UNDERSTANDING THE RESPONSE OF THE SHOULDER COMPLEX TO THE DEMANDS OF REPETITIVE WORK

UNDERSTANDING THE RESPONSE OF THE SHOULDER COMPLEX TO THE DEMANDS OF REPETITIVE WORK

By ALISON C. MCDONALD, B. Sc

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AUTHOR: Alison C. McDonald, B. Sc. Honours (University of Waterloo)

SUPERVISOR: Dr. Peter J. Keir

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ABSTRACT

Repetitive work is common in the workplace and can lead to the development of muscle fatigue. The purpose of this thesis was to improve our understanding of muscular and kinematic adaptation strategies of the shoulder complex throughout the process of fatigue and recovery. To achieve this I completed 6 studies, three studies investigating various aspects of repetitive work and fatigue and three methodological studies that were needed to interpret results. The muscular and kinematic effects of repetitive work were first investigated by incorporating a fatigue protocol between pre- and post-fatigue, simulated, repetitive work (Chapter 2). Fatigue is a complex process and how fatigue develops has been shown to influence its effects. To address this, Chapter 6 and 7 respectively, investigated the response to dynamic and static, fatiguing, repetitive work performed until participants reached termination criteria. Electromyography (EMG) was used throughout this thesis to assess muscle activity, which presented challenges because of its time consuming MVE protocols, the effects of myoelectric fatigue on its interpretation and between participants, fatigue developed in different muscles and at different rates, making comparisons between individuals challenging. For more efficient data collection, a method was developed to reduce the number of maximum voluntary exertions (MVE) required to elicit repeatable, maximum shoulder muscle activity, without eliciting muscle fatigue (Chapter 3). Methods were developed (Chapters 4 and 5) to mitigate the effects of myoelectric fatigue on EMG data and to calculate a multi-muscle fatigue score. This improved interpretation of how prolonged repetitive work impacted load sharing in the shoulder muscles and allowed the calculation of a multi-muscle fatigue score. Overall, this thesis found that the response to repetitive work is complex, multi-faceted and varies between individuals. Repetitive work impacts kinematics, muscle activity, muscle fatigue, strength, affective valence and perceived mental and physical fatigue in both static and dynamic work tasks (Chapters 2, 6, 7). Participants utilized the degrees of freedom in the shoulder complex and use coordinated compensation strategies to maintain their task performance, both following muscle fatigue (Chapter 2) and while developing muscle fatigue (Chapter 6, 7). These responses changed over time, as different muscles fatigued and recovered and were variable between individuals (Chapters 2, 6, 7). Removing fatigue artifacts from the EMG showed that muscle activity changes observed are due to load sharing between the musculature of the shoulder complex (Chapter 6, 7). Participants can adapt to the challenge of fatiguing, repetitive work and individuals will use different, coordinated strategies to maintain task performance.

Keywords: Shoulder, Fatigue, Electromyography, Repetitive Work

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THESIS FORMAT AND ORGANIZATION

This thesis contains the PhD work completed by Alison C. McDonald. This thesis has been prepared in a "sandwich" format, as outlined in the McMaster School of Graduate Studies' Guild for the Preparation of Thesis. The thesis begins with an introductory chapter, reviewing foundational literature on functional shoulder anatomy, movement of the shoulder complex and repetitive workplace tasks.

Chapter 2 uses a fatigue protocol to investigate the kinematic and muscular response of the shoulder complex to repetitive work tasks. The recovery process is evaluated over 60 minutes of post-fatigue work and compared to pre-fatigue performance. This work has been published in the Journal of Electromyography and Kinesiology. The remaining studies in this thesis were developed to address limitations from this work.

Chapters 3-5 propose and evaluate data collection and analysis methods to advance shoulder fatigue research methodology. Chapter 3 evaluates a series of maximal voluntary exertion tests to elicit maximum muscle activity in muscles of interest in this thesis, in a manner that is time efficient and does not elicit muscle fatigue. Chapter 4 proposes and evaluates a method to normalize EMG data to mitigate fatigue effects on EMG amplitude during repetitive work. Chapter 5 proposes a function to calculate a multi-muscle fatigue score, allowing for a global evaluation of shoulder muscle fatigue.

Chapters 6 and 7 utilize the methodology developed in Chapters 3-5 to evaluate the effects of repetitive work during dynamic (Chapter 6) and static (Chapter 7) tasks.

The concluding chapter of this thesis, Chapter 8, summarizes the findings from all six studies and discusses some challenges and limitations that were encountered throughout the thesis.

CONTRIBUTIONS TO PAPERS WITH MULTIPLE AUTHORS

Chapter 2

McDonald, A.C., Tse T.F.C., Keir P.J., 2016. Adaptations to isolated shoulder fatigue during simulated repetitive work. Part II: Recovery. Journal of Electromyography and Kinesiology. 29: 42-49.

Contributions

This study was conceived by Alison C. McDonald and Dr. Peter J. Keir. Method development and data collection were conducted equally by Alison McDonald and Calvin Tse, with input from Dr. Keir. Data analysis, interpretation and manuscript preparation was completed by Alison McDonald, with input from Dr. Keir and Calvin Tse. All co-authors contributed to this manuscript.

Chapter 3

Alison C. McDonald, Michael W. L. Sonne & Peter J. Keir (2017) Optimized maximum voluntary exertion protocol for normalizing shoulder muscle activity, International Biomechanics, 4:1, 9-16.

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Chapter 4

McDonald, A.C., Mulla, D.M., Stratford, P.W., Keir. P.J. Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work. Submitted June 29, 2017, to the Journal of Electromyography and Kinesiology.

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This study was conceived by Alison C. McDonald, with input from Dr. Peter Keir. Data collection was completed by Alison McDonald and Daanish M. Mulla. Methodology was developed by Alison McDonald, with input from Dr. Keir. Statistical analysis and interpretation was completed by Paul W. Stratford and Alison McDonald. The manuscript was written by Alison McDonald, with input from Daanish Mulla and Dr. Keir. All co-authors contributed to this manuscript.

Chapter 5

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Chapter 7

McDonald, A.C., Brown, D.M.Y., Bray, S.R., Keir. P.J. The multi-faceted response to static, repetitive work in the shoulder. Prepared for submission to Ergonomics.

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This study was conceived by Alison C. McDonald and Dr. Peter J. Keir. Methodology was developed by Alison C. McDonald, with major contributions from Dr. Peter Keir. Denver M.Y. Brown and Dr. Steven Bray contributed expertise and methodology for the psychological measures used in this study. Data were collected, analysed and interpreted by Alison McDonald, with input from Dr. Keir. The manuscript was written by Alison McDonald with input from Denver Brown and Dr. Keir. All co-authors contributed to this manuscript.

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CHAPTER SEVEN

LIST OF ABBREVIATIONS

ABD .... Abduction

AC .... Acromioclavicular joint

AD .... Anterior deltoid

Adel .... Anterior deltoid

AHD .... Acromiohumeral distance

ANOVA .... Analysis of variance

Bi .... Biceps brachii

CMN .... Cubic model normalizing

CMRR .... Common-mode rejection ratio

COV .... Coefficient of variation

EMG .... Electromyography

FAS .... Felt arousal scale

FLEX .... Flexion

FON .... Fatigue only normalizing

FS .... Feeling scale

FSQ .... Fatigue state questionnaire

GH .... Glenohumeral

ICC .... Interclass Correlation Coefficient

IM .... Individual muscle test

ID .... Infraspinatus

Infra .... Infraspinatus

ISB .... International Society of Biomechanics

JASA .... Joint Analysis of EMG Spectrum and Amplitude

Lats .... Latissimus dorsi

LD .... Latissimus dorsi

LMN.. Linear model normalizing

Lowtrap ....Lower trapezius

LT...Lower trapezius

Ltrap ....Lower trapezius

MAD .... Median absolute deviation

MD .... Middle deltoid

Mdel .... Middle deltoid

MDF ....Median power frequency

MF... Mental fatigue

Midtrap..Middle trapezius

MM ....Multi-muscle test

MMFS ....Multi-muscle fatigue score

MnPF..Median power frequency

MPF..Median power frequency

MT..Middle trapezius

Mtrap..Middle trapezius

MUAP ....Motor unit action potential

MVC .... Maximal voluntary contraction

MVE..Maximum voluntary exertion

PC... Clavicular head of pectoralis major

PD .... Posterior deltoid

Pdel .... Posterior deltoid

PecC .... Clavicular head of pectoralis major

PecClav .... Clavicular head of pectoralis major

PecS..Sternal head of pectoralis major

PecStern ....Sternal head of pectoralis major

PF ....Post-fatigue

PFBN.. Points forward/backward normalizing

PFN ....Points forward normalizing

PPM..Pearson product moment

PRE .... Pre-fatigue

PS ....Sternal head of pectoralis major

RM ....Repeated measures

ROF...Rating of fatigue

RPE ....Rating of perceived exertion

RPF ....Rating of perceived fatigue

RVE..Reference exertion

SA .... Serratus anterior

SAS ....Subacromial space

SC... Sternoclavicular joint

SCCS.. State Self-Control Scale

SCS ....Brief Self-Control Scale

sEMG ....Surface electromyography

Sert .... Serratus anterior

SN ....Standard normalizing

SU .... Supraspinatus

Tri..Triceps brachii

Uptrap ....Upper trapezius

US .... Ultrasound

UT ....Upper trapezius

Utrap ....Upper trapezius

VAS..Visual analogue scale

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CHAPTER ONE:
1.0 INTRODUCTION AND GENERAL LITERATURE REVIEW

The shoulder complex has six degrees of freedom, making it the most mobile group of joints in the human body. This anatomy provides a large functional range of motion for the hand through coordinated movement of the joints. The muscles of the shoulder complex have overlapping functions that allow a variety of recruitment strategies to accomplish a desired task. They function together to actively maintain stability and moments throughout the large range of motion. Although the muscles generate force throughout the range of motion, upper extremity posture affects their ability to generate forces and moments efficiently by changing moment arms, lines of action and muscle lengths. The coordinated function of the complex is relatively well understood; however, understanding how the components respond to perturbations, such as repetitive work and fatigue, can provide insight into the nature of work-related musculoskeletal disorders and their prevention.

A thorough understanding of the functional anatomy of the shoulder complex is vital for understanding shoulder mobility, function and injury mechanisms. The three bones in the complex are the clavicle, scapula and humerus (Figure 1.1). The clavicle connects the upper limb to the trunk through two joints, the sternoclavicular joint (SC) and the acromioclavicular joint (AC). The clavicle functions to increase glenohumeral (GH) range of motion and to transmit force from the upper limb to the axial skeleton (Moore & Dalley, 2006). The scapula is a flat, triangular shaped bone that provides sites for muscle attachment; scapula movement affects muscle moment arms and lines of action. The scapular spine thickens and stiffens the scapula, and extends anteriorly and laterally, to form the acromion (Halder & Itoi, 2000). Individual variation in acromion shape allows classification into three groups based on their cross-sectional morphology: flat, curved or hooked (Gill et al, 2002; Halder & Itoi, 2000). The shape of the acromion affects the subacromial space (SAS), which may be related to the development of rotator cuff pathologies (Gill et al, 2002; Gohlke et al, 1993; Halder & Itoi, 2000). The humerus is largest bone in the upper limb and articulates with the glenoid cavity of the scapula, forming the glenohumeral joint. The curved shape of the humeral head compared to the relatively flatter glenoid surface makes the GH joint non-conforming. As with acromion shape, there is individual variation in the shape of these surfaces as well (McPherson et al, 1997). The total area of the glenoid surface is approximately 28% of the humeral head surface, which allows for the large mobility of the joint, but has direct implications on joint stability (Jobe & Iannotti, 1995; Veeger & Van der Helm, 2007).

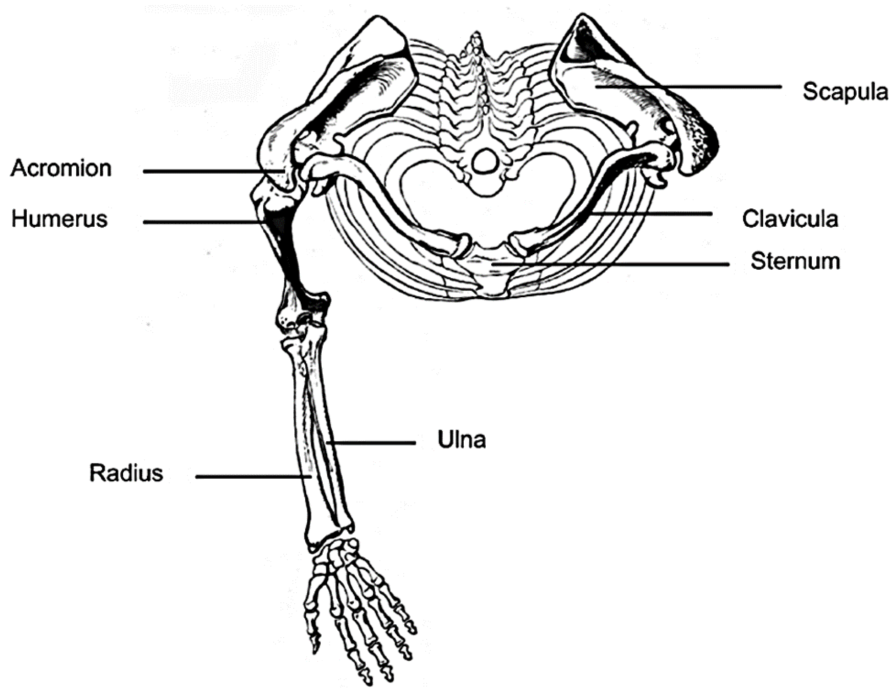

Figure 1.1: An overhead view of the bones of the shoulder complex (Image from Veeger & Van der Helm, 2007)

Shoulder instability has been defined mechanically and clinically. The mechanical definition states that an unstable joint will not return it its original position with a small perturbation, whereas, a clinically unstable joint is one where there the displacement of the humerus, in any direction, is deemed to be too large (Veeger & Van der Helm, 2007). For the purpose of this thesis, I will be using the clinical definition of shoulder stability. Maintaining shoulder stability relies on contributions from the muscles and ligaments surrounding the complex. There are two AC ligaments, four SC ligaments and six GH ligaments, which are important contributors to shoulder stability when approaching the end ranges of motion. In the mid-ranges of motion, the shoulder muscles provide active stability for the complex (Labriola et al, 2004; Warner et al, 1999). The rotator cuff muscles (supraspinatus, infraspinatus, subscapularis, teres minor) (Figure 1.2) form a half circle around the GH joint and produce smaller antagonist moments, making them effective stabilizers (Veeger & Van der Helm, 2007). These muscles typically function together throughout humeral elevation. The direction of the applied external force, the posture of the upper extremity and the plane of motion, will effect which rotator cuff muscles are most effective at maintaining joint stability (Baeyens et al, 2001; Blasier et al, 1997; Chen et al, 1999; Chopp et al, 2010; Karduna et al, 1996; Kuechle et al, 1997; Lee et al, 2000; Symeonides, 1972; Wuelker et al, 1998). All four rotator cuff muscles produce inferior shear forces to resist superior humeral translation, however, contributions are posture dependent. During abduction movements, the infraspinatus and subscapularis aid the supraspinatus in resisting humeral translation after approximately 10° of elevation has occurred (Lee et al, 2000; Wickham et al, 2010). Contributions to stability are also dependent on humeral rotation. In a anatomical humeral rotation, subscapularis and supraspinatus produce posterior shear to resist anterior translation, while infraspinatus and teres minor produce anterior shear to resist posterior translation (Blasier et al, 1997; Lee et al, 2000; Motzkin et al, 1998; Oversen & Nielsen, 1986). With external rotation, the infraspinatus and supraspinatus produce superior shear, and resist inferior translation (Lee et al 2000; Soslowsky et al, 1997). The other shoulder muscles assist the rotator cuff muscles with these stabilizing roles, however, they also generate large antagonist moments, requiring compensations to be made by the other shoulder muscles (Veeger & Van der Helm, 2007). The deltoid can create superior translation of the humerus to counteract the applied inferior load during abduction (Halder et al, 2001; Kelkar et al, 2001). The middle deltoid is effective in this role because of its line of action and large cross-sectional area (Halder et al, 2001). The deltoid also stabilizes the shoulder anteriorly when the humerus is abducted and externally rotated (Kido et al, 2003). The teres major is not an active stabilizer during movement, but it does provide support in static positions (Inman et al, 1944). The role of the biceps brachii muscle in active stabilization is dependent on posture (Blasier et al, 1997; Itoi et al, 1993; Pagnani et al, 1996). In neutral and external humeral rotations, the long head of the biceps helps to prevent posterior dislocation, but with internal rotation the biceps can contribute to posterior dislocation (Blasier et al, 1997; Pagnani et al, 1996). In all degrees of humeral rotation, the biceps also acts as an inferior stabilizer (Pagnani et al, 1996; Soslowsky et al, 1997). In flexion, as the humerus approaches its end range of motion (~120°), the role of the biceps is less apparent (Itoi et al, 1993). Along with these stabilizing roles, these muscles work together to produce moments throughout the large range of motion and different force directions afforded by the mobility of the shoulder.

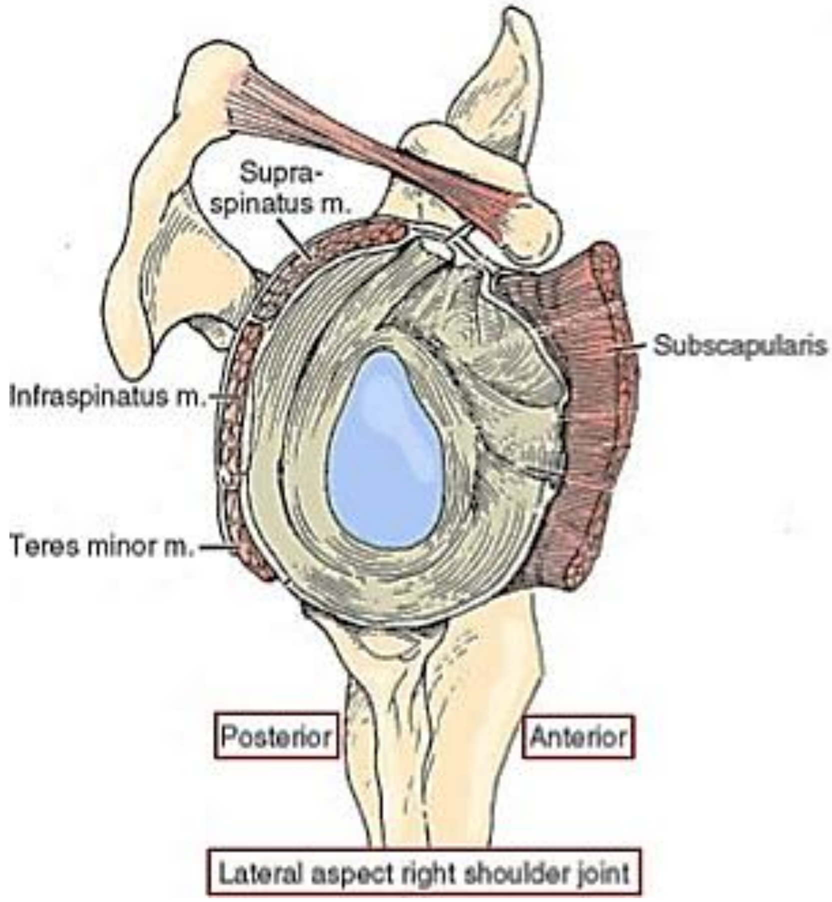

Figure 1.2: Cross section of the four rotator cuff muscles (infraspinatus, supraspinatus, teres minor, subscapularis) surrounding the glenoid cavity from a lateral aspect of the right shoulder joint (Image from Rockwood et al, 2009).

Changes in humeral posture affect the external load placed on the shoulder complex and the ability to generate moments by changing muscle moment arms (Figure 1.3), lines of action and muscle lengths (Garg et al, 2005; Mathiassen & Winkel, 1990). These effects are evident in investigations of shoulder strength that have found that posture, force direction and task constraints have significant impacts on individual's maximal moment generating capacity (Chaffin et al, 1983; Chow & Dickerson, 2009; Das & Wang, 2004; Gielo-Perczak, 2009; Gielo-Perczak et al, 2006; Grant & Habes, 1997;

Haslegrave et al, 1997; Imrhan & Ramakrishnan, 1992; Kumar, 1995; MacKinnon, 1998; Mital & Genaidy, 1989; Roman-Liu & Tokarski, 2005). Since multiple muscles perform similar functions, posture can change the set of muscles contributing to an exertion or task. The muscles used to elevate the humerus will change depending on the plane of motion and arm elevation. The deltoid has a large moment arm for humeral elevation, which makes it an efficient elevator. The portion (anterior, middle, posterior) of the deltoid most effective is largely dependent on the plane and degree of elevation (Kuechle et al, 1997). The supraspinatus moment arm is greatest in the first 30° of abduction, and peak muscle activations occur around 80° of forward flexion and 100° of abduction (Inman et al, 1944; Kuechle et al, 1997). Different portions of the pectoralis major muscle are also active during sagittal plane elevation, with peak activity between 75° and 115° of elevation (Inman et al, 1944). These large muscles can create accessory moments in unintended directions; these moments then need to be compensated for through a coordinated effort by other muscles. For example, in glenohumeral flexion, the pectoralis major creates an internal rotation moment about the vertical axis, requiring an accompanying external rotation moment (Veeger & Van der Helm, 2007). Along with their stabilizing functions, the rotator cuff muscles also have roles in movement and torque production. The infraspinatus is primarily an external rotator, but also acts as a humeral elevator for the first 30-50° of elevation, after which, it functions as a depressor. The subscapularis is primarily an internal rotator but in humeral elevation initially acts as a humeral depressor and then switches to an elevator function with greater degrees of humeral elevation (Kuechle et al, 1997). Shoulder muscle activity patterns also affect scapular motion, which has a direct impact on humeral range of motion.

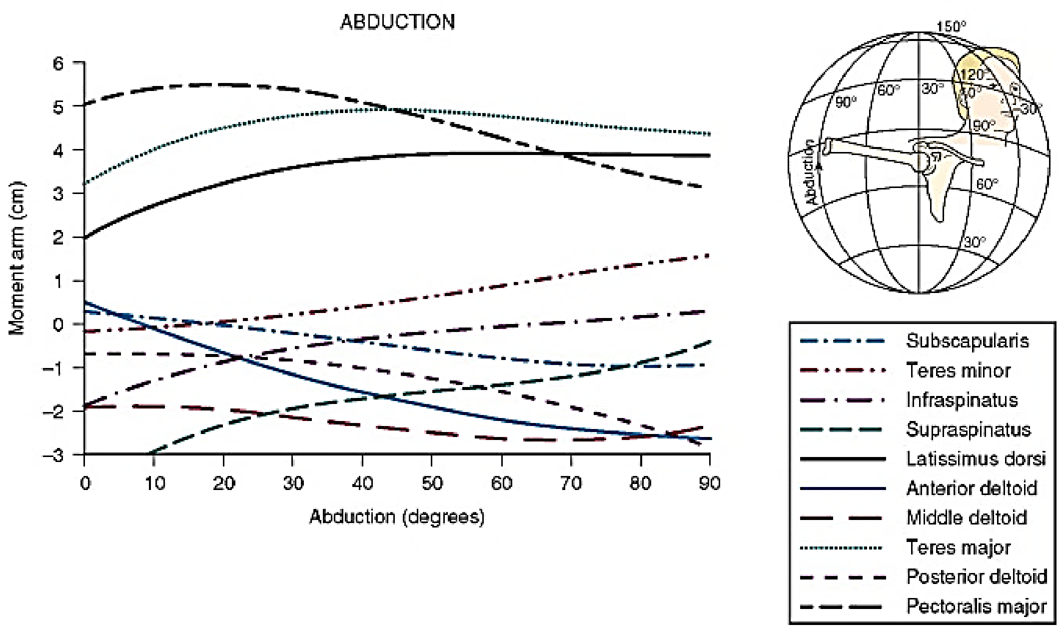

Figure 1.3: Changes in shoulder muscle moment arms with abduction will change their moment production efficiency and capability (Image adapted from Keuchle et al, 1997 in Rockwood et al, 2009)

The coordinated movement of the three shoulder joints, also referred to as the shoulder rhythm, is essential for the full range of motion (Inman et al, 1944). The absolute movement of the bones varies between individuals with different ranges of motion, but the relative relationships between the bones, with respect to each, are similar across individuals (Hogfors et al, 1991; Inman et al, 1944). For the first 30° of abduction or 60° of flexion, there is generally little scapular motion, however, depending on the resting position of the scapula, there may be some oscillations and possible downward rotation in some individuals (Borsa et al, 2003; Inman et al, 1944; Van der Helm & Pronk, 1995). After the initial phase of elevation, the movement between the bones is consistent and for every 15° of elevation approximately 10° occurs at the glenohumeral joint and 5° is from upward rotation at the scapula and thorax (Borsa et al, 2003; Inman et al, 1944). Overall, approximately 100-120° of elevation results from glenohumeral motion with the remaining 60° results from scapular motion (Inman et al, 1944; Magermans et al, 2005; Van der Helm & Pronk, 1995). The scapula moves in two translational axes and three rotational axes, and typically these movements do not occur independently (Halder & Itoi, 2000). Full scapular rotation is dependent on motion at both the SC and AC joints (Inman et al, 1944; Van der Helm & Pronk, 1995). For the first 90° of humeral elevation, there is about 4° of clavicular elevation for every 10° of humeral elevation, and above 90° there is little SC joint motion (Inman et al, 1944). There is about 20° of total motion at the AC joint, occurring within in the first 30° and after 135° of glenohumeral elevation (Inman et al, 1944). These movements of the AC and SC joints are critical for scapular rotation, which is what allows for full glenohumeral range of motion (Nakazawa et al, 2011; Teece et al, 2008). The muscle attachments on the scapula make individual shoulder rhythms sensitive to factors like muscle fatigue and changes in activation patterns (Ebaugh et al, 2006; Hogfors et al, 1991; Joshi et al, 2011).

The muscular attachments on the scapula (Figure 1.4) cause its orientation on the thorax to be influenced by changes in muscle activity (Ebaugh et al, 2005). The levator scapulae, upper trapezius and upper portions of serratus anterior cause upward rotation and elevation of the scapula during humeral elevation and at rest (Halder & Itoi, 2000; Inman et al, 1944, Moore & Dalley, 2006). The lower trapezius and lower portions of the serratus anterior have the opposite function, causing depression and downward rotation

(Halder & Itoi, 2000; Moore & Dalley, 2006). The lower and middle portions of the trapezius are more active in abduction than flexion, allowing the scapula to migrate forward during flexion which has to be compensated for by the serratus anterior (Inman et al, 1944).

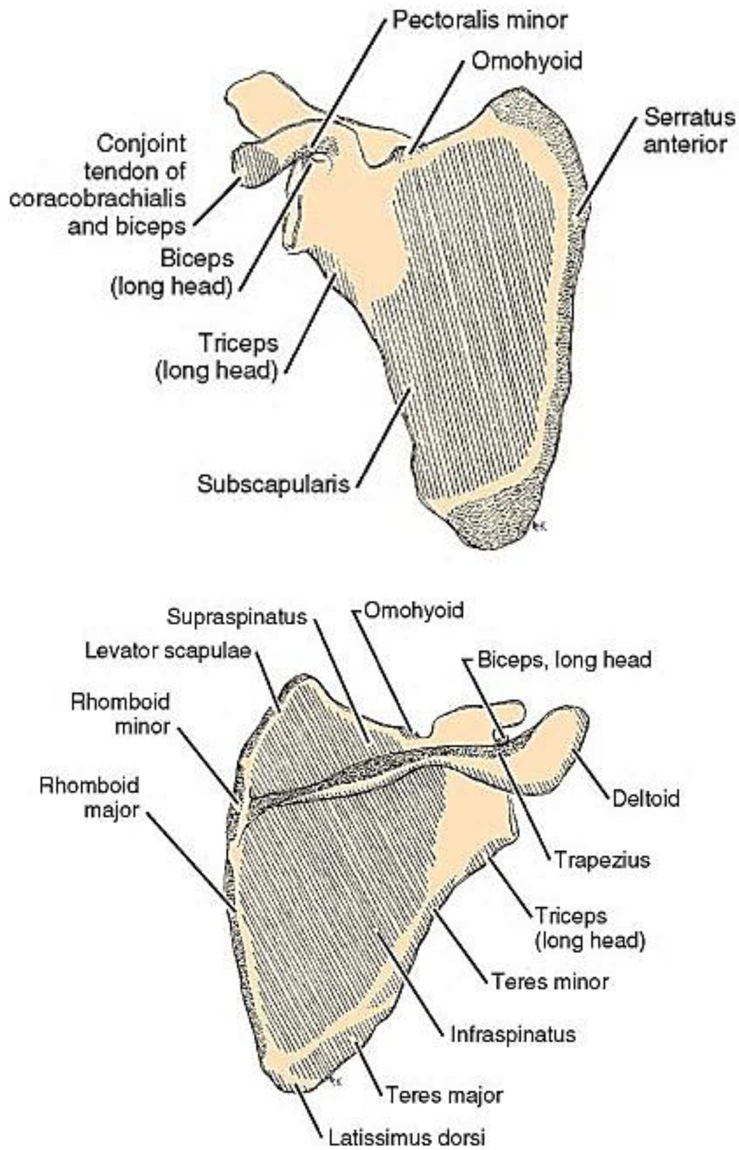

Figure 1.4: Anterior (a) and posterior (b) view of muscle origins and insertions (striped patterns) on the scapula (Image from Rockwood et al, 2009)

Scapular position is also important because of its impact on the width of the subacromial space. The subacromial space (SAS) is the space between the acromion and the humeral head. Its width is highly variable between individuals; it is affected by gender, arm position, scapular rotation, lateral acromial angle, acromial shape and muscle activity (Banas et al, 1995; Chopp & Dickerson, 2012; Graichen et al, 2001; Graichen et al, 2005). The subacromial space encompasses rotator cuff tendons and reduced subacromial space is associated with rotator cuff pathology (Banas et al, 1995). Inferior rotation, anterior tilt and protraction are scapular motions that act together to reduce the subacromial space (Banas et al, 1995; Michener et al, 2013). Humeral head position is also an important contributor to changes in the subacromial space width. Adduction causes inferior and anterior translation of the humeral head, thus increasing the subacromial space. Abduction has the opposite effect on the humerus, causing superior translation and reducing the subacromial space. Superior glenohumeral translation is greatest between 60-120° of abduction, which would impact the effects of elevated work on muscle activity and subacromial space (Graichen et al, 2005). The rotator cuff muscles are important for actively resisting superior humeral migration during humeral elevation and muscle fatigue within this group can result in subacromial space reduction (Chen et al, 1999; Chopp et al, 2010).

Repetitive work, elevated arm postures, constrained workplaces and periods of sustained muscle activity are risk factors for the development of neck and shoulder pain and disorders in the workplace (Bostad et al, 2009; Hanvold et al, 2012; Nordander et al, 2009; Ohlsson et al, 1994; Ostensvik et al, 2009; Svendsen et al, 2004). Repetitive work and overuse have been associated with damage to tendons, muscle, bone, vessels and nerves (Barr & Barbe, 2002; Fedorczyk et al, 2010; Silverstein et al, 1986). Manipulating "work" rates and force levels in a rat model has led to an exposure-response relationship to repetitive work, that is both force and work rate dependent (Barbe et al, 2008; Kietrys et al, 2012). Along with an inflammatory response, motor control changes have also been observed with a rat model, including decreased task participation and changes in movement patterns (Barbe et al, 2003; Barr et al, 2004; Kietrys et al, 2011). These adaptations could have been the result of skill acquisition, pain and/or fatigue or may be behavioral adaptations to allow for recovery. Repetitive work can impair recovery mechanisms if adequate opportunities for rest within, and between, work shifts are not provided (Elliott et al, 2008; Elliott et al, 2009).

When performed repetitively, low-level exertions can lead to the development of muscle fatigue. Muscle fatigue can be objectively measured by evaluating the electromyographic (EMG) signal as seen by increased EMG amplitude and decreased frequency content (Al-Mulla et al 2011, Clancy et al, 2005; Fuglevand et al, 1993; Hagg, Luttmann & Jager, 2000; Hagg & Suurkula, 1991). The Joint Analysis of EMG Spectrum and Amplitude (JASA) method proposed that combining the changes in the time and frequency domains can be used to differentiate between force-related and fatigue-related changes to the EMG signal (Luttrman et al 2010). During evaluation of repetitive work, it is challenging to interpret changes in EMG activity and distinguish between amplitude changes from load sharing between muscles and from fatigue.

The degrees of freedom in the shoulder complex makes multiple kinematic and muscle activation strategies possible. This may be protective against injury as it allows for load sharing, however, the shoulder complex remains a frequently injured area of the body (WSIB 2016). This large potential for variability in task execution also results in individual differences in movement strategies (Gates & Dingwell, 2008; Lucidi & Lehman, 1992; Mathiassen & Aminoff, 1997; Srinivasan & Mathiassen, 2012). These different strategies may contribute to some individuals being more susceptible to upper extremity workplace injuries, and could contribute to an explanation of why some workers become injured and others do not, when their workplace exposures are the same. Further understanding into the response of individuals to repetitive work demands will aid in the development of workplace design strategies aimed at reducing future workplace upper extremity injuries.

1.1 Thesis overview

The purpose of this thesis is to improve our understanding of muscular and kinematic adaptation strategies of the shoulder complex throughout the process of fatigue and recovery. To achieve this, I have completed 6 studies (Figure 1.5) aimed at eliciting these changes with repetitive, dynamic and static work and developing methods to evaluate these data. Specifically, the study objectives were to:

1. To evaluate how the response of the shoulder complex (kinematics, muscle activity, strength, perceived fatigue) changed over one hour of simulated, repetitive work following a fatigue protocol (Chapter 2). This study is part II of a two-part series, Part I contains complete study methods and is included in Appendix G of this thesis.
2. To compare maximum muscle activity elicited from a previously published multi-muscle test MVE protocol against a protocol of individual muscle MVE tests. Furthermore, we aimed to evaluate the reliability of these protocols, and to determine the MVE test protocol, consisting of both multi-muscle and individual muscle tests, that should be used in future shoulder investigations.
3. To evaluate the efficacy of normalizing EMG data to repeated, static, submaximal exertions to mitigate fatigue artifact, allowing EMG data to be used to evaluate load sharing between muscles over time. Five novel normalization methods were created and evaluated compared to a standard method of normalizing EMG to submaximal reference exertions.

4. To develop a function to quantify shoulder muscle fatigue using relative changes in EMG amplitude and frequency in multiple shoulder muscles over time that could be applied to the evaluation of workplace tasks.
5. To simultaneously evaluate muscular and kinematic adaptations during dynamic, fatiguing, repetitive work.
6. To evaluate the multi-faceted effects of static, repetitive work on the load sharing strategies between the muscles of the shoulder, subacromial space width and perception of fatigue and affective valence.

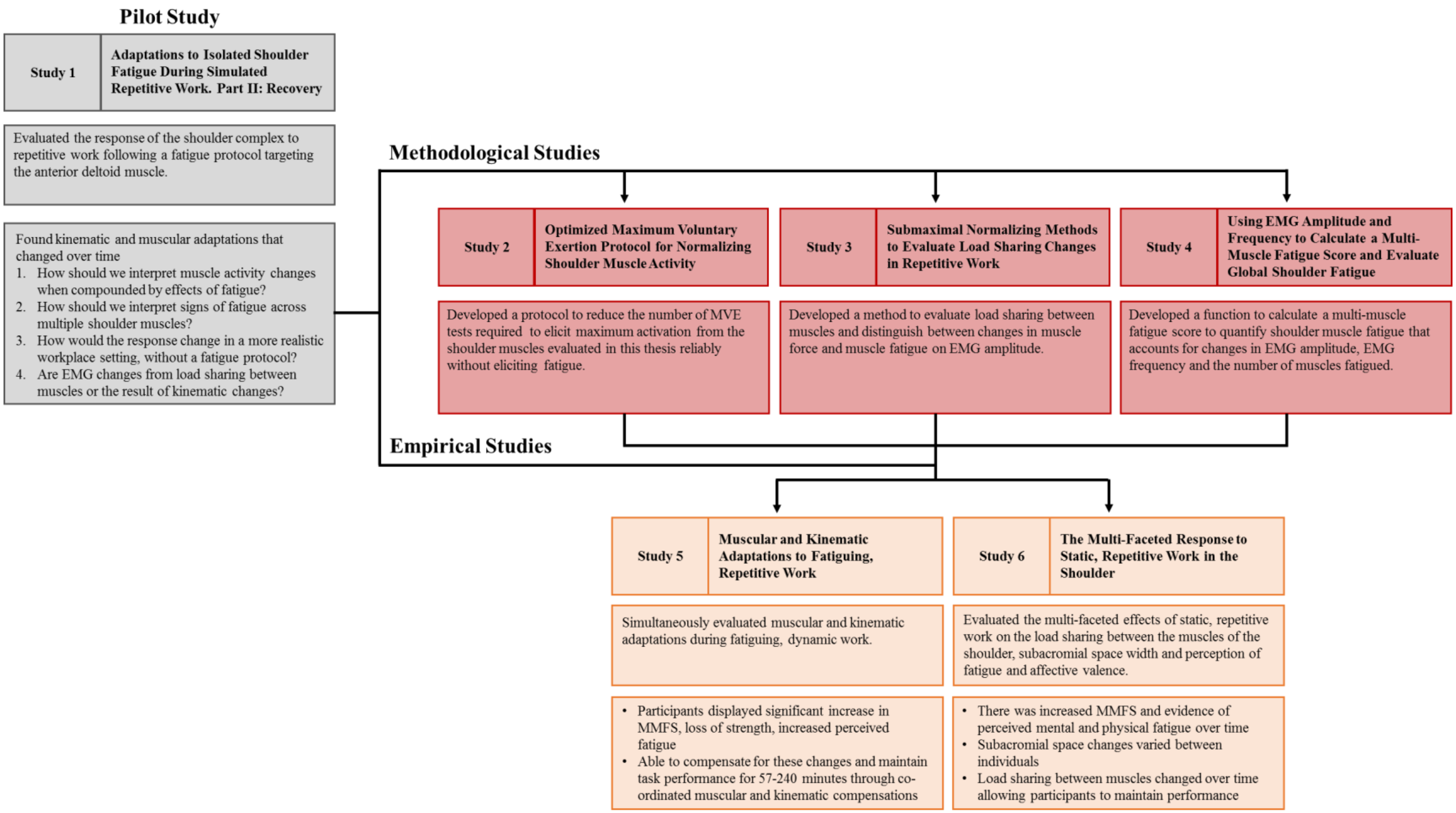

Figure 1.5: Flow diagram outlining the six studies completed in this thesis.

CHAPTER TWO

Adaptations to isolated shoulder fatigue during simulated repetitive work. Part II: Recovery

Alison C. McDonald, Calvin T.F. Tse, Peter J. Keir

Department of Kinesiology, McMaster University, Hamilton, Ontario, Canada

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2.1 Abstract

The shoulder allows kinematic and muscular changes to facilitate continued task performance during prolonged repetitive work. The purpose of this work was to examine changes during simulated repetitive work in response to a fatigue protocol. Participants performed 20 one-minute work cycles comprised of 4 shoulder centric tasks, a fatigue protocol, followed by 60 additional cycles. The fatigue protocol targeted the anterior deltoid and cycled between static and dynamic actions. EMG was collected from 14 upper extremity and back muscles and three-dimensional motion was captured during each work cycle. Participants completed post-fatigue work despite EMG manifestations of muscle fatigue, reduced flexion strength (by 28%), and increased perceived exertion (~3 times). Throughout the post-fatigue work cycles, participants maintained performance via kinematic and muscular adaptations, such as reduced glenohumeral flexion and scapular rotation which were task specific and varied throughout the hour of simulated work. By the end of 60 post-fatigue work cycles, signs of fatigue persisted in the anterior deltoid and developed in the middle deltoid, yet perceived exertion and strength returned to pre-fatigue levels. Recovery from fatigue elicits changes in muscle activity and movement patterns that may not be perceived by the worker which has important implications for injury risk.

2.2 Introduction

Industrial workplaces are often characterized by low load, repetitive and prolonged tasks. Repetitive work, elevated arm postures, constrained workplaces and periods of sustained muscle activity act in combination as risk factors for developing shoulder pain and disorders, stressing the need to understand muscular and kinematic responses to these exposures (Hanvold et al, 2012; Ferguson et al, 2013; Nordander et al, 2009; Svendsen et al, 2004). Much of the evidence for the risk of repetitive injuries in the workplace comes from cross-sectional and longitudinal epidemiological studies, making it difficult to characterize causal relationships. Understanding responses to repetitive work can become even more challenging as workers' functional capacities change with muscle fatigue.

Muscle fatigue can be defined as a combination of increased perceived effort and an eventual decline force production ability (Enoka and Stuart, 1992). Muscle fatigue can be quantified through changes in muscle activity and maximum force output (Enoka and Duchateau, 2008). Quantifying the development of fatigue and variations in fatigue during workplace tasks is difficult. Workers are typically required to generate submaximal efforts, allowing tasks to be successfully completed even in the presence of fatigue-reduced muscle capacity. With fatigue, the force generating capacity of muscle is reduced, effectively increasing the relative demands of the task. A change in capacity would impact the maximal acceptable effort for a task, which is also dependent on the duty cycle, or the percentage of time a worker is actively engaged in task, further emphasizing the effect of fatigue on workers (Potvin, 2012).

Repetitive work has been shown to impair recovery when rest opportunities within the day and between work shifts are inadequate (Elliott et al, 2008; Elliott et al, 2009). Both muscle fatigue and recovery are time-dependent processes but each proceeds at a different rate (Lucidi and Lehman, 1992; Vollestad et al, 1988), with recovery statistically modeled to occur 10-15 times slower than the fatiguing process itself (Frey-Law et al, 2012). Long recovery times for fatigued muscles is especially relevant in the shoulder given the frequent demands placed on the postural and stabilizing muscles of the shoulder complex, specifically, the rotator cuff muscles (Karduna et al, 1996; Labriola et al, 2004). In an endurance study of the trapezius muscle, individuals exhibiting greater changes in muscle activity had longer endurance times than those with more uniform activity, suggesting that variability in load distribution may allow for recovery during a sustained exertion (Farina et al, 2008). Numerous multi-joint movement strategies are possible with the large range of motion of the shoulder and additional degrees of freedom from the elbow and forearm (Culham & Peat, 1993). In the upper extremity, the presence of muscle fatigue also impacts kinematics, such as scapulohumeral rhythm, scapular motion and glenohumeral range of motion (Endo et al, 2001; McQuade et al, 1998).

In the workplace, the specific changes in both scapulothoracic and glenohumeral kinematics would be impacted by task design, as they are sensitive to the angle of humeral elevation during simple movements (Ebaugh et al, 2006a; Ebaugh et al, 2006b; Tsai et al, 2003). In more complex repetitive pointing tasks, participants maintained performance with strategies such as changing their inter-segmental movement timing and compensating for altered shoulder position by varying their elbow and wrist movements

(Fuller et al, 2009; Fuller et al, 2011). Changes in muscle activity have been assessed by variability and co-dependence between muscles and joints (Fedorwich et al, 2013). Several measures of variability in motion and muscle activity have been related to fatigue, experience, and pain; however, outcomes seem to be dependent on the specific tasks and variables measured (Fedorwich et al, 2013; Fuller et al, 2011; Madeleine et al, 2008a; Madeleine et al, 2008b; Qin et al, 2014). Although these immediate responses to fatigue protocols have been examined, how these adaptations change over time remains unknown. Understanding the response over time will give insight into the changing demands of repetitive work.

This study is the second part of an investigation examining the immediate and prolonged kinematic and muscular responses to muscle fatigue during repetitive work. In the first paper, we examined the immediate response in the first eight minutes of "fatigued" work. The purpose of this paper was to focus on how the response of the shoulder complex changed over one hour of simulated repetitive work. We hypothesized that throughout the post-fatigue period, muscular and kinematic adaptations would occur to reduce the load on the fatigued muscles. We also hypothesized that kinematics and muscle activity would return to pre-fatigue values by the end of 60 work cycles after the fatiguing protocol.

2.3. Methods

2.3.1 Participants

Twelve right-hand dominant men (20-24 years, 76.5 ± 8.5 kg, 177.9 ± 6.8 cm), free from upper limb or shoulder pathologies within the last year, participated in this study. The Hamilton Integrated Research Ethics Board approved this study and all participants provided informed written consent. Participant information including age, mass, height, umbilicus height, and acromioclavicular height were recorded.

2.3.2 Protocol

Detailed methods are described in the companion paper (Tse et al, 2016) and are included in the supplementary files (Appendix G). In brief, participants performed 20 simulated, repetitive work cycles before and 60 identical work cycles after a fatiguing protocol. Each work cycle consisted of four tasks performed on a custom apparatus: (1) handle pull (2 kg, 10 repetitions), (2) cap rotation (6 revolutions - 3 clockwise, 3 counter-clockwise), (3) drill press (50% of maximum in the anterior axis, 10-seconds), (4) handle push (2 kg, 10 repetitions). Each work cycle was repeated every 60 seconds; participants were instructed to perform the tasks at their own pace within the 60 seconds. The tasks were chosen to simulate industrial tasks and specific durations were designed to create an 80-90% duty cycle. The fatigue protocol targeted the anterior deltoid and cycled between a static hold (60 s at 45° of glenohumeral flexion) and a dynamic task (20 repetitions of glenohumeral flexion from 0° to 90°) using 25% of their maximum isometric flexion strength. Participants repeated this cycle until one of two stoppage criteria were met: (1) verbal declaration of inability to continue, (2) failure to perform either task with adequate form despite verbal encouragement (Ebaugh et al., 2006a). To quantify fatigue and recovery throughout the protocol, a maximal flexion exertion (digital force gauge, Mark-10, NY, USA) and a static, submaximal 5-second exertion were performed at four time points (baseline, pre-fatigue, post-fatigue, post 60 work cycles). Surface EMG (Trigno, Delsys, Natick, MA, USA) was used to measure muscle activity from 14 muscles (primarily on the right side except where noted): anterior, middle and posterior deltoid, biceps brachii, triceps brachii, bilateral upper and lower trapezius, infraspinatus, latissimus dorsi, sternal and clavicular heads of pectoralis major, serratus anterior. A passive motion capture system using eleven 4-megapixel resolution cameras and 30 reflective markers placed on the upper extremity and trunk were used to track three-dimensional motion during the work cycles (Cortex v4.1.1.1408 and Raptor-4 cameras, Motion Analysis Corp., Santa Rosa, CA). EMG and kinematic data were recorded continuously for each 60 second work cycle. Participants were asked for their rating of perceived exertion (RPE) every second work cycle.

2.3.3 Data Analysis

Work cycles were divided into the four constituent tasks with the handle push and handle pull tasks further divided into load and return phases. Only tasks 1, 3 and 4 were included in the analysis. Task 2 was included to increase the duty cycle and add a complex task above shoulder height; the cap rotation set-up did not include force or position data and was not analyzed. EMG data were linear enveloped (dual pass, 2nd order Butterworth filter, f_c = 4 Hz) and were normalized to maximal EMG from maximal voluntary exertions. Mean muscle activity and median absolute deviation (MAD) were calculated for each muscle in each task. Median absolute deviation (MAD) was calculated as a measure of variability in both muscle activity and joint angles (Bosch et al, 2012). Marker data were imported into Visual 3D (C-Motion, Germantown, MD, USA) and the following segments were modeled: pelvis, thorax, clavicle, scapula, humerus, forearm and hand. Local coordinate systems were computed in accordance with ISB recommendations and joint angles were calculated for each task (Wu et al, 2002; Wu et al, 2005). Joint angles were dual-pass filtered with a 2nd order Butterworth filter (f_c =10 Hz). Mean and MAD were also calculated for each joint angle for each task. There were no significant differences between the last 8 pre-fatigue work cycles that were greater than 0.5% MVE or 1°, thus they were averaged to generate one pre-fatigue value for each variable. To evaluate the changes throughout the post-fatigue work-cycles, the mean of every second set of four work cycles was computed (Figure 2.1). Using the reduced data set, one-way repeated measures ANOVAs were performed on the mean and MAD for each muscle and joint angle in each task. Preplanned comparisons were made between pre-fatigue and post-fatigue variables using Tukey's HSD tests. Effect sizes were calculated with Eta-squared (η^2) tests and are reported for significant variables. All statistical analyses used an alpha level of 0.05 (SPSS v20.0, IBM, NY, USA).

To assess changes in EMG frequency in the static reference contractions, a power spectral analysis was performed on the middle 3-second window for each muscle using a Fast Fourier Transformation and the median power frequency (MDF) was calculated (0.125 s sliding rectangular window and 0.0625s window overlap) for each muscle. The mean normalized EMG amplitude was also calculated for this central 3-second window.

Repeated measures ANOVAs were performed on EMG amplitude and MDF between the four reference exertions. Muscles were considered to be fatigued if there was a statistically significant increase in normalized EMG amplitude and decrease in median frequency. A muscle was considered to have "recovered" when it no longer exhibited these statistically significant manifestations of fatigue. Repeated measures ANOVAs were also performed on the four maximal flexion exertions to evaluate changes in strength. Preplanned comparisons were made between the pre-fatigue values and the other three time points with Tukey's HSD tests and an alpha level of 0.05.

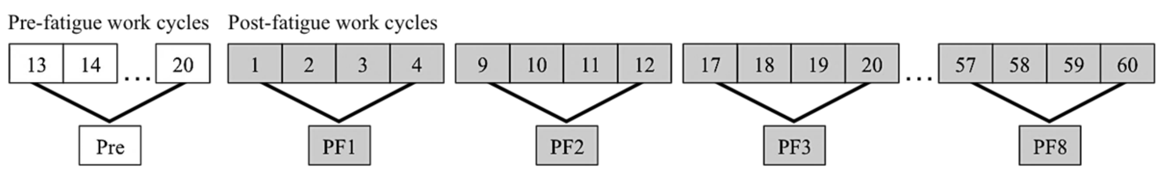

Figure 2.1: The final 8 pre-fatigue work cycles were averaged to calculate the "Pre" value for each variable in the statistical analysis. For the port-fatigue trials, every second set of 4 work cycles were averaged to calculate 8 post-fatigue values for each variable (PF1-PF8) for the statistical analysis.

2.4 Results

2.4.1 Fatigue Following Fatigue Protocol

The fatigue protocol invoked muscle fatigue in the anterior deltoid, posterior deltoid, latissimus dorsi, and serratus anterior muscles as seen by significantly decreased MDF and increased normalized amplitude ($p < 0.05$) in the reference contraction following the fatigue protocol (Table 2.1). The anterior deltoid and serratus anterior continued to exhibit significant fatigue following the 60 post-fatigue work cycles ($p < 0.05$) while the latissimus dorsi and posterior deltoid no longer exhibited signs of fatigue

($p > 0.05$). The middle deltoid exhibited muscle fatigue following the 60 post-fatigue work cycles ($p < 0.05$).

Table 2.1. Muscle fatigue was quantified as a statistically significant ($p < 0.05$) decrease in median power frequency and increase in normalized EMG amplitude during the submaximal, static reference exertions. Only muscles with significant changes are included in this table.

Time Point	Muscle	MnPF (% change)	Amplitude (% change)
1. Fatigue after pre-fatigue work	n/a	n/a	n/a
2. Fatigue with protocol	Anterior deltoid	-13.3±13.9%	46.8±60.7%
	Posterior deltoid	-10.4±15.9%	65.4±85.7%
	Latissimus dorsi	-10.1±13.1%	44.3±34.1%
	Serratus anterior	-9.8±9.2%	47.7±49.0%
3. Recovered with post-fatigue work	Posterior deltoid	-0.5±10.5%	115.2±28.6%
	Latissimus dorsi	-3.5±11.6%	32.3±32.7%
4. Fatigued after post-fatigue work	Anterior deltoid,	-13.0±13.9%	37.3±32.6%
	Middle deltoid	-9.7±10.4%	42.6±27.2%
	Serratus anterior	-5.3±8.4%	51.0±32.3%

Glenohumeral flexion strength was significantly reduced immediately following the fatigue protocol yet, by the end of the 60 post-fatigue work cycles, strength had returned to baseline (Figure 2.2). Participant ratings of perceived exertion (RPE) increased significantly immediately following the fatigue protocol (5.9 ± 2.1) ($p < 0.05$) but, by the end of the 60 post-fatigue work cycles (2.2 ± 1.6), RPE was no longer significantly different than the pre-fatigue scores (2.4 ± 1.2).

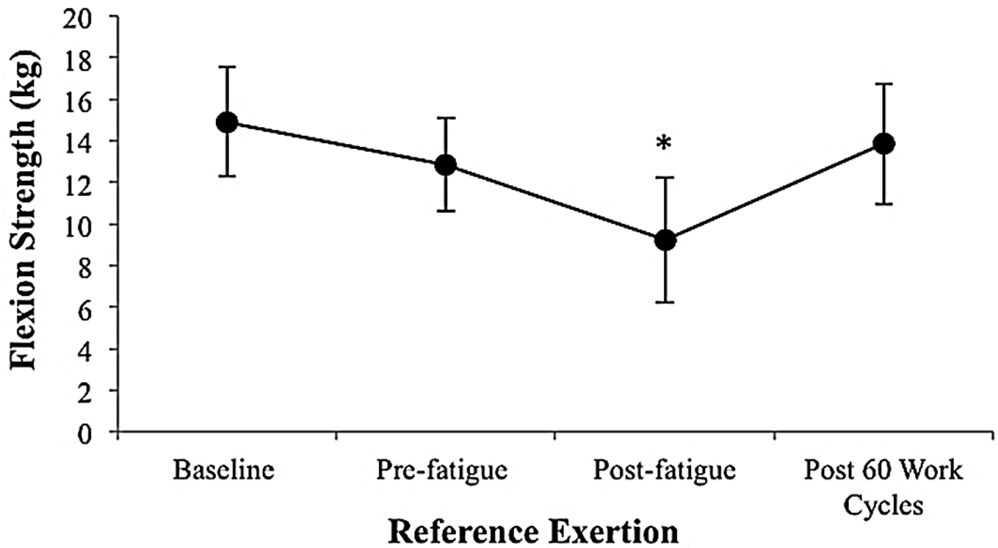

Figure 2.2: Changes in strength with fatigue were quantified with maximal, static flexion exertions at 45° glenohumeral flexion (kg). Participants had a significant reduction in strength immediately following the fatigue protocol and strength recovered by the end of the 60 post-fatigue work cycles. Statistically significant differences denoted with *.

2.4.2 Kinematic and Muscle Activity Changes During Post-Fatigue Work Cycles

Following the fatigue protocol, joint angles and normalized muscle activity changed throughout the post-fatigue work cycles. The specific joint angles and muscles affected by the fatigue protocol varied based on the task and time point. Summaries of statistically significant mean changes are presented below and magnitudes of all variables are presented in the supplementary tables (Appendix H). Values presented in text depict maximum joint angle and muscle activity changes (all means and standard deviations are found in the supplementary material with significant values presented in Tables 2.2-2.4). While the magnitude of the changes was small, the effect sizes of the significant variables ranged from medium to large for both kinematics (η^2=0.164-0.393) and muscle activity (η^2=0.118-0.785). Changes in variability (MAD) were inconsistent for both joint angles and muscle activity and did not align with mean changes, thus these data are not summarized below but are provided in the supplementary tables (Appendix H).

2.4.3 Task 1: Handle Pull

In the block of cycles directly following the fatigue protocol, significant increases in relative left trunk bend (11.2 - 13.4°) and posterior deltoid (7.8 - 8.5 %MVE) muscle activity were seen in the handle pull task (Table 2.2). As work cycles continued, posterior deltoid activity (7.82 – 10.15 %MVE) remained elevated while wrist extension (19.3 – 16.0°), elbow flexion (43.5 – 37.4°) and glenohumeral flexion (29.9 – 26.3°) decreased ($p < 0.05$). By the end of the 60 post-fatigue cycles there was reduced wrist extension (19.8 - 16.0°), elbow flexion (43.5 - 37.7°) and glenohumeral flexion (29.9 - 26.0°) ($p < 0.05$) while trunk rotation returned to its pre-fatigue angle. In the return phase of the task, no changes were seen immediately following the fatigue protocol, but, as work continued, there was reduced wrist extension (20.6 – 14.3°) and elbow flexion (44.8 – 36.3°) along with increased posterior deltoid activity (7.63 – 9.90 %MVE) ($p < 0.05$).

Table 2.2: Statistically significant changes in mean joint angle in handle pull task (Task 1) in the post fatigue (PF1-PF8) work cycles compared to pre-fatigue work cycles. Significant changes are denoted with *, rows with no * had a main effect and no significant post hoc tests. Magnitudes of the significant and non-significant changes can be found in the supplementary tables (Appendix H).

	Angle	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Pull	Wrist Extension		*	*	*	*	*	*	*
	Elbow Extension		*		*		*		*
	GH Extension					*	*		*
	Absolute Right Trunk Bend								
	Relative Left Trunk Rotation	*							
Return	Wrist Extension		*	*	*	*	*	*	*
	Elbow Extension		*	*		*	*	*	
	Absolute Right Trunk Bend								
	Muscle	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Pull	Posterior Deltoid	*	*	*	*	*	*	*	*
	Right Upper Trapezius					*	*		
Return	Posterior Deltoid		*	*	*	*	*	*	*

2.4.4 Task 3: Drill

In the block of work cycles immediately following the fatigue protocol, there were significant increases in anterior (13.1 - 17.1 %MVE) and middle (17.2 - 20.1 %MVE) deltoid activity as well as decreased glenohumeral flexion (31.5 - 23.9°) (Figure 2.3, Table 2.3). These changes are seen graphically in Figure 2.3 from their initial pre-fatigue values as they rise and fall throughout the recovery process (from the first post-fatigue block, PF1, to the last block, PF8). As work cycles progressed, increased scapular superior rotation (32.2 – 38.0°) and sternoclavicular retraction (4.4 – 8.9°) were also seen, accompanied by increased infraspinatus (10.2 – 13.4 %MVE) and posterior deltoid activity (8.0 – 10.8 % MVE). In the final work cycles, changes in the scapular and sternoclavicular angles persisted; however, there was no longer a significant reduction in glenohumeral flexion ($p < 0.05$). Along with these statistically significant changes, there were several trends in joint angle changes during this task. These joint angle changes work in concert, and vary throughout the post-fatigue work cycles to maintain hand position (Figure 2.4).

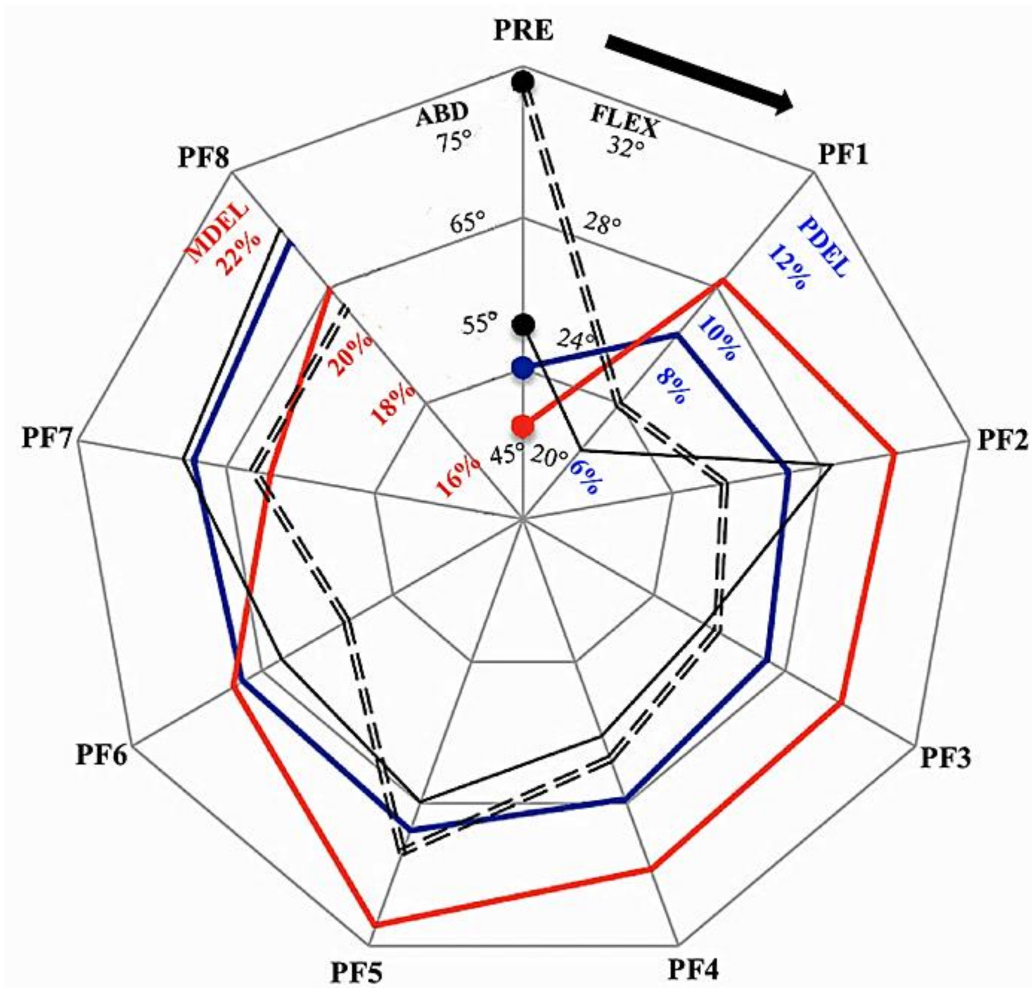

Figure 2.3: In the drill task during the post-fatigue work cycles, glenohumeral flexion (dashed black line) decreased, which coincided with increased glenohumeral abduction (solid black line), middle deltoid (red line) and posterior deltoid (blue line) muscle activity. By the end of the post-fatigue work cycles the middle deltoid exhibited significant signs of muscle fatigue ($p < 0.05$). Axes have been adjusted for each variable and work cycles are shown from pre-fatigue (PRE) to post-fatigue 8 (PF1-PF8) in a clockwise direction.

