprioceptive portion of this dissertation has two aims. Aim 3 is to determine the relationship between joint position sense and force sense at the shoulder (Chapter IV). Aim 4 is determine how the central nervous system resolves a functional and sensory disassociation between shoulders (Chapter V).

Suprascapular Nerve Block

A nerve block procedure, a type of regional anesthesia, is a medical procedure that prevents signals from travelling along a nerve. Anesthetic is injected around the nerve. Action potentials are propagated down a neuron by the influx of sodium when the nerve is depolarized. Sodium enter the neuron via voltage gated sodium channels. Local anesthetic binds to sites in the inner membrane. This generates a positive electrostatic field that repels sodium ions (Vadhanan, Tripaty, & Adinarayanan, 2015). This prevents transmission of both afferent and efferent nerve impulses. Nerve blocks are commonly performed for pain relief or desensitization of an area undergoing a surgical procedure. The choice of anesthetic is dependent on the desired outcome. Lidocaine is a fast acting anesthetic but the effects can be extended through the use of epinephrine, which constricts vessels at the site of injection, delaying the clearance of the anesthetic (Hadzic, 2007).

The suprascapular nerve is a mixed nerve arising from the superior trunk of the brachial plexus. It supplies innervation to 70% of the GH joint, the AC joint and the supraspinatus and infraspinatus muscles. There is no cutaneous innervation. The nerve travels through the scapular notch, which is also the site of injection. The site is determined by bisecting a line marking the scapular spine. The entry mark is 2-3cm toward the middle and upper quarter (Hadzic, 2007).

Acknowledgement of Co-Authored Material

This dissertation consists of previously published (and unpublished) co-authored material. Chapters II and IV have been published with Dr Andrew Karduna as a co-author for his contribution to editing, experimental design, advising, and project conception. In addition to Dr Andrew Karduna, chapters III and V will also be co-authored by Dr Peter Kosek who advised on and performed the nerve block procedure in both studies.

CHAPTER II

DELTOID EMG RELIABILITY IN ISOMETRIC RAMP CONTRACTIONS

The experiment described in this chapter was developed with Dr Andrew Karduna who contributed substantially to this work by assisting with experimental conception, editing and advising throughout the project. I was the primary contributor to the development of the experiment design, data collection and analysis, and write up.

Introduction

Electromyography (EMG) is a tool to determine the electrical behavior of muscles during a contraction. EMG can be measured simultaneously with an applied force to determine the relationship of EMG and an external load. This relationship may change depending on the rate of force development (Ricard et al., 2005), joint angle (Saito & Akima, 2013), muscle fiber properties (Kupa, Roy, Kandarian, & De Luca, 1995) and the particular muscle under investigation (Lawrence & De Luca, 1983).

Reliability of EMG variables and measured isometric force has been demonstrated in a multitude of muscles, such as the quadriceps (Kollmitzer, Ebenbichler, & Kopf, 1999) and triceps brachii (Yang & Winter, 1983). Establishing an EMG amplitude and isometric force curve requires repeated isometric contractions at each force level. This procedure is both time consuming and can introduce fatigue which is known to elicit changes in the EMG signal (Merletti, Rainoldi, & Farina, 2004). A potential solution to this problem is to utilize isometric ramp contractions (IRCs), which give a continuous EMG amplitude curve with isometric force. The deltoid is a superficial muscle of the shoulder and is often investigated using EMG (de Witte et al., 2014; McCully et al., 2007). It would be important to know if IRCs could be used to reduce overall testing time for this muscle.

The purpose of this study is to determine the reliability of EMG amplitude during IRCs from the middle deltoid. Since the EMG signal may change at different muscle lengths and rate of force increase, possibly affecting reliability, these factors will be included in the analysis. We hypothesize that IRCs will be reliable for sub-maximal isometric contractions. We further hypothesize that EMG amplitude will increase with load and elevation angle.

Materials and Methods

Subjects

Twenty two subjects between 18 and 35 years old (11 male, 11 female, age: 20.2 ± 1.2 years, 20 right handed, 2 left handed) were tested. Subjects self-reported hand dominance by indicating which hand was used to write. Exclusion criteria included: 1) previous shoulder or neck injuries, 2) current shoulder or neck pain, 3) humeral elevation ROM less than 135° and 4) pregnancy. The subjects were briefed on the purpose and the experimental procedure prior to the start of the experiment and completed an informed consent form. The experiment received ethical clearance from the Internal Review Board at the University of Oregon.

Experimental set up

The force acting on a wrist cuff was recorded using a uni-axial load cell (Lebow Products, Troy, MI. Model 3397-50). Force data and surface electromyography (EMG) were sampled at 1,000 Hz and processed with custom LabVIEW software (LabVIEW v13.0, National Instruments, Austin, Tx).

Surface EMG signals from the middle deltoid of the dominant limb were recorded with oval, bipolar Ag/AgCl, conductive solid gel electrode pairs (Bio Protech Inc, Wonju, Korea). The skin surface was cleaned with rubbing alcohol and Nuprep gel (Weaver and Company, Aurora, CO). The electrodes were placed 2cm below the acromion process on the middle deltoid. The electrodes were positioned along the estimated muscle fiber direction with an inter-electrode distance of 2cm. The ground electrode was fixed over the acromion process of the ipsilateral scapular. The deltoid EMG was collected with the Myopac Jr unit (Run Technologies, Mission Viejo, CA). This unit provided signal amplification (gain = 1,000 dB), band pass filtering (10-1,000 Hz) and CMMR of 110 dB.

Protocol

The subjects stood so that their arm was elevated in the scapular plane and that the styloid process of the ulna was placed on the far edge of the load cell surface in the 'thumbs up' position and the elbow fully extended (Figure 2.1). Three angles of humeral elevation selected for testing were: 30°, 60°, and 90°. The angle and height of the load cell was adjusted for each testing angle so that the forearm was flush with the surface of the load cell. A single five second maximal voluntary contraction (MVC) of shoulder abduction for each humeral elevation angle was recorded prior to the first session of testing. Following the MVC's, IRCs were recorded.

The IRC protocol was repeated twice at two rates of loading: absolute and loading rates relative to the subject's MVC. The absolute loading rate was set at 15 N/s and relative loading was set at 14.3% MVC/s. This level was based on pilot subjects MVC and set so that the loading rates were unlikely to exceed 30N/s.

The subjects performed three IRC trials at each angles of shoulder elevation. The order that the angles were tested was block randomized. Each loading rate was tested completely before repeating the same angle testing order for the second rate of loading (18 contractions total). The loading rate conditions order was also block randomized. After the trials at all contractions were completed in the first session, subjects waited 15 minutes and the protocol was repeated.

An LCD monitor presented visual feedback of force output that consisted of 3 lines. Biofeedback of the subject's force output was represented with a dynamic pink line across the width of the graph and required loading rate was presented by 2 limit green lines across the width of the graph. The limit lines were separated by a space representing 10 N. The limit lines would move up the graph at either the absolute rate of 15 N/s or the relative rate at the onset of the trial from a point representing -40 N. Fifty percent force MVC for each humeral elevation angle (30°, 60° and 90°) was represented with a static red line across the width of the graph.

Subjects were instructed to relax the arm at the beginning of each trial and to increase the force applied on the load cell to keep the dynamic force line between the 2 moving limit lines. The limit lines were set to increase at either the relative or absolute rate. If the dynamic force line left the boundaries set by the limit lines, the trial was repeated. Each trial was separated by a 1 minute rest period and a 2 minute rest period was given between angle changes.

Figure 2.1: Experimental setup for reliability. 1) Bracket and load cell set flush with forearm at 60° humeral elevation 2) Middle deltoid electrodes 3) Floor taping. Tape was placed at 60° from the coronal plane. The subject's feet and pelvis were placed parallel to these lines with the shoulder in line with the bracket. This placed the subject's arm in the scapular plane, approximately 30° anterior to the coronal plane.

Normalization

Electromyography amplitude data and force data for each subject at each angle were normalized with respect to their maximum values obtained during a maximal voluntary contraction performed at 30°, 60° and 90° of humeral elevation (Doheny, Lowery, FitzPatrick, & O'Malley, 2008). The 5 second MVC contraction was smoothed using a 300ms RMS window. The first 2.5 seconds and the last 1 second was trimmed. The mean of the remaining 1.5 seconds was used for normalization.

Data analysis

Electromyography amplitude was smoothed using a running 300 ms RMS window. After normalization, the custom Labview program would search the data for the first instance that the subject reached one of the predetermined force level values (10%, 20%, 30% & 40% MVC). The associated smoothed and normalized EMG amplitude value was extracted for each force level from every trial. Three EMG amplitude values were obtained for each force level from the 3 trials and were averaged for each angle and loading rate.

Statistical Analysis

Statistical analysis was performed using SPSS version 22.0 (SPSS Inc., Chicago, IL). Reliability for EMG amplitude at 30°, 60° and 90° elevation and for absolute and relative rates were assessed at 10%, 20%, 30%, and 40% of max force between the first and second testing sessions via a two-way mixed effects ICC(2,3) model. A 2-way repeated measures ANOVA was used to assess the effect of elevation angle and force level on EMG amplitude using the first session's data. Follow up comparisons were performed using a Bonferroni adjustment.

Results

Intraclass correlation coefficient (ICC) values (**Error! Reference source not found.**) were higher than 0.8 for all angles and rates except for 90° elevation during relative rate. Due to the drop in reliability for the relative rate at 90° elevation, we utilized the first session's absolute rate data for further analysis. The main effect for force level on EMG amplitude was significant, $p < 0.01$ (Greenhouse Geisser adjustment) with a follow-up contrast demonstrating a significant ($p < 0.001$) linear increase in EMG amplitude with force level. The main effect for elevation angles on EMG amplitude was significant, $p < 0.01$. Follow up pairwise comparisons were performed with a Bonferonni adjustment ($\alpha = 0.017$). Deltoid EMG amplitude was found to significantly different between 30° and 90° ($p < 0.017$) elevation and 60° and 90° ($p < 0.017$) but not between 30° and 60° ($p = 0.0171$) (Figure 2.2).

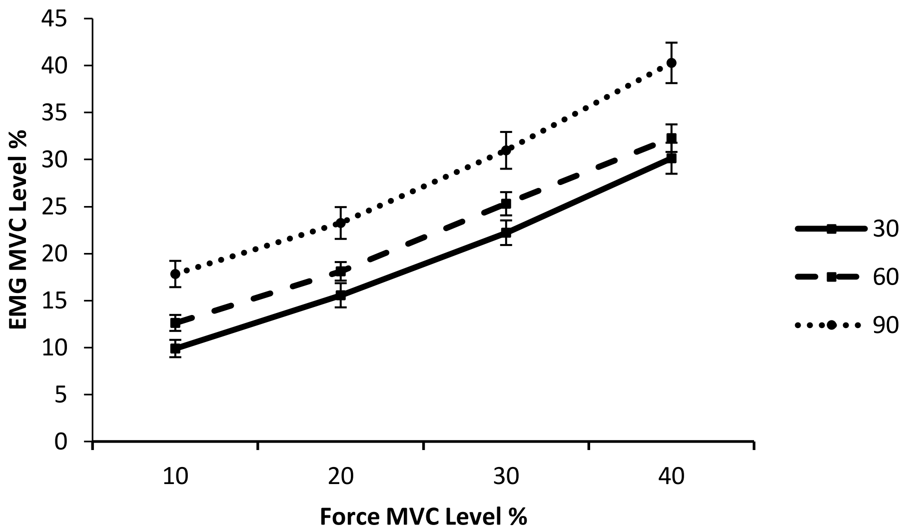

Figure 2.2: Effect of angle and external load on deltoid EMG at a rate of 15N/s. Error bars are standard error of the mean.

Table 2.1: ICC Values for EMG at Different %MVC Force Levels for Each Rate of Force Application

	Angle	10%	20%	30%	40%
Absolute Rate	30°	0.94	0.93	0.93	0.85
	60°	0.92	0.89	0.87	0.89
	90°	0.96	0.93	0.95	0.95
Relative Rate	30°	0.94	0.94	0.84	0.85
	60°	0.93	0.90	0.86	0.83
	90°	0.93	0.89	0.69	0.67

Discussion

The purpose of this study was to assess the reliability of deltoid EMG amplitude during isometric ramp contractions and determine if EMG amplitude and isometric force relationship was affected by shoulder elevation angle and rate. We found that reliability was better during the absolute rate of force application (15 N/s) and did not drop below an ICC value of 0.87. This is comparable to reliability during isometric contractions at the triceps brachii and quadriceps (Kollmitzer et al., 1999; Yang & Winter, 1983). We selected to further examine the absolute rate of force application because it was more reliable (or as reliable) as the relative rate in 9 of the 12 conditions and because of ICC values below 0.70 were found at 90° elevation for the relative rate. Subjects reported that ninety degrees of elevation was the most uncomfortable position from which to apply a force.

Although we found an effect for both shoulder angle and force level, neither was associated with a drop in reliability. The change in EMG amplitude and isometric force curve with increased elevation angles reflects an effect of the change in muscle length or the need to overcome a greater amount of baseline torque at higher angles of elevation (De Wilde, Audenaert, Barbaix, Audenaert, & Soudan, 2002). The effect of increasing EMG amplitude with increased force is consistent with previous research (Lawrence & De Luca, 1983; Madeleine, Bajaj, Søgaard, & Arendt-Nielsen, 2001).

Using the IRC method, three data points for each force level can be obtained in about five minutes using a 2 minute rest period and the same elevation angle. Another methodology using isometric contractions combined with a regression analysis is able to strongly predict EMG at loads above 60% in the quadriceps muscles (Stock, Drusch, & Thompson, 2014). This method would take an estimated 11 minutes and records a single isometric contraction data point at intervals of 10% MVC up to 60% MVC. These time estimates to not include the measurements of MVCs.

Researchers should consider EMG variables of interest when selecting a contraction type. A frequency analysis cannot be performed with a Fouier transform algorithm during an IRC due to the non-stationarity of the signal. However this can be solved by using a wavelet and principle component analysis (Qi, Wakeling, Green, Lambrecht, & Ferguson-Pell, 2011). In the case of a frequency domain investigation, an isometric contraction type protocol may be less complex.

Conclusion

A set rate of 15N/s for all subjects was reliable across all angles while the reliability for the rate based on MVC had reduced reliability at 90° elevation but was still reliable overall. We concluded that EMG amplitude during isometric ramp contractions are reliable up to 40% MVC and that either a set rate or a rate relative to a subject's MVC can be used. This provides an additional tool for researchers to obtain a continuous EMG amplitude and isometric force curve without the use of regression, increased time efficiency, multiple data points and reduced fatigue induced signal alterations.

Limitations

This study utilized a single MVC at each humeral elevation angle to normalize the force and EMG data. While subjects were given practice MVC attempts, it is possible that the recorded MVC may underestimate actual MVC. The effect of angle and load should be interpreted with this in mind. The position of the subjects in this study was controlled and the results are not necessarily applicable to movements occurring outside the scapular plane or at higher rates of load application.

Acknowledgements

Thanks to Katya Trousset for her assistance in data collection.

Bridge

The isometric ramp contraction technique was determined to be reliable in this study. This technique reduces the length of time to establish a continuous isometric contraction and EMG amplitude curve compared to step level isometric contractions. This experimental protocol is used to collect data used calculate supraspinatus contribution in the next chapter.

CHAPTER III

SUPRASPINATUS CONTRIBUTION

The experiment described in this chapter was developed with Dr Andrew Karduna who contributed substantially to this work by assisting with experimental conception, editing and advising throughout the project. Dr Peter Kosek further contributed to the study by advising on and performing the nerve block procedure. I was the primary contributor to the development of the experiment design, data collection and analysis, and write up.

Introduction

Shoulder complaints have a prevalence between 7% - 49% (Brox, 2003) and pain often persists beyond a year after the initial insult (Chard et al., 1991). The supraspinatus muscle unit is a common site of injury for these shoulder complaints (Fehringer, Sun, VanOeveren, Keller, & Matsen, 2008; Milgrom, Schaffler, Gilbert, & van Holsbeeck, 1995). In order to prevent and treat supraspinatus injuries effectively, the mechanical behavior of this muscle needs to be understood. Determining the mechanical behavior of the supraspinatus is difficult due to the large number of muscles crossing the glenohumeral joint and changing muscle moment arms throughout the range of motion (Ackland et al., 2008). This makes inverse dynamics calculations statistically indeterminate and the computation of individual muscle forces difficult.

Functionally, the supraspinatus helps elevate the arm and counteracts the superiorly directed forces generated by the deltoid and keep the joint reaction force directed into the glenoid cavity (de Witte et al., 2014; Yanagawa et al., 2008). Paralyzing the supraspinatus by suprascapular nerve block results in an approximate drop in maximal torque capacity of 50% (Howell et al., 1986). A suprascapular neve block also causes superior migration of the humeral head (J. G. S. Juan, Kosek, & Karduna, 2013) and a compensatory increase in deltoid electromyographic (EMG) amplitude (McCully et al., 2007). As there is a direct relationship between EMG amplitude and force generated by a muscle (Kuriki, Mello, & Azevedo, 2012), the magnitude of the force that the deltoid applies on the humerus increases under these conditions. These studies provide evidence that the supraspinatus assists with both humeral elevation and glenohumeral stabilization. A final note on supraspinatus function is that it is reported in anatomy texts as an initiator of shoulder abduction (Marieb & Hoehn, 2016; Moore et al., 2014). However, this in not supported by the evidence, as the supraspinatus is activated before movement begins, but not before than the deltoid (Reed et al., 2013).

The specific torque contribution of the supraspinatus has not been determined, but some models indicate the deltoid provides the majority of the elevation torque (Wuelker, Plitz, Roetman, & Wirth, 1994; Yanagawa et al., 2008). Based on the shoulder mechanics and muscle anatomy it would be expected that the deltoid contributes more than the supraspinatus to shoulder elevation, particularly at higher elevation angles. This is because the cross-sectional area of the deltoid is much greater than the supraspinatus (Aluisio, Osbahr, & Speer, 2003; Bouaicha et al., 2016) and the change in mechanical advantage through the range of motion (Ackland et al., 2008). The deltoid moment arm increases with humeral elevation, while the supraspinatus moment arm decreases.

The maximal torque producing capacity of the arm is only reduced by 50% with either a suprascapular or axillary nerve block (Howell et al., 1986). Since these nerves innervate the supraspinatus and deltoid respectively, their relative contribution to shoulder elevation torque is approximately 50% during a maximal isokinetic contraction. A cadaveric study supports this finding, with a doubling of deltoid force needed to initiate shoulder abduction (Thompson et al., 1996). In contrast, a computation model indicates that the deltoid is the primary mover while the supraspinatus seemed to apply most of its force to stabilize the humeral head into the glenoid cavity (Yanagawa et al., 2008). This model also has supporting cadaveric evidence where researchers concluded that the supraspinatus function more to compressed the humeral head into the glenoid than generate and elevation torque (Wuelker et al., 1994).

EMG studies have demonstrated differing behavior of the supraspinatus. One observation indicates that when external load is increased, deltoid activation increases, while supraspinatus activation remains constant (de Witte et al., 2014). However, during scapular plane abduction at different maximal voluntary contraction levels, loads and speeds, all shoulder muscles increased their activity, including the supraspinatus (Alpert, Pink, Jobe, McMahon, & Mathiyakom, 2000; Reed, Cathers, Halaki, & Ginn, 2016). Increased activation of a muscle indicates that it is contracting with more force, but the amount of force cannot be determined from EMG amplitude alone. Further examinations of the supraspinatus are necessary, as our current understanding of the supraspinatus function is not consistent in the literature. Quantifying the non-pathological contribution of the supraspinatus may be useful in early identification of shoulder dysfunction and improve treatment plans.

An approach to calculate supraspinatus contribution to glenohumeral abduction is to utilize the compensatory deltoid EMG, seen after a suprascapular nerve block, and externally measured force to indirectly calculate the torque contribution of the supraspinatus muscle at submaximal levels. This method would require that the middle deltoid EMG and external MVC curve to be determined. The purpose of the study is to determine the contribution of the supraspinatus muscle to humeral elevation during isometric ramp contractions at three angles of humeral elevation. We hypothesize that the supraspinatus contribution will decrease with increasing humeral elevation and %MVC.

Methods

Subjects

Nine subjects initially enrolled in the experiment but data was utilized from seven subjects between 18 and 35 years of age (4 males, 3 females, age: 24.9 ± 3.6 years, mass: 76.5 ± 11.0 kg, height: 178 ± 12.2 cm, all right hand dominant). Exclusion criteria included: 1) previous shoulder or neck injuries, 2) current shoulder or neck pain, 3) humeral elevation ROM less than 135°, 4) previous syncope due to needle insertion, 5) known allergic reaction to anesthetic, 6) BMI greater than 30 or 7) pregnancy. Subjects were briefed on the purpose and the experimental procedure prior to the start of the experiment and provided informed consent. The experiment received ethical clearance from the Internal Review Board at the University of Oregon.

Experimental set up

The force acting on the forearms immediately proximal to the radius styloid process was recorded using a uni-axial load cells (Lebow Products, Troy, MI. Model 3397-50). Force data were sampled at 2000 Hz with custom LabVIEW software (LabVIEW v12.0, National Instruments, Austin, Tx). The load cell was offset at each angle to read 0 N. The forearm was flush with the surface of the load cell and secured with custom non-elastic lifting Velcro™ straps.

