The Effect of Neck Muscle Fatigue on Shoulder Humeral Rotation Proprioception

By

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A thesis submitted to the School of Graduate and Postdoctoral Studies in partial fulfillment of the requirements for the degree of

Master of Health Science in Kinesiology

Department of Kinesiology/Faculty Of Health Science

Ontario Tech University (UOIT)

Oshawa, Ontario, Canada

November, 2019

THESIS EXAMINATION INFORMATION

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Master of Health Sciences in **Kinesiology**

Thesis title: **The Effect of Neck Muscle Fatigue on Shoulder Humeral Rotation Proprioception**

An oral defense of this thesis took place on November 8th, 2019 in front of the following examining committee:

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The above committee determined that the thesis is acceptable in form and content and that a satisfactory knowledge of the field covered by the thesis was demonstrated by the candidate during an oral examination. A signed copy of the Certificate of Approval is available from the School of Graduate and Postdoctoral Studies.

ABSTRACT

Fatigue of the cervical extensor muscles (CEM) is thought to disrupt afferent feedback from postural mechanoreceptors as it has been observed to cause decrements in upper limb proprioceptive accuracy. There has been limited research to quantify this interaction at the shoulder; thus further investigation into the effect of neck fatigue on shoulder joint position sense (JPS) accuracy is necessary. In this thesis, study one determined that high variability in upper limb DOFs led to no effect of neck fatigue on humeral rotation proprioception. Study two identified wrist deviation as a significant contributor to end effector variability when performing unconstrained humeral rotation. Wrist deviation will require a sufficient method of constraint in order to make significant observations of end effector position during humeral rotation possible.

Keywords: Proprioception; Shoulder Joint Position Sense (JPS); Cervical Extensor Muscles (CEM); Upper Limb Degrees of Freedom (DOF); Humeral Rotation.

CO-AUTHORSHIP STATEMENT

The manuscripts described in Chapters 3 and 4 of this thesis were authored with co-supervisors Dr. Nicholas J. La Delfa and Dr. Bernadette Murphy. I, **Matthew S. Russell**, was the primary contributor to all aspects of these works, including study conception & design, data collection, data analysis and manuscript preparation.

Dr. La Delfa lent direction and expertise towards study conception and design pertinent to the field of biomechanics. **Dr. Murphy** provided direction and expertise towards study conception and design pertinent to the field of neuroscience. Both supervisors also provided extensive edits to both manuscripts, but all written content presented in this thesis is the work of my own.

AUTHOR’S DECLARATION

I hereby declare that this thesis consists of original work of which I have authored. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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The research work in this thesis that was performed in compliance with the regulations of Ontario Tech’s Research Ethics Board/Animal Care Committee under **REB Certificate number #15033 & #15034**.

Matthew Stephen Russell

STATEMENT OF CONTRIBUTIONS

The work described in Chapter 3 was performed within the Rehabilitative Neuroscience Laboratory at the University of Ontario, Institute of Technology, directed by Dr. Bernadette Murphy. The work described in Chapter 4 was performed within the Occupational Neuromechanics and Ergonomics Laboratory at the University of Ontario, Institute of Technology, directed by Dr. Nicholas La Delfa. I conducted all data collection, analysis, and primary interpretation of the results at the advisement of my co-supervisors. Dr. La Delfa and Dr. Murphy provided feedback and made minor editorial adjustments to the manuscripts contained within.

I hereby certify that I am the sole author of this thesis. I have used standard referencing practices to acknowledge ideas, research techniques, or other materials that belong to others. Furthermore, I hereby certify that I am the sole source of the creative works and/or inventive knowledge described in this thesis.

DEDICATION

This thesis is dedicated to my brother, Nathan Russell.

It has always been my pride to watch you surpass me at every endeavor.
I cannot wait to see what you accomplish.

ACKNOWLEDGEMENTS

While this thesis is meant to consolidate the knowledge I have gained throughout my masters, it fails to truly capture all that I have been so fortunate to experience. I would therefore like to first acknowledge the contributions of the following individuals as best as I can within the schools stipulated 2-page limit.

Foremost I want to thank my co-supervisor Dr. Nicholas La Delfa. From day one you surely took on the toughest job at OntarioTech; collectively mentoring me and Dan. I have to say as much as the Fourier analysis assignment was a real page turner, it was undoubtedly the Christmas parties, conference road trips, and Croatian rock star moonlighting that made these two years so memorable. Nick in your short time with the faculty you've already managed to raise an incredibly lovely family while concurrently fostering a true 'lab family' environment with your friendly and inviting personality. Perhaps your most significant impact has been not to my thesis but to being a true role model for a young academic like myself.

To my other co-supervisor Dr. Bernadette Murphy: none of my incredibly valuable experiences these past years would have happened without your belief in me from the very beginning. At the time I began my masters we had regrettably been unacquainted with each other, yet you took a chance and made a premature evaluation of my character. I realized then that words could not properly convey how much your support from the very beginning truly meant, so in response I have done my best to live up to your expectations and create a body of work you would be proud of. I hope I proved you right.

To Dr. Paul Yielder I owe my thanks for first instilling my fascination in neuroscience. Your influence during my fourth year of undergrad helped put me on the path of an academic. Some of the most acute moments of learning during my graduate degree consisted of our conversations at conferences and backyard barbeques alike.

To Fahima Wakeely and Halima Hossinzehi (listed alphabetically, not in order of importance): over two years I have had the distinct pleasure to watch you both progress from my students, to my colleagues, to now two of my dearest friends. I am genuinely delighted to know that you will both continue to uphold the lab community which has always meant so much to me. Take care of mom and dad.

To Andrew Cregg and Ryan Foley: without both of your collective help and patience, I would surely still be trying to figure out how First Principles works. Though neither of you signed up for it, you both did much to mentor both Dan and myself and I thank you both for turning every problem into an opportunity to teach us.

To my senior colleagues (in experience, not in citizen status): Heather McCracken, Antonia Karellas, Mahboobeh Zabihosseinian, and Mrs. Annie Goncin. Thank you for all being incredible role models and always taking the time to step across the hall and offer your experiential advice. I can attest that it goes a long way in making new grad students feel reassured.

To the members of my graduate student cohort: Brianna, Tracey, Andrea, Nik, Matt, Jason, and so many others: It was the best of times, it was the worst of times, and it was wholly too short. But we certainly made the most of it and I thank you for the many great memories.

To the members of the junior graduate cohort: Adam Murphy, Rufeyda Cosgun, and Praveen Sanmugananthan: believe me when I say that the worst has certainly passed. I know you are all going to have an incredible experience this coming year and I wish you the very best.

To my loving parents Steve and Liz, and to my brother Nate: thank you for seeing me this far. The last two years have meant so much to me, though I have always lamented what time was lost that could have been spent together. You have all played such a large part in helping me get by the last two years as you have always helped me. Enduring a commitment like this has helped me realize what an important part of my life you all are and I now intend on making up for the lost time together promptly.

Andrew Varley: Once there was a time that I believed I would never achieve in my academics, until you brought out my potential. I very much accredit the man and academic I now am today to all the things you have taught me. Thank you.

Lastly, to Daniel Abdel-Malek. Two years ago I was excited to join the lab together as friends, though we have now certainly parted as brothers. I can't wait to see what the next chapter has in store for you, as I know that you're going to help so many people as a doctor. As long as they've got barbells and Aviation in Nova Scotia, I know you're going to thrive.

I pursued a master's degree to learn about my field of study and the world of research, but I never expected I would learn so much about myself in the process. Everyone in this section of acknowledgments has played an important part not only throughout my masters, but also in my life. I thank you all for always bringing out the best in me.

Oh, and follow @thekinbasement.

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LIST OF ABBREVIATIONS AND SYMBOLS

AMEDA	Active Movement Extent Discrimination Apparatus
ATP	Adenosine Triphosphate
CEM	Cervical Extensor Muscles
CNS	Central Nervous System
CPGS	Chronic Pain Grade Scale
DOF	Degree(s) of Freedom
DOMS	Delayed Onset Muscle Soreness
DV	Dependent Variable
EDH	Edinburgh Handedness Inventory
EMG	Electromyography
GTO	Golgi Tendon Organ
ISB	International Society of Biomechanics
JPE	Joint Position Error
JPS	Joint Position Sense
JPR	Joint Position Reproduction
LMN	Lower Motor Neuron
MVC	Maximum Voluntary Contraction
NDI	Neck Disability Index
PV	Predictor Variable
TTDPM	Threshold to Detect Passive Movement
UMN	Upper Motor Neuron
AE	Absolute Error
CE	Constant Error
VE	Variable Error

ElbFlex	Elbow Flexion
ElbPro	Elbow Pronation
HumNegElv	Humeral Negative Elevation
HumPOE	Humeral Plane of Elevation
HumRot	Humeral Rotation
WrstFlex	Wrist Flexion
WrstPro	Wrist Pronation
α	Alpha
β	Beta
γ	Gamma
γ_2	Second-Order Gamma
α-MN	Alpha Motor Neuron
γ-MN	Gamma Motor Neuron
β-Weight	Beta Weight

CHAPTER 1.

THESIS INTRODUCTION

The muscles of the cervical spine play a crucial role in the perception of the head's orientation in 3D space, and also in the construction of an accurate internal map of the body stored in the brain known as the body schema. Like many neuropsychological frameworks, the definition of body schema has received many updates over the years as more research continuously unveils a greater understanding of our subconscious cortical function. Neck extensor muscles have a uniquely dense array of intramuscular mechanoreceptors which are used by the central nervous system (CNS) to reference accurate head and neck posture (Jull, Falla et al. 2007). This perceived sense of head and neck posture allows the CNS to update body schema in reference to the limb segments distal from the neck (Knox and Hodges 2005). Body schema is the cortical 'map' of our body's position in space relative to the features of our surrounding environment, which is constructed through a network of interacting cortical and subcortical structures (Holmes and Spence 2004, Proske 2015). Based on body schema, our CNS maps our limbs positions in their environment, which it can reference to facilitate accurate motor planning (Contessa, De Luca et al. 2016). Proprioceptive acuity also arises from body schema, which references body posture to predict limb orientation (Proske and Gandevia 2009). Local limb orientation is signaled by the coordination of various proprioceptive neurons which are specialized to reference articular changes within the somatosensory tissue they are embedded within. In skeletal muscle fibers, muscle spindles relay perceived muscle length, while Golgi tendon organs (GTO's) embedded in muscle tendon relay information on muscle tension (Jerosch and Prymka 1996, Ribeiro and Oliveira 2011, Proske 2015). Non-contractile tissues also contribute to proprioceptive afference, especially during passive joint movement where muscle spindle activity regulated by gamma motoneuron feedback is reduced and GTOs are mostly inactive (Collins, Refshauge et al. 2005). This means that passive movement sensation mainly subsists on cutaneous and joint capsule Ruffini corpuscles which sense compression, stretch, and torsion of their respective tissues (Forget and Lamarre 1987, Collins and Prochazka 1996, Collins, Refshauge et al. 2005).

Recent studies have confirmed that perceived head and neck posture can disrupt limb proprioception (Knox and Hodges 2005, Thigpen, Padua et al. 2010). Additionally, afferently disruptive stimuli at the neck such as pain (Barker 2011), tendon vibration (Knox, Cordo et al. 2006), and fatigue (Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2017, Zabihhosseinian, Yielder et al. 2019) have been linked to upper limb proprioceptive decrements similar to changes observed with altered head posture. Decrements in sensorimotor function due to altered sensory feedback to the neck include disruptions in balance and posture (Gosselin, Rassoulian et al. 2004), changes in gait mechanics (Schieppati, Nardone et al. 2003), deficits in motor learning and retention (Baarbé, Yielder et al. 2015), impedances to spatial orientation (Schmid and Schieppati 2005), altered limb mechanics (Falla, Bilenkij et al. 2004, Baarbé, Murphy et al. 2015), and most prominently decrements in limb proprioceptive accuracy (Haavik and Murphy 2011, Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2017, Reece 2019, Zabihhosseinian, Yielder et al. 2019).