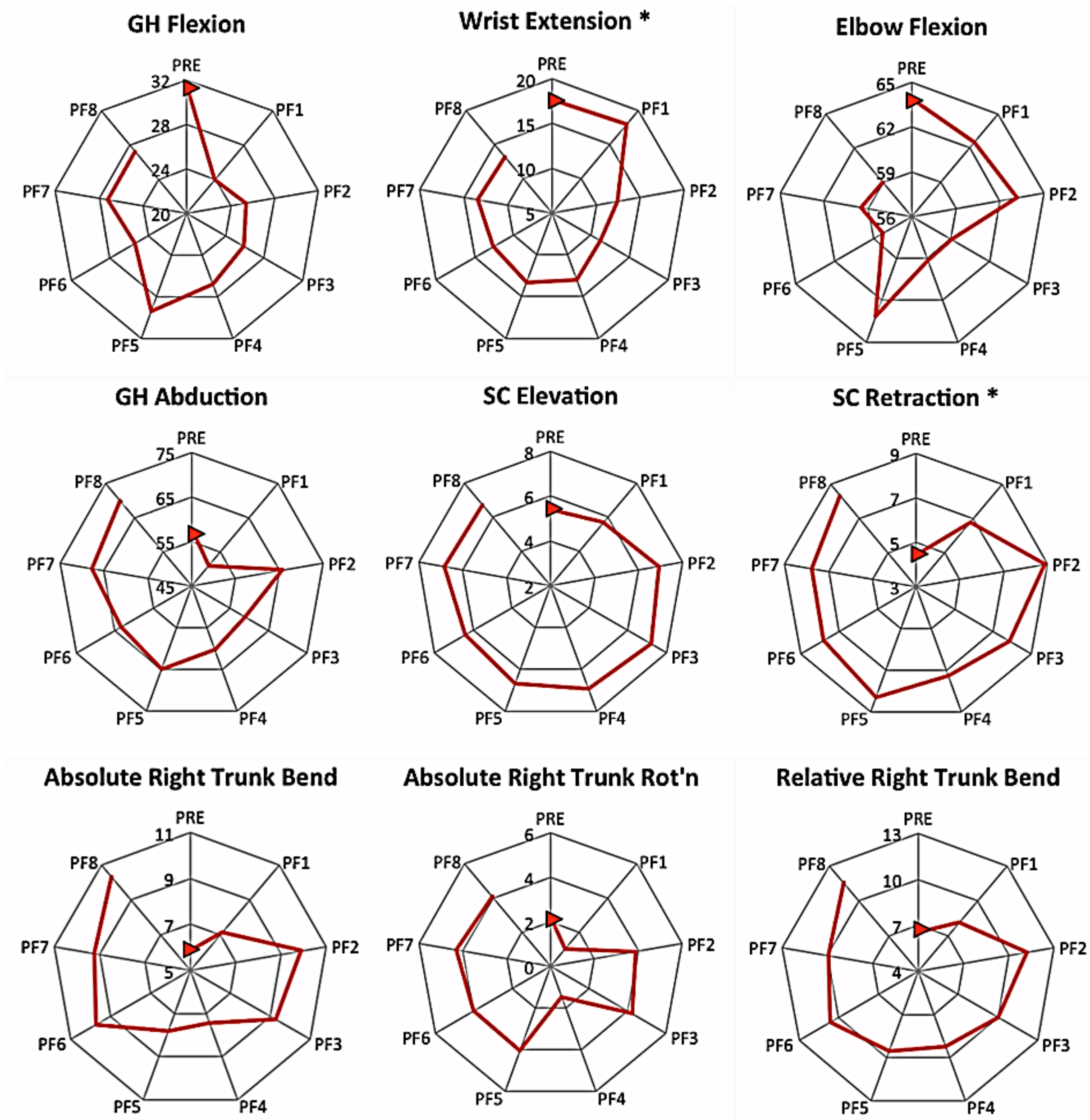

Figure 2.4: Following the fatigue protocol, in the drill task (Task 3), participants reduced their glenohumeral flexion angle to reduce the external moment and demands on the fatigued anterior deltoid muscle. To maintain task performance, they compensated for this change through changes in several other joints angles. Axes are in degrees and scales have been adjusted in each plot. Work cycles are shown from pre-fatigue (PRE) to post-fatigue 8 (PF1-PF8) in a clockwise direction.

Table 2.3: Statistically significant changes in mean joint angle in the drill task (Task 3) in the post fatigue (PF1-PF8) work cycles compared to pre-fatigue work cycles. Significant post hoc changes are denoted with *, rows without notation had a main effect but no significant post hoc tests. Magnitudes of the significant and non-significant changes can be found in the Supplementary Tables (Appendix H).

	Angle	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Drill	GH Extension	*	*	*	*		*		
	Scapular Inferior Rotation		*	*	*	*	*	*	*
	SC Protraction		*	*		*	*	*	*
	Muscle	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Drill	Anterior Deltoid	*	*	*	*	*	*	*	*
	Infraspinatus		*	*	*	*			
	Latissimus Dorsi								
	Middle Deltoid	*	*	*	*	*	*		*
	Posterior Deltoid		*	*	*	*	*	*	*
	Pectoralis Major Sternal						*	*	*
	Triceps					*			*

2.4.5 Task 4: Handle Push

Immediately following the fatigue protocol, there was a significant reduction in glenohumeral flexion (32.4 – 29.0°) with increased scapular superior rotation (28.2 – 31.1°) and increased sternoclavicular retraction (1.9 – -0.7°). These kinematic changes were accompanied with increased triceps activity (12.5 – 16.1 %MVE) (Table 2.4). As the work cycles progressed, posterior deltoid activity (8.2 – 10.0 %MVE) also increased while middle deltoid (23.6 – 19.0 %MVE) and latissimus dorsi activity (4.4 – 3.7 % MVE) decreased. By the end of the 60 work cycles, the initial kinematic changes were accompanied by increased absolute right trunk bend (-1.2 – 0.7°) and increased relative left trunk bend (0.04 – -2.0°) ($p < 0.05$).

During the return phase of this task, there was an immediate increase in the activity of the triceps (5.9 – 7.4 %MVE), middle deltoid (15.8 – 17.6 %MVE) and posterior deltoid (7.7 – 8.5 %MVE), as well as increased superior scapular rotation (28.9 – 31.7°) and decreased sternoclavicular protraction (2.2 – 0.0°). More kinematic changes developed as the work cycles progressed including reduced elbow (46.7 – 37.1°) and glenohumeral flexion (33.7 – 26.6°) and increased scapular superior rotation (28.9 – 33.5°). These kinematic changes persisted throughout the final work cycles ($p < 0.05$).

Table 2.4: Statistically significant changes in mean joint angle in handle push task (Task 4) in the post fatigue (PF1-PF8) work cycles compared to pre-fatigue work cycles. Significant changes are denoted with *, rows without notation had a main effect with no significant post hoc tests. Magnitudes of the significant and non-significant changes can be found in the Supplementary Tables (Appendix H).

Angle		PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Push	Wrist Extension			*	*		*		
	GH Extension	*	*	*	*	*	*	*	*
	Scapular Inferior Rotation	*	*	*	*	*	*	*	*
	SC Protraction	*	*	*		*	*	*	*
	Absolute Right Trunk Bend					*			*
	Relative Right Trunk Bend								*
Return	Elbow Extension		*	*	*	*	*	*	*
	GH Extension		*	*	*	*	*	*	*
	Scapular Inferior Rotation	*	*	*	*	*	*	*	*
	Scapular Internal Rotation			*	*	*	*	*	*
	SC Protraction	*	*	*	*	*	*	*	*
	Absolute Right Trunk Bend					*			*
	Relative Right Trunk Bend					*			*
Muscle		**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Push	Latissimus Dorsi				*	*	*	*	*
	Middle Deltoid			*	*	*	*	*	*
	Posterior Deltoid			*	*	*	*	*	*
	Triceps	*			*	*	*	*	
Return	Latissimus Dorsi								
	Middle Deltoid	*	*						
	Posterior Deltoid	*	*	*	*	*	*	*	*
	Triceps	*	*	*	*	*	*	*	*

2.5 Discussion

Following the fatigue protocol, all participants completed the 60 post-fatigue work cycles, despite physiological signs of muscle fatigue, reduced strength and increased ratings of perceived exertion. During the post-fatigue work cycles, joint angles and muscle activity changed over time. By the end of the protocol, strength and perceived exertion had returned to pre-fatigue levels, yet some muscles exhibited signs of fatigue. After the additional hour of work, the latissimus dorsi and posterior deltoid showed recovery, fatigue persisted in the anterior deltoid and serratus anterior, and fatigue had developed in the middle deltoid. Based on this evidence, participants were able to use the mobility of the shoulder complex and upper extremity to adapt to reduced physical capacity and allow recovery in some muscles. Despite the constrained nature of the pushing and pulling tasks, we saw significant kinematic and muscular changes during the post-fatigue work cycles. In the drill task, participants reduced their glenohumeral flexion angle, effectively reducing the demand on the fatigued muscle. Amongst other kinematic changes, participants increased glenohumeral abduction to compensate for this change. This action also corresponded with increased middle and posterior deltoid activity (Figure 2.3). The fatigue protocol induced fatigue in the anterior and posterior deltoids; as time progressed, the posterior deltoid recovered, which allowed it to compensate for reduced middle deltoid capacity. The greatest abduction occurred during the later post-fatigue work cycles (PF7-PF8), corresponding with increased posterior deltoid activity (Figure 2.3). Given that that the posterior deltoid recovered, its increased activity is likely an increased contribution rather than a fatigue induced EMG amplitude artifact. Conversely, the higher activity in the anterior deltoid can be attributed to increased EMG amplitude due to fatigue. Although the observed changes were quite small in magnitude, they were much larger than the variations seen in the pre-fatigue work cycles, and likely impose greater risk for injury.

With the reduction in glenohumeral flexion following the fatigue protocol, we observed increased glenohumeral abduction. However, abduction alone was likely not adequate to maintain the hand position required for the elevated drill task. To maintain task performance, participants appeared to employ a strategy that involved small compensations in several joint angles. These small joint angle changes acted together, suggesting a coordinated effort to maintain the required hand position. These strategies appeared to change as the work cycles progressed, suggesting that the degrees of freedom of the upper extremity and torso allow for multiple ways to perform the task successfully (Figure 2.4). Kinematic adaptations and the prioritization of maintaining task performance are supported by previous reports in which the performance of simple and multi-joint tasks was maintained in the presence of fatigue (Bosch et al, 2012; Côté et al, 2002; Forestier et al 1998, Fuller et al, 2011). Although not all of the kinematic changes in the current investigation were statistically significant, the data suggest that these small changes acted in concert to place the participant's hand in the required location to complete the task.

Glenohumeral changes were not found in isolation but were accompanied by changes in scapular kinematics as well. Scapular rotation is known to affect the subacromial space (SAS), which is important given that decreased SAS width is associated with rotator cuff disorders and likely a mechanism of shoulder injury (Banas et al, 1995). In the current investigation, changes in scapular position occurred during the drilling (Task 3) and pushing tasks (Task 4). In these tasks, we saw a combination of increased superior rotation and decreased anterior tilt and internal rotation. These rotations can increase SAS width and similar changes to scapular kinematics with rotator cuff fatigue have been found previously (Chopp et al, 2011; Endo et al, 2001; McQuade et al, 1998). These findings suggest that our participants developed an response that increased the SAS in the presence of fatigue and reduced physical capacity. Due to the importance of maintaining the SAS during repetitive work, future work should aim to investigate the dependence of these adaptations on task design and their persistence over longer durations.

We evaluated variability in both motion and muscle activity as variability has become a focus in workplace exposures, fatigue and injury risk. We found that the changes in variability (median absolute deviation) observed throughout the post-fatigue work cycles were dependent on the variable (muscle, joint angle) and the work-cycle. The usefulness of exposure variability on fatigue appears to be dependent on definitions and measurement of both variability and fatigue, as well as the specific tasks involved (Luger et al, 2014). Variability seems inherent in the mobility of the shoulder complex, which affords many kinematic and muscle strategies to complete a given task (Srinivasan & Mathiassen, 2012). Multiple control strategies create challenges when using job rotation to increase exposure variability when muscle overlap between tasks is difficult to determine (Keir et al, 2011). In simulated cutting tasks, experienced butchers had greater kinematic variability but reduced EMG variability (Madeleine et al, 2008a). Our mixed findings agree with the previous studies and support the notion that variability with fatigue and recovery will be dependent on the context of its measurement (Qin et al, 2014).

There are limitations to the current study. The multiple degrees of freedom of the upper extremity and trunk made it possible for participants to combine multiple small changes in joint angles and muscle activity to compensate for fatigue. Although we believe that these adaptations are important to understand, they are difficult to quantify with traditional statistical analyses. Another challenge in this analysis was the vast amount of data, and thus only summary variables have been presented. Future work will aim to develop analyses methods that are more sensitive to small changes and can incorporate more data to better understand the complex response to fatigue over time.

2.6 Conclusion

In the presence of muscle fatigue and reduced muscle capacity, participants were able to coordinate muscular and kinematic adaptations to maintain task performance for an hour of simulated work. These findings highlight the importance of not only examining the immediate response to fatigue but also how the response changes over time. We also found that, while individuals continued working with signs of muscle fatigue, they did not perceive fatigue, as reflected in their lowered perceived exertion scores. Adaptations to isolated muscle fatigue change over time, kinematic and muscular changes allow recovery but individuals may not perceive existing fatigue, which may contribute to overuse injuries in the workplace.

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CHAPTER THREE

Optimized maximum voluntary exertion protocol for normalizing shoulder muscle activity

Alison C. McDonald, Michael W.L. Sonne, Peter J. Keir

Department of Kinesiology, McMaster University, Hamilton, Ontario, Canada

Key words: shoulder muscles; maximum voluntary exertion; EMG; normalizing

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3.1 Abstract

Muscle activity is most often normalized to maximal activation found using isometric maximum voluntary exertions (MVE) in specific postures and direction of exertion for each muscle. This can present challenges in the shoulder complex due to the large number of muscles. The objective of this investigation was to compare maximum activity for the shoulder muscles elicited from a multi-muscle maximum voluntary exertion (MVE) test protocol versus individual muscle MVE tests and to determine the reliability of these protocols. Ten healthy males participated and muscle activity was recorded from 12 trunk, and upper extremity muscles. Participants performed 12 individual muscle, and 4 multi-muscle MVEs three times each. Peak sEMG amplitudes for each muscle were calculated from the 4 multi-muscle tests and the 12 individual muscle tests and paired sample t-tests were used to compare maximum sEMG amplitude between the two muscle test protocols. Reliability was evaluated using Interclass Correlation Coefficients. Individual muscle test maximum sEMG amplitudes differed significantly from the multi-muscle test protocol in 3 of 12 muscles ($p<0.05$). In the muscles that did not attain statistical significance, maximum amplitude differences of 6-15% were found. There was high reliability (ICC values: 0.831-0.986) and no significant differences between the second and third repetitions of the protocol. Since differences of 6-15% could have functional significance, 8 MVE tests (3 multi-muscle, 5 individual muscle) were selected for future use. Using two repetitions of this reduced test set will save time and reduce risk of pain and injury during experimental protocols.

3.2 Introduction

Surface electromyography (sEMG) is an important tool in many therapeutic and rehabilitation applications of the shoulder. Surface EMG provides non-invasive information on the amplitude and timing of muscle activity as well as muscle fatigue (De Luca 1997). Many muscle and subject specific factors affect the sEMG signal, making it essential to normalize the signal when comparing between individuals or muscles (Veiersted 1991; Mathiassen et al. 1995; De Luca 1997). Muscle activity is most often normalized to maximal activation found using isometric maximum voluntary exertions (MVE) in specific postures and direction of exertion for each muscle.

The muscles of the shoulder complex are challenging to obtain MVEs for normalizing because of their number and multiple functions. Multiple MVEs for each of the shoulder muscles can cause pain and discomfort, tissue trauma, delayed soreness, and are very time consuming during experimental protocols (Veiersted 1991; Bao et al. 1995; Mathiassen et al. 1995). Repeated maximal exertions can also lead to muscle fatigue, which can be identified through increases in sEMG amplitude and decreases in frequency. To reduce the number of MVEs required, tests to elicit maximum activation from multiple muscles simultaneously have been used. Kelly et al (1996), concluded that four of the 27 exertions they tested were necessary to maximally activate the 8 shoulder muscles examined; however, they did not compare these results to individual muscle tests. Maximum activation is more dependent on exertion direction than posture itself (Chopp et al. 2010), making this an important limitation in the application of this work. Boettcher et al (2008), also developed a protocol of four tests to elicit maximal activity from a large subset of the shoulder muscles. Although they examined many postures and exertion directions, they did not include traditional individual muscle tests in their design. Previous work examined the utility of combining multiple and single muscle tests, and although they have found this method to be successful, only a subset of shoulder muscles were examined (Chopp et al. 2010; Rota et al. 2013). Attempts to expand this four test protocol to include rhomboid major and teres major demonstrated individual muscle tests elicited greater activation levels for these muscles (Ekstrom et al. 2005). However, research to date has not provided insight into how activation levels differ between multi-muscle and individual muscle tests.

The purpose of this investigation was to compare maximum activity elicited from a previously published multi-muscle test MVE protocol (Boettcher et al. 2008) against a protocol of individual muscle MVE tests. Furthermore, we aimed to evaluate the reliability of these protocols, and to determine the MVE test protocol, consisting of both multi-muscle and individual muscle tests, that should be used in future shoulder investigations. We hypothesized that the two test protocols (individual muscle and multi-muscle) would elicit comparable maximum values and would be reliable.

3.3 Methods

Ten right-handed men participated in the laboratory study (23.6 ± 3.4 years, 179.0 ± 4.8 cm, 79.4 ± 12.6 kg), this sample size is consistent with previous literature (Yang and Winter 1983). All participants were free from upper extremity pathologies within the last year and recruited from the university population. The study was approved by the

Hamilton Integrated Research Ethics Board. Participants provided informed, written consent prior to participation, completed the protocol in a single visit and were free to withdraw from the study at any time. The protocol and analyses are outlined in Figure 3.1.

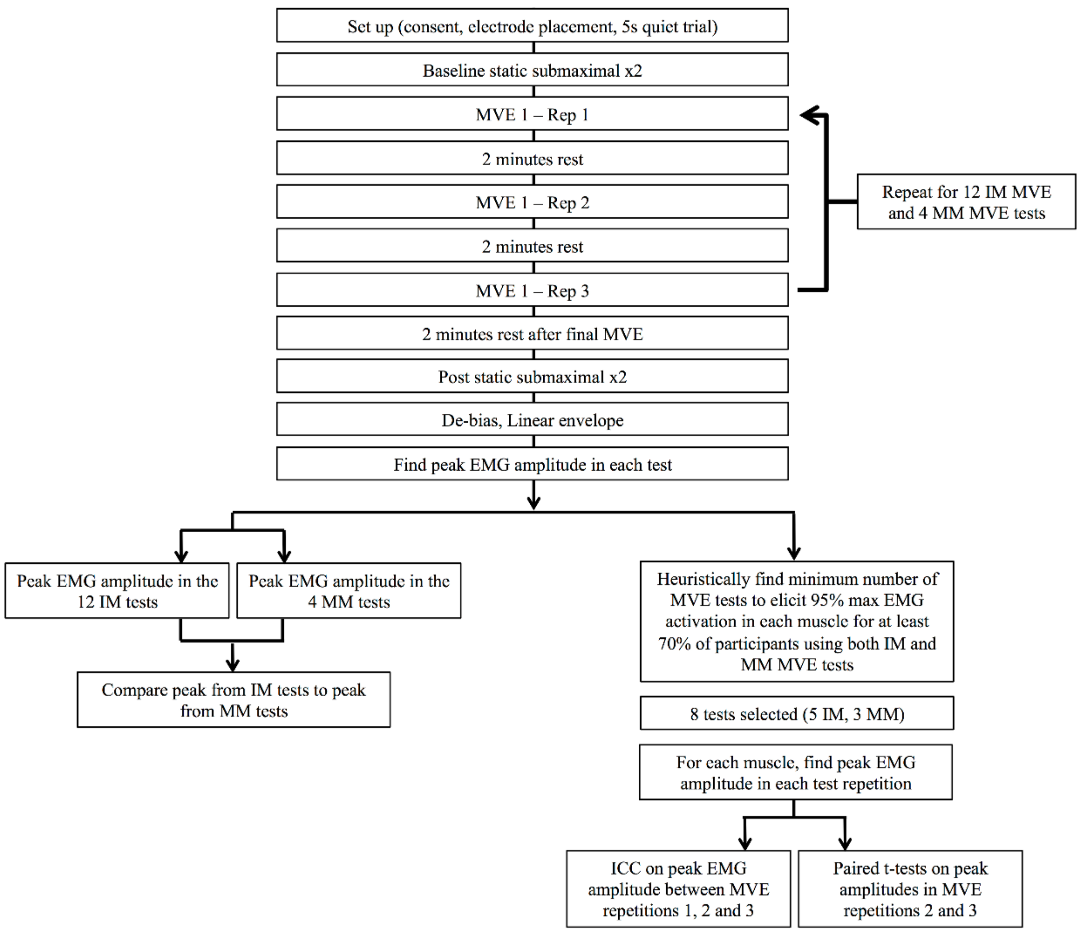

Figure 3.1: Overview of study protocol and analyses. IM=individual muscle test, MM=multi-muscle test.

Muscle activity was recorded from 12 right trunk and upper extremity muscles (anterior, middle and posterior deltoids, infraspinatus, supraspinatus, upper, middle and lower trapezius, latissimus dorsi, serratus anterior and the clavicular and sternal heads of the pectoralis major) using a wireless surface EMG system (Trigno, Delsys Inc., Natick, MA, USA). Electrodes sites were located with guidance from the literature (Ekstrom et al. 2005; Waite et al. 2010; Hodder and Keir 2013) and confirmed with manual palpation. Sites were shaved and cleansed with isopropyl alcohol prior to electrode placement. Electrodes were oriented parallel to muscle fibres. EMG signals were sampled at 1926 Hz, differentially amplified (input impedance $10^{15}\Omega$, CMRR > 80 dB), band-pass filtered (20-450 Hz), and converted with a 16-bit card with a ±5 V range.

Following electrode placement, a five-second quiet trial was collected. To evaluate muscle fatigue development in the anterior and middle deltoid muscles, participants performed two static submaximal reference exertions. Participants elevated their arm (1) to 90° in the sagittal plane, and (2) to 90° in the frontal plane, supporting only the weight of their arm. In these postures, a hand load is not required to elicit muscle activity over 15-20% MVE (Oberg et al. 1994; Antony and Keir 2010). Following the reference exertions, participants performed 12 individual muscle, and 4 multi-muscle MVEs (Table 3.1). Each MVE was repeated 3 times for a total of 48 maximal exertions. Postures were confirmed with a manual goniometer. Each 5 second MVE was followed by 2 minutes of rest between repetitions of the same MVE. No rest was given between MVEs for different muscles (eg. moving between the MVE for anterior deltoid and posterior deltoid), and order of exertions was block randomized for each participant. Two minutes following the final maximum exertion, the two submaximal reference exertions were repeated to evaluate muscle fatigue over the course of the protocol. Participants did not report any pain during the MVE protocol.

Table 3.1: Test postures and exertions for the individual muscle (IM) and the multi-muscle (MM) test protocols (Ekstrom et al. 2005; Dark et al. 2007; Boettcher et al. 2008; Waite et al. 2010; Hodder and Keir 2013). Test postures were confirmed with a manual goniometer.

Test	Posture	Exertion
Anterior deltoid (IM)	Straight arm with 45° flexion	Shoulder flexion with resistance at wrist
Middle deltoid (IM)	Straight arm with 45° abduction	Shoulder abduction with resistance at wrist
Posterior deltoid (IM)	Shoulder 90° abduction; elbow 90°	Horizontal Extension with resistance at wrist
Infraspinatus (IM)	Arm at side with 90° elbow flexion	External Rotation with resistance at wrist
Supraspinatus (IM)	Straight arm with slight abduction	Abduct
Upper trapezius (IM)	Arm abducted to 90° with neck side-bent, rotated to the opposite side and extended	Abduct
Middle trapezius (IM)	Abduct 120°, thumb pointing backward	Exert backwards/lateral scapular rotation
Lower trapezius (IM)	Arm abducted to 90°, elbow flexed to 90°	Squeeze scapula together with resistance proximal to the humerus
Latissimus Dorsi (IM)	Arm abducted 90°, elbow flexed to 90°	Shoulder adduction and extension with resistance under elbow
Serratus (IM)	Arm abducted 90°, elbow 180°	Push forward in horizontal flexion
Pectoralis Major Sternal Head (IM)	Shoulder ~90°; elbows ~90°	Bilateral palm press
Pectoralis Major Clavicular Head (IM)	Shoulder 90° flexion; elbow 90°	Horizontal (axial plane) adduction with resistance proximal to elbow
Empty Can (MM)	Arm abducted 90°, 30° cross flexion, humerus internally rotated (thumb pointing down)	Flex and abduct with resistance at wrist
125° Flexion (MM)	Arm flexed 125°, forearm pronated, elbow 180°	Flexion with resistance above elbow, pressure on inferior angle of scapula
Palm Press (MM)	Arm flexed 90°, elbow 20°, forearm semi prone	Horizontal adduction, resistance at heel of palm
Internal Rotation (MM)	Arm abducted 90°, cross flex 30°, elbow 90°	Internal rotation with residence at wrist

3.3.1 Data Analysis

Bias was removed from raw sEMG data by subtracting the mean of the quiet trial for each muscle. The sEMG data were linear enveloped with a 2nd order, 4 Hz dual-pass Butterworth filter. Single peak values for each muscle were extracted from each test and used in subsequent evaluations. To quantify fatigue development, the median power frequency (MDF) of the middle and anterior deltoid muscles was calculated from a middle 3-second window in each of the pre- and post-test submaximal trials using the raw sEMG signal. Myoelectric fatigue was defined as an 8% decrease in the MDF from the pre- to the post-test exertion (Öberg et al. 1990).

The sEMG data were initially divided into two sets for the analysis; peak amplitudes for each muscle were calculated from the 4 multi-muscle tests and from the 12 individual muscle tests. Paired sample t-tests were used to compare each muscle's maximum sEMG amplitude from the multi-muscle tests with the maximum amplitude from the individual muscle tests. Within each test set (individual muscle test set and multi-muscle test set), the maximum amplitude obtained for each muscle in any of the 3 repetitions was used, regardless of which specific MVE test or repetition elicited the value.

The specific MVE test that elicited the peak amplitude for each muscle varied between individual participants, thus based on preliminary results, a post hoc criterion was developed heuristically to select a minimum number of tests applicable to the most participants. This criterion was based on a trade off between minimizing the number of tests required and maximizing muscle activity across all of the included muscles. The criteria ensured the set of tests selected obtained 95% of each muscle's maximum amplitude for at least 70% of participants. Once the optimal set of tests were selected, reliability was evaluated using Interclass Correlation Coefficients (ICC, two-way random effects model) and the set of tests selected were analyzed collectively for their reliability. Reliability was assessed between repetitions two and three only, allowing participants to use the first repetition of the test to familiarize themselves with the required action. Paired t-tests were used to evaluate differences between the maximum activities obtained in the second and the third repetitions of each test. All statistical tests were conducted in SPSS (v20.0, IBM, NY, USA) with $\alpha=0.05$.

3.4 Results

Individual muscle (IM) specific tests elicited 29-60% greater activation for the infraspinatus, posterior deltoid, and latissimus dorsi muscles ($p < 0.05$) than the set of multi-muscle (MM) MVE tests. For the remaining 9 muscles, there were no statistically significant differences in the maximum sEMG amplitudes obtained from the two sets of tests (IM and MM) ($p > 0.05$), however, these maximum sEMG amplitudes differed by 6-15% (Table 3.2).

Table 3.2: Maximum EMG voltage for each muscle from the 12 individual muscle tests (IM) and the 4 multi-muscle tests (MM). Muscles with significantly different mean values are denoted with *. The muscles % difference values that are less than 100% had greater max values with the individual muscle (IM) tests and those that are greater than 100% had greater values from the multi-muscle (MM) tests.

Muscle	**Test**	**Mean (V)**	**SD**	**MM max as a % of IM max**
Anterior Deltoid (AD)	IM	0.215	0.095	87%
	MM	0.188	0.075	
Middle Deltoid (MD)	IM	0.129	0.087	91%
	MM	0.117	0.053	
Posterior Deltoid* (PD)	IM	0.217	0.091	71%
	MM	0.154	0.064	
Infraspinatus* (IN)	IM	0.112	0.077	70%
	MM	0.079	0.059	
Supraspinatus (SU)	IM	0.169	0.113	91%
	MM	0.154	0.105	
Upper Trapezius (UT)	IM	0.117	0.073	85%
	MM	0.100	0.067	
Middle Trapezius (MT)	IM	0.123	0.097	89%
	MM	0.110	0.085	
Lower Trapezius (LT)	IM	0.100	0.074	89%
	MM	0.090	0.072	
Latissimus Doris* (LD)	IM	0.047	0.019	40%
	MM	0.019	0.011	
Serratus Anterior (SA)	IM	0.158	0.152	106%
	MM	0.168	0.155	
Pectoralis Major Sternal (PS)	IM	0.055	0.038	108%
	MM	0.060	0.037	
Pectoralis Major Clav (PC)	IM	0.081	0.045	111%
	MM	0.090	0.065	

A set of eight tests was selected from the 16 exertions that obtained 95% of each muscle's maximum for at least 70% of the participants (Figure 3.2). These tests included the individual tests for the anterior and posterior deltoids, infraspinatus, latissimus dorsi and upper trapezius muscles and the 125° flexion, empty can and palm press multi-muscle tests (Table 3.1). The specific test that elicited the maximum amplitude for each muscle varied between participants (Figure 3.3). ICC between the three repetitions of each muscle's maximum amplitude in the set of 8 tests listed above ranged from 0.831-0.986 (excellent according to Fleiss 1986) (Table 3.3). Paired t-tests showed that there were no significant differences between the maximum amplitudes obtained in the second and third repetitions of the tests ($p > 0.05$) (Figure 3.4).

No signs of myoelectric fatigue were found in the submaximal reference contractions of anterior or middle deltoid muscles following the 48 maximum exertions (less than 8% decrease in MDF (Öberg et al. 1990)).

Table 3.3: ICC values (mean measures) and 95% confidence intervals showing the reliability between the three repetitions of each muscle's maximum amplitude in the set of 8 tests.

Muscle	ICC (Mean measures)	95% Confidence Interval	
		Lower Bound	Upper Bound
Anterior Deltoid	0.966	0.902	0.991
Middle Deltoid	0.952	0.858	0.987
Posterior Deltoid	0.979	0.939	0.994
Infraspinatus	0.983	0.95	0.995
Supraspinatus	0.986	0.959	0.996
Upper Trapezius	0.97	0.912	0.992
Middle Trapezius	0.985	0.955	0.996
Lower Trapezius	0.993	0.979	0.998
Latissimus Dorsi	0.979	0.939	0.994
Serratus Anterior	0.995	0.985	0.999
Pectoralis Major-Clavicular	0.995	0.986	0.999
Pectoralis Major-Sternal	0.936	0.814	0.983

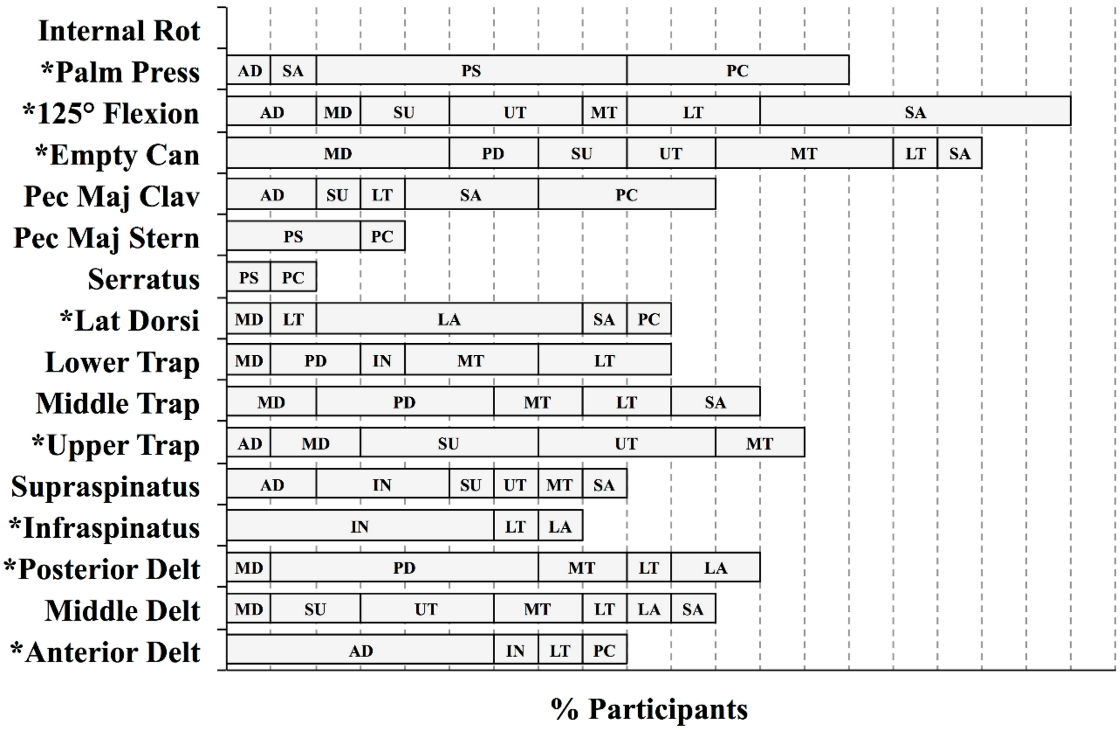

Figure 3.2: Percent of participants that achieved at least 95% of the listed muscles maximum amplitude with each test. Each vertical dashed line represents 10% of the participants. Tests marked with * are the ones selected for the recommended test protocol for future work. For example, using the palm press test, anterior deltoid was activated to greater than or equal to 95% of maximum in 10% of participants, serratus anterior in 10%, pectoralis major sternal 70% and pectoralis major clavicular in 60% of participants.

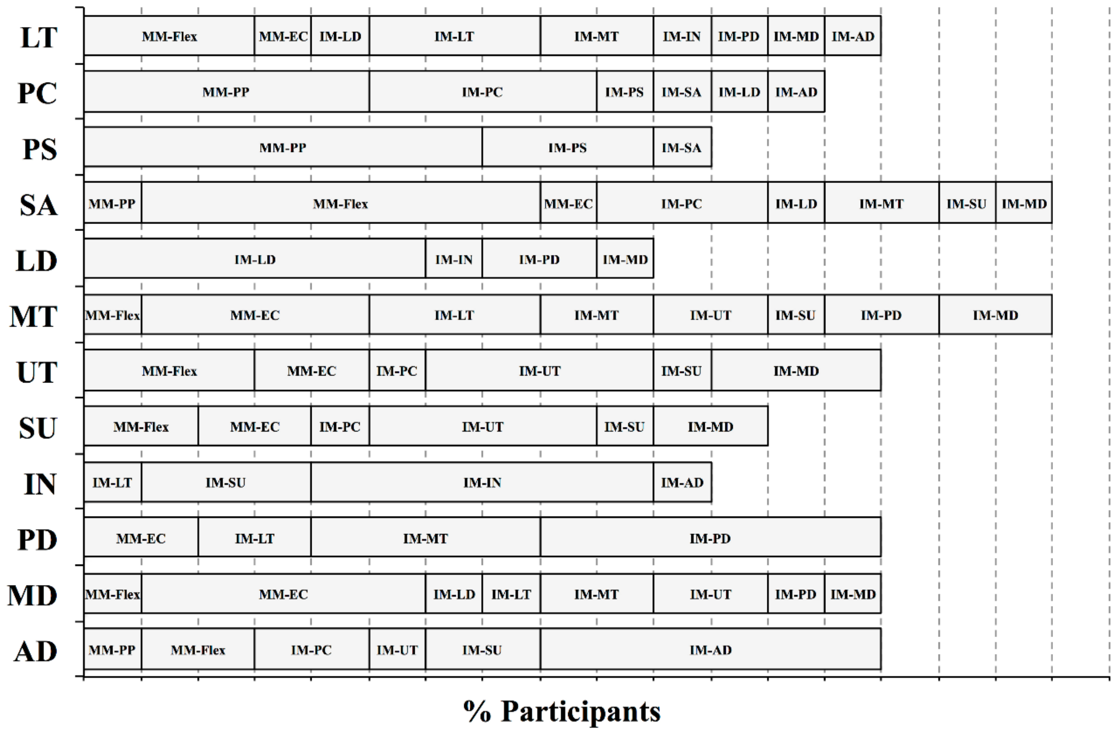

Figure 3.3: The percent of participants that had at least 95% of the listed muscles maximum amplitude with the listed tests. Each vertical dashed line represents 10% of the participants. For example, the latissimus dorsi was activated to greater than or equal to 95% of maximum activation for 60% of participants in the IM-LD test, 10% in the IM-IN test, 20% in the IM-PD test and 10% of participants in the IN-MD test.

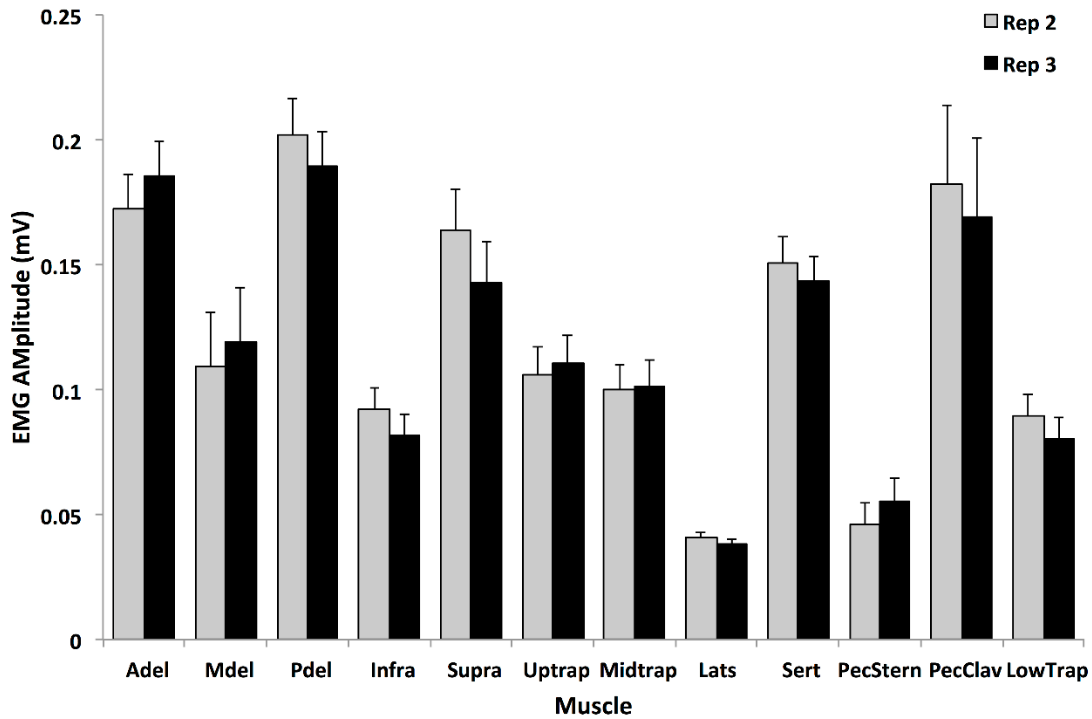

Figure 3.4: Comparison of maximum sEMG amplitude between the second and third repetitions of the recommended 8-test protocol for each muscle. Reliability of maximum sEMG amplitude (V) between the second and third repetitions. There were no statistically significant differences in the maximum values elicited between the second (grey bars) and third (black bars) repetitions of the protocol ($p > 0.05$). Error bars depict the within subject standard deviations between the two repetitions.

3.5 Discussion

The aims of this investigation were to determine if a previously published set of 4 multi-muscle MVE tests could effectively replace individual shoulder muscle MVE tests for a more time efficient and safe protocol for experimental studies, and to examine the reliability of these values between multiple repetitions of each exertion. The statistical analysis revealed few differences between the two test sets (IM and MM), however, large differences in the actual values suggest that this 4 MVE test protocol might have limitations in practical and research applications. To develop a protocol that would elicit maximum activation in all muscles in the fewest number of tests, a heuristic post hoc analysis was completed to select a set of individual and multi-muscle MVE tests. Reliability between multiple repetitions of the exertions was found to be excellent and that participants were able to complete 48 maximum exertions within one test session without developing signs of myoelectric fatigue when given 2 minutes of rest between each exertion.

Although there were few statistical differences between the two test sets (12 IM vs 4 MM tests), we found differences of 6-15% MVE between statistically non-significant values. Underestimating maximum activity by as little as these statistically non-significant levels can lead to overestimating submaximal sEMG amplitude, which, depending on the amplitude and the research question or ergonomic assessment goals, may lead to a functional difference or increased variability inherent to the normalization process (Yang and Winter 1983). For example, Jonsson (1978), recommended that static submaximal work not exceed 2-5% MVC, thus even small overestimations of muscle activations can have significant implications in the design and evaluation of return to work task assessments. Using upper extremity muscles, including a subset of the shoulder muscles, previous work found a combination of strength exercises and individual muscle tests was the best way to normalize sEMG data (Rota et al. 2013). Investigations involving the shoulder complex often require a larger number of shoulder muscles than included in Rota et al (2013); the current investigation expands this literature by recommending a set of 8 tests to normalize 12 shoulder muscles.

Selecting only one test for each muscle may be challenging due to between subject variability in shoulder muscle activation patterns (Nieminen et al. 1993), which was confirmed in our study as demonstrated by the between subject differences in which MVE tests elicited the maximum amplitude for each muscle (Figure 3.3). The multi-muscle tests elicited high sEMG amplitudes; however, because of between subject variability and large differences in each individual muscle function, these tests alone were not sufficient to elicit maximum sEMG amplitude from all muscles examined. Although 3 of the 4 multi-muscle tests evaluated (empty can, palm press, 125° flexion) were selected as part of the 8 test set, reducing the protocol to only 4 tests may prevent finding true maximum activity level, while 12 individual tests may prove too lengthy. The proposed 8 MVE tests (empty can, palm press, 125° flexion, anterior deltoid, posterior deltoid, latissimus dorsi, upper trapezius, infraspinatus) represent a trade-off between the convenience of 4 multi-muscle tests and the specificity of 12 individual muscle tests. This proposed test protocol produced higher maximum amplitudes than either of the other two test sets alone. Although the criterion used to select the 8-test set did not require maximal activation for every muscle in all participants, it performed better than the individual muscle tests and multi-muscle tests alone. Depending on the muscle, the proportion of participants that obtained their maximum with this 8 test set ranged from 50-90%. Depending on the muscle, from 30% to all of the participants reached their maximum in the individual muscle protocol while 0-70% of participants reached their maximum with the multi-muscle protocol. The combination of multi-muscle and individual muscle tests elicited greater muscle activation from all the test muscles, compared to either the multi-muscle or the individual muscle tests alone.

The reliability of the maximum amplitude from the 8 recommended tests was excellent, suggesting 2 repetitions of each test is sufficient for obtaining maximum sEMG amplitude. Previous work has shown that shoulder MVE repeatability was dependent on the muscle tested (Fischer et al. 2011). The ICCs found in this investigation ranged from 0.831 to 0.986. According to Fleiss (1986), these would be considered excellent (0 – 0.4 weak, 0.4 – 0.75 fair to good, and greater than 0.75 excellent). This shows that the recommended protocol elicited reliable maximum amplitude from all of the muscles included in the investigation with only 2 repetitions of each test. Collecting MVEs for every muscle is very time consuming in shoulder research because of the large number of muscles typically included in these investigations. By reducing the number of repetitions for each test from 3 to 2, we are able to reduce the time of the MVE portion of the data collections for 2-3 minutes/muscle tested, which can have a sizable impact in shoulder muscle assessment, rehabilitation, and research. It was confirmed with EMG amplitude and frequency, that the 48 MVE protocol did not elicit muscle fatigue when 2 minutes rest was given between each exertion. Although this protocol focused on shoulder muscles, the recommendation of 2 minutes of rest between exertions can be applied to different muscle groups.

There are limitations to the current investigation. The order of the MVE tests were randomized for each participant and it is possible that two tests targeting the same muscle could have directly followed each other. We controlled order effects by randomizing the test order between participants and confirmed that fatigue did not develop with this protocol by evaluating changes in median power frequency with submaximal exertions before and after the MVE tests. Although there were several different options for shoulder muscle MVE tests in the published literature, a set of 16 tests were selected for this investigation. Since participants were completing 48 maximum exertions with the current protocol we were not able to include and evaluate multiple tests for each muscle. Only healthy males were included in this investigation and future investigations are required to confirm that this test protocol is appropriate for female participants and clinical populations as well.

3.6 Conclusion

Two repetitions each of eight tests (empty can, palm press, 125° flexion, anterior deltoid, posterior deltoid, latissimus dorsi, upper trapezius, infraspinatus) are recommended to effectively generate repeatable, maximal muscle activation in the examined 12 shoulder muscles using sEMG. The findings also show that, with two minutes of rest between each maximum exertion participants are able to complete at least 48 maximum exertions without signs of myoelectric fatigue. This shoulder muscle normalizing protocol can be incorporated into other experimental protocols to elicit maximum sEMG amplitude in a more time efficient manner than previous protocols.

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CHAPTER FOUR

Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work

Alison C. McDonald[1], Daanish M. Mulla[1], Paul W. Stratford[2], Peter J. Keir[1]

[1] *Occupational Biomechanics Laboratory, Department of Kinesiology, McMaster University, Hamilton, ON, Canada, L8S 4K1*

[2] *School of Rehabilitation Science, Institute for Applied Health Sciences, McMaster University, Room 403, 1400 Main Street West, Hamilton, Ontario, L8S 1C7, Canada.*

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4.1 Abstract

The relationship between EMG and muscle force changes with muscle fatigue, making interpretation of load sharing between muscles over time challenging. The purpose of this investigation was to evaluate the efficacy of normalizing EMG data to repeated, static, submaximal exertions to mitigate the fatigue artifact on EMG amplitude. Participants completed simulated repetitive work tasks, in 60-second work cycles, until exhaustion and surface EMG was recorded from 11 muscles. Every 12 minutes, participants completed a series of 4 submaximal reference exertions. Reference exertion EMG data were used in 6 normalizing methods including 1 standard (normalized to initial reference exertion) and 5 novel methods: (i) *Fatigue Only,* (ii) *Linear Model, (iii) Cubic Model,* (iv) *Points Forward, and (v) Points Forward/Backward*. EMG data were normalized to each novel method and results were compared to the *Standard Method*. The significant differences between the novel methods and the *Standard Method* were dependent on the muscle and the number of time points in the analysis. Correlation analysis showed that the predicted cubic model points correlated better to the actual data points than the linear predicted values. This novel method to create “fatigue debiased” ratios may better reflect the changing muscular loads during repetitive work.

4.2 Introduction

Electromyography (EMG) can be used to reflect the internal load on muscles and is dependent on both the external load from the task, and individual internal factors, making it a valuable tool in muscle load sharing and fatigue research. The individual factors that affect EMG, such as subcutaneous tissue, recording electrode distance, and muscle fibre composition, make signal normalization imperative to compare contraction levels recorded across tasks, muscles, and individuals. Two common methods for normalization include using a maximum voluntary exertion (MVE) or a reference exertion (RVE). The MVE method, assuming a linear EMG to force relationship, is better for estimating peak values, but may overestimate sub-maximal values (Jonsson, 1982). Comparatively, a 30% reference exertion may be better for estimating sub-maximal values (up to 40-50% MVE), but can underestimate electrical activity at higher contraction levels (Jonsson, 1982). Submaximal exertions have also been found to have more repeatable EMG signals between sessions, which could make normalizing to these values a reliable between-session method (Yang & Winter, 1983). With the common occurrence of low-level exertions in the modern workplace, estimation of low-level muscular load may have a greater applicability in this setting (Das & Sengupta, 1996).

When performed repetitively, low-level exertions can lead to the development of muscle fatigue. The measurable effects of muscle fatigue on EMG have been well documented. Muscle fatigue can be observed through increased EMG amplitude and decreased frequency content (Clancy et al, 2005; Fuglevand et al, 1993; Hagg, Luttmann & Jager, 2000; Hagg & Suurkula, 1991). The Joint Analysis of EMG Spectrum and

Amplitude (JASA) method proposes using changes in amplitude and the spectral parameters to differentiate between force-related and fatigue-related changes to the EMG signal (Luttmann et al, 2000). This method has not been well operationalized and has seen limited use, but can theoretically integrate EMG amplitude and frequency changes. Signal stationarity is required for these methods, however, there are challenges when interpreting EMG parameters. With changes in load sharing between synergistic muscles, signal stationarity is lost; thus differences cannot be conclusively attributed to physiological changes (Duchêne & Goubel, 1990).

Muscle fatigue alters the relationship between EMG and muscle force (Dideriksen et al 2010), making interpretation of muscle load sharing relationships challenging. Load sharing between the shoulder muscles appears to be relatively consistent between people at low contraction intensities, slow movement speeds, and within a limited range of motion (Laursen et al, 1998). With fatigue, load sharing between forearm, lower leg, and respiratory muscles can change, and in some instances, occur in the absence of changes in motion (Bonnard et al, 1994; Duchêne & Goubel, 1990; Lucidi & Lehman, 1992; Roussos et al, 1979). Given that the anatomy of the shoulder allows multiple ways to distribute loads between muscles, load-sharing adaptations to fatigue would be expected. A fatigue protocol directed at the serratus anterior resulted in increased trapezius activity during humeral elevation (Szucs et al, 2009). Similarly, a fatigue protocol targeting the infraspinatus was found to alter trapezius activity during humeral elevation and lowering (Joshi et al, 2011). These examples of load sharing during simple movement tasks suggest that load sharing could also exist during repetitive work, which could have implications for the process of fatigue and recovery. Repeated test contractions have been used to compare mean power frequency changes in past studies where the work task was dynamic in nature (Hagg et al, 1987). The challenge in understanding how muscle load sharing changes with repetitive work and fatigue is that if myoelectric fatigue is present, then we are unable to distinguish EMG amplitude changes from load sharing between muscles or the result of fatigue (Tse et al, 2016). Changes to EMG amplitude are dependent on both the level of fatigue and how fatigue is developed, preventing the use of a simple scale factor to compensate for these time-varying effects of fatigue (Dideriksen et al 2010). The purpose of this investigation was to evaluate the efficacy of normalizing EMG data to repeated, static, submaximal exertions to mitigate the fatigue artifact, allowing EMG data to be used in musculoskeletal models. Five novel normalization methods were created and evaluated compared to a standard method of normalizing EMG to submaximal reference exertions. We hypothesized that normalizing EMG data to reference exertions taken throughout a fatigue protocol would minimize the fatigue artifact from the EMG data and that there would be differences between the 5 novel methods, allowing us to propose one method to normalize EMG data to allow muscle activity ratios to reflect load sharing changes under fatigue conditions.

4.3 Methods

4.3.1 Participants

Twenty right hand dominant men, free of upper extremity injury or pain in the last year participated in this study. The study was approved by the Hamilton Integrated Research Ethics Board (HIREB) and participants provided informed, written consent.

4.3.2 Instrumentation

Surface electromyography was recorded from 11 muscles on the right side using silver-contact wireless bipolar bar electrodes with fixed 1 cm inter-electrode spacing (Trigno, Delsys Inc., Natick, MA, USA). Muscles included: anterior deltoid (Adel), middle deltoid (Mdel), posterior deltoid (Pdel), infraspinatus (Infra), upper trapezius (Utrap), middle trapezius (Mtrap), lower trapezius (Ltrap), latissimus dorsi (Lats), serratus anterior (Sert), sternal head of pectoralis major (PecS), clavicular head of pectoralis major (PecC). Data for another investigation were simultaneously recorded on the biceps brachii, triceps brachii, and supraspinatus but were not included in this analysis. EMG signals were differentially amplified (CMRR > 80 dB, input impedance $10^{15}\Omega$), band-pass filtered (20-450 Hz) and 16-bit converted at 1926 Hz (± 5 V range).

4.3.3 Protocol

Data analyzed in this work were collected as a part of a larger study investigating muscular and kinematic adaptations to repetitive, upper extremity work. Participants performed repetitive 60-second work cycles consisting of four tasks designed to promote upper extremity fatigue. Participants were asked to continue these tasks as long as possible. Work cycle EMG and kinematic data were collected every 3 minutes. Every 12 minutes, participants were asked to complete a series of reference exertions that included one maximal exertion and four submaximal exertions (30% MVC) (Figure 4.1). This cycle (12 work cycles + reference exertions) continued until one of the termination criteria for the fatigue protocol was met: 1) participant declared they could no longer continue, 2) participant was no longer able to successfully complete the designated work tasks, 3) participant was no longer able to maintain 30% MVC for any of the 4 reference exertions. Since the purpose of this analysis was to develop an EMG normalizing method, the work cycle EMG data will only be used as data to test the proposed normalizing methods. Since it was a static task, the drill task was selected for this analysis. EMG and kinematic adaptations will be evaluated in a future publication.

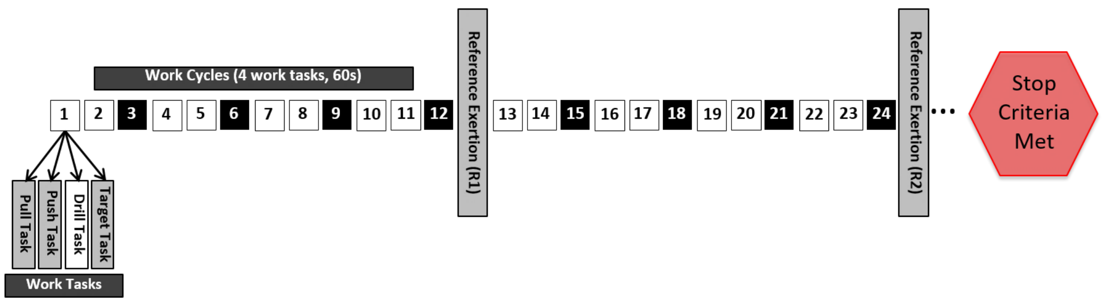

Figure 4.1: Protocol overview: The numbered black and white boxes represent the 60-second work cycles. Work cycle data were collected every 3 minutes (black boxes). Each work cycle consisted of 4 work tasks; the drill task was selected for this analysis. A series of reference exertions (submaximal and maximal) were completed every 12 work cycles. This sequence was continued until one of 3 stop criteria was met.

4.3.4 Reference Exertion Protocol

A series of four reference exertions were used to activate all of the muscles included in this study (Boettcher et al, 2008) (Table 4.1, Figure 4.2). A manual goniometer confirmed each reference posture and marks were made on the wall to help position the participants quickly between exertions. Participants performed maximal voluntary exertions in each of the four reference exertion postures. Maximum force in each posture was measured using a force transducer (Mark-10, Copiague, NY, USA) and 30% of each exertion was calculated. Every 12 minutes, participants performed 4 submaximal (30% MVC), 3-second exertions and one maximal glenohumeral flexion exertion (shoulder flexed to 90°, elbow extended), participants were given visual feedback of their force level. Once the participant met the 30% target force level, EMG was recorded for 3 seconds.

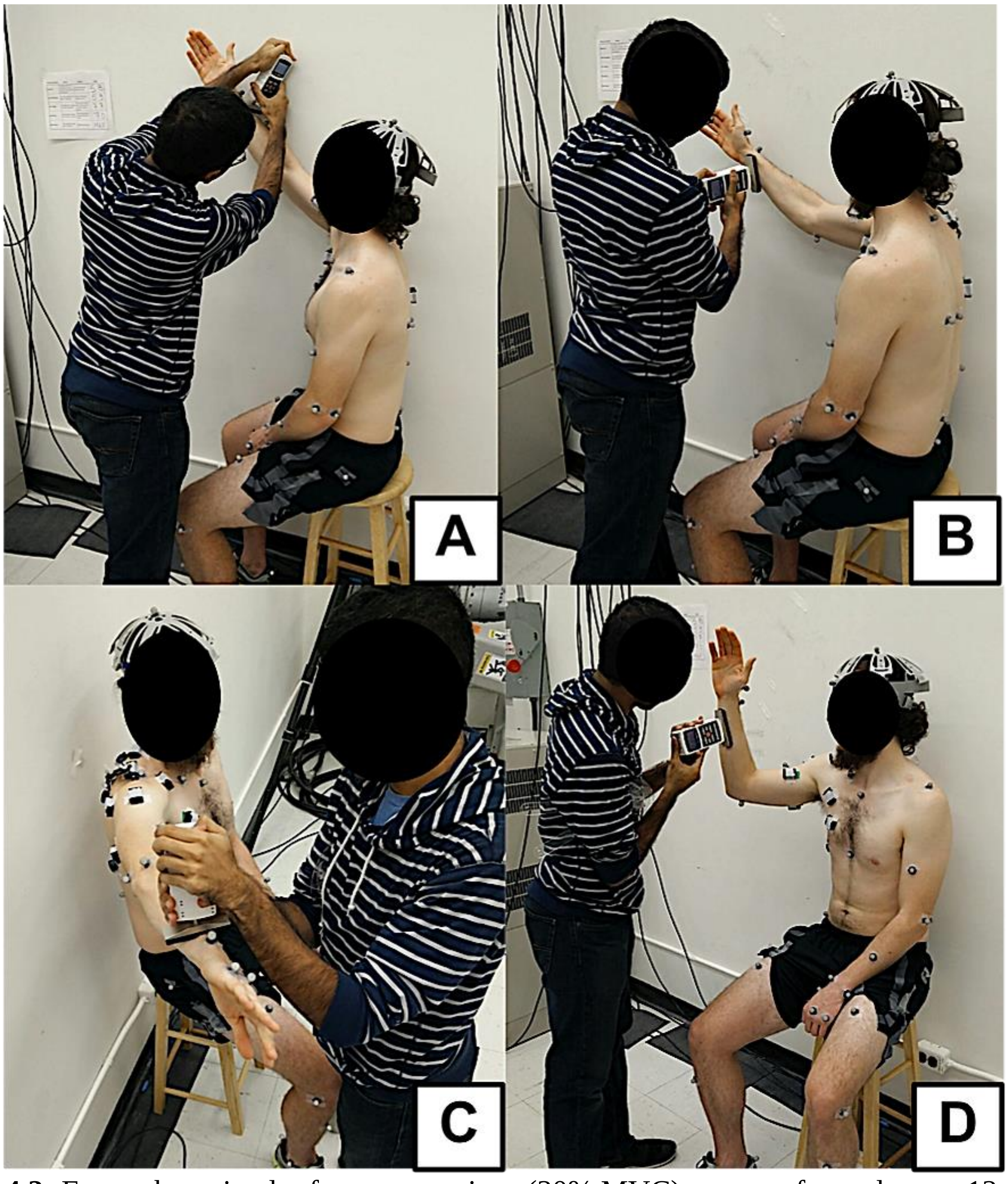

Figure 4.2: Four submaximal reference exertions (30% MVC) were performed every 12 work cycles. (A) 125° glenohumeral flexion (B) Palm press (C) Empty can (D) Internal rotation.

Table 4.1: Postures and force directions for four submaximal reference exertions that were performed at 30% MVC every 12 work cycles. Postures were selected to elicit activation from all 11 muscles of interest (Boettcher et al, 2008).

Reference Exertion	Exertion Description	Muscle Evaluated
Empty Can Test	Shoulder abducted 90° in plane of scapula, internally rotated and elbow extended. Arm abducted as resistance applied at wrist	Posterior deltoid
Internal Rotation	Shoulder abducted 90° in plane of scapula with neutral humeral rotation and elbow flexed 90°. Arm internally rotated as resistance applied at wrist	Latissimus dorsi
Flexion	Shoulder flexed 125° as resistance applied above elbow with subject sitting in an erect posture with no back support	Anterior deltoid, middle deltoid, infraspinatus, serratus anterior, upper trapezius, middle trapezius, lower trapezius
Palm Press	Shoulder flexed 90° with resistance applied at the heel of hand on force transducer and elbows flexed 20° as arms are horizontally adducted	Sternal head of pectoralis major, clavicular head of pectoralis major

4.3.5 Data Analysis

EMG from the middle second from each submaximal reference exertion was used in the analysis. Fatigue was quantified by performing a power spectral analysis using a Fast Fourier Transformation and the median power frequency (MPF) was calculated (0.125 s sliding rectangular window and 0.0625 s window overlap). Fatigue was defined as an 8% or greater decrease in MPF between reference exertions (Öberg et al, 1990). For the amplitude analysis, the EMG data from the submaximal exertions were full wave rectified and linear enveloped with a dual-pass Butterworth filter (2nd order, f_c = 4 Hz).

Submaximal EMG data were used to create 6 normalizing methods (1 standard method and 5 novel methods, Figure 4.3).

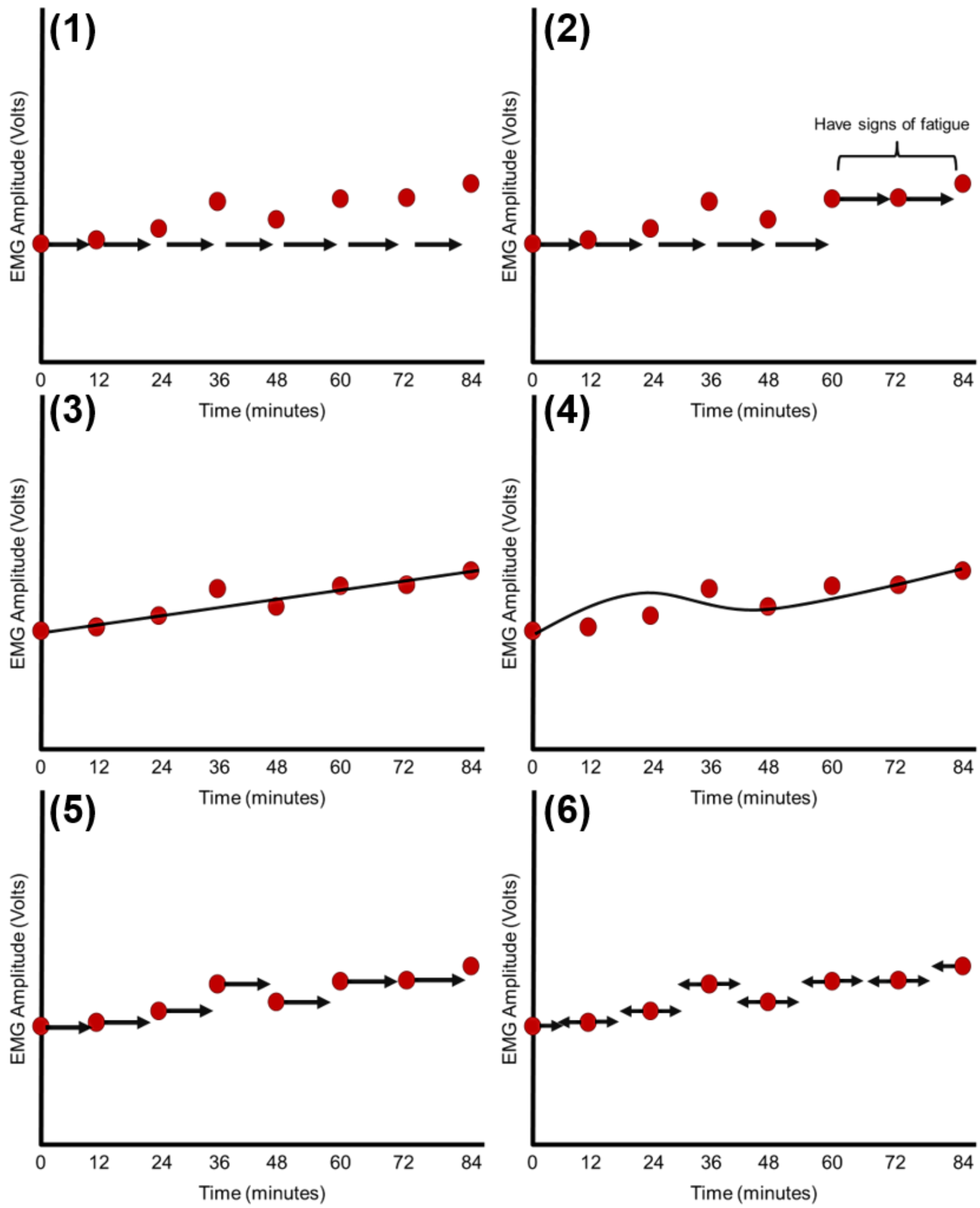

Figure 4.3: Schematic representation of the 6 normalizing methods (1) Standard Normalizing (SN), (2) Fatigue Only Normalizing (FON), (3) Linear Model Normalizing (LMN), (4) Cubic Model Normalizing (CMN), (5) Points Forward Normalizing (PFN), (6) Points Forward/Backward Normalizing (PFBN).