Surface EMG signals from the anterior deltoid, middle deltoid and posterior deltoid of both arms were recorded with oval, bipolar Ag/AgCl, conductive solid gel electrode pairs (Bio Protech Inc, Wonju, Korea). Only the middle deltoid EMG was used in this study. The skin surface was cleaned with rubbing alcohol. On the anterior deltoid, the electrodes were placed 4 cm below the clavicle on the anterior aspect of the arm, on the middle deltoid electrodes were placed 2 cm below the acromion process and on the posterior deltoid electrodes were placed 2 cm below the lateral border of the scapula spine and angled obliquely. The electrodes were position along the muscle fiber direction with and inter-electrode distance of 2 cm. The ground electrode was fixed over the right lateral malleolus. The deltoid EMG was collected with a Myopac Jr unit (Run Technologies, Mission Viejo, CA) and sampled at 2000 Hz. This unit provided signal amplification (gain = 1000), band pass filtering (10-1000Hz) and CMMR of 110 dB.

Maximal Voluntary Contractions

Prior to the ramp contraction protocol, a series of 5 second maximal voluntary contractions (MVCs) were taken. Subjects were verbally instructed on how to perform a MVC and a practice attempt was given prior to recording. To generate maximum deltoid muscle activity, a unique position was used to target each part of the deltoid (C. E. Boettcher, Ginn, & Cathers, 2008). For the anterior deltoid, the subject performed resisted flexion with shoulder flexed to 90° in the sagittal plan and the elbow flexed to 90° with the forearm vertical. For the middle deltoid, the subject performed resisted abduction with the arm was abducted 90° and elbow flexed to 90° with the forearm vertical. For the posterior deltoid, the subjected performed resisted horizontal extension with the arm abducted to 90° and the elbow flexed to 90° and parallel with the floor.

Additional MVCs were taken for external rotation and abduction in the scapula plane (30° anterior to the coronal plane) for 3 humeral elevation angles: 30°, 60° and 90°. The subjects stood so that their arm was abducted in the scapula plane and that the styloid process of the ulna was placed on the far edge of the load cell surface in the ‘thumbs up’ position and the elbow fully extended. The angle and height of the load cell were adjusted to the appropriate angle being tested and then zeroed. For external rotation, the shoulder was slightly abducted and the elbow flexed to 90°. A towel was placed under the arm to help prevent the subject from abducting their arm during the MVC. If the arm did abduct, the towel would fall to the ground and the MVC was repeated. The height of the load cell and foot positions of the subjects was marked for each abduction angle and external rotation so the position can be replicated in the ramp contraction trials and when the protocol was repeated after the suprascapular nerve block. Subjects were given two attempts for each MVC position with a two minutes rest between each attempt. If the MVC was performed incorrectly, feedback was given to the subject and a third MVC taken. The first 2.5 seconds and the last one second of force data were removed. The mean of the remaining 1.5 seconds was averaged and used as subject's MVC for force and EMG. For MVCs targeting the different sections of the deltoid, the MVC that resulted in the highest EMG amplitude was used for EMG normalization. The MVC with the largest force for each humeral elevation angle was considered for further analysis and MVC force normalization. The mean of the two MVCs after the nerve block were used to determine the subject's MVC.

Ramp Contraction Protocol

Following the collection of all MVCs, a ramp contraction protocol was performed. The ramp contractions were performed at a loading rate of 15 N/s (see Chapter II) up to 60% of the subject's MVC. The subjects performed three trials at each of the following angles of shoulder elevation in the scapula plane: 30°, 60° and 90°. The subjects stood with their feet in the foot positions marked for each during the MVC collection trials and the load cell at the marked heights. The order that the angles were tested was block randomized. A total of nine ramp contraction, three at each angle was collected.

An LCD monitor presented visual feedback of force output that consisted of three lines. All lines were represented on a graph covering approximately half the monitor. Subject force output was represented with a dynamic red line across the width of the graph and the required loading rate was presented by two limit green lines across the width of the graph. The limit lines were separated by a space representing 10 N. The limit lines would move up the graph at either the absolute rate of 15 N/s or the relative rate at the onset of the trial from a point representing -40 N. Sixty percent force MVC for the angle under testing was represented with a static white line across the width of the graph. The display of the feedback graph was reset after each trial.

Subjects were instructed to completely relax the arm prior to the start of the trial, but to gradually increase the force applied on the load cell to keep the dynamic force line between the two moving limit lines. Subjects were given practice attempts until they could successfully keep the dynamic force line between the limit lines for the duration of the trial. If the dynamic force line dramatically left the boundaries set by the limit lines, the trial was repeated. Each trial was separated by a one minute rest period and a two minutes rest period was given between elevation angle changes. Subjects were encouraged to keep their elbow straight throughout the trial and abduct so their arm would elevate in the scapula plane if it were able to move.

Following the collection of all ramp contractions the suprascapular nerve block was performed. External rotation MVCs were taken at five minutes, ten minutes and then every two minutes. Once two external rotation MVCs were recorded below 50% preblock external rotation MVC, the protocol would be repeated.

Data Analysis

Electromyography amplitude was smoothed using a running 300 ms RMS window. EMG data were normalized to the highest EMG recorded for position targeting the anterior, middle and posterior deltoid. Force data were normalized to the peak force at each angle. To quantify the effect of the suprascapular nerve block on the middle deltoid EMG amplitude and %MVC, the program searched for EMG amplitude between 10% and 55%, in 5% increments, and extracted the associated %MVC force level. The mean of the three trials for each humeral elevation angle, before and after the suprascapular nerve block, was calculated.

The program searched the ramp force data before the nerve block for the first instance that the subject reached one of the predetermined force level values between10% and 55% MVC, in 5% increments. The middle deltoid EMG amplitude for each level was extracted and the mean of three trials was calculated. The EMG value for each force level was then used to search the middle deltoid EMG amplitude data in the ramp trials after the nerve block, and extracted the corresponding force level. The difference between the force level at the same middle EMG amplitude represented the contribution of the supraspinatus muscle (Figure 3.1). This value was then divided by the sum of pre-block force level (10% to 55%) and baseline force on the load cell due to the weight of the arm to determine the percent contribution of the supraspinatus at each force level. Baseline force was calculate with anthropometric measurements (D. Winter, 2005) and normalized to MVC. This was repeated for each humeral elevation angle: 30°, 60° and 90°. A specified restriction was that the subject needed to overcome the weight of the arm and apply a positive force on the load cell. This was because subjects that completely relaxed prior to the start of each trial bent their elbow and other subjects did not completely relax so that not all the weight of the arm was measured by the load cell.

$$\%\, supraspinatus = \frac{A - B}{A + |C|}$$

Where *A* is pre-block force level (10% to 55%), *B* is the post block force for the same deltoid EMG amplitude and *C* is the baseline force of the arm on the load cell. All forces normalized to MVC.

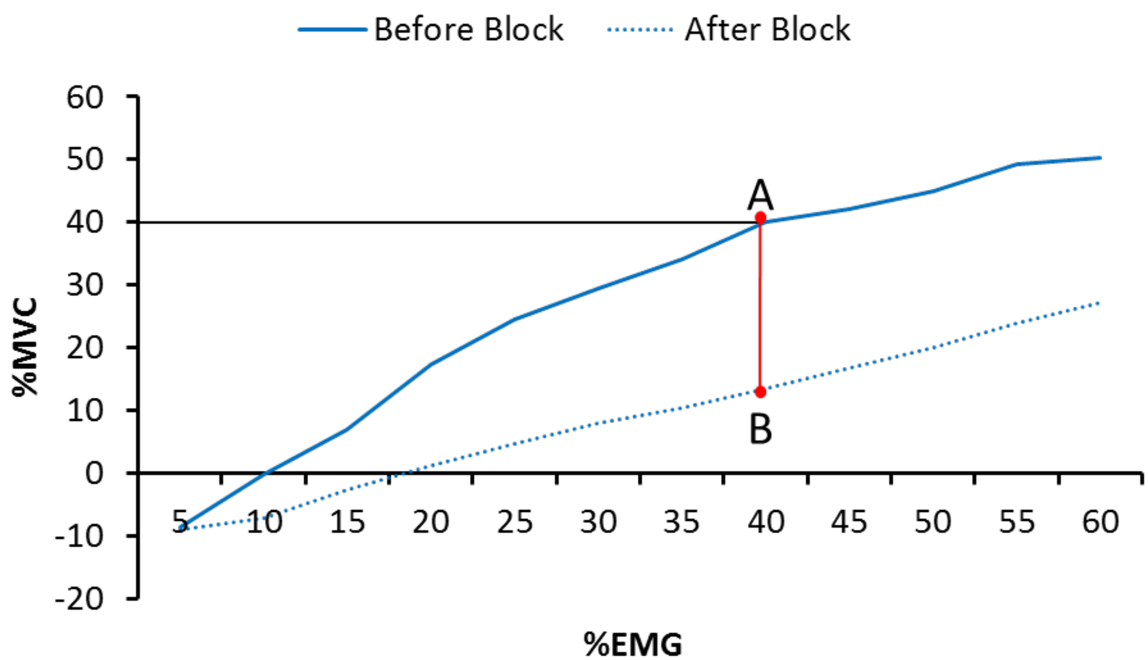

Figure 3.1: A schematic plot to demonstrating the calculation of supraspinatus contribution. The difference between *A* and *B* for the same EMG amplitude, measured at 40% MVC before the block, represents the contribution of the supraspinatus muscle to elevation torque. The difference was then divided by the sum of *A* and normalized baseline resting force at the load cell (calculated from anthropometry). Analysis is performed only when both lines are greater than 0% MVC.

Statistical Analysis

Statistical analysis was performed using SPSS version 22.0 (SPSS Inc., Chicago, IL). To quantify the effects of the suprascapular nerve block on MVC forces, a paired *t* test was conducted on each MVC position: external rotation and maximal abduction force at each elevation angle (30°, 60° and 90°).

A 2-way repeated ANOVA was conducted to determine the effect of %MVC force level (10%:55% in 5% increments) and humeral elevation angle (30°, 60° and 90°) on the percent contribution of the supraspinatus during isometric ramp contractions.

Results

Two subjects were removed from the analysis, thus reducing the total subjects analyzed to seven as reported in subject demographics. One subject was removed because he was unable to activate his deltoid above 20% MVC after the nerve block. The second subject was removed based on a clinical observation of an unusual scapula position by the doctor and not completing the MVC procedure correctly after the nerve block.

Maximal Voluntary Contractions

Subjects demonstrated a significant reduction in the blocked condition for all MVCs. For external rotation, 30° elevation, 60° elevation and 90° elevation there was a 68%, 64%, 71% and 66% reduction respectively ($p \leq 0.001$ for all tests) (Figure 3.2).

Supraspinatus Contribution to Abduction in the Scapular Plane

A positive force value was present for all subjects in the non-blocked and blocked conditions from 30% to 50% MVC. The interaction between humeral elevation angle and load was not significant, $p = 0.23$. The main effect of angle and load was not significant, $p = 0.67$ and $p = 0.13$, respectively (Figure 3.3).

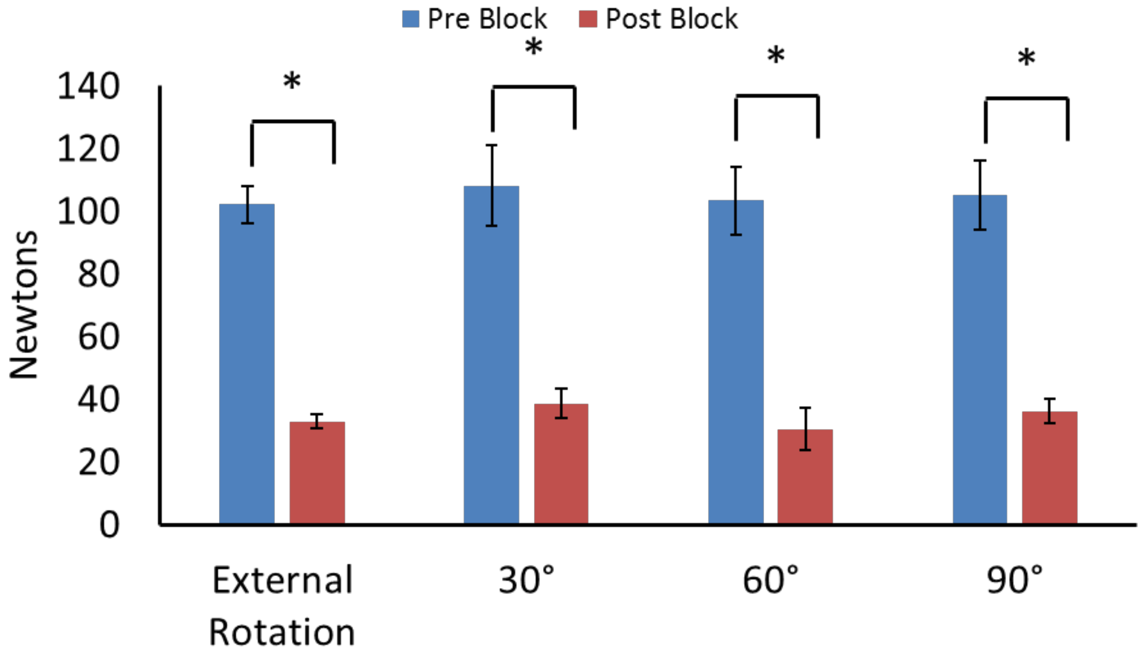

Figure 3.2: Change in maximal voluntary measured force after the suprascapular nerve block

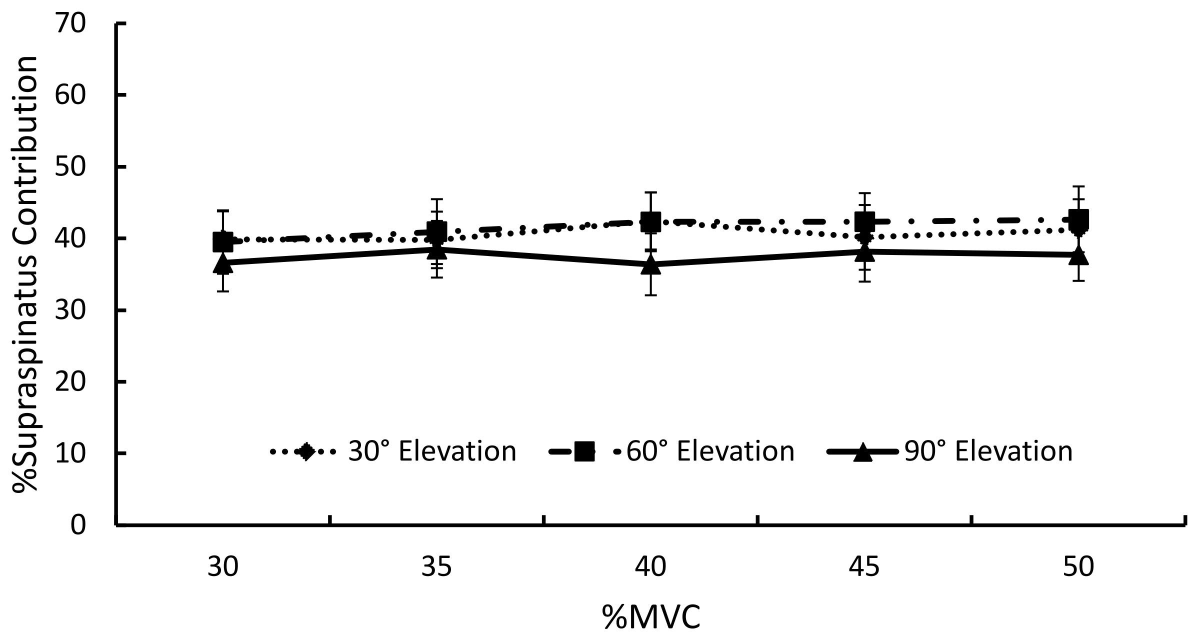

Figure 3.3: Percent contribution of the supraspinatus to the %MVC at each. Error bars are standard error of the mean.

Discussion

The purpose of the study was to determine the contribution of the supraspinatus muscle to external force production. We hypothesized that the supraspinatus contribution will decrease with increasing humeral elevation and %MVC. The suprascapular nerve block resulting in a significant reduction in external rotation MVC and at each angle of abduction in the scapular plane. A similar drop in MVC was seen at each humeral elevation angle. This resulted in a significant decrease in MVC for a given a specific level of deltoid EMG amplitude (%MVC) after the suprascapular nerve block. The criteria for an effective block was a minimum drop of 50% in external rotation. This indicates that all subjects experience an effective suprascapular nerve block. The drop in maximum shoulder elevation is greater than was reported in a previous nerve block study (Howell et al., 1986) but the external rotation reduction was slightly less (McCully et al., 2007).

The data did not support for our hypothesis – the contribution of the supraspinatus did not change significantly for either angle or %MVC. The results indicate that the supraspinatus contributes 36% - 43% of the torque needed to elevate the shoulder. This amount is still more than is predicted by a computational and cadaveric model (Wuelker et al., 1994; Yanagawa et al., 2008). These models predicted that the deltoid is the prime mover with the supraspinatus contributing mainly to compressive forces at the glenohumeral joint. These results also do not perfectly agree with Howell et al., (1986) where the supraspinatus contributes 50% at maximum contraction. However, Howell et al., (1986) used a maximal isokinetic contraction while we used sub-maximal isometric ramp contractions. This could account for the potential differences. Our result is also consistent with the cross-sectional area of the two muscles. The deltoid is substantially larger than the supraspinatus and capable of producing far more force (Aluisio et al., 2003; Bouaicha et al., 2016).

We did not find any effect of angle in this study. The deltoid has a larger moment arm at higher elevation than the supraspinatus. The deltoid's moment arm will decrease while the supraspinatus' will increase with decreasing elevation (Ackland et al., 2008). While some evidence of an angle effect is seen by visually inspecting the graph, this study lacks the statistical power to detect the different. A post-hoc power analysis indicated a small angle effect size of 0.26. A sample of 25 would be needed to achieve a power of 0.8.

It is not clear how much the supraspinatus should contribute to shoulder elevation. Quantifying non-pathological mechanical contribution of the supraspinatus may be useful in diagnosing shoulder dysfunction and improve treatment plans. Current supraspinatus rehabilitation focuses on reducing the activation of surrounding musculature to isolate the supraspinatus and reducing superiorly directed forces at the glenohumeral joint (C. Boettcher, Ginn, & Cathers, 2009; Reinold et al., 2007; Yasojima et al., 2008). Basing rehabilitative exercises solely on activation patterns may not optimize supraspinatus behavior, prevent injury or promote recovery.

Limitations

Subjects in this study were young and asymptomatic with no history of upper extremity injury or pathology. The assumption is that the supraspinatus is functioning normally. The results can only be applied to this population.

In calculating the supraspinatus contribution, only middle deltoid EMG amplitude was used. EMG data from the anterior and middle deltoid, collected in this study, can be incorporated into computational modelling to determine the contribution of the supraspinatus. Modelling results can then be compared to the results of this study. The method of calculation was not able to determine supraspinatus contribution for 0%-29% MVC. A positive force on the load cell was needed to complete the calculation. This was because subjects that completely relaxed prior to the start of each trial bent their elbow and other subjects did not completely relax so that not all the weight of the arm was measured by the load cell. It is for the same reason that baseline force was calculate from anthropometry. Lastly, isometric ramp contractions were used to establish the MVC and EMG curve. Ramp isometric contractions may not represent the concentric and eccentric functioning at the shoulder.

Conclusions

The supraspinatus contributes 36% - 43% of the elevation torque at 30% to 50% MVC. Previous nerve block research found that the supraspinatus and deltoid contribute equally to elevation at maximal contractions. Maximal contractions may therefore not characterize the behavior of the supraspinatus at submaximal levels. However, these results are also different from computational and cadaveric modelling studies indicating the supraspinatus is not a significant contributor to shoulder elevation. Additional research is needed to clarify the role of the supraspinatus muscle.

Bridge

The supraspinatus muscle plays an important mechanical role at the shoulder – elevating the arm and stabilizing the humeral head in the glenoid fossa. It is also one of the most often injured locations at the shoulder. These injuries may compromise the ability to effectively know the position and status (proprioception) of the shoulder. The next chapter investigates the relationship between two sub-modalities of proprioception at the shoulder – joint position sense and force sense – and the implications for future assessment.

CHAPTER IV

RELATIONSHIP BETWEEN JOINT POSITION SENSE AND FORCE SENSE

The experiment described in this chapter was developed with Dr Andrew Karduna who contributed substantially to this work by assisting with experimental conception, editing and advising throughout the project. I was the primary contributor to the development of the experiment design, data collection and analysis, and write up.

Introduction

Proprioception is the ability to determine the movement status and position of a limb in space without the use of vision. Poor proprioception at a joint may result in the increased likelihood of injury (Blasier et al., 1994) and proprioceptive deficits in an injured population have been documented by comparing a dysfunctional or injured joint to either the unaffected side or to a healthy population (Anderson & Wee, 2011; Kim et al., 2014; Maenhout, Palmans, De Muynck, De Wilde, & Cools, 2012; Relph, Herrington, & Tyson, 2014). However, different tests are administered to test proprioception depending on whether the researcher is assessing joint position sense (JPS), force sense (FS) or kinesthesia. While JPS can be evaluated using passive or active protocols, FS is always assessed with an active muscle contraction (Han et al., 2015; U. Proske & Gandevia, 2012). Force sense can be evaluated with ipsilateral and contralateral remembered protocols or concurrent contralateral protocols.