Neck pain is one example of altered sensory input from the neck that is highly prevalent, impacting 40-60% of all people in their lifetime (Hogg-Johnson, Van Der Velde et al. 2008), with incidence expected to rise (Ming, Närhi et al. 2004). Individuals with recurrent pain have been found to display decreased upper limb proprioception (Falla, Bilenkij et al. 2004, Barker 2011, Haavik and Murphy 2011). Neuromuscular fatigue is a common precedent to chronic pain and it has also been implicated in the onset of various performance decrements (Merton 1954, Bigland-Ritchie, Dawson et al. 1986, Skinner, Wyatt et al. 1986, Allen, Lannergren et al. 1995, Carpenter, Blasier et al. 1998, Ellenbecker and Roetert 1999, De Ruiter, Elzinga et al. 2005, Allen, Lamb et al. 2008, Amann 2011, Emery and Cote 2012, Contessa, De Luca et al. 2016). Locally induced muscle fatigue has been observed to lead to significant decreases in maximum force output (Merton 1954), time to fatigue for subsequent endurance tasks (Amann 2012), and proprioception (Lee, Liau et al. 2003, Enoka and Duchateau 2008, Emery and Cote 2012). Proprioceptive decrements due to fatigue arise from a combination of neuromuscular factors, notably: type III/IV afferent inhibition following nociceptive responses to delayed-onset muscle soreness (DOMS), inflammation (Gandevia 2001, Hyldahl and Hubal 2014), alterations in the muscles normal force-length relationship (Riemann and Lephart 2002), and mechanical damage sustained to fusimotor muscle segments from eccentric loading (Laszlo 1992, Torres, Vasques et al. 2010). In typical skeletal muscle, joint proprioceptive performance decrements are observed following muscle fatigue of the local prime movers, although it has also been shown that these local decrements to proprioception can be compensated for at the distal limbs (Emery and Cote

2012). However, fatigue of neck muscles is unique in its observed capacity to induce widespread systemic sensorimotor decrements (Zabihhosseinian 2014).

Neck fatigue has also been shown to result in sensorimotor decrements in all joints of the upper limb (Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2017, Reece 2019, Zabihhosseinian, Yielder et al. 2019), with the majority of research focused on the elbow as the local joint mechanics are the simplest to quantify. However, the shoulder is responsible for the greatest range of upper limb movement and it is foundational to the orientation of the distal upper limb joints (Halder, Itoi et al. 2000). It is due to the appreciable degrees of freedom (DOF) at the shoulder that the joint mechanics here can be very difficult to quantify (Halder, Itoi et al. 2000). Therefore the shoulder has been comparatively under researched compared to the elbow, with only a couple of studies in literature examining the impact of any form of altered sensory input to the neck on shoulder proprioception and mechanics (Zabihhosseinian, Holmes et al. 2017, Zabihhosseinian, Yielder et al. 2019). The shoulder also relies on axioscapular muscles to contribute to scapulohumeral rhythm for the purpose of unlocking the glenohumeral joint for greater mobility (Prescher 2000). It can therefore be assumed that fatigue of neck musculature may have a confounding effect on scapular orientation, and subsequently directly impact many planes of motion for the shoulder. This can make it difficult to isolate proprioceptive decrements in the shoulder which are directly resultant of altered body schema. There has yet to be a study examining the effect of neck muscle fatigue on a shoulder proprioceptive task that does not also implicate the locally fatigued axioscapular musculature. Thoracohumeral rotation may provide a shoulder plane of motion for analysis which is not also heavily influenced by axioscapular mechanics.

Therefore, the overall objective of this thesis was to explore altered mechanics in a humeral rotation joint position re-creation task that would result from the proprioceptive decrement symptomatic of neck muscle fatigue.

1.1. Objectives of Thesis

1. To explore the decrement in thoracohumeral proprioception following fatigue of the cervical extensor muscles (CEM).
2. To determine the accuracy of a novel Shoulder Joint Position Sense (JPS) Measurement Device in quantifying humeral rotation about the thorax.

1.2. Hypothesis of Thesis

The following hypotheses were tested, where H_0 represents the null hypothesis and H_A represents an alternative hypothesis:

$H1_0$: Thoracohumeral joint position reproduction (JPR) will not change significantly following the induction of significant CEM fatigue.

$H1_A$: Thoracohumeral JPR will be significantly less accurate following the induction of significant CEM fatigue.

$H2_0$: The Shoulder JPS Measurement Device will not accurately capture unconstrained humeral rotation about the thorax.

$H2_{A1}$: The Shoulder JPS Measurement Device will capture unconstrained humeral rotation about the thorax with the confounding effect of unconstrained motion at one or more upper-limb joints being significantly responsible for measurement inaccuracies.

$H2_{A2}$: The Shoulder JPS Measurement Device will accurately capture unconstrained humeral rotation about the thorax without being confounded by other upper-limb joint DOF.

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CHAPTER 2.

QUANTIFYING THE EFFECT OF NECK MUSCLE FATIGUE ON ALTERED BODY SCHEMA & PROPRIOCETION: A REVIEW OF THE LITERATURE

The overall objectives of this thesis were: 1) to explore the decrement in thoracohumeral proprioception following significant fatigue of the CEM, and 2) to determine the reliability and validity of a novel Shoulder JPS Measurement Device in accurately quantifying thoracohumeral rotation. The hypothesis for study one was that neck muscle fatigue would alter sensory feedback from the neck, thus depreciating the accuracy of shoulder joint proprioception reflecting disruptions to internal body schema. The hypothesis for study two was that at least one other cardinal plane of upper limb motion besides glenohumeral rotation would explain the variance in performance of a shoulder joint position matching task. This thesis will further our understanding of the neurophysiological impact of neck muscle fatigue on upper limb motor task performance and the way in which neck muscle fatigue impacts neurophysiological and behavioral responses.

2.1. INTRODUCTION TO LITERATURE REVIEW

This literature review covers the background literature relevant to the two studies of this thesis. It begins with an overview of neuromuscular fatigue mechanisms and what is known about changes in sensory feedback from fatigued muscle. It then discusses the concept of body schema and the role of proprioception in creating that schema. Then the impact of altered neck sensory feedback on upper limb proprioception is discussed. Next the technological considerations for proper description and quantification of the shoulder kinematics are reviewed. Finally, we review the anatomical considerations relevant to the chain of upper-limb musculature originating from the axioscapular muscles.

2.2. MUSCLE FATIGUE

2.2.1. Defining Muscle Fatigue

Muscle fatigue is defined as an exercise-induced reduction in maximal voluntary muscle force (Gandevia 2001). Critical to the definition of muscle fatigue are the three following factors (Williams and Ratel 2009): 1) There is a decline in one or more of the biological systems, be it peripheral or central, 2) The decline is reversible, and 3) The decline may or may not occur before an observable performance or task failure occurs. Muscle fatigue has been defined in numerous ways (see table 2.1) but the consistent feature is an inability to maintain force production. Peripheral fatigue describes factors which contribute to decrements in muscle output that are involved in physiological processes distal to the neuromuscular junction. Central fatigue describes factors which contribute to decrement in muscle performance that are involved in neurophysiological processes within the motoneurons and CNS. In most volitional cases of fatigue onset, fatigue presents as a combination of both factors (Gandevia 2001).

Table 2.1: Definitions of Fatigue in Literature. Adapted from (Williams and Ratel 2009)

A reversible state of force depression, including a lower rate of rise of force and a slower relaxation.	**(Fitts and Holloszy, 1978)**
The failure to maintain a required or expected force.	**(Edwards, 1981)**
Muscle fatigue is a decline in the maximal contractile force of the muscle.	**(Vøllestad, 1997)**
The inability to maintain of a physiological process to continue functioning at a particular level and/or the inability of the total organism to maintain a predetermined exercise intensity.	**(Fifth International Symposium on Biochemistry of Exercise, 1982)**
Reduction in the maximal force generating capability of the muscle during exercise.	**(Miller et al., 1995)**
Any reduction in the force-generating capacity (measured by the maximum voluntary contraction), regardless of the task performed.	**(Bigland-Ritchie and Woods, 1984)**
A loss of maximal force generating capacity.	**(Bigland-Ritchie et al., 1986)**
A condition in which there is a loss in the capacity for developing force and/or velocity of a muscle, resulting from muscle activity under load which is reversible by rest.	**(NHLBI, 1990)**
Any reduction in a person's ability to exert force or power in response to voluntary effort, regardless of whether or not the task itself can still be performed successfully.	**(Enoka and Stuart, 1992)**
Any exercise-induced reduction in the maximal capacity to generate force or power output.	**(Vøllestad, 1997)**
Intensive activity of muscles causes a decline in performance, known as fatigue.	**(Allen and Westerblad, 2001)**
Performing a motor task for long periods of time induces motor fatigue, which is generally defined as a decline in a person's ability to exert force.	**(Lorist et al., 2002)**
The development of less than expected amount of force as a consequence of muscle activation.	**(McCully et al., 2002)**
Fatigue is known to be reflected in the EMG signal as an increase of its amplitude and a decrease of its characteristic spectral frequencies.	**(Kallenberg et al., 2007)**

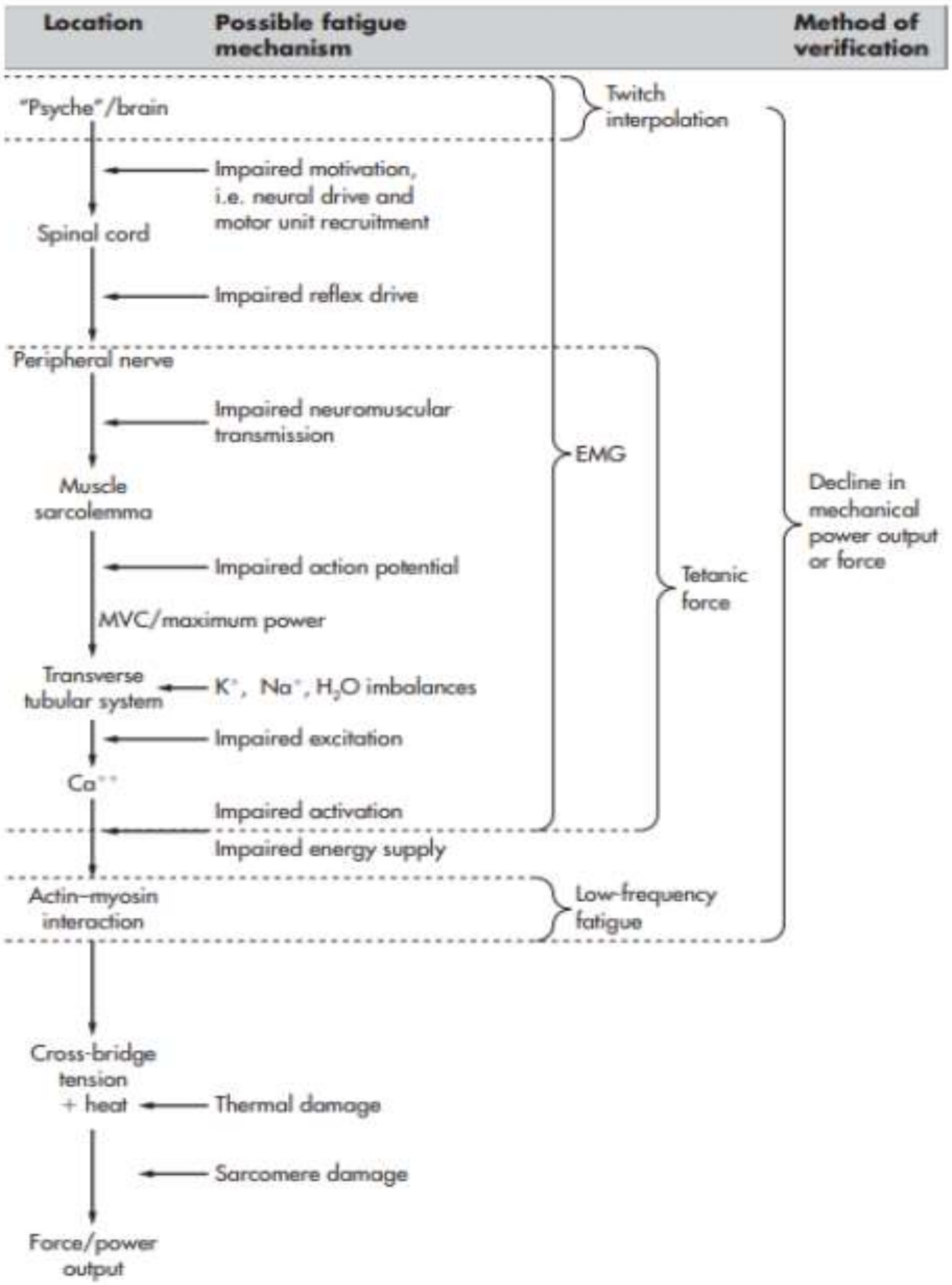

Figure 2.1: Descending Chain of Neuromuscular Fatigue Factors. Adapted from (Williams and Ratel 2009)

2.2.1.1. Metabolic and Mechanical Factors Influencing Fatigue of Muscle Tissue

Foundational work in 1904 by Mosso was the first of its kind to show that fatigue occurs in the muscle as well as the CNS (Mosso 1904). Mosso electrically stimulated a finger muscle nerve and observed that fatigue was still induced rapidly (Mosso 1904). At the physiological level, fatigue manifests in the muscle tissue due to a number of factors. Enoka and Duchateau (2008) summarize the physiological constituents contributing to muscle fatigue as a combination of intracellular metabolite buildup due to blood flow occlusion, mechanical damage to sarcomeres, and the decline of finite muscle glycogen and oxygen levels (Enoka and Duchateau 2008).