EMG data from the simulated drill task was used to test the 6 normalizing methods. This task was static, making it well suited for developing this method. Drill task EMG data were full wave rectified and linear enveloped using the same filter design as the submaximal exertions. The work task EMG was normalized to the first method and the mean EMG value for each work cycle was calculated for each muscle. These steps were repeated for each of the 5 novel normalizing methods.

4.3.5.1 Normalization Methods (Figure 4.3)

1) *Standard Normalizing (SN)*: All points were normalized to the mean EMG amplitude calculated from the submaximal reference exertions at baseline.
2) *Fatigue Only Normalizing (FON)*: When signs of myoelectric fatigue were present during a submaximal reference exertion for a specific muscle (≥8% decrease in MPF), all points for the next block of work tasks were normalized to the mean EMG amplitude calculated from the reference exertion. All muscles not exhibiting signs of fatigue during a submaximal reference exertion were normalized as in (1) *SN method*.
3) *Linear Model Normalizing (LMN)*: For each muscle, a least squares regression model was used to create a linear function to predict the submaximal EMG amplitude for every third minute. Work task EMG was normalized to the predicted corresponding submaximal amplitude.
4) *Cubic Model Normalizing (CMN)*: For each muscle, a least squares regression model was used to create a cubic function to predict the submaximal EMG

amplitude for every third minute. Work task EMG were normalized to the predicted corresponding submaximal amplitude.

5) *Points Forward Normalizing (PFN)*: For each muscle, points were normalized to the mean EMG amplitude from the submaximal reference exertion performed before the block of 12 work cycles.
6) *Points Forward and Backward Normalizing (PFBN)*: For each muscle, points from the first 6 work cycles were normalized to the mean EMG amplitude calculated from the submaximal reference exertion performed before the block of 12 work cycles and the last 6 work tasks were normalized to the mean EMG amplitude calculated from the submaximal reference exertion performed after the block of 12 work cycles.

4.3.6 Statistical Analysis

Mixed-effects modelling was used to identify differences in mean EMG amplitude between participants and between the normalizing methods; data points were removed if not physiologically possible. To test for between participant differences in mean EMG amplitude, a random intercept model was compared to a random intercept, random slope model, with both models including only the main effect of time. Since not all participants performed the same number of work cycles, a spline analysis was performed to divide the work cycles into five equal sample-size time intervals: work cycles 3-24, 27-45, 48-72, 75-114, 117-240. Significant differences in the fit of the models (random intercept vs. random intercept, random slope) were assessed using the chi-squared test statistic obtained from a likelihood-ratio test. The analysis was performed for each of the 11 muscles and 6 normalizing methods separately, with all models using an independent covariance structure.

A random intercept, random slope model including time, normalizing method, and the two-way interaction between time and method was used to test for differences in mean EMG amplitude between normalizing methods. Since participants completed different numbers of work cycles, two models were used to analyze two sets of time points: (a) the first 72 work tasks only, allowing 75% of participants to have complete data sets; (b) all time points, where only 2 participants had complete data sets. In both cases, time was modelled as a categorical variable as the study employed a fixed-occasion design (i.e., every third work cycle collected). Normalizing method was coded as a dummy variable. The analysis was performed for each of the 11 muscles, with all models using an independent covariance structure. Significant effects of method were further analyzed post-hoc using Sidak's adjustment for multiple-comparisons, with all novel methods compared to only the (1) *Standard Normalizing Method*.

To compare the linear and cubic model predictions, pearson product moment (PPM) correlations were used to assess how well the predicted points from the linear and cubic normalization models fit the reference exertion data points. Paired samples t-tests were used to evaluate changes in glenohumeral flexion strength between the baseline and the final reference exertions. Statistical tests were conducted in SPSS Statistics (v20.0, IBM, NY, USA) and Stata (v13.1, TX, USA) with $\alpha = 0.05$.

4.4 Results

Participants performed the work cycles for 57 to 240 minutes and completed 5 to 20 sets of reference exertions before meeting one of the termination criteria. The muscles that displayed signs of myoelectric fatigue and the time points where fatigue differed between participants (Appendix, Table AI.1). Mean strength during the maximal reference exertion significantly decreased from 108.8±19.2 N to 79.7±28.6 N ($p<0.05$) by the end of the protocol. Significant differences existed between participants and between the methods ($p<0.05$); the specific details are outlined below.

4.4.1 Between Participant Differences

Across the five novel normalization methods (FON, LMN, CMN, PFN, PFBN), there were significant between participant differences in mean EMG amplitude across all the muscles (Table 4.2, Figure 4.4). However, the between participant differences, as estimated by the variances in the slopes of these models, were very small and ranged from 8.6×10^{-5} to 2.89×10^{-8}.

Method	Adel		Mdel		Pdel		Infra		Utrap		Mtrap		Ltrap		Sert		Lats		PecC		PecS	
	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2	Var (time)	LR chi2
(1) SN	2.47E-06	131.2	6.95E-07	179.4	6.3E-06	1192.3	2.27E-06	191.0	1.87E-07	40.9	2.73E-06	151.6	1.27E-06	241.3	1.17E-06	317.1	8.55E-05	87.9	4.15E-06	507.5	1.78E-05	187.8
(2) FON	8.26E-06	318.6	1.72E-06	99.2	4.96E-06	75.2	1.39E-06	122.4	2.89E-08	4.2	2.96E-06	164.1	7.83E-07	29.1	9.25E-07	174.6	0.000057	69.9	4.49E-06	454.6	2.72E-05	65.1
(3) LMN	5.21E-06	360.4	4.19E-07	146.5	1.4E-06	279.8	1.64E-06	139.7	4.64E-08	33.8	9.05E-07	140.0	9.36E-07	185.4	7.86E-07	267.2	6.04E-05	85.1	3.07E-06	579.6	1.64E-05	196.4
(4) CMN	6.52E-06	475.2	6.77E-07	218.0	2.16E-06	420.1	2.24E-06	157.2	5.35E-07	232.2	9.38E-07	103.7	7.7E-06	600.2	1.1E-06	297.3	5.78E-05	109.7	3.62E-06	555.5	3.61E-05	439.4
(5) PFN	5.07E-06	166.5	8E-07	37.5	4.71E-06	226.7	8.47E-07	41.0	2.54E-07	26.2	1.29E-06	159.8	7.1E-07	13.1	7.92E-07	90.18	0.000041	47.3	3.15E-05	497.1	3.95E-05	87.1
(6) PFBN	4.13E-06	117.2	8E-07	42.3	3.65E-06	181.5	1.06E-06	56.4	8.36E-08	19.1	1.17E-06	148.9	2.29E-06	31.9	5.38E-07	68.0	3.34E-05	42.8	3.03E-05	460.3	4.08E-05	84.7

Table 4.2: Between subject differences across all muscles and normalization methods, as indicated by a significant chi-squared (LR chi2) test statistic obtained from the likelihood-ratio test when comparing a random intercept model versus a random intercept, random slope model ($p<0.05$). Although statistically significant, the differences between subjects, as estimated by variances (var) in the slope (i.e., rate of change in mean EMG amplitude over time), were very small ($8.6x10^{-5}$ to $2.89x10^{-8}$).

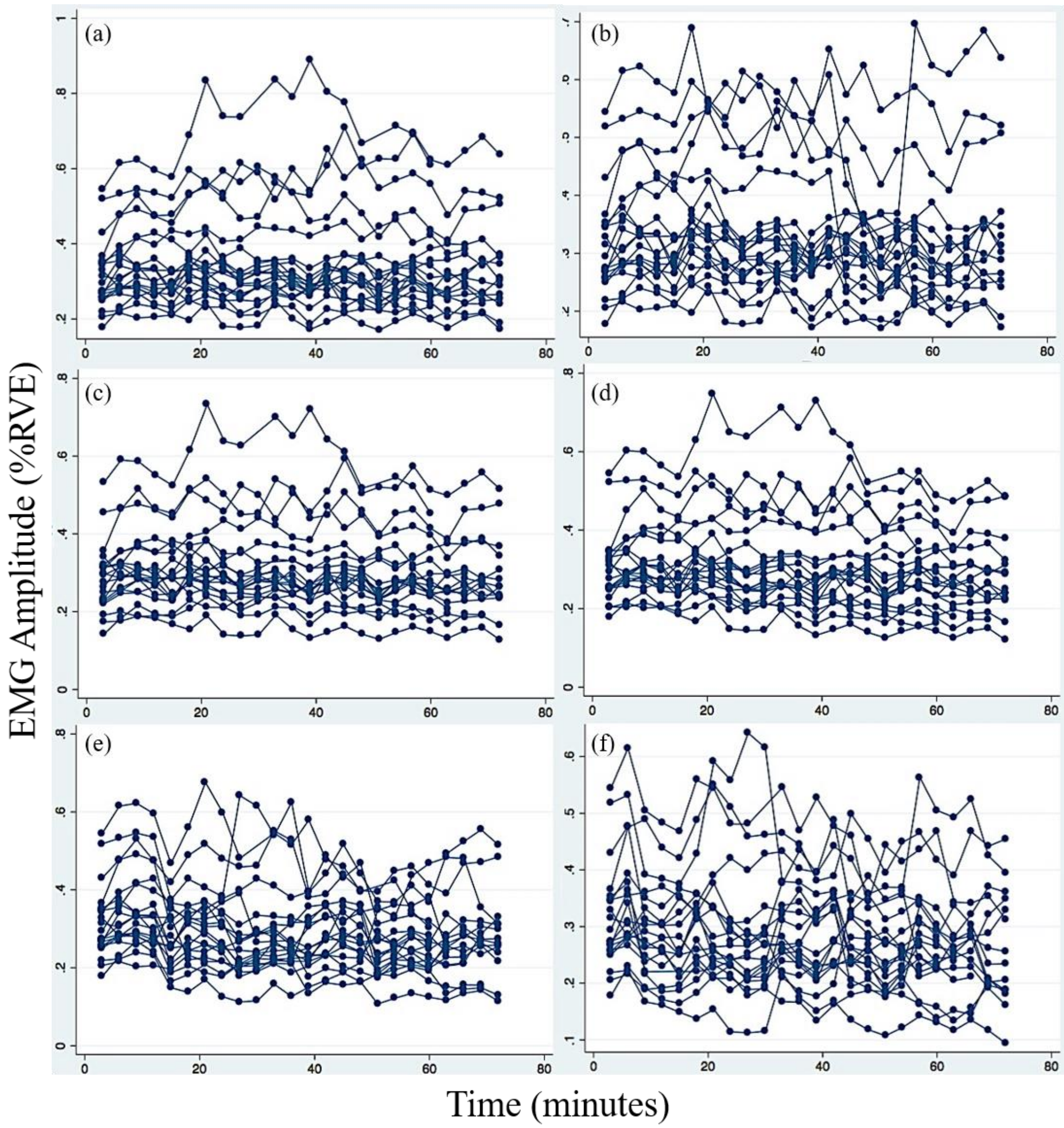

Figure 4.4: Between subject differences in the relationship between EMG values and time across all 6 methods in the middle deltoid. Although differences were statistically significant between people, the slopes of these lines were very small. (a) Standard Normalizing (SN), (b) Fatigue Only Normalizing (FON), (c) Linear Model Normalizing (LMN), (d) Cubic Model Normalizing (CMN), (e) Points Forward Normalizing (PFN), (f) Points Forward/Backward Normalizing (PFBN).

4.4.2 Between Methods Differences

The significant differences in mean EMG amplitude between the novel methods and the (1) *SN Method* were dependent on the muscle and the number of time points in the analysis (Table 4.3, Figure 4.5). Significant differences from *SN* method were the consistent between the *FON (2), LMN (3), and CMN (4)* methods. For these 3 methods, there were mean EMG amplitude differences in the posterior deltoid muscle for both time point analyses (all time points and first 72 time points only). There were differences in the middle deltoid that were only present in the model including all the time points.

Table 4.3: Chi-squared values for post-hoc tests to assess differences between normalizing methods. All methods were compared to the standard normalizing method. Significant differences ($p<0.05$) were dependent on muscle and on the number of time points included in the analysis and are denoted with *.

Musc.	FON vs Standard		LMN vs Standard		CMN vs Standard		PFN vs Standard		PFBN vs Standard	
	72 time points	All time points	72 time points	All time points	72 time points	All time points	72 time points	All time points	72 time points	All time points
Adel	29.24	8.48	24.03	6.37	21.88	5.04	*57.97	*17.31	*44.48	19.8
Mdel	22.33	*22.47	6.75	*17.21	20.78	*24.29	*76.5	*58.21	*50.51	57.72
Pdel	*57.39	*149.39	*52.64	*118.2	*72.45	*130.72	*263.15	*299.56	*236.74	302.07
Infra	6.54	1.77	4.00	2.21	1.32	1.29	14.8	10.24	17.22	14.25
Utrap	2.97	0.18	1.32	1.75	4.65	2.59	11.95	*16.87	10.56	12.9
Mtrap	1.08	0.02	7.16	1.10	13.19	2.73	32.78	4.87	24.07	3.19
Ltrap	8.42	3.48	0.62	1.54	15.01	2.28	22.63	6.90	20.16	8.95
Lats	2.08	0.67	0.78	0.86	3.56	2.89	6.17	2.61	4.11	3.06
Sert	6.46	1.35	1.98	2.18	5.98	2.89	17.15	*17.81	13.11	8.64
PecC	1.26	0.27	0.40	3.84	3.52	7.39	19.66	*20.35	30.03	12.69
PecS	32.07	6.82	0.33	0.16	4.16	1.88	*44.72	12.59	*42.51	7.93

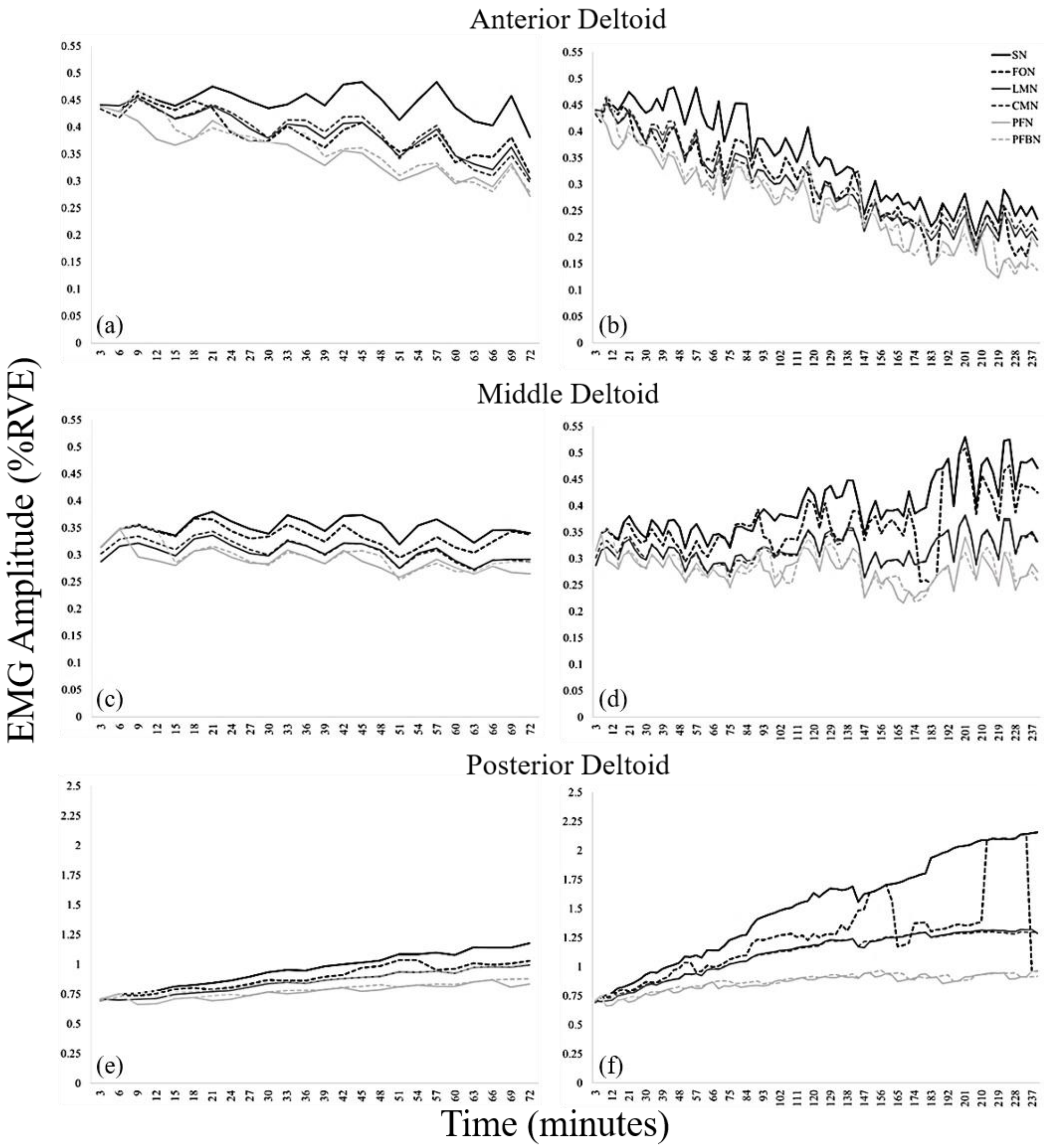

Figure 4.5: Adjusted predictions of 6 normalization methods for the three heads of the deltoid muscle (a) anterior deltoid (72 time points), (b) anterior deltoid (all time points), (c) middle deltoid (72 time points) (d), middle deltoid (all time points) (e), posterior deltoid (72 time points), (f) posterior deltoid (all time points). The methods include: Standard Normalizing (SN), Fatigue Only Normalizing (FON), Linear Model Normalizing (LMN), Cubic Model Normalizing (CMN), Points Forward Normalizing (PFN), Points Forward/Backward Normalizing (PFBN).

Pearson product moment correlations between the measured reference exertion EMG and the points predicted with the linear model and the cubic models showed that the cubic model points correlated better to the actual data points than the linear model points (Table 4.4). Using *PFN (5)*, there were significant differences from *SN (1)* for both time point analyses in the anterior, middle and posterior deltoid EMG (Figure 4.5). When only the first 72 time points were analyzed, there were significant differences in the pectoralis major muscle (sternal head) and when all time points were included there were significant differences in the upper trapezius, serratus anterior, and the pectoralis major (clavicular head). When the *PFBN Method (6)* was tested, there were only significant differences from the *SN Method (1)* in the anterior, middle and posterior deltoid, and pectoralis major muscles (sternal head) when the first 72 time points were analyzed. There were no significant differences when all time points were included in the analysis.

Table 4.4: Pearson product moment correlations values between the measured EMG values during the reference exertions and the linear model and cubic model predictions at those points. Significance values show that for all muscles, the values predicted with the cubic model had greater correlations to the measured values than the ones predicted with the linear model ($p<0.05$).

Muscle	Linear-Mean	Linear-SD	Cubic-Mean	Cubic-SD	Sig (2-tailed)
Adel	0.57	0.27	0.72	0.18	0.002
Mdel	0.52	0.25	0.74	0.19	0.000
Pdel	0.82	0.22	0.89	0.18	0.005
Infra	0.46	0.26	0.66	0.19	0.001
Utrap	0.41	0.28	0.72	0.21	0.000
Mtrap	0.40	0.28	0.72	0.19	0.000
Ltrap	0.37	0.22	0.68	0.18	0.000
Lats	0.36	0.22	0.67	0.20	0.000
Sert	0.31	0.25	0.64	0.17	0.000
PecC	0.31	0.29	0.59	0.22	0.000
PecS	0.33	0.21	0.64	0.19	0.000

4.5 Discussion

The purpose of this work was to evaluate the efficacy of five novel EMG normalizing functions to mitigate the expected increase in EMG amplitude indicative of myoelectric fatigue, allowing us to distinguish changes in EMG due to load sharing from changes due to myoelectric fatigue. EMG amplitude from 11 shoulder muscles were normalized using five novel normalization methods and compared to a standard normalizing method. Significant differences were dependent on the muscle examined, normalizing method used, and the number of time points in the analysis.

Although there were significant between participant differences in all of the muscles examined, individual differences in slopes of the mixed effects model were very small (estimated variances in slopes ranged from 8.6×10^{-5} to 2.89×10^{-8}). Thus, we do not believe that these are meaningful differences and believe that the results are applicable across all of the participants. Participants had varying endurance times and muscles displayed fatigue at different time points throughout the protocol. These differences likely explain some of the individual model differences.

When comparing the effects of each novel normalizing method to the standard normalizing method, we found that the significant differences were dependent on both the muscle and the number of time points included in the analysis. Results were consistent between the FON, LMN, and CMN methods, with significant differences in the posterior deltoid muscle. The Fatigue Only normalizing method was labour intensive, as it combined participant specific EMG amplitude and frequency analyses. This method, along with the PFN and PFBN methods were sensitive to variability in an individual reference exertion. The Linear and Cubic Model methods had significant differences that were consistent with the Fatigue Only method but were less sensitive to variability in individual reference exertions as the models were fit to the data over the entire endurance time. Since the predicted points from the Cubic Model were better correlated to the measured data than the Linear Model points, we recommend this method as a novel method to compensate for the effects of fatigue on EMG amplitude.

The shoulder musculature is complex and understanding load sharing relationships between these muscles is challenging, especially in the presence of muscle fatigue. Previous work has attempted to correct for the influence of fatigue in other joints. Regression equations have been developed to estimate the effect of fatigue on EMG amplitude during sustained static and dynamic and intermittent elbow flexion exertions (Hagberg, 1981). These equations have been used to predict what the expected increase in EMG resulting from muscle fatigue is; therefore, the other changes in the EMG amplitude during a prolonged trunk exertion could be attributed to changes in muscle force (O'Brien & Potvin, 1997). Na et al (2014) used the first dorsal interosseous in an index finger abduction fatigue protocol to demonstrate the utility of a time-varying gain factor to rescale the EMG-force relationship as it changed over the collection period. Similarly, a time-varying gain factor method has also been used to adjust a more complex EMG-force model of the trunk based on time and the previous exertions (Sparto & Parnianpour, 1998). In another trunk muscle modelling application, the maximum muscle stress was adjusted based on a function of changes in mean power frequency during a prolonged exertion, authors found reduced mean error between the measured and predicted net moments with the adjustment (Haddad et al, 2013). A more complex application of time-varying EMG normalization methods is seen in EMG-based control systems, such as those used with prosthetics. These systems rely on a consistent relationship between EMG and torque/intended motion and need to be designed to be used over long periods of time and under a variety of conditions. Lalitharatne et al (2013) developed a fuzzy rule based scheme to account for time-varying changes in EMG with fatigue that was effective at controlling flexion and extension of a robotic elbow joint under fatigued conditions. This method is mathematically complex and required a customized fuzzy rule scheme for each individual. The methods proposed in the current work attempt to compensate for the effects of fatigue to understand time-varying changes in load sharing in a complex group of muscles and across a variety of motions; a benefit of this method is that it can be applied to any task or muscle groups.

The fatigue normalizing methods proposed and evaluated in this work have a variety of applications. Low-load, repetitive work is common in the workplace, however, our understanding of the load sharing and muscular changes over time is limited by our ability to infer muscle force estimates from EMG. The method proposed in this paper allows for greater understanding of how these relationships change as capacity changes over the course of a work task or workday and may give greater insights into injury development over time. This method has been developed for the shoulder muscles, however the concept could be easily applied to other joints and muscle groups, allowing further understanding of load sharing changes over time across the body. EMG-force modelling is an important tool to incorporate task variability and individual variability into muscle force estimates, the ability to incorporate time-varying EMG parameters into these models is an important step to improve their application to real work tasks. Future work should attempt to evaluate the applicability of these EMG normalizing methods for use in muscle models.

There are several limitations to the current work that should be considered. These methods were created using a subset of shoulder muscles and have not been tested on other muscle groups. However, the shoulder musculature is very complex, and if these methods can be successfully applied to this group of joints then they should be effective in other regions as well. Also, the static reference exertions need to be performed throughout the fatiguing work and therefore must be built into the experimental protocol, this limits the ability to use these methods on previously collected data. This work can be used to guide future research on prolonged, repetitive work. Finally, as we cannot directly measure muscle force, there is no gold standard to compare these results to and ensure their validity. This is a limitation to all EMG-force applications and should not prevent further exploration into this and other methods to evaluate load-sharing changes over time.

In conclusion, using a cubic regression model to create a time-varying factor to normalize EMG shows promise as a novel method to mitigate the effects of myoelectric fatigue on EMG amplitude. Applying this method to EMG data from a simulated work task had a significant effect on the mean EMG amplitude when compared to the standard normalizing method. The cubic model normalizing method can be applied to a variety of research applications that attempt to improve our understanding of load sharing changes between muscles during fatiguing, prolonged or repetitive tasks.

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CHAPTER FIVE

Using EMG amplitude and frequency to calculate a multi-muscle fatigue score and evaluate global shoulder fatigue

Alison C. McDonald, Daanish M. Mulla, Peter J. Keir

Occupational Biomechanics Laboratory, Department of Kinesiology, McMaster University, Hamilton, ON, Canada, L8S 4K1

*Corresponding Author:

Peter J. Keir, PhD
McMaster University
Department of Kinesiology
Ivor Wynne Centre, room 212
1280 Main Street West
Hamilton, ON, Canada, L8S4K1
Telephone: 905-525-9140 ext. 23543
Email: pjkeir@mcmaster.ca

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5.1 Abstract

Background: Evaluating both frequency and amplitude components of the electromyographic signal provides a more complete evaluation of muscle fatigue than either variable alone, however; little effort has been made to combine time and frequency domains for the evaluation of myoelectric fatigue.

Objective: The objective of this analysis was to develop a function to quantify fatigue in multiple shoulder muscles by generating a single score using relative changes in EMG amplitude and frequency over time.

Method: Surface EMG was measured from 14 shoulder muscles while 20 participants performed simulated, repetitive work tasks until exhaustion. The function was generated to calculate a multi-muscle fatigue score (MMFS) based on changes in EMG frequency, amplitude and the number of muscles showing signs of myoelectric fatigue. The utility of the function was evaluated through changes in the MMFS over time and by evaluation of the relationships between MMFS and changes in strength and MMFS and perceived fatigue (RPF) over time.

Results: MMFS scores significantly increased over time ($p<0.05$) with significant relationships between MMFS and strength changes and RPF ($p<0.05$).

Conclusion and application: The MMFS function provides a method for assessing myoelectric fatigue development that is sensitive to time and to perception of fatigue. The MMFS function is easily adaptable to different muscle groups and workplace tasks to provide a single score that represents global muscle fatigue. By generating a single score,

this function allows for comparisons between workplace tasks, which can aid in workplace design to mitigate the development of fatigue.

Précis: EMG amplitude and frequency were combined to develop a function to quantify fatigue in multiple muscles with a single score. The MMFS assesses myoelectric fatigue development and is sensitive to time and perception of fatigue and can be applied to evaluation of fatigue development in the workplace.

5.2 Introduction

Repetitive and prolonged work is common in the workplace. Repetitive demands continuously challenge the same muscles, and over time, can lead to the development of muscle fatigue; thus upper extremity muscle activity during these tasks is of ergonomic interest. The kinematic and muscular degrees of freedom in the shoulder complex allow for multiple strategies between and within individuals to complete a functional task at the hand (Tse et al, 2016; McDonald et al 2016). Along with individual differences, differing strategies cause fatigue to develop in different muscles and at varying rates across people, making it challenging to compare multi-muscle fatigue states between people by using individual muscles (Gerdle et al 2000). Fatigue begins to accumulate at the start of a contraction and progresses continuously throughout without muscular rest. Assessing the fatigue state at the beginning and end of an endurance task limits the information obtained on the progression and evolution of fatigue throughout the task (Bonato et al 1996).

A common way to objectively measure muscle fatigue is by evaluating the electromyographic (EMG) signal. With myoelectric fatigue, there are changes in both time and frequency domains of the EMG signal (Al-Mulla et al 2011). As a muscle fatigues, the EMG signal is affected by firing rate, recruitment changes, the shape of the motor unit action potentials (MUAP), synchronization of MUAPs, and conduction velocity (De Luca 1979), leading to a measurable increase in EMG amplitude and a decrease in EMG frequency (Hagberg 1981, Potvin and Bent 1997, Navaneethakrishna and Ramakrishnan 2014). These changes in the EMG signal (i.e. increased amplitude, decreased frequency) have been well documented and are well accepted, however challenges in interpretation remain. Although the combination of EMG frequency and amplitude changes provides a more complete evaluation of muscle fatigue, little effort has been made to combine the time and frequency domains to evaluate fatigue. MacIssac and colleagues (2006), created a model to estimate a fatigue index based on changes in both amplitude and frequency parameters. The statistical model was limited to one muscle (biceps brachii), assumed a linear progression of fatigue from start to failure, and also required task and subject specific training. Another statistical model combined the two EMG parameters for 3 knee extensors using partial least squares regression to predict peak torque over time during a dynamic task. The authors found that only the frequency domain parameters were required for the prediction. Although not statically significant, the authors noted that the group r^2 value was greater when both parameters were included and that there was variability in specific subjects' EMG amplitude responses over time (Gerdle et al 2000). Although these methods evaluate both myoelectric parameters well, they are limited to a small number of muscles and relatively simple joints. The Joint Amplitude and Spectral Analysis method was proposed to distinguish between force-related and fatigue-related changes to the EMG spectrum (Luttmann et al, 2000). This method evaluates muscles individually and fatigue related changes are required to have increased EMG amplitude and decreased EMG frequency. The shoulder complex has many degrees of freedom and during repetitive work tasks, individuals develop muscle fatigue in different muscles and at different rates, making it challenging to quantify overall fatigue level when evaluating muscles individually.

The behaviour of EMG signal and the progression of fatigue over time is variable between individuals (Gonzalez-Izal et al 2010, Gerdle et al 2000) and there is limited work that has quantified fatigue across multiple muscles with a single value. Quantifying fatigue scores that combine multiple muscles over time can allow for more effective comparisons between people for the evaluation of workplace tasks. Roger and MacIsaac (2010) measured 7 channels of EMG (on one muscle), calculated a fatigue estimate for each channel and then used the highest one for the fatigue estimate of the given contraction. If applied to a group of muscles, this method would account for differing muscles being fatigued between individuals; however, the actual score for each person or task is only dependent on one muscle and wouldn't reflect the fatigue state of the entire shoulder complex. Gonzalez-Izal and colleagues (2010) used artificial neural networks, a statistical modeling technique, to predict force loss during knee extensions using myoelectric fatigue indicators in multiple knee extensor muscles. The model had to be trained for each subject and each task, making practical applications challenging. Bilateral trunk muscle fatigue has been evaluated by calculating a fatigue gain factor based on median frequency change from baseline, using the mean from the right and left trunk muscles (Haddad et al 2012). The assumption of symmetry between the muscles was accepted based on the symmetric task studied, however this would not be an appropriate assumption in the shoulder, making the evaluation of upper extremity fatigue across individuals during repetitive work tasks challenging. Overall, the current models are not able to effectively evaluate fatigue development in the group of muscles in the shoulder complex or are too mathematically and functionally burdensome to apply to the evaluation of workplace tasks. The purpose of this analysis was to develop a function to quantify shoulder muscle fatigue using relative changes in EMG amplitude and frequency in multiple shoulder muscles over time that could be applied to the evaluation of workplace tasks.

5.3 Methods

5.3.1 Participants

Twenty right-handed men from the university population participated in this study (height: 179.3±5.1 cm; weight: 84.6±13.6 kg; age: 23±4 years). All participants were free from upper extremity pain or injury within the last year. This protocol was approved by the McMaster Research Ethics Board and prior to beginning the protocol all participants provided informed, written consent.

5.3.2 Instrumentation and Protocol

After providing informed written consent, anthropometric measures were taken (weight, height, shoulder height, umbilicus height, arm length). The data used to create the MMFS function were collected as part of a larger investigation on the effects of repetitive work on the shoulder complex. Bipolar surface electrodes (Trigno, Delsys Inc., Natick MA, USA) were affixed to 14 muscles (biceps brachii, triceps brachii, anterior deltoid, middle deltoid, posterior deltoid, infraspinatus, supraspinatus, upper trapezius, middle trapezius, lower trapezius, latissimus dorsi, serratus anterior, sternal head of the pectoralis major, and clavicular head of the pectoralis major). Prior to electrode placement, sites were located with guidance from literature (Ekstrom et al 2005; Waite et al 2010; Hodder and Keir 2013) and manual palpation. The skin was shaved and cleansed with isopropyl alcohol. EMG signals were sampled at 1926 Hz, differentially amplified (input impedance $10^{15}\Omega$, CMRR > 80 dB), band-pass filtered (20-450 Hz), and converted with a 16-bit card (±5 V range). Participants completed a 10 second rest trial and then began a maximum voluntary exertion (MVE) protocol. Each participant completed two sets of 8 different MVE tests with one minute of rest between exertions (McDonald et al 2017). As part of the collection, participants were also instrumented with a full body kinematic marker set (62 reflective markers, 11 Raptor-4 cameras, Motion Analysis Corporation, Santa Rosa, CA).

Participants performed 60 second work cycles (4 work tasks/cycle) that were scaled to participant's anthropometrics and strength (described in Chapter 6). They continued performing the repetitive work tasks until one of three stoppage criteria were met: (1) verbally declared they were unable to continue, (2) unable to meet any of the work task demands, or (3) unable to maintain 30% MVC during any of the 4 submaximal reference exertions. Every 3 minutes, participants were prompted to provide a rating of perceived fatigue (RPF) on a 0-10 scale and EMG, kinematic, and task data were collected. Every 12 minutes, participants stopped performing the work tasks and completed 4 submaximal (30% MVC) and 1 maximal reference exertion (Table 5.1) to evaluate fatigue development. To set the 30% target force for each of the reference exertions, maximal exertions were performed in each of these postures (Mark-10 Corporation, Copiague, NY, USA). Each test was repeated twice, with 2 minutes of rest between exertions. Tests were repeated if the values were not within 10%, and the mean of the 2 values within 10% was used to calculate the 30% submaximal targets.

Table 5.1: Four submaximal reference exertions were performed every 12 work cycles to elicit activation from the 14 muscles of interest and were performed at 30% MVC (Boettcher et al, 2008).

Reference Exertion	Exertion Description
Empty Can Test	90° shoulder abduction in plane of scapula with internal humeral rotation and elbow extended. Arm abducted with resistance applied at the wrist.
Internal Rotation	90° shoulder abduction in plane of scapula with neutral humeral rotation and 90° elbow flexion. Arm internally rotated with resistance applied at wrist.
Flexion	Seated with erect back posture, no back support, 125° shoulder flexion with thumb pointing upwards and elbow extended. Arm flexed with resistance applied above the elbow.
Palm Press	90° shoulder flexion and 20° elbow flexion. Horizontal adduction with resistance applied at the heel of hand.

5.3.3 Data Analysis

The EMG from the central three seconds of each submaximal exertion was used for the amplitude and frequency analyses. A spectral analysis was performed using a Fast Fourier Transformation and the median power frequency (MPF) was calculated using a 0.125 s sliding rectangular window with 0.0625 s window overlap. For the amplitude analysis, the raw EMG data (MVE, submaximal) were full wave rectified and linear enveloped with a dual-pass Butterworth filter (2nd order, f_c = 4 Hz). The peak EMG amplitude for each muscle was obtained from the series of MVEs and the submaximal EMG data were linearly normalized to 100% MVE of each muscle. Throughout the protocol, the EMG data from the submaximal reference exertions were evaluated relative to the values in the baseline reference exertions (taken prior to beginning the work protocol).

The fatigue quantifying function expressed in Equation 1 calculates a multi-muscle fatigue score (MMFS) by accounting for changes in EMG frequency, amplitude and the number of muscles showing signs of myoelectric fatigue. A muscle was classified as showing signs of muscle fatigue and included in the MMFS calculation only if there was both an increase in EMG amplitude and a decrease in EMG frequency. A multiplier was developed for the equation to shape the curve by accounting for the number of muscles that contributed to MMFS. This was done so that a summated score that was created with a greater number of muscles would have a lower fatigue score than if the same score was calculated with a smaller number of muscles. The magnitude of the multiplier changes depending on the number of muscles included in the analysis and the number of them exhibiting signs of fatigue. Fewer muscles exhibiting signs of fatigue results in a greater multiplier value. The shape of the multiplier curve is dependent on the number of fatigued muscles included in the analysis, the fewer muscles included, the "sharper" the shape of the curve.

$$MMFS = \sum_0^{n_f} \left(\left| AEMG_{i_f} - AEMG_{i_b} \right| + \left| \%\Delta MPF_{i_f} \right| \right) \times \tanh\left(\frac{N}{n_f} \Big/ \sqrt{N} \right) \qquad \textbf{(1)}$$

Where, MMFS = multi-muscle fatigue score, $AEMG_{i_f}$= average EMG for each fatigued muscles (% MVE),, $AEMG_{i_b}$= average EMG for each fatigued muscles at baseline (% MVE), $\%\Delta MPF_{i_f}$= percent change median power frequency for each fatigue muscle from baseline, N = the total number of muscles measured, nf = number of muscles showing signs of muscle fatigue (increase in EMG amplitude and decrease in MPF) .

To evaluate the MMFS function, a MMFS was calculated at three time points for each participant: (1) after the first 12 work cycles, (2) after the middle work cycle, and (3) after the last work cycle. To evaluate the progression of fatigue over time, a repeated measures ANOVA was performed on the fatigue scores calculated for these 3 time points. The number of muscles exhibiting signs of fatigue and included in the analysis were also evaluated at each time point with a repeated measures ANOVA. Post hoc analyses were completed using Tukey's HSD tests. To evaluate how well myoelectric fatigue related to participants' perception of fatigue and changes in strength Pearson Product Moment correlations were performed between the MMFS and ratings of perceived fatigue (RPF) and between the MMFS and flexion strength. Statistical tests were conducted in SPSS Statistics (v20.0, IBM, NY, USA) with $\alpha = 0.05$.

5.4 Results

Participants performed the work cycles for 57 to 240 minutes and completed 5 to 20 sets of reference exertions before meeting one of the termination criteria. The calculated MMFS, RPF values and the change in glenohumeral strength from baseline for each reference exertion, across all participants is presented in Table 5.2. Fatigue scores increased significantly between the first (31.8 ± 14.6), middle (47.6 ± 25.3), and final set of reference exertions (58.6 ± 35.5) ($p<0.05$) (Figure 5.1). There were no significant differences between the middle and the last work cycles. There were no significant differences in the number of muscles exhibiting signs of fatigue between the first (4.7 ± 1.6), middle (4.9 ± 2.9), and final (5.3 ± 2.4) work cycles ($p>0.05$).

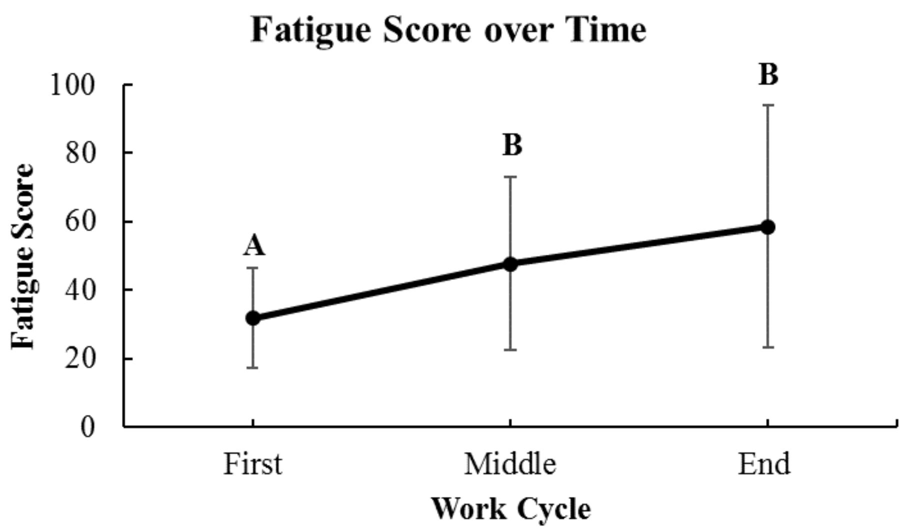

Figure 5.1: Mean Fatigue Score over time across all participants (standard deviation error bars). There is a significant increase in fatigue between the first, middle and last work cycles ($p<0.05$). There were no differences between the middle and end scores.

Table 5.2: Participants ratings of perceived fatigue, changes in glenohumeral flexion strength (kg) and fatigue score for all reference exertion time points. Changes in strength are relative to each participants baseline strength measurement.

Reference Variables for each Participant	Work Cycle	12	24	36	48	60	72	84	96	108	120	132	144	156	168	180	192	204	216	228	240
	RPF	3.0	5.0	7.0	9.0	10.0															
P16	Strength	-13.8	-9.8	-14.0	-30.8	-20.5															
	FS	54.6	52.4	60.2	58.7	87.6															
	RPF	3.0	3.0	4.0	9.0	9.0															
P20	Strength	-23.2	-38.3	-22.7	-48.2	-65.3															
	FS	34.6	69.0	98.1	141.0	157.1															
	RPF	5.0	7.0	8.0	9.0	9.0															
P9	Strength	-51.1	-55.3	-58.4	-42.6	-67.6															
	FS	62.6	79.0	82.7	100.1	120.0															
	RPF	5.0	6.0	8.0	9.0	9.9	10.0														
P3	Strength	-8.0	-18.3	-10.7	-2.5	-13.6	-15.4														
	FS	43.5	27.5	22.8	25.8	31.1	31.1														
	RPF	1.0	3.5	6.0	8.0	9.0	9.5														
P13	Strength	-25.0	-32.8	-48.6	-68.4	-75.1	-78.7														
	FS	54.8	59.1	68.3	95.9	65.2	49.7														
	RPF	4.0	5.0	6.0	7.5	9.0	9.5														
P14	Strength	-4.9	-6.5	-4.7	-13.8	-23.0	-40.8														
	FS	8.8	43.3	46.2	28.0	25.8	7.2														
	RPF	5.0	7.0	8.0	8.0	9.0	10.0														
P22	Strength	-23.0	-24.5	-8.5	-8.2	-25.0	-16.5														
	FS	32.6	15.0	4.7	14.7	19.5	43.4														
	RPF	3.5	5.0	6.0	7.0	8.5	9.5	10.0													
P2	Strength	2.2	-2.0	-0.4	-13.4	-14.5	-42.1	-56.6													
	FS	18.5	56.7	33.6	47.4	57.7	44.4	44.0													
	RPF	4.0	4.0	6.0	7.0	9.0	9.0	8.0													
P18	Strength	-5.8	-4.2	-23.9	9.4	11.4	8.9	1.8													
	FS	28.8	43.2	67.4	16.8	35.8	21.3	29.5													
	RPF	2.0	4.0	5.0	6.0	8.0	9.0	9.5													
P21	Strength	1.3	-1.3	-3.3	-1.6	-18.3	-27.4	-4.5													
	FS	31.5	48.7	33.8	75.8	64.1	51.9	60.8													
	RPF	7.0	8.0	8.0	7.0	8.0	8.0	8.0	9.0												
P12	Strength	-14.5	-21.4	-16.5	-20.5	4.7	2.7	-2.0	-21.6												
	FS	36.7	13.1	23.9	5.8	16.4	7.0	0.0	12.7												
	RPF	2.0	2.0	3.0	3.0	4.5	6.5	7.0	9.0	10.0											
P23	Strength	-6.7	-23.2	-15.2	-30.1	-6.7	-10.7	-29.9	-28.1	-72.2											
	FS	25.8	32.4	33.7	29.3	37.2	22.7	28.5	16.1	34.5											
	RPF	1.0	2.0	3.0	4.0	4.0	5.0	6.0	8.0	9.0	9.0	9.5									
P10	Strength	-6.7	3.3	4.2	-21.6	-7.1	-14.3	-14.3	-14.3	-27.0	-31.9	-37.0									
	FS	29.5	31.4	25.0	40.2	47.6	46.3	32.7	48.7	62.1	74.5	66.1									
	RPF	2.0	3.0	4.0	5.0	7.0	7.0	7.0	7.0	8.0	9.0	10.0									
P4	Strength	-9.8	-18.1	-3.6	-14.3	-13.4	-28.3	-6.9	-29.4	-22.1	-24.7	-54.2									
	FS	11.6	27.0	19.9	n/a	43.0	33.0	36.1	51.1	41.3	36.3	43.8									
	RPF	0.5	1.0	3.0	3.0	3.0	3.0	3.0	4.0	5.0	6.5	8.5	9.5								
P5	Strength	16.5	3.8	15.4	9.4	13.2	4.2	-7.6	-6.2	-22.5	-12.0	-58.9	N/A								
	FS	20.7	31.9	34.2	29.9	37.9	49.7	38.4	34.7	20.6	44.6	56.7	51.5								
	RPF	3.0	5.0	5.0	6.0	6.0	7.0	7.0	7.0	8.0	8.0	9.0	10.0								
P8	Strength	-7.1	-15.2	-8.0	-10.5	-8.0	-7.1	-1.6	-5.4	-12.5	-8.2	-17.2	-29.0								
	FS	30.9	12.8	30.9	13.7	17.7	13.2	24.9	37.7	36.7	47.8	44.9	34.6								
	RPF	1.0	1.0	2.0	3.0	3.0	4.0	5.0	7.0	8.0	9.0	9.0	10.0								
P15	Strength	-14.9	-22.5	-24.1	-19.4	-40.4	-16.3	-37.5	-53.5	-52.8	-64.0	-65.1	-82.3								
	FS	13.6	34.6	51.2	55.4	39.8	56.3	63.4	70.7	69.2	74.0	79.7	83.9								
	RPF	4.5	4.5	5.0	3.5	2.5	3.0	4.0	5.0	5.0	6.0	6.0	7.0	8.0	9.0	9.0					
P19	Strength	0.7	0.7	-2.0	8.2	12.9	-3.6	5.1	-15.8	-13.4	-25.2	14.3	-13.6	-2.5	-43.9	-74.0					
	FS	44.4	64.0	44.5	40.2	36.2	75.2	48.7	44.1	49.1	48.4	44.8	46.4	51.5	47.7	67.7					
	RPF	2.0	3.0	4.0	5.0	7.0	7.0	8.0	8.0	8.0	7.0	8.0	8.0	8.0	8.0	8.0	8.0	8.0	8.0	8.0	8.0
P6	Strength	-49.9	-27.9	-31.9	-41.0	-51.7	-32.1	-37.7	-41.7	-37.9	-28.5	-43.3	-47.0	-45.5	-45.5	-54.8	-56.0	-43.9	-58.0	-63.3	-56.4
	FS	31.9	34.8	41.9	48.3	53.4	27.0	27.1	46.1	43.8	71.3	54.8	82.9	88.8	100.2	99.5	75.6	83.6	81.7	N/A	79.8
	RPF	0.0	1.0	1.0	2.0	3.0	4.0	5.0	5.0	5.0	6.0	6.0	7.0	8.0	9.0	9.0	9.0	9.0	9.0	9.0	9.0
P7	Strength	-17.2	-28.3	-21.2	-16.5	-19.8	-10.0	-15.2	-8.2	-23.6	-30.1	-27.2	-26.1	-30.3	-30.3	-29.0	-43.9	-42.4	-82.9	-39.5	-37.5
	FS	21.3	19.2	1.9	23.6	22.2	36.9	28.1	40.8	57.1	50.2	48.7	31.5	36.6	55.9	42.7	58.6	47.0	60.7	66.8	66.5

Pearson product moment correlations indicate that there is a statistically significant ($p<0.05$), weak, positive ($r=0.298$) relationship between MMFS and participant rating of perceived fatigue over time (Figure 5.2a). There was also a statistically significant ($p<0.05$), moderate, negative ($r=-0.510$) relationship between the MMFS and the change in glenohumeral flexion strength from baseline (Figure 5.2b).

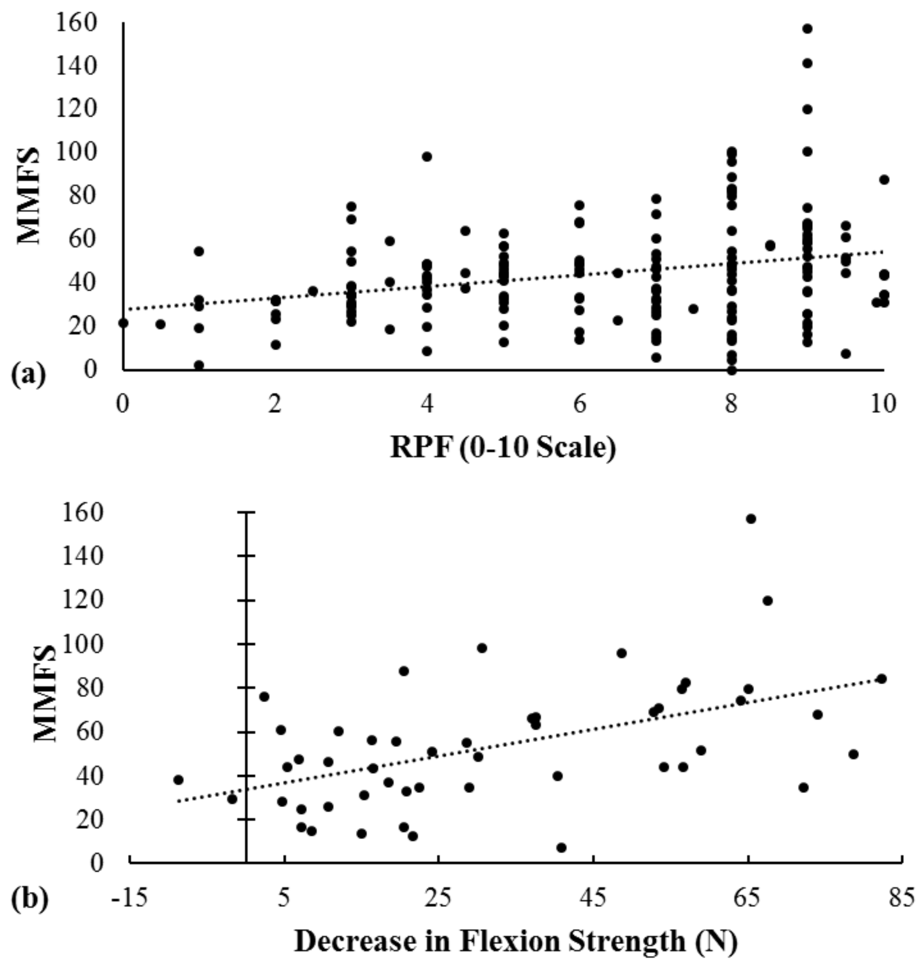

Figure 5.2: (a) Relationship between MMFS and participants Rating of Perceived Fatigue (RPF) for all 20 participants. Pearson production moment correlations revealed a significant, weak, positive correlation between these two variables. (b) Relationship between MMFS and change in glenohumeral flexion strength (N) for all 20 participants. Pearson production moment correlations revealed a significant, moderate, negative correlation between these two variables. Linear lines of best fit are included for both relationships.

5.5 Discussion

The number of muscles in the shoulder complex and large variability in muscle activity patterns across people make it challenging to effectively evaluate fatigue development. The purpose of this work was to generate a single value to quantify overall shoulder muscle fatigue based on changes in EMG amplitude, EMG frequency and the number of muscles exhibiting signs of muscle fatigue.

This work combines changes in EMG amplitude and frequency to evaluate muscle fatigue. This new method operationalizes the fatigue quadrant of the Joint Analysis of EMG Spectrum and Amplitude (JASA) method (Luttmann et al, 2000) and expands it to include multiple muscles. The JASA method proposed that changes in amplitude and the spectral parameters could differentiate between force-related and fatigue-related changes to the EMG spectrum, providing a construct to aid in understanding EMG amplitude and frequency changes. Application of these theories to the current work was effective at generating a MMFS that could evaluate fatigue across multiple different muscles and was sensitive to time. Repetitive work tasks were completed until exhaustion and the MMFS increased over time, while the number of muscles exhibiting fatigue across participants did not change. As previous reports have concluded, the muscular response to fatigue is variable over time (Gonzalez-Izal et al 2010, Gerdle et al 2000); this is displayed in the variability in numbers of muscles fatigued between individuals and was not unexpected.

There was a significant relationship between the MMFS calculated in this analysis and the perceptual ratings of fatigue (RPF) given by the participants during the simulated work tasks and changes in glenohumeral flexion strength. The reduction of strength is a measurable indicator and consequence of multi-muscle shoulder fatigue that has direct impacts on workplace ergonomics. The correlation between shoulder strength and the MMFS indicates that the score effectively captures the reduced capacity of the shoulder with fatigue. Ratings of perceived fatigue have been shown to relate to physiological indicators (heart rate, oxygen uptake, respiratory exchange ratio, carbon dioxide production, ventilation rate, blood lactate concentration) of fatigue (Micklewright et al, 2017). The rating of fatigue (ROF) scale created by Micklewright and colleagues was shown to have good face validity and high convergent validity during ramped cycling until exhaustion, recovery from this exercise and during activities of daily living. Although the correlation between the MMFS and RPF was weak in this analysis, fatigue is a global construct and muscle fatigue is only one aspect impacting perception. The statistically significant correlation between these variables suggests that the function represents the muscle fatigue portion of fatigue perception well as perception of fatigue alone may not be sensitive enough to detect all indications of myoelectric fatigue, especially during recovery conditions (McDonald et al, 2016). Using the MMFS function to quantify myoelectric fatigue in combination with perception can give a more complete analysis of fatigue development during repetitive work tasks.

There are limitations to this work that should be considered. Although the function produces a single value to represent muscle fatigue, there is little known about the meaning of the absolute score value. The value is sensitive to the number of muscles and the specific muscles included and therefore its use should be limited to comparing the fatigue development between and within individuals over time within the same muscles. Since a purpose of this function was to reduce the degrees of freedom within the shoulder complex, it does not give insight into which muscles are exhibiting signs of fatigue. Although these specific details are lost with the use of this function, this allows for comparisons amongst the between and within subject variability in the response of specific muscles. There were 14 muscles included in the development of this function and although that does not include all the muscles in the shoulder complex, the function has the flexibility to allow for as many or as few muscles as desired to be included in an analysis.

The multi-muscle fatigue score function proposed and evaluated in this work provides a method for assessing myoelectric fatigue development over time that is sensitive to time and to perception of fatigue. The function considers the changes in EMG amplitude, EMG frequency, number of muscles included in the analysis and the number of these muscles exhibiting signs of fatigue. These parameters make its applications adaptable to the evaluation of many different muscle groups and workplace tasks.

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CHAPTER SIX

Muscular and Kinematic Adaptations to Fatiguing, Repetitive Work

Alison C. McDonald, Daanish M. Mulla, Peter J. Keir

Occupational Biomechanics Laboratory, Department of Kinesiology, McMaster University, Hamilton, ON, Canada, L8S 4K1

*Corresponding Author:

Peter J. Keir, PhD
McMaster University
Department of Kinesiology
Ivor Wynne Centre, room 212
1280 Main Street West
Hamilton, ON, Canada, L8S4K1
Telephone: 905-525-9140 ext. 23543
Email: pjkeir@mcmaster.ca

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6.1 Abstract

Repetitive work is common in the modern workplace and the effects are often studied using fatigue protocols; however, there is evidence that the way fatigue develops will impact the response. The purpose of this work was to simultaneously evaluate muscular and kinematic adaptations during fatiguing, repetitive work to exhaustion. We measured muscle activity in 13 muscles in the shoulder and trunk and captured full body kinematics while participants completed simulated, repetitive work tasks. Every 12 minutes, data were collected to quantify fatigue. This sequence continued until they reached the termination criteria. Participants displayed significant signs of muscle fatigue, loss of strength and increased perceived fatigue ($p<0.05$). Analysis revealed a significant effect of time on posture and muscle activity that was both task and time dependent and variable both between and within individuals. Participants were able to compensate for reduced physical capacity and maintain task performance through coordinated compensation strategies.

6.2 Introduction

The large range of motion and degrees of freedom of the upper extremity, including the shoulder complex, elbow, and wrist, create a large working area for the hand and opportunities for many multi-joint movement strategies. Kinematic changes due to fatigue have been investigated during both simple movements and in more complex multi-joint movements. Following a shoulder fatigue protocol, scapulothoracic and glenohumeral kinematics changes observed during simple arm elevation tasks were found to be sensitive to humeral elevation angle (Ebaugh et al, 2006b; Joshi et al, 2011; Tsai, et al, 2003). In response to a fatigue protocol targeting the infraspinatus via external rotation, more clavicular retraction and less humeral external rotation was observed during humeral elevation movements (Ebaugh et al, 2006a). In more complex tasks involving multiple joints, participants utilize multi-joint coordination strategies to maintain task performance with fatigue (Heiderscheit, 2000). During a repetitive pointing task, participants maintained performance by changing their elbow and wrist movements to compensate for altered shoulder position (Fuller et al, 2009). In a more skilled, ball tossing movement, participants were still able to maintain performance in the presence of fatigue (Huffenus et al 2006). When successful throws were compared, it was observed that performance was maintained in a different manner post fatigue compared to pre fatigue. After the fatigue protocol, movement between the joints was more rigid, which simplified the movement by reducing the degrees of freedom of the arm system thus task complexity.

To investigate how fatigue affects movements in the workplace, simple simulated workplace tasks such as sawing, hammering, and screwing have been examined. Task completion performance after a fatigue protocol was maintained with experienced carpenters; however, there were slight changes in how the tasks were performed: slower sawing pace, faster screwing pace, and slight performance decrements in the hammering task. However, movement strategies were not evaluated as they did not monitor kinematics (Hammarskjöld & Harms-Ringdahl, 1991). In another repetitive hammering investigation, greater kinematic changes were found at the wrist and elbow joints than at the shoulder complex following a fatigue protocol. Although the fatigue protocol was designed to target the shoulder complex, fatigue was not evaluated quantitatively and thus, muscles surrounding the elbow joint may have fatigued to a greater extent than the shoulder. Since the reaction force at the hand is working through a closed chain, changes at the wrist and elbow would also impact the shoulder (Côté et al, 2005). With repetitive sawing, participants tended to move their shoulder closer to the end target after a fatigue protocol, but overall, the movement amplitude and duration in a sawing task was not affected. Further investigation of individual joint changes revealed that decreased elbow movement amplitude and increased amplitude in all other joints were observed. Although the changes in the other joints were small, when combined, they could compensate for large changes in elbow posture, leading to the conclusion that there was a coordinated multi-joint strategy employed to maintain task performance (Côté et al, 2002). These examples show that people can compensate for fatigue and maintain performance during repetitive tasks within the large working volume of the hand through multi-joint coordination strategies within the upper extremity. There has been limited work measuring detailed upper body kinematic changes, including the scapula and trunk, in response to simulated workplace tasks and in conjunction with changes in muscle activity.

A previous study in our laboratory investigated the effects of anterior deltoid fatigue on kinematic and muscular adaptations during repetitive work (McDonald et al, 2016; Tse et al, 2016). Following the fatigue protocol, participants reduced their glenohumeral flexion angle and made a combination of scapular changes to maintain an elevated shoulder posture. In addition to these kinematic changes found across participants, individual adaptation strategies were employed. Variability is a common factor investigated in association with fatigue, and has also been shown to both increase and decrease in different participants (Gates & Dingwell, 2008). This large variability would also suggest that with the number of available compensation strategies, how fatigue develops, in which muscles and when, might be an important factor in the progression of compensatory changes.

A challenge in evaluating the literature and understanding the relationship between fatigue and repetitive work is the sensitivity of the kinematic changes to how the muscles of the shoulder complex are fatigued. Voluntary fatigue leads to a reorganization of the multi-joint coordination strategies to maintain task performance but with electrically stimulated fatigue, participants are more likely to use the same joint coordination strategy and instead compensate with increased muscle activity (Huffenus et al, 2006; Huffenus & Forestier, 2006). During voluntary fatigue, there is active involvement of the central nervous system that does not occur with stimulated fatigue, which can potentially lead to different motor control strategy outputs with the two strategies (Huffenus et al, 2006). There are also differences in energy metabolism with the two mechanisms (Vanderthommen et al, 2003). Movement constraints can also lead to different kinematic changes with fatigue than unconstrained or free movements. This will change the mechanisms contributing to fatigue as a task progresses and therefore produce different outcomes (Amasay & Karduna, 2009; Enoka & Stuart, 1992). The dependency of the response to how muscle fatigue develops makes it challenging to arrive at a complete understanding of the fatigue and recovery process in the shoulder complex. In the work preceding this study, fatigue was induced with a fatigue protocol aimed at the anterior deltoid muscle (McDonald et al, 2016; Tse et al, 2016). Compensatory changes observed are likely different than those that would occur if fatigue had developed while performing the simulated work tasks alone. The purpose of this work was to simultaneously evaluate muscular and kinematic adaptations during fatiguing, repetitive work to exhaustion. We hypothesized that people will use coordinated compensation strategies to maintain performance for a limited time and that there will be variability in the responses across individuals.

6.3 Methods

Eighteen right-handed males (height: 179.0±5.3 cm; weight: 84.1±14.2 kg; age: 23±4 years) free of any upper extremity injuries/pain in the past 12 months participated in this study. Participants provided informed, written consent before beginning the protocol and the study was approved by the Hamilton Integrated Research Ethics Board (HIREB).

6.3.1 Instrumentation

Muscle activity was recorded from 13 shoulder/trunk muscles – biceps brachii (Bi), triceps brachii (Tri), anterior (Adel), middle (Mdel) and posterior (Pdel) deltoids, infraspinatus (Infra), upper trapezius (Utrap), middle trapezius (Mtrap), lower trapezius (Ltrap), latissimus dorsi (Lats), serratus anterior (Sert), sternal head of pectoralis major (PecS), clavicular head of pectoralis major (PecC) – on the right side, using silver-contact wireless bipolar bar electrodes (fixed 1 cm inter-electrode spacing) (Trigno, Delsys Inc., Natick, MA, USA). EMG signals were differentially amplified (CMRR > 80 dB, input impedance $10^{15}\Omega$), band-pass filtered (20-450 Hz), and 16-bit converted at 1926 Hz (± 5 V range). A quiet trial was performed and followed by two repetitions each of 10 maximal voluntary exertions (MVE), with 2 minutes of rest provided between each MVE (McDonald et al, 2017). Full body kinematics were recorded (50 Hz) with 11 cameras (Raptor-4, Motion Analysis Corporation, Santa Rosa, CA), and 72 reflective markers placed on anatomical landmarks and a custom-designed scapular tracker (Karduna et al, 2003, Tse et al, 2016).