There is likely a relationship between these sub-modalities, since similar pathways and sensory receptors are active during when each is assessed. This is particularly true during active protocols, where muscle tension must be developed in both protocols. The processes active in FS may therefore also be playing some role in JPS (J. A. Winter, Allen, & Proske, 2005). The processes would include gathering and integrating sensory information from receptors located in the periphery (muscle spindles, Golgi Tendon organs, cutaneous receptors and joint receptors) and the centrally generated sense of effort (U. Proske & Gandevia, 2012).

With one exception for ankle eversion, studies investigating the relationship between JPS and FS have not found a significant or high correlation between the two sub-modalities. (Docherty, Arnold, Zinder, et al., 2004; Kim et al., 2014; Li, Ji, Li, & Liu, 2016). Deficits in proprioception may place an individual at greater risk of injury (Blasier et al., 1994). If different aspects of proprioception are unrelated, then a single test may not identify all proprioceptive deficits. Undetected deficits cannot be treated, leaving the individual at risk of injury. Complicating the assessment of proprioception is that the sub-modalities may be affected differently if there is an injury. For example, at the ankle, FS deficits are evident but JPS is unaffected by functional ankle instability (Docherty, Arnold, Gansneder, Hurwitz, & Gieck, 2004).

It is possible that previous studies have not detected a correlation between JPS and FS due to significant methological differences, in terms of joint angle and load. Previous work from our lab has demonstrated that both load and angle have significant effects on JPS at the shoulder and elbow, with decreased errors at higher angles of shoulder elevation (King, Harding, & Karduna, 2013; Suprak, Osternig, van Donkelaar, & Karduna, 2006). To the authors knowledge, the effect of different joint angles and loads have yet to be investigated in a FS protocol. Of the studies investigating a correlation between JPS and FS, only Li et al., (2016) conducted a FS protocol at the same joint targets used in the JPS protocol. This study also found no relationship between JPS and FS.

Force targets are typically set at percentages of a maximum voluntary contraction, while JPS targets are set at specific angles (Docherty, Arnold, Zinder, et al., 2004; Kim et al., 2014; Li et al., 2016). When targets are set in this way, the amount of torque generated during FS protocols will be higher than in JPS protocols. This is because in FS the muscles must first overcome the weight of the limb before a force can be applied while in JPS only the limb's weight needs to be supported. Additionally, force control in the upper extremity has been shown to be less variable than in the lower extremity (Christou, Zelent, & Carlton, 2003). This lower degree of variability may make the joints of the upper extremity more appropriate to examine the relationship between JPS and FS.

The purpose of this study is to determine if JPS and FS are related at the shoulder when external load and joint angle are the same between the protocols. The second purpose of this study is to determine the effect of angle and load on FS. It is hypothesized that there will be a positive correlation between JPS and FS. It is further hypothesized that the behavior of JPS and FS will be the same and error will decrease at higher angles and loads for both.

Material and Methods

Subjects

Eighteen healthy subjects (9 male, 9 female, age: 25.8 ± 3.6 years, weight: 70.7 ± 11.8 kg, height: 171.1 ± 8.6 cm, 16 right handed and 2 left handed) were tested. One additional subject was tested, however due to equipment error, the subject did not complete the experiment. Subjects self-reported hand dominance by indicating the hand they used to write. Subjects were included in the study if they were between the ages of 18 and 40. Exclusion criteria included: previous shoulder or neck injuries that required medical attention, current shoulder or neck pain, humeral elevation range of motion less than 135° and, pregnancy. The subjects were briefed on the purpose and the experimental procedure prior to the start of the experiment and completed an informed consent form. The experiment received ethical clearance from the Internal Review Board at the University of Oregon and all subjects provided written consent.

Instrumentation

The force acting on the forearm immediately proximal of the ulna styloid process was recorded using a uni-axial load cell (Lebow Products, Troy, MI. Model 3397-50) during FS. Force data were sampled at 100 Hz with custom LabVIEW software (LabVIEW v12.0, National Instruments, Austin, Tx). The forearm was flush with the surface of the load cell, in the 'thumbs up' position with the elbow fully extended and secured with custom non-elastic lifting Velcro™ straps. The load cell was adjusted for each humeral elevation angle being tested (50°, 70° and 90°). A head mounted display (Z800, eMagine, Bellevue, WA) provided visual guidance to targets during the JPS and FS protocols and was modified to block all vision of the shoulder and arm and external light sources.

Thoracic, scapular and humeral kinematics were sampled at 120 Hz with a magnetic tracking device (Polhemus Liberty, Colchester, VT), which included a transmitter, three sensors, and a digitizer. The sensors were mounted on the manubrium of the sternum, the flat area of the acromion, as well as on the distal humerus via a custom-molded Orthoplast™ cuff and Velcro™ strap (Ludewig and Cook 2000; Suprak et al. 2006). The transmitter was positioned posterior and contralateral to the testing arm of the subject. The subject sat on an ergonomically designed kneeling chair (Better Posture Kneeling Chairs, Jobri®, Konawa, OK) for both protocols.

Anatomic landmarks were palpated and digitized, using the standards recommended by the International Society of Biomechanics (ISB) (Wu et al. 2005), with the first option used to set the humeral coordinate system. Based on the ISB standard, for humerothoracic motion, the following Euler sequence was used: plane of elevation, elevation, and axial rotation (Wu et al., 2005).

Protocol

Testing was completed in a single session. After weight, height, arm length (acromion process to radial styloid process) and hand length (midpoint between styloid processes to knuckle II) were measured with a tape measure, subjects sat on an ergonomic kneeling stool with no back support, with minimized tactile cues from the back during testing. The calculation of force targets in the FS and weights in JPS was based on anthropometric measurements (D. Winter, 2005), so that the target torque experienced at the shoulder was equal between the two protocols. Target torque was a function of baseline torque, which was considered to be the torque with no extra weights. The order of protocol (JPS and FS) was randomized.

We used an active angle reproduction JPS protocol previously developed in our lab, with minor modifications (King et al., 2013). This protocol uses horizontal white lines on the head mounted display to guide the subject to a target joint angle. The instructions, timing and visual feedback were therefore consistent between the JPS and FS protocols. In both JPS and FS, the subject would memorize a target humeral elevation angle or force for 3 seconds. The subject would then return the arm to the side for JPS and relax the shoulder muscles for FS. The time to relax was 2 seconds. The final phase would require the subject to reproduce the humeral elevation angle or force without visual feedback. The subject would push a trigger held in their free hand when they believed they had reproduced the target. The first modification for this study was that the white lines on the display during the target memorization period did not disappear until the subject was instructed to relax. The second was that the line guiding the subject to a target would represent the glenohumeral elevation angle during JPS and force applied to the load cell during FS. Prior to the first instance of the JPS or FS protocol, six practice trials were given.

For the JPS protocol, there were three target humeral elevation positions (50°, 70°, and 90°) in the scapular plane (35° ± 4° anterior of the coronal plane). Each target position was repeated four times, resulting in 12 trials. The JPS protocol was repeated for each external load (120%, 140% and 160% of baseline torque) with a break of five minutes between each block of 12 trials. In each repetition of the JPS protocol a weight was placed on the wrist that would increase the torque at each angle to 120%, 140% and 160% of baseline torque (Figure 4.1B). The order of testing for targets and weights was randomized.

For the FS protocol there were three force targets with the arm secured to the load cell with a modified none elastic lifting strap at the humeral elevation angle of interest (50°, 70°, and 90° ± 1° elevation) in the scapular plane (Figure 4.1A). This was done with real-time kinematic feedback from the magnetic tracking system so that testing angles in the FS protocol would be the same as the target angles in the JPS protocol. Each target force was repeated four times, resulting in 12 trials. The FS protocol was repeated for each humeral elevation angle (50°, 70°, and 90°) with a break of five minutes between each block of 12 trials. The force targets were calculated to be the applied force to the load cell would be 120%, 140% and 160% of baseline torque. As with the JPS protocol, the order of testing for targets was randomized.

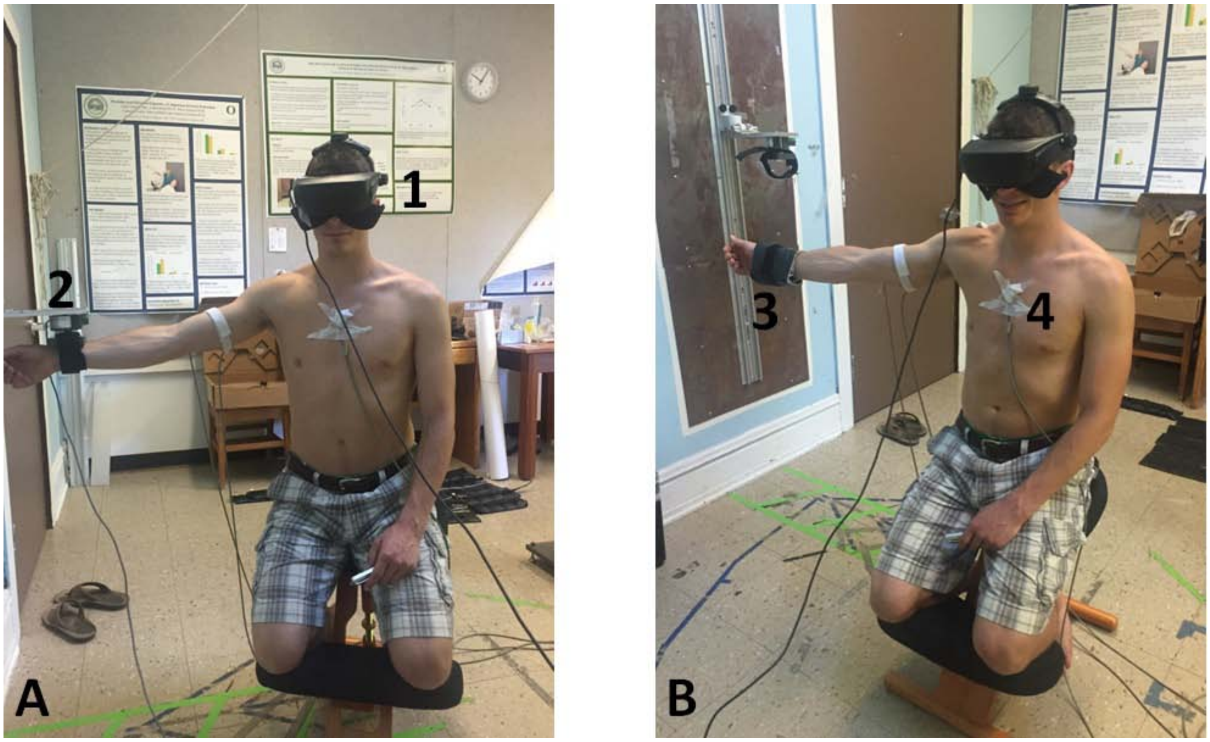

Figure 4.1: Experimental Setup for FS (A) and JPS (B). 1) Head-mounted display 2) Bracket and load cell set flush with the forearm at 90° humeral elevation 3) Wrist weight 4) Thoracic magnetic sensor. Additional sensors were placed on the distal humerus and acromion process. All testing was performed in the scapula plane.

Data Analysis

Angles from the JPS protocol were converted into torque values (Nm) as follows. The wrist weight torque was calculated using the humeral elevation angle, arm length and mass (kg) of the wrist weight. Torque due to the weight of the arm was added to the torque due to the wrist weight to get the total torque value for the presented and reproduced, when the subject pressed the trigger, angles. In the FS protocol the force level during memorization and the reproduced force level, when the subject pressed the trigger, were measured in Newtons. These forces were then converted into torque values (Nm) using the force measured from the load cell acting perpendicularly to the forearm and the length of the arm. The torque measured from the load cell was added to baseline torque for the current humeral elevation angle. The average of the torque during the memorization period was termed presented torque and the instantaneous torque when the subject pressed the trigger button was termed reproduced torque.

RMS error was calculated for each angle and load level (eg. 70° at 120% of baseline torque) in both protocols and normalized to presented torque level (120%, 140% and 160% of baseline torque).

$$\text{RMS error} = \sqrt{\sum \left(\frac{(x_i - T)}{T} * 100\right)^2 / n}$$

Where x_i is reproduced torque, T is presented torque, and n is the number of trials. The average of the four trials at each load and angle combination was calculated. In almost all cases, there were four viable trials. However in some instances subjects did not perform the protocol correctly (eg. relaxing during memorization) or the trigger did not register, resulting in three viable trials in 17 (11%) cases for FS and five (3%) cases for JPS. In these cases the average of three trials was calculated.

Correlation coefficients were calculated for averaged subject score (normalized RMS error score across all conditions) and for each condition. The averaged subject score was calculated by averaging each subject's normalized RMS error score for each condition to indicated ability across all angles and loads for each protocol. All other correlation coefficients were calculated for each condition between the 2 protocols. Eg 50° humeral elevation at 120% during JPS was correlated with 50° humeral elevation at 120% during FS.

Statistical Analysis

Statistical analysis was performed using SPSS version 22.0 (SPSS Inc., Chicago, IL). A Pearson correlation coefficient was calculated to determine the relationship between the subjects averaged score across all JPS conditions and averaged score across all FS conditions in RMS error normalized to target. The correlation between overall JPS and overall FS normalized RMS error was determined by Pearson correlation analysis with a linear regression model with normalized RMS error data averaged for each subject across all loads and angles. Additional Pearson correlations were calculated for each angle and load condition between JPS and FS normalized RMS error.

A 3-way repeated measures ANOVA was used to assess the effect of protocol (JPS and FS), elevation angle (50°, 70° and 90°) and load (120%, 140% and 160% of baseline torque) on normalized RMS error. A significant 3-way interaction will have two follow up 2-way ANOVAs run for JPS and FS to assess the effect of elevation angle (50°, 70° and 90°) and load (120%, 140% and 160% of baseline torque) on normalized RMS error. In the case of a significant 2-way interaction, follow up comparisons of simple effects using a Bonferroni adjust are run. Follow up comparisons are run for significant main effects using a Bonferroni adjustment. An *apriori* alpha level of 0.05 was set for all tests.

Results

Correlations between JPS and FS

No significant correlation for overall normalized RMS error between FS and JPS was found, $r = -0.019$, $p = 0.941$ (Figure 4.2). Neither was there a significant correlation for normalized RMS error between FS and JPS at any angle or load combination (Table 4.1).

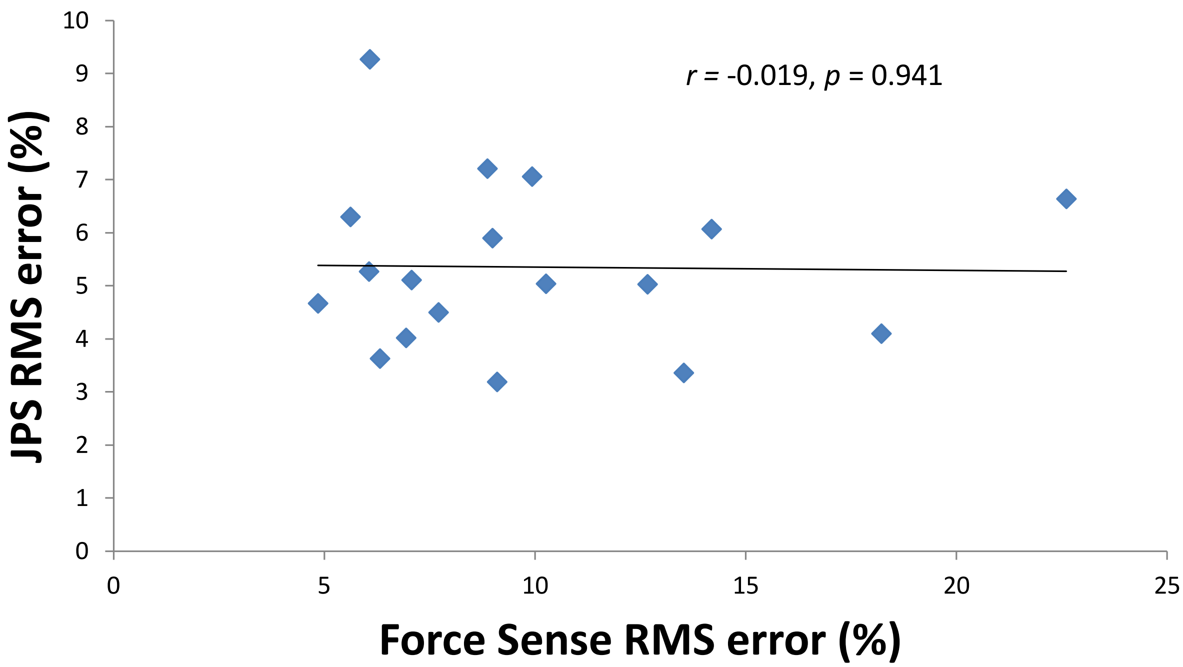

Figure 4.2: Averaged JPS and FS RMS error across all angles and loads

JPS and FS Behavior

A significant 3-way interaction for protocol by angle by load was found on normalized RMS error $F(4, 68) = 22.9$, $p = 0.024$. The follow up 2-way ANOVA for normalized RMS error in the JPS protocol did not demonstrate a significant interaction effect $F(4, 68) = 0.5$, $p = 0.74$ or main effect for load $F(2, 34) = 0.8$, $p = 0.47$ but the main effect for angle was significant $F(2, 34) = 209.0$, $p < 0.001$ after a Greenhouse-Geisser adjustment. A follow up contrast demonstrated a significant ($p < 0.001$) linear decrease in error with increasing humeral elevation (Table 4.2).

Table 4.1. Pearson correlation coefficients between JPS and force sense for each shoulder elevation angle and load.

Angle	Load	Correlation Coefficient
50°	120%	-0.12
50°	140%	0.30
50°	160%	-0.01
70°	120%	-0.03
70°	140%	-0.15
70°	160%	0.11
90°	120%	0.19
90°	140%	-0.42
90°	160%	0.35

Note: No correlations were significant

The follow up 2-way ANOVA for normalized RMS error in the FS protocol found a significant interaction between load and angle $F(4, 68) = 40.0$, $p = 0.014$. Follow up simple effects tests for angle by load showed significant decreases in normalized RMS error at 50° from 120% to 140% ($p = 0.001$), from 120% to 160% ($p < 0.001$) and from 140% to 160% ($p < 0.01$); at 70° from 120% to 160% ($p = 0.001$) and from 140% to 160% ($p = 0.019$). No significant decrease in normalized RMS error was found for any load at 90°. Follow up simple effects tests for load by angle only showed a significant decrease in normalized RMS error at 120% from 50° to 90° ($p = 0.034$) (Table 4.3).

Table 4.2. RMS error normalized to target means and standard error of the mean between angles and loads for joint position sense.

	50° Elevation (SEM)	70° Elevation (SEM)	90° Elevation (SEM)	Total
120% Baseline	12.1 (1.1)	3.4 (0.3)	0.6 (0.1)	5.4 (0.5)
140% Baseline	11.7 (0.7)	3.4 (0.3)	0.5 (0.1)	5.2 (0.4)
160% Baseline	12.4 (0.9)	3.5 (0.3)	0.6 (0.1)	5.5 (0.4)
Total	12.1 (1.1)*	3.4 (0.3)*	0.6 (0.1)*	5.4

Note: * indicates significant difference between angles ($p < 0.05$)

Table 4.3. RMS error normalized to target means and standard error of the mean between angles and loads for force sense.

	50° Elevation (SEM)	70° Elevation (SEM)	90° Elevation (SEM)	Total
120% Baseline	14.9 (2.0)*[a]	12.4 (1.7)[b]	10.6 (1.5)*	12.6 (1.7)
140% Baseline	10.9 (1.4)[a]	9.6 (1.5)[c]	8.7 (1.3)	9.7 (1.4)
160% Baseline	7.2 (1.1)[a]	6.6 (0.9)[b c]	8.7 (1.3)	7.5 (1.1)
Total	11.0 (1.5)	9.5 (1.4)	9.3 (1.4)	9.9

Note: [a, b, c] indicates significant load difference with Bonferonni adjustment ($p < 0.05$), * indicates significant angle difference with Bonferonni adjustment ($p < 0.05$)

Discussion

The primary purpose of the study was to determine if JPS and FS at the shoulder are related when load and angle are controlled between the two protocols. The secondary purpose was to determine the effect of load and angle on FS. We found no significant relationship between FS and JPS overall, or at each angle and load combination (Table 4.1). Our correlation coefficient of -0.019 is consistent with the previous studies at the knee and ankle where non-significant correlations of 0.01-0.35 have been found (Docherty, Arnold, Zinder, et al., 2004; Kim et al., 2014; Li et al., 2016). The exception being the correlation coefficient of 0.65 between JPS and FS during eversion at the ankle in the study by Docherty et al., (2004). Contrary to our hypothesis, accounting for load and angle between JPS and FS indicated that JPS performance predicts < 0.001% of the variance in FS. The individual correlation values for each angle and load varied across the conditions, but all were not significant and within the range previously reported by the other studies.