During exercise, our breathing rates increase in an attempt to supply more oxygen to our working muscles (Vatner and Pagani 1976). This is because our bodies prefer to subsist on aerobic energy systems whenever possible, in order to minimize the production of lactate and other acidic metabolites in the muscle (Vatner and Pagani 1976). But as relative task demand intensity increases, so too does the local tissue requirement for oxygen (Vatner and Pagani 1976). While greater limits to our cardiovascular systems potential to transport oxygen are observed in more elite athletes, the aerobic energy system eventually capitulates at the aerobic threshold (AeT) around 60% of an individual's maximum aerobic capacity (Viitasalo, Luhtanen et al. 1985). As intermuscular pressure begins to rise it can obstruct the delivery of oxygen to the local tissues due to the occlusion of local blood vessels (Allen, Lamb et al. 2008). At such a point, the body is required to begin to supplement energy demands in the absence of fully oxygenated tissue. In normally oxygenated muscle tissue, the Krebs cycle metabolizes pyruvate from glycolysis to create adenosine triphosphate (ATP) (Stryer 1995). This process also produces the additional substrates NADH and $FADH_2$ which ultimately get converted to more ATP through the electron transport chain (Stryer 1995). The current estimate of ATP yield from glycolysis, the Krebs cycle, and the electron transport chain in oxygenated tissue is approximately 30 ATP per glucose with consideration given to what energy is also required to drive these metabolic processes (Meurant 2012). However, when requisite oxygen is not present to promote sufficient aerobic respiration, pyruvate is instead converted into lactate (Ross 2003). Lactate cannot enter the Krebs cycle, and therefore has no further steps through which to provide ATP in the mitigation of oxygen. However producing lactate does at least allow glycolysis to continue to produce some energy for the cell (Vatner and Pagani 1976). For this reason, aerobic cellular respiration is considered superiorly efficient to anaerobic glycolysis due to their respective energy yields.

The buildup of lactate in muscle cells is often attributed to DOMS, however literature would convey that this is a misappropriation (Cheung, Hume et al. 2003). Lactate accumulation in muscle cells only last for about 1-3 minutes, at which point the local tissue pH becomes so acidic that the metabolites responsible for glycolysis begin to fail, and the tissue can no longer generate sufficient ATP for energy (Cheung, Hume et al. 2003). Moreover, type III/IV afferents usually inhibit motor output of a sufficient contraction before this physiological limit can occur (Gandevia, Allen et al. 1996). This feedback loop acts as a protective mechanism to beget volitional muscle fatigue, which immediately reduces the hydrostatic pressure within the muscle as blood vessels become less occluded (Allen, Lamb et al. 2008). In healthy individuals, blood supply to skeletal muscle collapses when muscle force output exceeds approximately 50% of maximum (Allen, Lamb et al. 2008). In most cases, reaching the point of fatigue will also directly preclude a recovery phase from the exercise which will promote excess post-exercise oxygen consumption (Gaesser and Brooks 1984). The increase in tissue perfusion in combination with the rise in baseline oxygen consumption will also aid in the transition of lactate back to pyruvate, allowing some metabolic uptake to complete the Krebs cycle and the remainder being transported to the liver by the blood (Gaesser and Brooks 1984). This process circumvents the instigation of rhabdomyolysis unless extreme mechanical muscle damage is induced (Pearcey, Bradbury-Squires et al. 2013).

A primary characteristic of skeletal muscle is its ability to adapt to chronic demands and increase its performance capacity (Enoka and Duchateau 2008). However, the complete understanding of the molecular events which drive sarcomere remodeling still remain partially unclear (Orfanos, Gödderz et al. 2016). During eccentric exercise, the muscles are forcibly lengthened. This process can overload the composite units of sarcomeres: the myofibrils. When these contractile units reach mechanical failure in their forcibly extended eccentric state, they can potentially rupture (Orfanos, Gödderz et al. 2016). The lesions produced at the site of rupture are found across the sarcomere post exercise and they appear as focal disruptions in the myofibril pattern. The decline in force post exercise is thought to be heavily attributed to these lesions, which interrupt connections between Z-discs, reducing the contractile capacity of the damaged segments (Orfanos, Gödderz et al. 2016). The mechanical rupturing of cells within post exercise muscle tissue also promotes an acute local inflammatory response. This acute muscle inflammation response has also been linked to post exercise decreases in muscular potential by instigating further tissue breakdown (Carroll, Taylor et al. 2016).

2.2.1.2. Factors Influencing Lower Motor Neuron Neuromuscular Fatigue

In volitional movement, muscle tissue is stimulated by excitatory chemical synapse at the junction between the sarcoplasmic reticulum and the lower motor neuron (LMN) axon (Porter and Lemon 1993). The site for electrochemical transmission from nervous tissue to muscle tissue is referred to as the neuromuscular junction. In vertebrates, motor neurons primarily release acetylcholine (ACh) which diffuse across the synaptic channel and bind to receptors in the sarcolemma (Adam and De Luca 2003). The summative binding of ACh molecules on the sarcolemma depolarizes muscle tissue, and when action potential threshold is reached, excitation-contraction coupling occurs (Dale, Feldberg et al. 1936).

As the definition of muscle fatigue is a decline in muscle performance associated with prolonged activation, muscle tissue components explored earlier help explain the development of fatigue over repeated or isometric exertions, which can interrupt the energy pathways of muscle tissue (Allen, Lamb et al. 2008). However, what is yet to be explained is the seemingly immediate and progressive decrease in muscle force generation overserved after even a single tetanic contraction. Figure 2.2 below adapted from (Allen, Lannergren et al. 1995) illustrates the force records from Flexor Digitorum Brevis across repeated short-duration tetanic contractions. The top panel is an illustration of the normal cascade of force output across subsequent contractions, whereas the bottom panel illustrates the same Flexor Digitorum Brevis tissue in the presence of cyanide to inhibit mitochondrial oxidative phosphorylation. The comparison is clear to illustrate the performance benefits provided by aerobic energy pathways, however in either panel, the progressive and immediate decline from the first tetanic MVC suggests that there are contributions to fatigue beyond intramuscular components.

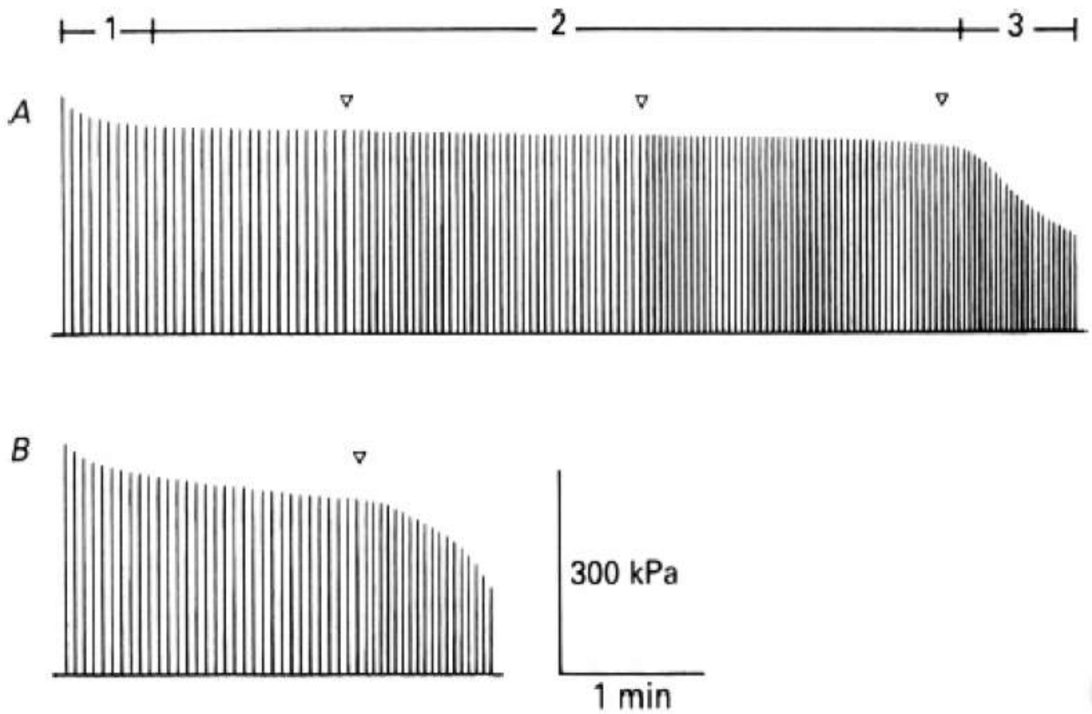

Figure 2.2: Aerobic and Anaerobic Fatigability of Flexor Digitorum Brevis. Adapted from (Allen, Lannergren et al. 1995)

At the neurophysiological level, elements of neuromuscular fatigue arise at both the LMN and upper motor neuron (UMN) (Gandevia 2001). Traits of fatigue, attributable to defects in LMN and muscle function are characterized as peripheral fatigue, whereas declines in output from the UMN and supraspinal anatomy are termed central fatigue.

Eberstein and Sandow (1973) were among the first to implicate the current model of excitation-contraction coupling contributing to fatigue by perfusing a fatigued muscle with caffeine, facilitating the expenditure of calcium ions from the sarcoplasmic reticulum to endure muscle contraction (Eberstein 1963). A decade later, Burke et al. (1973) stimulated a cat muscle specimen to exhaustion – identified by the intramuscular depletion of glycogen (Burke, Levine et al. 1973). Together these two studies provided the initial understanding of differences in fatigability between slow and fast twitch muscle fibers and their parent motoneurons; where fast twitch motoneurons fatigue absolutely (Burke,1973), and slow twitch motoneurons can present as essentially unfatigable (Eberstein and Sandow,1963). However, where the distinction in twitch recruitment arises is when selecting the appropriate force output to a given task (Allen, Lamb et al. 2008). Due to their relatively high resistance to fatigue, slow twitch motoneuron fibers make the obvious choice to recruit for very low intensity tasks such as activities of daily living (Henneman 1957). The tradeoff is that while slow twitch motoneurons induce the slowest rate of fatigue, their maximal force threshold is heavily tempered. In contrast, as motoneurons of exceedingly faster twitch are recruited they also demonstrate a capacity for greater muscle fiber recruitment and subsequently greater force output (Potvin and Fuglevand 2017). This delicate balance of selecting the most efficient harmony of motoneurons to a task is regulated at the UMN level and will be discussed in that section, however it is important to understand here that while UMN's optimize LMN selection and modulation to be their most efficient, once task intensity exceeds the capacity of fatigue-resistant motoneurons, an eventual progression towards motoneuron fatigue initiates (Potvin and Fuglevand 2017).

LMN fatigue is described previously by Enoka and Duchateau as the slowing of the release of neurotransmitters at the neuromuscular junction (Enoka and Duchateau

2008). The progressive decline in neurotranismitter and chemotransmitter concentration at the sarcoplasmic reticulum is met with a corresponding reduction in muscle tissue excitability (Enoka and Duchateau 2008). This processes continues through a series of inhibitory and excitatory modulations intended by the UMN to continually optimize the recruitment of vital LMNs and their muscle fiber pools. However this is ultimately a terminal process. The process by which motoneuron pools are modulated and optimized for task demands will be discussed in the following section on UMN fatigue properties.

2.2.1.3. Factors Influencing Upper Motor Neuron Neuromuscular Fatigue

Where the LMN is the point for the integrative transduction of signals to the muscle tissue, the UMN is the section of the neuromuscular system from which these modulatory signals originate (Allen, Lamb et al. 2008). This comes primarily in the form of the upregulation and downregulation of LMN activity which is termed as nervous excitation and nervous inhibition (Allen, Lamb et al. 2008). The purpose of targeted LMN excitation and inhibition is to manipulate the threshold to which LMNs will achieve action potentials. Through this process the CNS can set a hierarchical order in which muscle fibers will be recruited to a task, and what summative and temporal potential they will be recruited to (Allen, Lamb et al. 2008). The information to the CNS which evaluates this neuromuscular modulation is based on an afferent:efferent feedback loop, where continuous sensory afference of muscle fatigue informs and updates a compensatory motor plan (Contessa, De Luca et al. 2016). A number of different symptoms of neuromuscular fatigue can affect the compensatory and contributory mechanisms of UMN factors in fatigability. The interrelationship between exercise characteristics, physiological responses, and fatigue induction currently remains to be fully investigated (Carroll, Taylor et al. 2016). However, the primary known factors which modulate LMN excitability; namely type III/IV afferent feedback, compensatory motor unit recruitment, and central drive, will be discussed here.