6.3.2 Protocol

Following EMG and motion capture set-up, participants completed a series of baseline reference measures. These include four submaximal (30% MVC) exertions (125° glenohumeral flexion, empty can, palm press, and internal rotation; McDonald et al, *Thesis Chapter 4: Submaximal Normalizing Methods to Evaluate Load Sharing Changes* *in Repetitive Work*), one maximum glenohumeral flexion exertion (shoulder flexed to 90°, elbow extended), and a rating of perceived fatigue (RPF) on a scale from 0-10. Participants then began a series of cyclic work tasks on a custom apparatus designed to simulate repetitive work and induce upper extremity muscle fatigue (Figure 6.1). The apparatus was instrumented with two 6-dof-force transducers (MC3-500, MC3-100 AMTI, Watertown, MA), 2 linear potentiometers, and was scaled to the participant's anthropometrics and strength. Participants stood on a force plate (ORF-6, AMTI, Watertown, MA) and the apparatus was positioned at 60% of each participant's arm reach. Foot placements were marked on the force plate to ensure that participants maintained the same arm reach throughout the repetitive work. Tasks 1 and 2 were positioned above the umbilicus by one-half the vertical distance between the umbilicus and AC joint, and tasks 3 and 4 were adjusted to this same vertical distance above the AC joint. Each work cycle (WC) was 60-seconds long and comprised of 4 tasks: (Task 1) a weighted pulling task (60% MVC, 14 seconds), (Task 2) a weighted pushing task (60% MVC, 14 seconds), (Task 3) an anterior drill press task (50% MVC, 10 seconds), and (Task 4) an up/down target-matching task with two force levels (15, 35%, 16 seconds). The remaining 6 seconds were built into the work cycle as rest effectively creating a 90% duty cycle. Participants were given visual feedback (Labview, National Instruments, Texas, USA) to maintain task performance. After 12 work cycles, participants repeated the reference measures listed above (4 submaximal, 1 maximal, RPF), and as quickly as possible, resumed another set of 12 work cycles. The sequence of 12 work cycles, followed by reference measures continued until one of three termination criteria were met: (1) participant was no longer able to maintain task performance, (2) participant was no longer able to maintain the 30% force level for any one of the submaximal reference exertions, or (3) participant verbally declared that they were unable to continue.

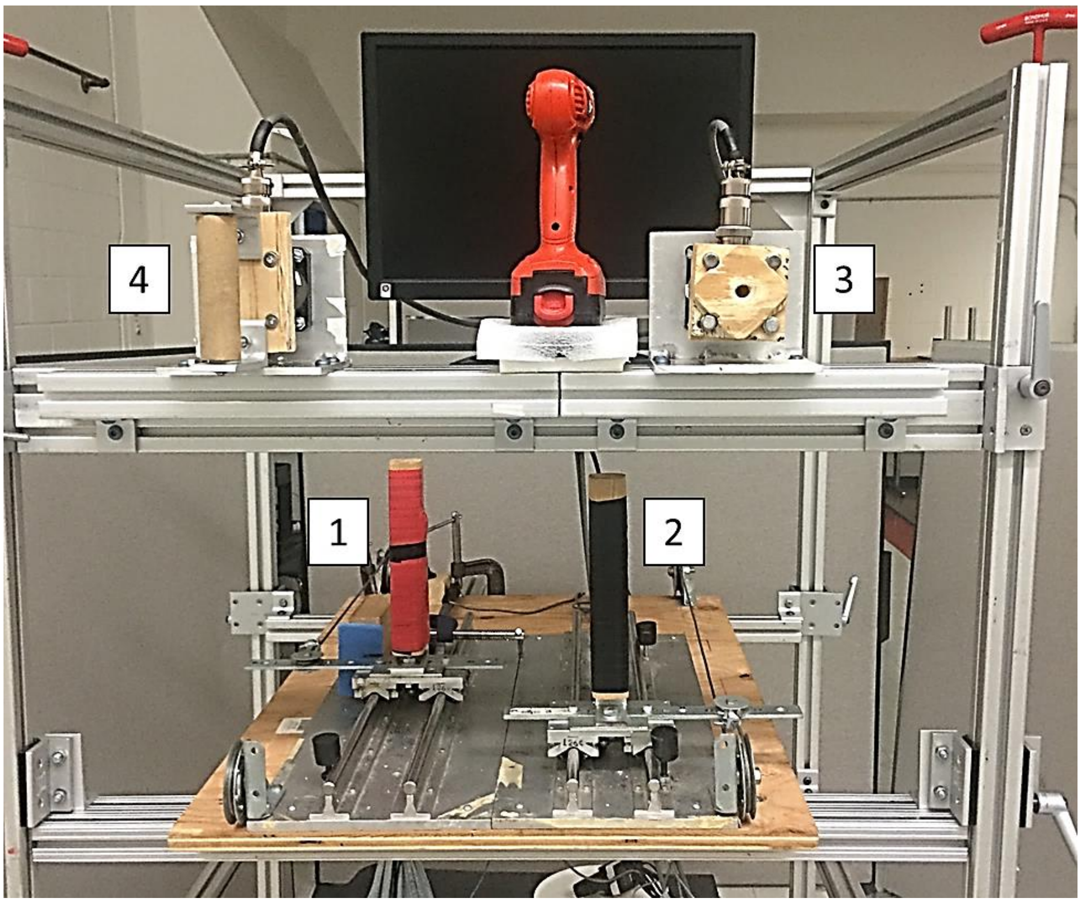

Figure 6.1: The work cycle apparatus consisted of four tasks: (1) pull task (60% MVC, 14 seconds), (2) push task (60% MVC, 14 seconds), (3) drill task (50% MVC, 10 seconds), and (4) up/down target task with 2 force levels (15%/35% MVC, 16 seconds). Each work cycle was 60 seconds long and participants continued working until one of the termination criteria were met.

6.3.3 Data Analysis and Statistics

Work cycles (WC) were broken down into the four work tasks, with the push and pull tasks further divided into their concentric and eccentric phases. Since participants completed different numbers of work cycles, 5 WCs were selected for the analysis for each participant (first, 25%, 50%, 75% endurance time, last WC). The target task was excluded from this analysis since targets were presented to participants randomly, and therefore the selected WCs for analysis are not the same targets between participants.

EMG data from the work cycles and the 4 submaximal reference measures were linear enveloped (dual pass, 2nd order BW filter, fc=4 Hz). Normalizing functions were created for each muscle using the EMG amplitude data from the submaximal reference exertions and least squared cubic regression models (McDonald et al, *Thesis Chapter 4: Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work*). Work cycle EMG data were normalized to these functions to mitigate the fatigue artifact on EMG amplitude. Summary variables were calculated (mean, coefficient of variation (COV)) for each muscle and each work cycle. Repeated measures ANOVAs were conducted on the five selected work cycles (first, 25%, 50%, 75%, last) for each muscle, variable and task. Post-hoc analyses were performed with Tukey's HSD tests.

A power spectral analysis was performed on the reference exertion EMG using a Fast Fourier Transformation. The median power frequency (MPF) was calculated (0.125 s sliding rectangular window and 0.0625 s window overlap). To quantify overall muscle fatigue, the change in reference exertion EMG amplitude and MPF across all fatigued muscles (increased EMG amplitude and decreased MPF during reference exertion) were combined to generate a multi-muscle fatigue score (MMFS) (McDonald et al, *Thesis Chapter 5: Using EMG amplitude and frequency to calculate a multi-muscle fatigue score and evaluate global shoulder fatigue*). To assess between participant differences in myoelectric fatigue development across muscles, individual muscle fatigue was quantified as an 8% or greater decrease in MPF relative to the baseline exertion. Changes in strength and RPE over time were assessed for the first, middle and last work cycles using repeated measures ANOVAs and Tukey's HSD post hoc tests.

For the kinematic analysis, marker data were exported to Matlab (Mathworks Inc., USA) and the upper extremity segments were modelled according to ISB standards (Wu et al, 2002; Wu et al, 2005). Three-dimensional joint angles were calculated for the wrist (hand relative to forearm), elbow (forearm relative to upper arm), shoulder (upper arm relative to trunk) and trunk (thorax relative to global axis), and dual-pass filtered (2^{nd} order, Butterworth filter, f_c=10 Hz).

Summary variables were calculated (mean, coefficient of variation (COV)) for each muscle and each work cycle. Repeated measures ANOVAs were conducted on the five selected work cycles for each joint angle, task and variable and post hoc analysis was performed with Tukey's HSD tests. All statistical tests were conducted in SPSS Statistics (v20.0, IBM, NY, USA) with $\alpha = 0.05$.

6.4 Results

Participants maintained task performance for 57-240 minutes of repetitive work. Changes in RPF, strength, and MMFS were evidence that the protocol was effective at inducing upper extremity fatigue. Ratings of perceived fatigue significantly increased between baseline (3.0±1.8) and the middle WC (6.7±1.5), and between the middle and final (9.4±0.7) WCs ($p<0.05$). Glenohumeral flexion strength significantly decreased over time, from the first (123.0±19.4 N) to the middle (104.9±24.0N), and last work cycles (80.2±29.3 N) ($p<0.05$). MMFS was significantly higher in the middle (49.3±6.4) and final WCs (59.8±8.8) than the first WC (33.2±3.59) ($p<0.05$). Posture and muscle activity changed over time as participants performed the work tasks. The specific changes in means and coefficient of variation were task, variable (joint angle, muscle) and time dependent, and are described in the following sections.

6.4.1 Drill Task

There were a number of significant changes in mean joint angles over time that were used to maintain task performance during the drill task. There was a main effect of time on humeral flexion angle, which trended to a lower flexion angle over time (Figure 6.2a). There were also main effects on external humeral rotation, leftward trunk lateral rotation, and trunk extension (Figure 6.2b) and wrist ulnar deviation (Figure 6.2c) ($p<0.05$). The muscular changes that accompanied these postural adaptations included a reduction in anterior deltoid activity, and increased posterior deltoid and infraspinatus activity (Figure 6.2d) ($p<0.05$). There was a main effect of time for middle deltoid activity ($p<0.05$).

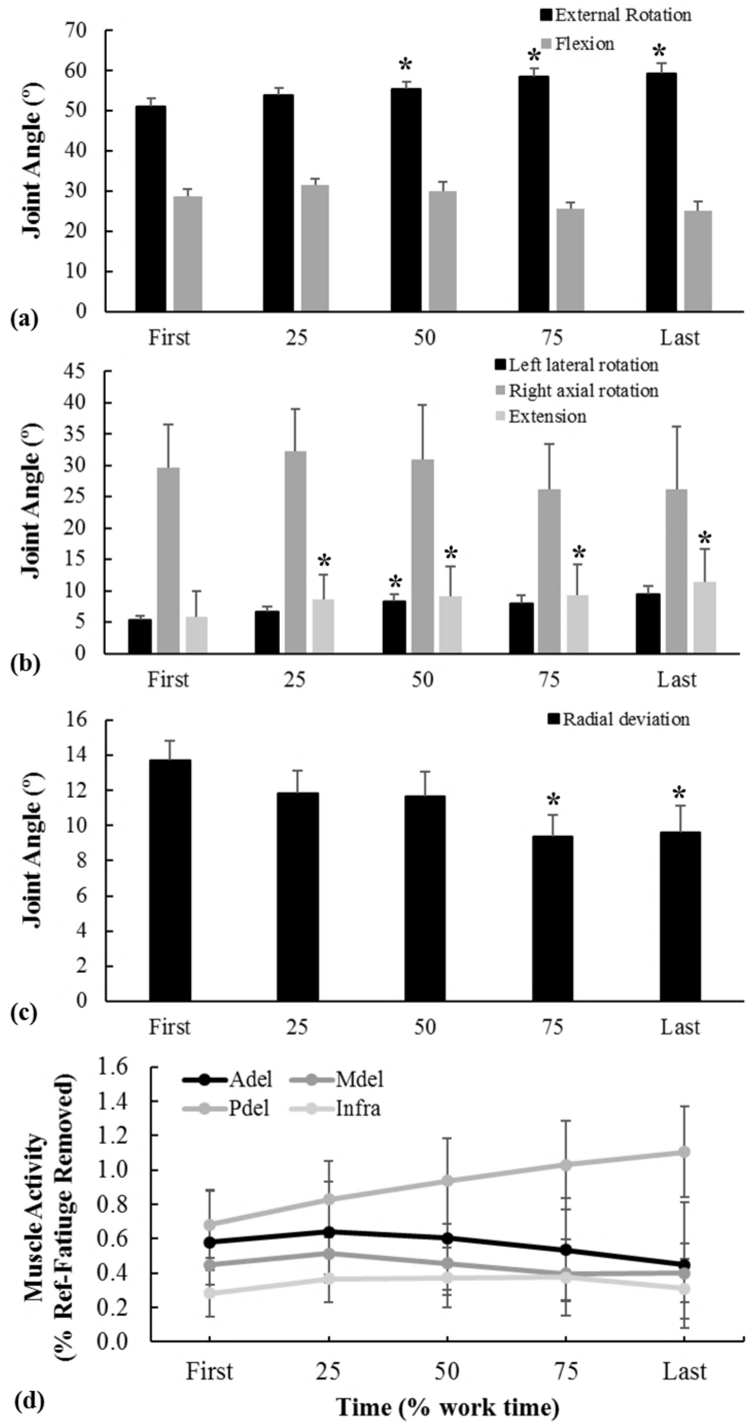

Figure 6.2: Significant changes in kinematics and muscle activity for the drill task. There were significant changes over time in (a) humeral angles, (b) thorax angles, (c) wrist angles and (d) muscle activity. Muscle activity changes are reflective of changes in muscular loading since EMG data were normalized to a function created to mitigate fatigue artifacts. Significant changes are denoted with *.

6.4.2 Pull Task

The pull task was divided into the concentric and eccentric phases of the movement for the analysis. Participants responded to the repetitive tasks by increasing elbow extension (Figure 6.3c) over time during the concentric phase of the task and decreasing it in the eccentric phase of the task ($p<0.05$). Over time, there was also an increase in leftward trunk lateral rotation (Figure 6.3b) during both phases of the task and increased humeral abduction during the concentric portion only (Figure 6.3a) ($p<0.05$). Muscle activity changes were similar in both phases of the pull task. When normalized to remove the fatigue artifact, there was a decrease in triceps activity that was compensated for by an increase in posterior deltoid activity ($p<0.05$) (Figure 6.3d,e). In the concentric phase of the pull task, there was a significant main effect of time on mean middle trapezius activity, with no significant post hoc tests. However, there were trends towards increased activity between the first and 25% WC, followed by decreased activity for the cycles after 25%. There was decreased biceps activity in the eccentric phase of the task only ($p<0.05$).

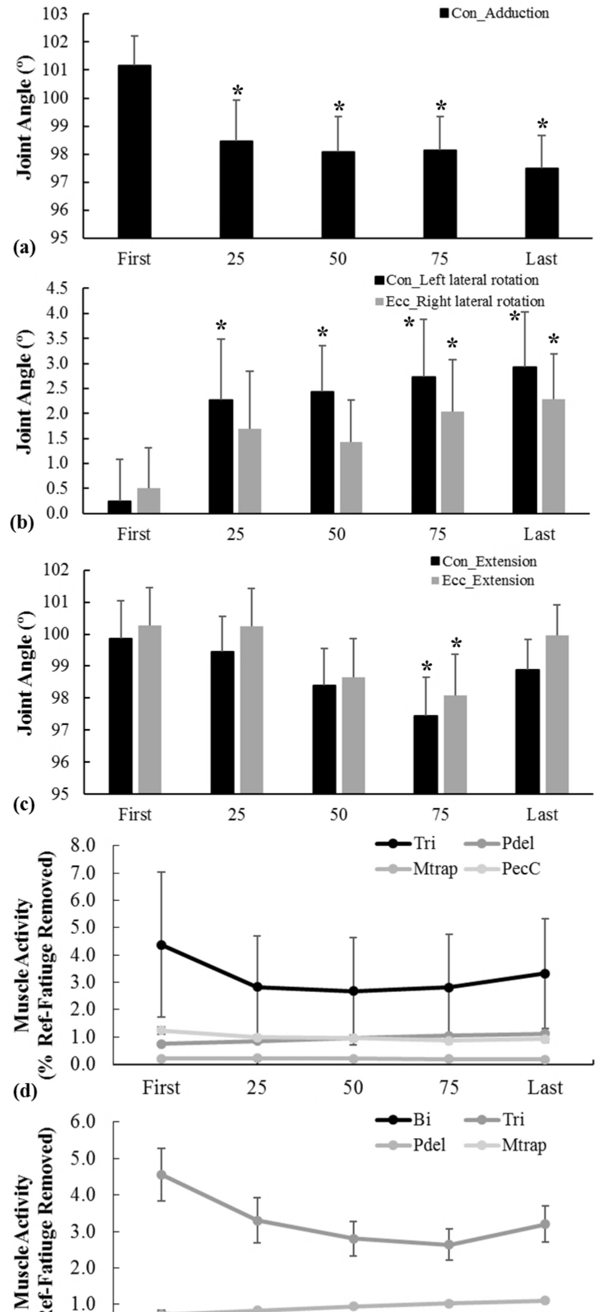

Figure 6.3: Significant changes in kinematics and muscle activity for the pull task. There were significant changes over time in (a) humeral angles, (b) thorax angles, (c) elbow angles (d) muscle activity in the concentric phase of the task and (e) muscle activity in the eccentric phase of the task. Muscle activity changes are reflective of changes in muscular loading since EMG data were normalized to a function created to mitigate fatigue artifacts. Significant changes are denoted with *.

6.4.3 Push Task

During the push task, the postural changes observed over time were similar between the concentric and eccentric phases of the task. During both phases, there was an increase in humeral flexion (Figure 6.4a), combined with a reduction in leftward trunk axial rotation (Figure 6.4b) ($p<0.05$). In the concentric phase (only), there was a main effect of time on humeral abduction (Figure 6.4a) without significant post hoc tests. Similarly, there were muscle activity changes consistent across both phases of the task (Figure 6.4c,d). Over time, triceps and posterior deltoid activity increased to maintain the performance of this task ($p<0.05$). In the eccentric phase only, there was a reduction in activity in the clavicular head of the pectoralis major ($p<0.05$). In the concentric phase, there was a main effect of time on mean middle deltoid activity.

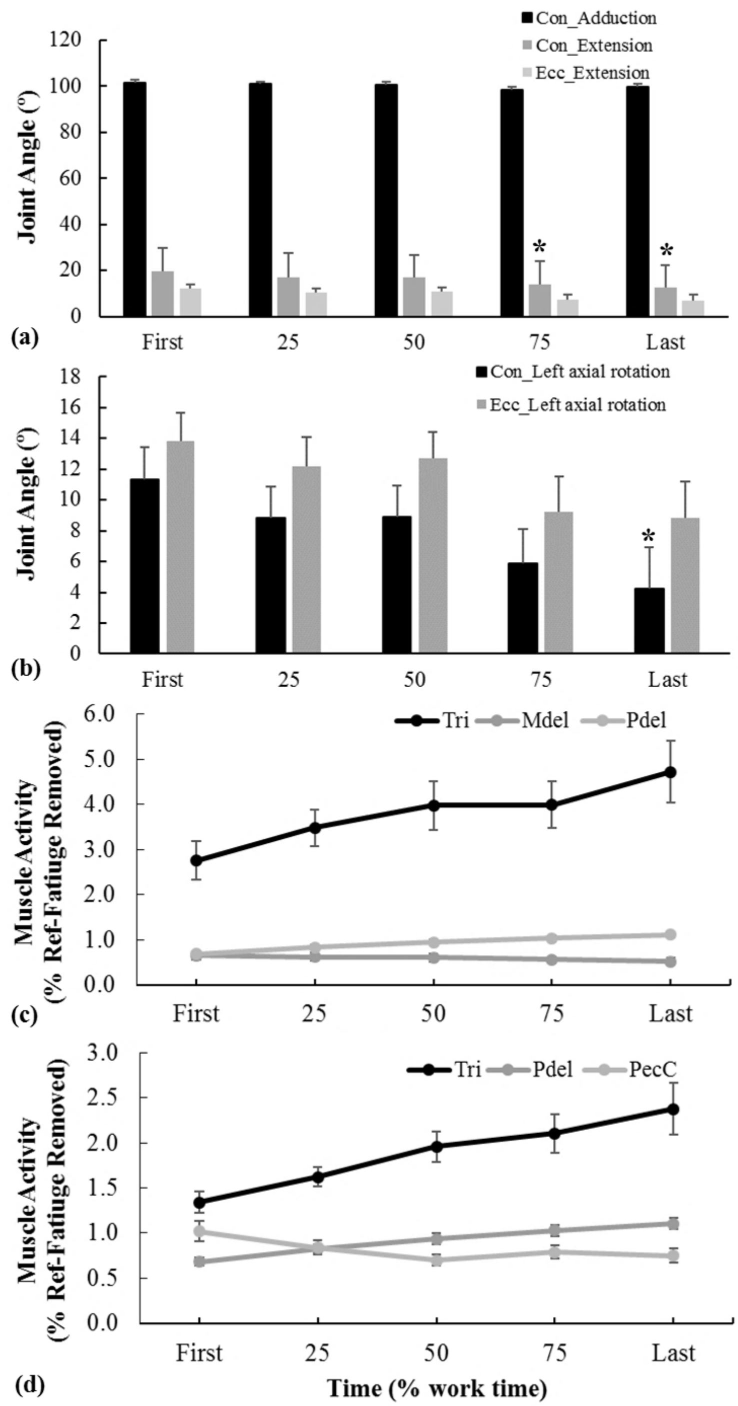

Figure 6.4: Significant changes in kinematics and muscle activity for the push task. There were significant changes over time in (a) humeral angles, (b) thorax angles, (c) muscle activity in the concentric phase of the task and (d) muscle activity in the eccentric phase of the task. Muscle activity changes are reflective of changes in muscular loading since EMG data were normalized to a function created to mitigate fatigue artifacts. Significant changes are denoted with *.

6.4.4 Variability

There was variability in task endurance and kinematic/muscular adaptations between- and within-participants. Task performance endurance time ranged from 57-240 minutes between participants. Between-participant variability was assessed with between participant coefficient of variation (COV), which was indicative of differences in posture and muscle activity strategies to perform the work tasks across individuals. Across the 10 joint angles and 5 time points evaluated, the between participant COV ranged from 2.9-1523% in the concentric phase of the push task, 3.1-10418.9% in the eccentric phase of the push task, 4.1-4984% for the concentric phase of the pull task, 4.0-7179.9% in the eccentric phase of the push task and 5.8-166% in the drill task. Across the 13 muscles, the between participant COV activity range from 23.7-348% in the concentric phase of the push task, 0.4-386% in the eccentric phase of the push task, 21.0-635.2% for the concentric phase of the pull task, 23.2-357% in the eccentric phase of the push task and 23.8-343% in the drill task. The variability in strategy was also displayed when and in which muscles participants developed myoelectric fatigue (≥8% decrease in MPF between submaximal reference exertions) (refer to Appendix K in *Thesis Chapter 4: Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work*). There was also evidence of within-participant variability, as indicated by significant changes in within participant coefficients of variation. Postural variability tended to increase between the first and final two analyzed work cycles (75 and 100% work time). In the drill task, there was increased variability in humeral ab/adduction and internal/external rotation angles, and thorax axial rotation ($p<0.05$). In the pull task, there was increased variability in the elbow flexion/extension, humeral internal/external rotation and flexion/extension angles, and thorax axial rotation during the concentric phase of the task with time. During the eccentric phase, there was increased variability in only the humeral ab/adduction angle. There were no changes in postural variability in either phase of the push task. Significant changes in muscle activity variability tended to decrease over time. There were main effects of time on the clavicular head of the pectoralis major (drill task), latissimus dorsi (pull task (concentric and eccentric phases)), and biceps (pull task (concentric phase only)). There were decreases in posterior deltoid activity variability in both phases of the pull task and the concentric phase of the push task ($p<0.05$). The only significant increase in muscle activity variability was between the first and middle work cycles in the lower trapezius during the eccentric phase of the push task ($p<0.05$).

6.5 Discussion

The goal of this investigation was to build on previous work examining the response to a fatigue protocol by simultaneously evaluating muscular and kinematic adaptations of the upper extremity during fatiguing, repetitive work. As we hypothesized, participants used coordinated compensation strategies to maintain task performance for 57-240 minutes. Participants used variable but coordinated adaptations that allowed muscles to fatigue and recover during the trial. There was evidence of both between- and within-participant variability in these responses over time. Postural and muscular changes were dependent upon task and time.

There were coordinated adaptations over time that allowed participants to maintain performance of the drilling task as muscles fatigued and recovered. Postural changes during this task included a reduction in humeral flexion angle that was compensated for by external humeral rotation, ulnar deviation, and increased trunk extension and leftward lateral rotation. All together, these four changes could help raise the right upper extremity to continue to grasp and elevate the drill to the required position as the shoulder flexion angle decreased. These changes may have been the result of fatigue developing in the anterior deltoid muscle, evidenced by reduced activity in this muscle over time. Previous work in our laboratory (Tse et al, 2016) examined the kinematic and muscular response to an anterior deltoid fatigue protocol during repetitive work tasks. The drilling task in that investigation was the same as the current study, allowing comparisons of the responses. Interestingly, both protocols resulted in fatigue-induced reductions in humeral flexion during the drill task. In the current work, there were significant, compensatory postural changes that compensated for the reduced flexion angle. Comparatively, when a fatigue protocol induced anterior deltoid fatigue, although not significant, there were trends towards the same changes. Fatigue protocols are a common method to examine the response to muscle fatigue (Ebaugh et al, 2006a,b; Chopp et al, 2011; Chopp-Hurley et al, 2016; Côté et al, 2002; Côté et al, 2005; Joshi et al, 2011; McDonald et al, 2016; Tsai et al, 2003; Tse et al, 2016). This method has several advantages in that it is time efficient, easily standardized, and can be used to assess any work task. This comparison suggests that the effects may be more salient when fatigue is task-induced; however, the overall message obtained from the postural findings may be similar.

The pushing and pulling tasks were quite constrained and were physically challenging (60% MVC). This likely afforded fewer opportunities for postural adaptations than the drilling task. Despite these constraints, there were still fatigue-induced postural changes observed during these tasks. As opposed to the drilling task, during the pushing task, there was a reduction in leftward trunk lateral rotation and an increase in humeral flexion, maintained through both phases of the task. These findings are in line with previous studies observing kinematic adaptations with repetitive work. Côté and colleagues (2002, 2005) examined repetitive hammering and sawing tasks and found coordinated, multi-joint kinematic changes following fatigue protocols. Similarly, kinematic adaptations throughout the upper extremity joints have been observed during repetitive, light assembly tasks in which participants appeared to utilize the ability to modify wrist and elbow angles to compensate for fatigue in their shoulders (Qin et al, 2014). The current study shows that these adaptations also develop during heavier, repetitive work. In conjunction with the current work, these studies show individuals are able to make muscle fatigue-induced postural adaptations in order to maintain task performance during repetitive work, with specific changes being workplace and task dependent.

In conjunction with postural adaptations, there were several muscle activity adaptations over time. Interestingly, once the effects of fatigue on the EMG amplitude were removed (through the normalizing method), posterior deltoid activity still increased over time in all 3 tasks. Increased activation of this muscle was also observed in previous work (Tse et al, 2016). However, EMG in that investigation was normalized to initial MVE, making it unclear to what extent the observed activity changes were the result of fatigue artifact or increased muscle force. The normalization method used in the current work allows us to interpret these changes as due to an increase in the contribution of this muscle to the performance of tasks over time. During the tasks, as other muscles begin to fatigue, the posterior deltoid appears to compensate for the reduced capacity in other muscles to maintain the task performance requirements. In conjunction with increased activity in this muscle, there was also less variability in the posterior deltoid during the pulling task and in the concentric phase of the push task. To evaluate the relationship between the posterior deltoid and endurance time, a Pearson Product Moment correlation was run for each task on the change in posterior deltoid activation between the first and last WC and endurance time. There was a significant correlation between the change in posterior deltoid muscle activity and endurance time in both phases of the pulling task ($r=0.617$ (con), $r=0.482$ (ecc). Although the current study was not designed to fatigue the anterior deltoid, the tasks required GH flexion and the posterior deltoid appears to be an important muscle in maintaining this posture as other muscles develop fatigue.

There were also differences in several types of variability throughout the analysis. Although the tasks were scaled to each participant's anthropometrics and strength, some participants had four times greater endurance time than others. Mental imagery of an endurance task has been shown to increase muscle activity and reduce performance on the physical performance of the endurance task (Graham et al, 2014). These findings suggest that interactions between mental and physical factors may have contributed to the number of work cycles each participant was able to complete. The between-participant variability in posture and muscle activity may have also contributed to these differences in endurance time. The anatomy of the shoulder complex affords many ways to complete functional tasks at the hand. Variability between individuals is a common challenge in investigations of upper extremity repetitive work. In a repetitive sawing task, some participants responded to the demands of repetitive work by increasing their movement amplitude and speed, while others, had the opposite response (Gates & Dingwell, 2008). In a repetitive pointing task, investigators associated motor control differences between individuals with their muscle activity patterns (Bosch et al, 2012). Participants employed different movement/muscular strategies at the start of these tasks, which led to differences in myoelectric fatigue across individuals and as a result, variability in postural and muscular adaptations to fatigue. Within subject variability may be participant dependent as well. In a study examining pipetting tasks, it was found that some individuals had variable movement patterns while others did not (Sandlund et al 2017). In the workplace, although individuals are exposed to the same workplace demands, injuries are often sustained by only certain workers (Kilbom & Persson, 1987). The between-participant variability observed during these simulated workplace tasks may help explain the highly differential risk/incidence of injuries across individuals.

There were limitations to this work that should be considered when interpreting and applying the results. All of the participants included in this investigation were men, which may limit applications to female employees in the workplace. There were also only four work tasks examined: push, pull, drill, target matching. Although the specific joint angle and muscle activity changes are not directly applicable to a specific workplace, the fact that participants were able to adapt to multiple, constrained tasks and maintain their task performance suggest that this would happen in the workplace as well. Statistical limitations prevented the inclusion of all the work cycles in the analysis. Selecting five work cycles throughout the protocol was not inclusive of all of the data; however, it gave an indication of how task performance and fatigue varied throughout repetitive work. All of the work tasks in this investigation were dynamic in nature, thus allowing for simultaneous kinematic and muscular adaptations. Although this is realistic to the workplace, it is unclear if muscle activity changes can occur in the absence of the kinematic changes that alter the moment demands at each joint. Future work examining static work tasks would aid in better understanding the complex, multi-faceted response.

6.6 Conclusion

Participants were able to maintain task performance in the presence of myoelectric fatigue, perceived physical fatigue, and reduced physical capacity, for 57-240 minutes. Over time, they employed kinematic and muscular adaptation strategies that were task and time dependent. These findings suggest that the adaptations in the workplace will also be dependent on the specific task demands. The posterior deltoid appears to be an important muscle in facilitating these adaptive strategies and future work should investigate the contributions of this muscle further. Variability between participants may aid in explaining why only certain workers develop musculoskeletal disorders when exposed to the same workplace physical demands as other individuals.

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CHAPTER SEVEN

The multi-faceted response to static, repetitive work in the shoulder

Alison C. McDonald, Denver M.Y. Brown, Steven R. Bray, Peter J. Keir

Department of Kinesiology, McMaster University, Hamilton, Ontario, Canada

*Corresponding Author:

Peter J. Keir, PhD
McMaster University
Department of Kinesiology
Ivor Wynne Centre, room 212
1280 Main Street West
Hamilton, ON, Canada, L8S4K1
Telephone: 905-525-9140 ext. 23543
Email: pjkeir@mcmaster.ca

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7.1 Abstract

It is a challenge to understand the multi-faceted effects of fatigue in repetitive and prolonged workplace tasks. The purpose of this investigation was to evaluate the effects of static, repetitive work on the load sharing strategies between shoulder muscles, subacromial space width and perception of fatigue and affective valence. Electromyography was measured from 13 muscles while participants completed static, repetitive tasks until exhaustion or refusal to continue. Every 12 minutes, reference measures were completed to quantify fatigue, subacromial space, perceived physical and mental fatigue, and affective valence. Over time there was evidence of myoelectric, perceived mental fatigue and perceived physical fatigue ($p<0.05$). Subacromial space changes varied between individuals and load sharing between muscles changed over time allowing participants to maintain task performance in the presence of reduced muscle capacity ($p<0.05$). These findings support a holistic approach to workplace assessments to capture the multi-faceted and variable response to repetitive demands.

Keywords: Shoulder, Electromyography, Fatigue, Repetitive work, Static work

Practitioner Summary

Fatigue is multi-faceted and assessment of workplace tasks should incorporate both physical and mental constructs to better understand the effects of repetitive work tasks. Load sharing occurs between muscles to maintain task performance, and individual assessments can capture the variable response to these tasks between workers.

7.2 Introduction

Coordinated function of the shoulder muscles during repetitive and fatiguing work is essential for the full function of the shoulder complex in the workplace. Load sharing between muscles appears to be relatively consistent between people at low contraction intensities, slow movement speeds, and within a limited range of motion (Laursen et al, 1998). With fatigue, load sharing has been shown to change in the forearm, lower leg, and respiratory muscles, and in some instances, occurred without changes in motion (Bonnard et al, 1994; Duchêne & Goubel, 1990; Lucidi & Lehman, 1992; Roussos et al, 1979). Given that the anatomy of the shoulder allows multiple ways to distribute loads between the muscles, load sharing adaptations to fatigue should be expected. In a shoulder simulation that acted to paralyze teres minor and posterior deltoid, the surrounding musculature, including the supraspinatus, triceps, middle deltoid and latissimus dorsi, compensated with increased activity (Crouch et al, 2013). A similar response has been shown experimentally when a suprascapular nerve block of the infraspinatus and supraspinatus muscles resulted in an increase in deltoid activity during humeral abduction (McCully et al, 2007). Load sharing has also been observed in less traumatic examples of physiological muscle fatigue in primary movers. A fatigue protocol directed at the serratus anterior resulted in increased trapezius activity during humeral elevation (Szucs et al, 2009), while a fatigue protocol for the infraspinatus altered trapezius activity during humeral elevation and lowering (Joshi et al, 2011). These examples during simple movements suggest that load sharing likely also exists during repetitive work, which would have implications for the processes of fatigue and recovery.

Upper extremity muscle activity during repetitive and prolonged work is an area of interest in ergonomic research. The superficial location and propensity for injury of the upper trapezius has made it the focus of many of these investigations (Farina et al, 2008; Holtermann & Roeleveld, 2006; Jensen & Westgaard, 1997; Mathiassen & Winkel, 1996; Palmerud et al, 1995; Samani et al, 2010; Thorn et al, 2007). When participants used visual feedback to reduce upper trapezius activity, muscle activity increased in the serratus anterior, anterior deltoid, infraspinatus and other sections of the trapezius (Palmerud et al, 1995; Palmerud et al, 1998). In an endurance study, individuals with larger shifts in activity throughout the muscle, measured with a multi-electrode array, had greater endurance time during static exertions than those with more uniform activation patterns (Farina et al, 2008). These changes in muscle activity patterns suggest that a neuromuscular control strategy may be employed to redistribute the load within a muscle and recruit synergistic muscles to provide opportunities for recovery. While these compensation strategies have been demonstrated, a meta-analysis of endurance times across the main joints in the body found the shoulder complex to be the most fatigable, with a nonlinear relationship between endurance time and task intensity (Frey-Law & Avin, 2010). The trapezius is an important muscle related to injury development, however, activity from more muscles is needed to understand function of the shoulder complex. During sustained exertions, recovery from fatigue has been modeled to occur 10-15 times slower than muscle fatigue (Frey-Law et al, 2012). The long recovery time for fatigued muscles has implications for postural muscles in the shoulder complex that often may not have adequate rest, making the prevention of muscle fatigue in the workplace difficult (Lucidi & Lehman, 1992).

The rotator cuff musculature is essential for maintaining shoulder stability and function, thus its response to repetitive work and fatigue is integral. A possible outcome of rotator cuff muscle fatigue is a reduction in subacromial space (SAS). The subacromial space lies between the acromion and the humeral head and it encompasses rotator cuff tendons; reduced SAS width is associated with rotator cuff pathologies, such as supraspinatus tears (Banas et al, 1995). The width of the subacromial space is highly variable, affected by gender, lateral acromial angle, acromial shape, arm position, scapular rotation, and muscle activity (Banas et al, 1995; Graichen et al, 2001; Graichen et al, 2005; Chopp et al, 2012). Muscle activity affects the subacromial space through its effect on scapular kinematics and superior humeral head translation. Previous work in our laboratory has shown scapular kinematic changes that may increase the SAS during simulated repetitive work following a fatigue protocol (Tse et al, 2016); however, the effect of these changes on the width of the space was not quantified. Chopp et al (2010) examined two fatigue protocols and both induced scapular changes that also may have been protective of SA impingement. Tsai et al (2003) also found changes in scapular kinematics with fatigue and a relationship between fatigue (decrease in strength) and posterior tilt. Abduction and deltoid muscle activity can cause superior translation of the humeral head, which also impacts SAS, if the rotator cuff muscles do not provide sufficient inferior translations to resist it (Chen et al, 1999; Chopp et al, 2010; Graichen et al, 2005). By measuring scapular kinematics alone, especially considering each axis individually, only speculations can be made about the effects on SAS width. Chopp-Hurley et al (2016) measured SAS width before and after fatigue, but for safety, were limited to a small number of x-rays for each participant. Ultrasound (US) imaging can be used to measure the subacromial space in vivo and without any added risk from multiple measures. Excellent inter- and intra-rater reliability of the US measure of acromiohumeral distance (AHD) on a phantom shoulder has been found (coefficient of variation < 3%), suggesting that this tool can be used effectively to assess changes to the space with muscle fatigue (McCreesh et al, 2014).

Fatigue is multi-faceted and previous work suggests that performance during repetitive work may not only be limited by physical constraints but may have a psychological component as well. McDonald et al (2017; *Thesis Chapter 6-Muscular and Kinematic Adaptations to Fatiguing, Repetitive Work*) examined repetitive work until exhaustion, and although tasks were scaled to each participants anthropometrics and strength, some participants had four times greater endurance times than others. Psychological factors to consider include perceived mental and physical fatigue, affective valence, self-control and fatigue state. Mental workload has been shown to reduce endurance times while performing moderate physical workload tasks and can cause greater strength decrements than performing physical work alone (Mehta and Agnew, 2012). In a recent meta-analysis, Clarkson et al (2016) revealed that mental fatigue stemming from tasks requiring cognitive control has a medium to large effect on subsequent self-control task performance. Self-control refers to the capacity to resist temptations by regulating attention, emotion and behavior (Baumeister and Vohs, 2016), and is associated with many positive outcomes in life (Tangney and Boone 2004). Positive affect has been examined in workplace for many years and is generally associated with positive outcomes, including goal setting and action, work engagement and better decision making and conflict resolution and reduced absence from work (Barsade and Gibson 2007). The Circumplex Model of Affect measures two basic dimensions of affect (affective-valence (pleasure-displeasure) and perceived activation) that when combined represent different affective states (Ekkekakis et al 2008). Combining these psychological measures with more common biomechanical variables impacted by fatigue may help to explain some of the inter-individual variance in the response to fatiguing, repetitive work. The purpose of this investigation was to evaluate the multi-faceted effects of static, repetitive work on the load sharing strategies between the muscles of the shoulder, subacromial space width, perception of fatigue and affective valence. We hypothesized that during static, repetitive work tasks there would be evidence of load sharing between the shoulder muscles, that the SAS would decrease over time and that repetitive work would cause perceived mental and physical fatigue, negative affect and a reduction in self-control.

7.3 Methods

7.3.1 Participants

Right hand dominant men (n=20), free of upper extremity injury or pain in the last year were recruited to participate in this study. The study was approved by the Hamilton

Integrated Research Ethics Board (HIREB) and participants provided informed, written consent prior to beginning the protocol.

7.3.2 Instrumentation

Muscle activity was recorded from 13 muscles using surface electromyography (EMG) (Trigno, Delsys Inc., Natick, MA, USA). Electrodes were silver-contact wireless bipolar bar electrodes with fixed 1 cm inter-electrode spacing and signals were differentially amplified (CMRR > 80 dB, input impedance $10^{15}\Omega$), band-pass filtered (20-450 Hz), and 16-bit converted at 1926 Hz (± 5 V range). Muscles were all on the right side and included: biceps brachii (Bi), triceps brachii (Tri), anterior (Adel), middle (Mdel) and posterior (Pdel) deltoids, infraspinatus (Infra), upper trapezius (Utrap), middle trapezius (Mtrap), lower trapezius (Ltrap), latissimus dorsi (Lats), serratus anterior (Sert), sternal head of pectoralis major (PecS), and clavicular head of pectoralis major (PecC). Ten maximal voluntary exertions were performed twice with 2 minutes of rest between each repetition (McDonald et al, 2017).

The apparatus was instrumented with a 6 degree-of-freedom force transducer (MC3-500, AMTI, Watertown, MA, USA) attached to a wooden handle and participants were seated on a height adjustable chair placed on a force plate (OR6-5-1, AMTI, Watertown, MA). The apparatus was adjusted such that participants gripped the handle with a power grip, with trunk upright at 90°, arm positioned in 60° glenohumeral (GH) flexion, slight GH abduction, and 90° elbow flexion. Posture was measured with a manual goniometer and tracked throughout the protocol using electromagnetic sensors (FASTRAK, Polhemus, Colchester, VT, USA) placed on the wrist, elbow, shoulder, and trunk. Two static maximal voluntary contractions (MVC) were completed in 4 directions (up, down, push, pull) with 2 minutes of rest between each MVC. The mean MVC for each direction was used to scale 4 work tasks to 30% MVC.

A sonographic system (Vivid Q BT10, GH Healthcare, Milwaukee, WI) was used to image the superior aspect of the humeral head and the inferior aspect of the acromion in two seated postures (neutral: 0° GH abduction, abducted: 60° GH abduction), confirmed with a manual goniometer. Two postures (neutral, abducted) have been selected to see if there is an interactive effect of fatigue and glenohumeral elevation on the subacromial space. The subacromial space (SAS) was defined as the tangential distance from the humeral head to the edge of the acromion on the longitudinal sonogram and measured by placing a 12 MHz linear probe (i12L) along the longitudinal axis of the humerus on the lateral aspect of the shoulder (Desmeules et al, 2004). For measurement repeatability, the probe location was outlined on the participant's arm in each posture and a custom probe holder was attached to the ultrasound probe (Figure 7.1). A digital inclinometer (Bubble Level App v. 1.7 on iPhone 5s, Apple Inc. CA, USA) was attached to the probe holder and used to ensure a consistent probe angle between repeated SAS measurements.

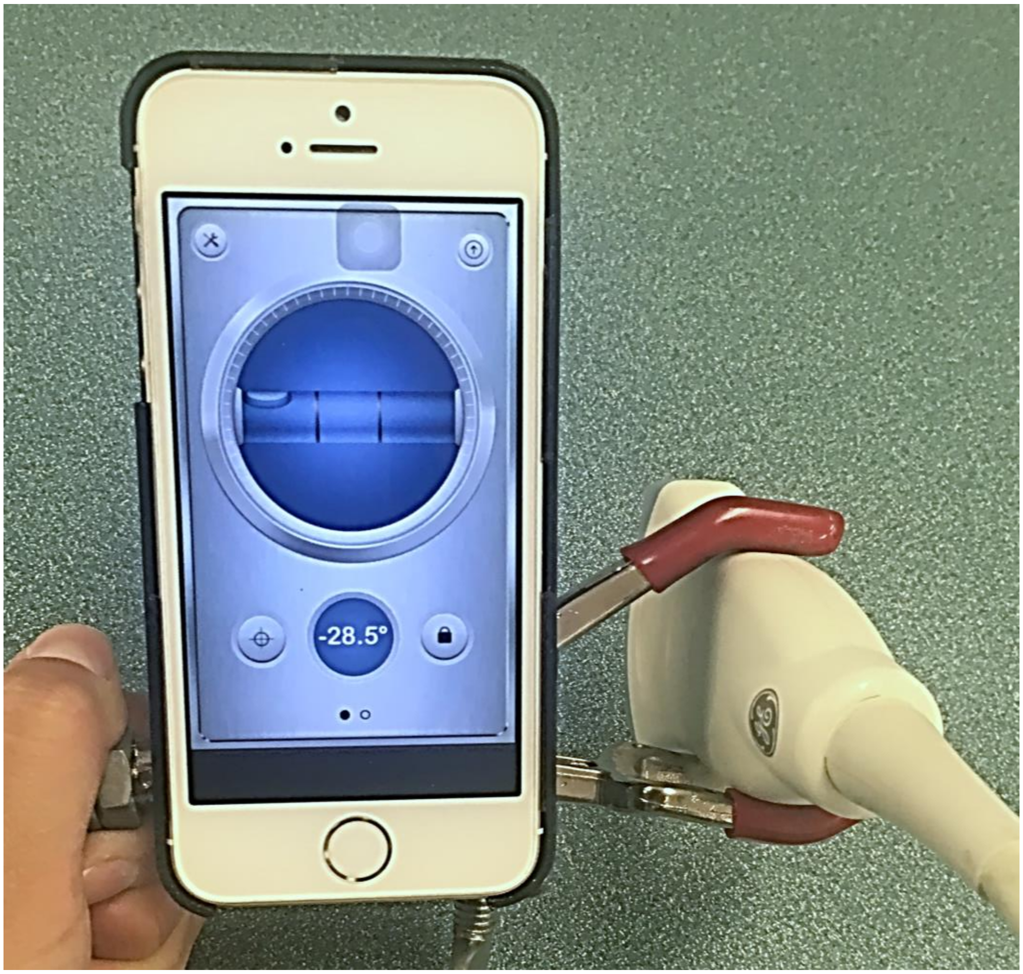


Figure 7.1: A custom US probe holder was created to fix a digital inclinometer to the US probe, allowing a consistent probe angle between repeated images to ensure measured difference in SAS were the result of fatigue and not from measuring different anatomical landmarks.

7.3.3 Protocol

Participants began by completing 3 questionnaires, the Brief Self-Control Scale (SCS, Appendix J.1) (Tangney et al, 2004) and State Self-Control Scale (SCCS, Appendix J.2) (Ciarocco et al, 2007) to evaluate baseline self-control, and the Fatigue State Questionnaire (FSQ, Appendix J.3) (Greenberg et al, 2016) to evaluate fatigue. Following set up, baseline reference measures were recorded. The reference measures included 4 submaximal exertions, 1 maximal exertion, 4 perceptual questions, and 2 ultrasound SAS measures (described in detail below). Next, participants began 12 minutes of static work. Each work cycle was 60 seconds long and was comprised of 4 static tasks (up, down, push, pull) at 30% MVC. Each task was 12 seconds long with 3 seconds of rest. All tasks were completed in the static posture (60° glenohumeral (GH) flexion, slight GH abduction, and 90° elbow flexion). If participants moved more than 5° or 5 cm, they were verbally coached back to their initial posture. Immediately following the 12 minutes of static work, participants completed the reference measures and resumed another 12 minutes of work as quickly as possible. This cycle continued until one of four termination criteria were met: (1) participant declared they could no longer continue, or participant was unable to maintain (2) any one of the submaximal reference exertions, (3) any one of the work tasks, or (4) the required posture. Following termination of the work tasks, participants completed a final set of reference exertions and completed the SCCS and FSQ questionnaires for a second time.

7.3.4 Reference Measures

Reference measures were taken at baseline, every 12 minutes throughout the protocol and after the final work cycle (McDonald et al 2017-Thesis Chapter 4: *Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work*). The four submaximal exertions were 5 seconds long and completed at 30% MVC, a force transducer (Mark-10, Copiague, NY, USA) and manual goniometer were used to confirm force and posture for each exertion. These exertions were used to normalize EMG and evaluate myoelectric fatigue over time. To evaluate changes in strength, participants performed one maximal glenohumeral flexion exertion (90º GH flexion, 180º elbow extension). Subacromial space width was measured in two postures to evaluate changes in the space width over time. Each set of reference measures was completed with 4 perceptual questions. Participants were asked to rate their perceived physical fatigue (RPF) on a 0-10 scale (Appendix J.4) and their perceived mental fatigue using a visual analogue scale (VAS) by placing an 'X' on a 100 mm line with anchors ranging from 0 (none at all) on the left hand side to 100 (maximal) on the right hand side (Wewers and Lowe, 1990). Scores were calculated by measuring the distance (in millimeters) that the 'X' was placed from the left side of the scale. To generate the Circumplex Model of Affect, participants also rated their feelings (FS) on a -5 (very bad) to +5 (very good) scale (Hardy & Rejeski, 1989) (Appendix J.5) and their arousal (FAS) from 1 (low arousal) to 6 (high arousal) (Svebak & Murgatroyd, 1985) (Appendix J.6).

7.3.5 Data and Statistical Analyses

All EMG data were linear enveloped and dual-pass filtered (2nd order BW, $f_c = 4$ Hz). For each muscle, the central 3 seconds of the submaximal reference exertions were used with a least squares regression model to create a cubic function to predict the submaximal EMG amplitude for every 3rd minute. The work task EMG were normalized to the predicted amplitude from the corresponding time point. This was done to mitigate fatigue artifact over time, allowing us to distinguish between load sharing changes between muscles over time and changes in EMG from myoelectric fatigue artifacts (McDonald et al 2017-*Thesis Chapter 4: Submaximal Normalizing Methods to Evaluate Load Sharing Changes in Repetitive Work*). Mean, coefficient of variation (COV) and range variables were calculated for each muscle at 20% endurance time intervals. Repeated measures (RM) ANOVAs were used to evaluate differences in EMG summary variables (mean, COV, range) over time at 1, 20, 40, 60, 80, 100% of endurance time for each muscle and post hoc analyses were performed with Tukey's HSD tests. Based on muscle function, muscles were grouped for each task (Table 7.1) and Pearson product moment correlations were done between all muscle pairs within the task specific muscles groups.

A power spectral analysis was done using a Fast Fourier Transformation and the median power frequency (MPF) was calculated (0.125 s sliding rectangular window and 0.0625 s window overlap) on the central 3 seconds of the submaximal reference exertions. Myoelectric fatigue across all the muscles included in this investigation was quantified with a multi-muscle fatigue function, accounting for changes in EMG frequency, amplitude, the number of muscles included in the analysis and the number of these muscles exhibiting signs of fatigue (both an increase in EMG amplitude and decrease in EMG frequency) (McDonald et al-*Thesis Chapter 5: Using EMG amplitude and frequency to quantify global muscle fatigue across multiple muscles: A shoulder example*). The changes in myoelectric fatigue score over time and in glenohumeral flexion strength were evaluated with a RM ANOVA and Tukey's post hoc tests between the first, middle and final work cycle fatigue scores.

Table 7.1: Based on their function the measured muscles were grouped by task and posture. Muscles with multiple functions were included in all relevant groups.

Task	Muscles
Up	Biceps, anterior and middle deltoid, upper trapezius, sternal and clavicular heads of pectoralis major
Down	Latissimus dorsi, posterior deltoid, triceps
Push	Triceps, anterior deltoid, sternal and clavicular heads of pectoralis major
Pull	Biceps brachii, middle trapezius, infraspinatus, latissimus dorsi
Posture	Anterior, middle and posterior deltoids, infraspinatus, upper trapezius, serratus anterior, sternal and clavicular heads of pectoralis major

Ultrasound measurements were made offline using onscreen calipers on EchoPAC Software (General Electric Healthcare) with 0.01 mm precision. SAS was measured as the shortest two-dimensional, linear distance between the humeral head and the anterior-inferior tip of the acromion (Michener et al, 2015). One researcher who was blinded to the image order made all measurements. Each image was measured twice, on separate days, and the average of the two measurements was used (Michener et al, 2015). Images were removed from the analysis if visible landmarks differed from the baseline image, indicating a change in view. Probe position and angle were consistent between measurements; however, there were instances where participants posture changed during the reference measures as they fatigued, leading to the removal of these images. To evaluate between day reliability of the two within posture measurements, mixed model ICC with 95% confidence intervals and Chronbach's alpha values were assessed. To evaluate changes in SAS width over time, repeated measures ANOVAs and Tukey's HSD post hoc tests were performed between the 1st, middle and last work cycles in the neutral and abducted views. A paired samples t-test was used to evaluate the effect of posture on SAS.

The scores for the negatively phrased questions on the State SCCS (1-4,6-7,9-10) and the SCS scale (1-4,6,8-9,11-12) were reversed and single scores were calculated for each questionnaire by summing the responses and dividing by the total number of questions. Each question on the FSQ was analysed individually. To evaluate changes in perceived self-control and fatigue state before and after the work task protocol, paired sample t-test were used between the pre and post FSQ and SCCS scores. Pearson product moment correlations were used to examine the relationship between the pre-fatigue SCCS, SCS and FSQ scores and endurance time. The RPF, MF, FS and FAS ratings were fit to the best polynomial (1-7) based on r2 values (0.033-1), specific to each variable and each participant. This was done to allow predicted points at 20% endurance time internals, since these variables were only measured every 12 minutes and participants had 3-12 scores each. The effect of time on FS, FAS, RPF and MF was evaluated with repeated measures ANOVA's and Tukey's HSD post hoc tests. The Circumplex Model of Affect was created using mean scores from the FS (x-axis) and FAS (y-axis) scales. Pearson product moment correlations assessed the relationship between the rate of change in RPF and MF and endurance time. All statistical tests were conducted in SPSS Statistics (v20.0, IBM, NY, USA) with $\alpha = 0.05$.

7.4 Results

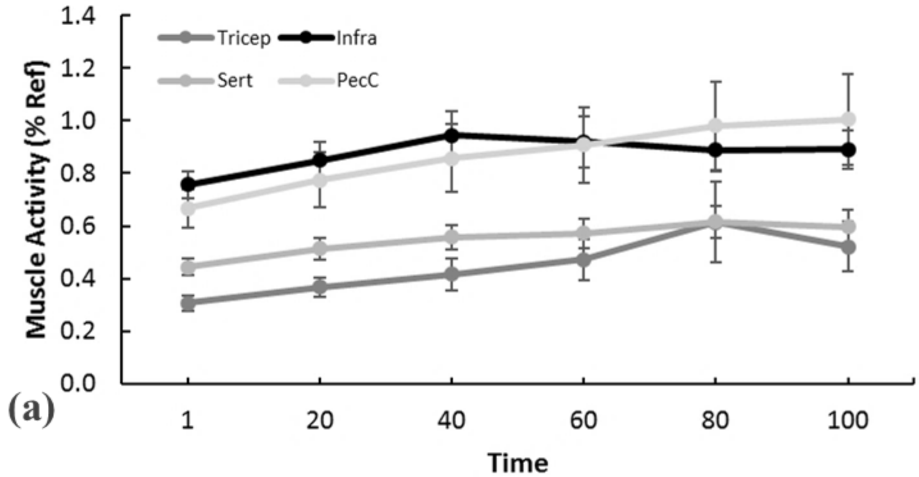
Participants were able to maintain task performance and the required static posture for 36 to 150 minutes of static, simulated work. Following the repetitive work protocol, participants showed signs of increased multi-muscle myoelectric fatigue (MMFS) between the first (43.8 ± 22.7) and last (89.6 ± 20.4) work cycles and a significant decrease in glenohumeral flexion strength (pre: 105.2 ± 77.8 N, post: 77.8 ± 20.6 N) ($p<0.05$). The specific effects of static, repetitive work on muscle activity, subacromial space and perceived fatigue and affect are described below.

7.4.1 Muscle Activity Changes During Static Work

Performing the static, repetitive work tasks had significant effects on summary muscle activity variables (mean, COV, range) that were dependent on task, time and muscle (Table 7.2) ($p<0.05$). During the up task, there was a significant increase in bicep variability between the first and the last work cycle. Once the fatigue artifact was removed, there were significant increases in mean EMG activity of the triceps, infraspinatus, serratus anterior and the clavicular head of the pectoralis major (Figure 7.2a). The range in muscle activity during the task also increased in the serratus anterior during this task ($p<0.05$). While pushing down, posterior deltoid activity variability increased between the first and last work cycle. There were significant changes in mean upper trapezius activity, range of latissimus dorsi activity, and both mean and range in the bicep muscle ($p<0.05$) (Figure 7.2b). During the push task, there was increased activity in the posterior deltoid over time ($p<0.05$) and no changes in the range or variability (p>0.05). There were no significant changes in the summary variables during the pulling task (p>0.05).

Table 7.2: Mean, coefficient of variation and range EMG variables that showed significant differences over time (p<0.05) in each muscle and task. Muscle activity was normalized to the cubic reference function to mitigate the effects of fatigue on EMG amplitude. Values that were significantly different from the baseline value according to the post hoc tests are denoted with *.

Mea.	Task	Musc.	Mean ± std (% baseline reference function)-Fatigue removed					
			First	20	40	60	80	Last
COV	Down	Pdel	24.63 ± 11.73	26.34 ± 12.0	25.29 ± 9.53	31.31 ± 12.19	26.80 ± 12.37	**34.24 ± 17.73***
	Up	Bi	23.89 ± 2.60	38.75 ± 4.59	34.88 ± 3.10	32.52 ± 3.87	37.73 ± 3.93	**40.66 ± 5.03***
Mean	Down	Bi	0.15 ± 0.06	0.23 ± 0.08	0.26 ± 0.06	0.36 ± 0.10	0.37 ± 0.09	**0.46 ± 0.12***
		Utrap	0.08 ± 0.24	0.04 ± 0.01	0.02 ± 0.003	0.02 ± 0.004	0.02 ± 0.005	0.02 ± 0.005
	Push	Tri	0.61 ± 0.07	0.74 ± 0.07	0.83 ± 0.10	0.83 ± 0.09	0.83 ± 0.10	0.87 ± 0.12
		Pdel	0.05 ± 0.01	**0.05 ± 0.00***	**0.05 ± 0.01***	**0.05 ± 0.00***	**0.05 ± 0.01***	**0.06 ± 0.01***
	Up	Tri	0.31 ± 0.03	0.37 ± 0.04	0.42 ± 0.06	0.47 ± 0.08	0.61 ± 0.15	0.52 ± 0.09
		Infra	0.76 ± 0.05	0.85 ± 0.07	0.95 ± 0.09	0.92 ± 0.10	0.89 ± 0.08	0.89 ± 0.07
		Sert	0.44 ± 0.03	0.51 ± 0.04	0.56 ± 0.05	0.57 ± 0.06	**0.62 ± 0.06***	**0.60 ± 0.06***
		Pecc	0.67 ± 0.07	0.78 ± 0.10	0.86 ± 0.13	0.91 ± 0.14	0.98 ± 0.17	1.00 ± 0.17
Range	Down	Bi	0.23 ± 0.10	0.41 ± 0.11	0.64 ± 0.19	0.69 ± 0.21	0.77 ± 0.25	0.90 ± 0.25
		Lats	3.15 ± 0.42	3.36 ± 0.41	4.43 ± 0.62	4.64 ± 0.65	5.53 ± 0.99	4.40 ± 0.65
	Up	Tri	0.31 ± 0.03	0.37 ± 0.04	0.42 ± 0.06	0.47 ± 0.08	0.61 ± 0.15	0.52 ± 0.09
		Infra	0.76 ± 0.05	0.85 ± 0.07	0.95 ± 0.09	0.92 ± 0.10	0.89 ± 0.08	0.89 ± 0.07
		Sert	0.44 ± 0.18	0.51 ± 0.04	0.56 ± 0.05	0.57 ± 0.06	**0.62 ± 0.06***	**0.60 ± 0.06***
		Pecc	0.67 ± 0.07	0.78 ± 0.10	0.86 ± 0.13	0.91 ± 0.14	0.98 ± 0.17	1.00 ± 0.17



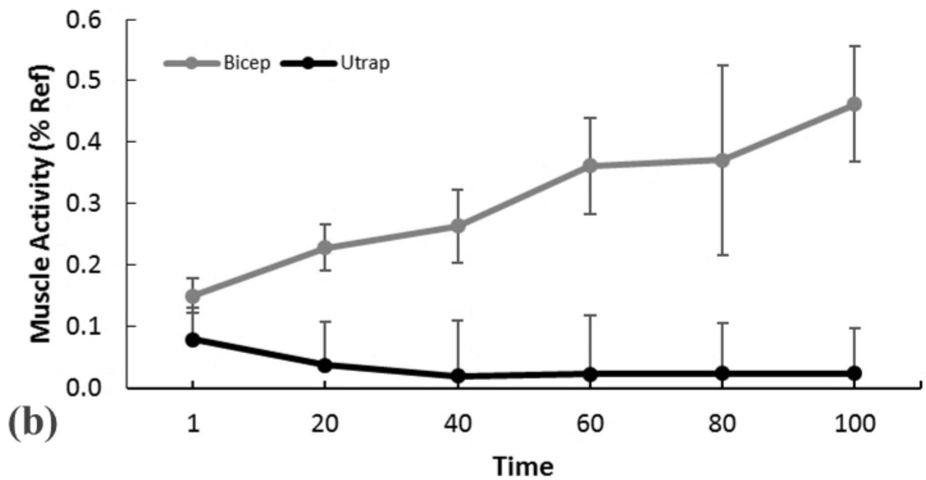


Figure 7.2: Mean EMG activity for the (a) triceps, infraspinatus, serratus anterior and clavicular head of the pectoralis major during the up task and for the (b) biceps and upper trapezius during the down task in 20% intervals over time. Only muscles with statistically significant mean changes are displayed.

7.4.2 Load Sharing Changes between Muscles

Negative correlations between muscle pairs were indicative of load sharing relationships between muscles and were task dependent. During the up task, there were significant, weak, negative correlations between the upper trapezius muscle and the anterior deltoid (r=-0.174), middle deltoid (r=-0.169), and clavicular head of the pectoralis major (r=-0.226, Figure 7.3a) ($p<0.05$). During the push and pull tasks, there was load sharing between the anterior deltoid and triceps (r=-0.149, Figure 7.3b), and between the middle trapezius and biceps (r=-0.203, Figure 7.3c) respectively. There were no significant load sharing changes to maintain performance during the down task ($p>0.05$).