The low correlation values may be due to the differing behavior between FS and JPS. We hypothesized that both FS and JPS, error would decrease at higher loads and angles of elevation. However, only an effect of angle was found in JPS and an interaction effect was found in FS. The effect of decreasing error in JPS with increased elevation found in this study is consistent with previous research in our laboratory (Suprak et al., 2006). However the effect of load previously observed (Suprak, Osternig, van Donkelaar, & Karduna, 2007) was not seen. Decreased error in that study was seen between 110% of baseline torque and, 130% and 140% at 50° humeral elevation. This study utilized 120%, 140% and 160% and no effect of load was seen for JPS at any humeral elevation angle. It would seem that a slight advantage is gained at very light additional loads (110%) but loads between 120% to 160% of baseline torque do not provide any improvement or decrement.

Conversely, for FS, an interaction effect for load by angle was found. For the load analysis, no follow up tests were significant for 90° humeral elevation. At 70° humeral elevation normalized RMS error decreased from 120% to 160% and from 140% to 160%. At 50° humeral elevation normalized RMS error decreased from 120% to 140%, 120% to 160% and 140% to 160%. In the angle analysis, the only follow up test that was significant showed a decrease in normalized RMS error between 50° to 90° for the 120% load.

A number of factors may be at play causing this interactive result between angle and load. Mechanically, the moment arms of the deltoid and supraspinatus lengthen and shorten respectively with higher elevations (Ackland et al., 2008). The activation characteristics and length-tension relationship of muscles also changes with joint angles (Rassier, MacIntosh, & Herzog, 1999; Saito & Akima, 2013). The greatest torque is encountered when the arm is at 90° humeral elevation. This means that the targets set as a function of baseline torque would be greatest at this angle. These mechanical changes make it difficult to determine if the afferent feedback and sense of effort at the 50° and 70° humeral elevation loads are equivalent to that at 90° humeral elevation. The feedback at lower angles may be more distinctive between loads.

Sense of effort also makes contributions to JPS accuracy. In a contralateral JPS protocol, decreased accuracy is seen when the reference arm is supported and muscle are relaxed. If additional weight is placed on the reference arm, errors are made in the direction of the movement caused by the contracting muscles (J. A. Winter et al., 2005). In our study, force targets in FS and weights for JPS were calculate so that the conditions would have the same torque output. It is unlikely that sense of effort is responsible for the differing behavior and lack of correlation between the two protocols.

The lack of relationship in normalized RMS error between the two protocols may also be due to variations in feedback from the peripheral mechanoreceptors during isometric and concentric contractions. When contracting a muscle both alpha and gamma (fusimotor) motor neurons fire simultaneously, which is referred to as alpha-gamma co-activation (Vallbo, 1970). This prevents a muscle spindle from becoming slack and unable to respond to the muscle lengthening. During isometric contractions, the agonist muscle's muscle spindles firing increases at an inconsistent rate (Vallbo, 1974). In this case the alpha-gamma co-activation may be attempting to shorten intrafusal muscle fibers that are remaining at the same length and applying tension on the muscle spindle increasing its firing rate. This would be the only signal from muscle spindles since during an isometric contraction the antagonist muscle will also remain at a constant length. However it is argued that these fusimotor induced signals from muscle spindles during an isometric contraction are filtered out because they do not induce an illusion of movement at the joint (D. McCloskey, Gandevia, Potter, & Colebatch, 1983).

The incoming afferent information may therefore be very different between the two protocols. The brain may implement different strategies to interpret the afferent information during a FS compared to a JPS task. Given the lack of a correlation in the present study and others, it is possible that JPS and FS are independent of each other. A single proprioceptive test, JPS or FS, would be insufficient to quantify and individual's proprioception because they are independent.

It may be more important to determine which test best correlates with performance based outcomes or injury risk. Results from studies investigating knee JPS with functional based outcomes have shown that JPS can be unaffected but have different functional outcomes between healthy and affected sides (Kafa, Ataoglu, Hazar, Citaker, & Ozer, 2014; Naseri & Pourkazemi, 2012; Yosmaoglu, Guney, & Yuksel, 2013). However, free throw percentage in basketball is higher in athletes with better upper extremity JPS (Kaya, Callaghan, Donmez, & Doral, 2012; Sevrez & Bourdin, 2015). FS and performance based outcomes warrant further investigation.

Limitations

Anthropometric measurements were made according to the protocol by Winter (1995). These were used to calculate the baseline torque, force targets during FS and weights for JPS. This is a limitation in the study since the force vector applied on the load cell may not result in the exact same torque as the gravity directed force vector from the wrist weight. Further, the cutaneous feedback at the wrist between the two protocols may also have differed. Lastly laboratory testing of proprioception may not represent proprioceptive ability during daily movements.

Conclusions

The lack of a relationship between JPS and FS at the shoulder is not due to differing angles and loads between the different modality testing protocols. The relationship at the shoulder is also no different from what has been found in joints of the lower extremity. This may be due to FS and JPS being affected differently by load and angle. Lastly, assessment of only a single sub-modality of proprioception may not be sufficient to quantify an individual's proprioception.

Bridge

The reasons why JPS and FS are not related have not yet been identified. It may be related to the manner in that central mechanism process incoming afferent information. Movement is absent during FS protocols possibly reducing the importance of afferent signals coming from muscle spindles on muscle length. FS may be more reliant on sense of effort to match forces. Chapter V investigates the possible contribution of sense of effort and afferent information in a contralateral force matching task.

CHAPTER V

FORCE REPRODUCTION BEFORE AND AFTER A SUPRASCAPULAR NERVE BLOCK

The experiment described in this chapter was developed with Dr Andrew Karduna who contributed substantially to this work by assisting with experimental conception, editing and advising throughout the project. Dr Peter Kosek further contributed to the study by advising on and performing the nerve block procedure. I was the primary contributor to the development of the experiment design, data collection and analysis, and write up.

Introduction

Proprioception is the awareness of our body and limbs in space without visual perception. The sensory information comes from mechanical receptors located in muscles, tendons, joint capsules, ligaments and the skin (Riemann & Lephart, 2002). In addition, the centrally generated sense of effort is also thought to contribute to proprioception (D. I. McCloskey et al., 1974). Various contralateral force matching experiments have been designed in order to investigate the role of afferent information and sense of effort to proprioception.

In case studies where sensory information has been lost due to deafferentation, subjects are still able to match forces between sides using only the sense of effort (Lafargue, Paillard, Lamarre, & Sirigu, 2003; Luu, Day, Cole, & Fitzpatrick, 2011). In healthy subjects, the sense of effort has been demonstrated to be important where a side is matched to a eccentrically fatigued side or contralateral muscle pairs are match with differing muscle length (Cafarelli & Bigland-Ritchie, 1979; Carson et al., 2002; Uwe Proske et al., 2004). When matching contralateral muscles at different lengths, different force levels would be produced but the level of activation between the muscles was the same. When eccentrically fatigue, subjects would overshoot the reference force when the reference limb was the fatigued side and undershoot when it was the matching side. Matching between sides using the centrally generated sense of effort has been termed the central corollary discharge hypothesis. An important observation made by Carson et al., 2002 was that the undershoot was absent when the force produced was normalized to the muscle's post eccentric fatigue maximal voluntary contraction. They proposed that the damage to muscle fibres from the eccentric contraction altered the gain of the relationship between motor command sent to the muscle and the corresponding sense of effort. More effort was perceived for the same level of muscle activation. Additional evidence supporting that these changes come from a central source is that after eccentric exercise no abnormal function of Golgi tendon organs or muscle spindles was found in cats (Gregory, Brockett, Morgan, Whitehead, & Proske, 2002; Gregory, Morgan, & Proske, 2004). However, Luu et al., 2011 found that after fatigue, *deafferented* subjects overestimated the load, consistent with sense of effort based predictions but healthy subjects did not.

Deafferented subjects have only the sense of effort to match forces but healthy subjects still have afferent proprioceptive information available to them. It is therefore possible that both pathways contribute to force matching ability in healthy subjects. Also in Luu et al. (2011)'s study, healthy subjects performed a matching protocol while the reference side underwent a sustained contraction to fatigue while matching with the opposite limb periodically. Subjects gradually increased the amount of force applied by the matching side but did not overestimate the target to the extent that would be predicted if only sense of effort were used. Since the matching side did gradually overestimate the load some sense of effort role is implicated.

A functional and sensory imbalance between side can also be accomplished at the shoulder using a suprascapular nerve block. Shoulder abduction requires contraction of the supraspinatus and deltoid muscles. These two muscles are innervated by the suprascapular and axillary nerves respectively. A suprascapular nerve block would leave the deltoid muscle unaffected but paralyze the supraspinatus and prevent any afferent information being transmitted from the muscle. Because the block is performed on one side, this will create a functional and sensory disassociation between the shoulders.

Functionally a suprascapular nerve block will reduce the torque producing capacity of the shoulder by an estimated 50% (Howell et al., 1986). In order to produce an equal torque about each shoulder, the deltoid will have to compensate for the loss of the supraspinatus muscle and increase its force on the blocked side. This will however create a sensory imbalance between shoulders since the tension in each deltoid is substantially different. This compensation would also result in increased deltoid EMG amplitude/motorneuron drive (McCully et al., 2007) and associated sense of effort (Morree, Klein, & Marcora, 2012) compared to the same load on the opposite side. This compensation can help determine what information is used by the central nervous system to complete a contralateral force matching task. If sense of effort or deltoid tension (from Golgi tendon organs) are the dominant sources of afferent information to match the load, then the subject would dramatically underestimate the load and deltoid EMG would be the same before and after the block.

The purpose of the study is to determine how the central nervous system accounts for a functional and sensory disruption between sides in a contralateral force matching task. When the left/unblocked side is the reference side and the right/blocked side is the matching side: we hypothesize that subjects will undershoot the pre-block torque error and no change in deltoid EMG error. When the right/blocked side is the reference side and the left/unblocked side is the matching side: we hypothesize that subjects will overshoot the pre-block error and no change in deltoid EMG error. We further hypothesize that the any change in error will be dramatically greater for torque error than EMG error with either side as the reference. Lastly, we hypothesize that the force matching accuracy will be worse after the nerve block with either side as the reference.

Methods

Subjects

Eight subjects initially enrolled in the experiment but data was utilized from seven subjects between 18 and 35 years old (3 males, 4 females, age: 22.4 ± 3.6 years, weight: 67.0 ± 10.0 kg, height: 171.9 ± 7.8 cm, all right hand dominant) were included in the study. Exclusion criteria included: 1) previous shoulder or neck injuries, 2) current shoulder or neck pain, 3) humeral elevation ROM less than 135°, 4) previous syncope due to needle insertion, 5) known allergic reaction to anaesthetic, 6) BMI greater than 30 or 7) pregnancy. Subjects were briefed on the purpose and the experimental procedure prior to the start of the experiment and provided informed consent. The experiment received ethical clearance from the Internal Review Board at the University of Oregon.

Experimental set up

The force acting on the forearms immediately proximal of the ulna styloid processes was recorded using a uni-axial load cells (Lebow Products, Troy, MI. Model 3397-50). Force data were sampled at 2000 Hz with custom LabVIEW software (LabVIEW v12.0, National Instruments, Austin, Tx). The forearm was flush with the surface of the load cell and loosely secured with custom non-elastic lifting Velcro™ straps.

Surface EMG signals from the anterior deltoid, middle deltoid and posterior deltoid of the both shoulder were recorded with oval, bipolar Ag/AgCl, conductive solid gel electrode pairs

(Bio Protech Inc, Wonju, Korea). The skin surface was cleaned with rubbing alcohol. On the anterior deltoid, the electrodes were placed 4 cm below the clavicle on the anterior aspect of the arm; on the middle deltoid, electrodes were placed 2 cm below the acromion process; and on the posterior deltoid, electrodes were placed 2 cm below the lateral border of the scapula spine and angled obliquely. The electrodes were position along the muscle fiber direction with an inter-electrode distance of 2 cm. The ground electrode was fixed over the right patella. The deltoid EMG was collected with the Myopac Jr unit (Run Technologies, Mission Viejo, CA) and sampled at 2000 Hz. This unit provided signal amplification (gain = 1000), band pass filtering (10-1000Hz) and CMMR of 110 dB.

Subjects were presented with three force targets with the arms secured to the load cells in the sagittal plane. The load cells were angled 20° above parallel to the ground. This achieved an estimated 70° humeral elevation when the wrist was flush with the load cell. Each target force was repeated four times, resulting in 12 trials. The force targets were presented in a randomized order. The force targets were calculated to be the applied force to the load cell would be 120%, 140% and 160% of baseline torque using anthropometric equations from Winter (2005). Baseline torque is the torque at the shoulder due to the weight of the arm at 70° humeral elevation.

Vision of the environment during the protocol was occluded with a head mounted display (Z800, eMagine, Bellevue, WA) with modifications to prevent influence of external light sources. The display provided visual guidance to targets during the force matching protocol while blocking all vision of the shoulder and arms.

Maximal Voluntary Contractions

Prior to the contralateral force matching protocol, a series of 5 second maximal voluntary contractions (MVCs) were taken. Subjects were verbally instructed on how to perform an MVCs and a practice attempt was given prior to recording. MVCs were recorded for external rotation on the right/blocked side only. The shoulder was slightly adducted and elbow flexed to 90°. A towel was placed under the arm to help prevent the subject from abducting their arm during the MVC. If the arm did abduct, the towel would fall to the ground and the MVC was repeated. MVCs were then recorded for both arms at 70° humeral elevation in the sagittal plane (the testing position for the force matching protocol). Subjects were given 2 attempts for each MVC position with a 2 minute rest between each attempt. If the MVC was performed incorrectly or there was a non-maximal effort, feedback was given to the subject and a third MVC taken. The first 2.5 seconds and the last 1 second of force data were trimmed. The mean of the remaining 1.5 seconds was averaged and used to represent the subject's MVC. The MVC with the largest force was considered for further analysis.

Contralateral Force Matching Protocol

Testing was completed in a single session. Weight, height, arm length (acromion process to radial styloid process) and hand length (midpoint between styloid processes to knuckle II) were measured with a tape measure. Following these anthropometric measures, MVCs before the suprascapular nerve block were recorded.

The subjects were positioned so that they stood with their forearms flush with the surface of the load cells and arms parallel to each other in the sagittal plane. The load cells were positioned shoulder width apart. A custom non-elastic strap was used to loosely secure the wrists to the load cells. The subject was presented with a black screen with 2 white horizontal lines across the middle two thirds of the head mounted display. A dynamic read line represented the force applied to the reference load cell. The target represented randomly changed to one of the three force targets after each trial and the subjects was provided with no knowledge of their results.

Subjects were verbally instructed on how to perform the protocol. Subjects were asked to maintain both their arms in the ‘thumbs up’ position. Subjects were instructed to maintain the red line between the two white lines to become accustomed to the reference force. The acclimation time was 1.5 seconds. After the 1.5 seconds, the program would initiate the ‘find target’ command and the subject would attempt to reproduce the reference force target with the contralateral arm while maintaining the reference force with the other. Visual feedback for the force generated by the reference arm remained on the heads up display. When the subject felt they had replicated the force, they verbally signalled the researcher to press the trigger and record the force level. A ‘relax’ cue was initiated by the computer and a 15 second count down timer appeared on the display, which indicated the time until the next trial began. Prior to the first instance of the force matching protocol six practice trials were given. During the practice trials, the researcher provided verbal feedback and answered any questions. The force output of these trials was visually inspected to ensure that the subject had understood the instructions before recording the experimental trials. No feedback was given to subjects regarding their performance.

The left arm was always used as the reference for the first set of 12 trials. The right arm was used as the reference for the second set of 12 trials. Following the collection of both sets of 12 trials, the nerve block was performed (see ‘Nerve Block Procedure’).

The external rotation MVC was used to determine whether the block had been effective and the post-block force matching protocol could proceed. The criteria to proceed with testing was a 50% reduction in external rotation for two consecutive external rotation MVCs. Five minutes after the block was completed, the subject's external rotation was tested. If the external rotation force was still above 50% MVC, the subject was retested after another 5 minutes. From that point on, the subject's external rotation was tested every 2 minutes until 2 consecutive external rotation maximal contractions measurements were below 50% MVC. Following this, 2 maximal contractions were recorded at 70° humeral elevation in the sagittal plane for the right/blocked side. The force matching protocol was then repeated. Upon completion of the protocol, 2 maximal contractions were recorded at 70° humeral elevation in the sagittal plane for the right/blocked side and another 2 external rotation maximal contractions measurements. These contractions were used to ensure that the block was still effective at the conclusion of testing.

Suprascapular Nerve Block Procedure

A suprascapular nerve block was performed by a board certified anaesthesiologist. The subject was seated for the procedure with the head flex slightly to the contralateral side. Ultrasound imaging was used to visualize the scapula notch where the suprascapular nerve travels. The ultrasound gel served as a conductive medium and surface preparation. A 3 inch block needle was advanced toward the scapula following the angle the spine makes with the body of the scapula. The advancing needle was observed on the ultrasound until it reached the scapula notch. At this point the lidocaine and epinephrine (1.5%, 1:200,000) was injected. The needle was removed and the subject was allowed to remain seated.

EMG Normalization

EMG amplitude was normalized to the highest recorded MVC at 70° humeral elevation for each side. The raw EMG from each part of the deltoid (anterior, middle and posterior) was smoothed using a 300 ms RMS window. The first 2.5 seconds and the last 1 second was trimmed. The mean of the remaining 1.5 seconds was used for normalization. EMG recorded during the trial was smoothed using a 300 ms RMS window.

Data Analysis

Forces from the load cells were converted to torque (Nm). The torque measured from each load cell was added to baseline torque for the arm at 70° humeral elevation. The total torque for the reference side was termed reference torque and the total torque for the matching side was termed matching torque at the time the trigger was pressed. Likewise, EMG for the reference side was termed reference EMG and the matching side was termed matching EMG at the time the trigger was pressed.

Error was calculated as a percent of the reference for each trial.

$$\% \text{ Error} = \frac{X_{match} - X_{ref}}{X_{ref}} \; X \; 100$$

Where *X* is torque, anterior deltoid EMG, middle deltoid EMG or posterior deltoid EMG. When torque is the variable, a positive value indicates that subject overestimated the target and a negative an underestimate of the target. When EMG is the variable, a positive value indicates that the matching muscle was more active than the reference muscle and a negative value indicates the matching muscle was less active than the reference muscle. The average error of the 4 trials at each load were calculated when the left and right were the reference.

To assess the accuracy of subjects the root mean square (RMS) error was calculated and normalized to baseline torque.

$$\text{RMS error} = \sqrt{\sum \left(\frac{(X_{match} - X_{ref})}{b} * 100 \right)^2 / n}$$

Where X is torque, b is baseline torque of the arm at 70° humeral elevation and n is the number of trials. Again, the average error of the 4 trials at each load were calculated when the left and right were the reference.

Statistical Analysis

Statistical analysis was performed using SPSS version 22.0 (SPSS Inc., Chicago, IL). To quantify the effects of the suprascapular nerve block on maximal voluntary contraction forces, a paired *t* test was conducted on maximal external rotation and shoulder flexion, before and after the block. A third paired *t* test was conducted to compare the sagittal flexion MVC between left and right sides.

The following statistical analyses were conducted first with the left/unblocked side as the reference side and then repeated with the right/blocked side as the reference. To test our first hypothesis on the direction of error (undershoot and overshoot) before and after the nerve block, a 2-way repeated measures ANOVA was used to assess the effects of the condition (non-blocked vs blocked) and load (120%, 140%, and 160% of baseline torque) on each dependent variable (torque error, anterior deltoid EMG error, middle deltoid EMG error and posterior deltoid EMG error).

To test our second hypothesis on the magnitude of the change in error due to the nerve block, a 2-way repeated measures ANOVA was used to assess the effect of each error parameter (torque, anterior EMG, middle EMG, and posterior EMG) and load (120%, 140%, and 160% of baseline torque) on the change in error. In the case of a significant main effect for change in error, a simple contrast comparison between torque error change and EMG error change (anterior deltoid, middle deltoid and posterior deltoid) was performed planned.

To test the final hypothesis on force matching accuracy, a 2-way repeated measures ANOVA was used to assess the effect of condition (non-blocked vs blocked) and load (120%, 140%, and 160% of baseline torque) on torque root mean square (RMS) error normalized to baseline torque.

Results

One subject was removed from all analysis due to an error during target load assignment reducing the total subjects to seven as reported in subject demographics.

Maximal Voluntary Contraction

The block resulted in a significant reduction in external rotation (57%) and sagittal plane flexion (52%) torque on the blocked (right) side ($p < 0.001$) (Figure 5.1).