Group III/IV muscle afferents, termed ‘ergoreceptors’ are known to play a key role in regulating LMN changes to muscle fiber activation (Gandevia 2001). Group III/IV afferents originate at skeletal muscle tissue and synapse at various sites within the CNS to provide inhibitory feedback in the regulation of central drive (Hill 1938). This was first observed during maximal isometric exercise of a single muscle (Gandevia 2001). One of the primary instigators to toggle the firing of III/IV afferents is the occlusion of blood flow to muscle tissue (Gandevia, Allen et al. 1996). Studies manually arresting blood flow to the distal limb observed decrements in central motor drive and subsequently voluntary muscle activation which remained consistent until circulation was restored, reinstating the normative frequency of III/IV afferent feedback (Gandevia, Allen et al. 1996). A study by Amann (2011) suggested that III/IV afferents had a critically relevant inhibitory effect on the regulation of central motor drive when they measured III/IV afferent feedback in subjects performing a 5 km cycling time trial (Amann 2011). The participants had been administered intrathecal fentanyl to block the attenuation of III/IV muscle ergoreceptors and the results showed that the CNS tolerated a substantially higher power output in addition to elevated factors of peripheral fatigue manifestation (Amann 2011). Results from a meta-analysis of relevant literature by Amann (2012) leads the author to propose that the CNS processes neural feedback from type III/IV afferents and adjusts central motor drive to LMNs to confine the development of peripheral skeletal muscle fatigue as a protective mechanism (Amann 2012).

As muscle fatigue develops as a result of sustained contraction, active motor units exhibit an increased rate of temporal and spatial summation thereby progressively contributing additional motor units to sustain force output. (Adam and De Luca 2003, De Ruiter, Elzinga et al. 2005, Contessa, De Luca et al. 2016, Potvin and Fuglevand 2017). These adaptations result from an increase in the excitation of the LMN pool to maintain muscle output towards task demands, despite fatigue induced reduction in tissue capacity and active LMN twitch (Bigland-Ritchie, Dawson et al. 1986, Potvin and Fuglevand 2017). The process of recruiting supplementary LMNs to persist in meeting peripheral demands as initial motor units approach failure represents the UMN's selection of the most appropriate LMNs to meet the task (Adam and De Luca 2003, Contessa, De Luca et al. 2016, Potvin and Fuglevand 2017). This process has been shown to mediate the necessity for increased central motor drive from the UMNs and cortex, and therefore may be one example of the CNS finding the most efficient method of task mediation (Adam and De Luca 2003, Potvin and Fuglevand 2017). However once the LMN reaches collective exhaustion, the only way in which the UMNs can continue to meet task demands at such a critical point in fatigue development is to increase descending central motor drive (Contessa, De Luca et al. 2016).

Central motor drive is considered the primary supraspinal influence on muscle fatigability. During exercise, there are observed modulations in kephalinergic, dopaminergic, and serotonergic systems in the brain (Hoffmann, Terenius et al. 1990, Bailey, Davis et al. 1993, Gandevia 1998). These are understood to control vigilance and motivation at the basal ganglia and also influence neuroendocrine alterations in the availability of substrates necessary for muscle contraction (Gandevia 1998). In humans, cortical output monosynaptically excites most spinal neurons (Porter and Lemon 1993). With the processes that govern inhibition as an aside, supraspinal UMN centers which alter corticomotoneruonal output directly impact LMN excitability (Gandevia 2001). In humans, the magnitude of EMG responses during submaximal contractions have been shown to decline in response to transcranial magnetic stimulation of the cortex (Zanette, Bonato et al. 1995). This requires an increase in descending corticomotoneruonal output to achieve consistent excitability in LMN's (Zanette, Bonato et al. 1995). It is this descending magnitude of corticomotoneuron potential which is termed central motor drive (Gandevia 2001).

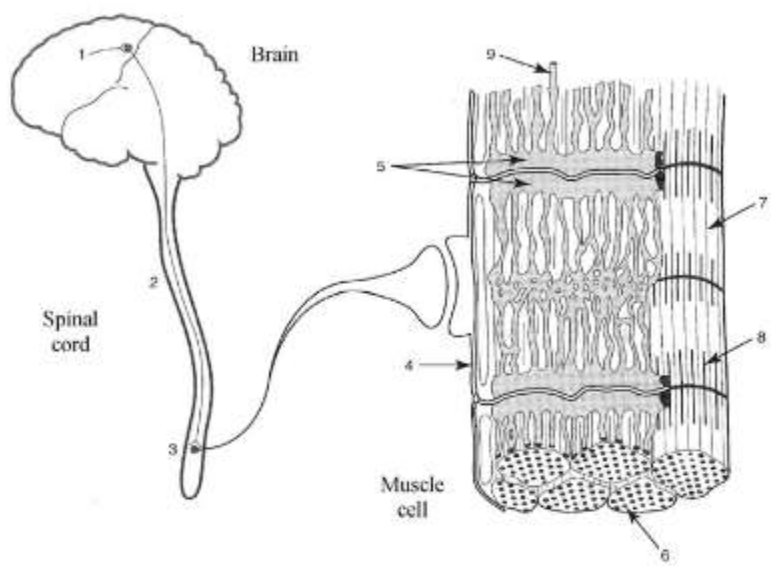

Figure 2.3: Locations in the Neuromuscular System which Affect Fatigability . Adapted from (Williams and Ratel 2009). 1. Cortical Output, 2. Descending Motor Drive, 3. LMN Selection, 4. Synaptic Transmission, 5. Signal Propagation, 6. Sarcomere Rupture, 7. Damage to Actin Filaments, 8. Damage to Myosin Filaments, 9. Damage to Filament Cross-Bridges.

2.2.2. Evaluating Neuromuscular Muscle Fatigue

Neuromuscular fatigue can develop at multiple sites in the neuromuscular system (Fig. 2.4). Therefore, in order to accurately quantify and describe the different symptoms of fatigue, there exists a diverse array of measures (Gandevia 2001, Williams and Ratel 2009). Common methods and protocols utilized in the quantification of neuromuscular fatigue include muscular force output and tetanic force, muscle electromyography (EMG), and twitch interpolation (TI) (Chaffin 1973, Vøllestad 1997).

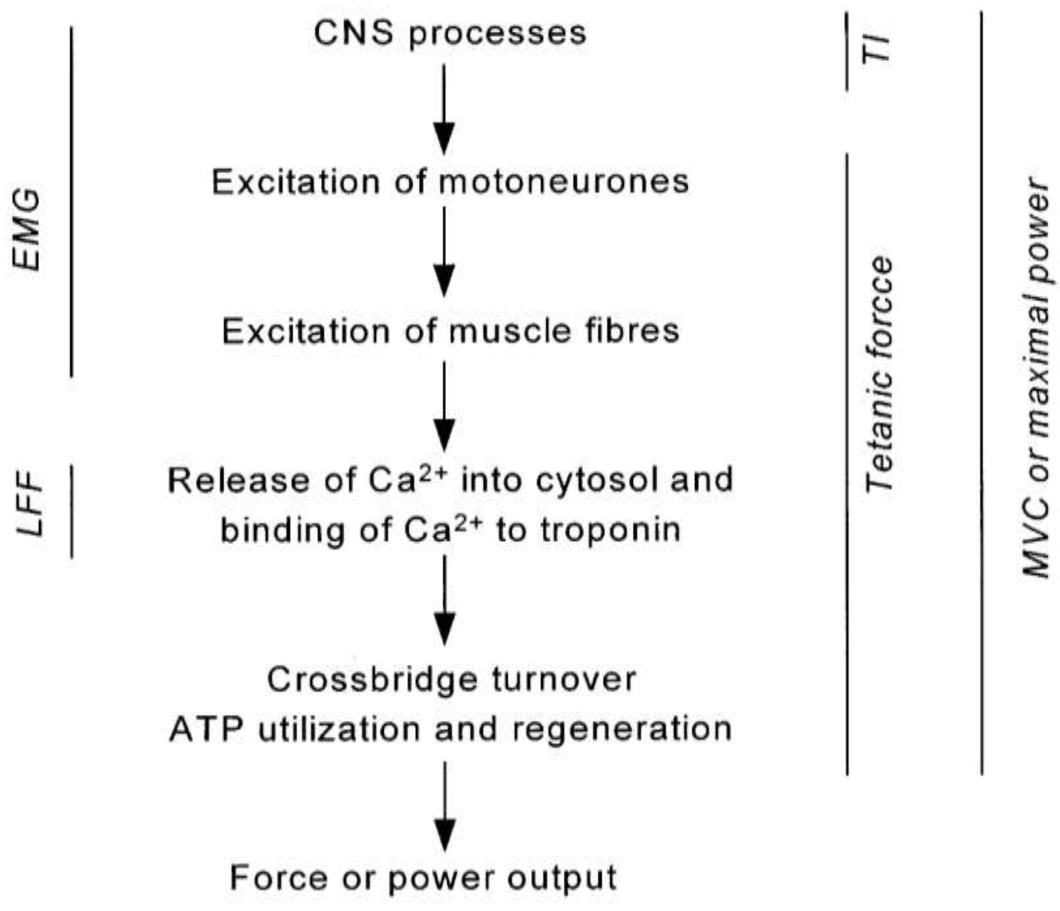

Figure 2.4: Schematic of Various methods to Quantify Muscle Fatigue and where they interact with the Neuromuscular System. Adapted from (Vøllestad 1997) where EMG stands for electromyography, LFF stands for low-frequency fatigue, and TI stands for twitch interpolation, and MVC stands for maximum voluntary contraction.

Measurement of muscular force output is one of the most common measures of muscle fatigue. This is because it represents a direct assessment of the multiple factors in the neuromuscular system which contribute to total force generating capacity (Vøllestad 1997). To get a reliable estimate of the total force output for a given task, participants are asked to contract an agonist muscle along its common moment trajectory to their maximum exertion. This method only describes the participant's maximum volitional contraction (MVC), as their true physiological maximum will likely be inhibited by lack of motivation driving descending output from the UMN's as well as inhibitory safety mechanisms at the LMN and muscle (Gandevia, Allen et al. 1995, Vøllestad 1995, Windhorst and Boorman 1995). Gandevia (1995) suggests the best method to instigate a true maximum force output is to stimulate a maximum evocable voluntary contraction. This is done by electrically stimulating the nerve. This process utilizes tetanic activation of the LMN and subsequent muscle to bypass the CNS, eliminating the fatiguing factors from sub-optimal central drive (Gandevia, Allen et al. 1995).

Muscle electromyography (EMG) represents the most common indirect method for quantifying muscle fatigue (Vøllestad 1997). There are two types of muscle electromyography: surface electromyography and intramuscular or indwelling electromyography (Bourne, Choo et al. 2011). Surface EMG is non-invasive, as the procedure involves adhering bi-polar electrodes to the surface of the skin that overlays the belly of the muscle under analysis (De Ruiter, Elzinga et al. 2005, Bourne, Choo et al. 2011). The surface electrode picks up the electrical potential propagating along the muscle fibers; however, surface EMG is also sensitive enough to pick up many of the other electrical potentials in that region including nearby muscles (Backus, Tomlinson et al. 2011).

Intramuscular EMG involves inserting an electrode needle into the tissue in question and is considered to be significantly more specific, as it greatly reduces signal artifacts, muscle crosstalk, and impedance from peripheral tissues (Backus, Tomlinson et al. 2011). Both EMG techniques utilize the electrodes to pick up electrical activity from the motor units as they depolarize to produce motor unit action potentials. The amplitude and power spectrums from EMG can be used to estimate the magnitude of activation and what fiber types are being recruited respectively (Vøllestad 1997). This method is considered indirect because EMG readings do not give a clear representation of an individual's state of fatigue (Vøllestad 1997). As neuromuscular fatigue is induced, EMG amplitude will eventually increase due to increased descending motor drive from the supraspinal regions (Vøllestad 1997, Allen, Lamb et al. 2008). The EMG frequency spectrum will also shift as additional larger motor units are selectively recruited to sustain the required task demands under fatiguing conditions. As such, EMG readings will not give an accurate depiction of how close the participant is to volitional fatigue as the scale of readings is technically continuous. Additionally, fatigue will inevitably skew neuromuscular responses to a task, making subsequent tasks of the same collection period difficult to compare to each other (Williams and Ratel 2009, Bourne, Choo et al. 2011). Moreover, it is suggested that EMG between different collections may not always be accurate as minute changes in electrode position can intercept significantly different electromotor signals (Bourne, Choo et al. 2011, Gardiner 2011).