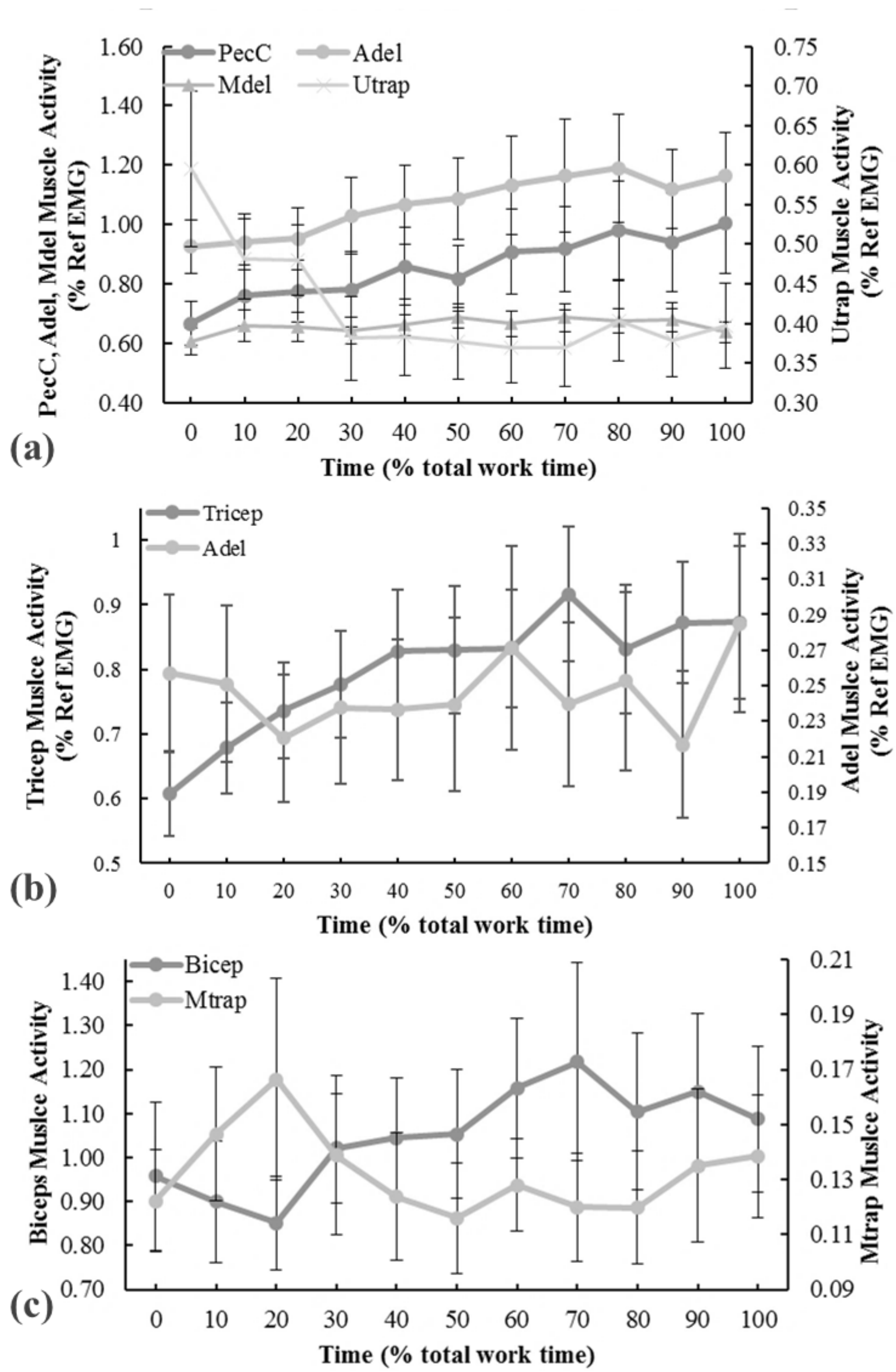


Figure 7.3: Muscle pairs with significant, negative correlations during the (a) up, (b) push, and (c) pull tasks, indicative of load sharing between these muscle pairs to maintain task performance over time. Mean muscle activity across participants, normalized to the cubic reference functions, is shown for each muscle in 10% time intervals from work cycle 1 to the final work cycle.

7.4.3 Subacromial Space

Between day repeatability of the SAS width measurements was excellent in both the neutral (Chronbach's alpha=0.97, ICC=0.93) and abducted (Chronbach's alpha=0.99, ICC=0.98) postures ($p<0.01$) (Figure 7.4). Subacromial space width was affected by time and posture. In the abducted posture, there were no significant changes in width over time (First: 0.81 ± 0.21 cm, middle: 0.84 ± 0.20 cm, last: 1.11 ± 0.15 cm) ($p>0.05$), and in the neutral posture there was a significant decrease in width between the middle (1.63 ± 0.18 cm) and last (1.11 ± 0.15 cm) time points ($p<0.05$), and no significant differences from the first measure (1.14 ± 0.14 cm) ($p>0.05$) (Figure 7.5). The SAS width was greater in the neutral posture (1.18 ± 0.16 cm) than the abducted posture (0.85±0.20 cm) ($p<0.05$).

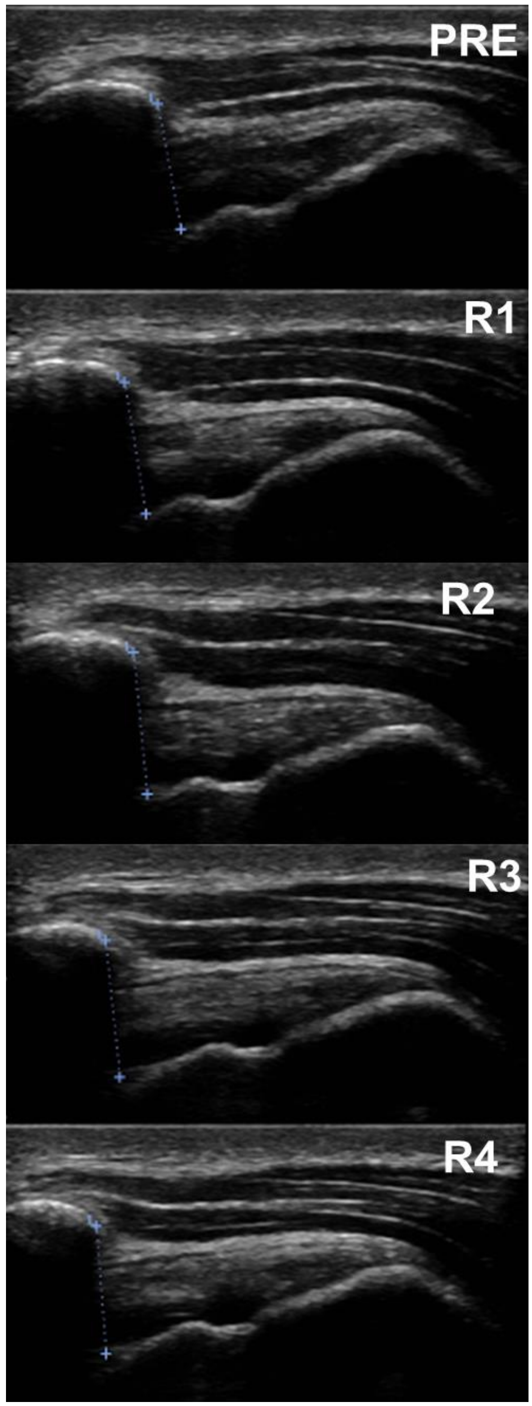


Figure 7.4: Ultrasound images of SAS from one participant over time in the neutral (0°) posture. The + symbols on each image are placed on the tip of the acromion and the humeral head and the blue line between them is the measured SAS width. The image labeled “Pre” is the baseline image and labels R1-R4 are during reference exertion 1 (12 min) to reference exertion 4 (48 minutes).

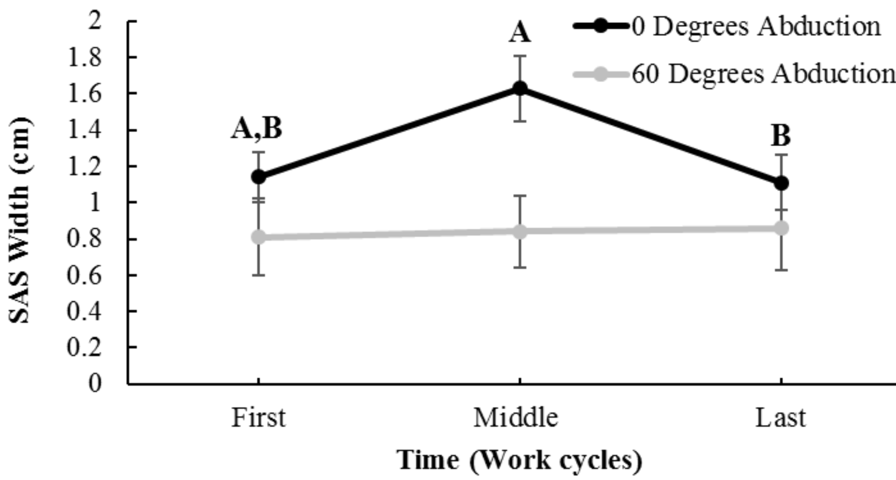


Figure 7.5: Subacromial space decreased between the middle and last time points in the neutral posture only (p<0.05). There were no other significant effects of time between measures in either posture (p>0.05).

7.4.4 Effects of Repetitive work on Perceptions of Fatigue and Affect

Participants' perception of both physical and mental fatigue increased over time (p<0.05) (Figure 7.6). There were no significant relationships between the rates of change in perceived mental or physical fatigue and endurance time (p>0.05). Participants; fatigue state (FSQ) also increased after the repetitive work protocol, evidenced by significant increased scores on three (FSQ1,2,4) of the four FSQ questions following the termination of the work protocol (p<0.05). Participant perceived affect changed over time: there were significant increases in felt arousal (FAS) and decrease in feeling (FS) between baseline and all other time points (p<0.05). Together, these variables create the Circumplex model of affect and indicate that feelings are moving on a trajectory from the deactivated/pleasant quadrant (calm, relaxed, serene, contented) to the activated/unpleasant quadrant (upset, stressed, nervous, tense) (Figure 7.7). There were no significant relationships between baseline fatigue state (FSQ score) or baseline trait or state self-control (SCS, SCCS scales) and endurance time ($p>0.05$). The repetitive work protocol had a significant, negative effect on perceived self-control (SCCS pre=5.0 ± 0.8; SCCS post=3.4 ± 1.3) ($p>0.05$).

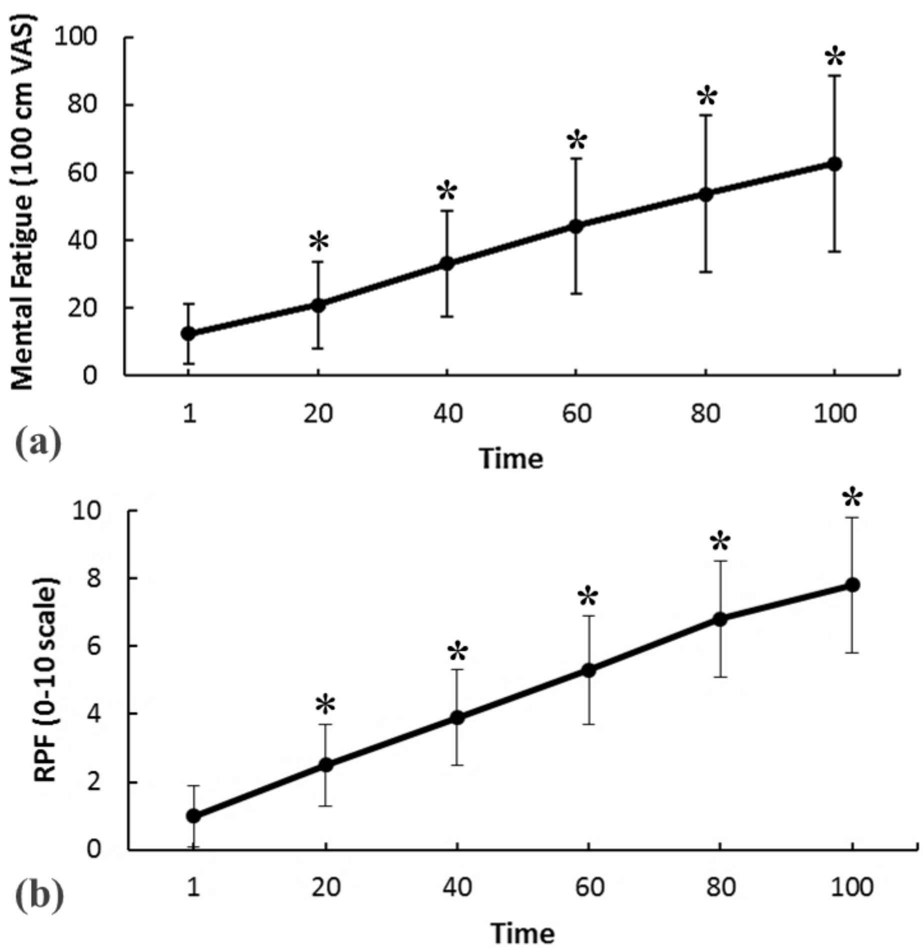


Figure 7.6: (a) Mental Fatigue was measured on a 100 mm VAS and increased over time. (b) Ratings of Perceived Fatigue were given on a 0-10 scale and increased over time. For both scales, significant increases with respect to the baseline score are denoted with *.

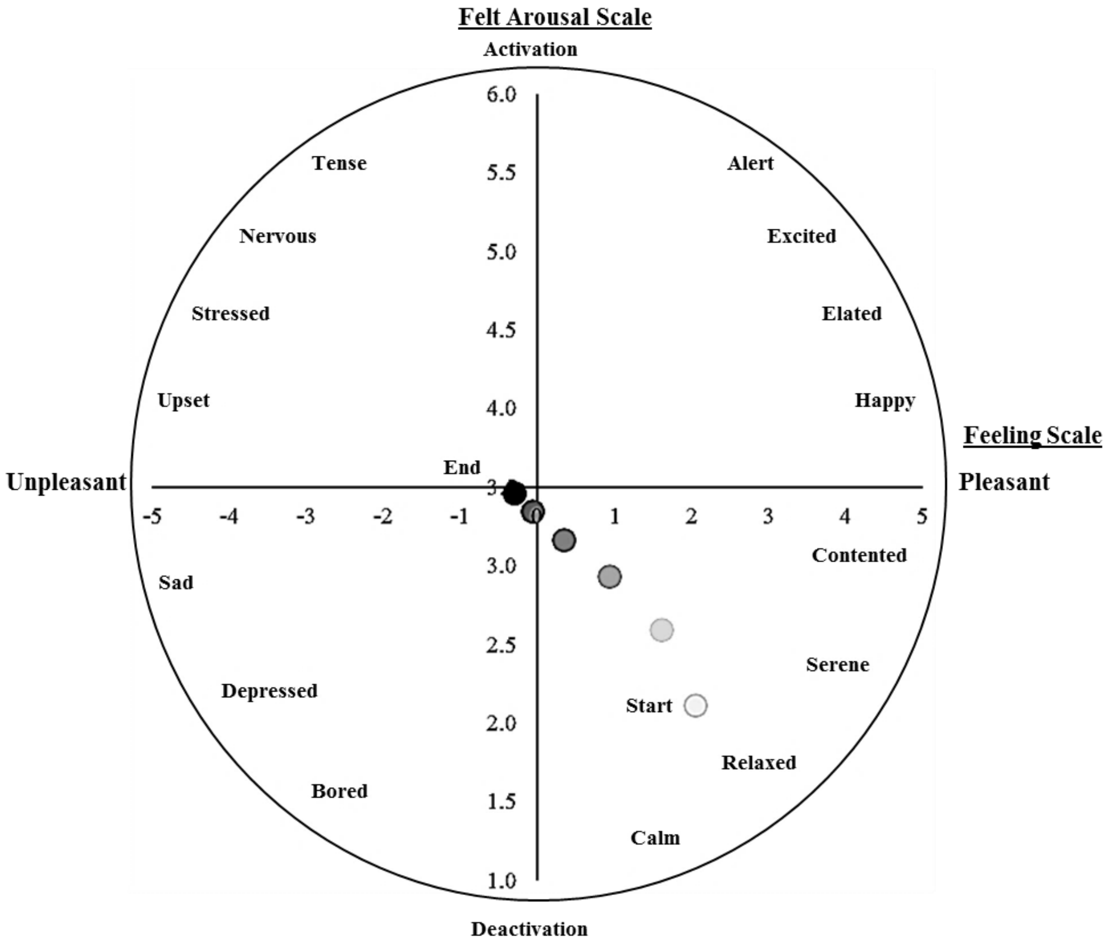


Figure 7.7: The Circumplex model of affect is created by combining scores from the Feeling Scale (x-axis, Unpleasant – Pleasant) and the Felt Arousal Scale (y-axis, Deactivated – Activated). Average scores across participants move on a trajectory from the pleasant-deactivated quadrant towards the unpleasant-activated quadrant over time. Each mark on the figure represents 20% endurance time intervals from the baseline scores to the final scores and all values, on both scales were significantly different from the baseline scores ($p<0.05$).

7.5 Discussion

The aim of this investigation was to combine several methods to evaluate the multi-faceted effects of static, repetitive work on the shoulder. As we hypothesized, there was evidence of load sharing between the shoulder muscles to maintain task performance as fatigue developed. The normalizing procedure used in this investigation mitigates the effects of fatigue on EMG amplitude, making the significant EMG changes observed representative of changes in load sharing over time. Participants perceived greater mental and physical fatigue over time with a negative effect on their affective valence and reported self-control. Contrary to our hypothesis, we did not see the expected decrease in SAS width over time, particularly in the 60° abduction posture.

As fatigue developed, participants showed that they could utilize the redundant nature of the shoulder musculature to share the task demands between muscles, which likely allowed for active muscle recovery from fatigue to maintain task performance longer. Previous work in our laboratory found muscle activity and kinematic changes during repetitive work tasks (Tse et al 2016, McDonald et al 2016, McDonald et al 2017: Thesis Chapter 6: *Muscular and kinematic adaptations to fatiguing, repetitive work*). Since the kinematic adaptations changed the moment demands at the shoulder, it was unclear if muscular changes were in response to new moment demands or a neuromuscular control strategy varying muscle activation to allow for recovery. Finding muscle activity changes during static work as well suggests that at least some of the changes during dynamic work may be a control strategy. The shoulder complex requires active stability in mid-range of postures (Labriola et al, 2004; Warner et al, 1999) and constant balancing of muscle moments to maintain a static posture. Varying muscle activation to maintain task performance also requires adjustments in surrounding musculature to maintain the postural demands. While we found changes in muscle activation patterns, we did not find as many changes as were expected. Maintaining the postural demands of this task likely impacted the possible muscle activity patterns available to complete these work tasks.

The rotator cuff muscles are well designed to maintain shoulder stability. Because of their small cross-sectional areas, they do not generate large moments that need to be balanced by other muscles surrounding the GH joint (Veeger & Van der Helm, 2007). Fatigue in this muscle group can impact the subacromial space as seen with the inter-individual variability in SAS changes following a fatigue protocol (Chopp-Hurley et al 2016). This study found that although there were no significant mean differences in SAS measures across the population, 39-57% of individuals displayed fatigue related changes that were considered disadvantageous to the SAS (Chopp-Hurley et al 2016). In the current study, we did not see significant group changes in the SAS space width between the baseline and final SAS measurements, however, 41-63% (in abducted posture and neutral posture respectively) of individuals had reduced SAS width between these two time points. This supporting the idea that when evaluating workplace injury risk, it may be more advantageous to consider individual responses and risk factors, rather than only the population response. Another factor to consider is the cumulative exposure to repetitive task demands. Ettinger et al (2012) examined scapular kinematic changes following a day of dental hygiene work and found greater anterior tilt by the end of the day in hygienists with greater work experience. Anterior tilting of the scapular is an important change because it can decrease SAS width (Banas et al, 1995). These findings, combined with the absence of the expected significant SAS width changes in the current investigation, suggest that repeated exposure to repetitive demands may be more detrimental to the ability to maintain the SAS width than a single exposure to fatiguing work.

In addition to the physical response to repetitive work, this study also aimed to improve our understanding of the psychological response to these demands. As we hypothesized, there was an increase in perceived physical fatigue (RPF) over time in response to the static tasks. Perceived physical fatigue has been shown to be related to physiological measures of fatigue during exercise tasks (Micklewright et al, 2017). The results of the current study support its use for workplace fatigue as well, especially given that the RPF is a very simple metric to administer and interpret. Perceived mental fatigue also increased with time during the work tasks. Although the rate of change in perceived mental or physical fatigue was not significantly related to endurance time, previous work has shown mentally demanding tasks impact physical endurance (Mehta and Agnew 2012). The tasks in the current study were not designed to be mentally demanding, and perhaps future work should incorporate mental demands more realistic to workplace tasks. Mental fatigue was assessed with a 100 mm VAS and scores across participants increased from 12.4 ± 8.9 mm at baseline to 62.6 ± 25.8 mm by the end of the protocol. One participant however, began the protocol with a baseline mental fatigue score of 72 mm, interestingly, this was the only participant that was unable to complete the first set of

12 work cycle (data were removed from the analysis). This supports further investigation into the interaction between mental and physical fatigue in the workplace. The fatiguing work protocol also impacted participants' affective valence and perceived self-control. Although the current study did not assess a post-task measure of self-control performance, the current findings are in line with effects demonstrated by Clarkson and colleagues (2016) and show that within a work environment mental fatigue has negative downstream effects on perceived self-control abilities. These variables have been studied in the workplace and positive affect and self-control are associated with successful workplace attributes, including goal setting and directed action, work engagement, higher income, better negotiation, conflict resolution and decision making skills, and reduced absenteeism and intension to turnover (Barsade and Gibson 2007, Bledow et al 2011, Duckworth and Gross 2014). The current work makes an interesting contribution to this literature by showing the direct effect of a challenging simulated work task on these variables, outside of confounding factors present in a field setting.

There are some limitations to this work that should be considered in the interpretation and application of the results. The results from this study come from a single exposure to a physically demanding simulated work protocol. Although most work tasks can be performed for more than 36 minutes before exhaustion, this protocol was designed to induce fatigue and measure the impacts within a time frame that was feasible within the constraints of a controlled, laboratory setting. Although extreme care was taken to ensure that there was consistency between each ultrasound measure, several images were removed from the analysis based on visual inspection from the researcher performing the offline measurements. As fatigue accumulated, some participants were no longer able to maintain a fully upright torso during the ultrasound measurements and thus, although the probe was in the same location and orientation between measures, the view obtained in the image was different. This was a conservative decision and future work should aim to address this postural limitation. Due to the repetitive and prolonged work task procedure only muscles accessible with surface EMG were included in this investigation. Although this only allowed for one rotator cuff muscle to be included, evaluating 13 muscles gives a good representation of the response of the shoulder complex. The results from this study only examine one posture and one set of work tasks, expanding these methodologies to more tasks will aid in greater understanding of workplace responses.

7.6 Conclusion

The response to static, repetitive work and fatigue development is multi-faceted and variable between individuals. The findings from this investigation show that there are physical and mental effects to repetitive work tasks that display inter-individual variability over time. These findings support a wholistic approach to the assessment of workplace demands and support previous suggestions that evaluating individual responses to physical demands may be a beneficial approach to prevent workplace injuries.

7.7 References

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CHAPTER 8: DISCUSSION

The shoulder complex is a flexible, mobile group of joints and muscles that afford many kinematic and muscular strategies for completing functional tasks at the hand. This makes it challenging to predict and understanding how low load, repetitive work tasks will be completed challenging. Although there are ample opportunities for variability and load sharing within the complex, workplace shoulder injuries remain prevalent (WSIB 2016). The purpose of this thesis was to improve our understanding of muscular and kinematic adaptation strategies of the shoulder complex throughout the process of fatigue and recovery. This was accomplished with a combination of empirical studies aimed at eliciting fatigue through different approaches and methodological studies to better understand any fatigued-induced changes in kinematic and muscular strategies.

8.1 Methodological contributions to understanding fatigue development

The methodological contributions in this thesis were developed to address some of the challenges faced while evaluating fatigue. Electromyography is a tool used for measuring muscle activity and from that, estimating muscle force. This presents challenges since the EMG and force relationship is affected by myoelectric fatigue, causing increases in EMG amplitude and decreases EMG frequency that are unrelated to the muscle force. This made it difficult to interpret the EMG changes found in Chapter 2, as fatigue artifacts in the EMG made it unclear as to what extent the EMG changes observed during the post-fatigue work were the result of increased muscle force (indicative of load sharing) or myoelectric fatigue. This led to the development of the novel EMG normalizing methods presented in Chapter 4. Normalizing the work cycle

EMG data to a cubic function created with repeated, static, submaximal exertions was done to mitigate the fatigue artifact in EMG, allowing us to distinguish between fatigue changes and muscle force changes in Chapters 6 and 7. This method was developed using the data from the dynamic work investigation in Chapter 6. To further evaluate this method, we have normalized the EMG data from the static work cycles in Chapter 7 to the set of maximal voluntary exertions (developed in Chapter 3) and compared these results to the data normalized with the fatigue mitigating method (developed in Chapter 4 method). For both normalizing methods, repeated measures ANOVAs were completed on the mean EMG from the first, 20%, 40%, 60%, 80% and last work cycles for each muscle and task. The relationships between muscle activity and time were dependent on the muscle, task and the normalizing method (Table 8.1, Table 8.2). When accounting for fatigue using the novel EMG normalizing method, there were fewer muscles with significantly greater activity over time. The differences in these results suggest that accounting for the effects of fatigue on EMG amplitude is a crucial step when interpreting EMG data over time. The normalizing process developed in this thesis can also be applied to other joints and muscle groups and merits further investigation.

Table 8.1: Mean EMG after normalizing to maximum voluntary exertions. Only muscles with a significant main effect of time are included in the table and time points with significant post hoc tests are denoted * and bold font.

Measure	Task	Muscle	Mean and std dev (% MVE)					
			0	20	40	60	80	100
Mean	Down	Bi	1.0 ± 0.4	1.5 ± 0.4	1.7 ± 0.4	2.1 ± 0.5	2.5 ± 0.6	**3.1 ± 0.8**
		Tri	7.5 ± 1.0	8.2 ± 1.1	8.2 ± 1.1	8.4 ± 1.1	9.1 ± 1.2	**9.7 ± 1.4**
		Adel	0.2 ± 0.1	0.6 ± 0.2	0.5 ± 0.1	0.5 ± 0.1	0.5 ± 0.1	0.5 ± 0.1
		Utrap	1.9 ± 0.5	1.3 ± 0.3	0.6 ± 0.10	0.7 ± 0.1	0.7 ± 0.1	0.7 ± 0.2
		Lats	1.7 ± 0.2	1.7 ± 0.3	1.7 ± 0.3	1.7 ± 0.2	1.8 ± 0.3	1.9 ± 0.3
		PecC	3.2 ± 0.4	3.6 ± 0.5	3.6 ± 0.5	3.9 ± 0.6	4.1 ± 0.6	3.9 ± 0.5
	Pull	Utrap	3.4 ± 1.0	2.7 ± 0.5	1.1 ± 0.2	1.5 ± 0.3	1.1 ± 0.3	1.6 ± 0.5
		Lats	4.6 ± 0.5	**6.8 ± 0.9***	5.9 ± 0.6	6.1 ± 0.8	6.3 ± 0.8	6.5 ± 0.7
	Push	Tri	4.1 ± 0.7	5.0 ± 0.9	5.9 ± 1.3	6.1 ± 1.2	6.1 ± 1.3	6.8 ± 1.6
		Adel	5.3 ± 0.7	4.9 ± 0.7	5.5 ± 0.7	6.2 ± 0.9	6.2 ± 0.9	7.2 ± 1.1
		Pdel	0.8 ± 0.1	0.8 ± 0.1	0.9 ± 0.1	0.9 ± 0.1	1.0 ± 0.1	1.2 ± 0.2
		Infra	5.2 ± 0.7	4.6 ± 0.6	5.3 ± 0.6	5.5 ± 0.7	6.1 ± 0.9	6.8 ± 0.9
	Up	Tri	2.1 ± 0.3	2.6 ± 0.4	3.0 ± 0.6	3.5 ± 0.8	4.2 ± 1.1	4.1 ± 1.0
		Adel	19.5 ± 1.6	20.9 ± 1.7	24.1 ± 2.3	26.5 ± 3.0	**27.6 ± 3.5***	**27.8 ± 3.2***
		Mdel	17.0 ± 1.3	19.0 ± 1.5	20.4 ± 1.4	21.4 ± 1.7	**22.8 ± 2.0***	**22.3 ± 1.8***
		Pdel	2.0 ± 1.2	2.3 ± 0.2	2.5 ± 0.3	2.6 ± 0.3	2.7 ± 0.3	**2.8 ± 0.3***
		Infra	17.4 ± 1.7	19.9 ± 1.8	22.8 ± 2.3	23.1 ± 2.6	23.1 ± 2.4	**23.8 ± 2.5***
		Ltrap	25.5 ± 2.0	25.3 ± 2.2	27.8 ± 2.3	30.3 ± 2.8	31.0 ± 2.6	30.2 ± 3.0
		Sert	17.3 ± 1.6	20.1 ± 1.9	22.0 ± 2.1	23.3 ± 2.8	**24.8 ± 2.9***	**24.1 ± 3.0***
		Pecc	11.1 ± 1.0	12.7 ± 1.2	13.9 ± 1.6	14.7 ± 1.6	**16.1 ± 1.9***	**16.7 ± 2.2***

Table 8.2: Mean EMG after normalizing to submaximal reference function to remove fatigue artifacts. Only muscles with a significant main effect of time are included in the table and time points with significant post hoc tests are denoted * and bold font.

Measure	Task	Muscle	Mean and std dev (% baseline reference function)-Fatigue removed					
			0	20	40	60	80	100
Mean	Down	Bi	0.1 ± 0.1	0.2 ± 0.1	0.3 ± 0.1	0.4 ± 0.10	0.4 ± 0.1	**0.5 ± 0.1***
		Utrap	0.1 ± 0.2	0.04 ± 0.01	0.0 ± 0.003	0.02 ± 0.004	0.02 ± 0.005	0.02 ± 0.005
	Push	Tri	0.6 ± 0.1	0.7 ± 0.1	0.8 ± 0.10	0.8 ± 0.1	0.8 ± 0.10	0.9 ± 0.12
		Pdel	0.1 ± 0.01	**0.1 ± 0.00***	**0.1 ± 0.01***	**0.05 ± 0.00***	**0.1 ± 0.01***	**0.1 ± 0.01***
	Up	Tri	0.3 ± 0.03	0.4 ± 0.04	0.4 ± 0.1	0.5 ± 0.1	0.6 ± 0.2	0.5 ± 0.1
		Infra	0.8 ± 0.1	0.9 ± 0.1	1.0 ± 0.1	0.9 ± 0.10	0.9 ± 0.1	0.9 ± 0.1
		Sert	0.4 ± 0.03	0.5 ± 0.04	0.6 ± 0.1	0.6 ± 0.1	**0.6 ± 0.1***	**0.6 ± 0.1***
		Pecc	0.7 ± 0.1	0.8 ± 0.1	0.9 ± 0.1	0.9 ± 0.14	1.0 ± 0.2	1.0 ± 0.2

Another challenge in interpreting the results of this work was the variability in which muscles and at which time points fatigue developed between participants (Table AJ.1). Creating the multi-muscle fatigue score (MMFS) in Chapter 5 allowed us to evaluate overall shoulder fatigue state between participants. This function uses a combination of amplitude and frequency changes, and similar to the normalizing method in Chapter 4, it can be applied to other muscle groups and tasks.

Not all methods investigated during the development of this thesis were successful in aiding in the interpretation of the results. Since the process of fatigue and recovery is dynamic, summary variables, such as the mean, COV, and MAD, do not capture all of the pertinent information. Statistical parameter mapping (SPM) is conceptually identical to univariate procedures, however each test results in a test statistic as a function of time and uses random field theory to assess the significance of the statistical field (Pataky et al, 2000). This allows for the statistical evaluation of complete data sets over time, instead of reducing each data set to a single composite score, such as the mean. The kinematic changes that were described in Chapter 6, were also evaluated with SPM repeated measures ANOVAs and t-test post hoc tests. The SPM method showed interesting results for humeral internal and external rotation during the drilling task (Figure 8.1). The SPM analysis indicated that there were no significant differences between the first work cycle versus the 25% work cycle (Figure 8.1a), but there were differences between the first work cycle and the 50%, 75% and final work cycles. The analysis revealed that the differences between the first and 50% work cycle occurred at 3 short intervals throughout the task (Figure 8.1b). By the 75% work cycle there was more humeral external rotation for more than the first half of the drill task (Figure 8.1c) and by the final work cycle there was more external rotation for the entire drill task (Figure 8.1d) compared to the first WC. In the traditional analysis of the mean humeral internal/external rotation, significant differences from the first work cycle were identified in the 50%, 75% and final work cycles as well; however, the SPM procedure gives us more information about the phase of the task where these changes occurred, which could be an important for certain interpretations. For example, in a more dynamic task, where the mean joint angle is less representative of the whole task, this method may be very useful in identifying which phases of a task may be putting workers at greater risk of injury. Although this specific joint angle showed interesting findings with the SPM analysis, the majority of the angles did not have statistically significant post hoc tests. Furthermore, this method remains subject to similar constraints as traditional methods, susceptible to type I/II errors and therefore we were still restricted to limiting the number of comparison and thus the number of work cycles analysed. Although this method shows promise and warrants further investigation, it was not effective in this thesis and therefore was not included in Chapter 6.

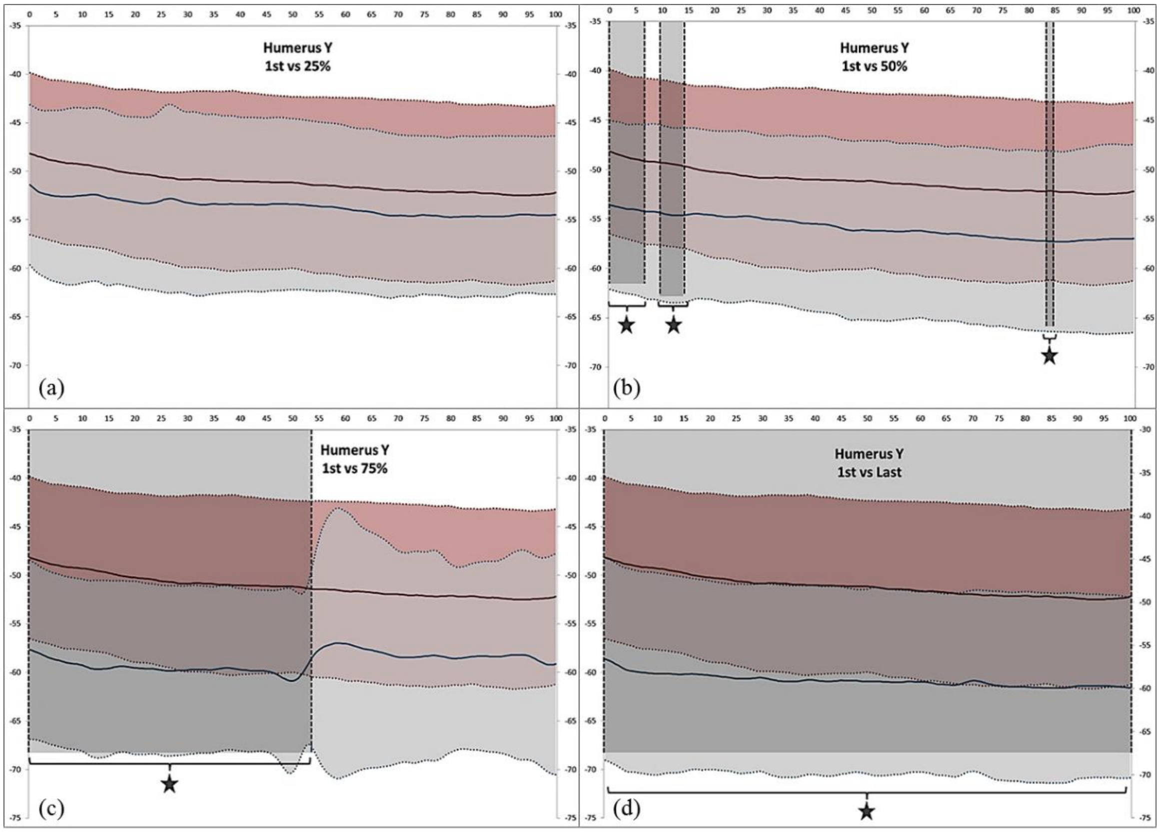

Figure 8.1: Humeral external rotation during the drill task. The mean joint angle for the first work cycle is plotted as the red line in all figures and the standard deviation with the red shading. The mean external rotation angles for the (a) 25%, (b) 50%, (c) 75%, and (d) final work cycles are plotted with blue lines and their standard deviations are plotted with blue shading. Significant differences identified with the SPM procedure are highlighted in gray and denoted with star icons.

8.2 Between Participant Variability

Between participant variability was evident throughout this thesis. The purpose of Chapter 3 was to select a set of maximal voluntary exertions that would reliability elicit maximum activation from all the shoulder muscles of interest, with the fewest exertions possible. This was done to reduce data collection time and decrease the risk of fatigue development before initiating the fatiguing work protocols. Unexpectedly, there was a high degree of between participant variability in which test elicited maximum activation for each muscle was unexpected. To manage this variability, we heuristically developed a set of criteria to select a set of tests that would elicit at least 95% of maximum recorded activity for each muscle, for at least 70% of the participants. Although this method has limitations, the selected set of 8 tests, which combined both multi-muscle and individual tests, performed better across the participants than either the multi-muscle or individual muscle test protocols alone. Future work could investigate other multi-muscle and individual muscle tests; however, I believe the issue of between participant variability would likely still exist. It would likely require a very large test set to elicit maximum activation from all muscles and participants, increasing the time demands and risk of developing muscle fatigue. These factors were considered when selecting the number of MVE tests.

The between participant variability in the response to repetitive work was quantified in Chapter 6. Endurance time ranged from 57-240 minutes across participants, even though tasks were scaled to individual anthropometrics and strength. This finding prompted us to include psychological measures for the static work study in Chapter 7. Although we did not find significant correlations between the rate of change in perceived fatigue or baseline state or trait self-control scores and endurance time, this is a relationship that should be investigated further in the future. There was also evidence of large between participant variability in both muscle activity and joint angles throughout the dynamic work tasks in Chapter 6. Between participant joint angle coefficients of variation (COV) ranged from 3-10419%, depending on the angle and task. Between participant muscle activity COV ranged from 0.4-635.2%, depending on the muscle and task. These tasks were designed to be constrained, which makes the observed range of possible strategies interesting. Future work should aim to explain what drives the variability between participants.

8.3 Within Participant Variability

Within participant variability has been challenging to quantify throughout this thesis. Although changes in coefficient of variation (COV) were observed in joint angles and muscle activity over time in Chapter 6, it is difficult to meaningfully interpret the impact that changes in a single joint angle and axis has on the entire system. Statistical methods used in this thesis limited the ability to quantify the net effects of all the joint angle and muscular changes together. Figure 8.2, Figure 8.3 and Figure 8.4 (refer Appendix K for example with labelled markers) depict how small changes, across multiple joints, combine to create whole body adaptations and postural variability in three representative participants (Figures for the remaining participants are in Appendix K). Eleven markers are plotted for each figure to show the torso and right upper extremity during the Chapter 6 drilling task, for the 5 work cycles (first, 25%, 50%, 75%, last WC) analyzed. For each work cycle, the first frame is plotted in green, every 50 frames (1 per second) is plotted in progressively darker shades of gray, and the last frame of the task is plotted in red. For each figure, the right side view shows forward and backward torso lean, glenohumeral flexion/extension and elbow flexion/extension changes. The back view effectively shows left- and rightward trunk bend and glenohumeral ad/abduction. Based on visual inspection of each participant's figures, there does not appear to be a relationship between postural variability and endurance time (included on each figure).

There are however, between participant differences in how much postural variability occurred and when. Figure 8.2, shows a participant who completed 72 minutes of work with very low postural variability visible throughout. Alternatively, Figure 8.3 shows a participant who completed 63 minutes of work and had seemingly high postural variability throughout all work cycles. In the first, 50% and 75% WC the participant appears to be leaning forward over time during the task. Alternatively, during the last work cycle, the participant appears to be leaning both forward and backward to maintain their drill task performance. From the back view, they appeared to be static for the first WC, tried leaning to the left for the 25% WC, right for the 50% and 75% WCs and by the last WC were leaning in both directions. Figure 8.4 is an example of a participant who completed 180 minutes of work. In the first WC, they leaned both forward and backwards during the task. They did not display much postural variability for the 25% WC. For the 50% and 75% WC, they appear to lean further backward over time and then leaned forward during their final drill task. This participant had little observable variability from the back view. There are many possible explanations for this variability, including learning, distributing the loads to different structures, fatigue and changing priorities over time. Future work should attempt to quantify and explain this variability, both within participants and between participants.

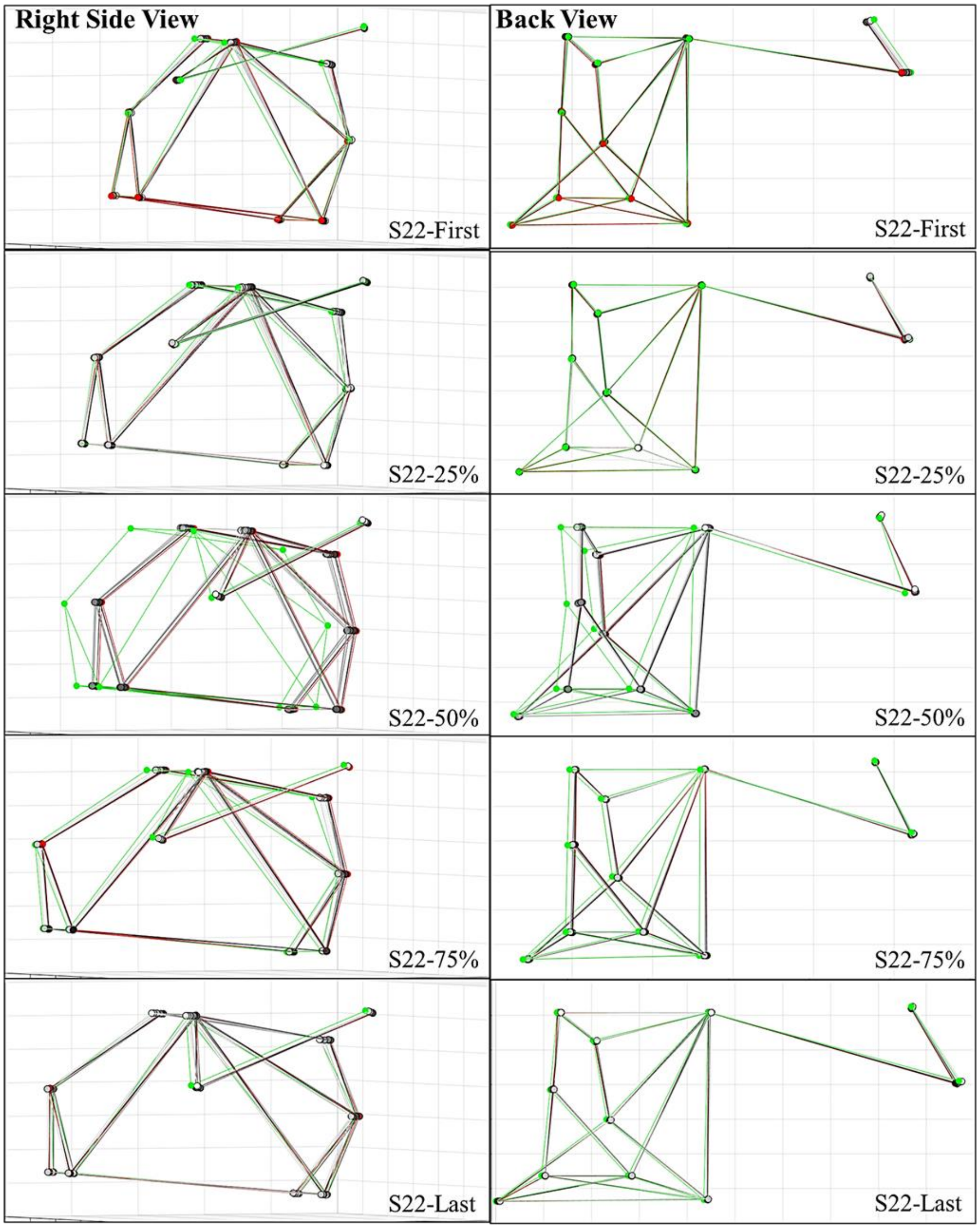

Figure 8.2: Torso and right upper extremity for Participant 22 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one per second) and the last frame is plotted in red.

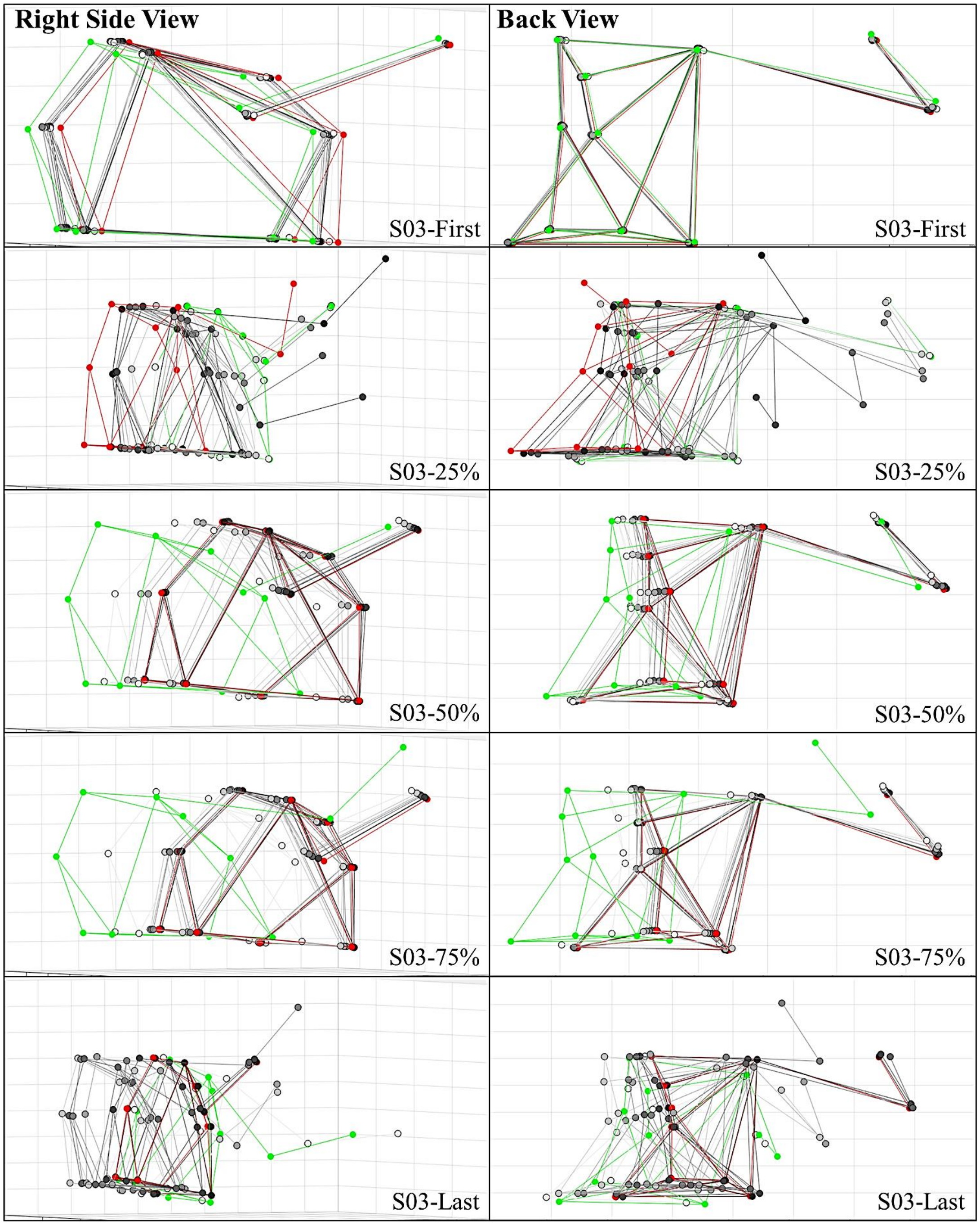

Figure 8.3: Torso and right upper extremity for Participant 3 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one per second) and the last frame is plotted in red.

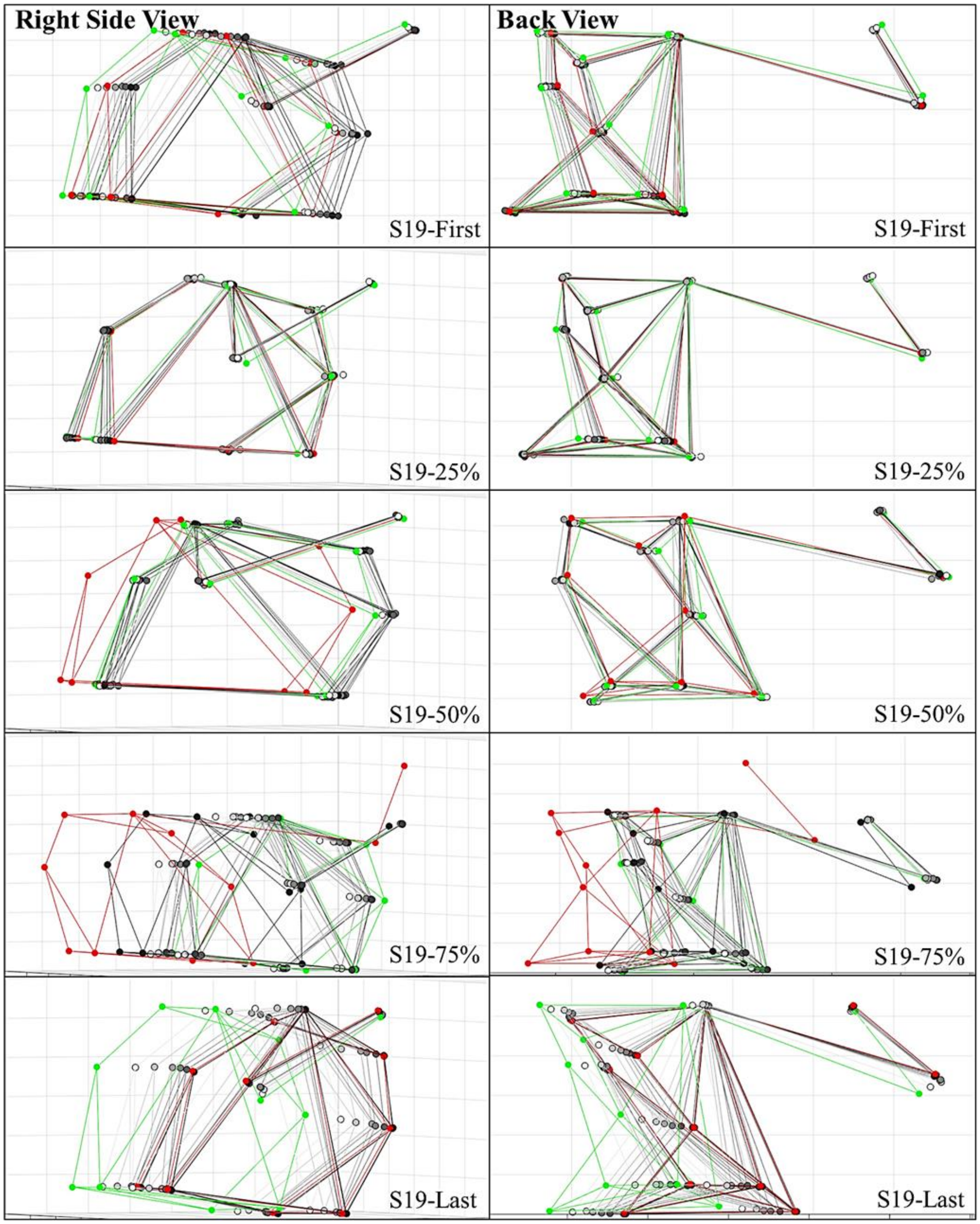

Figure 8.4: Torso and right upper extremity for Participant 3 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one per second) and the last frame is plotted in red

8.4 Response to Static and Dynamic Repetitive Work

The response to repetitive work is complex, multi-faceted and varies between individuals and over time. This thesis has shown that it impacts kinematics, muscle activity, muscle fatigue, strength, affective valence and perceptions of mental and physical fatigue. Throughout the thesis we have found that participants were able to adopt kinematic and muscular strategies to maintain task performance as their capacities changed during the process of fatigue and recovery. Muscular and kinematic changes to maintain task performance occur during repetitive work both following muscle fatigue (Chapter 2) and while developing fatigue (Chapters 6,7). These changes were sensitive to task, time and individual differences (Chapters 2,6,7) and muscular changes occur both with (Chapter 2,6) and without (Chapter 7) kinematic changes. Although there were statistically significant changes reported in these chapters, there were also smaller changes across multiple joint angles that were not in themselves statistically significant, but when considered together likely amounted to functional changes. Future work should aim to develop methods to quantify how subtle joint angle changes across different segments, which may not be statistically significant on their own, can combine together to produce functionally significant deviations in overall posture.

Participants were not necessarily able to perceive all these changes during recovery. In Chapter 2, participants rating of perceived exertion returned to baseline by the end of the post-fatigue work; however, signs of myoelectric fatigue remained in a number of muscles, with additional muscles developing fatigue during this post-fatigue work. During fatiguing work (Chapters 6, 7), participants were able to perceive mental and physical fatigue development; therefore, this perceptual relationship may be different during recovery. These perceptual differences could have implications in workplace settings and warrant further investigation.

Throughout this thesis it is evident that all individuals do not all respond the same way to the demands of repetitive work and the process of fatigue as well as recovery. In the workplace, although many workers are exposed to the same demands, some develop musculoskeletal disorders and others do not (Kilbom & Persson, 1987). This may be the result of individual specific compensation strategies. Statistical methods generally limit the interpretation of results to the average response and this may be causing us miss valuable information about individuals that may be at risk of workplace injuries. A specific example in this thesis was the variable changes in subacromial space width over time. Although there were no significant changes to the width of the space from baseline over time in the abducted or neutral postures, 41-63% of participants had reduced space over time (in the abducted posture and neutral posture respectively). By only focusing our interpretation on the mean findings these potentially at risk individuals are missed. Future work can aim to investigate how we can identify and include these individuals in the design and evaluation of workplace tasks.

8.5 Conclusions

This thesis has found that participants were able to adapt to the demands of repetitive work through coordinated kinematic and muscular strategies, allowing them to maintain task performance as their capacities changed during the process of fatigue and recovery. The response of the shoulder complex to repetitive work is multi-faceted and the specific muscular, kinematic and perceptual responses were dependent on task demands, time and individual variability. The methods that I developed in this thesis have allowed us to interpret the observed changes in muscle activity with repetitive work. Future work should further validate these methods and aim to understand what drives the observed variability both between and within individuals, as this likely has significant implications in the development of workplace injuries.

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APPENDICES

APPENDIX A: ETHICS APPROVAL FOR CHAP 2

Hamilton Health Sciences McMaster University Inspiring Innovation and Discovery St. Joseph's Healthcare Hamilton

Hamilton Integrated Research Ethics Board
AMENDMENT REQUEST

REB Project #: 09-548
Principal Investigator: Dr. Peter Keir

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

Document(s) Amended with version # and date:

- Protocol Amendment - Detailed Protocol #FW3-Fatigue Dated: 28 October, 2013
- Consent Form Amendment - Participant Information Sheet/Consent Form Dated: 28 October, 2013
- Recruitment Ad - Advertisement Flyer
- Administrative Change - Additional funding from Centre for Research Expertise for the Prevention of Musculoskeletal Disorder
- Administrative Change - Add Co-investigators Calvin Tse and Alison McDonald
- Administrative Change - Delete Co-investigator Samantha Ebata
- Other - PI's letter dated October 28, 2013 clarifying amendment

Research Ethics Board Review
(this box to be completed by HIREB Chair only)

[X] **Amendment approved as submitted**

[ ] **Amendment approved conditional on changes noted in "Conditions" section below**

[ ] **New enrolment suspended**

[ ] **Study suspended pending further review**

Level of Review:

[ ] Full Research Ethics Board
[X] Research Ethics Board Executive

The Hamilton Integrated Research Ethics Board operates in compliance with and is constituted in accordance with the requirements of: The Tri-Council Policy Statement on Ethical Conduct of Research Involving Humans; The International Conference on Harmonization of Good Clinical Practices; Part C Division 5 of the Food and Drug Regulations of Health Canada, and the provisions of the Ontario Personal Health Information Protection Act 2004 and its applicable Regulations; For studies conducted at St. Joseph's Hospital, HIREB complies with the health ethics guide of the Catholic Alliance of Canada

Raelene L Rathbone
~~Suzette Salama PhD., Chair~~
Raelene Rathbone, MB, BS, MD, PhD, Chair

11/12/2013
Date

All Correspondence should be addressed to the HIREB Chair(s) and forwarded to:
HIREB Coordinator
293 Wellington St. N, Suite 102, Hamilton ON L8L 8E7
Tel. 905-521-2100 Ext. 42013 Fax: 905-577-8378

APPENDIX B: ETHICS APPROVAL FOR CHAP 3-6

Hamilton Health Sciences McMaster University Inspiring Innovation and Discovery St. Joseph's Healthcare Hamilton

Hamilton Integrated Research Ethics Board
AMENDMENT REQUEST

REB Project #: 09-548

Principal Investigator: Dr. Peter Keir

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

Document(s) Amended with version # and date:

- Protocol Amendment - Protocol #:FW5-Fatigue Dated: 12 February, 2015
- Consent Form - Participant Information Sheet/Consent Form Dated: 12 February, 2015
- Recruitment Ad - Advertisement Flyer
- Other - PI's Letter dated February 12, 2015 clarifying the amendment

Research Ethics Board Review
(this box to be completed by HIREB Chair only)

[X] **Amendment approved as submitted**

[ ] **Amendment approved conditional on changes noted in "Conditions" section below**

[ ] **New enrolment suspended**

[ ] **Study suspended pending further review**

Level of Review:

[X] Full Research Ethics Board
[ ] Research Ethics Board Executive

Conditions:
The Hamilton Integrated Research Ethics Board operates in compliance with and is constituted in accordance with the requirements of: The Tri-Council Policy Statement on Ethical Conduct of Research Involving Humans; The International Conference on Harmonization of Good Clinical Practices; Part C Division 5 of the Food and Drug Regulations of Health Canada, and the provisions of the Ontario Personal Health Information Protection Act 2004 and its applicable Regulations; For studies conducted at St. Joseph's Hospital, HIREB complies with the health ethics guide of the Catholic Alliance of Canada.

Raelene L Rathbone
Suzette Salama PhD., Chair
Raelene Rathbone, MB, BS, ~~MD~~, PhD, Chair

04 March, 2015
Date of REB Meeting

All Correspondence should be addressed to the HIREB Chair(s) and forwarded to:
HIREB Coordinator
293 Wellington St. N, Suite 102, Hamilton ON L8L 8E7
Tel. 905-521-2100 Ext. 42013 Fax: 905-577-8378

APPENDIX C: ETHICS APPROVAL FOR CHAPTER 7

Hamilton Health Sciences McMaster University Inspiring Innovation and Discovery St. Joseph's Healthcare Hamilton

Hamilton Integrated Research Ethics Board
AMENDMENT REQUEST

REB Project #: 09-548

Principal Investigator: Dr. Peter Keir

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

Document(s) Amended with version # and date:

- Protocol Amendment - Protocol #:FW6-Fatigue Dated: 10 February, 2016
- Recruitment Ad - Advertisement Poster
- Consent Form - Participant Information Sheet/Consent Form Dated: 10 February, 2016
- Administrative Change - Delete Co-investigator Calvin Tse
- Other - Summary and Rationale of Changes dated February 10, 2016

Research Ethics Board Review
(this box to be completed by HIREB Chair only)

[X] **Amendment approved as submitted**

[ ] **Amendment approved conditional on changes noted in "Conditions" section below**

[ ] **New enrolment suspended**

[ ] **Study suspended pending further review**

Level of Review:

[ ] Full Research Ethics Board
[X] Research Ethics Board Executive

Conditions:
The Hamilton Integrated Research Ethics Board operates in compliance with and is constituted in accordance with the requirements of: The Tri-Council Policy Statement on Ethical Conduct of Research Involving Humans; The International Conference on Harmonization of Good Clinical Practices; Part C Division 5 of the Food and Drug Regulations of Health Canada, and the provisions of the Ontario Personal Health Information Protection Act 2004 and its applicable Regulations; For studies conducted at St. Joseph's Hospital, HIREB complies with the health ethics guide of the Catholic Alliance of Canada

S. Salama

Suzette Salama PhD., Chair
Raelene Rathbone, MB, BS, MD, PhD, Chair

2/29/2016
Date

All Correspondence should be addressed to the HIREB Chair(s) and forwarded to:
HIREB Coordinator
293 Wellington St. N, Suite 102, Hamilton ON L8L 8E7
Tel. 905-521-2100 Ext. 42013 Fax: 905-577-8378

APPENDIX D: CONSENT FORM FOR CHAPTER 2

McMaster University

Inspiring Innovation and Discovery

PARTICIPANT INFORMATION SHEET

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

Calvin Tse, BSc Kin (Hons) Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Alison McDonald, MSc, PhD Candidate, Department of Kinesiology, Faculty of Science, McMaster University

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

You are being invited to participate in a study conducted by Calvin Tse and Alison McDonald because you are either a healthy male or are currently awaiting rotator cuff repair surgery. The study will help us to learn more about the coordination of shoulder muscles and how they adapt to muscle fatigue.
In order to decide whether or not you want to be a part of this research study, you should understand what is involved and the potential risks and benefits. This form gives detailed information about the research study, which will be discussed with you. Once you understand the study, you will be asked to sign this form if you wish to participate. Please take your time to make your decision. Feel free to discuss it with your friends and family.

WHY IS THIS RESEARCH BEING DONE?

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

WHAT IS THE PURPOSE OF THIS STUDY?
The purpose of this study is to measure the effort of the muscles surrounding the shoulder during a cyclic repetitive task simulating work on an automotive assembly line to better understand how muscle activity changes with time.

WHAT WILL MY RESPONSIBILITIES BE IF I TAKE PART IN THE STUDY?
If you volunteer to participate in this study, we will ask you to do the following things:

STUDY PROTOCOL

In these studies, we are interested in measuring muscle activity during various tasks using the arm and hand. To measure muscle activity we use surface electrodes to measure the superficial muscles of the shoulder and upper arm. These electrodes will only monitor the electrical activity of the muscle of interest and will not transmit an electrical signal to the body. Surface electrodes are small circular self adhesive pads with a conductive gel in the middle. The skin over each muscle of interest will be cleaned with alcohol and two surface electrodes will be placed. For this study, the following muscles of the right side of the participant will be investigated: three heads of deltoid, upper and lower trapezius, and biceps. Once the electrodes are placed, you will be asked to complete a series of reference contractions and maximal efforts for each muscle being investigated. These will be static efforts against resistance.

You will be asked to perform 4 work-related tasks with the right arm in a clockwise direction at your own pace within a 1-minute cycle continuously for a maximum of for 20 minutes followed by a series of actions that will fatigue the muscle on the front of your right shoulder, followed by performing the 4 tasks again for one hour. The 4 tasks will consist of a two-finger pull, pipe connector push, turning a knob and an anterior push while holding a drill. Thus the duty cycle (the ration between work and rest in each cycle) will be self-determined. You must fully complete each task before moving on to the subsequent task or you will be required to repeat the task. A beep will indicate the start of each cycle. You will also be asked to rate your level of exertion for the task every 5-minutes. At the end of the test, you will be asked to perform another set of maximal contractions.

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

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

1. You may experience mild discomfort or skin irritation from being shaved and cleansed in preparation for electrode placement. This is usually very mild and clear within 24 hours.
2. There may be discomfort related to the delayed onset of muscle soreness associated with maximal and isometric contractions of the arm muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.
3. Maximal effort isometric contractions are associated with an increase in blood pressure. If you have received medical clearance and/or are already physically active, the risks are minimal. The researchers' first priority as an investigator is to maintain the emotional, psychological, and physical health of those participating in the study.

HOW MANY PEOPLE WILL BE IN THIS STUDY?

Fifteen men from the University population will be recruited for this study.

WHAT ARE THE POSSIBLE BENEFITS FOR ME AND/OR FOR SOCIETY?
Participants will receive no direct benefits from participating in this study. However, participants should know that their willingness to serve as a subject for this experiment will help develop an understanding of shoulder mechanics during rotator cuff injury, which may benefit individuals in the future.