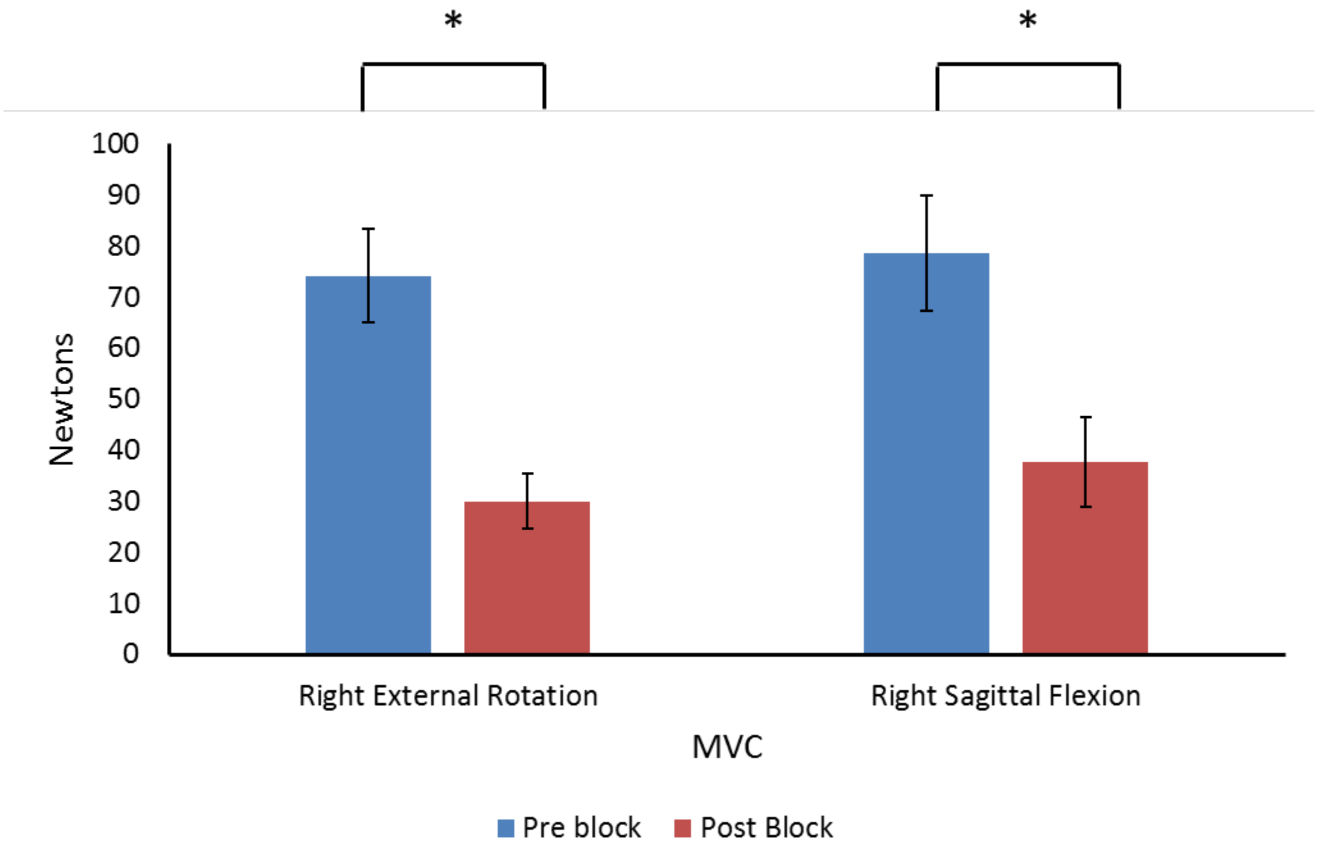

Figure 5.1: Changes in maximal voluntary contraction after the suprascapular nerve block.

Force Matching

In almost all cases, the mean of the four viable trials were calculated. However, in some instances subjects did not perform the protocol correctly (eg. relaxing during acclimation or the trigger did not register) resulting in 3 viable trials in 4/48 (8%) cases and 2 viable trials in 2/48 (4%) cases. In these cases, the average of the viable trials were calculated.

Left/Unblocked Side as Reference

Torque Error

The interaction between the condition (blocked vs non-block) and load was significant ($p = 0.02$). Follow up simple effects for condition at each load only demonstrated significance at 160% where subjects significantly undershot the blocked ($M = -10.3$, $SD = 10.5$) compared to non-block ($M = 1.3$, $SD = 15.3$) condition (Figure 5.2).

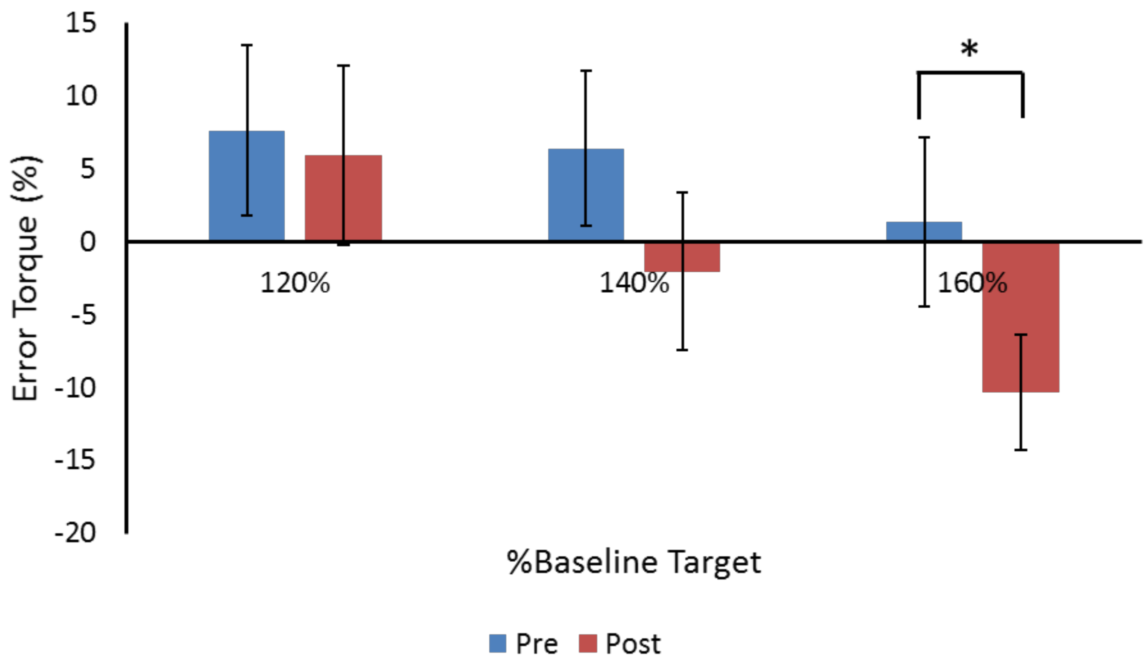

Figure 5.2: Torque error (%) before and after the suprascapular nerve block with the left/unblocked side as reference.

Anterior Deltoid EMG Error

The interaction between the condition and load was not significant ($p = 0.68$). The main effect for condition was significant ($p = 0.039$). The matching anterior deltoid demonstrated higher activation error in the blocked condition ($M = 71.8$, $SD = 24.7$) than the non-blocked condition (*M* = 19.3, *SD* = 17.3) (Figure 5.3). The main effect for load was significant (p = 0.02).

Middle Deltoid EMG Error

The interaction between the condition and load was not significant (p = 0.55). The main effect for load was not significant (p = 0.07). The main effect for condition was significant (p = 0.038). The matching middle deltoid demonstrated higher activation error in the blocked condition (*M* = 133.0, *SD* = 51,7) than the non-blocked condition (*M* = 10.0, *SD* = 15.0) (Figure 5.3).

Posterior Deltoid EMG Error

The interaction between the condition and load was not significant (p = 0.64). The main effect for load was not significant (p = 0.19). The main effect for condition was not significant (p = 0.05).

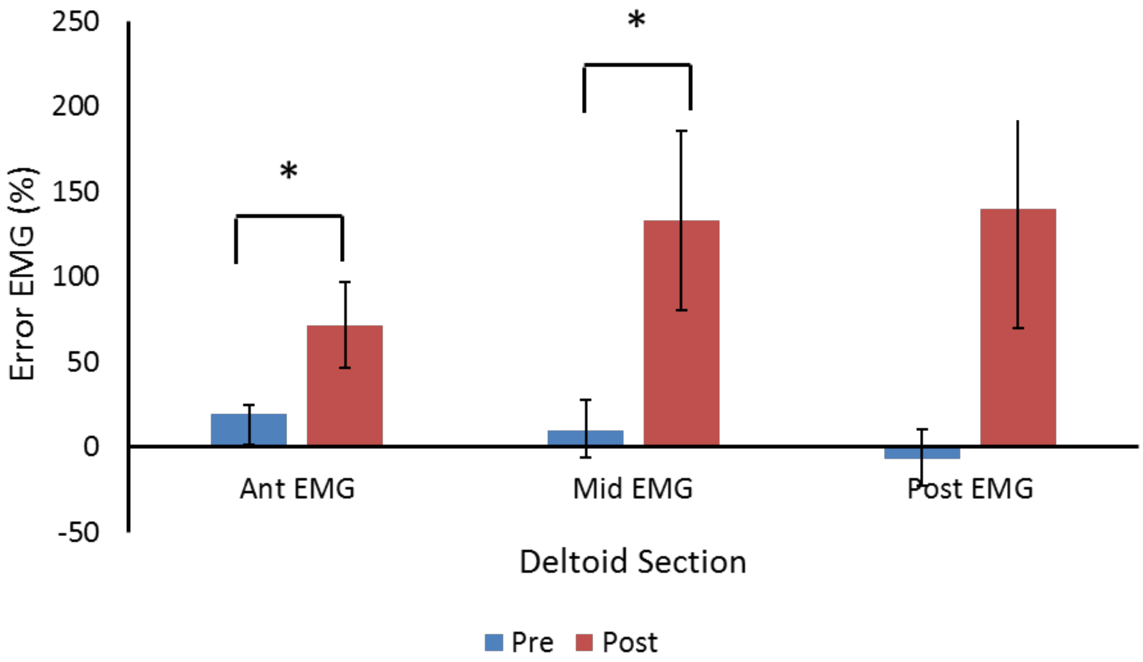

Figure 5.3: EMG error (%) before and after the suprascapular nerve block with the left/unblocked side as reference.

Change in Error

The interaction between the error parameters and load was not significant (p = 0.63) (Greenhouse Geisser). The main effect for load was not significant (p = 0.56). The main effect for error parameter was significant (p = 0.03) (Greenhouse Geisser). The follow up simple contrast demonstrated a significantly larger change in anterior deltoid EMG error (M = 52.5, SD = 19.9, p = 0.01); middle deltoid EMG error (M = 123.0, SD = 46.4, p = 0.025); and posterior deltoid EMG error (M = 147.2, SD = 59.1, p = 0.04) than the change in torque error (M = -7.3, SD = 4.2) (Figure 5.4).

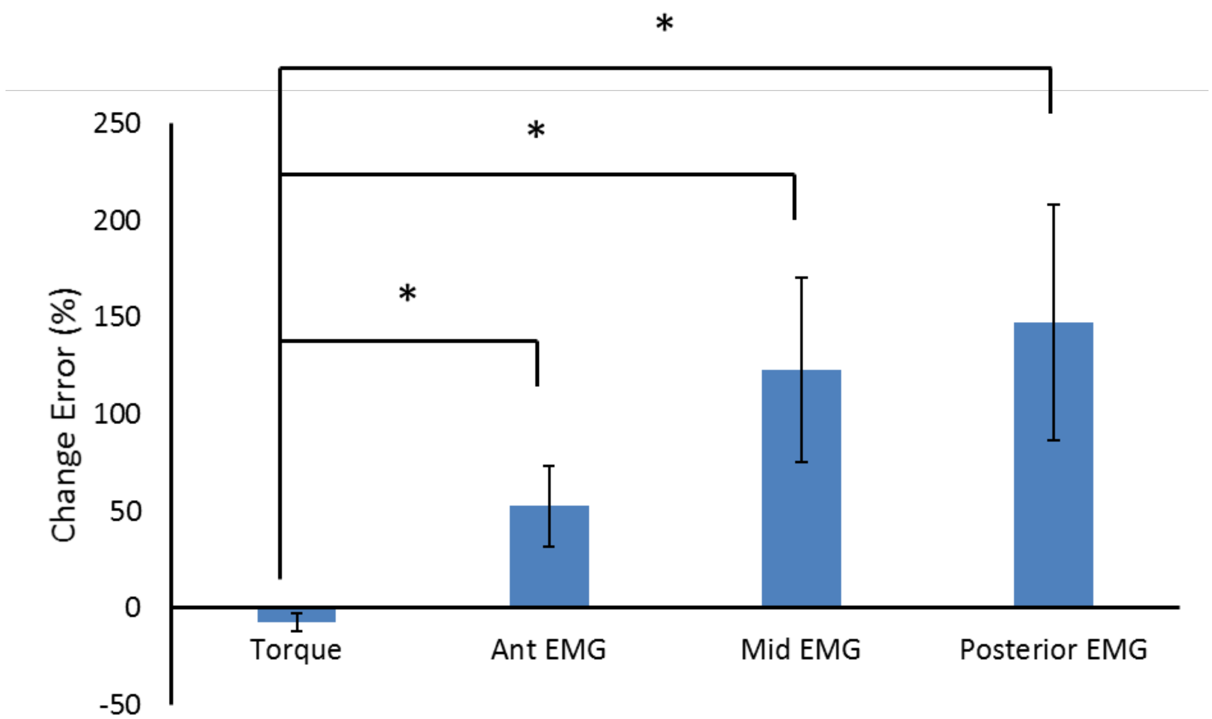

Figure 5.4: Comparison of the change in error (%) from before to after the suprascapular nerve block between torque and EMG amplitude with the left/unblocked side as reference.

Force Matching Accuracy

The interaction between the condition and load was not significant (p = 0.96). The main effect for load was not significant (p = 0.43) and the main effect for condition was not significant (p = 0.78) (Figure 5.5).

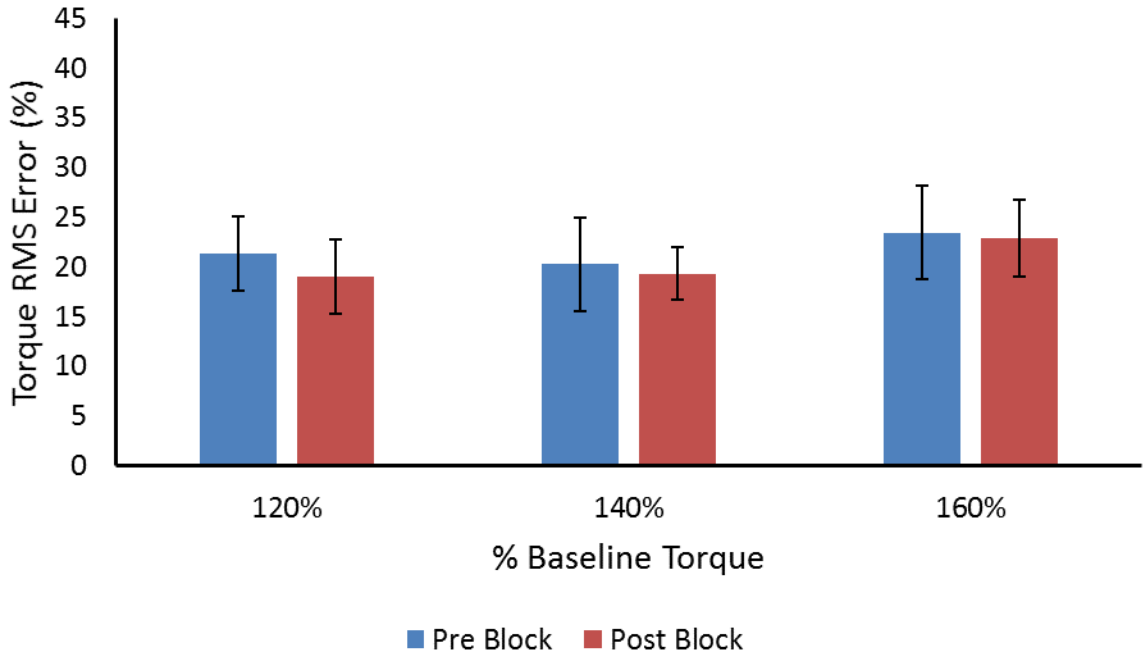

Figure 5.5: RMS torque error (normalized to baseline torque) before and after the suprascapular nerve block with the left/unblocked side as reference.

Right/Blocked Side as Reference

Torque Error

The interaction between the condition and load was not significant (p = 0.064). The main effect for load was not significant (p = 0.38). The main effect for condition was significant (p = 0.02). Subjects had a significantly higher overshoot error in the blocked condition (M = 20.6, SD = 4.9) compared to the non-blocked condition (M = 6.1, SD = 1.2) (Figure 5.6).

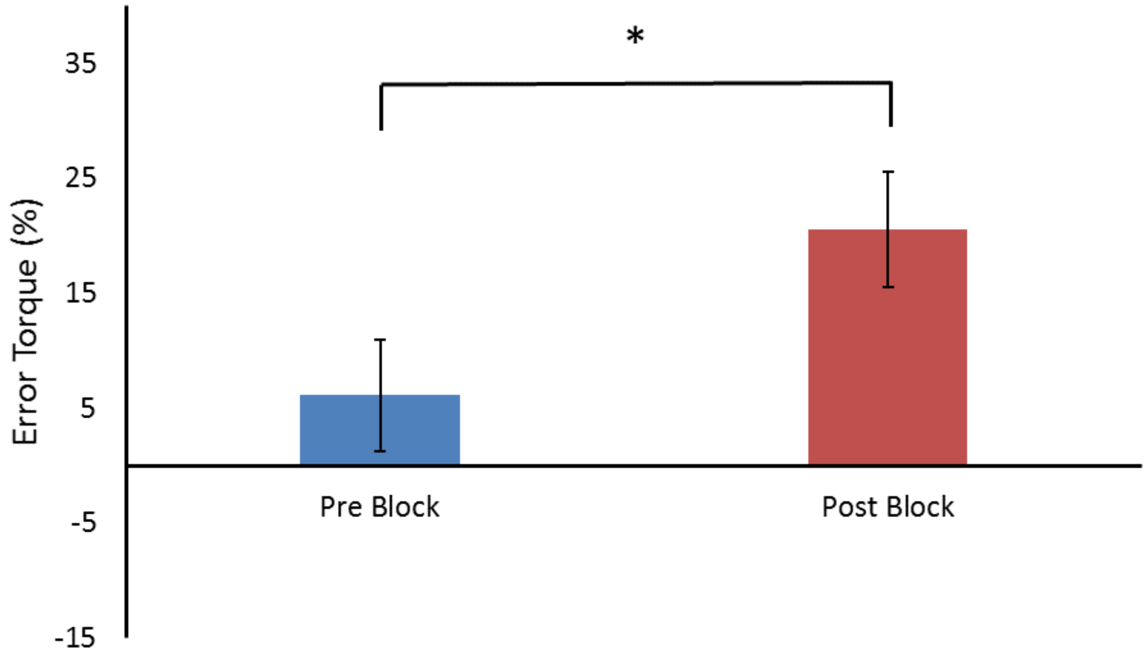

Figure 5.6: Torque error (%) before and after the suprascapular nerve block with the right/blocked side as reference.

Anterior Deltoid EMG Error

The interaction between the condition and load was not significant ($p = 0.67$). The main effect for load was not significant ($p = 0.28$). The main effect for condition was significant ($p = 0.02$). The matching anterior deltoid demonstrated higher activation error in the blocked condition (*M* =*-44.4*, *SD* = 12.4) than in the non-blocked condition (*M* = -12.4, *SD* = 12.0) (Figure 5.7).

Middle Deltoid EMG Error

The interaction between the condition and load was not significant ($p = 0.49$). The main effect for condition was significant ($p = 0.01$). The matching middle deltoid demonstrated higher activation error in the blocked condition (*M* =-68.7, *SD* = 22.5) than the non-blocked condition (*M* = 13.9, *SD* = 14.4) (Figure 5.7). The main effect for load was significant ($p = 0.01$) (Greenhouse-Geisser adjustment).

Posterior Deltoid EMG Error

The interaction between the condition and load was not significant ($p = 0.79$). The main effect for condition was significant ($p = 0.005$). The matching poster deltoid demonstrated higher activation error in the blocked condition (*M* =-70.0, *SD* = 35.0) than the non-blocked condition (*M* = 7.2, *SD* = 23.4) (Figure 5.7). The main effect for load was not significant ($p = 0.058$) (Greenhouse-Geisser adjustment).

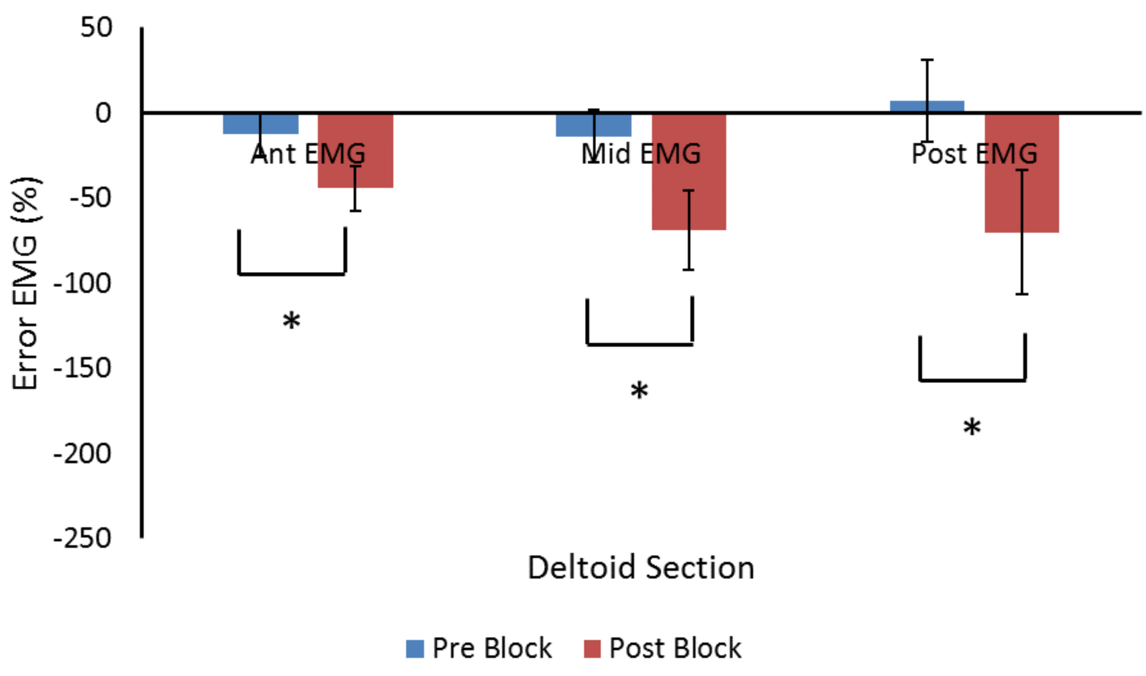

Figure 5.7: EMG error (%) before and after the suprascapular nerve block with the right/blocked side as reference.