Where force output and EMG reflect measures of fatigue quantification at the muscle tissue and LMN respectively, twitch interpolation is the logical inclusion of an assessment for UMN and cortical effects on fatigue (Bigland-Ritchie, Dawson et al. 1986, Vøllestad 1997). This method is a modified version of the tetanic stimulation method presented previously (Merton 1954, Bigland-Ritchie, Dawson et al. 1986, Gandevia and McKenzie 1988). For this protocol, participants are motivated to produce a muscular force which can be anywhere along their spectrum of potential (Gandevia and McKenzie 1988). During this muscular contraction, the parent LMN, UMN, or cortical region, is electrically stimulated which results in a brief force increment (Gandevia and McKenzie 1988). The observed increment in force generation represents the force reserve (Fig. 2.5). The relationship between unfatigued and fatigued twitch interpolation can be extrapolated to deduce the relative output of the submaximal force compared to the maximal evoked potential (Gandevia and McKenzie 1988). Stimulating at different regions in the ascending chain of the neuromuscular system integrates more structures into the twitch stimulated differential, giving a clearer representation of what structures may be contributing to the symptoms of fatigue (Gandevia and McKenzie 1988).

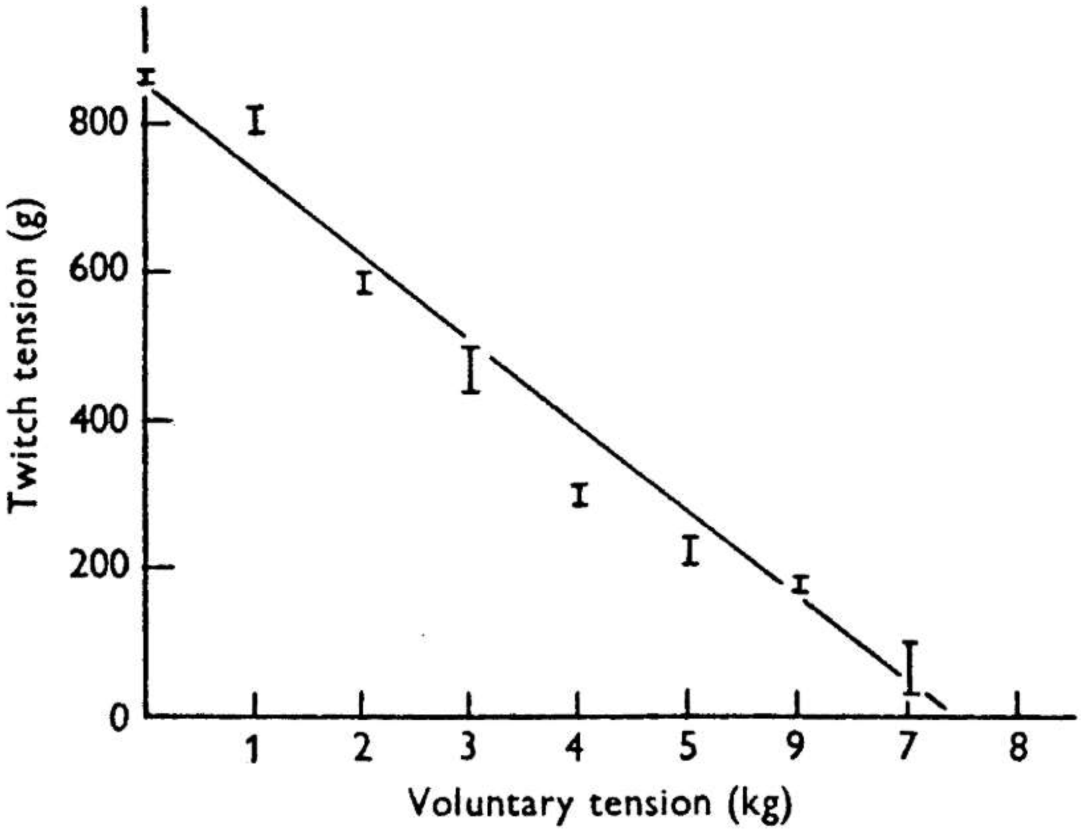

Figure 2.5: The Relationship between Twitch Force and Voluntary Force. Adapted from (Merton 1954)

2.2.3. Effects of Fatigue on Afferent Feedback

Afferent proprioceptive feedback relays from impulses transmitted by specialized mechanoreceptors to the CNS. Proprioceptive afference relays information on JPS, joint movement sense, and muscle tension. Several studies have examined proprioceptive feedback from muscle tissues after inducing a fatiguing stimulus to them. This has been observed in upper limb musculature, lower limb musculature, and postural reflex muscles (Barrack, Skinner et al. 1983, Worringham, Stelmach et al. 1987, Blasier, Carpenter et al. 1994, Voight, Hardin et al. 1996, Rozzi, Lephart et al. 1999, Barden, Balyk et al. 2004, Gosselin, Rassoulian et al. 2004, Letafatkar, Alizadeh et al. 2009, Torres, Vasques et al.

2010, Wong, Wilson et al. 2011, Cuğ, Ak et al. 2012, Hyldahl and Hubal 2014, Proske 2015, Zabihhosseinian, Holmes et al. 2015). Acute effects of muscle fatigue on proprioception are typically studied under eccentrically loaded muscles with the understanding that eccentrically overloaded muscles will produce significantly more tissue micro-trauma (Ribeiro and Oliveira 2011, Hyldahl and Hubal 2014). The hypothesized interaction here is that the inflammatory cascade which results from tissue trauma releases bradykinin and chemokine substrates that initiate a nociceptive DOMS response (Hyldahl and Hubal 2014). This is to suggest that the proprioceptive decrement to local tissues is related to the amount of mechanical tissue trauma rather than fatigue symptoms proximal to the neuromuscular junction. Riberio and Oliveira (2011) observed that decreased proprioceptive JPS accuracy due to muscle fatigue may leave impairments up to 24 hours (Ribeiro and Oliveira 2011).

Proprioceptive deficits associated with fatigue return to normal following recovery from fatigue, however when stimuli overloads tissues to the point of injury, proprioceptive decrements remain unless the structures are repaired. Lephart et al (1994) examined individuals with healthy, unstable, and surgically repaired shoulders. Their findings suggested that individuals with unstable shoulders were significantly less accurate than healthy and surgically repaired shoulders, which showed no significant interactions (Lephart, Warner et al. 1994).

2.3. BODY PERCEPTION

In humans there are 3 main reflections of somatic condition. These conditions are interoception, exteroception, and proprioception (Craig 2003). Of these sensory frameworks, exteroception and proprioception are concerned with the perception of the body and its orientation to itself, its segments, and its environment. When executing motor tasks, our body perception is constantly being referenced by matching kinesthetic feedback with our audiovisual afference. Together, this creates a cortical multisensory map of our body's projection in space known in literature as our body schema (Head and Holmes 1911, Proske 2015).

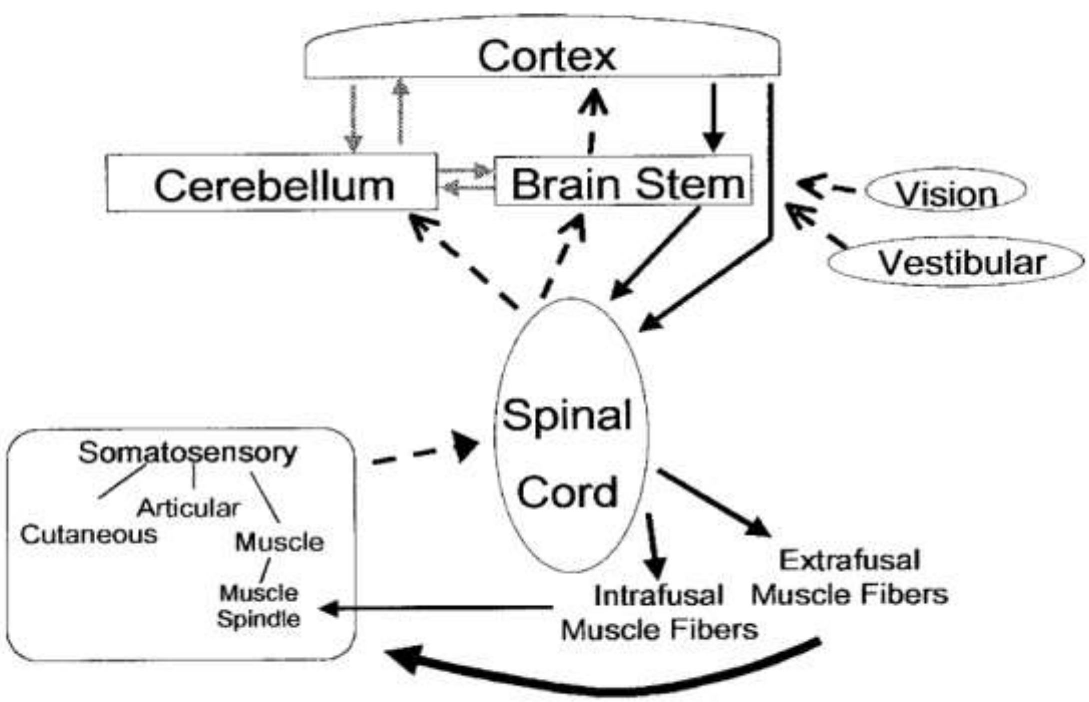

Figure 2.6: Afferent and Efferent Integration in the Human Sensorimotor System. Adapted from (Riemann and Lephart 2002)

Interoception, as the word's root affixes would suggest, describes the body and CNS's latent potential for internally projected perception (Craig 2003). Converging evidence across neuropsychological and psychophysical literature indicates that humans, as well as primates, display a distinct ability to reflect homeostatic afferent activity concerning the physiological conditions of the tissues in their body (Craig 2003). Interestingly, there has long been literature recognizing the binary relationship between parasympathetic afferents and efferents (Cannon 1939), however only in more recent literature has the correlation been extended to sympathetic efferents, and the afferents they would logically synapse with (Craig 2003). The interoceptive system is hypothesized to arise from these sympathetic afferents in the autonomic tracts. Interoceptive representation engenders 'feelings' from the body which may include pain, temperature, itch, muscular and visceral sensations, hunger, and thirst (Craig 2008). The sensory afference regarding these needs is used to construct the subjective image of self-awareness that is entity (Craig 2008).

Where interoception is our perception of the environment within our body, exteroception is our perception of our body within its environment; its external perception (Proske 2015). Exteroception relies heavily on our auxiliary senses of body segment position which would include: visual feedback, auditory feedback, and vestibular sensation (Proske 2015). While these organs have primary functions to identify and understand points of focus within our environment, they provide a secondary subconscious function to supplement our body schema with cues to refine our position sense (Proske 2015).

2.3.1. Position Sense

Awareness of JPS in constructed through a harmony of exteroceptive and proprioceptive afferents. Proske (2015) explains that proprioception relays the contribution of tactile afferents such as muscle spindles, cutaneous receptors, and GTOs to our schema of position sense (Proske 2015). Exteroception involves the auxiliary receptors which integrate our movements without having a direct neuromuscular connection. While both systems contribute to our overall framework of position sense, proprioceptors are especially attuned to consolidate the intrinsic schematic of our body segments positions relative to each other, whereas exteroceptors are more specialized in perceiving our body's position within its environment (Proske 2015). One study reduced radiocarpal proprioception by introducing local anesthetic and found a significant decrease in proprioceptive capacity without occluding vision (Moberg 1985). Jerosch (1995) compared elbow proprioception in healthy individuals and professional table tennis players and found that both groups significantly overestimated joint angles in the absence of vision (Jerosch, Thorwesten et al. 1995). Interestingly, this trend did not recur for joint angles of approximately 90 degrees, suggesting a codominant function between visual feedback and afference. Lastly, a novel study by Goble and Brown in 2008 came to the finding that in the presence of vision, dominant arms performed more accurately at reproducing joint angles, whereas in the absence of vision non-dominant arms performed more accurately (Goble and Brown 2008). This suggests that dominant arms may be attenuated by having a greater reliance on visual feedback, and non-dominant arms may be selectively advantageous when having to rely solely on proprioception. Additionally, while exteroception refines our position sense with additional ques to our body's relative positon, exteroceptive feedback is largely subjective to our attentional foci, whereas proprioceptive feedback is relatively continuous and therefore more heavily relied upon (Proske 2015).