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

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

WHAT INFORMATION WILL BE KEPT PRIVATE?
Your data will not be shared with anyone except with your consent or as required by law. All personal information such as your name, address, phone number or email will be removed from the data and will be replaced with a number. A list linking the number with your name will be kept in a secure place, separate from your file. The data, with identifying information removed will be securely stored in a locked office in the research office and on an encrypted hard drive. The data for this research study will be retained for ten years.
For the purposes of ensuring the proper monitoring of the research study, it is possible that a member of the Hamilton Health Sciences/FHS McMaster University Research Ethics Board, a Health Canada representative may consult your research data. However, no records which identify you by name or initials

will be allowed to leave the institution/university/hospital. By signing this consent form, you authorize such access.

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

CAN PARTICIPATION IN THE STUDY END EARLY?

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

WILL I BE PAID TO PARTICIPATE IN THIS STUDY?

If you agree to take part, we will reimburse you $20 for your time.

WILL THERE BE ANY COSTS?

Your participation in this research project may involve additional costs of parking for the duration of the study collection (approximately 3 hours).

WHAT HAPPENS IF I HAVE A RESEARCH-RELATED INJURY?

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

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

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

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

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

CONSENT STATEMENT

CONSENT STATEMENT

SIGNATURE OF RESEARCH PARTICIPANT/LEGALLY-AUTHORIZED REPRESENTATIVE*

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

Name of Participant

Name of Legally Authorized Representative

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

Consent form administered and explained in person by:

Name and title

Signature Date

SIGNATURE OF INVESTIGATOR:

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

Signature of Investigator Date

APPENDIX E: CONSENT FORM FOR CHAPTER 3-6

McMaster University

Inspiring Innovation and Discovery

PARTICIPANT INFORMATION SHEET

Title of Study: The effect of simulated repetitive work on muscle activity and load distribution in the shoulder complex

Daanish Mulla, BSc Kin (Hons) Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Calvin Tse, BSc Kin (Hons) Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Alison McDonald, BSc, PhD Candidate, Department of Kinesiology, Faculty of Science, McMaster University

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

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You are being invited to participate in a study conducted by Alison McDonald because you are a healthy male. The study will help us to learn more about the coordination of shoulder muscles and how they adapt to muscle fatigue.

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

WHY IS THIS RESEARCH BEING DONE?

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

WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to measure the effort of the muscles surrounding the shoulder during a cyclic repetitive task aimed at fatiguing the rotator cuff muscles to better understand how muscle activity and kinematics changes with time.

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

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

STUDY PROTOCOL

In these studies, we are interested in measuring muscle activity and kinematics during various tasks using the arm and hand. To measure muscle activity we use surface electrodes to measure the superficial muscles of the shoulder and upper arm. These electrodes will only monitor the electrical activity of the muscle of interest and will not transmit an electrical signal to the body. Surface electrodes are small circular self adhesive pads with a conductive gel in the middle. The skin over each muscle of interest will be cleaned with alcohol and two surface electrodes will be placed. For this study, the following muscles of the right side of the participant will be investigated: three heads of deltoid, upper, middle and lower trapezius, biceps brachii, infraspinatus, latissimus dorsi, triceps brachii, pectoralis major and supraspinatus. Once the electrodes are placed, you will be asked to complete a series of reference contractions and maximal efforts for each muscle being investigated. These will be static efforts against resistance.

You will be asked to repetitively perform four tasks aimed at fatiguing the rotator cuff muscles and simulating industrial work. You will also be asked to rate your level of exertion for the task after each duty cycle. At the end of the test, you will be asked to perform another set of maximal and submaximal contractions to quantify muscle fatigue.

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

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

4. You may experience mild discomfort or skin irritation from being shaved and cleansed in preparation for electrode placement. This is usually very mild and clear within 24 hours.
5. There may be discomfort related to the delayed onset of muscle soreness associated with maximal and isometric contractions of the arm muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.
6. Maximal effort isometric contractions are associated with an increase in blood pressure. If you have received medical clearance and/or are already physically active, the risks are minimal. The researchers' first priority as an investigator is to

maintain the emotional, psychological, and physical health of those participating in the study.

HOW MANY PEOPLE WILL BE IN THIS STUDY?

Twenty men from the University population will be recruited for this study.

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

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

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

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

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

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

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

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

WILL I BE PAID TO PARTICIPATE IN THIS STUDY?

If you agree to take part, we will reimburse you $30 for your time.

WILL THERE BE ANY COSTS?

Your participation in this research project may involve additional costs of parking for the duration of the study collection (approximately 4 hours on one day~~s~~).

WHAT HAPPENS IF I HAVE A RESEARCH-RELATED INJURY?

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

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

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

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

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

CONSENT STATEMENT

SIGNATURE OF RESEARCH PARTICIPANT/LEGALLY-AUTHORIZED REPRESENTATIVE*

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

____

Name of Participant

____

Name of Legally Authorized Representative

______________________________________ ____________________

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

Consent form administered and explained in person by:

____

Name and title

______________________________________ ____________________

Signature Date

SIGNATURE OF INVESTIGATOR:

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

______________________________________ ____________________

Signature of Investigator Date

APPENDIX F: CONSENT FORM FOR CHAPTER 7

PARTICIPANT INFORMATION SHEET

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

Daanish Mulla, BSc Kin (Hons) Candidate, Department of Kinesiology, Faculty of Science, McMaster University

Alison McDonald, BSc, PhD Candidate, Department of Kinesiology, Faculty of Science, McMaster University

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

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You are being invited to participate in a study conducted by Alison McDonald because you are a healthy male. The study will help us to learn more about the coordination of shoulder muscles and how they adapt to muscle fatigue.

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

WHY IS THIS RESEARCH BEING DONE?

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

WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to measure the effort of the muscles surrounding the shoulder during cyclic repetitive tasks aimed at fatiguing the rotator cuff muscles to better understand how muscle activity changes with time. To investigate changes in subacromial space from repetitive work.

WHAT WILL MY RESPONSIBILITIES BE IF I TAKE PART IN THE STUDY?
If you volunteer to participate in this study, we will ask you to do the following things:

STUDY PROTOCOL

In these studies, we are interested in measuring muscle activity during various tasks using the arm and hand. To measure muscle activity we use-surface electrodes-to measure the superficial muscles of the shoulder and upper arm. These electrodes will only monitor the electrical activity of the muscle of interest and will not transmit an electrical signal to the body. Surface electrodes are small circular self adhesive pads with a conductive gel in the middle. The skin over each muscle of interest will be cleaned with alcohol and two surface electrodes will be placed. For this study, the following muscles of the right side of the participant will be investigated: three heads of deltoid, upper, middle and lower trapezius, biceps brachii, infraspinatus, latissimus dorsi, triceps brachii, pectoralis major (sternal and clavicular insertions), serratus anterior, and supraspinatus. Once the electrodes are placed, you will be asked to complete a series of reference contractions and maximal efforts for each muscle being investigated. These will be static efforts against resistance.

You will be asked to repetitively perform tasks in four hand locations aimed at fatiguing the rotator cuff muscles-will also be asked to rate your level of exertion, level of perceived activation, and levels of pleasure-displeasure for the task after each duty cycle. At the end of each cycle, you will be asked to perform another set of maximal and submaximal contractions to quantify muscle fatigue and the experimenter will use ultrasound to measure your subacromial space (in your shoulder).

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

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

7. You may experience mild discomfort or skin irritation from being shaved and cleansed in preparation for electrode placement. This is usually very mild and clear within 24 hours.
8. There may be discomfort related to the delayed onset of muscle soreness associated with maximal and isometric contractions of the arm muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.
9. Maximal effort isometric contractions are associated with an increase in blood pressure. If you have received medical clearance and/or are already physically active, the risks are minimal. The researchers' first priority as an investigator is to maintain the emotional, psychological, and physical health of those participating in the study.
10. Allergic contact dermatitis from the Aquasonic Ultrasound transmission gel is possible, however very rare as the gel is water soluble and hypoallergenic. In the case of discomfort experienced while using the gel the experiment will be terminated.

HOW MANY PEOPLE WILL BE IN THIS STUDY?

Twenty men from the University population will be recruited for this study.

WHAT ARE THE POSSIBLE BENEFITS FOR ME AND/OR FOR SOCIETY?
Participants will receive no direct benefits from participating in this study. However, participants should know that their willingness to serve as a subject for this experiment will help develop an understanding of shoulder mechanics during rotator cuff injury, which may benefit individuals in the future.

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

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

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

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

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

CAN PARTICIPATION IN THE STUDY END EARLY?

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

WILL I BE PAID TO PARTICIPATE IN THIS STUDY?

If you agree to take part, we will reimburse you $40 for your time.

WILL THERE BE ANY COSTS?

Your participation in this research project may involve additional costs of parking for the duration of the study collection (approximately 4 hours on one day).

WHAT HAPPENS IF I HAVE A RESEARCH-RELATED INJURY?

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

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

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

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

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

CONSENT STATEMENT

SIGNATURE OF RESEARCH PARTICIPANT/LEGALLY-AUTHORIZED REPRESENTATIVE*

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

____

Name of Participant

____

Name of Legally Authorized Representative

_______________________________________ ___________________

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

Consent form administered and explained in person by:

____

Name and title

_______________________________________ ___________________

Signature Date

SIGNATURE OF INVESTIGATOR:

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

_______________________________________ ___________________

Signature of Investigator Date

APPENDIX G: This article has been printed "with permission" by the publisher Elsevier, Journal of Electromyography and Kinesiology

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Adaptations to isolated shoulder fatigue during simulated repetitive work. Part I: Fatigue

Calvin T.F. Tse, Alison C. McDonald, Peter J. Keir *

Occupational Biomechanics Laboratory, Department of Kinesiology, McMaster University, Hamilton, ON L8S 4K1, Canada

A R T I C L E I N F O

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Keywords:
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Muscle fatigue
Shoulder
Scapular kinematics
Control strategy

A B S T R A C T

Upper extremity muscle fatigue is challenging to identify during industrial tasks and places changing demands on the shoulder complex that are not fully understood. The purpose of this investigation was to examine adaptation strategies in response to isolated anterior deltoid muscle fatigue while performing simulated repetitive work. Participants completed two blocks of simulated repetitive work separated by an anterior deltoid fatigue protocol; the first block had 20 work cycles and the post-fatigue block had 60 cycles. Each work cycle was 60 s in duration and included 4 tasks: handle pull, cap rotation, drill press and handle push. Surface EMG of 14 muscles and upper body kinematics were recorded. Immediately following fatigue, glenohumeral flexion strength was reduced, rating of perceived exertion scores increased and signs of muscle fatigue (increased EMG amplitude, decreased EMG frequency) were present in anterior and posterior deltoids, latissimus dorsi and serratus anterior. Along with other kinematic and muscle activity changes, scapular reorientation occurred in all of the simulated tasks and generally served to increase the width of the subacromial space. These findings suggest that immediately following fatigue people adapt by repositioning joints to maintain task performance and may also prioritize maintaining subacromial space width.

1. Introduction

Numerous industrial jobs consist of low load, repetitive tasks, which have been identified as increasing risk for developing workplace injuries (Nordander et al., 2009). Along with tissue damage, altered movement patterns have been observed with repetitive work in animal models. Movement pattern changes have the potential to increase, or decrease, exposure over time, and also provide opportunity for rest and recovery (Barbe et al., 2003; Coq et al., 2009; Elliott et al., 2008). The mobility afforded by the human shoulder and the upper extremity may allow workers opportunities to use kinematic and muscle recruitment strategy changes to adapt to the demands of repetitive work, especially when compromised with fatigue.

The shoulder complex achieves its large range of motion through the simultaneous motion of its three joints (Inman et al., 1944). Vital to maintaining this range of motion is proper motion of the scapula (Inman et al., 1944; Picco et al., 2010; van der Helm et al., 1995). Scapular position also impacts the space between the acromion and the humeral head that encompasses rotator cuff tendons, known as the subacromial space (SAS) (Banas et al., 1995). The SAS is highly variable between individuals and affected by arm position, scapular rotation, and muscle activity (Banas et al., 1995; Chopp and Dickerson, 2012; Graichen et al., 2005). Muscle attachment sites on the scapula cause its orientation to be affected by changes in muscle activity patterns and fatigue (Ebaugh et al., 2005). For example, a fatiguing external rotation protocol has been shown to lead to increased scapular external rotation, upward rotation, and decreased posterior tilt during humeral elevation (Ebaugh et al., 2006a,b; Tsai et al., 2003). The SAS can also be affected with rotator cuff muscle fatigue and humeral head migration during humeral elevation (Chen et al., 1999; Chopp et al., 2011).

Repetitive work and muscle fatigue can lead to other kinematic changes in the upper extremity as well. Following fatigue protocols, scapulothoracic and glenohumeral changes have been found to be sensitive to elevation angle (Ebaugh et al., 2006a,b; Tsai et al., 2003). Adaptations have also been observed in more complex tasks involving multiple joints and specific performance demands. For example, with repetitive pointing, participants changed their wrist and elbow movements to compensate for altered shoulder position (Fuller et al., 2009). People appear to prioritize

* Corresponding author at: McMaster University, Department of Kinesiology, Ivor Wynne Centre, room 212, 1280 Main Street West, Hamilton, ON L8S4K1, Canada. Tel.: +1 905 525 9140x23543.
E-mail address: pjkeir@mcmaster.ca (P.J. Keir).

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performance of endpoint location with compensations made via greater variability in movement trajectories in other axes, changes in endpoint velocity, movement time and kinematics (Bosch et al., 2012; Fuller et al., 2011; Luger et al., 2015). Many of these studies have demonstrated that movement changes occur as muscle fatigue develops during repetitive work, how they react to a strong fatigue stimulus remains unknown.

Allowing workers to maintain performance and quality standards as well as understanding adaptations during workplace tasks is important for worker health and applications such as job design. An early study found that experienced carpenters were able to maintain performance during simulated sawing, drilling and hammering tasks despite completing a fatigue protocol; however their kinematics were not monitored, so the strategies used to maintain task performance are not known (Hammarskjiild & Harms-Ringdahl, 1992). Other investigations of repetitive hammering and sawing have observed multi-joint changes that allowed participants to maintain performance following a fatigue protocol (Côté et al., 2002, 2005). During a sawing task there was a large change in elbow range of motion that was offset by a number of concurrent small changes in other joints (Côté et al., 2002). Changes in movement patterns may provide objective signs of reduced capacity. Given that changes in posture are often observable, they could be used as indicators of fatigue in an occupational setting. Previous work in our laboratory has evaluated changes to EMG and kinematics during one hour of simulated repetitive work and found that adaptations occurred before signs of muscle fatigue (Ebata, 2012). Muscle fatigue development while performing low load tasks can be time consuming and thus fatigue protocols can be used to accelerate the development of muscle fatigue. Understanding these strategies following a fatigue protocol leads to the purpose of this investigation, which was to examine the muscular and kinematic adaptation strategies that develop in response to isolated muscle fatigue while performing simulated repetitive work.

This is the first of a two-part communication of our investigation aimed at understanding the immediate and longer-term adaptations to fatigue. This paper will focus on the immediate effects of fatigue in the first 8 min of repetitive work following a fatigue protocol. The companion paper will examine the response over one hour of repetitive work (McDonald et al., 2016). We hypothesized that after fatiguing the anterior deltoid; participants would be able to maintain performance through changes in muscle recruitment and kinematics of the upper extremity and trunk. The kinematic changes should act to reduce the shoulder moment demands and muscle activity will support the mechanical needs at the shoulder.

2. Methods

2.1. Participants

Twelve right-hand dominant men, free from upper limb or shoulder pathologies in the past year were recruited from the university population. The study was approved by the Hamilton Integrated Research Ethics Board (HIREB). Participants provided written consent, age (20–24 years) and anthropometric measurements including mass (76.5 ± 8.5 kg), height (177.9 ± 6.8 cm), umbilicus height (108 ± 5 cm), and acromioclavicular height (148 ± 6 cm).

2.2. Instrumentation

Eleven cameras (Raptor-4, Motion Analysis Corporation, Santa Rosa, CA) sampled 26 reflective markers (at 100 Hz) placed on specific anatomical landmarks of the pelvis, thorax, and right upper extremity (Wu et al., 2002, 2005). A scapular tracker with 4 reflective markers was affixed to the skin over the spine of the scapula and the acromion and used to capture subcutaneous scapular movements (Fig. 1) (Karduna et al., 2001; Parel et al., 2013). Temporary markers were placed on the skin on the inferior angle and root of the scapular spine and were removed after relationships were established with static calibration trials.

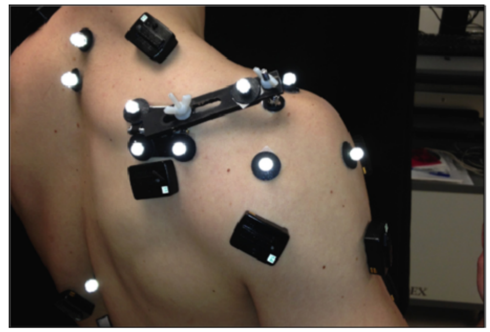

Fig. 1. A scapular tracker with four reflective markers was used track the three-dimensional motion of the scapula. The hinge conformed to the mid portion of the scapular spine and the adjustable arm extended to the area above the postero-lateral aspect of the acromion process. The footpad was affixed to the skin over the acromion and conformed to the angle of the acromion with a ball and socket joint (modeled after Karduna et al., 2001).

Surface electromyography (EMG) of 14 muscles was recorded using silver-contact wireless bipolar bar electrodes with fixed 1 cm inter-electrode spacing (Trigno, Delsys Inc., Natick, MA, USA). Muscles on the right side included pectoralis major (sternal and clavicular head), latissimus dorsi, serratus anterior, infraspinatus, anterior deltoid, middle deltoid, posterior deltoid, biceps brachii, triceps brachii. The upper and lower trapezius were monitored bilaterally. EMG signals were differentially amplified (CMRR > 80 dB, input impedance 10^{15} Ω), band-pass filtered (20–450 Hz), sampled at 2000 Hz and were converted with a 16-bit card with a ±5 V range. Prior to electrode placement, sites were shaved and skin scrubbed with isopropyl alcohol. Electrodes were placed parallel to muscle fibers (with guidance from Perotto & Delagi, 2005). Maximum voluntary exertions were performed to elicit maximal activity from each muscle. Each of the 14 tests were repeated twice and the peak muscle activity across all tests was used to normalize EMG data (Hodder & Keir, 2013; Perotto & Delagi, 2005).

2.3. Simulated work protocol

Participants completed two blocks of simulated repetitive work separated by an anterior deltoid focused fatigue protocol. Each work cycle was 60 s in duration. Twenty (20) cycles were completed before the fatigue protocol ("pre-fatigue") and 60 cycles were completed after the fatigue protocol ("post-fatigue"). Simulated work was performed at a workstation comprised of 4 tasks: (1) handle pull (2 kg, 10 repetitions), (2) cap rotation (3 clockwise & 3 counter-clockwise repetitions), (3) 10 s anterior drill press (50% MVC), (4) handle push (2 kg, 10 repetitions) (Fig. 2). Tasks 1 and 4 were positioned above the umbilicus by one-half the vertical distance between the umbilicus and AC joint and tasks 2 and 3 were adjusted to this same vertical distance above the AC joint for each participant. Based on previous work from our laboratory, these tasks were selected as a sample of possible industrial tasks and the apparatus was positioned such that the completion

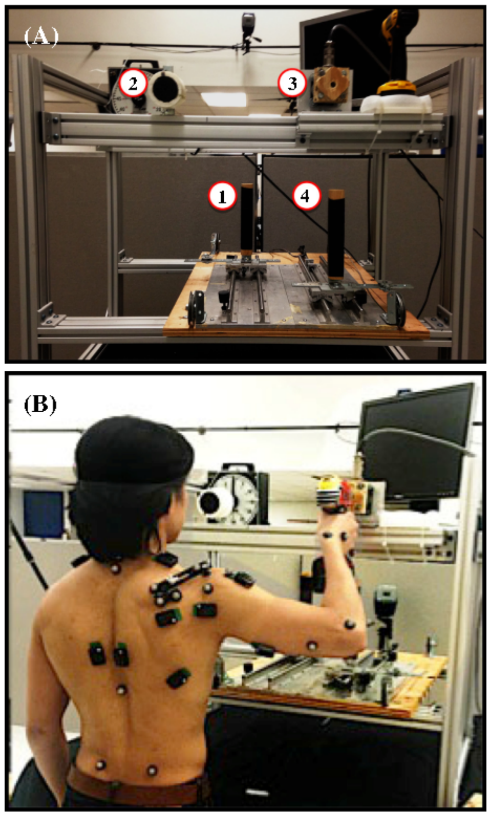

Fig. 2. (A) Simulated work apparatus. Each 60-s work cycle included (1) handle pull (2 kg, 10 repetitions), (2) knob rotation (6 revolutions-3 clockwise, 3 counter-clockwise) (3) drill press (50% MVE in Z-axis, 10 s), (4) handle push (2 kg, 10 repetitions). The apparatus was elevated so the upper two tasks were above shoulder height. (B) Image of participant performing the anterior drill press task in the simulated work apparatus.

of these tasks would challenge the anterior deltoid muscle (Ebata, 2012). Participants were instructed to complete the tasks at their own pace within each 60-s work cycle. Push and pull displacements were recorded using linear potentiometers and drill press forces and moments were recorded with a 6-degree of freedom force transducer (MC3-500, AMTI, Watertown, MA, USA). Visual force feedback was provided on a computer monitor to assist participants in maintaining the required force output during the drill press using custom Labview software (National Instruments, Texas, USA). All data were sampled at 2000 Hz.

2.4. Fatigue protocol

The fatigue protocol targeted the anterior deltoid (AD) as it is a primary mover for many of the tasks. The protocol included repeated static and dynamic forward flexion exertions in the sagittal plane (adapted from Ebaugh et al., 2006a,b). Resistance was set at 25% of maximal forward flexion force determined using a handheld force gauge before initiating the pre-fatigue work cycles (M5-200, Mark-10 Corporation, Copiague, NY, USA). The resistance for each participant was created by filling a sack with the appropriate weight of lead shot and was secured to the ulnar surface of the distal forearm with compression wrap. The protocol started with 20 repetitions of forward glenohumeral flexion from 0° (at side) to 90° of elevation, followed by a 60 s isometric hold at 45° of glenohumeral flexion. All exertions were performed with the hand open, forearm in neutral rotation (thumb pointing up), and elbow in full extension. This combination of flexion and external rotation has been shown to elicit considerable AD muscle activity (Escamilla et al., 2009). Participants repeated the static and dynamic exertions until one of two stoppage criteria were met: (1) verbal declaration of inability to continue, (2) failure to perform either task with adequate form despite verbal encouragement (Ebaugh et al., 2006a,b). Failure was defined as the inability to preserve 45° of forward flexion for the full 60 s during static exertions, or inability to reach 90° for two consecutive repetitions during dynamic exertions. Verbal feedback was provided to minimize extraneous postural movements such as trunk extension or contralateral leaning.

2.5. Quantifying fatigue

To assess the level of fatigue, subjective and quantitative measures were collected. Ratings of perceived exertion (RPE) (0 = "nothing at all", 10 = "extremely strong") were collected based on the Borg CR-10 scale throughout the protocol (Robertson & Noble, 1997). In addition, maximal and submaximal reference flexion exertions were collected immediately before and after the pre-fatigue and post-fatigue work protocols. Maximal forward flexion strength was measured at 45° of glenohumeral elevation in the sagittal plane using a handheld force gauge (M5-200, Mark-10 Corporation, Copiague, NY, USA). For the submaximal static reference exertions participants elevated their arm to 90° in the sagittal plane for 5 s. Both exertions were performed with the hand open, forearm in neutral rotation (thumb pointing up), and elbow in full extension. EMG amplitude and median power frequency were evaluated with a submaximal reference exertion (5 s arm elevation with middle finger pointed at the drill press task). Criteria for muscles to be considered fatigued were a significant increase in mean normalized amplitude and a significant decrease in median frequency during the central 3 s of the submaximal reference exertion.

2.6. Data analysis

Marker data were exported into Visual 3D (C-Motion, Germantown, MD, USA) and segments were modeled in accordance with ISB recommendations (Wu et al., 2002, 2005). Segments included the pelvis, thorax, clavicle, scapula, humerus, forearm, and hand. Angles were calculated in three dimensions for the wrist/forearm, elbow, humerothoracic shoulder (humerus relative to thorax), glenohumeral shoulder (humerus relative to scapula), sternoclavicular (clavicle relative to thorax), scapular (scapula relative to thorax), relative thorax (thorax relative to pelvis), and absolute thorax (thorax relative to global axis). Processed data were dual-pass Butterworth filtered (2nd order, f_c = 10 Hz). Raw EMG data were full wave rectified and low pass filtered with a 2nd order Butterworth filter (f_c = 4 Hz) and normalized to 100% MVE. To quantify muscle fatigue, median power frequency (MDF) and normalized EMG amplitude were calculated for the submaximal static reference exertions. A power spectral analysis was performed on the middle 3 s window for each muscle using a Fast Fourier Transformation and the median power frequency (MDF) was calculated (0.125 s sliding rectangular window and 0.0625 s window overlap).

The current analysis focused on the immediate effects of the fatigue protocol and thus only describes the data collected from the final eight pre-fatigue cycles and first eight post-fatigue work cycles. To evaluate the effects of the fatigue protocol on kinematics and muscle activity the post-fatigue work cycles were compared to the pre-fatigue work cycles. Analysis of the full 60 post-fatigue cycles may be found in the companion paper (McDonald et al., submitted). One-way repeated measure ANOVAs found no statistical differences between the 8 pre-fatigue work cycles, thus a mean was calculated for each variable and a single pre-fatigue value was compared to each post-fatigue work cycle. Handle pull and push tasks were decomposed into their respective "loaded" and return phases resulting in five analyzed tasks: drill press, handle pull, handle pull return, handle push, handle push return.

A series of one-way repeated-measures ANOVAs and pre-planned comparisons with Tukey's HSD tests were conducted to compare pre-fatigue mean EMG and joint angles during each task to the post-fatigue values. Effect sizes were calculated and Eta-squared (η^2) values were reported for significant variables. Effect sizes were interpreted as small (0.01–0.08), medium (0.09–0.24) or large ($\geqslant$0.25) (Field, 2013). All statistical tests were conducted in SPSS Statistics (v20.0, IBM, NY, USA) with $\alpha = 0.05$.

3. Results

3.1. Effects of fatigue protocol

The duration of the fatigue protocol ranged from 5 to 30 min (14.62 ± 9.15 min, mean ± standard deviation). Ratings of perceived exertions increased from 1.8 ± 0.9 (weak) in the pre-fatigue trials to 5.9 ± 2.1 (strong to very strong) immediately after the fatigue protocol ($p < 0.05$). During the fatigue protocol, RPE scores reached near maximal levels for all participants. Maximum flexion strength decreased from 14.9 ± 2.6 kg at the end of the pre-fatigue work cycles to 9.2 ± 3.0 kg immediately after the fatigue protocol ($p < 0.001$). Fatigued muscles, defined as those that demonstrated both frequency and amplitude changes following the fatigue protocol, were anterior deltoid, posterior deltoid, latissimus dorsi, and serratus anterior ($p < 0.05$).

3.2. Kinematic adaptations

Significant changes in joint kinematics between pre-fatigue and post-fatigue work cycles were found following the fatigue protocol (Table 1, Supplementary Tables S1–S5). While the magnitude of the changes was small, the effect sizes of the significant kinematic variables ranged from medium to large ($\eta^2 = 0.12–0.34$). During the drill task, there was a significant reduction in mean glenohumeral flexion angle post-fatigue versus pre-fatigue trials (by as much as 9°) (Fig. 3). Superior scapular rotation (+3.4°), sternoclavicular elevation (+0.6°) and retraction (+3.2°) increased significantly in post-fatigue trials (Fig. 4). In the pulling task, there was a significant reduction of 3.0° in absolute trunk flexion post-fatigue. In the return phase of the pulling task, anterior scapular tilt increased significantly (+1.3°) with a significant decrease in sternoclavicular elevation (2.8°). In both phases of the pushing task, glenohumeral flexion decreased by 5.4° while scapular superior rotation increased by about 4° ($p < 0.05$). For the return phase, significant changes were found in glenohumeral adduction, glenohumeral internal rotation, humerothoracic extension, and relative trunk bending.

3.3. Muscular adaptations

Significant changes in muscle activity were found in the pulling and pushing tasks in the post-fatigue work cycles (Table 2). As with the kinematic changes, the magnitudes of the significant EMG changes were small as well but the effect size ranged from medium to large ($\eta = 0.19–0.49$). In various work cycles, during both phases of the pulling task, muscle activity increased significantly in the posterior deltoid (1.3–1.5% MVE) and sternal head of pectoralis major (~0.7% MVE). In the return phase of the pushing task, posterior deltoid and triceps brachii muscle activity increased significantly by 1.0% and 1.9% MVE, respectively. However, triceps activity was increased only in the work cycle immediately following the fatigue protocol, while the posterior deltoid activity remained elevated for all 8 post-fatigue work cycles. A comprehensive summary of EMG variables can be found in Supplementary Tables S6–S10.

Table 1
Statistically significant changes in mean joint angle in the post-fatigue (PF1–PF8) work cycles compared to pre-fatigue work cycles. Significant changes are denoted with bold font, rows without notation had a main effect and no significant post hoc tests. Complete mean data with standard deviation are provided in the supplementary tables.

Task	Angle	PRE	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Task 1 pull	Scapular anterior tilt	2.7 ± 3.9	3.3 ± 4.1	3.6 ± 4.0	3.7 ± 4.1	3.7 ± 4.1	2.5 ± 4.5	2.8 ± 3.7	2.8 ± 4.0	2.9 ± 4.1
	Absolute trunk flexion	9.3 ± 6.1	9.8 ± 5.7	7.8 ± 8.3	**6.3 ± 7.8**	**6.4 ± 7.5**	9.3 ± 6.0	6.8 ± 7.4	7.5 ± 6.2	**6.3 ± 5.4**
Task 1 return	Humeral internal rotation	63.7 ± 9.7	60.2 ± 10.8	63.0 ± 10.2	62.9 ± 9.4	63.6 ± 10.5	63.1 ± 9.5	66.1 ± 6.9	64.5 ± 7.3	64.9 ± 8.7
	Scapular anterior tilt	2.8 ± 3.8	**4.2 ± 4.1**	3.7 ± 3.9	3.5 ± 4.0	3.6 ± 3.9	3.1 ± 4.3	2.6 ± 3.9	2.8 ± 3.7	2.7 ± 3.9
	SC elevation	11.5 ± 3.1	**8.7 ± 3.8**	9.9 ± 4.5	10.1 ± 4.2	10.0 ± 3.9	10.0 ± 3.6	12.1 ± 3.9	11.3 ± 5.2	11.3 ± 5.2
Drill	GH flexion	31.5 ± 8.5	**24.8 ± 11.1**	**23.7 ± 10.4**	**23.0 ± 9.9**	**24.1 ± 9.5**	**24.3 ± 8.2**	**24.6 ± 8.2**	**22.5 ± 7.7**	**24.1 ± 8.3**
	Scapular superior rotation	32.3 ± 10.8	**35.7 ± 10.4**	34.1 ± 11.0	35.0 ± 10.3	**35.2 ± 10.6**	**35.7 ± 10.3**	**35.3 ± 10.9**	**35.5 ± 10.7**	**35.7 ± 10.5**
	SC elevation	5.4 ± 5.6	**6.0 ± 5.3**	5.1 ± 5.6	5.5 ± 5.2	6.1 ± 5.9	7.4 ± 6.4	7.3 ± 7.1	7.1 ± 6.8	7.3 ± 6.7
	SC retraction	4.4 ± 5.7	**7.6 ± 5.8**	6.0 ± 7.3	**7.1 ± 6.9**	6.5 ± 6.9	6.4 ± 5.9	5.6 ± 6.3	**6.5 ± 6.4**	6.0 ± 6.5
	Relative right trunk rotation	5.2 ± 4.2	7.2 ± 6.4	6.6 ± 5.6	6.9 ± 5.6	5.9 ± 6.1	5.5 ± 6.0	3.9 ± 6.2	5.3 ± 6.2	5.2 ± 4.9
Task 4 push	GH flexion	32.4 ± 9.8	30.4 ± 9.5	29.9 ± 10.3	**28.5 ± 9.9**	**27.1 ± 10.0**	**29.4 ± 8.2**	**28.8 ± 10.7**	**28.1 ± 9.3**	**28.6 ± 9.0**
	Scapular superior rotation	28.2 ± 10.6	**31.7 ± 10.8**	**30.8 ± 10.4**	**31.2 ± 10.8**	**30.6 ± 10.8**	**32.0 ± 11.5**	**31.3 ± 10.6**	**31.6 ± 11.0**	**31.1 ± 9.7**
	Relative left trunk rotation	1.6 ± 2.8	0.2 ± 3.7	2.0 ± 3.8	2.4 ± 3.4	1.2 ± 3.9	0.6 ± 3.4	3.2 ± 4.3	2.5 ± 3.8	2.1 ± 3.8
Task 4 return	GH abduction	49.0 ± 17.7	47.2 ± 21.6	37.4 ± 33.4	36.4 ± 36.6	33.1 ± 33.3	47.3 ± 26.7	27.8 ± 50.2	26.6 ± 45.0	28.5 ± 44.4
	GH flexion	33.7 ± 11.0	34.0 ± 10.4	30.1 ± 9.1	30.4 ± 11.5	30.7 ± 11.4	31.2 ± 9.9	**28.3 ± 8.8**	**28.4 ± 9.3**	**28.5 ± 8.8**
	GH internal rotation	35.2 ± 18.7	34.9 ± 22.4	26.2 ± 32.6	25.2 ± 35.2	22.3 ± 32.3	34.7 ± 26.6	17.3 ± 47.2	15.6 ± 42.5	18.1 ± 40.4
	Humeral flexion	55.0 ± 9.9	59.6 ± 5.1	51.2 ± 13.8	53.9 ± 11.0	54.8 ± 10.9	57.5 ± 6.1	48.6 ± 11.6	50.0 ± 11.2	49.7 ± 10.7
	Scapular superior rotation	28.9 ± 10.5	**32.8 ± 10.4**	30.3 ± 11.3	**31.8 ± 10.4**	**31.9 ± 10.4**	**32.9 ± 11.2**	31.2 ± 10.3	**31.7 ± 10.7**	31.2 ± 9.8
	Relative right trunk bend	0.5 ± 2.1	2.3 ± 2.7	1.2 ± 2.6	0.7 ± 2.8	1.2 ± 3.6	0.2 ± 3.3	0.4 ± 4.1	0.4 ± 3.6	0.7 ± 3.4

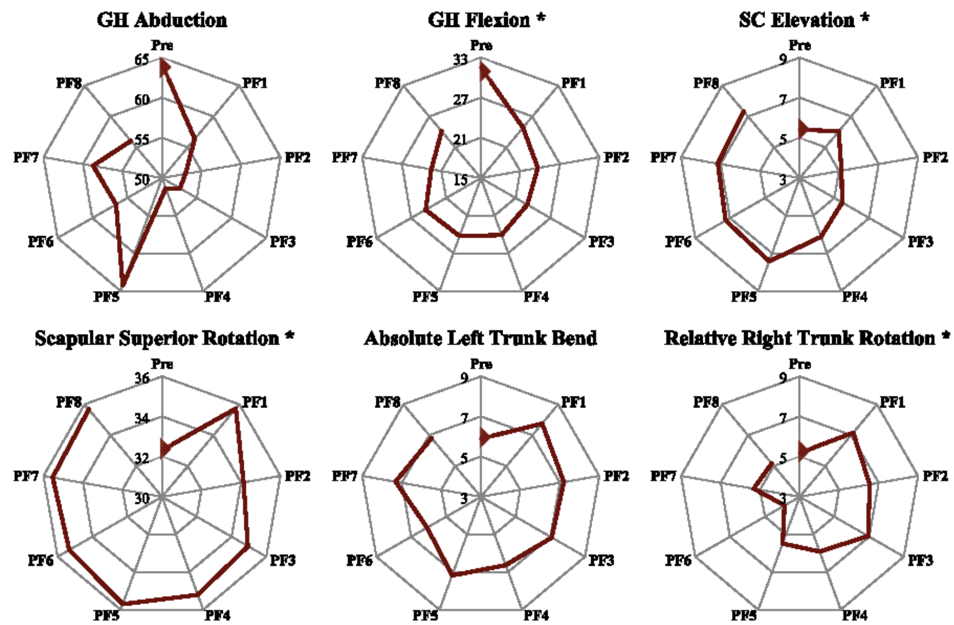

Fig. 3. Polar plots displaying mean joint angle values during pre-fatigue and post-fatigue work cycles during the drill task. The combination of these kinematic changes reduced the demand on the fatigued anterior deltoid while continuing to fulfill the task demands of the drill task. Axes are in degrees and scales have been adjusted in each plot. Work cycles are shown from Pre-fatigue (PRE) to Post-fatigue 8 (PF1-PF8) in a clockwise direction. Angles with significant post hoc tests are indicated with *.

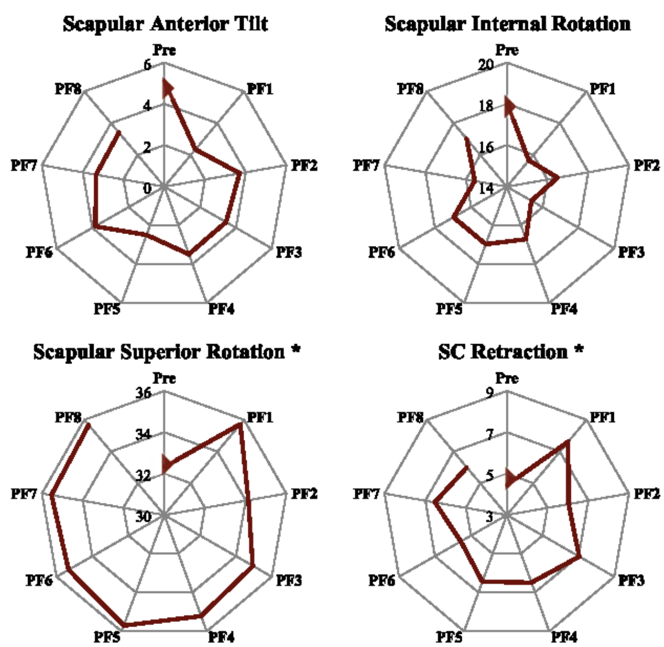

Fig. 4. Polar plots comparing mean angle values associated with scapular reorientation during pre-fatigue and post-fatigue work cycles of the drill task. The increase in scapular superior rotation and sternoclavicular retraction and decrease in scapular anterior tilt and internal rotation all contribute toward expanding subacromial space width. Axes are in degrees and scales have been adjusted in each plot. Work cycles are shown from Pre-fatigue (PRE) to Post-fatigue 8 (PF1–PF8) in a clockwise direction.

Table 2
Statistically significant changes in mean muscle activity in the post-fatigue (PF1–PF8) work cycles compared to pre-fatigue work cycles. Significant changes are denoted with bold font, rows without notation had a main effect but no significant post hoc tests. Complete mean data and standard deviations are found in the supplementary tables.

Task	Muscle	PRE	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Task 1 pull	Posterior deltoid	7.8 ± 4.4	8.6 ± 4.8	8.4 ± 4.6	8.5 ± 4.6	8.6 ± 4.6	8.7 ± 4.6	9.0 ± 4.4	9.2 ± 4.4	**9.3 ± 4.2**
	Pectoralis major sternal	3.9 ± 2.5	4.5 ± 2.7	4.4 ± 2.8	**4.6 ± 2.7**	4.4 ± 2.4	4.1 ± 2.4	3.7 ± 2.5	3.8 ± 2.4	3.7 ± 2.5
Task 1 return	Posterior deltoid	7.6 ± 4.1	8.3 ± 4.5	8.3 ± 4.5	8.3 ± 4.5	8.3 ± 4.6	8.4 ± 4.5	8.8 ± 4.2	**8.9 ± 4.3**	**8.9 ± 4.3**
	Pectoralis major sternal	3.8 ± 3.4	3.9 ± 2.8	4.2 ± 3.5	**4.6 ± 2.9**	4.3 ± 3.4	4.3 ± 3.4	3.7 ± 3.1	3.5 ± 3.3	3.6 ± 3.1
Task 4 return	Posterior deltoid	7.7 ± 4.1	**8.5 ± 4.4**	**8.6 ± 4.3**	**8.5 ± 4.4**	**8.6 ± 4.4**	**8.7 ± 4.4**	**8.5 ± 4.5**	**8.5 ± 4.5**	**8.5 ± 4.5**
	Triceps	5.9 ± 2.7	**7.8 ± 3.7**	7.4 ± 3.2	7.3 ± 3.5	7.0 ± 3.1	7.2 ± 3.4	6.7 ± 3.2	7.1 ± 3.5	6.1 ± 3.9

4. Discussion

The aim of this investigation was to identify adaptive changes in joint kinematics and muscle activity immediately following isolated anterior deltoid fatigue. All participants maintained task performance during post-fatigue simulated work despite reaching failure during the fatigue protocol. Some adaptations were task-specific while others acted across multiple tasks. Identifying adaptive changes in movement patterns is a critical step toward recognizing objective signs of reduced capacity in workers that may lead to injury. Changes in posture are often observable and could be a used as indicators of fatigue in an occupational setting.

In the current investigation, completion of the drill task required the generation of an anterior force while elevating the arm to position the hand and drill above shoulder height. With anterior deltoid fatigue, participants opted for a strategy of reduced glenohumeral flexion and increased relative rightward trunk rotation. This kinematic combination reduces the external flexion moment demands about the shoulder while facilitating successful completion of the task despite significant muscle fatigue. Previous studies have shown similar load-reducing strategies, which may represent attempts to search for alternative movement patterns that reduce the demands from fatigued muscles. Fuller et al. (2011) found that participants were able to continue performing a repetitive anterior–posterior reaching task despite fatigue by increasing shoulder and elbow movement variability in the medial–lateral and superior–inferior planes. With repetitive hammering, participants compensated for shoulder fatigue by decreasing elbow and shoulder movements while increasing trunk movement (Côté et al., 2005). Individuals displayed different kinematic changes following the fatigue protocol, making it difficult to predict specific joint angle changes across the entire group. It is possible that multiple small single joint changes sum to create a more meaningful resultant effect on the movement envelope (Fig. 3) and that individuals apply the degrees of freedom of the upper extremity differently to adapt to fatigue.

Scapular kinematics are an important area of focus considering many shoulder pathologies are linked to changes in SAS width, which can be altered with scapular orientation and muscle action (Chopp and Dickerson, 2012; Chopp et al., 2011). Motions that reduce SAS width include scapular inferior and internal rotation, scapular anterior tilt, and/or superior translation of the humeral head (Banas et al., 1995; Chopp and Dickerson, 2012; Solem-Bertoft et al., 1993). In the current investigation, the increased scapular anterior tilt observed during the pulling task may be considered a maladaptive consequence of fatigue. However, this was only seen in the first post-fatigue work cycle and returned to pre-fatigue values by the eighth post-fatigue work cycle, suggesting a transient effect. To offset this potentially maladaptive effect, other kinematic changes may have acted to increase SAS width. In both the drill press and pushing tasks, participants exhibited increased superior rotation (Fig. 4). This change represents a possible impingement sparing adaptation that expands the SAS by shifting the anterolateral aspect of the acromion away from the humeral head (Ludewig and Cook, 2000; Solem-Bertoft et al., 1993). These findings support previous investigations that revealed impingement-sparing scapular adaptations following fatigue protocols in simple tasks (Chopp et al., 2011; McQuade et al., 1998).

In consideration of related movements that affect the scapula, there was a significant increase in sternoclavicular retraction that occurred in conjunction with the scapular changes during the drill task. Since the clavicle joins the scapula at the acromioclavicular joint, these movements are linked, thus the scapula would retract with the clavicle, resulting in the decreased scapular internal rotation observed (Fig. 4) and possibly increasing the SAS. During active arm elevation sternoclavicular retraction has been found to be greater than when moved passively (Ludewig et al., 1996). Similarly, other SAS expanding scapular and clavicular movements have been found during active arm elevation versus passive motion, suggesting that active muscle forces are critical in scapulothoracic adaptations to fatigue (Ebaugh et al., 2005).

We evaluated muscle activity to provide insight to potential muscle recruitment compensation strategies. In the pushing and pulling tasks, significant changes were observed in posterior deltoid, pectoralis major and triceps muscle activity following the fatigue protocol. Considering that these tasks were physically constrained, a reorganization of muscle activity may suggest a redistribution of workload in response to fatigue. In muscles that demonstrated significant signs of muscle fatigue (anterior deltoid, latissimus dorsi, serratus anterior), an increase in EMG amplitude was expected but not observed during post-fatigue work cycles. The absence of increased EMG amplitude suggests that these muscles were less active and potentially "protected" by other muscles to allow recovery following. Similar multi-joint, multi-muscle strategies to redistribute demands have been observed in a repetitive hopping task, in which participants increased vastus lateralis activity to compensate for the fatigued gastrocnemius (Bonnard et al., 1994).

There are limitations to the current investigation. First, the post-fatigue changes were elicited using an anterior deltoid fatigue protocol. While likely not representative of fatigue that may develop in the workplace, it provided valuable insight into possible adaptations to fatigue. Second, we did not use fine wire EMG, thus the infraspinatus, available with surface EMG, was the only rotator cuff muscle included in this investigation. These muscles are important in workplace shoulder injuries and future work should attempt to include them to comprehensively examine muscle activity redistribution strategies. This study was limited to relatively young male participants, thus caution should be used when applying to all members of the workforce. Finally, due to the large volume of data, only the mean changes were presented, thus some within cycle information may have been lost.

In conclusion, following the onset of fatigue in anterior deltoid, participants were able to successfully complete post-fatigue work cycles by changing muscle recruitment and kinematic strategies. The findings suggest that people adapt to fatigue by repositioning joints to minimize the workload on fatigued muscles. Scapular reorientation occurred in all of the simulated tasks and generally served to increase the width of the subacromial space. While means and joint ranges of motion provided some insight, they did not always describe a clear picture statistically, thus innovative analyses of joint motion are warranted. Further research of posture and task demands on scapular orientation would provide insight into the mechanisms leading to the development of workplace shoulder injury with fatigue.

Conflict of interest

The authors declare that there are no conflicts of interest.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jelekin.2015.07.003.

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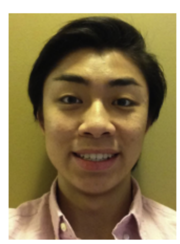


Calvin Tse received his Bachelor of Science Kinesiology (Hons., 2014) from McMaster University. He is continuing his studies at McMaster University as an MSc student in the Department of Kinesiology. He hopes to continue his research in upper extremity biomechanics and digital modelling.

Alison McDonald received a Bachelor of Science (Kinesiology, Honors Cooperative Program, 2011) from the University of Waterloo. She started her MSc at McMaster University in 2011 and is currently a PhD student holding an NSERC post-graduate scholarship.

Peter Keir received his PhD from the University of Waterloo. He is currently a Professor in the Department of Kinesiology at McMaster University in Hamilton, Ontario. His research examines upper extremity mechanics and function using EMG, imaging and modelling to determine the mechanisms of work-related musculoskeletal disorders, with an emphasis on carpal tunnel syndrome and muscle-related injuries of the arm and hand. He is a former President of the Canadian Society for Biomechanics.

APPENDIX H: SUPPLEMENTARY TABLES FROM CHAPTER 2

Supplementary Table 1.1. Mean and standard deviation of kinematics for drill task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value. All values are in degrees.

Angle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Wrist Ulnar Deviation	11.91±12.23	9.60±13.28	16.72±5.80	16.16±9.03	15.02±9.28	14.45±11.20	15.01±8.00	15.27±7.60	15.15±7.26
Wrist Extension	17.72±12.94	18.02±13.13	12.43±7.50	11.25±9.02	13.05±6.44	13.33±8.38	12.62±7.39	13.44±7.61	13.25±7.39
Wrist (Forearm) Pronation	-6.07±4.75	-6.51±4.54	-6.27±5.22	-4.92±6.35	-5.72±5.30	-6.78±5.13	-5.85±4.86	-5.46±5.52	-5.60±4.98
Elbow Extension	-63.89±9.40	-62.53±10.80	-63.16±9.76	-59.07±7.40	-59.11±9.16	-63.16±10.22	-58.24±12.24	-59.50±9.12	-59.02±9.82
GH Adduction	-57.92±32.53	-50.99±39.52	-65.75±43.58	-58.85±33.98	-60.22±46.94	-64.92±39.99	-63.45±42.56	-67.86±37.56	-70.05±41.47
GH Extension	-31.51±8.51	**-23.90±9.66**	**-25.41±7.66**	**-25.89±10.60**	**-26.75±8.83**	-29.38±9.52	**-25.39±8.92**	-27.22±8.98	-27.28±8.56
GH Internal Rotation	42.41±32.42	38.56±39.37	50.24±43.21	48.38±45.00	43.75±45.66	48.31±38.08	46.66±41.29	50.23±36.15	52.19±40.13
Humeral Adduction	-61.22±14.26	-62.75±14.37	-65.22±17.48	-61.83±21.81	-64.24±24.22	-65.30±27.00	-66.36±19.43	-66.22±21.30	-70.23±21.94
Humeral Extension	-58.74±10.58	**-52.58±8.66**	-56.35±7.58	-55.41±8.51	-55.08±8.84	-59.04±9.99	-56.07±10.17	-58.91±8.21	-58.93±8.14
Humeral Int. Rotation	70.45±11.74	71.23±12.81	68.82±12.65	67.32±13.38	68.47±13.09	67.58±13.87	69.19±12.61	68.53±13.84	70.38±12.80
Scapular Inf. Rotation	-32.33±10.84	-35.00±10.41	**-37.03±10.69**	**-36.90±11.27**	**-36.99±11.53**	**-37.53±11.93**	**-37.32±10.63**	**-37.35±10.41**	**-37.97±11.01**
Scapular Anterior Tilt	5.12±3.65	3.25±4.84	3.19±3.68	3.22±4.48	3.52±3.89	3.59±3.73	3.55±3.98	3.71±4.12	3.37±3.91
Scapular Int. Rotation	18.07±5.01	16.06±6.92	13.55±5.69	15.57±7.13	15.39±6.94	13.72±7.41	14.35±5.95	14.11±5.84	13.25±5.91
SC Depression	-5.42±5.60	-5.69±5.31	-6.93±6.84	-7.20±7.10	-6.92±6.63	-6.69±7.04	-6.40±6.99	-6.84±6.46	-6.73±6.41
SC Protraction	-4.44±5.70	-6.80±6.57	**-8.89±7.40**	**-7.85±7.34**	-7.26±6.42	**-8.29±7.26**	**-7.82±6.73**	**-7.77±5.89**	**-8.34±5.84**
Absolute Right Trunk Bend	5.88±4.68	7.17±7.02	9.89±6.23	9.29±5.28	7.44±5.22	7.81±5.30	9.75±5.19	9.25±4.22	10.32±4.62
Absolute Trunk Flexion	1.40±5.61	3.11±7.71	2.72±6.11	3.25±7.11	2.25±6.15	2.83±7.15	1.66±7.25	2.00±5.76	1.67±6.14
Absolute Left Trunk Rotation	-2.21±7.16	-1.00±7.35	-3.88±5.90	-4.24±6.95	-1.46±6.89	-4.05±8.84	-4.02±8.61	-4.30±11.12	-4.09±10.46
Relative Right Trunk Bend	-6.52±5.10	-8.18±8.18	-11.24±5.51	-10.04±5.08	-9.25±4.41	-9.56±4.32	-10.61±5.51	-9.97±5.30	-11.56±4.87
Relative Trunk Flexion	-14.80±4.60	-15.90±6.24	-16.44±3.67	-17.36±3.72	-15.99±4.04	-17.23±4.04	-15.80±4.47	-16.36±3.32	-15.79±3.79
Relative Left Trunk Rotation	-4.56±3.70	-6.08±5.70	-8.17±4.60	-7.78±4.21	-6.83±5.00	-7.90±5.28	-7.21±3.58	-7.23±3.16	-8.42±3.15

Supplementary Table 1.2. Mean and standard deviation of kinematics for the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value. All values are in degrees.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Deviation	2.35±3.56	2.56±5.44	2.95±4.05	3.17±2.55	4.36±4.24	2.59±3.99	4.11±3.98	2.26±4.04	3.91±4.48
Wrist Extension	19.82±4.78	18.80±5.37	**16.03±5.51**	**14.65±5.79**	**15.34±7.16**	**14.97±7.94**	**16.11±6.69**	**13.62±5.21**	**15.95±6.12**
Forearm Pronation	-4.74±5.63	-4.20±5.15	-4.03±4.20	-3.31±3.86	-3.54±3.66	-4.40±3.93	-4.23±4.72	-3.71±4.01	-4.59±4.18
Elbow Extension	-43.53±8.21	-42.70±13.82	**-37.42±8.09**	-36.20±12.40	**-37.84±11.31**	-34.50±10.84	**-39.01±12.09**	-33.77±7.69	**-37.65±11.07**
GH Adduction	-29.45±22.96	-34.24±15.95	-31.68±22.09	-33.34±28.40	-31.54±33.83	-30.70±26.38	-27.49±24.56	-34.70±22.50	-35.82±23.98
GH Extension	-29.89±6.38	-30.86±6.72	-27.84±7.00	-29.15±5.09	-29.40±6.69	**-26.29±5.10**	**-26.53±5.77**	-28.47±5.19	**-26.04±4.58**
GH Internal Rotation	16.74±22.67	21.74±15.95	18.60±22.40	22.05±26.08	20.06±30.65	19.15±24.70	16.24±23.63	23.18±22.59	23.09±23.01
Humeral Adduction	-40.85±4.20	-39.87±7.43	-43.14±5.74	-43.57±6.87	-43.47±8.55	-44.39±4.94	-42.69±5.30	-42.91±5.54	-46.50±6.65
Humeral Extension	-53.87±8.18	-56.63±8.94	-52.56±9.41	-54.81±9.39	-54.89±9.68	-51.25±9.09	-51.92±7.28	-54.98±7.52	-53.30±6.87
Humeral Int. Rotation	64.05±9.56	64.71±9.27	66.33±8.85	**68.23±9.93**	**67.95±10.59**	**67.58±9.37**	66.68±9.57	**68.41±9.95**	**68.24±8.83**
Scapular Inf. Rotation	-35.73±10.20	-36.14±10.93	-35.10±9.69	-36.73±9.71	-37.40±10.17	-35.99±9.91	-37.23±10.49	-36.62±9.82	-36.56±9.18
Scapular Anterior Tilt	2.69±3.86	3.57±4.04	2.93±3.73	2.65±3.90	2.59±4.45	2.42±4.60	2.16±4.26	2.61±3.95	2.48±3.93
Scapular Int. Rotation	29.10±4.06	29.22±7.31	30.11±3.95	30.01±4.54	29.46±4.79	29.35±6.16	29.08±4.65	30.16±5.26	28.95±5.46
SC Depression	-11.82±2.79	-11.06±4.13	-11.20±4.83	-11.80±5.34	-11.47±4.46	-10.81±5.47	-11.46±5.39	-12.10±5.40	-11.36±4.68
SC Protraction	3.53±5.15	3.91±4.79	4.09±3.90	3.39±5.36	2.81±6.25	3.54±5.53	2.81±5.57	3.58±5.62	2.88±5.46

Absolute Right Trunk Bend	-0.86±4.51	-2.12±3.66	-1.57±4.13	-1.17±3.45	0.39±5.31	-0.63±5.12	-0.22±5.29	-0.10±4.85	0.46±5.34
Absolute Trunk Flexion	-9.28±6.08	-7.56±6.97	-7.25±4.04	-6.93±4.04	-6.34±7.54	-7.24±5.52	-6.62±5.55	-8.24±6.43	-7.72±6.36
Absolute Left Trunk Rotation	23.61±5.67	22.76±6.39	24.74±6.30	25.19±5.94	22.34±5.83	23.50±7.42	22.05±6.23	22.77±6.80	25.53±7.05
Relative Right Trunk Bend	0.97±4.43	1.69±4.10	1.31±4.53	0.70±4.08	-0.75±5.43	0.47±4.46	-0.02±4.42	0.05±4.26	-0.18±3.89
Relative Trunk Flexion	-6.56±4.85	-7.91±5.74	-8.02±4.41	-8.89±3.64	-9.15±4.66	-8.43±4.46	-8.40±4.69	-7.44±4.38	-7.45±4.95
Relative Left Trunk Rotation	11.15±3.84	**13.38±3.20**	11.76±3.45	11.07±3.59	9.55±4.79	11.00±3.81	10.25±3.71	10.94±3.87	10.67±4.21

Supplementary Table 1.3. Mean and standard deviation of kinematics for the return phase of the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value. All values are in degrees.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Deviation	2.34±4.18	3.07±4.94	2.89±3.41	3.20±2.52	4.93±4.49	3.03±3.92	4.77±4.23	3.25±4.19	3.92±5.04
Wrist Extension	20.56±5.09	20.82±5.61	**15.74±5.21**	**14.26±7.04**	**16.09±7.17**	**15.80±8.73**	**15.38±6.41**	**15.16±6.68**	**16.12±5.77**
Wrist (Forearm) Pronation	-4.86±5.64	-4.10±5.19	-3.66±4.00	-3.02±3.59	-3.86±3.75	-4.28±3.79	-4.16±4.85	-3.53±3.78	-4.63±4.72
Elbow Extension	-44.77±7.63	-49.23±13.41	**-37.51±10.42**	**-36.28±14.93**	-40.00±11.55	**-36.85±12.66**	**-37.62±11.69**	**-38.32±9.06**	-38.78±10.76
GH Adduction	-26.51±21.31	-22.66±22.79	-28.27±21.31	-30.21±18.77	-33.68±24.31	-23.74±31.84	-31.73±23.84	-23.53±27.08	-29.42±33.05
GH Extension	-29.77±6.26	-29.52±7.97	-26.55±5.31	-26.96±4.43	-28.22±7.80	-26.37±7.82	-28.58±6.42	-26.55±6.63	-27.03±5.87
GH Internal Rotation	14.00±21.76	10.65±22.40	16.10±21.57	19.24±17.98	21.96±22.46	12.38±30.25	20.18±22.28	12.94±25.39	16.51±32.53
Humeral Adduction	-40.43±4.04	-38.52±7.40	-42.17±6.97	-42.82±7.17	-42.44±8.50	-44.49±6.23	-41.56±5.83	-43.98±7.70	**-45.86±6.90**
Humeral Extension	-53.26±7.99	-51.77±11.19	-49.68±9.10	-52.15±8.38	-54.92±8.23	-50.19±12.86	-54.53±9.21	-49.45±9.49	-51.83±10.36
Humeral Int. Rotation	63.68±9.66	62.40±9.83	65.50±9.46	**67.39±9.80**	**66.92±9.80**	**67.37±9.74**	66.15±9.67	**68.20±10.46**	**66.97±9.26**
Scapular Inf. Rotation	-35.75±10.15	-35.77±11.03	-34.86±9.93	-36.23±9.94	-37.24±10.46	-36.19±9.97	-37.11±10.34	-35.84±9.96	-36.38±9.12
Scapular Anterior Tilt	2.82±3.81	3.74±3.95	3.20±3.88	2.86±3.69	2.57±4.31	2.56±4.36	2.18±4.28	3.29±4.10	2.58±4.05
Scapular Int. Rotation	28.72±4.53	27.80±6.58	29.15±4.79	29.21±4.81	29.35±4.74	28.97±5.22	29.77±4.43	28.60±5.45	28.54±5.87
SC Depression	-11.48±3.12	-9.68±3.93	-10.31±4.90	-10.86±6.26	-11.22±5.35	-10.83±4.96	-11.43±5.51	-10.42±5.24	-10.94±5.02
SC Protraction	3.38±5.08	3.48±4.17	3.78±4.26	3.13±5.74	2.78±6.55	3.36±5.69	3.36±5.65	3.34±5.35	2.86±5.50
Absolute Right Trunk Bend	-0.68±4.54	-1.61±3.52	-1.20±4.09	-0.95±3.46	0.54±5.33	-0.75±5.12	-0.36±5.18	0.23±4.77	0.61±5.43
Absolute Trunk Flexion	-9.29±6.35	-6.86±6.85	-6.28±4.13	-6.14±3.82	-6.38±7.70	-7.11±5.77	-7.01±5.61	-7.27±6.50	-7.55±6.92
Absolute Left Trunk Rotation	22.70±4.48	20.71±5.39	23.16±7.09	23.97±5.86	21.79±5.05	23.35±6.62	22.41±6.59	21.26±6.86	24.97±7.15
Relative Right Trunk Bend	0.76±4.26	1.01±3.79	0.74±4.56	0.28±3.95	-0.91±5.25	0.36±4.47	0.22±4.44	-0.55±4.01	-0.39±3.94
Relative Trunk Flexion	-6.69±5.04	-8.37±5.91	-8.69±4.55	-9.47±3.7	-9.13±4.90	-8.74±4.56	-8.12±4.69	-8.18±4.39	-7.58±5.30
Relative Left Trunk Rotation	10.55±3.22	11.79±2.95	10.89±3.55	10.28±3.26	9.16±4.56	10.85±3.62	10.58±4.28	9.90±4.28	10.31±4.88

Supplementary Table 1.4. Mean and standard deviation of kinematics for the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value. All values are in degrees.