Change in Error

The interaction between the error parameter and load was not significant (p = 0.56) (Greenhouse Geisser). The main effect for load was not significant (p = 0.89) (Greenhouse Geisser). The main effect for error parameter was significant (p < 0.001). The follow up simple contrast found a significantly larger change in anterior deltoid EMG error (M = -32.0, SD = 10.3, p < 0.001); middle deltoid EMG error (M = -54.8, SD = 15.26, p = 0.001); and posterior deltoid EMG error (M = -77.2, SD = 18.0, p = 0.001) than the change in torque error (M = 14.5, SD = 4.7) (Figure 5.8).

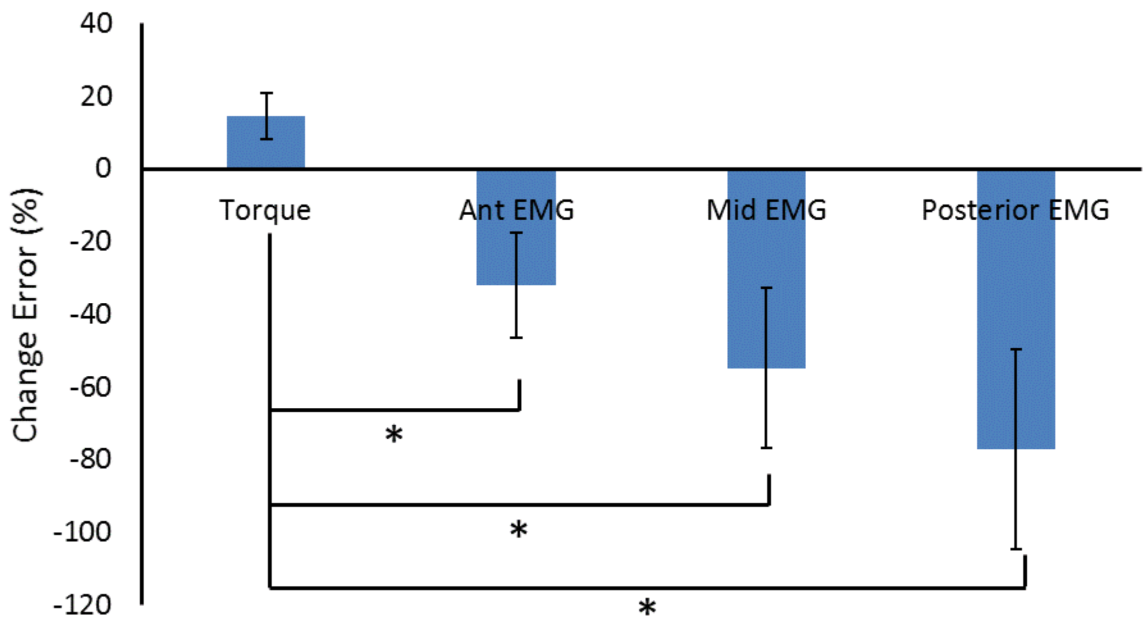

Figure 5.8: Comparison of the change in error (%) from before to after the suprascapular nerve block between torque and EMG amplitude with the right/blocked side as reference.

Force Matching Accuracy

The interaction between the condition and load was not significant (p = 0.83). The main effect for load was not significant (p = 0.39) and the main effect for condition was not significant (p = 0.08) (Figure 5.9).

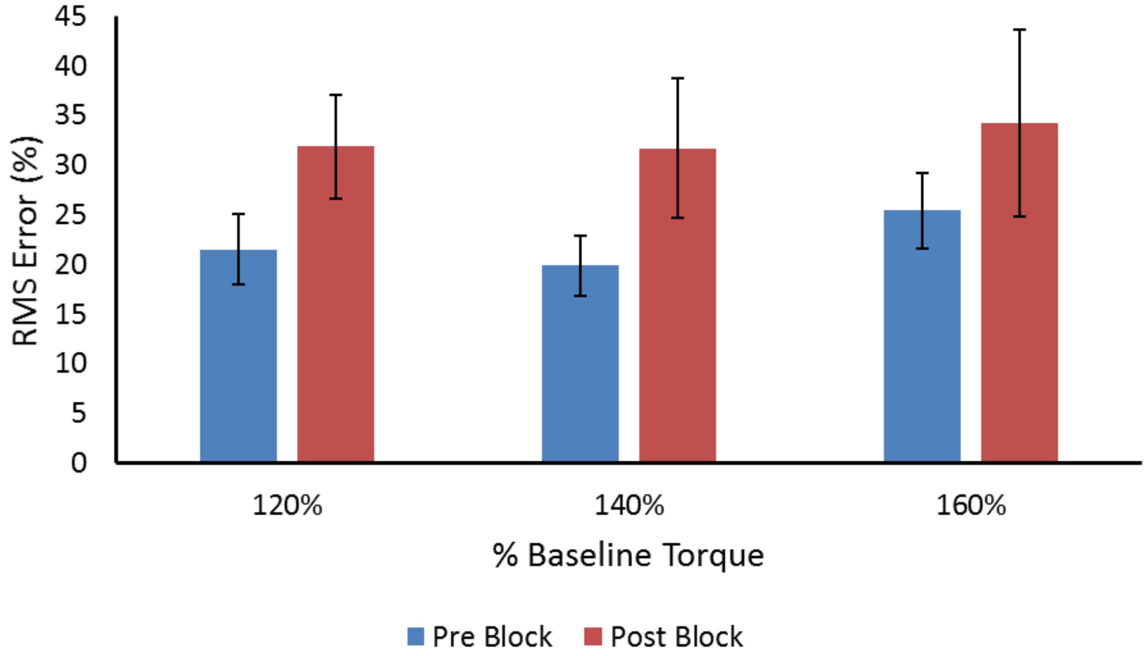

Figure 5.9: RMS torque error (normalized to baseline torque) before and after the suprascapular nerve block with the right/blocked side as reference.

Discussion

The purpose of the study was to determine how the central nervous system accounts for a functional and sensory disruption between sides in a contralateral force matching task. We hypothesized that the subject would undershoot the reference target after the nerve block when the left/unblocked side was the reference and EMG would remain the same and the opposite when the right/blocked side was the reference.

For the left side as refence, subjects only significantly undershot the reference when the load target was 160% of baseline torque after the suprascapular nerve block (Figure 5.2). A much greater change due to block is observed in EMG matching errors across the deltoid muscle (Figure 5.3). In fact, the change score on EMG is seven times greater for the anterior, 17 times greater for the middle and 20 times greater for the posterior deltoid. Even though some effect is observed for torque at the 160% baseline load, the effect is relatively small when considering the EMG changes. The same effect but in the opposite direction is observed when the right/blocked side is the reference. In this case, the change score for the anterior, middle and posterior was 2, 4 and 5 times greater than the change in torque error respectively. The reason for the magnitude difference in error change in EMG between the left and right sides is due to the normalization to the reference value. For the left, the reference remains the same before and after the block while for the right, EMG dramatically increases. Matching torque appears to be more important than matching muscle activity of the deltoid muscles and associated sense of effort (Morree et al., 2012).

Other studies have demonstrated support for the central signal hypothesis. A biceps brachii fatigue force matching model (L. A. Jones & Hunter, 1983). EMG amplitude from the fatigue side was able to predict the force produced by the matching side. Using an eccentric fatigue model, Carson et al., (2002) observed that subjects significantly undershot the reference load when the unfatigued side was the reference. They also saw an increase in EMG amplitude in the matching side but this was not significantly different compared to the reference side's EMG. They hypothesized that the eccentric contractions (through muscle fibre damage) modified the gain between the descending motor command and the perceived amount of effort. This hypothesized change in the gain between the motor command and sense of effort would not have occurred with the suprascapular nerve block. The ratio between motor output and sense of effort would remain the same. Although the force results from that study and the present study occur in the same direction, the effect is smaller in the present study and small in comparison to the changes we observed in EMG. If EMG and associated sense of effort were the primary sources of matching information, it would have led to a dramatic underestimation of force when the left was the reference and an overestimation when the right was the reference. Instead we see a dramatic change in motor output by increases in deltoid EMG but no change in force matching error. Our force error observations are very similar to the curare method by Luu et al., (2011). The subjects' error moved in the direction with hypothesized using sense of effort feedback, but not to the extent that would be expected if only sense of effort was used.

Even though subjects' error direction changed in accordance with our hypothesis – undershot when the left/unblocked was reference and over shot when right/blocked was reference – their overall accuracy in the performance of the force matching task did not statistically change. This further indicates that torque, and not motor neural input to a muscle or sense of effort, is being used to match forces during this task. It should be noted that when the right/blocked side was the reference, accuracy was worse for the group but not statistically significant.

In the absence of vision, the CNS can make use of afferent feedback, motor command or both to estimate the forces. Possible sources of afferent information include: muscle spindles and GTOs within the deltoid muscle; sense of effort in parallel with the motor command and efferent copy; cutaneous receptors located at the wrists; and possibly the lack of information from the supraspinatus muscle on the blocked side. Barring the cutaenous receptors - because error was not great after the block and pressure on the skin would remain almost the same – the suprascapular nerve block would have cause significant disruption to all other sources of afferent information during the task.

The deltoid muscles generated vastly different tensions and the sense of effort would have been far greater on the right side, as evident from the substantially greater activation. The role of muscle spindles in force matching protocols has not been specifically identified. Although muscle spindle firing rates do increase during an isometric contraction it is expected that these signals are filtered out (D. McCloskey et al., 1983). The last afferent consideration is cutaneous receptors, which should not be affected by the suprascapular nerve block. Removal of cutaneous afferents can result in either underestimation (L. Jones & Piateski, 2006) or overestimation (Monzee, Yves, & Smith, 2003) of forces depending on the on whether feedback is attenuated or completely removed. In our study, subjects tended undershot the load with the left/unblocked as the reference and significantly overshot by 21% of reference when the right/blocked was the reference. This overshoot is still well below what would be expected if only sense of effort was being utilized to match the loads. Cutaneous feedback would still have been able to determine a difference between sides. The possibility exists that it may not be sensitive enough or be weighted as heavily by the CNS to prevent error moving in the direction consistent with sense of effort matching but could prevent excessive errors.

Another potential option for the CNS system is to use total afferent feedback. This would require interpretation of the total tension in all muscles or the total strength of the of the motor command (sense of effort) to the contracting muscles. Using this type of model and the left side (unblocked) as the reference, the summed tension of the left supraspinatus and deltoid would need to be the same as the summed tension from the right deltoid and paralyzed supraspinatus. Summed tension could be represented as number of type Ib afferents per second from a muscle group. The same would apply to sense of effort. Error in the system would increase since GTO and muscle spindle firing in the deltoid may not correlate exactly with the supraspinatus. This may also be the reason why subjects were successful at completing the force matching task after the block.

Conclusion

Our results do not support a central corollary discharge model for matching forces between left and right sides. Subjects are able to successfully match forces between sides even after the torque producing capacity of one side is reduced by more than 50% and afferent information from the supraspinatus is lost. This study provides evidence that afferent information is the more likely source used to match contralateral forces. It is not certain how this is done since the afferent information is also mismatched between sides. It is possible that the central nervous system is able to rapidly recalibrate and reinterpret incoming afferent information.

Limitations

In this study, we have assumed that the nerve block does cause any central changes other than increasing the motor drive and perceived effort to the deltoid muscle. It is not certain whether central drive to the block supraspinatus remained unaffected and could not reach the muscle due to the block or if it was reduced, or completely stopped.

CHAPTER VI

CONCLUSIONS

There are many possible causes of shoulder injury. These factors can be stand alone or form a positive feedback injury loop leading to further injury, pain and dysfunction. The experiments described previously were conducted on healthy subjects with no previous or current pain or injury at the shoulder. The population was selected to investigate some of the unknowns in the absence of any pathology. It would be interesting to see similar experiments conducted in a pain or injury population.

As mentioned in the introduction, the dissertation focused on two functional aspect of the shoulder: the supraspinatus muscle and shoulder proprioception. To investigate the supraspinatus the reliability of an isometric ramp technique was investigated. The protocol was found to be highly reliable and time efficient, approximately nine time more efficient, compared to previous techniques. While specific to the deltoid muscle, this technique can likely be applied to any muscle of the body that is accessible by surface EMG.

The isometric ramp contraction technique was then used before and after a suprascapular nerve block to indirectly calculate the contribution of the supraspinatus muscle to submaximal shoulder elevation. The novel finding is that the supraspinatus contributes more than the deltoid up to 50% MVC at all the tested shoulder angles. This finding was surprising for several reasons. Previous research indicated that the supraspinatus contribution was 50% at max but contributed very little to shoulder elevation in other circumstances. The supraspinatus and deltoid contributions are also not specifically related to either their cross-sectional areas or to their moment arms. This observation opens the door for possible motor behavior retraining to optimize the timing of muscle activations to better utilize mechanical advantage. It is important to note that these observations are presumably representative of supraspinatus submaximal functioning in a healthy population. Due to the small sample size, follow up experiments should replicate the results before investing too heavily in the outcomes. The suprascapular nerve block with ramp contraction protocol can also be employed to study rotator cuff tear pathologies. The procedure is minimally invasive, provides pain relief and can determine the functioning of the supraspinatus in a pathological state.

Less enlightening were the joint position sense and force sense results. Like proprioceptive studies at other joints, there appears to be no relationship between joint position sense and force sense at the shoulder. The lack of relationship also appears unrelated to external load or testing angles. Since poor proprioception may form part of positive injury loop, the lack of relationship between different sub-modalities make its clinical interpretation difficult – is joint position sense or force sense important? Because there is no relationship, both would need to be assessed. This increases the amount of time and specialized equipment needed to assess an individual. The reason no relationship exists needs to be determined as well as the specific role the different modalities play in preventing and treating injuries.

From a neuroscience perspective, the suprascapular nerve block created a disassociation between left and right sides. Contrary to the central corollary discharge hypothesis, we found that subjects are not solely using sense of effort to match forces between sides. While subjects did make errors in the hypothesized directions for both sides, the magnitude of the torque errors was dwarfed by the mismatch in EMG between sides. Even though the error direction was in the direction we hypothesized, the accuracy of the task did not change. Afferent feedback appears to play a dominant role. The question that remains is how the central nervous system can successfully integrate mismatched afferent information between sides. Keeping in mind that they protocol was repeated immediately after the block and the system had recalibrated within that time. This also demonstrates rapid neuroplasticity of the central nervous system.

APPENDIX A

RELIABILITY INFORMED CONSENT FORM

UNIVERSITY OF OREGON

University of Oregon Department of Human Physiology

Informed Consent for Participation as a Subject in

"The Relationship between Deltoid EMG and Shoulder Abduction Force "

Investigator: Andrew Karduna, PhD

Introduction

You are being asked to take part in a research study of the relationship between deltoid EMG and shoulder abduction force. You were selected as a possible participant because you are generally in good health and have no shoulder problems. Up to 100 subjects may be recruited for this study. We ask that you read this form and ask any questions that you may have before agreeing to be in the study.

Purpose of Study:

The purpose of this study is to investigate shoulder muscle EMG and shoulder abduction force

Participants in this study are from the University of Oregon and Eugene communities.

Description of the Study Procedures:

If you agree to be in this study, we would ask you to participate in one session of testing. The session will last approximately one hour. We will ask you to do the following things, during the session:

First of all, you will be asked to elevate your arm as hard as you can against manual resistance, and the maximum force of shoulder abduction and EMG of the deltoid, infraspinatus, upper trapezius and seratus anterior will be recorded. After that you will elevate your arm against resistance until maximal force 6 times in 2 different positions. You will then wait 15 minutes. At this point another 6 trials in the 2 positions will be recorded.

The informed consent procedure and the completion of a subject intake form may take from 5 to 15 minutes. The measurement of maximum strength will take 5 minutes. The initial 6 trials will take approximately 10 to 15 minutes. The wait time between testing may be up to 15 minutes. The final 6 trials may take approximately 10 to 15 minutes.

Risks/Discomforts of Being in the Study:

The study has the following risks. You will experience the sensation of muscle fatigue during the maximal muscle contractions~~.~~. You may also feel light muscle soreness after the test (up to two days). However, because you are healthy and have no shoulder problems, the risk will be very low, equivalent to the sensation following intense physical exercise.

Benefits of Being in the Study:

The purpose of the study is to investigate the relationship between deltoid EMG and shoulder abduction force. There is no direct benefit to you by participating in this study. However, you understand that information gained in this study may help health care professionals develop better protocols to evaluate shoulder pathologies.

Payments:

You will receive the following reimbursement: You will be paid $15 for your participation in this study. This is to help defray costs incurred for participating such parking and transportation as well as your time. If you cannot complete the study, you will still be paid $15 for your time.

Costs:

There is no cost to you to participate in this research study.

Confidentiality:

The records of this study will be kept private. In any sort of report we may publish, we will not include any information that will make it possible to identify you as a participant. Research records will be kept in a locked file. All electronic information will be coded and secured using a password protected file.

Access to the records will be limited to the researchers; however, please note that regulatory agencies, and the Institutional Review Board and internal University of Oregon auditors may review the research records.

Voluntary Participation/Withdrawal:

Your participation is voluntary. If you choose not to participate, it will not affect your current or future relations with the University of Oregon. You are free to withdraw at any time, for whatever reason. There is no penalty or loss of benefits for not taking part or for stopping your participation.

Dismissal From the Study:

The investigator may withdraw you from the study at any time for the following reasons: (1) withdrawal is in your best interests (e.g. side effects or distress have resulted), or (2) you have failed to comply with the study requirements.

Disclaimer Statement and Compensation for Injury:

If you experience an emergency medical problem or injury as a direct result of your participation in this research, the investigators of the study will do everything they can to assist you. However, cost of care due to any injury will be covered by the participant and/or his/her insurance company.

Contacts and Questions:

The researcher conducting this study is Andrew Karduna. For questions or more information concerning this research you may contact him at (541) 346-0438, Department of Human Physiology, University of Oregon, Eugene OR, 97403. If you believe you may have suffered a research related injury, contact Andrew Karduna and he will provide you with further instructions.

If you have any questions about your rights as a research subject, you may contact: Research Compliance Services, University of Oregon at (541) 346-2510 or ResearchCompliance@uoregon.edu

Copy of Consent Form:

You will be given a copy of this form to keep for your records and future reference.

Statement of Consent:

I have read (or have had read to me) the contents of this consent form and have been encouraged to ask questions. I have received answers to my questions. I understand the risks and benefits of the study. I understand that my participation in this study is voluntary and that I choose to be in this study. I know I can stop being in the study at any time for any reason. I give my consent to participate in this study. I have received (or will receive) a copy of this form.

Signatures/Dates

Study Participant (Print Name)

Participant Signature **Date**

UNIVERSITY OF OREGON

Study 1

APPENDIX B

SUPRASPINATUS TORUQE INFORMED CONSENT FORM

University of Oregon Department of Human Physiology

Informed Consent for Participation as a Subject in

"Using a Nerve Block to Examine Shoulder Force and Proprioception"

Investigator: Andrew Karduna, PhD

Introduction

You are being asked to take part in a research study of sense of force, a component of proprioception (ability to know where your limb is without seeing it) and supraspinatus (a shoulder muscle) muscle torque. You were selected as a possible participant because you are generally in good health and have no shoulder problems. We ask that you read this form and ask any questions that you may have before agreeing to be in the study.

Purpose of Study:

The purpose of this study is to 1) determine the contribution of supraspinatus muscle torque to shoulder elevation and 2) the effect of load and angle on supraspinatus function. Participants in this study are from the University of Oregon and Eugene communities.

Description of the Study Procedures:

If you agree to be in this study, we would ask you to participate in one session of testing. The session will last approximately 3.5 hours. We may ask you to do the following things, during the session:

In the protocol you will be asked to elevate your arm as hard as you can against manual resistance, to measure the maximal force you can apply against a load cell at 3 different angles. After that, you will perform 9 isometric ramp contractions. An isometric ramp contraction is a contraction where your arm will not move but you apply a force. Your wrist will be strapped to a load cell. You will be asked to perform a ramp contraction 3 times at each angle to 65% of your maximum strength. You will be given visual feedback of the force you need to apply.