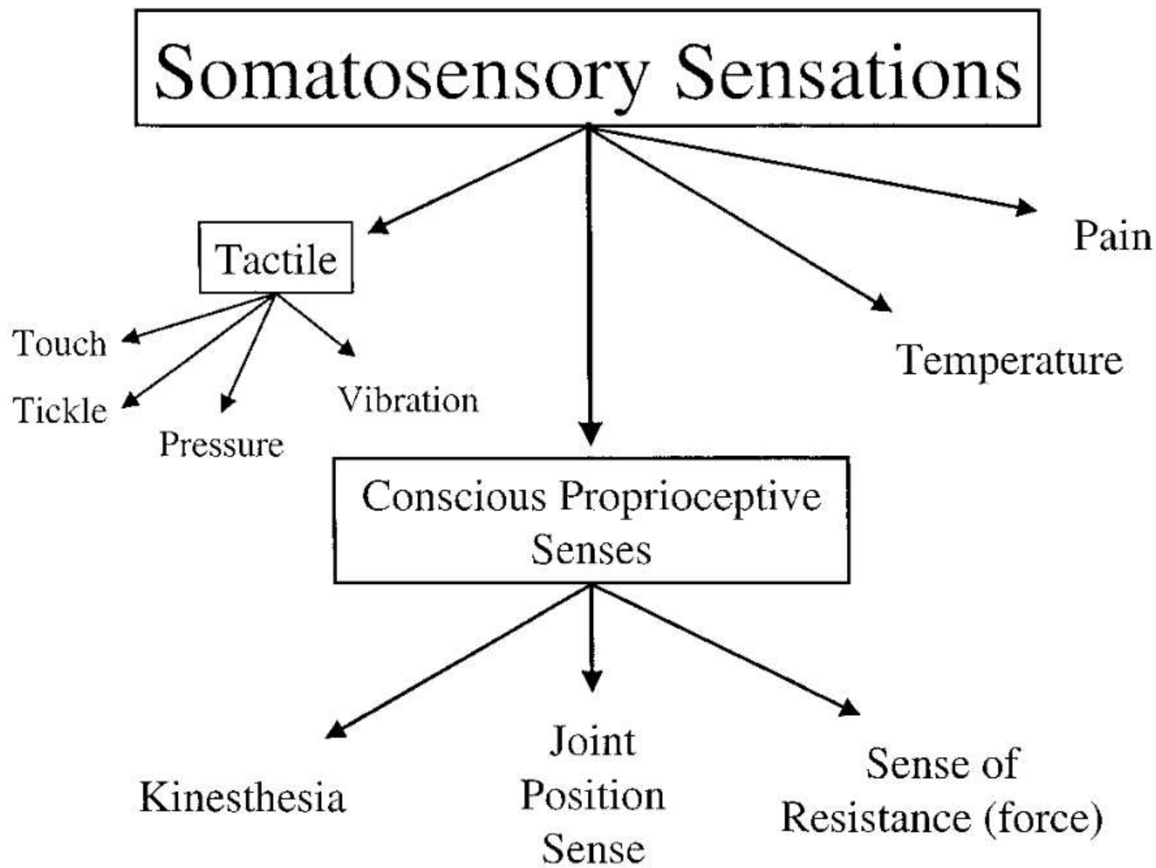

Figure 2.7: Overview of Somatosensory Sensations. Adapted from (Riemann and Lephart 2002)

Before the unique distinction of proprioception within the sensory schematic, Bastian (1880) introduced the term kinesthesia to describe perception encompassing the muscles, tendons and skin (Bastian 1880). In early motor behavior literature the prevailing hypothesis on subconscious positon sense was that there were no peripheral afferents responsible for signaling feedback of muscle movements and contractions

(Müller 1837). Rather this early framework failed to isolate the separate responsibilities between afferent and efferent projections, and it was believed that motor nerves were responsible for relaying feedback. It was not until two decades later that Sherrington (Sherrington 1900) challenged this hypothesis, and subsequently introduced the term proprioception (Sherrington 1906). In early research, Sherrington had observed that motor dural lesions in an anesthetized cat produced expected motor impairments, but failed to produce a decrement in muscle sensation (Sherrington 1898). This provoked Sherringtons hypothesis that muscle tissue contained an organ of sensation (Sherrington 1900).

Another century of literature has appreciated Sherringtons early hypothesis, however until only recently have muscle spindles been implicated into the framework of proprioception. For a significant amount of time, Duchennes early suggestions that joint receptors were the primary organ of afferent influence prevailed (Poore 1883). However, recent definitions of proprioception would not hold up to that model.

Riemann and Lephart (2002) first characterized the modern model of proprioception based on a meta-analysis of past studies (Voight, Hardin et al. 1996, Carpenter, Blasier et al. 1998, Riemann and Lephart 2002). Riemann and Lepharts model of proprioception proposed that there are three main sensations which combine to produce proprioception. These are: 1) a sense of tension (force), 2) a sense of movement, and 3) a sense of relative limb position or joint position sense (JPS). Under Riemann and Lepharts model, muscle spindles make an important contribution, as they are most notably responsible for utilizing alpha-gamma coactivation to detect changes in muscular resistance which are integral to our sense of tension along with contributions from

GTO's. Ruffini and Ruffini-like receptors in cutaneous tissue and joint capsule fibers contribute to our detection of relative changes in motion, direction, velocity, and acceleration. Riemann and Lepharts model distinguishes that our third sense of proprioception – joint position sense – is constructed based on the relationship between our senses of tension and movement.

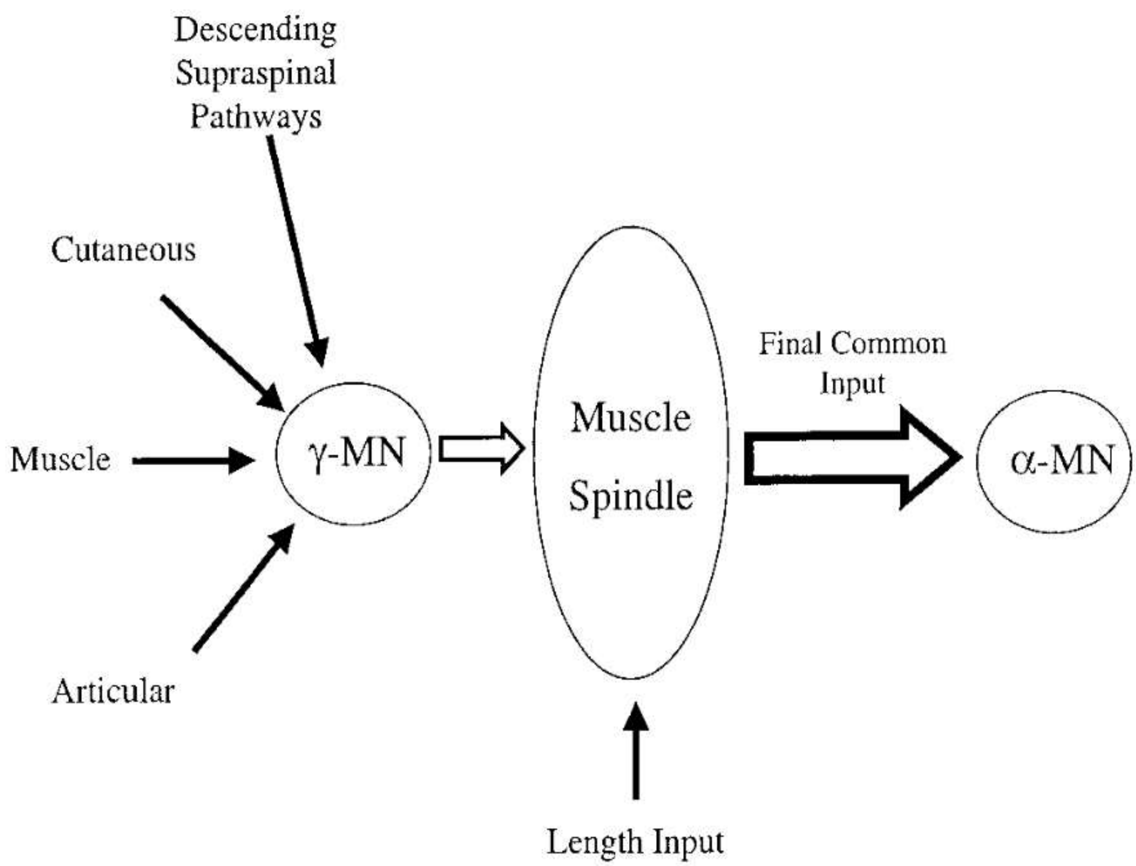

Figure 2.8: Factors Influencing Muscle Spindle Afference. Adapted from (Riemann and Lephart 2002)

2.3.2. Joint Position Sense

Based on Riemann and Lepharts model of proprioception physiology, Riberio and Oliveira conducted a 2011 review on factors which depreciate JPS (Ribeiro and Oliveira

2011). They compiled evidence that aging, muscle fatigue, active/passive muscle physiology, and cutaneous sensation availability all significantly decreased accurate cortical representation. Other literature has also implicated differences between dominant and non-dominant limbs (Goble, Lewis et al. 2006, Goble, Noble et al. 2009, Han, Waddington et al. 2016).

The effects of aging on nervous and biological function is well documented, yet the physiological relationship is not yet fully understood and often debated (Ribeiro and Oliveira 2011). Regardless, literature has been able to set a clear precedent that proprioception clearly increases as a function of age until the second decade of life, and then begins to decline as a function of age (Skinner, Barrack et al. 1984, Kaplan, Nixon et al. 1985, Pai, Rymer et al. 1997, Petrella, Lattanzio et al. 1997, Bullock-Saxton, Wong et al. 2001, Ribeiro and Oliveira 2011). Clinically this is primarily observed as the common deterioration of coordination and balance late in life, as well as through infantile development (Ribeiro and Oliveira 2011). Colledge et al. (1994) made a novel finding in their investigation of kinesthetic awareness at different age brackets (Colledge, Cantley et al. 1994). Their research determined that as age progresses, individuals rely more heavily on their kinesthetic and proprioceptive feedback to maintain their center of balance. However, this can be disadvantageous as aging populations also display a substantial decrease in proprioceptive faculty. These deleterious effects on proprioception are hypothesized to be heavily associated with the high incidence for falls that is so clinically relevant (Lord, Rogers et al. 1999). Skinner also hypothesizes that this balance inaccuracy promotes abnormal biomechanics in activities of daily living which in turn promotes degenerative joint diseases (Skinner, Wyatt et al. 1986).

Muscular fatigue is a common method of disrupting the length-tension relationship in literature because there have been many studies quantifying how to most reliably elicit muscular fatigue (Edmondston, Wallumrød et al. 2008). Inducing muscular fatigue causes cellular disruptions such as DOMS in the muscle tissue as well as fatigue of the efferent nervous projections. Fatigue induced effects of the UMN's can decrease concentration stamina, leading to decreased focus on proprioceptive afference (Thibault and Raz 2016). At the LMN region, type III/IV afferent feedback gain due to fatigue contributes to proprioceptive decrements by volleying with muscle spindles to progressively defacilitate the local alpha motoneruon pool (Taylor, Amann et al. 2016). At the peripheral muscle tissue, nociceptive pain accompanying DOMS increases muscle sensitization, however findings still suggest that this initiates a net decrease in accuracy. The hypothesized mechanism suggests that intrafusal and extrafusal muscle spindle damage accompanies contractile tissue trauma, again acting to defacilitate the motoneuron pool (Torres, Vasques et al. 2010, Hyldahl and Hubal 2014).