Angle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Wrist Ulnar Deviation	-4.12±7.07	-4.94±9.96	-2.67±8.24	-3.15±7.27	-4.05±6.22	-3.87±6.14	-4.14±6.18	-3.19±10.53	-3.22±8.72
Wrist Extension	29.13±8.66	30.16±8.26	24.76±9.28	**23.38±9.65**	**23.27±9.81**	26.09±11.01	**23.82±10.45**	24.24±10.89	24.46±9.44
Wrist Pronation	-5.62±5.65	-5.95±5.61	-5.44±4.42	-4.78±4.25	-5.20±4.71	-5.05±4.22	-5.40±4.22	-4.70±4.80	-5.28±3.53
Elbow Extension	-48.56±7.10	-47.79±8.25	-41.77±10.32	-35.53±10.69	-36.09±7.99	-36.43±12.31	-37.00±9.90	-39.36±8.89	-37.74±9.15
GH Adduction	-40.79±24.63	-33.45±26.80	-38.78±24.45	-24.40±46.64	-18.30±44.99	-33.11±33.89	-31.35±33.77	-26.44±38.30	-32.14±32.07
GH Extension	-32.44±9.79	**-28.96±9.33**	**-28.63±8.45**	**-29.19±7.72**	**-27.59±8.11**	**-28.44±7.78**	**-27.35±7.63**	**-26.78±6.87**	**-27.31±6.21**
GH Internal Rotation	26.36±26.19	21.11±26.83	22.79±27.96	12.78±47.83	6.40±47.47	21.23±35.21	19.04±36.72	13.84±40.20	20.03±32.62
Humeral Adduction	-51.96±10.49	-52.34±8.88	-50.51±7.76	-57.52±20.27	-60.59±19.64	-52.27±8.66	-57.95±13.93	-56.16±8.40	-54.73±7.24
Humeral Extension	-52.79±7.81	-51.13±8.42	-51.15±7.98	-49.36±10.73	-47.57±10.90	-51.35±6.59	-50.43±8.86	-48.34±8.65	-50.61±6.90
Humeral Int. Rotation	71.29±13.06	72.26±11.58	69.59±12.11	77.40±16.97	79.71±16.68	71.94±10.93	76.46±13.22	74.22±10.32	73.44±8.82
Scapular Inf. Rotation	-28.16±10.60	**-31.11±10.47**	**-31.69±11.36**	**-31.44±10.21**	**-31.29±11.29**	**-32.92±10.78**	**-33.06±10.68**	**-31.72±9.70**	**-32.53±10.13**
Scapular Anterior Tilt	7.40±4.03	6.93±4.64	5.99±4.72	6.62±3.02	6.75±3.32	5.84±4.29	6.55±3.11	6.50±3.66	6.20±3.53
Scapular Int. Rotation	25.62±3.82	24.10±3.89	24.77±4.12	23.81±3.45	23.87±4.11	24.09±4.90	22.90±4.64	22.94±4.98	24.22±4.86
SC Depression	-4.02±3.76	-4.39±4.52	-5.73±5.10	-4.77±5.38	-4.30±5.65	-4.77±5.25	-5.01±5.07	-4.35±4.84	-5.22±5.29
SC Protraction	1.90±5.55	**-0.50±6.02**	**-0.21±5.57**	**-0.43±5.98**	0.17±5.04	**-0.53±5.00**	**-1.01±5.37**	**-0.67±5.99**	**-0.34±4.44**
Absolute Right Trunk Bend	-1.22±2.02	-1.76±1.77	-0.74±2.94	-0.65±2.80	-0.50±3.31	**0.26±3.45**	-0.02±2.17	0.01±2.29	**0.73±2.96**
Absolute Trunk Flexion	-6.41±3.88	-6.28±4.33	-5.98±5.40	-5.97±4.30	-6.29±5.52	-7.47±5.77	-7.04±5.60	-5.73±3.77	-7.09±5.12
Absolute Left Trunk Rotation	8.59±5.25	7.86±6.12	7.93±6.08	8.08±5.77	9.06±5.69	9.55±5.25	8.23±6.74	9.69±4.19	10.25±5.15
Relative Right Trunk Bend	0.04±2.01	0.99±2.83	-0.60±4.14	-0.32±2.05	-0.26±2.74	-1.66±3.17	-1.08±2.74	-1.23±2.10	**-2.04±1.99**
Relative Trunk Flexion	-9.60±3.70	-9.30±3.20	-10.45±4.23	-10.63±4.22	-10.28±4.72	-9.25±6.51	-10.32±4.58	-10.98±3.69	-9.91±4.60
Relative Left Trunk Rotation	1.58±2.77	1.44±3.35	1.04±3.88	0.44±4.28	1.08±4.45	1.20±3.81	1.20±3.16	0.52±3.65	1.33±4.42

Supplementary Table 1.5. Mean and standard deviation of kinematics for the return phase of the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value. All values are in degrees.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Deviation	-4.28±7.52	-5.11±10.78	-3.05±8.70	-3.78±7.33	-3.74±6.23	-4.10±7.01	-3.79±7.15	-3.27±10.48	-3.52±9.18
Wrist Extension	29.16±8.29	29.56±8.45	24.71±10.69	25.22±9.92	22.91±10.25	26.87±10.38	24.75±10.63	24.30±11.34	24.54±9.44
Wrist Pronation	-6.13±5.57	-6.63±5.49	-5.61±4.16	-5.33±4.32	-5.21±4.54	-5.52±4.10	-5.81±4.38	-5.04±4.63	-5.19±3.51
Elbow Extension	-46.70±6.07	-44.16±6.95	**-38.35±8.93**	**-37.25±9.01**	**-37.06±8.61**	**-39.20±12.46**	**-38.04±9.85**	**-38.87±11.23**	**-38.66±9.65**
GH Adduction	-41.40±24.73	-37.51±27.98	-41.85±24.52	-25.02±45.73	-16.58±44.44	-33.97±28.31	-33.12±31.14	-24.33±41.41	-30.41±35.19
GH Extension	-33.69±11.02	-31.27±9.94	**-29.67±8.99**	**-30.01±9.68**	**-27.28±8.33**	**-28.32±8.39**	**-28.21±7.43**	**-27.14±7.39**	**-26.62±7.24**
GH Internal Rotation	27.25±26.18	25.24±27.92	25.74±28.87	13.22±46.98	5.05±46.86	22.50±30.08	20.96±34.26	12.43±43.08	17.84±37.18
Humeral Adduction	-51.07±11.63	-51.77±9.71	-50.52±8.03	-57.77±21.20	-61.91±19.76	-51.87±6.78	-56.29±14.08	-56.20±9.55	-55.48±7.73
Humeral Extension	-54.98±9.90	-54.86±8.84	-53.16±7.43	-50.85±12.20	**-47.00±11.07**	-51.35±8.36	-51.85±9.06	**-49.07±8.39**	-50.11±8.73
Humeral Int. Rotation	71.75±13.57	73.16±11.21	70.43±11.58	78.41±17.59	80.89±17.01	72.20±10.16	76.22±13.73	74.87±10.60	74.49±9.01
Scapular Inf. Rotation	-28.89±10.46	**-31.67±10.46**	**-32.23±11.28**	**-31.93±9.88**	**-31.35±11.28**	**-33.05±10.40**	**-33.45±10.67**	**-32.15±9.65**	**-32.60±10.38**
Scapular Anterior Tilt	7.14±4.10	6.66±4.56	5.74±4.63	6.49±3.16	6.86±3.31	5.98±4.28	6.43±2.90	6.21±3.40	6.30±3.36
Scapular Int. Rotation	26.52±4.42	25.50±3.64	25.64±4.57	**24.64±3.95**	**23.61±4.31**	**24.53±5.63**	**23.95±4.30**	**23.38±4.93**	**24.06±4.29**
SC Depression	-4.80±3.35	-5.53±5.04	-6.38±5.38	-5.31±5.30	-4.17±5.43	-5.01±4.55	-5.49±5.17	-4.79±5.54	-5.15±5.10
SC Protraction	2.22±5.71	**0.03±5.72**	**0.15±5.30**	**-0.01±5.84**	**0.24±4.98**	**-0.28±5.12**	**-0.57±5.27**	**-0.50±5.69**	**-0.29±4.30**
Absolute Right Trunk Bend	-1.59±1.88	-2.15±1.85	-1.22±3.13	-0.97±2.67	-0.59±3.25	**0.15±3.40**	-0.37±2.15	-0.29±2.19	**0.71±2.94**
Absolute Trunk Flexion	-6.89±4.19	-7.30±4.23	-6.55±5.35	-6.31±4.48	-6.32±5.56	-7.75±5.85	-7.12±5.51	-5.93±3.75	-7.46±5.37
Absolute Left Trunk Rotation	9.92±4.44	9.84±7.35	9.70±6.31	8.94±5.07	8.63±5.61	9.32±4.11	8.37±5.99	10.12±3.92	10.39±4.47
Relative Right Trunk Bend	0.45±2.11	1.35±2.75	-0.18±4.22	0.03±2.07	-0.10±2.85	**-1.72±3.48**	-0.71±2.61	-0.90±1.82	**-1.96±1.85**
Relative Trunk Flexion	-9.26±3.62	-8.56±3.16	-10.05±4.18	-10.29±4.07	-10.17±4.66	-8.88±7.09	-10.20±4.70	-10.85±3.79	-9.50±4.79
Relative Left Trunk Rotation	2.71±2.87	2.70±3.67	2.23±3.92	1.53±4.16	1.30±4.59	1.56±3.65	2.06±2.94	1.09±3.80	1.44±4.36

Supplementary Table 2.1. Mean and standard deviation normalized muscle activity for drill task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	13.12±4.09	**17.13±4.77**	**17.65±5.00**	**18.20±5.80**	**17.50±7.36**	**17.92±6.47**	**16.96±6.39**	**16.34±4.45**	15.80±4.46
Biceps	5.19±4.54	6.49±5.21	6.59±6.84	6.19±5.95	5.42±4.92	5.83±7.05	5.14±4.40	4.78±4.84	4.17±3.44
Infraspinatus	10.21±5.42	11.45±7.46	**12.67±6.97**	**13.37±7.50**	**12.77±8.01**	**13.32±8.32**	12.48±7.55	11.61±5.66	11.27±5.92
Lat. Dorsi	4.25±2.20	4.46±2.10	4.45±2.05	4.39±2.12	4.17±2.22	4.09±2.32	4.05±2.23	4.00±2.17	4.04±2.24
Left Lower Trap.	6.76±3.78	6.72±3.46	7.02±3.63	6.89±3.49	6.82±3.33	7.08±4.06	6.99±3.90	6.85±3.46	6.93±3.43
Left Upper Trap.	5.20±1.84	5.89±3.20	5.63±3.18	5.74±3.03	5.98±3.09	6.04±3.03	5.82±2.69	5.74±2.51	5.57±2.59
Middle Deltoid	17.21±5.39	**20.10±7.27**	**20.97±7.04**	**20.85±8.41**	**20.93±8.55**	**21.72±9.44**	**20.42±8.03**	19.38±6.82	**19.97±6.76**
Posterior Deltoid	7.99±3.88	9.15±4.21	**9.56±4.48**	**9.72±4.52**	**9.93±4.67**	**10.38±5.10**	**10.30±4.89**	**10.42±4.98**	**10.78±5.17**
Pec. Major Clavicular	11.30±6.14	10.56±4.84	10.35±6.11	11.51±6.24	10.26±6.33	11.05±6.95	10.65±6.61	10.43±6.52	9.48±5.27
Pec. Major Sternal	2.35±2.17	2.25±2.15	2.17±2.09	2.19±2.07	2.14±2.14	2.19±2.14	**2.12±2.19**	**2.03±1.96**	**2.08±2.08**
Right Lower Trap.	7.52±3.13	7.90±3.48	8.33±3.22	8.95±4.13	8.41±4.22	8.10±3.79	7.96±3.57	7.62±3.07	8.13±3.79
Right Upper Trap.	14.65±5.94	14.83±7.08	16.00±7.32	15.37±6.41	14.94±6.75	15.28±7.46	14.40±6.36	14.68±6.11	13.66±5.40
Serratus Anterior	20.63±8.77	22.08±9.11	22.74±8.95	22.56±8.73	21.08±8.94	21.66±8.52	21.08±9.03	21.21±8.82	20.77±8.04
Triceps	3.41±1.96	4.17±2.41	4.15±2.27	4.17±2.28	4.38±3.09	**4.61±3.17**	4.41±3.00	4.28±2.80	**4.51±3.02**

Supplementary Table 2.2. Mean and standard deviation normalized muscle activity for the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	4.24±5.08	3.40±3.28	3.52±3.85	3.41±4.02	4.12±4.22	3.07±3.01	3.96±4.06	3.30±3.55	2.74±1.75
Biceps	8.13±4.44	7.31±4.41	7.23±4.43	7.33±4.34	7.16±4.36	7.44±4.92	7.53±5.08	7.71±5.28	8.22±5.11
Infraspinatus	3.22±1.70	2.92±1.64	2.99±1.77	3.00±1.99	3.62±2.53	3.08±1.99	3.32±2.00	3.20±2.13	3.55±2.46
Lat. Dorsi	4.14±2.80	4.39±2.74	4.15±2.47	4.11±2.49	4.28±2.56	4.18±2.89	3.94±2.50	4.06±2.69	4.07±2.77
Left Lower Trap.	8.81±6.38	8.53±6.21	7.76±5.48	7.77±5.21	8.10±5.92	7.10±4.43	7.85±5.47	8.21±5.59	7.53±4.64
Left Upper Trap.	2.32±1.03	2.31±1.07	2.26±1.15	2.23±1.03	2.34±1.22	2.33±1.18	2.25±1.21	2.18±1.11	2.62±1.62
Middle Deltoid	8.32±5.45	7.93±5.75	8.02±5.82	7.94±5.69	8.62±5.21	7.96±5.68	8.33±5.32	8.01±5.69	8.65±5.53
Posterior Deltoid	7.82±4.44	**8.50±4.62**	**8.82±4.61**	**9.11±4.56**	**9.40±4.60**	**9.56±4.78**	**9.77±4.93**	**9.97±5.04**	**10.15±5.09**
Pec. Major Clavicular	5.93±3.16	5.75±3.28	5.33±2.95	5.25±2.95	5.36±2.82	5.42±3.33	5.59±2.91	5.49±3.24	6.24±4.51
Pec. Major Sternal	3.91±2.50	4.48±2.56	4.04±2.19	3.89±1.97	3.65±2.00	3.79±1.96	3.89±2.42	3.95±2.50	3.84±2.69
Right Lower Trap.	8.32±3.40	8.56±4.06	8.42±3.78	8.57±3.67	8.98±3.72	8.54±3.95	8.47±3.46	8.60±3.49	8.82±3.84
Right Upper Trap.	8.88±3.57	8.16±2.85	7.35±2.98	7.21±2.76	6.87±2.59	**6.31±2.72**	**6.60±2.55**	6.67±2.88	7.42±4.17
Serratus Anterior	3.04±0.90	2.79±1.32	2.65±0.92	2.36±0.95	3.65±3.48	2.51±1.18	3.12±1.90	2.58±1.22	2.67±1.47
Triceps	4.13±2.68	4.98±3.68	4.52±2.86	4.40±2.38	4.26±2.56	4.42±2.47	4.23±2.56	4.34±2.55	4.11±2.23

Supplementary Table 2.3. Mean and standard deviation normalized muscle activity for return phase of the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	3.01±1.15	2.52±1.33	2.53±1.31	2.37±1.31	3.20±3.02	2.32±1.25	2.99±2.25	2.39±1.29	3.36±3.71
Biceps	2.57±1.31	1.92±1.12	2.14±1.19	2.02±0.97	2.27±1.32	2.18±1.35	2.37±1.43	2.11±1.32	3.01±2.87
Infraspinatus	3.03±1.68	2.60±1.52	2.75±1.66	2.57±1.64	3.14±2.09	2.61±1.57	3.02±1.78	2.65±1.69	3.11±2.25
Lat. Dorsi	3.46±2.39	3.53±2.39	3.39±2.21	3.21±2.10	3.39±2.14	3.19±2.32	3.19±2.11	3.17±2.25	3.20±2.34
Left Lower Trap.	9.30±6.16	9.61±6.54	8.67±5.72	8.40±5.31	8.57±5.74	7.90±5.07	8.79±6.08	9.00±5.63	8.44±4.80
Left Upper Trap.	2.08±0.94	2.07±1.02	2.11±1.06	2.08±0.97	2.20±1.19	2.18±1.12	2.18±1.20	2.13±1.09	2.57±1.60
Middle Deltoid	8.85±5.11	8.14±5.65	8.24±5.79	8.04±5.70	8.67±5.24	8.08±5.68	8.37±5.40	8.06±5.69	8.70±5.58
Posterior Deltoid	7.63±4.13	8.27±4.53	**8.53±4.63**	**8.77±4.72**	**9.08±4.72**	**9.23±4.96**	**9.47±5.03**	**9.68±5.13**	**9.90±5.21**
Pec. Major Clavicular	4.53±2.98	4.55±3.09	4.48±3.04	4.36±2.91	4.53±2.80	4.55±3.36	4.96±3.20	4.55±3.25	5.35±4.63
Pec. Major Sternal	3.85±3.40	4.25±3.06	4.02±3.05	3.55±2.50	3.36±2.21	3.51±2.73	3.50±2.89	3.42±2.84	3.48±3.28
Right Lower Trap.	6.56±3.59	6.63±3.83	6.62±3.69	6.52±3.63	7.23±3.69	6.70±4.03	7.16±3.87	6.82±3.82	7.18±4.43
Right Upper Trap.	5.38±1.85	4.73±1.59	4.59±1.84	4.39±1.65	4.44±1.65	4.27±1.59	4.44±1.66	4.32±1.76	5.45±4.05
Serratus Anterior	3.26±1.70	2.86±1.61	2.87±1.34	2.45±1.17	3.65±3.55	2.49±1.16	3.37±1.97	2.64±1.10	2.73±1.47
Triceps	7.41±4.79	9.48±7.60	7.80±4.91	7.22±3.88	6.86±3.98	6.79±3.77	6.00±2.96	5.87±2.81	5.98±3.18

Supplementary Table 2.4. Mean and standard deviation normalized muscle activity for the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	14.06±5.11	13.15±5.90	14.52±5.75	13.74±5.71	13.43±6.14	13.71±5.87	13.90±5.97	13.67±5.62	12.71±6.52
Biceps	1.87±1.16	1.58±0.77	1.60±0.74	1.61±0.77	1.73±0.80	1.64±0.80	1.64±0.87	1.62±0.79	1.56±0.78
Infraspinatus	12.10±6.58	11.75±6.87	12.29±6.40	11.20±5.14	11.39±6.00	11.10±5.31	10.82±5.26	10.67±5.22	10.33±5.25
Lat. Dorsi	4.39±2.46	4.15±2.28	4.19±2.18	4.02±2.06	**3.95±2.26**	**3.71±2.08**	**3.81±2.09**	**3.83±2.16**	**3.65±2.06**
Left Lower Trap.	9.02±4.97	8.64±4.63	7.82±4.12	7.63±3.96	7.80±3.95	7.19±3.95	7.96±4.28	7.88±3.78	7.54±3.47
Left Upper Trap.	2.97±1.25	2.98±1.57	3.21±1.70	3.16±1.59	3.14±1.87	3.46±1.83	3.37±1.59	3.22±1.65	3.45±2.00
Middle Deltoid	23.61±9.16	22.61±10.51	22.02±10.08	**20.51±8.37**	**20.24±8.82**	**20.04±7.65**	**20.26±9.00**	**20.05±7.36**	**19.00±7.89**
Posterior Deltoid	8.16±4.07	9.07±4.43	9.09±4.50	**9.17±4.61**	**9.42±4.74**	**9.64±4.80**	**9.80±5.00**	**9.98±5.04**	**10.03±4.98**
Pec. Major Clavicular	6.21±3.58	5.84±3.88	6.15±3.01	6.53±3.14	6.66±3.79	6.81±3.61	6.75±3.36	6.59±3.55	6.05±3.45
Pec. Major Sternal	2.26±2.21	2.31±2.22	2.29±2.16	2.19±2.04	2.15±1.93	2.14±1.98	2.24±2.30	2.26±2.28	2.16±2.22
Right Lower Trap.	5.98±3.57	5.98±3.65	5.99±3.69	6.10±3.59	6.11±3.66	6.38±4.07	7.36±6.00	7.35±6.55	7.54±6.93
Right Upper Trap.	6.23±1.63	5.61±1.80	5.96±1.21	5.73±1.15	5.44±1.69	5.58±1.37	5.64±1.69	5.58±1.48	5.74±1.61
Serratus Anterior	18.72±9.81	17.33±10.21	17.96±10.71	16.73±9.95	16.56±9.57	15.98±8.77	16.13±9.83	15.34±8.65	15.48±9.53
Triceps	12.47±8.02	**16.09±9.21**	15.11±9.22	15.20±8.29	**15.31±9.12**	**15.47±8.31**	**15.41±8.07**	**15.53±7.67**	14.51±8.43

Supplementary Table 2.5. Mean and standard deviation normalized muscle activity for return phase of the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	12.16±3.56	12.89±5.00	13.60±4.35	12.95±4.78	12.52±5.70	13.21±5.21	13.15±5.61	13.43±4.84	13.06±6.72
Biceps	1.46±0.50	1.39±0.60	1.47±0.72	1.45±0.67	1.50±0.70	1.47±0.60	1.44±0.68	1.42±0.64	1.46±0.74
Infraspinatus	10.67±5.81	12.13±7.13	12.39±6.91	11.06±5.75	11.32±7.03	11.35±6.53	10.63±5.91	10.70±5.69	10.44±5.92
Lat. Dorsi	3.92±2.08	4.08±2.12	4.06±2.00	3.83±1.88	3.74±2.02	3.72±1.97	3.63±1.89	3.80±1.96	3.69±1.90
Left Lower Trap.	8.81±4.46	8.08±3.98	7.75±3.79	7.54±3.43	7.84±3.54	7.67±3.58	8.19±3.95	8.23±3.73	8.13±3.43
Left Upper Trap.	3.24±1.37	3.27±1.57	3.52±1.79	3.43±1.60	3.35±1.98	3.69±1.97	3.54±1.56	3.44±1.64	3.70±1.97
Middle Deltoid	15.75±5.88	**17.64±7.52**	**17.64±7.49**	16.17±6.60	15.93±6.59	16.37±6.19	16.37±6.34	15.94±6.21	15.41±6.04
Posterior Deltoid	7.65±4.12	**8.54±4.38**	**8.76±4.52**	**8.90±4.65**	**9.16±4.75**	**9.39±4.83**	**9.51±4.88**	**9.79±5.08**	**9.99±5.12**
Pec. Major Clavicular	8.00±3.63	7.26±3.85	7.61±2.85	7.42±2.83	7.30±3.52	7.64±3.13	7.31±3.17	7.95±3.48	7.34±3.18
Pec. Major Sternal	2.16±2.01	2.11±1.78	2.09±1.85	1.98±1.72	2.03±1.93	1.97±1.77	1.97±1.82	2.00±1.88	1.95±1.82
Right Lower Trap.	5.99±3.42	5.94±3.52	5.90±3.61	6.18±3.39	5.96±3.57	6.47±4.49	6.78±5.35	6.62±4.27	6.88±4.65
Right Upper Trap.	10.30±3.53	9.73±3.29	10.98±3.42	9.91±3.57	9.75±4.14	9.71±3.50	9.73±4.34	9.94±3.75	9.71±4.38
Serratus Anterior	14.89±6.70	16.44±8.09	16.60±7.77	14.78±6.65	15.09±7.73	15.41±7.14	14.98±7.93	14.75±6.79	14.90±7.20
Triceps	5.86±2.72	**7.36±3.32**	**7.46±3.59**	**7.39±3.23**	**7.34±3.55**	**7.44±3.32**	**7.50±3.05**	**7.50±3.08**	**7.22±3.36**

Supplementary Table 3.1. MAD and standard deviation of kinematics for drill task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Deviation	8.86±5.09	10.73±6.25	11.70±6.91	11.48±9.47	13.77±9.34	12.91±12.23	13.51±9.41	10.90±7.44	11.40±8.26
Wrist Extension	12.11±21.61	11.16±19.53	1.98±2.74	3.02±4.63	3.34±4.09	2.82±3.77	3.86±4.39	2.65±3.69	2.93±3.48
Wrist Pronation	8.29±6.88	8.46±9.63	12.22±8.34	**17.75±10.77**	**16.40±11.59**	**16.67±9.63**	13.56±9.67	**14.81±9.92**	**14.24±9.10**
Elbow Extension	110.30±43.28	102.08±42.79	126.30±20.48	120.95±36.19	111.65±30.33	124.42±42.20	119.20±24.19	120.29±27.96	123.64±28.83
GH Adduction	69.46±30.60	60.82±26.81	77.46±19.43	77.98±28.69	74.00±26.08	80.01±29.26	79.31±28.72	82.60±29.43	88.93±35.05
GH Extension	73.54±36.94	64.53±39.32	76.67±44.27	69.45±32.78	73.17±42.74	69.30±33.96	73.62±42.11	76.32±35.58	78.55±40.88
GH Internal Rotation	46.33±13.81	39.11±15.96	43.18±11.51	44.69±17.91	42.45±16.19	44.86±17.20	40.79±13.65	44.63±13.43	44.49±15.30
Humeral Adduction	27.65±12.62	31.19±15.16	32.78±25.83	37.66±32.93	34.02±30.56	36.32±33.80	34.38±33.42	36.41±33.14	36.95±34.95
Humeral Extension	73.40±22.40	71.48±22.79	80.37±14.59	76.38±15.20	76.33±21.22	76.73±20.30	79.63±16.51	79.98±17.94	82.29±18.99
Humeral Int. Rotation	78.51±6.75	73.26±10.03	70.46±12.18	**66.77±15.24**	**65.66±14.48**	**69.07±12.91**	**64.40±15.38**	**70.03±15.16**	**69.85±15.37**
Scapular Inf. Rotation	45.38±18.60	45.75±20.86	51.47±16.25	50.91±19.19	50.88±17.16	52.79±20.85	55.58±17.57	52.62±16.71	54.58±14.71
Scapular Anterior Tilt	41.59±19.49	43.00±19.54	52.48±12.57	48.35±18.78	47.79±19.72	46.62±19.40	45.70±18.39	47.57±18.62	47.93±19.88
Scapular Int.Rotation	17.98±10.36	18.68±10.94	15.04±6.51	14.75±9.29	13.99±8.13	13.21±8.32	**11.56±9.22**	12.68±9.29	12.89±8.28
SC Depression	29.41±25.25	29.13±34.91	36.84±42.52	42.07±44.77	43.18±36.46	42.16±33.30	46.39±34.78	38.83±32.04	43.58±32.63
SC Protraction	13.32±6.52	15.02±7.58	18.45±10.08	20.35±13.82	20.60±12.15	21.81±13.95	22.30±10.97	20.51±11.34	20.99±11.30
Absolute Right Trunk Bend	5.64±3.21	7.57±3.91	5.57±3.11	8.57±6.06	8.68±5.62	8.04±6.10	7.43±4.67	8.75±4.52	7.32±5.20
Absolute Trunk Flexion	8.75±6.35	8.45±7.23	12.12±8.49	12.76±11.07	12.90±9.86	14.72±10.58	11.57±9.60	11.16±8.58	9.70±9.81
Absolute Left Trunk Rotation	15.49±6.38	12.63±8.66	15.36±7.74	15.88±8.54	14.55±7.39	13.04±6.80	14.61±7.38	15.95±7.42	15.14±7.72
Relative Right Trunk Bend	20.97±8.79	18.96±10.05	23.50±9.13	27.30±11.35	22.80±11.03	23.87±15.21	22.52±9.85	25.25±14.76	24.47±14.47
Relative Trunk Flexion	21.55±12.80	23.51±13.17	29.42±5.79	29.33±8.84	28.22±8.79	27.27±9.20	28.50±10.52	27.86±10.34	29.09±8.84
Relative Left Trunk Rotation	28.34±13.50	29.44±14.83	34.68±7.53	35.02±10.37	32.99±10.21	32.97±12.80	31.15±10.51	31.84±9.58	32.14±8.44

Supplementary Table 3.2. MAD and standard deviation of kinematics for the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Angle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Wrist Ulnar Deviation	2.90±0.85	2.73±1.88	2.96±0.96	2.91±0.98	3.12±1.97	2.60±1.49	2.90±1.94	2.86±1.17	2.28±1.08
Wrist Extension	18.71±7.45	16.51±9.21	15.30±7.31	13.27±7.14	14.47±6.85	14.16±8.65	14.17±7.62	12.93±7.41	13.13±6.94
Wrist (Forearm) Pronation	7.38±5.42	6.82±6.64	6.75±5.23	6.54±4.71	8.59±6.03	7.58±4.93	8.87±6.78	7.19±4.52	9.32±6.96
Elbow Extension	46.59±13.40	42.78±22.71	39.08±14.26	39.11±18.81	42.51±16.87	35.96±16.39	43.65±18.94	34.53±12.17	41.82±18.94
GH Adduction	46.79±10.35	50.08±13.88	53.26±13.23	54.08±14.87	55.88±16.60	52.83±17.23	47.98±14.12	53.46±14.46	57.54±15.94
GH Extension	30.90±5.53	33.10±11.31	29.52±6.94	31.82±4.45	33.36±9.72	27.81±7.93	28.80±7.89	30.10±7.14	28.69±7.16
GH Internal Rotation	39.74±7.78	37.26±8.78	41.47±11.78	45.05±13.02	43.60±10.83	40.99±11.35	39.67±9.83	42.55±16.17	43.47±10.15
Humeral Adduction	42.16±5.87	41.34±7.81	44.36±7.44	45.80±7.05	**47.44±10.96**	45.28±7.42	45.70±6.29	43.96±6.92	**50.12±7.74**
Humeral Extension	55.68±7.56	60.64±12.95	54.88±9.43	57.91±11.30	58.07±13.11	53.19±10.20	54.17±6.85	57.06±5.92	56.50±7.09
Humeral Int.Rotation	61.48±11.42	62.80±9.43	63.49±10.61	65.73±10.39	63.98±12.57	64.65±11.05	63.33±12.04	66.71±11.93	64.66±11.24
Scapular Inf. Rotation	37.97±10.76	38.25±12.08	37.43±9.48	39.34±9.83	41.41±10.01	38.04±8.87	40.57±7.47	38.14±8.81	40.29±7.83
Scapular Anterior Tilt	4.82±4.28	4.64±3.47	5.01±3.55	4.81±2.64	5.94±4.06	5.33±2.07	6.29±3.26	4.89±2.43	5.89±2.84
Scapular Int. Rotation	27.19±5.32	28.57±8.91	28.13±6.20	27.45±4.49	25.21±7.50	27.43±8.27	25.68±6.98	28.98±8.27	25.51±7.04
SC Depression	13.90±4.80	13.15±6.36	13.36±5.08	14.25±6.29	15.18±4.94	12.78±5.81	14.43±5.52	13.50±4.79	15.03±5.72
SC Protraction	5.98±4.28	6.10±4.47	5.28±2.47	5.21±3.46	6.95±6.85	6.15±5.29	6.76±6.02	6.82±4.83	6.40±5.14
Absolute Right Trunk Bend	7.08±3.25	5.86±3.41	5.91±3.62	5.67±1.45	6.67±4.35	6.42±5.13	7.46±6.62	6.30±3.55	7.61±5.27
Absolute Trunk Flexion	11.68±5.18	11.08±5.57	9.94±4.54	9.90±4.90	10.37±5.68	9.05±4.80	10.61±5.30	10.53±4.79	11.79±6.59
Absolute Left Trunk Rotation	21.70±5.97	20.56±8.01	22.52±6.56	22.35±6.12	19.29±5.13	21.85±7.22	19.33±6.41	22.02±8.38	22.88±8.99
Relative Right Trunk Bend	4.74±2.22	6.43±5.37	4.29±2.78	5.19±3.25	6.84±5.95	3.76±2.89	4.92±3.44	4.65±2.30	5.54±3.02
Relative Trunk Flexion	9.80±5.00	10.94±6.79	10.98±5.10	11.80±4.26	13.73±7.40	11.75±4.44	13.03±5.33	10.25±4.63	12.05±5.20
Relative Left Trunk Rotation	10.12±4.79	11.72±4.24	9.39±4.62	9.08±3.50	8.56±4.50	9.14±4.41	8.05±3.79	10.12±5.37	8.89±4.46

Supplementary Table 3.3. MAD and standard deviation of kinematics for the return phase pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Dev	2.90±0.84	2.70±1.51	2.69±1.14	2.66±1.29	2.97±1.74	2.62±1.38	2.85±1.64	3.02±1.25	2.50±1.06
Wrist Extension	19.17±8.29	17.59±9.98	**13.97±5.81**	**11.86±7.46**	**14.96±6.95**	**14.98±8.97**	**13.07±7.29**	**14.10±8.42**	**13.72±7.61**
Wrist Pronation	7.69±5.83	7.10±7.08	6.41±4.85	5.81±4.31	9.40±6.21	7.92±4.97	9.34±6.37	7.82±4.70	9.31±7.12
Elbow Extension	48.28±13.06	53.89±21.12	39.90±17.72	37.24±22.48	45.67±16.92	40.27±17.61	41.23±20.09	42.85±11.66	43.05±18.58
GH Adduction	44.74±12.02	41.44±8.42	45.52±16.69	47.28±16.77	51.89±18.19	**54.06±18.03**	51.34±16.41	49.60±17.73	**55.49±19.43**
GH Extension	31.18±5.35	31.85±10.45	27.75±7.11	28.99±4.96	32.02±10.54	28.27±9.88	32.68±9.25	28.75±9.15	30.50±8.97
GH Int. Rotation	37.60±7.45	33.66±8.40	37.57±6.65	39.63±13.57	38.39±11.19	42.41±10.42	39.56±11.82	39.73±15.01	45.18±13.82
Humeral Adduction	41.87±6.18	40.65±7.60	43.46±7.78	44.78±7.10	46.61±11.26	46.08±8.36	45.64±5.44	45.78±8.55	**49.57±7.96**
Humeral Extension	55.29±7.18	53.89±13.58	51.03±10.82	55.13±11.11	59.05±9.93	52.11±14.03	59.07±8.82	50.18±10.74	54.77±11.54
Humeral Internal Rotation	60.90±11.34	59.58±10.75	62.44±10.44	65.22±10.08	62.35±11.78	64.39±11.17	62.44±12.41	64.72±12.02	63.09±12.28
Scap. Inf. Rotation	37.95±10.59	38.38±10.96	37.19±9.75	38.64±9.63	41.60±9.94	38.72±8.48	41.10±8.50	38.56±8.28	39.99±8.23
Scap. Ant. Tilt	5.15±4.48	4.28±3.37	4.22±2.96	4.44±2.70	6.16±4.23	5.70±1.67	6.27±3.27	5.66±2.23	6.26±2.90
Scap. Int. Rot.	26.71±7.08	25.80±8.43	26.66±5.95	26.70±5.01	24.86±8.26	26.33±7.30	26.04±7.19	25.38±7.69	25.06±7.36
SC Depress.	13.87±4.92	12.64±5.43	12.62±5.70	13.20±6.67	15.37±5.23	13.08±5.18	15.20±5.67	13.16±5.20	14.38±6.47
SC Protrac.	6.02±4.40	5.31±3.42	4.69±2.75	5.76±4.24	7.97±6.75	5.83±5.61	7.00±6.06	6.13±5.16	6.61±4.98
Ab. Rt. Trunk Bend	7.21±3.27	6.19±3.58	5.69±3.69	5.25±2.04	6.92±4.40	6.55±4.92	8.21±6.17	6.86±3.86	8.08±5.17
Ab.Trunk Flexion	11.85±5.55	10.78±5.34	9.27±4.77	8.71±4.30	11.04±5.51	9.53±4.87	10.98±5.42	10.69±4.95	11.69±6.97
Ab. Left Trunk Rot.	20.95±5.10	16.49±7.19	20.27±5.99	21.09±5.74	18.66±4.19	21.14±5.89	19.84±6.94	18.95±6.18	22.46±9.82
Rel. Rt. Trunk Bend	5.06±2.43	5.78±4.43	4.08±2.36	4.69±3.55	6.92±6.26	3.91±3.33	5.33±3.76	4.75±2.53	6.08±3.51
Rel. Trunk Flexion	10.49±5.10	12.04±6.59	11.35±5.49	12.20±4.85	14.04±8.01	12.34±4.50	13.30±6.51	12.19±4.67	12.23±6.20
Rel. Left Trunk Rotation	9.36±4.43	8.47±4.28	8.20±3.12	8.01±3.84	8.20±4.83	8.59±3.75	8.41±4.06	7.84±3.42	8.89±5.79

Supplementary Table 3.4. MAD and standard deviation of kinematics for the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Angle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Wrist Ulnar Deviation	3.47±1.58	3.07±1.53	3.58±1.47	3.88±2.18	3.94±2.63	4.16±1.87	4.12±2.39	3.12±1.05	3.95±1.73
Wrist Extension	37.36±14.80	34.52±18.46	31.72±13.77	28.45±17.81	29.75±15.28	33.29±15.76	31.31±15.27	32.76±17.99	31.18±14.59
Wrist (Forearm) Pronation	10.07±6.52	11.07±9.22	9.63±9.49	8.20±8.45	8.92±8.02	7.68±6.53	7.68±6.06	10.13±10.32	9.40±8.44
Elbow Extension	45.38±9.41	43.02±15.33	40.58±14.94	**32.48±13.58**	**31.91±10.83**	**31.02±15.44**	**31.68±10.89**	**35.96±13.47**	**34.38±12.78**
GH Adduction	49.05±16.85	43.85±15.12	48.46±17.03	52.60±22.16	55.21±21.11	49.05±16.20	49.11±17.83	49.39±19.15	51.67±19.32
GH Extension	27.84±11.12	25.57±13.68	25.87±12.63	26.03±11.81	**24.24±11.90**	**23.14±10.31**	**22.10±9.81**	**23.85±13.82**	**23.56±11.55**
GH Internal Rotation	45.89±17.50	41.05±14.76	41.50±17.88	48.32±26.57	53.57±22.45	46.33±19.08	47.40±19.45	45.98±20.24	47.82±22.09
Humeral Adduction	43.95±12.52	44.95±14.17	46.41±12.35	53.50±24.39	53.24±21.71	45.53±8.04	49.76±13.59	51.31±17.07	50.26±12.68
Humeral Extension	46.82±10.31	45.33±13.18	47.71±11.48	44.99±14.06	42.38±14.09	45.78±10.13	45.12±10.25	44.38±13.49	46.09±11.50
Humeral Internal Rotation	74.36±13.13	74.83±15.39	70.74±16.84	79.58±19.21	80.94±18.21	75.33±14.05	78.33±12.44	76.18±14.69	76.40±14.50
Scap. Inferior Rotation	22.88±10.99	**26.34±11.64**	**29.54±13.21**	**28.32±10.48**	**27.27±10.55**	**28.09±11.83**	**28.30±10.53**	**28.47±11.88**	**28.98±11.59**
Scap. Anterior Tilt	12.93±7.34	13.07±8.29	11.94±6.31	11.07±4.48	11.84±5.86	11.55±6.30	11.76±5.68	13.20±6.91	12.04±5.74
Scap. Internal Rotation	30.71±9.31	28.34±12.08	27.46±10.43	26.36±7.86	27.54±9.40	28.45±7.81	27.69±8.51	26.83±12.90	27.61±10.58
SC Depression	7.25±4.02	8.25±4.86	7.95±4.54	6.15±4.43	6.38±3.51	6.05±2.69	5.02±2.74	8.80±6.06	7.17±4.85
SC Protraction	8.31±6.96	7.92±7.52	7.74±5.63	6.17±4.74	6.52±5.30	6.44±4.14	6.41±4.83	9.01±7.65	7.21±4.72
Absolute Right Trunk Bend	8.55±5.12	9.89±6.14	7.98±5.93	8.12±5.65	8.09±6.92	7.95±5.14	6.97±4.52	9.60±8.82	9.35±7.00
Absolute Trunk Flexion	6.63±4.07	8.62±5.24	8.34±6.92	6.87±6.86	6.03±5.64	7.86±5.90	7.03±4.38	8.52±7.94	8.21±7.40
Absolute Left Trunk Rotation	14.50±4.97	14.38±8.89	12.97±5.83	13.36±6.07	14.36±6.80	14.71±6.77	13.44±7.24	15.04±8.51	14.94±7.67
Relative Right Trunk Bend	8.74±4.01	10.44±6.29	9.92±6.10	6.92±5.36	7.41±4.81	7.31±4.47	6.53±4.36	9.34±8.41	7.33±6.28
Relative Trunk Flexion	8.23±4.26	9.24±7.00	10.56±7.66	9.33±5.73	9.19±5.16	8.63±6.84	7.76±4.49	11.54±8.76	8.74±7.52
Relative Left Trunk Rotation	8.62±4.96	10.16±5.39	9.15±5.23	7.01±3.98	7.30±4.08	7.28±4.44	7.47±4.42	9.50±6.18	8.37±4.84

Supplementary Table 3.5. MAD and standard deviation of kinematics for the return phase of the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Angle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Wrist Ulnar Deviation	3.37±1.54	3.25±1.46	3.58±1.91	3.53±1.89	3.62±2.34	3.55±1.63	3.98±2.06	3.53±1.37	3.59±1.42
Wrist Extension	36.35±15.26	36.94±16.86	32.38±16.38	34.11±18.04	29.38±16.22	36.05±15.64	32.13±16.08	33.47±17.12	32.16±16.00
Wrist (Forearm) Pronation	10.28±6.90	11.54±9.99	9.71±9.14	9.21±7.90	9.19±8.12	9.42±8.07	9.53±7.40	10.52±9.75	9.69±8.67
Elbow Extension	42.86±10.92	39.46±11.79	34.59±13.88	34.14±12.19	32.19±10.49	36.73±14.28	33.02±10.12	36.42±14.69	35.29±12.89
GH Adduction	49.50±17.24	51.34±16.45	51.13±15.89	52.81±23.20	54.40±23.16	47.14±15.76	50.04±19.19	52.68±19.08	52.09±19.58
GH Extension	29.25±12.92	28.20±14.72	26.41±12.55	25.66±11.90	**23.88±12.18**	**23.60±12.66**	**23.74±11.79**	**24.20±13.93**	**22.64±13.09**
GH Internal Rotation	46.62±18.30	50.13±15.24	45.85±18.58	51.18±28.56	54.11±24.14	45.30±17.19	48.75±22.36	47.14±19.91	48.86±22.49
Humeral Adduction	43.06±14.11	45.57±15.37	45.26±12.63	51.49±25.00	55.09±22.34	44.84±8.44	48.69±14.13	51.05±16.14	51.38±13.56
Humeral Extension	49.80±11.53	49.85±14.79	49.37±9.77	45.78±13.03	42.24±14.33	45.63±13.21	47.07±12.76	45.68±14.40	45.56±13.99
Humeral Internal Rotation	74.32±14.07	77.81±15.13	72.71±17.33	81.34±18.92	81.62±18.42	75.71±14.83	77.61±14.65	76.00±15.29	77.62±14.84
Scapular Inferior Rotation	24.00±10.92	26.30±12.38	**29.00±13.14**	**27.59±10.78**	**27.52±11.10**	**27.91±11.11**	**29.26±10.89**	**29.22±12.51**	**28.99±12.54**
Scapular Anterior Tilt	12.64±7.87	14.21±7.13	12.56±6.65	12.13±6.21	11.84±5.99	12.57±7.79	11.99±6.43	13.02±6.37	12.65±6.34
Scapular Internal Rotation	31.34±10.35	30.46±12.52	29.33±11.54	28.79±9.31	27.10±9.95	29.52±10.18	28.10±8.78	27.37±11.21	27.60±10.26
SC Depression	6.99±3.86	**9.55±5.47**	7.75±4.03	6.24±4.29	6.83±3.68	6.74±4.35	5.97±3.39	8.81±5.74	7.26±5.70
SC Protraction	8.78±7.56	9.26±6.78	8.10±5.72	7.09±5.42	6.38±5.77	7.58±6.80	6.32±5.43	8.58±5.95	7.16±4.87
Absolute Right Trunk Bend	8.64±5.20	10.79±6.85	8.39±6.05	9.06±5.97	8.31±7.21	8.85±7.23	7.64±5.01	9.45±8.28	9.78±7.52
Absolute Trunk Flexion	6.92±4.24	8.89±6.63	8.22±6.53	6.53±6.80	6.26±5.90	8.12±6.88	7.79±5.53	8.17±8.41	7.82±8.00
Absolute Left Trunk Rotation	15.66±6.25	17.61±10.31	16.13±8.12	15.66±5.93	14.37±6.44	15.01±7.32	13.10±6.24	15.77±6.61	15.50±7.16
Relative Right Trunk Bend	9.22±4.42	11.85±6.30	10.86±6.01	8.43±5.84	7.55±5.08	8.01±6.04	7.14±4.86	9.23±8.09	7.74±6.48
Relative Trunk Flexion	8.28±4.99	9.37±7.91	10.24±7.66	8.56±5.48	9.30±5.22	9.93±7.38	8.31±5.70	10.48±9.49	8.88±7.86
Relative Left Trunk Rotation	9.60±5.61	11.97±5.06	10.44±6.35	9.18±5.16	7.94±4.00	7.99±5.56	7.94±4.40	9.50±5.64	8.93±4.58

Supplementary Table 4.1. MAD and standard deviation normalized muscle activity for drill task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	8.11±5.59	**11.18±6.26**	**12.43±5.83**	**12.73±6.96**	**12.31±8.43**	**13.01±7.35**	**11.96±7.39**	**11.54±5.88**	**11.47±5.84**
Biceps	0.56±0.48	1.07±1.26	1.04±1.69	1.04±1.60	0.70±0.80	0.99±1.94	0.70±0.85	0.70±1.07	0.64±0.74
Infraspinatus	6.83±5.47	7.87±6.51	**9.31±6.76**	**10.17±6.75**	**10.17±7.37**	**10.89±8.18**	8.91±7.52	8.65±5.67	8.49±6.02
Lat. Dorsi	3.73±3.25	4.05±3.61	4.41±5.39	4.25±4.24	3.75±3.38	4.16±5.27	3.67±2.92	3.51±3.65	3.16±2.14
Left Lower Trap.	3.32±2.50	3.83±2.51	4.31±3.97	4.15±3.13	3.82±3.08	4.39±3.65	3.94±3.18	3.42±2.34	4.05±2.91
Left Upper Trap.	3.71±1.98	3.93±2.05	5.27±4.90	4.32±2.73	4.61±2.47	5.05±3.29	4.34±2.36	4.50±2.36	4.34±2.42
Middle Deltoid	12.53±7.33	14.69±8.12	**16.48±8.41**	**16.25±9.53**	**16.30±9.72**	**17.35±10.11**	**15.70±9.41**	**15.30±7.82**	**15.84±7.68**
Posterior Deltoid	5.68±3.55	6.18±3.62	7.53±4.56	7.29±4.14	7.30±4.45	7.91±4.99	7.14±4.91	7.39±5.10	8.12±5.36
Pec. Major Clavicular	6.75±4.50	5.14±2.99	5.63±4.13	6.09±4.46	5.90±4.62	6.88±5.61	6.43±5.50	5.88±4.00	5.74±4.56
Pec. Major Sternal	3.58±4.02	4.44±4.73	4.88±6.27	4.50±5.23	3.74±4.28	4.14±5.87	3.60±3.71	3.26±4.20	2.58±2.50
Right Lower Trap.	4.55±2.66	4.81±3.15	5.78±4.42	6.29±4.46	5.19±4.17	6.11±4.88	5.26±3.32	5.19±2.71	5.48±3.27
Right Upper Trap.	9.71±6.89	8.56±7.71	10.09±8.11	9.39±7.20	9.52±7.01	9.76±6.21	9.33±6.77	9.76±6.26	9.50±6.21
Serratus Anterior	15.98±8.78	16.59±8.95	17.63±8.71	17.60±8.72	16.60±8.96	17.63±8.05	16.76±8.55	17.26±8.56	16.85±8.02
Triceps	4.02±3.49	4.97±3.85	5.40±5.17	4.91±4.25	4.59±3.74	4.79±5.14	4.25±3.13	4.15±3.55	3.70±2.57

Supplementary Table 4.2. MAD and standard deviation normalized muscle activity for the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	6.83±4.26	6.29±3.82	5.71±3.28	6.12±3.67	6.24±3.41	6.18±3.91	6.45±3.94	6.14±3.94	6.05±3.85
Biceps	3.94±2.80	3.80±3.21	3.67±2.90	3.81±3.07	4.08±3.22	4.02±3.58	4.09±3.58	4.26±3.83	4.28±3.47
Infraspinatus	5.51±3.94	5.11±3.69	4.56±3.00	4.88±3.44	4.67±3.12	5.09±4.21	5.13±4.03	5.11±3.95	5.38±3.97
Lat. Dorsi	5.58±2.95	4.89±2.85	4.48±2.13	4.78±2.51	4.45±2.15	5.09±3.13	4.99±3.09	5.40±3.08	5.83±2.89
Left Lower Trap.	6.25±4.21	6.00±4.52	5.32±3.82	5.69±3.96	5.81±4.57	5.85±3.32	6.09±4.46	6.13±4.52	6.17±3.54
Left Upper Trap.	5.70±4.31	5.25±3.88	4.66±3.63	4.94±4.06	4.57±3.38	5.11±4.46	5.18±4.19	5.31±4.52	5.52±4.29
Middle Deltoid	4.32±4.43	4.28±4.66	4.72±4.49	4.88±4.65	4.92±4.64	4.99±4.78	4.62±4.65	4.81±4.70	5.14±4.66
Posterior Deltoid	3.24±2.04	3.65±2.80	4.39±2.93	**4.56±2.78**	**4.54±2.90**	**4.98±2.96**	**4.59±3.10**	**4.77±3.20**	**4.91±3.15**
Pec. Major Clavicular	5.24±3.80	5.77±3.68	5.19±3.13	5.57±3.28	5.22±2.76	5.97±3.47	5.89±3.40	**6.12±3.64**	**6.49±3.18**
Pec. Major Sternal	6.02±3.50	5.53±3.79	4.97±2.92	5.27±3.53	4.84±2.90	5.41±4.08	5.28±3.97	5.66±4.00	6.18±4.01
Right Lower Trap.	4.76±3.19	5.39±3.11	5.06±2.93	5.53±2.97	5.50±2.95	6.03±3.20	5.79±3.03	5.83±3.26	6.41±3.49
Right Upper Trap.	4.25±1.75	4.04±1.35	4.00±1.53	4.38±1.70	4.36±1.97	5.07±2.09	4.55±1.98	4.78±1.98	5.15±1.76
Serratus Anterior	6.32±4.24	5.87±4.14	5.23±3.47	5.63±3.84	6.06±3.92	5.86±4.32	6.07±4.30	5.85±4.51	5.82±4.46
Triceps	5.38±3.32	4.85±3.30	4.36±2.75	4.45±2.90	3.88±2.18	4.51±3.42	4.50±3.41	4.69±3.34	5.14±3.41

Supplementary Table 4.3. MAD and standard deviation normalized muscle activity for the return phase of the pull task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	**Pre**	**PF1**	**PF2**	**PF3**	**PF4**	**PF5**	**PF6**	**PF7**	**PF8**
Ant. Deltoid	1.47±0.99	1.37±1.06	1.35±1.00	1.31±1.06	1.96±2.40	1.36±1.06	1.66±1.98	1.18±1.10	1.67±1.69
Biceps	0.76±0.65	0.41±0.47	0.47±0.48	0.46±0.49	0.47±0.55	0.49±0.54	0.55±0.55	0.45±0.44	0.70±0.93
Infraspinatus	1.66±1.27	1.55±1.22	1.62±1.28	1.56±1.24	1.91±1.61	1.62±1.21	1.77±1.51	1.47±1.37	1.86±1.56
Lat. Dorsi	1.89±2.06	2.15±1.99	1.89±1.85	1.76±1.80	1.90±1.82	1.76±1.87	1.82±1.83	1.69±1.95	2.04±2.28
Left Lower Trap.	7.24±6.37	7.99±6.39	7.00±5.39	6.96±5.26	7.03±5.59	6.26±4.87	7.04±5.71	7.58±5.63	6.65±4.97
Left Upper Trap.	1.01±0.72	0.89±0.52	0.88±0.59	0.96±0.54	0.96±0.62	1.01±0.67	1.09±0.72	1.04±0.85	1.43±1.28
Middle Deltoid	6.85±5.09	6.98±5.06	6.92±5.06	6.91±5.09	7.29±4.92	6.89±5.07	7.01±4.96	6.77±5.28	6.81±5.03
Posterior Deltoid	6.13±3.77	**7.22±4.06**	**7.43±4.13**	**7.66±4.37**	**8.01±4.44**	**8.12±4.50**	**8.43±4.46**	**8.54±4.72**	**8.30±4.75**
Pec. Major Clavicular	2.68±2.98	2.95±3.00	2.92±2.99	2.85±2.92	3.03±2.77	2.78±2.94	3.29±3.05	2.99±3.11	3.22±3.45
Pec. Major Sternal	2.47±3.13	2.44±2.90	2.33±2.84	1.90±2.27	1.85±1.74	2.00±2.54	1.93±2.61	1.77±2.68	2.59±3.26
Right Lower Trap.	4.60±3.90	5.01±3.86	5.01±3.78	4.92±3.66	5.59±3.50	5.07±3.84	5.44±3.63	5.25±3.75	5.35±4.08
Right Upper Trap.	2.93±1.66	2.91±1.65	2.76±1.80	2.60±1.62	2.68±1.62	2.48±1.66	2.65±1.58	2.66±1.63	3.00±2.25
Serratus Anterior	1.49±0.77	1.40±1.12	1.42±1.08	1.32±1.11	2.13±3.17	1.37±1.11	1.85±2.00	1.25±1.00	1.48±1.27
Triceps	4.78±3.08	6.67±6.06	5.33±3.58	4.85±3.06	5.01±3.32	4.91±3.10	4.16±2.43	3.88±2.34	4.63±2.57

Supplementary Table 4.4. MAD and standard deviation normalized muscle activity for the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Ant. Deltoid	10.00±4.22	8.91±5.27	9.98±5.87	10.01±5.74	9.14±5.40	9.07±5.15	9.45±5.47	9.24±5.56	8.65±6.31
Biceps	0.44±0.54	0.32±0.19	0.35±0.27	0.35±0.24	0.41±0.26	0.39±0.32	0.38±0.28	0.37±0.28	0.36±0.28
Infraspinatus	8.93±5.71	7.80±5.99	8.45±5.91	7.45±4.10	7.31±4.86	7.21±4.53	6.84±4.10	6.56±4.27	6.65±3.99
Lat. Dorsi	2.71±2.14	2.65±2.01	2.63±1.96	2.47±1.75	**2.30±1.86**	**2.12±1.77**	**2.16±1.74**	**2.19±1.91**	**2.11±1.72**
Left Lower Trap.	7.22±4.50	6.88±4.10	6.06±3.64	5.69±3.30	5.71±3.27	5.25±3.38	6.07±3.68	5.98±3.09	5.77±2.93
Left Upper Trap.	1.59±1.29	1.48±1.45	1.70±1.62	1.66±1.56	1.68±1.76	1.95±1.74	1.72±1.55	1.69±1.66	1.99±2.06
Middle Deltoid	21.85±9.38	20.80±11.05	20.47±10.35	**18.89±8.49**	**18.42±8.83**	**18.04±7.65**	**18.20±8.87**	**17.92±7.20**	**16.82±7.38**
Posterior Deltoid	6.26±4.03	**7.36±4.23**	**7.42±4.44**	**7.51±4.54**	**7.71±4.70**	**7.98±4.80**	**8.09±4.89**	**8.27±5.00**	**8.58±5.10**
Pec. Major Clavicular	3.22±2.90	3.07±3.28	3.54±2.85	3.90±2.71	3.68±3.02	3.94±3.06	4.10±2.88	4.04±3.19	3.62±3.16
Pec. Major Sternal	1.39±1.30	1.30±1.32	1.18±1.16	1.09±0.90	1.05±0.62	1.07±0.74	1.20±1.16	1.18±1.31	1.13±1.11
Right Lower Trap.	4.06±3.59	4.20±3.41	4.25±3.47	4.31±3.40	4.26±3.31	4.41±3.39	4.48±3.40	4.66±3.76	5.36±5.04
Right Upper Trap.	3.51±1.50	3.09±1.77	3.29±1.42	3.17±1.14	2.79±1.38	3.07±1.28	2.99±1.21	2.97±1.35	3.22±1.50
Serratus Anterior	16.79±9.45	15.03±9.67	15.34±10.58	14.34±9.75	13.87±9.27	13.17±8.44	13.37±9.15	12.26±8.33	12.64±9.02
Triceps	10.86±7.97	**15.00±9.07**	**14.00±9.50**	**14.13±8.08**	**14.35±9.13**	**14.40±8.36**	**14.23±8.07**	**14.09±7.51**	13.25±8.66

Supplementary Table 4.5. MAD and standard deviation normalized muscle activity for the return phase of the push task in pre-fatigue and post-fatigue work cycles. Rows shaded in grey represent main effects. Individual cells that are bolded represent significant post-hoc differences between post-fatigue work cycles and the pre-fatigue value.

Muscle	Pre	PF1	PF2	PF3	PF4	PF5	PF6	PF7	PF8
Ant. Deltoid	9.35±3.37	10.12±4.54	10.58±4.24	10.03±4.94	9.60±5.17	10.09±4.96	9.95±5.40	10.41±4.66	10.36±6.67
Biceps	0.30±0.14	0.28±0.16	0.31±0.17	0.29±0.19	0.29±0.17	0.26±0.11	0.27±0.13	0.24±0.10	0.25±0.13
Infraspinatus	8.39±5.48	9.51±6.94	9.96±6.99	8.70±5.39	9.24±7.01	9.18±6.45	8.43±5.91	8.53±5.63	8.58±6.01
Lat. Dorsi	2.58±1.79	2.80±1.76	2.76±1.66	2.58±1.61	2.54±1.64	2.46±1.70	2.40±1.68	2.55±1.80	2.45±1.72
Left Lower Trap.	7.44±4.23	6.61±3.82	6.30±3.60	5.99±3.02	6.39±3.12	6.21±3.31	6.87±3.92	6.83±3.39	6.76±3.18
Left Upper Trap.	1.96±1.44	1.93±1.51	2.23±1.76	2.19±1.63	2.12±1.90	2.41±1.99	2.27±1.50	2.13±1.80	2.38±2.07
Middle Deltoid	13.62±5.53	**15.59±6.97**	**15.78±6.91**	14.23±6.32	14.04±6.14	14.53±5.94	14.44±6.07	14.08±5.91	13.63±5.83
Posterior Deltoid	6.41±4.07	7.26±4.18	7.45±4.45	7.67±4.69	**8.00±4.71**	**8.17±4.90**	**8.37±5.04**	**8.58±5.13**	**8.79±5.23**
Pec. Major Clavicular	6.03±3.16	5.43±3.38	5.69±2.33	5.59±2.40	5.41±2.69	5.57±2.53	5.28±2.54	6.12±3.01	5.63±2.87
Pec. Major Sternal	0.90±1.06	0.82±0.85	0.74±0.87	0.66±0.75	0.79±0.78	0.78±0.78	0.76±0.92	0.75±0.97	0.78±0.93
Right Lower Trap.	4.59±3.42	4.55±3.42	4.51±3.49	4.78±3.32	4.66±3.41	4.75±3.49	4.63±3.53	4.83±3.44	5.15±3.99
Right Upper Trap.	8.82±3.69	8.42±3.50	9.43±3.69	8.41±3.79	8.38±4.24	8.33±3.61	8.30±4.38	8.44±3.90	8.26±4.56
Serratus Anterior	12.65±6.18	14.70±7.47	14.99±7.74	12.71±6.29	13.27±7.70	13.78±7.24	13.12±8.04	13.02±6.82	13.21±7.08
Triceps	3.76±2.01	**4.96±2.27**	**5.15±2.58**	**5.15±2.36**	**5.15±2.38**	**5.13±2.40**	**5.21±2.33**	**5.14±2.37**	**4.83±2.41**

APPENDIX I: SUPPLIMETNARY TABLE FROM CHAPTER 4

Table AI.1: Individual subject differences in muscle and time specific fatigue. There was variability between participants in which muscles and at which time points that they exhibited significant signs of myoelectric fatigue (≥8% decrease in MPF between submaximal reference exertions). Reference exertion numbers are indicated with the letter R#, reference exertions that were missing data are indicated in the table with brackets (R#). The total number of reference exertions each participant performed is indicated next to participant code.

	Muscle										
	Adel	Mdel	Pdel	Infra	Utrap	Mtrap	Ltrap	Lats	Sert	PecC	PecS
P2 (R7)	R2-R7	R2, R5-R6		R2			R1, R3-R7	R3,R5-R6			R1-R2, R5, R7
P3 (R6)			R3	R4	R5		R3-R4	R1-R7			R3,R6
P4 (R11)	(R4)	R5,R8 (R4)	R1-R3,R5-R6,R9,R11 (R4)	R6,R8 (R4)	R6,R10 (R4)	(R4)	R2,R8 (R4)	R1-R2, R5-R11 (R4)	R8 (R4)	R3, R5-R6 (R4)	R3, R5-R6, R8-R10 (R4)
P5 (R12)	R5, R7, R12	R1, R7-R9, R11-R12		R6, R12		R6		R1-R12	R9, R11	R2-R12	R3-R6, R9-R10
P6 (R20)	R5, R8, R10-R15, R18, R20 (R19)	R15 (R19)	R5, R7-R8, R10, R14-R17, R20 (R19)	(R19)	(R19)	(R19)	R1 (R19)	R10R5, R7-R18, R20 (R19)	(R19)	(R19)	R2, R20 (R19)
P7 (R20)	R10-R11, R15, R17, R19-R20	R13-R15, R17-R20	R20		R6, R8, R10			R1-R2, R4-R20			R6, R9, R16
P8 (R12)	R11, R1, R10-R11							R1, R2-R5, R7-R12			
P9 (R5)	R1-R2, R5	R3-R5	R1	R2-R5	R3-R4		R5	R1-R5	R5		R1-R3, R5
P10 (R10)	R1-R11	R9	R11					R2, R4, R6-R11	R1, R3, R7, R11		R5
P12 (R8)				R1, R6	R1, R4, R8			R2-R3, R6-R8	R2	R6	R4-R5, R7-R8
P13 (R6)		R4-R5	R2	R4		R4		R1-R6		R3,R5	R1-R2,R4-R6
P14 (R6)		R4		R2				R1-R6			
P15 (R12)	R4,R7-R12	R4,R8-R12	R2-R3,R5-R12					R1,R3-R5,R7-R8,R11-R12	R7-R8		R7,R12
P16 (R5)			R1-R5					R2-R3,R5			
P18 (R7)	R2-R3			R2-R3,R5-R6			R1	R2,R6		R7	R2-R7
P19 (R15)	R11-R12,R15	R5,R7-R12	R2,R6,R9,R11,R13,R15	R5-R15	R7	R9		R7,R9-R11	R2-R15	R1-R8,R10-R13	R4-R7,R11-R15
P20 (R5)	R1-R5	R2-R5		R2-R3,R5			R2,R5	R2-R5	R2-R5		
P21 (R7)			R2,R4-R7	R1-R6	R4-R5,R7			R4	R4-R5,R7	R7	
P22 (R6)	R6		(R6)	R2,R5			R1			(R3)	R1,R4-R6 (R3)
P23 (R9)					R5	R7,R9	R1-R9	R2,R7	R1,R3,R6,R8-R9		

APPENDIX J: QUESTIONAIRES FROM CHAPTER 7

AJ.1: BRIEF SELF-CONTROL SCALE

SCS Scale

Please answer the following items as they apply to you. There are no right or wrong answers. Please choose a number (1 – 5) that best represents what you believe to be true about yourself for each question. Use the following scale to refer to how much each question is true about you.

1	2	3	4	5
Not at all like me		**Sometimes like me**		**Very Much like me**

1. I have a hard time breaking bad habits. _____
2. I am lazy. _____
3. I say inappropriate things. _____
4. I do certain things that are bad for me, if they are fun. _____
5. I refuse things that are bad for me. _____
6. I wish I had more self-discipline. _____
7. People would say that I have iron self-discipline. _____
8. Pleasure and fun sometimes keep me from getting work done. _____
9. I have trouble concentrating. _____
10. I am able to work effectively toward long-term goals. _____
11. Sometimes I can’t stop myself from doing something, even if I know it’s wrong. ____
12. I often act without thinking through all the alternatives. _____
13. I am good at resisting temptation. _____

AJ.2: STATE SCCS

For each of the following statements, please indicate how true it is for you, using the following scale:

1	2	3	4	5	6	7
Not true						**Very true**

1. I need something pleasant to make me feel better. ______
2. I feel drained. ______
3. If I were tempted by something right now, it would be very difficult to resist. ______
4. I would want to quit any difficult task I was given. ______
5. I feel calm and rational. ______
6. I can’t absorb any information. ______
7. I feel lazy. ______
8. I feel sharp and focused. ______
9. I want to give up. ______
10. I feel like my will power is gone. ______

AJ.3: FATIGUE STATE QUESTIONNAIRE (FSQ)

Instructions: Please answer the following questions honestly and accurately about how you are feeling *right now*, in this present moment.