Following the completion of the experimental protocol a suprascapular nerve (a shoulder nerve) block will be performed. This procedure involves the injection of anesthetic into the shoulder by a board certified anesthesiologist and is commonly used for the treatment of shoulder related pain. The procedure temporarily paralyzes 2 supporting muscles in the shoulder. During this procedure you will sit with your head flexed forward. The anesthesiologist will palpate your shoulder and insert a needle inject anesthetic. We will then evaluate you to make sure the block was effective.

The experimental protocol done before the nerve block will then be repeated. The informed consent procedure and the completion of a subject intake form may take from 5 to 15 minutes. The protocol testing isometric ramp contractions will take 30 minutes. The nerve block procedure will also take 20 minutes. Repeating the protocol will take 30 minutes. Additional time is included to monitor your status after the experiment. The total testing time will not exceed 3.5 hours.

Female Subjects

An on-site pregnancy test that indicates negative is required to before beginning any experimental protocols.

Risks / Discomforts of Being in the Study:

The experimental protocol can result in slight muscle soreness but not more than is associate with normal exercise. However, because you are healthy and have no shoulder problems, the risk will be very low, equivalent to the sensation following physical exercise. The nerve block procedure has a slight risk of infection, syncope (passing out at the sight or insertion of the needle), bleeding and soreness at the site of injection. Other rare risks include tachycardia (rapid heartbeat) and seizure. Greater risks but even less likely are an allergic reaction and collapsed lung (pneumothorax).

Benefits of Being in the Study:

The purpose of the study is to investigate the supraspinatus torque and force sense at the shoulder. There is no direct benefit to you by participating in this study. However, you understand that information gained in this study may help health care professionals develop better protocols to evaluate shoulder pathologies.

Payments:

You will be compensated $75 for participating in the study. You may withdraw at anytime but this will affect your total compensation. You will receive $20 if you withdraw prior to the nerve block, $40 if you withdraw during or immediately after the nerve block without completing the second portion of the experiment. Please be aware, compensation for participation in research studies may be considered taxable income.

Costs:

There is no cost to you to participate in this research study.

Confidentiality:

The records of this study will be kept private. In any sort of report we may publish, we will not include any information that will make it possible to identify you as a participant. Research records will be kept in a locked file. All electronic information will be coded and secured using a password protected file. The archived data will not be destroyed because it is the intent of the PI to utilize these data in future research projects.

Access to the records will be limited to the researchers; however, please note that regulatory agencies, and the Institutional Review Board and internal University of Oregon auditors may review the research records.

Voluntary Participation / Withdrawal:

Your participation is voluntary. If you choose not to participate, it will not affect your current or future relations with the University of Oregon or Dr Kosek. You are free to withdraw at any time, for whatever reason. There is no penalty or loss of benefits for not taking part or for stopping your participation.

Dismissal From the Study:

The investigator may withdraw you from the study at any time for the following reasons: (1) withdrawal is in your best interests (e.g. side effects or distress have resulted), or (2) you have failed to comply with the study requirements.

Disclaimer Statement and Compensation for Injury:

If you experience an emergency medical problem or injury as a direct result of your participation in this research, the investigators of the study will do everything they can to assist you. However, cost of care due to any injury will be covered by the participant and/or his/her insurance company.

If you experience harm because of the project, you can ask the State of Oregon to pay you. A law called the Oregon Tort Claims Act limits the amount of money you can receive from the State of Oregon if you are harmed. If you have been harmed, there are two University representatives you need to contact. Here are their addresses and phone numbers:

General Counsel

Office of the President

University of Oregon

Eugene, OR 97403 (541) 346-

3082

Research Compliance Services

University of Oregon

Eugene, OR 97403

(541) 346-25

Contacts and Questions:

The researcher conducting this study is Andrew Karduna. For questions or more information concerning this research you may contact him at (541) 346-0438 or karduna@uoregon.edu, Department of Human Physiology, University of Oregon, Eugene OR, 97403. If you believe you may have suffered a research related injury, contact Andrew Karduna and he will provide you with further instructions.

If you have any questions about your rights as a research subject, you may contact: Research Compliance Services, University of Oregon at (541) 346-2510 or ResearchCompliance@uoregon.edu

Copy of Consent Form:

You may request a copy of this form to keep for your records and future reference.

Statement of Consent:

I have read (or have had read to me) the contents of this consent form and have been encouraged to ask questions. I have received answers to my questions. I give my consent to participate in this study. I have received (or will receive) a copy of this form.

Signatures/Dates

____

Study Participant (Print Name)

____

Participant Signature **Date**

APPENDIX C

JPS AND FS CORRELATION INFORMED CONSENT FORM

University of Oregon Department of Human Physiology

Informed Consent for Participation as a Subject in

"The Relationship between Force Sense and Joint Position Sense at the Shoulder"

Investigator: Andrew Karduna, PhD

Introduction

You are being asked to take part in a research study of sense of effort, a component of proprioception. Proprioception is the awareness of your arm's position without visual feedback. You were selected as a possible participant because you are generally in good health and have no shoulder problems. We ask that you read this form and ask any questions that you may have before agreeing to be in the study.

Purpose of Study:

The purpose of this study is to investigate the relationship between 2 sub-modalities of proprioception: force sense and joint position sense.

Participants in this study are from the University of Oregon and Eugene communities.

Description of the Study Procedures:

If you agree to be in this study, we would ask you to participate in one session of testing. The session will last approximately 2 hours. We will ask you to do the following things, during the session:

First of all, you will be asked to elevate your arm as hard as you can against manual resistance, to measure the maximal force you can apply against a load cell. The contractions will be isometric where the maximum force is developed by the muscle but no movement at the joint occurs. This will be done twice at three different shoulder angles totaling 6 maximal contractions. After that, the force sense of your shoulder will be tested. Your wrist will be strapped to a load cell. After that you will apply different levels of submaximal force (≤50% of maximum) up 20 times to the load cell (a force measuring device) for 3 different shoulder angles. The study is limited to maximum number of 60 trials total. You will be given visual feedback of the force level. You will memorize the amount of effort on each occasion and then reproduce that effort without visual feedback.

You well then perform a joint position sense task with different weights on the arm. You will elevate your arm to different angles with visual feedback. You will memorize the position on each occasion and then reproduce that position without visual feedback. The joint position sense task consists of 12-20 trials for each weight. You will perform the protocol 4 times with different weights.

The informed consent procedure and the completion of a subject intake form may take from 5 to 15 minutes. The measurement of maximum strength will take 5 minutes. The total testing time will not exceed 2 hours.

Risks / Discomforts of Being in the Study:

The study has the following risks. You may also feel light muscle soreness after the test (up to two days). However, because you are healthy and have no shoulder problems, the risk will be very low, equivalent to the sensation following physical exercise.

Benefits of Being in the Study:

The purpose of the study is to investigate the relationship between force sense and joint position sense, a components of proprioception, at different levels of force and shoulder abduction angles. There is no direct benefit to you by participating in this study. However, you understand that information gained in this study may help health care professionals develop better protocols to evaluate shoulder pathologies.

Payments:

You will receive the following reimbursement: You will be paid $15 via a check for your participation in this study. This is to help defray the costs incurred for participation such as parking and transportation as well as your time. If you cannot complete the study, you will not be paid.

Please note, compensation from participation in Human Subjects Research studies is taxable income. If your compensation totals $600 or more in a calendar year, the University of Oregon is required to report the income to the IRS. The University requires its departments to tack participant compensation and may contact you to complete a Form W-9 for tax reporting purposes. Because of the federal and University tracking requirements, your name will be associated with participation in research. Department and University administrators will have access to this information, but will not have access to research data.

Costs:

There is no cost to you to participate in this research study.

Confidentiality:

The records of this study will be kept private. In any sort of report we may publish, we will not include any information that will make it possible to identify you as a participant. Research records will be kept in a locked file. All electronic information will be coded and secured using a password protected file.

Access to the records will be limited to the researchers; however, please note that regulatory agencies, and the Institutional Review Board and internal University of Oregon auditors may review the research records.

Voluntary Participation/Withdrawal:

Your participation is voluntary. If you choose not to participate, it will not affect your current or future relations with the University of Oregon. You are free to withdraw at any time, for whatever reason. There is no penalty or loss of benefits for not taking part or for stopping your participation.

Dismissal from the Study:

The investigator may withdraw you from the study at any time for the following reasons: (1) withdrawal is in your best interests (e.g. side effects or distress have resulted), or (2) you have failed to comply with the study requirements.

Disclaimer Statement and Compensation for Injury:

If you experience an emergency medical problem or injury as a direct result of your participation in this research, the investigators of the study will do everything they can to assist you. However, cost of care due to any injury will be covered by the participant and/or his/her insurance company.

Contacts and Questions:

The researcher conducting this study is Dr. Andrew Karduna. For questions or more information concerning this research you may contact him at (541) 346-0438, karduna@uoregon.edu, Department of Human Physiology, University of Oregon, Eugene OR, 97403. If you believe you may have suffered a research related injury, contact Dr. Karduna and he will provide you with further instructions.

If you have any questions about your rights as a research subject, you may contact: Research Compliance Services, University of Oregon at (541) 346-2510 or ResearchCompliance@uoregon.edu

Copy of Consent Form:

You will be given a copy of this form to keep for your records and future reference.

Statement of Consent:

I have read (or have had read to me) the contents of this consent form and have been encouraged to ask questions. I have received answers to my questions. I give my consent to participate in this study. I have received (or will receive) a copy of this form.

Signatures/Dates

____

Study Participant (Print Name)

____

Participant Signature **Date**

Study 2

APPENDIX D

FORCE SENSE NERVE BLOCK INFORMED CONSENT FORM

University of Oregon Department of Human Physiology

Informed Consent for Participation as a Subject in

"Using a Nerve Block to Examine Shoulder Force and Proprioception"

Investigator: Andrew Karduna, PhD

Introduction

You are being asked to take part in a research study of sense of force, a component of proprioception (ability to know where your limb is without seeing it) and supraspinatus (a shoulder muscle) muscle torque. You were selected as a possible participant because you are generally in good health and have no shoulder problems. We ask that you read this form and ask any questions that you may have before agreeing to be in the study.

Purpose of Study:

The purpose of this study is to determine the source of proprioceptive feedback used by the CNS during a contralateral force matching task.

Participants in this study are from the University of Oregon and Eugene communities.

Description of the Study Procedures:

If you agree to be in this study, we would ask you to participate in one session of testing. The session will last approximately 3.5 hours. We may ask you to do the following things, during the session:

In the protocol you will complete a force matching task between your arms. Both your wrists will be attached to load cells (a force measuring device). For this portion your view will be occluded with a virtual reality headset which will guide you to a target force level with your dominant arm. You will be asked to replicate this force level with your other arm. You will perform 4 trials for each of the 3 target loads (12 trials total). The protocol will then be repeated using your non dominant hand as the reference.

Following the completion of the experimental protocol a topical anesthetic will be applied to both wrists to numb the skin and you will repeat the force matching task. After this a suprascapular nerve (a shoulder nerve) block will be performed. This procedure involves the injection of anesthetic into the shoulder by a board certified anesthesiologist and is commonly used for the treatment of shoulder related pain. The procedure temporarily paralyzes 2 supporting muscles in the shoulder. During this procedure you will sit with your head flexed forward. The anesthesiologist will palpate your shoulder and insert a needle inject anesthetic. We will then evaluate you to make sure the block was effective. The experimental protocol done before the nerve block will then be repeated. The force matching protocol will be repeated a total of 3 times.

The informed consent procedure and the completion of a subject intake form may take from 5 to 15 minutes. The protocol testing force sense will take 20 minutes. The nerve block procedure will also take 20 minutes. Repeating the protocol will take 20 minutes. Additional time is included to monitor your status after the experiment. The total testing time will be approximately 3.5 hours.

Female Subjects

An on-site pregnancy test that indicates negative is required to before beginning any experimental protocols.

Risks/Discomforts of Being in the Study:

The experimental protocol can result in slight muscle soreness but not more than is associate with normal exercise. However, because you are healthy and have no shoulder problems, the risk will be very low, equivalent to the sensation following physical exercise. The nerve block procedure has a slight risk of infection, syncope (passing out at the sight or insertion of the needle), bleeding and soreness at the site of injection. Other rare risks include tachycardia (rapid heartbeat) and seizure. Greater risks but even less likely are an allergic reaction and collapsed lung (pneumothorax).

Benefits of Being in the Study:

The purpose of the study is to investigate the supraspinatus torque and force sense at the shoulder. There is no direct benefit to you by participating in this study. However, you understand that information gained in this study may help health care professionals develop better protocols to evaluate shoulder pathologies.

Payments:

You will be compensated $75 for participating in the study. You may withdraw at anytime but this will affect your total compensation. You will receive $20 if you withdraw prior to the nerve block, $40 if you withdraw during or immediately after the nerve block without completing the second portion of the experiment. Please be aware, compensation for participation in research studies may be considered taxable income.

Costs:

There is no cost to you to participate in this research study.

Confidentiality:

The records of this study will be kept private. In any sort of report we may publish, we will not include any information that will make it possible to identify you as a participant. Research records will be kept in a locked file. All electronic information will be coded and secured using a password protected file. The archived data will not be destroyed because it is the intent of the PI to utilize these data in future research projects.

Access to the records will be limited to the researchers; however, please note that regulatory agencies, and the Institutional Review Board and internal University of Oregon auditors may review the research records.

Voluntary Participation/Withdrawal:

Your participation is voluntary. If you choose not to participate, it will not affect your current or future relations with the University of Oregon or Dr Kosek. You are free to withdraw at any time, for whatever reason. There is no penalty or loss of benefits for not taking part or for stopping your participation.

Dismissal From the Study:

The investigator may withdraw you from the study at any time for the following reasons: (1) withdrawal is in your best interests (e.g. side effects or distress have resulted), or (2) you have failed to comply with the study requirements.

Disclaimer Statement and Compensation for Injury:

If you experience an emergency medical problem or injury as a direct result of your participation in this research, the investigators of the study will do everything they can to assist you. However, cost of care due to any injury will be covered by the participant and/or his/her insurance company.

If you experience harm because of the project, you can ask the State of Oregon to pay you. A law called the Oregon Tort Claims Act limits the amount of money you can receive from the State of Oregon if you are harmed. If you have been harmed, there are two University representatives you need to contact. Here are their addresses and phone numbers:

General Counsel

Office of the President

University of Oregon

Eugene, OR 97403 (541) 346-

3082

Research Compliance Services

University of Oregon

Eugene, OR 97403

(541) 346-25

UNIVERSITY OF OREGON

Contacts and Questions:

The researcher conducting this study is Andrew Karduna. For questions or more information concerning this research you may contact him at (541) 346-0438 or karduna@uoregon.edu, Department of Human Physiology, University of Oregon, Eugene OR, 97403. If you believe you may have suffered a research related injury, contact Andrew Karduna and he will provide you with further instructions.

If you have any questions about your rights as a research subject, you may contact: Research Compliance Services, University of Oregon at (541) 346-2510 or ResearchCompliance@uoregon.edu

Copy of Consent Form:

You may request a copy of this form to keep for your records and future reference.

Statement of Consent:

I have read (or have had read to me) the contents of this consent form and have been encouraged to ask questions. I have received answers to my questions. I give my consent to participate in this study. I have received (or will receive) a copy of this form.

Signatures/Dates

____

Study Participant (Print Name)

____

Participant Signature **Date**

APPENDIX E

SUBJECT INTAKE FORM

Subject Intake Form

Subject Code ____________________

Date ________________________ Dominant Side ___________________

Right Arm Length_______________ Right Hand Length_________________

Age ___________________________ Gender _________________________

Weight _______________________ Height ____________________________

APPENDIX F

EDINBURG HANDEDNESS TEST

Edinburgh Handedness Inventory

Subject ID ___________________________

Please indicate your preferences in the use of hands in the following activities by *putting + in the appropriate column*. Where the preference is so strong that you would never try to use the other hand unless absolutely forces to, put ++. If any case you are really indifferent put + in both columns. Some of the activities require both hands. In these cases the part of the task, or object, for which hand preference is wanted is indicated in brackets.Please try to answer all the questions, and only leave a blank if you have no experience at all of the object or task.

	Left	Right
1. Writing		
2. Drawing		
3. Throwing		
4. Scissors		
5. Toothbrush		
6. Knife (without fork)		
7. Spoon		
8. Broom (upper hand)		
9. Striking Match (match)		
10. Opening box (lid)		
i. Which foot do you prefer to kick with?		
ii. Which eye do you use when using only one?		

REFERENCES CITED

Ackland, D. C., Pak, P., Richardson, M., & Pandy, M. G. (2008). Moment arms of the muscles crossing the anatomical shoulder. *Journal of Anatomy*, *213*(4), 383–390. http://doi.org/10.1111/j.1469-7580.2008.00965.x

Alpert, S. W., Pink, M. M., Jobe, F. W., McMahon, P. J., & Mathiyakom, W. (2000). Electromyographic analysis of deltoid and rotator cuff function under varying loads and speeds. *Journal of Shoulder and Elbow Surgery*, *9*(1), 47–58.

Aluisio, F. V, Osbahr, D. C., & Speer, K. P. (2003). Analysis of rotator cuff muscles in adult human cadaveric specimens. *American Journal of Orthopedics (Belle Mead, N.J.)*, *32*(3), 124–129.

Anderson, V. B., & Wee, E. (2011). Impaired joint proprioception at higher shoulder elevations in chronic rotator cuff pathology. *Archives of Physical Medicine and Rehabilitation*, *92*(7), 1146–1151.

Apreleva, M., Parsons, I. M., Warner, J. J., Fu, F. H., & Woo, S. L. (2000). Experimental investigation of reaction forces at the glenohumeral joint during active abduction. *Journal of Shoulder and Elbow Surgery / American Shoulder and Elbow Surgeons ... [et Al.]*, *9*(5), 409–17. http://doi.org/10.1067/mse.2000.106321

Blasier, R., Carpenter, J., & Huston, L. (1994). Shoulder proprioception. Effect of joint laxity, joint position, and direction of motion. *Orthopaedic Review*, *23*(1), 45–50.

Boettcher, C. E., Ginn, K. A., & Cathers, I. (2008). Standard maximum isometric voluntary contraction tests for normalizing shoulder muscle EMG. *Journal of Orthopaedic Research*, *26*(12), 1591–1597. http://doi.org/10.1002/jor.20675

Boettcher, C., Ginn, K., & Cathers, I. (2009). Which is the optimal exercise to strengthen supraspinatus? *Medicine and Science in Sports and Exercise*, *41*(11), 1979–83. http://doi.org/10.1249/MSS.0b013e3181a740a7

Borsa, P. A., & Lephart, S. M. (1994). Functional Assessment and Rehabilitation of Shoulder Proprioception for Glenohumeral Instability. *Journal of Sport Rehabilitation*. http://doi.org/Article

Bouaicha, S., Slankamenac, K., Moor, B. K., Tok, S., Andreisek, G., & Finkenstaedt, T. (2016). Cross-sectional area of the rotator cuff muscles in MRI - Is there evidence for a biomechanical balanced shoulder? *PLoS ONE*, *11*(6), e0157946. http://doi.org/10.1371/journal.pone.0157946

Brox, J. I. (2003). Shoulder pain. *Best Practice and Research: Clinical Rheumatology*, *17*(1), 33–56. http://doi.org/10.1053/yberh.2003.273

Cafarelli, E., & Bigland-Ritchie, B. (1979). Sensation of static force in muscles of different length. *Experimental Neurology*, *65*(3), 511–525. http://doi.org/10.1016/0014-4886(79)90040-2

Carson, R. G., Riek, S., & Shahbazpour, N. (2002). Central and peripheral mediation of human force sensation following eccentric or concentric contractions. *The Journal of Physiology*, *539*(Pt 3), 913–25. http://doi.org/10.1013/jphysiol.2001.013385

Chard, M. D., Hazleman, R., Hazleman, B. L., King, R. H., & Reiss, B. B. (1991). Shoulder disorders in the elderly: a community survey. *Arthritis and Rheumatism*, *34*(6), 766–769. http://doi.org/10.1002/art.1780340619

Christou, E. a, Zelent, M., & Carlton, L. G. (2003). Force control is greater in the upper compared with the lower extremity. *Journal of Motor Behavior*, *35*(4), 322–4.

Cordo, P., Gurfinkel, V. S., Bevan, L., & Kerr, G. K. (1995). Proprioceptive consequences of tendon vibration during movement. *Journal of Neurophysiology*, *74*(4), 1675–1688.

De Wilde, L., Audenaert, E., Barbaix, E., Audenaert, A., & Soudan, K. (2002). Consequences of deltoid muscle elongation on deltoid muscle performance: A computerised study. *Clinical Biomechanics*, *17*(7), 499–505. http://doi.org/10.1016/S0268-0033(02)00065-7

de Witte, P. B. B., Werner, S., ter Braak, L. M. M., Veeger, H. E. J. E. J., Nelissen, R. G. H. H. G. H. H., & de Groot, J. H. H. (2014). The Supraspinatus and the Deltoid – Not just two arm elevators. *Human Movement Science*, *33*(1), 273–283. http://doi.org/10.1016/j.humov.2013.08.010

Docherty, C. L., & Arnold, B. L. (2008). Force sense deficits in functionally unstable ankles. *Journal of Orthopaedic Research*, *26*(11), 1489–1493. http://doi.org/10.1002/jor.20682

Docherty, C. L., Arnold, B. L., Gansneder, B. M., Hurwitz, S. R., & Gieck, J. H. (2004). Individuals with functional ankle instability have contralateral force sense deficits but not joint position sense deficits. *Medicine & Science in Sports & Exercise*, *36*(5), s187.