Active and passive movement describe the two common methods of joint manipulation. Active joint movement is defined as manipulation of the joint performed by the work of the local musculature (Paillard and Brouchon 1968). This is the most common type of joint manipulation as it is the foundaiton for human movement. Passive movement is defined as a manipulation of the joint performed by the work of other forces external to the joint structure or individual (Paillard and Brouchon 1968). This is the categorization of movement that were to occur if the joint is manipulated by another individual. Numerous studies have examined the differences in kinesthetic awareness, movement accuracy, and JPS between active and passive manipulation of different joints

(Paillard and Brouchon 1968, Laufer, Hocherman et al. 2001, Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2017). All studies conclude that active movement is significantly more kinesthetically accurate then passive. Subsequent findings by Paillard and Brouchon (1968) also suggested that passive proprioceptive tests have a tendency to underestimate joint position relative to starting joint position (-18 mm), whereas active angle matching was significantly more accurate but may display a minor tendency to overestimate (+6 mm) (Paillard and Brouchon 1968). Laufer et al. (2001) postulates that the discrepancies between active and passive movement is due to their inherent physiological differences (Laufer, Hocherman et al. 2001). This is because the involvement of muscle contractions to manipulate joint angles during active movement increases the sensory afference from local muscle spindle fibers, improving JPS (Laufer, Hocherman et al. 2001). In passive movement there is inherently little or no muscle activation, and little to no muscle spindle afference, therefore the body only receives kinesthetic afference from cutaneous nerve endings and GTO's near end of range (Laufer, Hocherman et al. 2001). This same study by Laufer et al. was significant because they unveiled no significant effects of gender on active and passive movement differences (Laufer, Hocherman et al. 2001).

Much of the literature surrounding JPS assumes cutaneous afference to be a supplementary feedback mechanism (Dickinson 1976, Lephart, Warner et al. 1994, Collins and Prochazka 1996, Myers, Guskiewicz et al. 1999, Proske 2015). Studies have used comparisons between active and passive movement to distinguish joint proprioceptive trends and accuracy with and without muscle spindle afference respectively. Laszlo (1992) found that fusimotor activity is greatly diminished in passive movement, requiring a reliance on cutaneous feedback (Laszlo 1992). This shows our brains plasticity in being able to construct body schema from different sources of feedback. However as Han et al. (2016) recently pointed out, different sources of proprioceptive information may be processed at different areas in the brain, where hemispheric specialization could play a role in accuracy as well (Han, Waddington et al. 2016). Forget and Lamarre (1987) have a contribution in literature where they studied goal-directed movements of elbow flexion in normal human subjects as well as in patients deprived of proprioceptive and cutaneous feedback (Forget and Lamarre 1987). They found that the CNS had less coordination in sending ‘bursts’ of electromyographic activity to appropriately accelerate and decelerate the limb in the absence of peripheral feedback. The combined findings of these studies would suggest that the cortical schematic of JPS can subsist in the absence or reliance of cutaneous feedback, however JPS proves most accurate in combination with other afferents.

Data collected between dominant and non-dominant limbs has consistently found an advantageous performance of the non-dominant arm when compared to the dominant arm for joint positon matching tasks (Goble, Lewis et al. 2006, Goble, Noble et al. 2009, Han, Waddington et al. 2016). Han et al. (2016) tested three different joint angle matching methodologies designed to stimulate different cortical structures during proprioception; ipsilateral matching, contralateral matching, and contralateral-remembered matching (Han, Waddington et al. 2016). Across all three methods, they found a significant improvement in JPS accuracy in the non-dominant arm. This research may support earlier research by Goble and Brown (2008), and Jerosch and colleagues (1996), that suggests dominant arm position sense may be more heavily influenced by visual feedback, where non-dominant arm positon relies more strictly on proprioception (Jerosch and Prymka 1996, Goble and Brown 2008).

2.3.2.1. Absolute, Constant, and Variable Error in Proprioception Test Design

There are three classical methods used in psychophysical experiments and they are: 1) A method of adjustment, 2) A method of limits, and 3) A method of constant stimuli (Gescheider 2013). A method of adjustments would be a test were a participant is shown a stimulus, and then is required to increase or decrease their own response to match it accordingly. A method of limits would be a test where participants first deliver a response, and then indicate when an adjustable stimulus meets the same parameters. Lastly, a method of constant stimuli would involve a test where a target stimuli is presented randomly amongst distracting stimuli, and the participant must indicate which the proper target is. It would have it that the three proprioception methodologies in literature follow the constraints of these three tasks.

Perhaps the most common type of proprioception test in literature would be the joint angle re-creation test. There are two types of methodologies for this test. The first is the joint position re-creation (JPR) protocol, where participants are shown a joint positon and asked to then replicate the position shown to them either ipsilaterally or contralaterally. The second type of active test employs an active movement extent discrimination apparatus (AMEDA) (Waddington, Seward et al. 2000, Naughton, Adams et al. 2005, Han, Waddington et al. 2011, Han, Anson et al. 2013). An AMEDA test is a version of the JPR methodology that utilizes the uniquely designed AMEDA apparatus and stipulates that active movements are made under normal weight bearing, without physical constraints, and with the permission of vision (Waddington, Seward et al. 2000, Han, Waddington et al. 2011). AMEDA testing also ensures that information about test performance is given after each trial to allow participants to refine their performance.

The next most common proprioception test in literature would be the threshold to detect passive motion (TTDPM) test which embodies the principle of a method of limits. For a TTDPM methodology, participants have their limb and joint or interest supported in a starting posture. The posture is then manipulated, typically at a random interval of time after the start of the test, and at a relatively slow speed (Han, Waddington et al. 2016). The participant's job is to indicate when they first feel that their limb positon has been manipulated. Being solely a passive test, TTDPM relies heavily on cutaneous innervation, and findings would consolidate this by indicating the TTDPM is most sensitive near end of range, and is also more accurate at faster rates of joint angle change.

The final and least typical proprioception test in literature is the just noticeable difference test which represents a method of constant stimuli. The just noticeable difference procedure requires participants to match a target stimulus, while some sort of outlier exerts a distorting force (Han, Waddington et al. 2016). These tasks show the most utility by imbedding observations on proprioception adaptation in their study design, however they also require a large number of trials, making them unpopular in literature.

Perhaps the three most widely employed descriptive statistics in the field of motor behavior and motor control are constant error (CE), absolute error (AE), and variable error (VE) (Guth 1990). Error statistics have been used in literature for years (Chapanis, Garner et al. 1949, Granit 1972, Crabtree and Antrim 1988, Schmidt, Lee et al. 2018).

Together these three statistics can provide useful data on discrimination error, perceptual direction bias, and reliability (Guth 1990).

In proprioception literature for an active JPR task, constant error measures the deviation between a target angle, and the angle which is reproduced (Knox and Hodges 2005). Constant error is denoted as CE, and its score for a task provides information on the direction of the error. The formula for CE is $\sum[X-X0]/N$ [1].

Absolute error for a JPR task can be considered as the deviation between the target angle and the reproduced angle, except that unlike constant error, absolute error is irrespective of direction (Knox and Hodges 2005). Absolute error is denoted as AE, and it provides meaningful information on the general error of a task. The formula for AE is $\sum|[X-X0]/N|$; essentially AE = |CE [1].

Lastly, variable error for a JPR task is considered the standard deviation, or variability between a target and the reproduced angle(Knox and Hodges 2005). Variable error is denoted VE, and this measure provides information on the reproducibility of the results in a task. Larger variable error would suggest that there is a larger difference between error scores. The formula for VE is $\sqrt{[[\sum[X0-M]^2]/N]}$, essentially VE = standard deviation of CE [1].

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[1] Where X represents raw score, X_0 represents the desired criterion score, N represents number of trials, and M represents mean of the values.

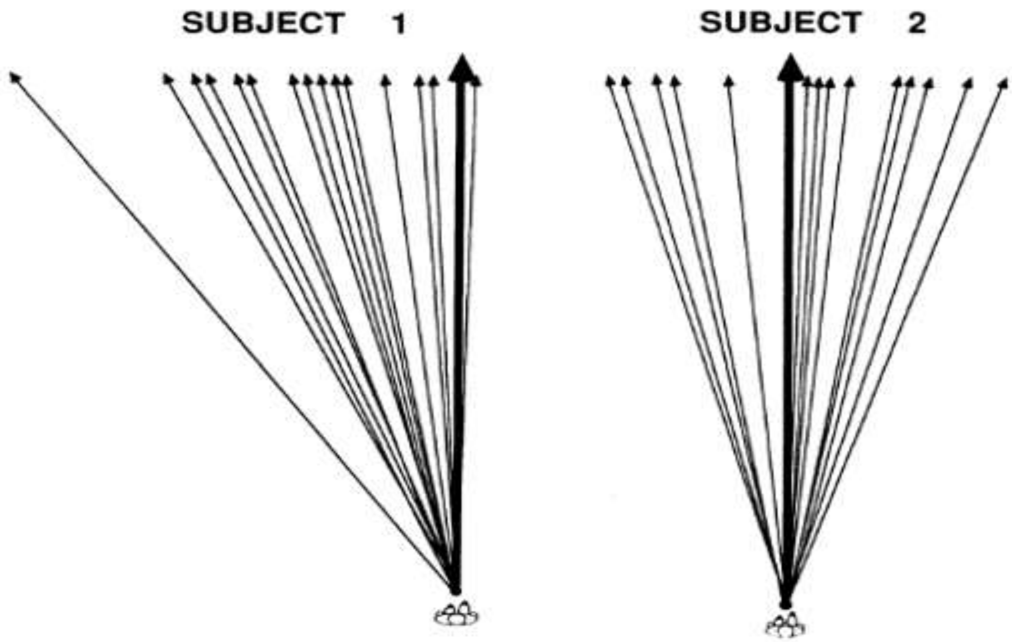

Figure 2.9: Illustration of Error Scores Example. Adapted from (Guth 1990).

*This figure displays two subjects error scores for a dart accuracy experiment. In this case, the subjects display similar absolute error, but subject 1 has greater constant error (towards the left direction)

2.4. EFFECT OF MUSCLE FATIGUE ON UPPER LIMB PROPRIOCEPTION

2.4.1. Altered Body Schema Effect on Joint Position Sense Error

Early psychologist Henry Head described body schema as a postural model for the body that actively modifies the "impressions produced by incoming sensory impulses in such a way the final sensations of position or of locality rise into consciousness charges with a relation to something that has gone before" (Gallagher 1995). Later, Schilder, a colleague of Head's would go on to elaborate that body schema incorporates a mental projection of our body image, its posture, and its immediate environment (Schilder 1935). As a postural model which functions to keep track of limb position, body schema plays an important role in directing motor commands to be fine-tuned to their orientation. Body schema plays an important role in characterizing ones adaptations to task performance based on their orientation, by constructing a pragmatic representation of the spatial properties of the body, including limb length, joint positions, and the shape of the body in space. Research has also suggested that body schema serves an important role for integrating the perception of tools that are being interacted with. Historically, body schema was considered consonant with body image. However as literature has developed and expanded, the relationship between the two terms has been more clearly defined. Body schema encompasses the sub-conscious and unconscious sensory-motor capacities that control movement and posture, where body image involves a person's conscious perception of their physical appearance. Body image does not drive motor behavior, and is more of a psychological representation of ones perceived aesthetic characteristics.

When body perception is functioning correctly, body schema is updated accurately during human movement. As elaborated previously, body schema arises from the conjunction of proprioception and exteroception (Enoka and Duchateau 2008). Together, these respective resources subconsciously track limb positions and body orientation; key contributions to the construction of body schema (Enoka and Duchateau 2008). A key physiological mechanism which helps integrate proprioception into body schema is in the dense array of sensory mechanoreceptors located in the postural neck musculature (Jull, Falla et al. 2007). One of the greatest concentrations of intramuscular mechanoreceptors is found in the spinal erector musculature (Jull, Falla et al. 2007). In addition to playing a fundamental role in maintaining body balance, the CNS references this specialized intramuscular array to provide feedback on body posture affecting distal limb orientation (Strimpakos, Sakellari et al. 2006). However it stands to reason that if this feedback afference is disrupted, it can disorient body schema and impact limb JPS accuracy (Letafatkar, Alizadeh et al. 2009).

In the section on proprioception many of the common influences which disrupt proprioception were listed. Among those listed was the influence of muscle fatigue on proprioceptive afferents, whereby type III/IV feedback gain due to fatigue and fusimotor spindle damage acts to inhibit muscle spindle facility (Taylor, Amann et al. 2016). During fatigue of the CEM, postural load can be transferred from active structures such as muscle to passive structures such as vertebrae and ligamentous tissue in an effort to balance destabilizing physical forces (Letafatkar, Alizadeh et al. 2009). Additionally, chronically fatigued tissue can display a higher expression of type 2 motor efferents which fatigue more rapidly, inducing a vicious cycle (Hyldahl and Hubal 2014). Lastly, tissue in chronically fatigued muscle can adapt to have higher concentrations of adipose interwoven with contractile segments, reducing the concentration of active fusimotor segments (Torres, Vasques et al. 2010). These factors can alter sensory input from the neck when acutely or chronically fatigued, potentially impairing sensory feedback to the CNS for directing limb position sense (Letafatkar, Alizadeh et al. 2009, Barker 2011).