1. How tired does your **body** feel right now?

1. Not at all
2. A little
3. Moderately
4. Very
5. Extremely

2. How tired does your **mind** feel right now?

1. Not at all
2. A little
3. Moderately
4. Very
5. Extremely

3. How awake do you feel right now?

1. Not at all
2. A little
3. Moderately
4. Very
5. Extremely

4. How slow and sluggish are you right now?

1. Not at all
2. A little
3. Moderately
4. Very
5. Extremely

AJ.4: RATING OF PERCEIVED FATIGUE SCALE

Please give a rating of perceived fatigue when prompted on the screen.

This rating should reflect your current perception of your force generating ability. This means we would like you to tell us how tired you feel in comparison to how you felt before we began the trial (ie complete rest).

It is important to focus on how you feel at each point in time when you are asked to give the rating. This means that your rating does not have to reflect the intensity of the effort but instead should provide an indication of how fatigued you are currently feeling.

Rating of Perceived Physical Fatigue

No fatigue at all	0
Very light fatigue	0.5
Light fatigue	1
Fairly fatigued	2
Moderately fatigued	3
Fatigued	4
Very fatigued	5
	6
Nearly exhausted	7
	8
	9
Absolutely exhausted	10

AJ.5: FEELING SCALE

Feeling Scale (FS)

(Hardy & Rejeski, 1989)

While participating in exercise, it is common to experience changes in mood. Some individuals find exercise pleasurable, whereas others find it to be unpleasant. Additionally, feeling may fluctuate across time. That is, one might feel good and bad a number of times during exercise. Scientists have developed this scale to measure such responses

+5 Very good

+4

+3 Good

+2

+1 Fairly good

0 Neutral

-1 Fairly bad

-2

-3 Bad

-4

-5 Very bad

AJ.6: FELT AROUSAL SCALE

FELT AROUSAL SCALE (FAS)

(Svebak & Murgatroyd, 1985)

Estimate here how aroused you actually feel. Do this by circling the appropriate number. By "arousal" we meant how "worked-up" you feel. You might experience high arousal in one of a variety of ways, for example as excitement or anxiety or anger. Low arousal might also be experienced by you in one of a number of different ways, for example as relaxation or boredom or calmness.

1 LOW AROUSAL

2

3

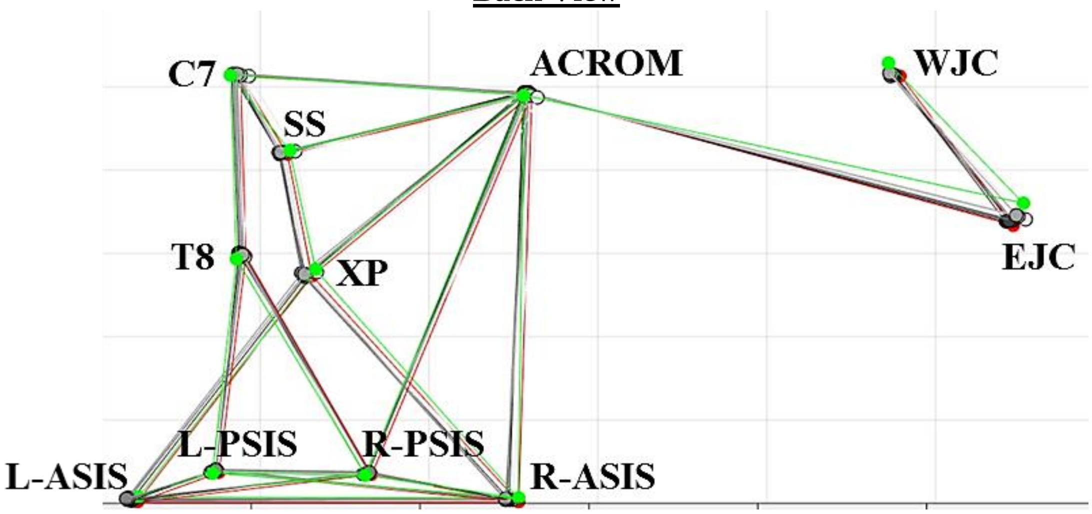
4

5

6 HIGH AROUSAL

APPENDIX K: VARIABILITY FIGURES FROM CHAPTER 8

Back View



Right Side View

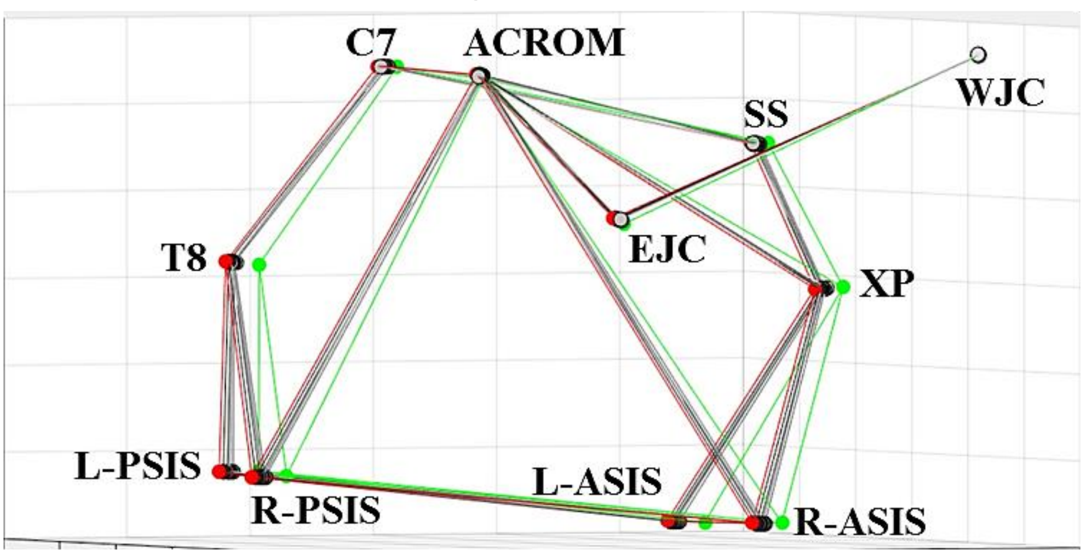


Figure AK.1: Torso and right upper extremity during drill task in Chapter 6. Eleven markers are plotted and labeled from a back view and a right side view: right acromion (ACROM), right elbow joint center (EJC), right wrist joint center (WJC), xiphoid process (XP), sternum (SS), C7, T8, Left and right posterior superior iliac spine (L-,R-PSIS), left and right anterior superior iliac spine (L-,R-ASIS) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

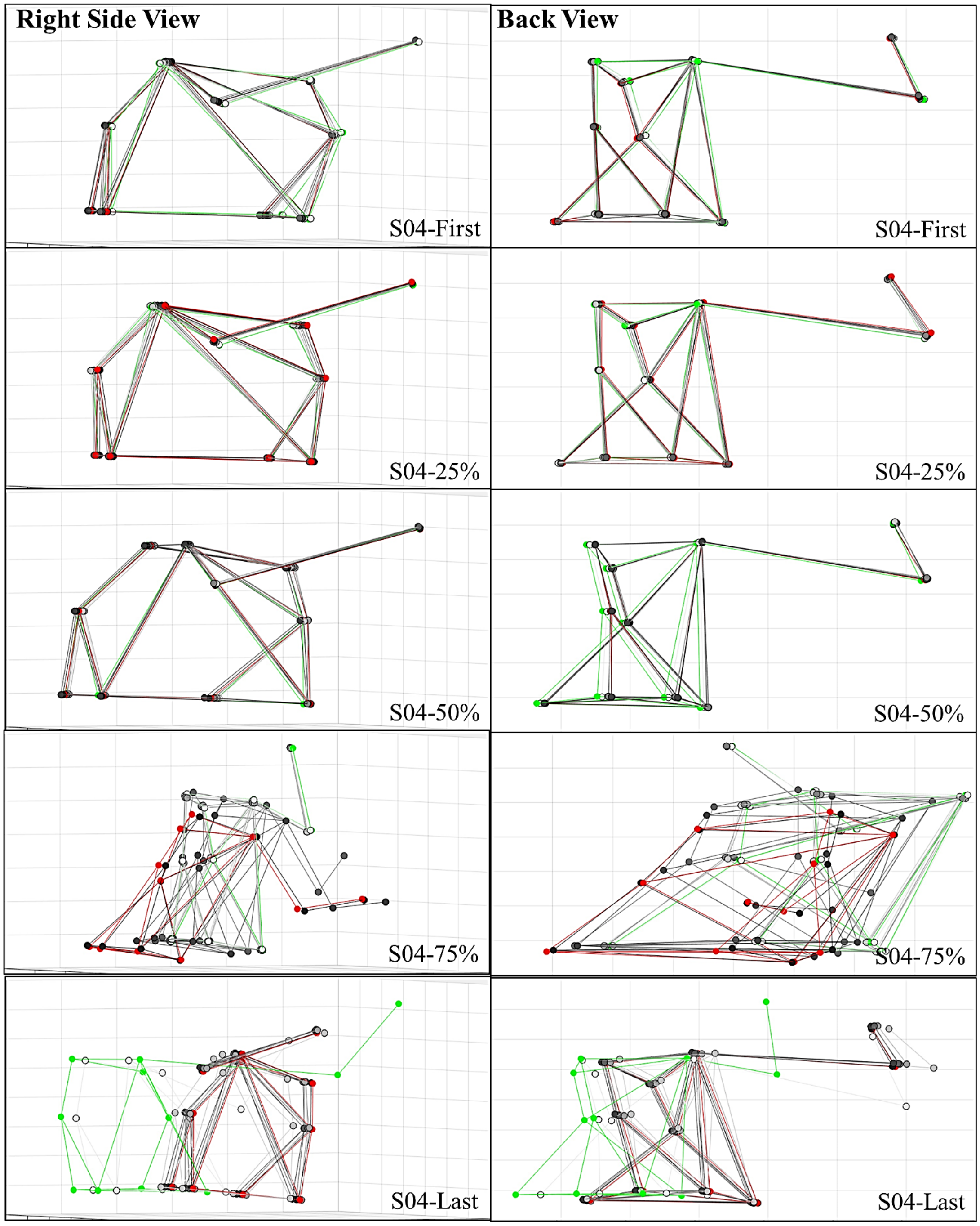


Figure AK.2: Torso and right upper extremity for Participant 4 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

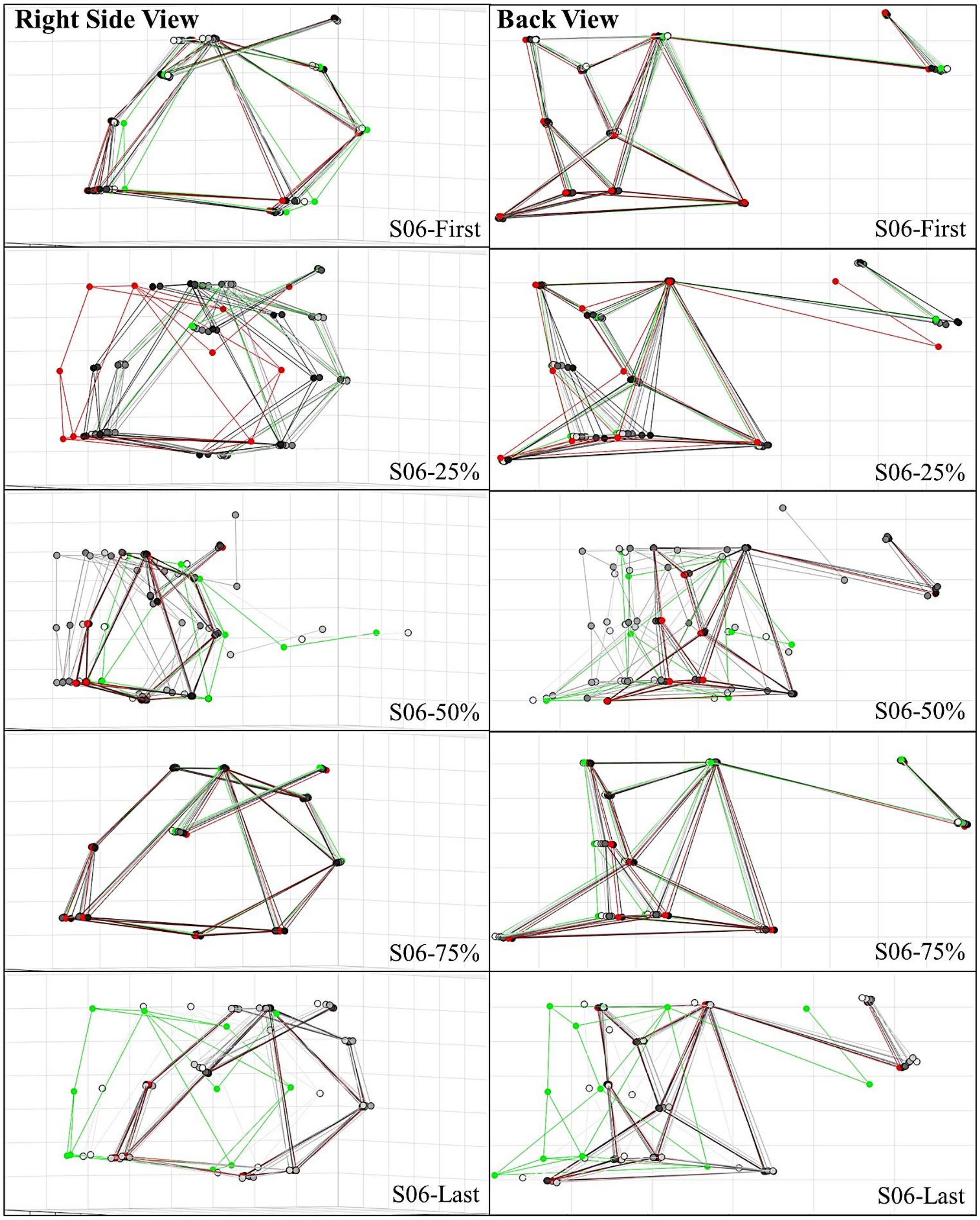


Figure AK.3: Torso and right upper extremity for Participant 6 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

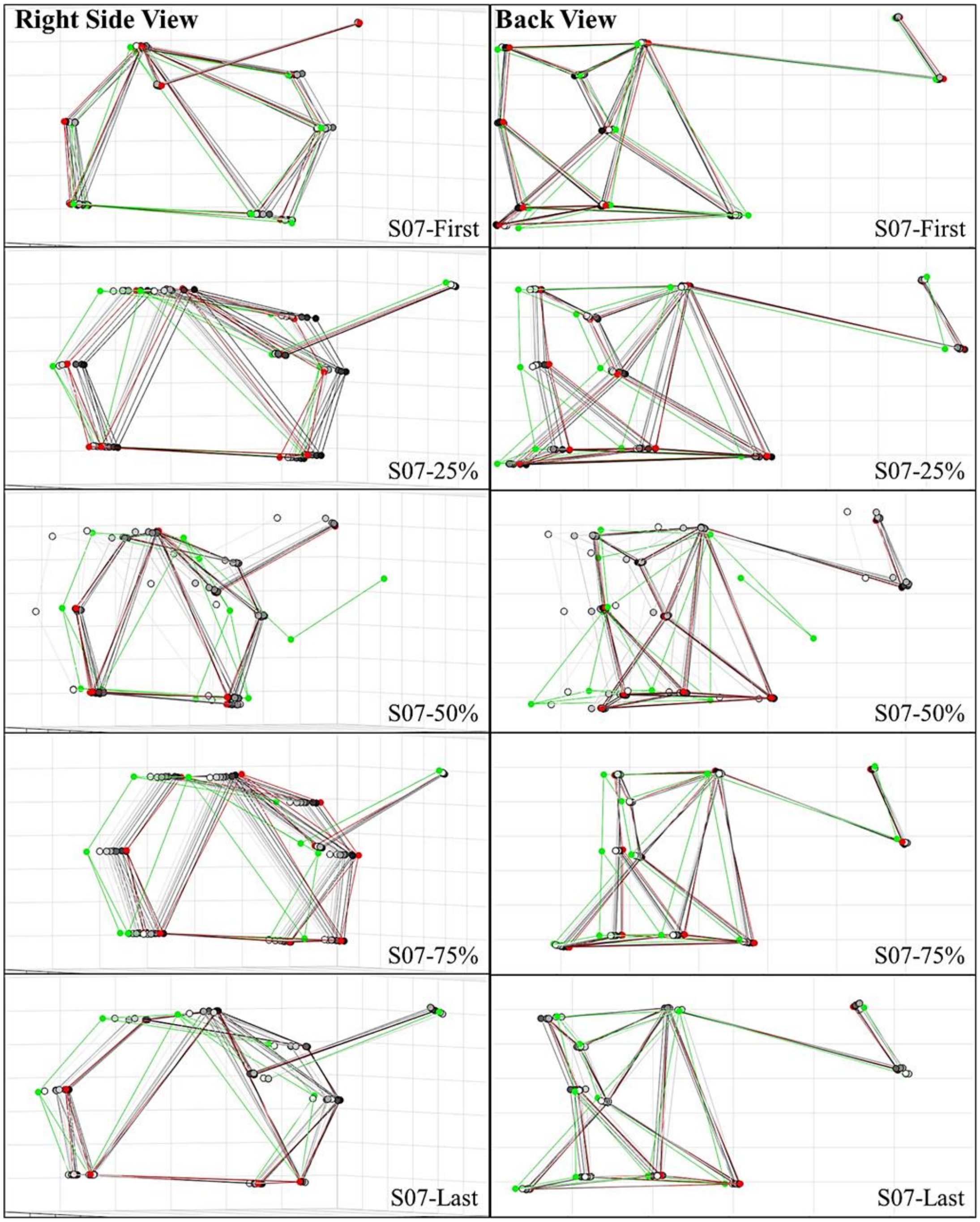


Figure AK.4: Torso and right upper extremity for Participant 7 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

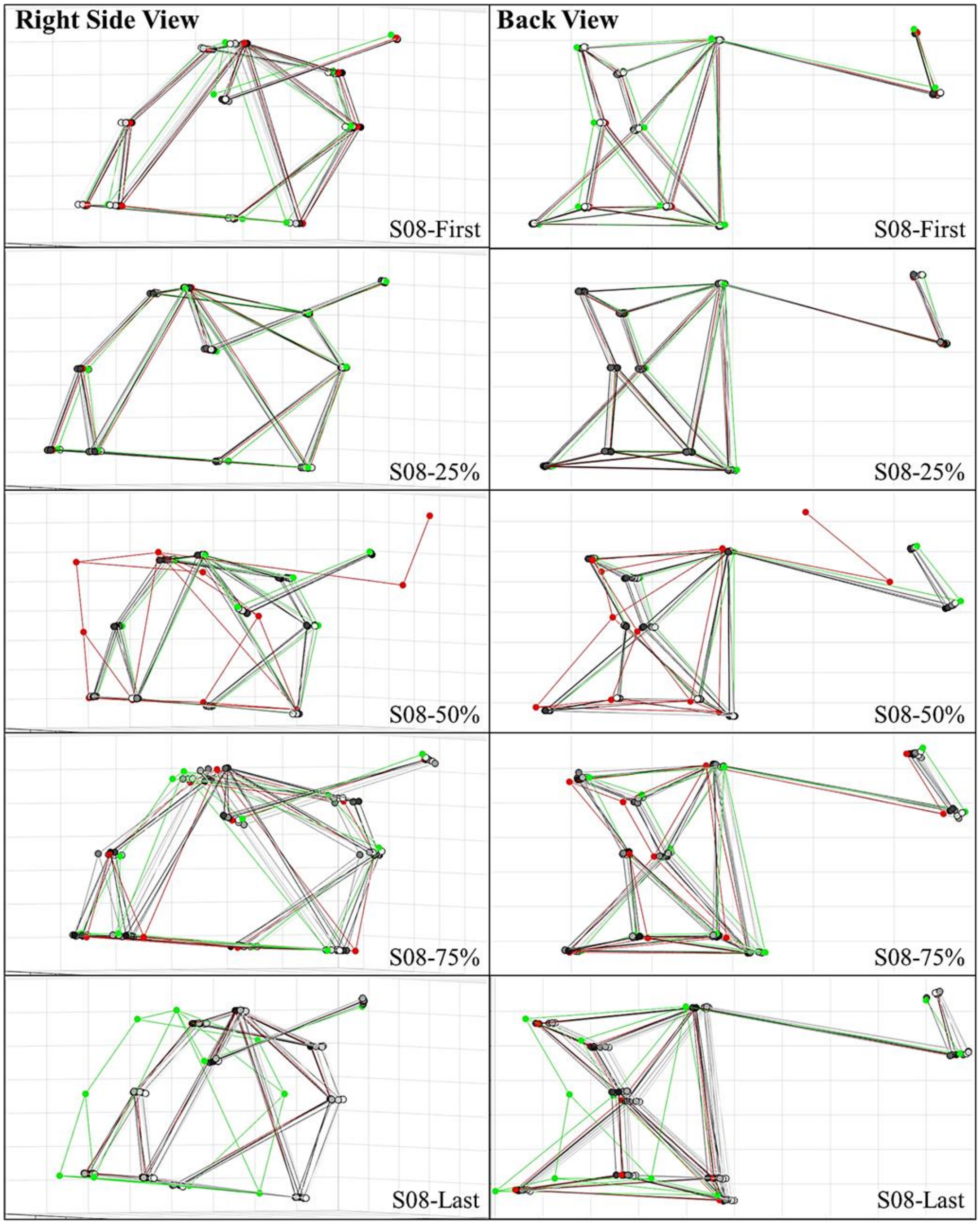


Figure AK.5: Torso and right upper extremity for Participant 8 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red

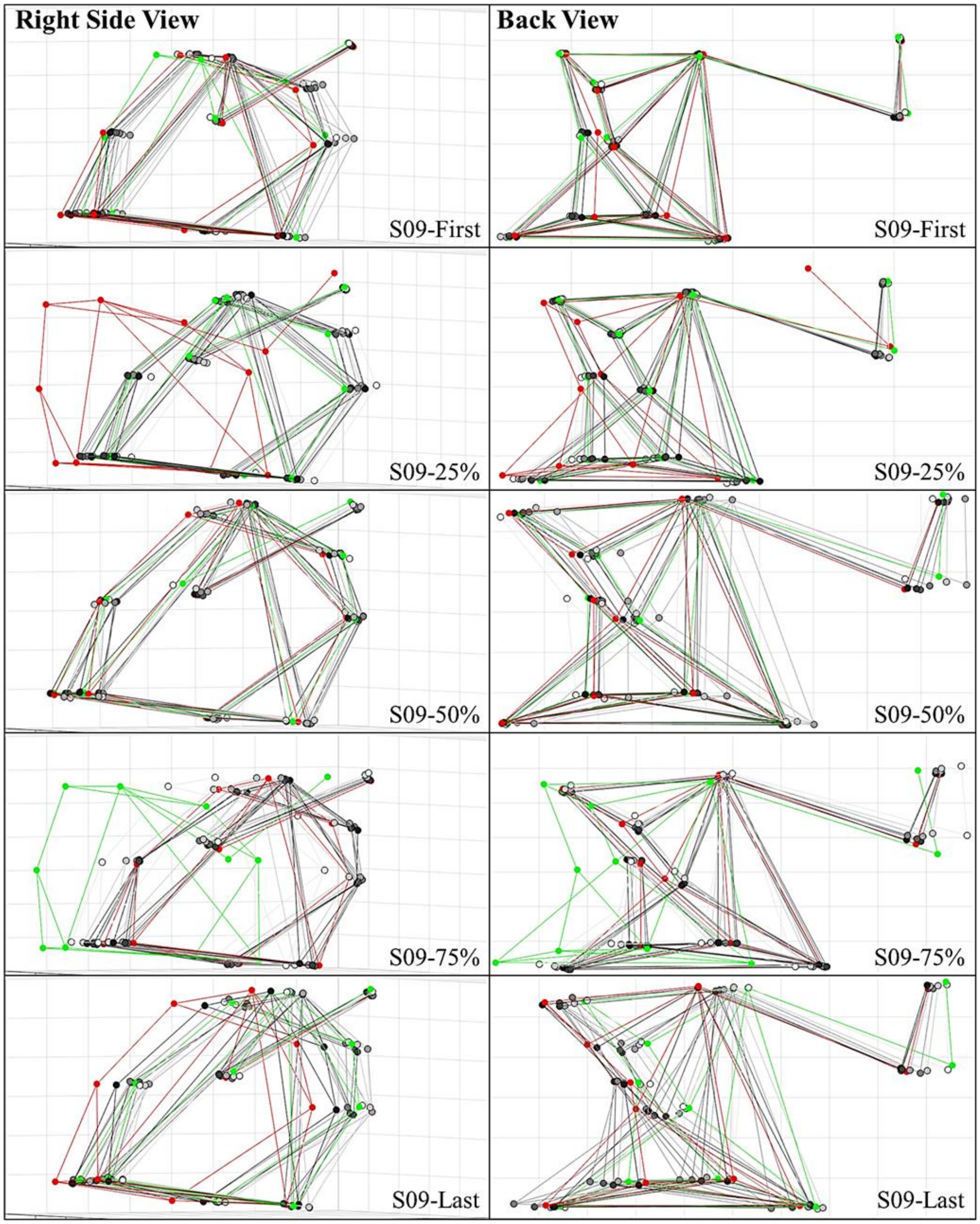


Figure AK.6: Torso and right upper extremity for Participant 9 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

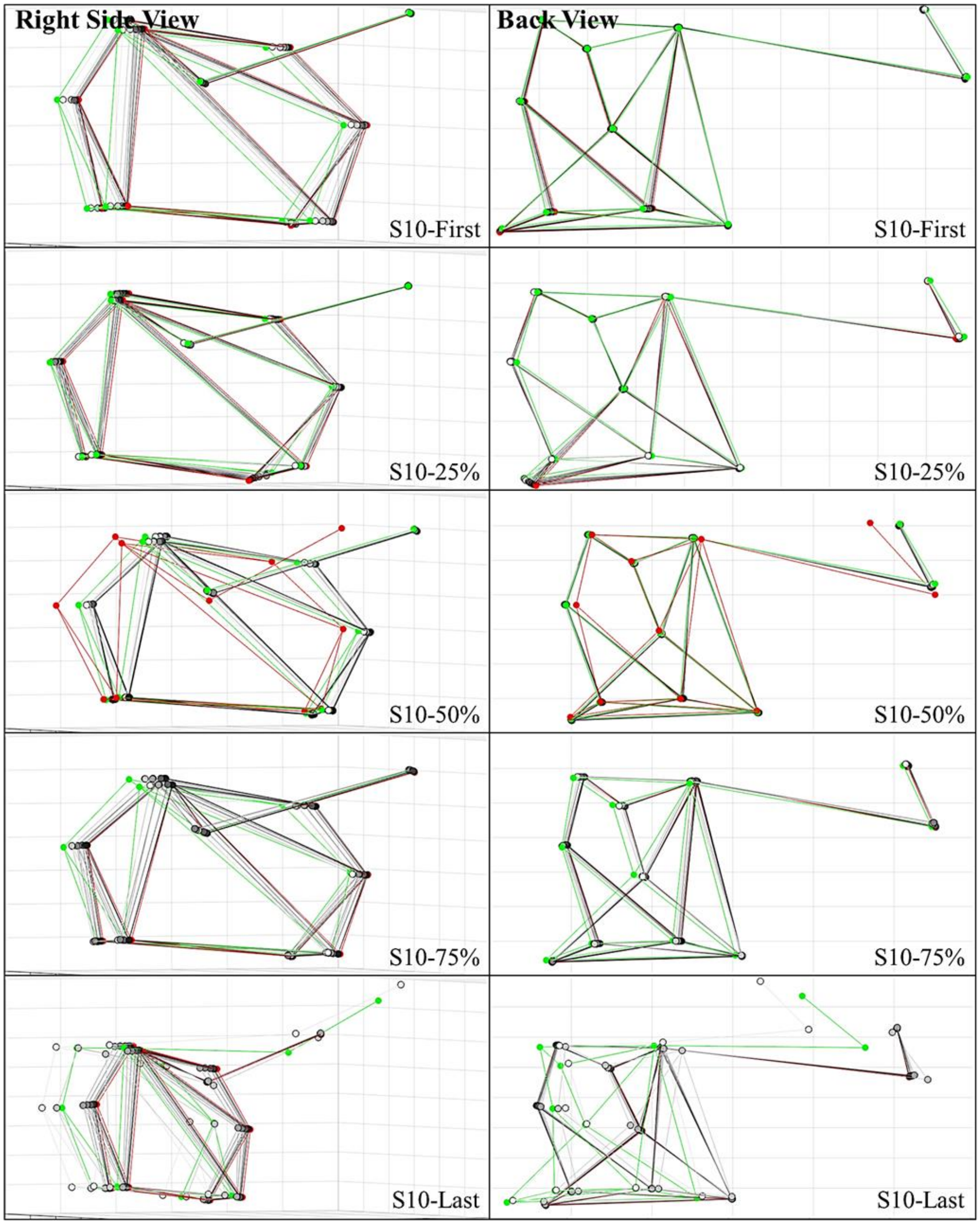


Figure AK.7: Torso and right upper extremity for Participant 10 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

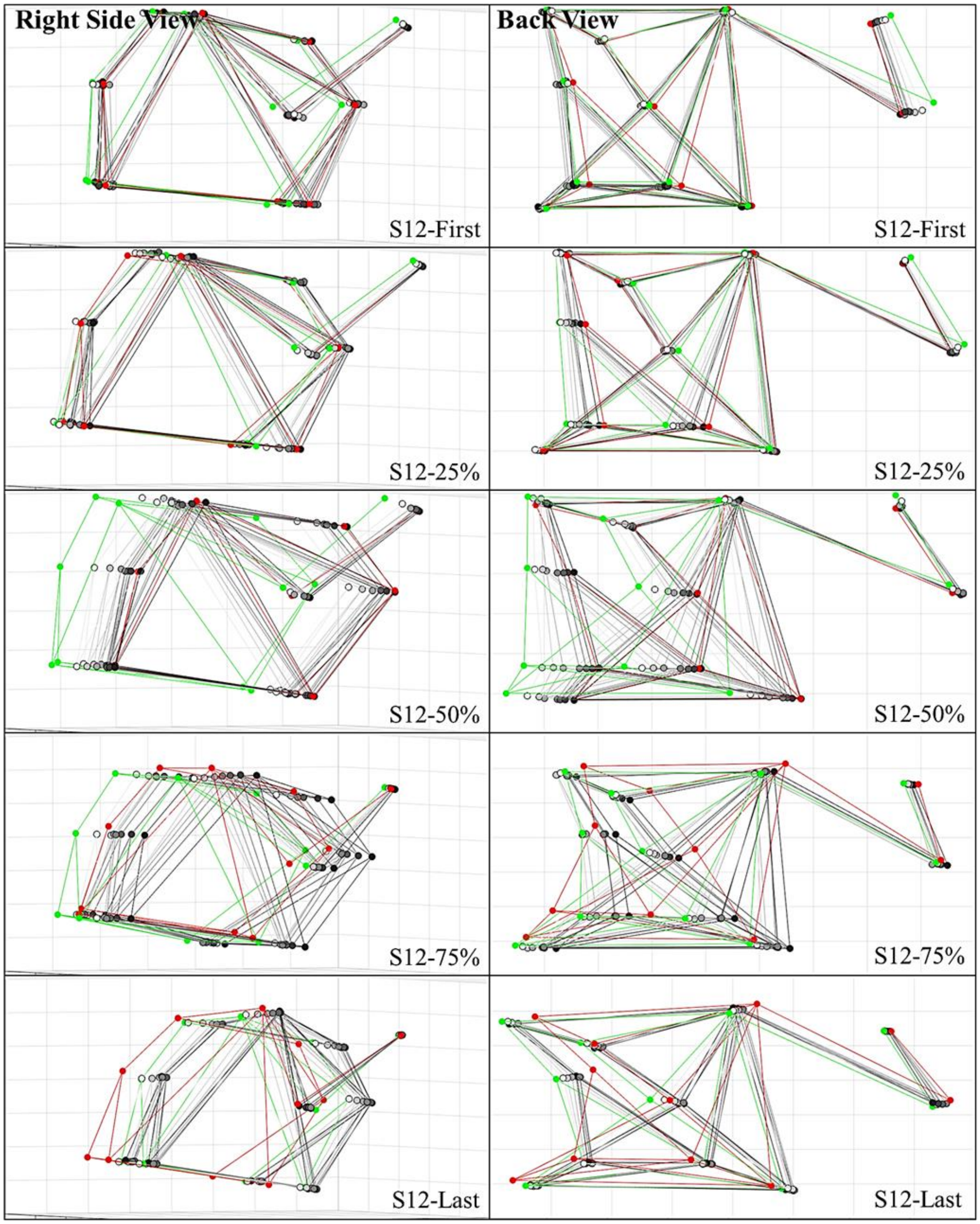


Figure AK.8: Torso and right upper extremity for Participant 12 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

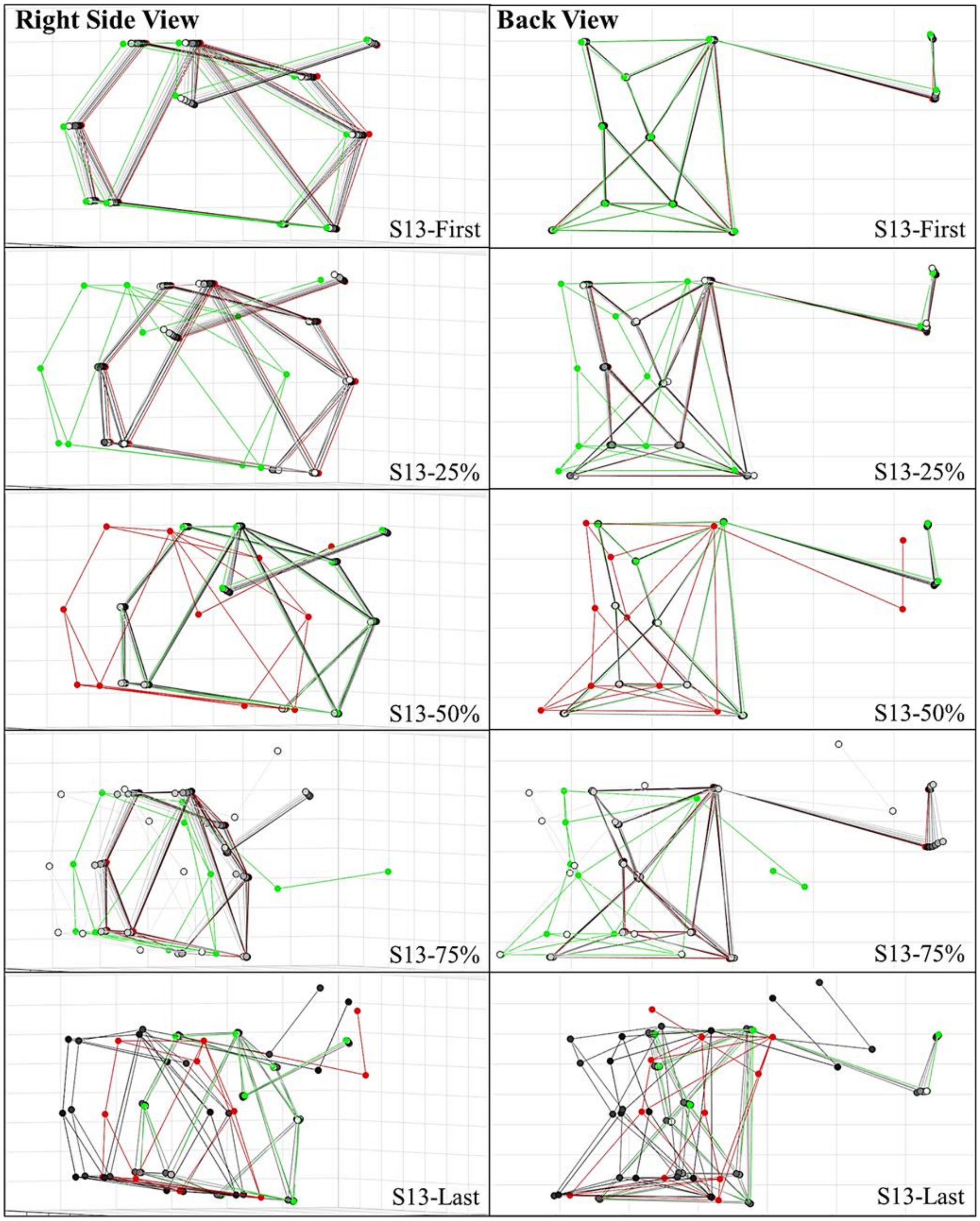


Figure AK.9: Torso and right upper extremity for Participant 13 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

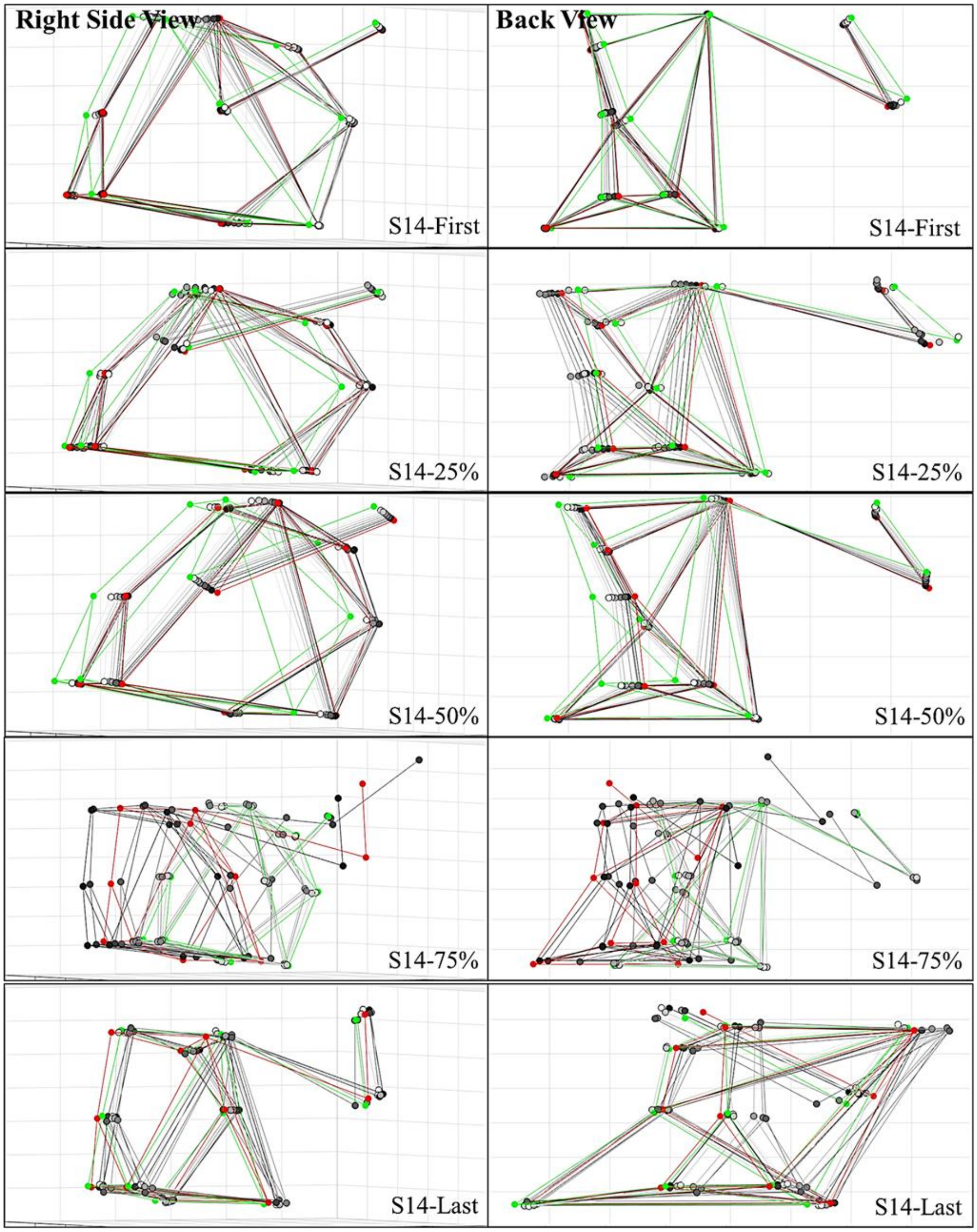


Figure AK.10: Torso and right upper extremity for Participant 14 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

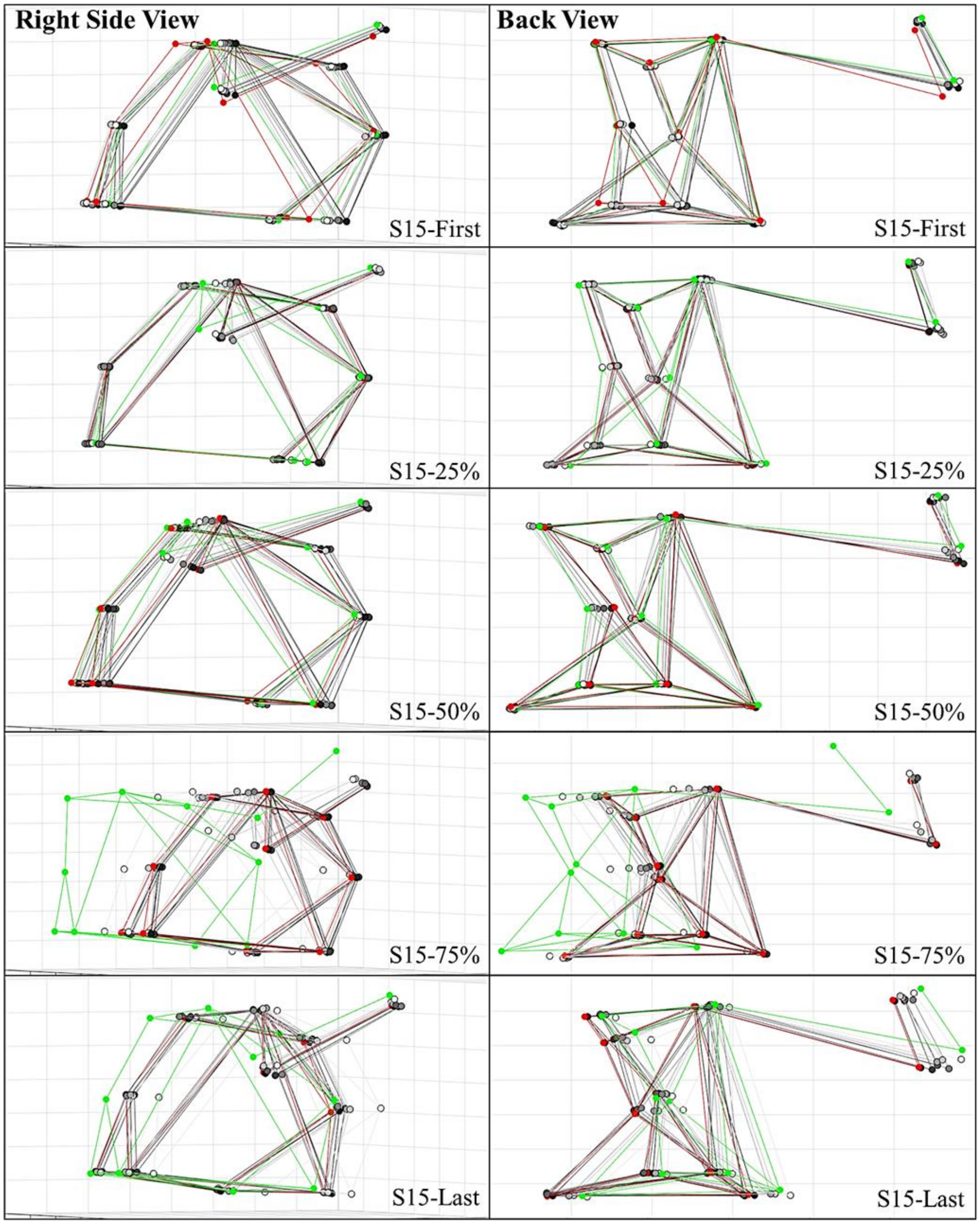


Figure AK.11: Torso and right upper extremity for Participant 15 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

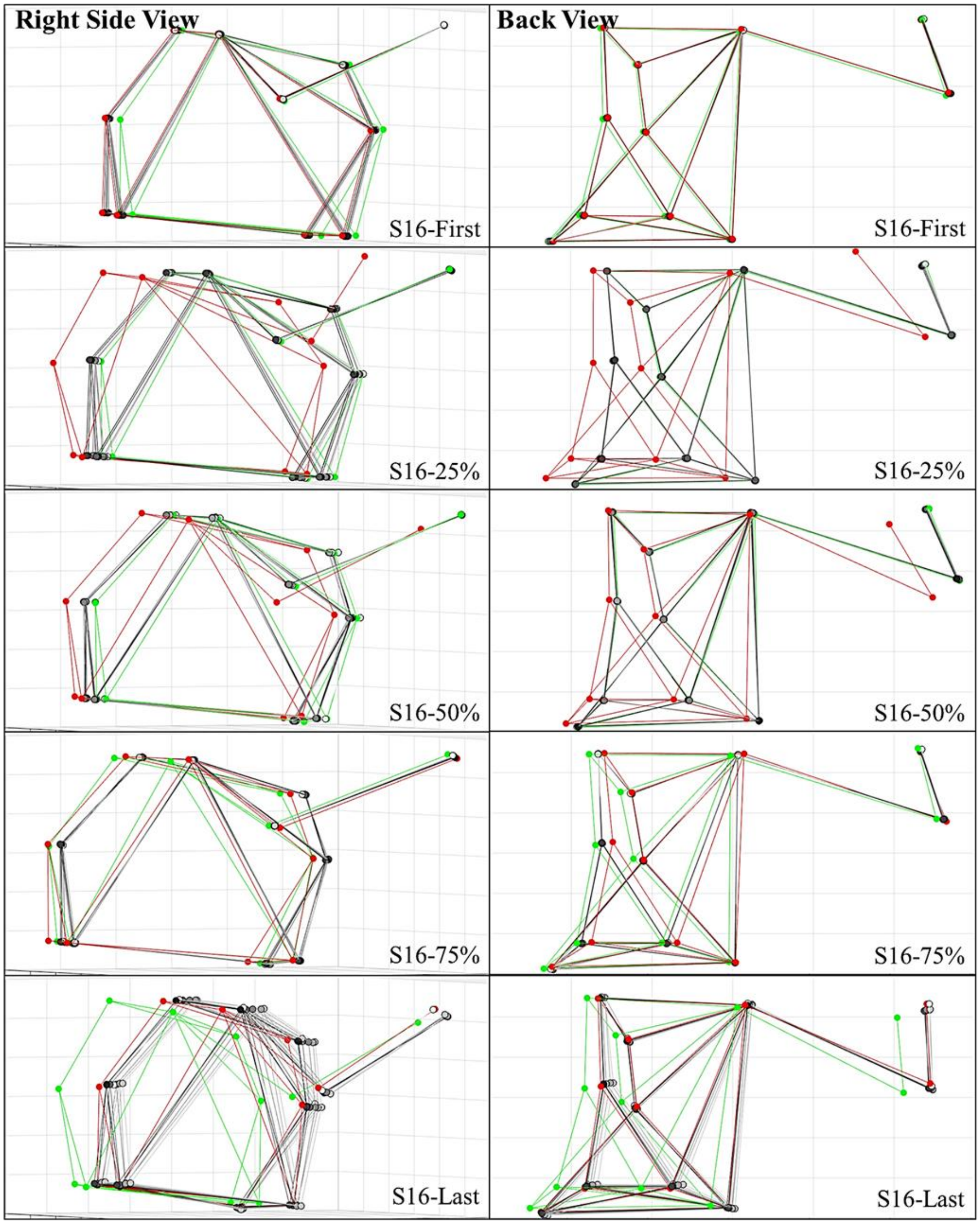


Figure AK.12: Torso and right upper extremity for Participant 16 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

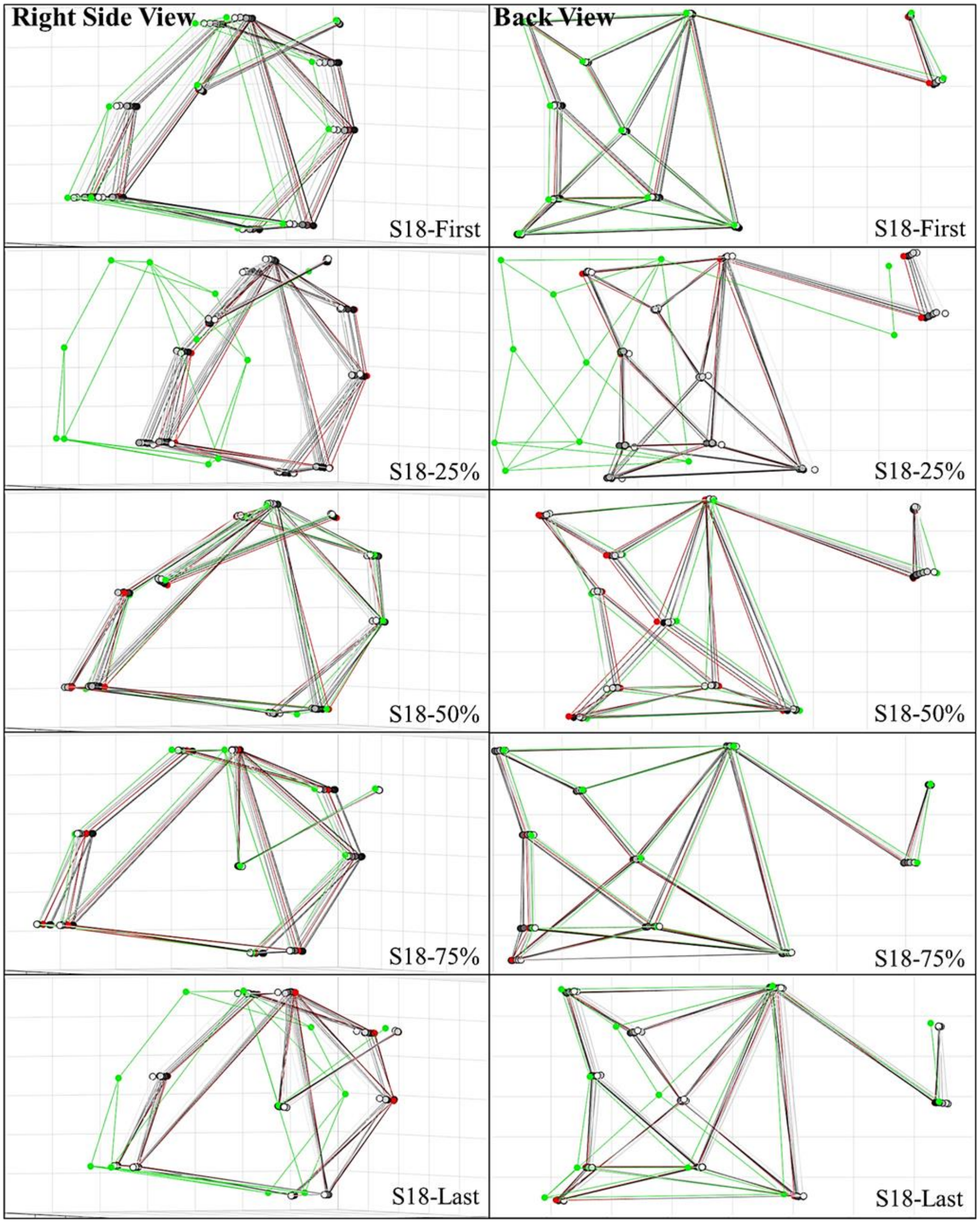


Figure AK.13: Torso and right upper extremity for Participant 18 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

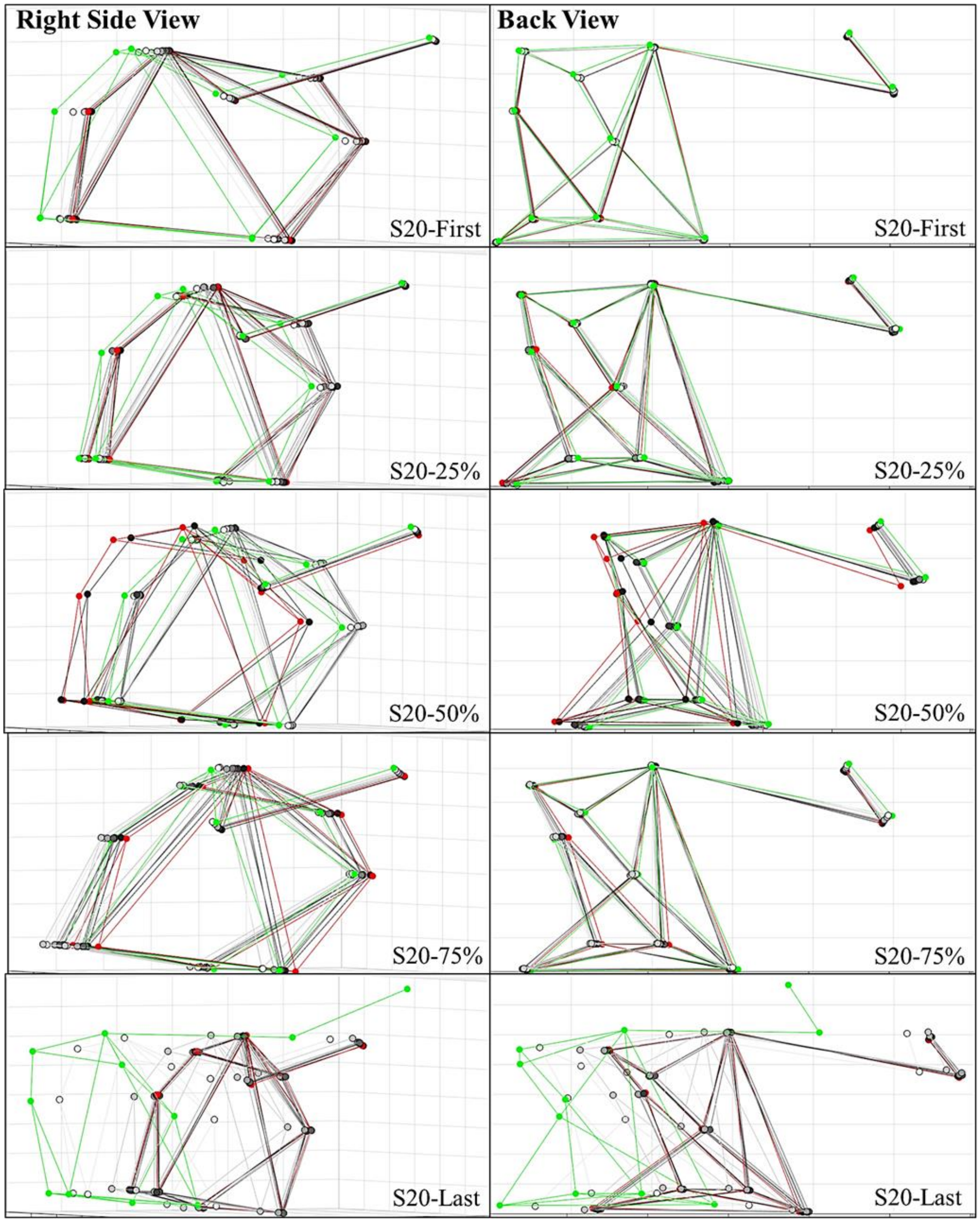


Figure AK.14: Torso and right upper extremity for Participant 20 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50^{th} frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

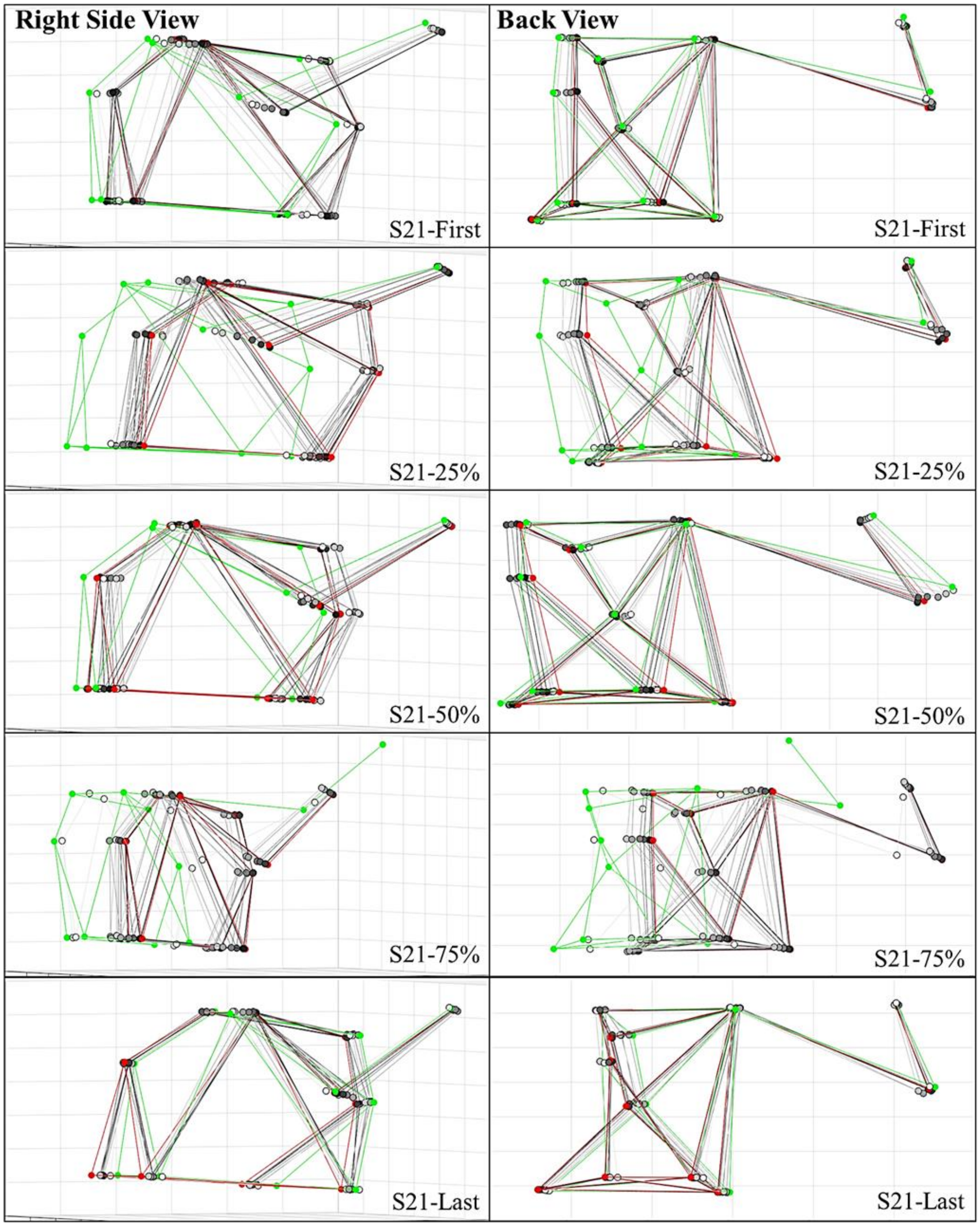


Figure AK.15: Torso and right upper extremity for Participant 21 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.

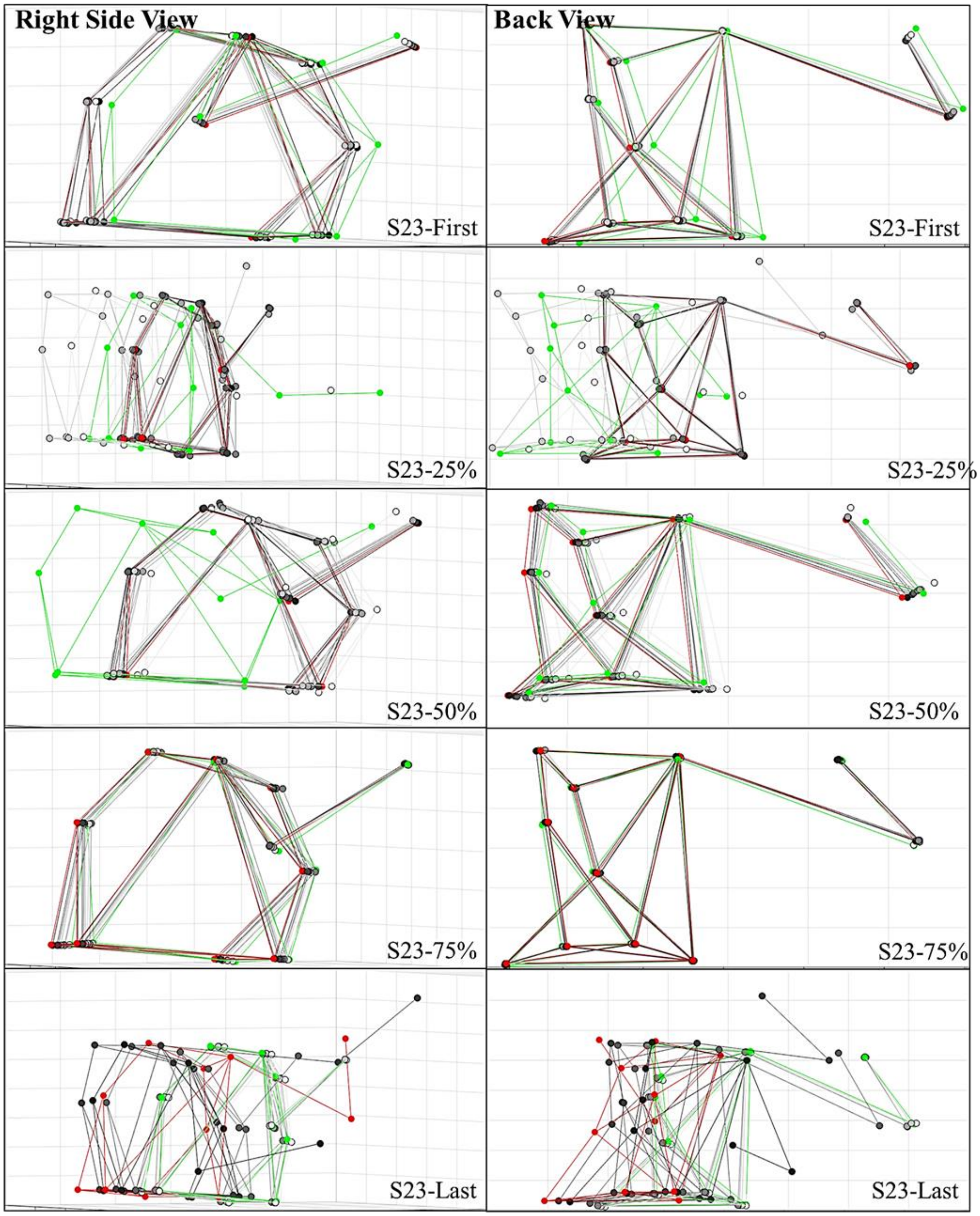


Figure AK.16: Torso and right upper extremity for Participant 23 during drill task in Chapter 6. Eleven markers are plotted (right acromion, right elbow joint center, right wrist joint center, xiphoid process, sternum, C7, T8, Left and right posterior superior iliac spine, left and right anterior superior iliac spine) for the first, 25%, 50%, 75% and last work cycles. For each work cycle, the first frame is plotted in green, every 50th frame is plotted in progressively darker shades of grey (one each second) and the last frame is plotted in red.