Docherty, C. L., Arnold, B. L., Zinder, S. M., Granata, K., & Gansneder, B. M. (2004). Relationship between two proprioceptive measures and stiffness at the ankle. *Journal of Electromyography and Kinesiology*, *14*(3), 317–324.

Doheny, E. P., Lowery, M. M., FitzPatrick, D. P., & O'Malley, M. J. (2008). Effect of elbow joint angle on force-EMG relationships in human elbow flexor and extensor muscles. *Journal of Electromyography and Kinesiology*, *18*(5), 760–770.

Dover, G., & Powers, M. E. (2003). Reliability of Joint Position Sense and Force-Reproduction Measures during Internal and External Rotation of the Shoulder. *Journal of Athletic Training*, *38*(4), 304–310.

Fallon, J. B., & Macefield, V. G. (2007). Vibration sensitivity of human muscle spindles and golgi tendon organs. *Muscle and Nerve*, *36*(1), 21–29. http://doi.org/10.1002/mus.20796

Fehringer, E. V., Sun, J., VanOeveren, L. S., Keller, B. K., & Matsen, F. A. (2008). Full-thickness rotator cuff tear prevalence and correlation with function and co-morbidities in patients sixty-five years and older. *Journal of Shoulder and Elbow Surgery*, *17*(6), 881–885. http://doi.org/10.1016/j.jse.2008.05.039

Gatti, C. J., Dickerson, C. R., Chadwick, E. K., Mell, A. G., & Hughes, R. E. (2007). Comparison of model-predicted and measured moment arms for the rotator cuff muscles. *Clinical Biomechanics*, *22*(6), 639–644. http://doi.org/10.1016/j.clinbiomech.2007.02.001

Gregory, J. E., Brockett, C. L., Morgan, D. L., Whitehead, N. P., & Proske, U. (2002). Effect of eccentric muscle contractions on Golgi tendon organ responses to passive and active tension in the cat. *The Journal of Physiology*, *538*(Pt 1), 209–218. http://doi.org/PHY_12785 [pii]

Gregory, J. E., Morgan, D. L., & Proske, U. (2004). Responses of muscle spindles following a series of eccentric contractions. *Experimental Brain Research*, *157*(2), 234–240. http://doi.org/10.1007/s00221-004-1838-9

Hadzic, A. (2007). *Textbook of Regional Anesthesia and Acute Pain Mangement* (1st ed.). New York, NY: Mcgraw-Hill.

Halder, a. M., Zhao, K. D., O'Driscoll, S. W., Morrey, B. F., & An, K. N. (2001). Dynamic contributions to superior shoulder stability. *Journal of Orthopaedic Research*, *19*(2), 206–212. http://doi.org/10.1016/S0736-0266(00)00028-0

Han, J., Waddington, G., Adams, R., Anson, J., & Liu, Y. (2015). Assessing proprioception: A critical review of methods. *Journal of Sport and Health Science*. http://doi.org/10.1016/j.jshs.2014.10.004

Hasvold, T., & Johnsen, R. (1993). Headache and neck or shoulder pain--frequent and disabling complaints in the general population. *Scandinavian Journal of Primary Health Care*, *11*(3), 219–24.

Howell, S. M., Imobersteg, A. M., Seger, D. H., & Marone, P. J. (1986). Clarification of the role of the supraspinatus muscle in shoulder function. *The Journal of Bone and Joint Surgery. American Volume*, *68*(3), 398–404.

Jones, L. A., & Hunter, I. W. (1983). Effect of Fatigue on Force Sensation. *Experimental Neurology*, *81*, 640–650.

Jones, L., & Piateski, E. (2006). Contribution of tactile feedback from the hand to the perception of force. *Experimental Brain Research*, *168*, 298–302.

Jørgensen, M., & Andersen, L. (2010). Muscle activity during functional coordination training: implications for strength gain and rehabilitation. *The Journal of …*, 1732–1739.

Juan, J. G. S., Kosek, P., & Karduna, A. R. (2013). Humeral Head Translation After a Suprascapular Nerve Block, 371–379.

Juan, J. S., Kosek, P., & Karduna, A. (2013). Humeral head translation after a suprascapular nerve block. *Journal of Applied …*.

Kafa, N., Ataoglu, M. B., Hazar, Z., Citaker, S., & Ozer, M. (2014). Relationship Between Functional Knee Joint Position Sense and Functional Performance Scores Following Anterior Cruciate Ligament Reconstruction (Pilot Study). *Orthopaedic Journal of Sports Medicine*, *2*(3 Suppl), 2016. http://doi.org/10.1177/2325967114S00278

Kakuda, N., & Nagaoka, M. (1998). Dynamic response of human muscle spindle afferents to stretch during voluntary contraction. *Journal of Physiology*, *513*, 621–628.

Kandel, E., Schwartz, J., & Jessell, T. (2000). *Principles of Neuroscience* (4th ed.). New York, NY: Mcgraw-Hill.

Kaya, D., Callaghan, M. J., Donmez, G., & Doral, M. N. (2012). Shoulder joint position sense is negatively correlated with free-throw percentage in professional basketball players. *Isokinetics and Exercise Science*, *20*(3), 189–196. http://doi.org/10.3233/IES-2012-0458

Kim, C.-Y., Choi, J.-D., & Kim, H.-D. (2014). No correlation between joint position sense and force sense for measuring ankle proprioception in subjects with healthy and functional ankle instability. *Clinical Biomechanics*, *29*(9), 977–983.

King, J., Harding, E., & Karduna, A. (2013). The shoulder and elbow joints and right and left sides demonstrate similar joint position sense. *Journal of Motor Behavior*, *45*(6), 479–86.

Kollmitzer, J., Ebenbichler, G., & Kopf, A. (1999). Reliability of surface electromyographic measurements. *Clinical Neurophysiology*, *110*, 725–734.

Kupa, E. J., Roy, S. H., Kandarian, S. C., & De Luca, C. J. (1995). Effects of muscle fiber type and size on EMG median frequency and conduction velocity. *Journal of Applied Physiology (Bethesda, Md. : 1985)*, *79*(1), 23–32.

Kuriki, H., Mello, E., & Azevedo, F. de. (2012). *The relationship between electromyography and muscle force*. (M. Schwartz, Ed.). InTech. http://doi.org/10.5772/25391

Lafargue, G., Paillard, J., Lamarre, Y., & Sirigu, A. (2003). Production and perception of grip force without proprioception: Is there a sense of effort in deafferented subjects? *European Journal of Neuroscience*, *17*(12), 2741–2749.

Lawrence, J. H., & De Luca, C. J. (1983). Myoelectric signal versus force relationship in different human muscles. *Journal of Applied Physiology (Bethesda, Md. : 1985)*, *54*(6), 1653–1659.

Lephart, S. M., Myers, J. B., Bradley, J. P., & Fu, F. H. (2002). Shoulder proprioception and function following thermal capsulorraphy. *Arthroscopy*, *18*(7), 770–778. http://doi.org/10.1053/jars.2002.32843

Li, L., Ji, Z.-Q., Li, Y.-X., & Liu, W.-T. (2016). Correlation study of knee joint proprioception test results using common test methods. *The Journal of Physical Therapy Science*, *28*, 478–482.

Linsell, L., Dawson, J., Zondervan, K., Rose, P., Randall, T., Fitzpatrick, R., & Carr, A. (2006). Prevalence and incidence of adults consulting for shoulder conditions in UK primary care; patterns of diagnosis and referral. *Rheumatology*, *45*(2), 215–221. http://doi.org/10.1093/rheumatology/kei139

Liu, J., Hughes, R. E., Smutz, W. P., Niebur, G., & Nan-An, K. (1997). Roles of deltoid and rotator cuff muscles in shoulder elevation. *Clinical Biomechanics*, *12*(1), 32–38. http://doi.org/10.1016/S0268-0033(96)00047-2

Luu, B. L., Day, B. L., Cole, J. D., & Fitzpatrick, R. C. (2011). The fusimotor and reafferent origin of the sense of force and weight. *The Journal of Physiology*, *589*(Pt 13), 3135–3147. http://doi.org/10.1113/jphysiol.2011.208447

Madeleine, P., Bajaj, P., Søgaard, K., & Arendt-Nielsen, L. (2001). Mechanomyography and electromyography force relationships during concentric, isometric and eccentric contractions. *Journal of Electromyography and Kinesiology*, *11*(2), 113–121.

Maenhout, A. G., Palmans, T., De Muynck, M., De Wilde, L. F., & Cools, A. M. (2012). The impact of rotator cuff tendinopathy on proprioception, measuring force sensation. *Journal of Shoulder and Elbow Surgery*, *21*(8), 1080–1086. http://doi.org/10.1016/j.jse.2011.07.006

Marieb, E., & Hoehn, K. (2016). *Human Anatomy and Physiology* (10th ed.). New York, NY: Pearson Education.

McCloskey, D., Gandevia, S., Potter, E., & Colebatch, J. (1983). Muscle Sense and Effort: Motor Commands and Judgements About Muscular Contractions. *Advances in Neurology*, *39*, 151–167.

McCloskey, D. I., Ebeling, P., & Goodwin, G. M. (1974). Estimation of weights and tensions and apparent involvement of a “sense of effort”. *Experimental Neurology*, *42*(1), 220–232. http://doi.org/10.1016/0014-4886(74)90019-3

McCully, S. P., Suprak, D. N., Kosek, P., & Karduna, A. R. (2007). Suprascapular nerve block results in a compensatory increase in deltoid muscle activity. *Journal of Biomechanics*, *40*(8), 1839–1846. http://doi.org/10.1016/j.jbiomech.2006.07.010

Meislin, R. J., Sperling, J. W., & Stitik, T. P. (2005). Persistent shoulder pain: epidemiology, pathophysiology, and diagnosis. *American Journal of Orthopedics (Belle Mead, N.J.)*, *34*(12 Suppl), 5–9.

Merletti, R., Rainoldi, A., & Farina, D. (2004). Myoelectric Manifestations of Muscle Fatigue. In *Electromyography: Physiology, Engineering and Noninvasive Application* (pp. 233–253). Hoboken, NJ: John Wiley & Sons, Inc.

Michener, L. a., McClure, P. W., & Karduna, A. R. (2003). Anatomical and biomechanical mechanisms of subacromial impingement syndrome. *Clinical Biomechanics*, *18*(5), 369–379. http://doi.org/10.1016/S0268-0033(03)00047-0

Milgrom, C., Schaffler, M., Gilbert, S., & van Holsbeeck, M. (1995). Rotator-cuff changes in asymptomatic adults. The effect of age, hand dominance and gender. *The Journal of Bone and Joint Surgery*, *77*(2), 296–8.

Milner-Brown, H. S., & Stein, R. B. (1975). The relation between the surface electromyogram and muscular force. *The Journal of Physiology*, *246*(3), 549–69. http://doi.org/10.1113/jphysiol.1975.sp010904

Monzee, J., Yves, L., & Smith, A. (2003). The Effects of Digital Anesthesia on Force Control Using a Precision Grip. *Journal of Neurophysiology*, *89*, 672–683.

Moore, K., Dalley, A., & Agur, A. (2014). *Clinically Oriented Anatomy* (7th ed.). Baltimore, MD: Wolters Kluwer.

Moritani, T., & DeVries, H. (1978). Reexamination of the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction. *American Journal of Physical Medicine*, *57*(6), 263–277.

Morree, H. M. De, Klein, C., & Marcora, S. M. (2012). Perception of effort reflects central motor command during movement execution, *49*, 1242–1253.

Naseri, N., & Pourkazemi, F. (2012). Difference in knee joint position sense in athletes with and without patellofemoral pain syndrome. *Knee Surgery, Sports Traumatology, Arthroscopy*, *20*(10), 2071–2076. http://doi.org/10.1007/s00167-011-1834-0

Oatis, C. (2009). *Kinesiology: The Mechanics and Pathomechanics of Human Movement* (2nd ed.). Baltimore, MD: Wolters Kluwer.

Otis, J. C., Jiang, C. C., Wickiewicz, T. L., Peterson, M. G., Warren, R. F., & Santner, T. J. (1994). Changes in the moment arms of the rotator cuff and deltoid muscles with abduction and rotation. *The Journal of Bone and Joint Surgery. American Volume*, *76*(5), 667–76.

Parsons, S., Breen, a., Foster, N. E., Letley, L., Pincus, T., Vogel, S., & Underwood, M. (2007). Prevalence and comparative troublesomeness by age of musculoskeletal pain in different body locations. *Family Practice*, *24*(June), 308–316. http://doi.org/10.1093/fampra/cmm027

Proske, U., & Gandevia, S. C. (2012). The Proprioceptive Senses: Their Roles in Signaling Body Shape, Body Position and Movement, and Muscle Force. *Physiological Reviews*, *92*(4), 1651–1697. http://doi.org/10.1152/physrev.00048.2011

Proske, U., Gregory, J. E., Morgan, D. L., Percival, P., Weerakkody, N. S., & Canny, B. J. (2004). Force matching errors following eccentric exercise. *Human Movement Science*, *23*(3–4 SPE. ISS.), 365–378. http://doi.org/10.1016/j.humov.2004.08.012

Pynn, L. K., & DeSouza, J. F. X. (2013). The function of efference copy signals: implications for symptoms of schizophrenia. *Vision Research*, *76*, 124–33. http://doi.org/10.1016/j.visres.2012.10.019

Qi, L., Wakeling, J. M., Green, A., Lambrecht, K., & Ferguson-Pell, M. (2011). Spectral properties of electromyographic and mechanomyographic signals during isometric ramp and step contractions in biceps brachii. *Journal of Electromyography and Kinesiology*, *21*(1), 128–135. http://doi.org/10.1016/j.jelekin.2010.09.006

Rassier, D., MacIntosh, B., & Herzog, W. (1999). Length dependence of active force production in skeletal muscle. *Journal of Applied Physiology*, *86*(5), 1445–1457.

Reed, D., Cathers, I., Halaki, M., & Ginn, K. (2013). Does supraspinatus initiate shoulder abduction? *Journal of Electromyography and Kinesiology*, *23*(2), 425–429. http://doi.org/10.1016/j.jelekin.2012.11.008

Reed, D., Cathers, I., Halaki, M., & Ginn, K. A. (2016). Does load influence shoulder muscle recruitment patterns during scapular plane abduction? *Journal of Science and Medicine in Sport*, *19*(9), 755–760. http://doi.org/10.1016/j.jsams.2015.10.007

Reinold, M. M., Macrina, L. C., Wilk, K. E., Fleisig, G. S., Dun, S., Barrentine, S. W., … Andrews, J. R. (2007). Electromyographic analysis of the supraspinatus and deltoid muscles during 3 common rehabilitation exercises. *Journal of Athletic Training*, *42*(4), 464–9.

Relph, N., Herrington, L., & Tyson, S. (2014). The effects of ACL injury on knee proprioception: A meta-analysis. *Physiotherapy (United Kingdom)*, *100*(3), 187–195.

Ricard, M. D., Ugrinowitsch, C., Parcell, a. C., Hilton, S., Rubley, M. D., Sawyer, R., & Poole, C. R. (2005). Effects of rate of force development on EMG amplitude and frequency. *International Journal of Sports Medicine*, *26*(1), 66–70.

Riemann, B. L., & Lephart, S. M. (2002). The sensorimotor system, part I: The physiologic basis of functional joint stability. *Journal of Athletic Training*, *37*(1), 71–79. http://doi.org/10.1016/j.jconhyd.2010.08.009

Saito, A., & Akima, H. (2013). Knee joint angle affects EMG-force relationship in the vastus intermedius muscle. *Journal of Electromyography and Kinesiology*, *23*(6), 1406–1412.

Sevrez, V., & Bourdin, C. (2015). On the Role of Proprioception in Making Free Throws in Basketball. *Research Quarterly for Exercise and Sport*, *1367*(May), 1–7. http://doi.org/10.1080/02701367.2015.1012578

Stock, M. S., Drusch, A. S., & Thompson, B. J. (2014). The minimum number of contractions required to examine the EMG amplitude versus isometric force relationship for the vastus lateralis and vastus medialis. *Journal of Electromyography and Kinesiology*, *24*(6), 827–834. http://doi.org/10.1016/j.jelekin.2014.10.001

Suprak, D. N., Osternig, L. R., van Donkelaar, P., & Karduna, A. R. (2006). Shoulder Joint Position Sense Improves with Elevation Angle in a Novel, Unconstrained Task. *Journal of Orthopaedic Research*, *24*(4), 559–568.

Suprak, D. N., Osternig, L. R., van Donkelaar, P., & Karduna, A. R. (2007). Shoulder joint position sense improves with external load. *Journal of Motor Behavior*, *39*(6), 517–525.

Suzuki, H., Conwit, R. a, Stashuk, D., Santarsiero, L., & Metter, E. J. (2002). Relationships between surface-detected EMG signals and motor unit activation. *Medicine and Science in Sports and Exercise*, *34*(9), 1509–1517. http://doi.org/10.1097/00005768-200209000-00018

Thompson, W., Debski, R., Boardman, D., Taskiran, E., Warner, J., Fu, F., & Woo, S. (1996). A biomechanical analysis of rotator cuff deficiency in a cadaveric model. *The American Journal of Sports Medicine*, *24*(4), 286–292.

Vadhanan, P., Tripaty, D. K., & Adinarayanan, S. (2015). Physiological and pharmacologic aspects of peripheral nerve blocks. *Journal of Anaesthesiology, Clinical Pharmacology*, *31*(3), 384–93. http://doi.org/10.4103/0970-9185.161679

Vallbo, A. B. (1970). Discharge Patterns in Human Muscle Spindle Afferents during Isometric Voluntary Contractions. *Acta Orthopaedica Scandinavica*, *80*(4), 552–566.

Vallbo, A. B. (1974). Human muscle spindle discharge during isometric voluntary contractions. Amplitude relations between spindle frequency and torque. *Acta Physiologica Scandinavica*, *90*(2), 319–336.

Vecchio, P., Kavanagh, R., Hazleman, B. L., & King, R. H. (1995). Shoulder pain in a community-based rheumatology clinic. *Rheumatology*, *34*(5), 440–442. http://doi.org/10.1093/rheumatology/34.5.440

Vuillerme, N., Teasdale, N., & Nougier, V. (2001). The effect of expertise in gymnastics on proprioceptive sensory integration in human subjects. *Neuroscience Letters*, *311*(2), 73–76. http://doi.org/10.1016/S0304-3940(01)02147-4

Wilson, L. R., Gandevia, S. C., & Burke, D. (1997). Discharge of human muscle spindle afferents innervating ankle dorsiflexors during target isometric contractions. *Journal of Physiology*, *504*(1), 221–232. http://doi.org/10.1111/j.1469-7793.1997.221bf.x

Winter, D. (2005). *Biomechanics and motor control of human movement* (3rd ed.). Hoboken, NJ: John Wiley & Sons, Inc.

Winter, J. A., Allen, T. J., & Proske, U. (2005). Muscle spindle signals combine with the sense of effort to indicate limb position. *The Journal of Physiology*, *568*(3), 1035–1046. http://doi.org/10.1113/jphysiol.2005.092619

Wu, G., van der Helm, F. C. T., (DirkJan) Veeger, H. E. J., Makhsous, M., Van Roy, P., Anglin, C., … Buchholz, B. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. *Journal of Biomechanics*, *38*(5), 981–992.

Wuelker, N., Plitz, W., Roetman, B., & Wirth, C. J. (1994). Function of the supraspinatus muscle. Abduction of the humerus studied in cadavers. *Acta Orthopaedica Scandinavica*, *65*(4), 442–6. http://doi.org/10.3109/17453679408995490

Yanagawa, T., Goodwin, C. J., Shelburne, K. B., Giphart, J. E., Torry, M. R., & Pandy, M. G. (2008). Contributions of the Individual Muscles of the Shoulder to Glenohumeral Joint Stability During Abduction. *Journal of Biomechanical Engineering*, *130*(2), 21024. http://doi.org/10.1115/1.2903422

Yang, J., & Winter, D. (1983). Electromyography reliability in maximal and submaximal isometric contractions. *Archives of Physical Medicine and Rehabilitation*.

Yasojima, T., Kizuka, T., NOGUCHI, H., Shiraki, H., MUKAI, N., & MIYANAGA, Y. (2008). Differences in EMG activity in scapular plane abduction under variable arm positions and loading conditions. *Medicine and Science in Sports and Exercise*, *40*(4), 716–21. http://doi.org/10.1249/MSS.0b013e31816073fb

Yosmaoglu, H., Guney, H., & Yuksel, I. (2013). Is there a relationship between tracking ability, joint positions sense, and functional level in patellofemoral pain syndrome? *Knee Surgery, Sports Traumatology*, *Arthroscopy*, *21*(11), 2564–2571. http://doi.org/10.1007/s00167-013-2406-2

Zbojniewicz, A. M., Maeder, M. E., Emery, K. H., & Salisbury, S. R. (2014). Rotator cuff tears in children and adolescents: Experience at a large pediatric hospital. *Pediatric Radiology*, *44*(6), 729–737. http://doi.org/10.1007/s00247-014-2875-6