In literature, altered sensory input from the neck due to pain has been observed to have multi-faceted implications on total body proprioceptive disturbances. In the upper limb, Haavik and Murphy (2011) observed significant impairments in elbow JPS in individuals with chronic sub-clinical neck pain, and these same participants saw a significant reduction in impairment following cervical spine manipulation. This suggests that improving the distorted input can restore the correct body schema relationships (Haavik and Murphy 2011). A later study by Baarbe (2016) extended on the findings by Haavik and Murphy, having participants with sub-clinical neck pain instead perform an upper-limb dart throwing task. Consolidating the findings by Haavik and Murphy, Baarbe observed greater elbow joint and forearm motor recruitment variability in participants with sub-clinical neck pain. Neck pain has also been shown to incite changes to cerebellar function and spatial awareness (Baarbé, Murphy et al. 2015). Baarbe (2015) demonstrated that participants with sub-clinical neck pain have significantly higher cerebellar inhibition than healthy controls. Interestingly, this study also demonstrated that following cervical manipulation, cerebellar inhibition was reduced to a level similar to that of participants who did not have sub-clinical neck pain (Baarbé, Yielder et al. 2015). Falla (2004-2005) and colleagues have also discovered that chronic neck pain reduces axioscapular muscle fiber conduction velocity (Falla and Farina 2005) and alters motor recruitment patterns of the cervical flexor (Falla, Jull et al. 2004) and upper limb muscles (Falla, Bilenkij et al. 2004). In the lower limb, indications of altered body schema due to neck fatigue have been observed to impair knee JPS as well as instigating multiple decrements in postural sway (Schieppati, Nardone et al. 2003), balance (Gosselin, Rassoulian et al. 2004), locomotive gait (Schmid and Schieppati 2005), and spatial orientation (Schmid and Schieppati 2005), though implications due to neck pain have yet to be researched.

Though the current model of upper limb disturbances due to CEM fatigue is predicated on the hypothesis of altered body schema to the CNS, one contradictory hypothesis is that upper limb and total body disruptions in performance could also be attributable to central fatigue arising from neck extension fatigue. The central fatigue following a neuromuscular fatiguing task is characterized by increased corticomotoneural inhibition. However a recent comprehensive review of the factors associated with central and peripheral neuromuscular fatigue following maximal and submaximal exercise suggests that fatigue induced either maximally or sub maximally takes an average of 2-3 minutes to CNS recovery (Allen, Lamb et al. 2008). Bortolotto et al. also contributes that in the case of submaximal contractions to fatigue, this phenomena more preferentially effects the motor neurons responsible for the contraction (Bortolotto, Cellini et al. 2000). This means that a submaximal CEM fatiguing protocol is very unlikely to significantly impact shoulder performance via central fatigue. On the chance that it does, any potential effects are mitigated after 2-3 minutes and the remaining decrement in performance would need to be otherwise substantiated.

2.4.2. Research at the Elbow Joint

In addition to the findings presented by Haavik and Murphy (2011), there have been other studies directed at quantifying the extent of upper limb proprioceptive deficit correlated with altered sensory input at the neck (Haavik and Murphy 2011). Typically, research directed at proprioceptive correlations in the upper limb focuses on interactions at the elbow as it represents the simplest upper limb joint to quantify. This is due to the comparatively simple DOF of elbow motion characteristic of a hinge joint, whereby the elbow can easily be locked into full supination, isolating the forearm into a plane of flexion and extension about the humerus.

Knox and Hodges examined changes in elbow JPS after manipulating head position (Knox and Hodges 2005). They found that participants performed significantly worse at repositioning their elbow to match target joint positions when their head and neck were in a position of flexion, rotation, or combined flexion/rotation than when their head and neck were in neutral posture. This study mitigated the potential effects of distracting from the target position during head and neck manipulation by moving the head and neck into posture during a break period between joint angle presentation and reproduction. The findings of this study are a novel contribution to the literature because they suggest that reduced proprioceptive performance may be attributable in part to the changes in interpretation of limb position brought about by manipulating neck posture. This plays into the overarching hypothetical framework of altered sensory input disrupting body schema in that other afferently disrupting stimuli such as fatigue, pain, or temperature can potentially cause the neck to feel like it is at a different posture, and thusly have performance decrements similar to those observed in this study.

Zabihhosseinian (2015) made the logical next step in building on the hypothetical framework for altered sensory feedback on body schema by testing proprioceptive error at the elbow both precluding and following a submaximal fatiguing stimulus induced to the neck (Zabihhosseinian, Holmes et al. 2015). This study found a significant increase to absolute joint position error (JPE), while variable and constant error did not change significantly. This study therefore determined that acutely induced neck fatigue significantly disrupts body schema leading to general decrements in JPE, but not necessarily influencing directional bias or variability. This study also complimented the earlier findings by Haavik and Murphy who found a significant interaction of elbow JPE with chronic neck pain.

Work by Baarbé (2015) and colleagues followed up on the isolated elbow JPE findings put forth by Zabihhosseinian and Haavik and Murphy by extending the DOF to a precision upper limb task; dart throwing (Baarbé, Murphy et al. 2015). In this study, participants with recurrent low grade sub-clinical neck pain showed an increase in the total distance of hand trajectory during the throw, as well as increased variability in elbow and forearm motor selection. Peak acceleration velocity of the shoulder and peak deceleration velocity of the wrist was also found to be faster in these participants. The findings here suggest that sensorimotor disturbances attributable to altered neck afference influence total neural control of the upper limb, shoulder and wrist inclusive.

Most recently, work by Reece (2019) extended the scope of upper limb research beyond the elbow to quantify wrist JPE in participants with chronic changes in neck sensory input in those with subclinical neck pain (Reece 2019). Their findings consolidated the previous literature by finding an effect of significantly higher matching error for a dynamic wrist tracking and stabilization task in subclinical neck pain participants than that found in controls. This study confirmed that altered sensory input at the neck affects body schema properties in the isolated wrist, as it has done in the elbow previously.

2.4.3. Research at the Shoulder

The literature reviewed thus far indicates significant gaps in our understanding of whether neck fatigue leads to shoulder proprioceptive disturbances. As of yet, no exact study to test this relationship exists, however there are other sources in literature which suggest that neck fatigue is likely to impact shoulder proprioception.

A 2010 study by Lewis and colleagues (2010) examined the implications on upper limb JPS in individuals with Complex Regional Pain Syndrome (Lewis, Kersten et al. 2010). As the authors state, this syndrome is typified by an intense general regional pain that can flare up chronically and seemingly at random. It has was found that that individuals suffering with this syndrome have a greater difficulty in perceiving their limbs position in space, which is thought to arise from disturbance in body schema. However instead of alterations to body schema being mediated by fatigue, pain is instead the mechanism for disturbing multisensory function. In this study, participants suffering from this syndrome were tasked to recreate arm postures from manipulating arm rotation about the shoulder. This study found a significantly higher JPE in individuals with Complex Regional Pain Syndrome compared to healthy controls, insisting that altered body schema may be as prevalent as disturbing shoulder JPS as it has been in the distal upper limb.

Following up on their significant findings at the elbow joint, Zabihhosseinian and colleagues continued their investigation into the disrupting proprioceptive effects of altered sensory input to the neck induced by fatigue, this time examining scapular and humeral kinematics (Zabihhosseinian, Holmes et al. 2017). Their study involved comparing individuals with subclinical neck pain and healthy controls in an unconstrained humeral elevation task, where participants raised their dominant right arm in the scapular plane to approximately 120 degrees elevation. This task was compared for both groups at baseline and following the induction of 70% submaximal acute neck muscle fatigue. This study discovered that precluding acute neck fatigue, the subclinical neck pain group trended towards more movement initiated at the scapulothoracic joint versus the glenohumeral joint. Following neck fatigue, control participants demonstrated a more abducted ‘scapular plane’ during their elevation task whereas subclinical neck pain participants did not demonstrate this same trend, with their end effector reaching approximately the same point during post fatigue trials. This study concludes by hypothesizing that the display in differential compensatory strategies between groups may suggest that chronic altered sensory input to the neck has resulted in an impaired adaptation to acute fatigue. It is also important to note that this study involves examining the kinematics of an axioscapular muscle task post neck fatigue, therefore while the switch to compensatory mechanics was instigated by local muscle fatigue, the difference in compensatory motor strategies between groups was hypothesized to arise from altered body schema.

Lastly, a most recent study by Zabihhosseinian (2019) continued their focus on properties of shoulder proprioception following altered sensory input to the neck induced by neck fatigue (Zabihhosseinian, Yielder et al. 2019). For this study, the method of quantifying shoulder motion was switched from using an experimental scapulohumeral kinematics framework, to focusing on end effector error for a shoulder tracing task. The task involved rotation of the shoulder to move an on-screen object to a target. This task was done with vision, and repeated in the absence of vision. Following induction of acute neck fatigue, participants ability to conceptualize the target task in the absence of vision was significantly impaired, suggesting incurred deficits in body schema which impacted upper limb performance accuracy and spatial orientation.

Building on progress of these studies, a potential next step in contributions to literature might involve shoulder tasks which isolate the humeral function of the shoulder from the scapular function. This may help to differentiate disruptions to shoulder mechanics and kinematics that are solely a product of altered body schema, and not axioscapular compensations due to neck muscle fatigue. One method by which this may be attempted without overly constraining the shoulder would be for the application of a shoulder rotation task, whereby the scapula will naturally be inclined to contribute very little as it is primarily involved in unlocking the glenohumeral joint during elevation and abduction (Prescher 2000, Yoshizaki, Hamada et al. 2009). However, designing unconstrained tasks and quantifying shoulder mechanics are two notable challenges to such a study, as this has been the primary reason why so little shoulder kinematics literature exists to date compared to literature examining the elbow and wrist (Halder, Itoi et al. 2000, Matsuki, Matsuki et al. 2011, Quental, Folgado et al. 2012).

2.5. KINEMATICS ASSESSMENT OF THE UPPER LIMB

2.5.1. History of Upper Limb Motion Analysis

The quantification of upper limb kinematics, particularly with regards to the shoulder complex, is complicated by the appreciable DOF's at the shoulder complex compared to other articulations in the body, such as the hip (Halder, Itoi et al. 2000). In an attempt to standardize the description of upper limb kinematics, the Standardization and Terminology Committee of the International Society of Biomechanics (ISB) set out to disseminate a communication to propose regulations for the definitions of upper limb motion as they had done previously with the lower limb (Wu, Siegler et al. 2002).

2.5.2. International Society of Biomechanics Standards for Describing Joint Movement

The ISB set out to formalize the definitions and reporting standards for human kinematics through their dual publications entitled "ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion" parts I and II (Wu, Siegler et al. 2002, Wu, van der Helm et al. 2005) as well as a precursory general report (Wu and Cavanagh 1995). The later of these instalments (part II) is focused on the joints of the upper limb: the shoulder, elbow, and radiocarpal joint (hereafter simplified as the wrist) (Wu, van der Helm et al. 2005). This publication breaks down the standardization and terminology of each articulation across three subsections: terminology, body segment coordinate recommendations, and joint coordinate system (JCS) and motion definitions. For the purposes of this review, only the subsections relevant to the shoulder will be discussed.

The terminology proposed to describe anatomical landmarks about the shoulder and upper limb are as follows in table 2.2 and compliment the illustration in figure 2.11.

Table 2.2: Landmarks for Kinematics Rigid Bodies (Wu, van der Helm et al. 2005)

Bone/Segment of Reference	Landmark	Definition
Thorax	C7:	Processus Spinosus (spinous process) of the 7th cervical vertebra
	T8:	Processus Spinosus (spinal process) of the 8th thoracic vertebra
	IJ:	Deepest point of Incisura Jugularis (suprasternal notch)
	PX:	Processus Xiphoideus (xiphoid process), most caudal point on the sternum
Clavicle	SC:	Most ventral point on the sternoclavicular joint
	AC:	Most dorsal point on the acromioclavicular joint (shared with the scapula)
Scapula	TS:	Trigonum Spinae Scapulae (root of the spine), the midpoint of the triangular surface on the medial border of the scapula in line with the scapular spine
	AI:	Angulus Inferior (inferior angle), most caudal point of the scapula
	AA:	Angulus Acromialis (acromial angle), most laterodorsal point of the scapula
	PC:	Most ventral point of processus coracoideus
Humerus	GH:	Glenohumeral rotation center, estimated by regression or motion recordings
	EL:	Most caudal point on lateral epicondyle
	EM:	Most caudal point on medial epicondyle
Forearm	RS:	Most caudal–lateral point on the radial styloid
	US:	Most caudal–medial point on the ulnar styloid

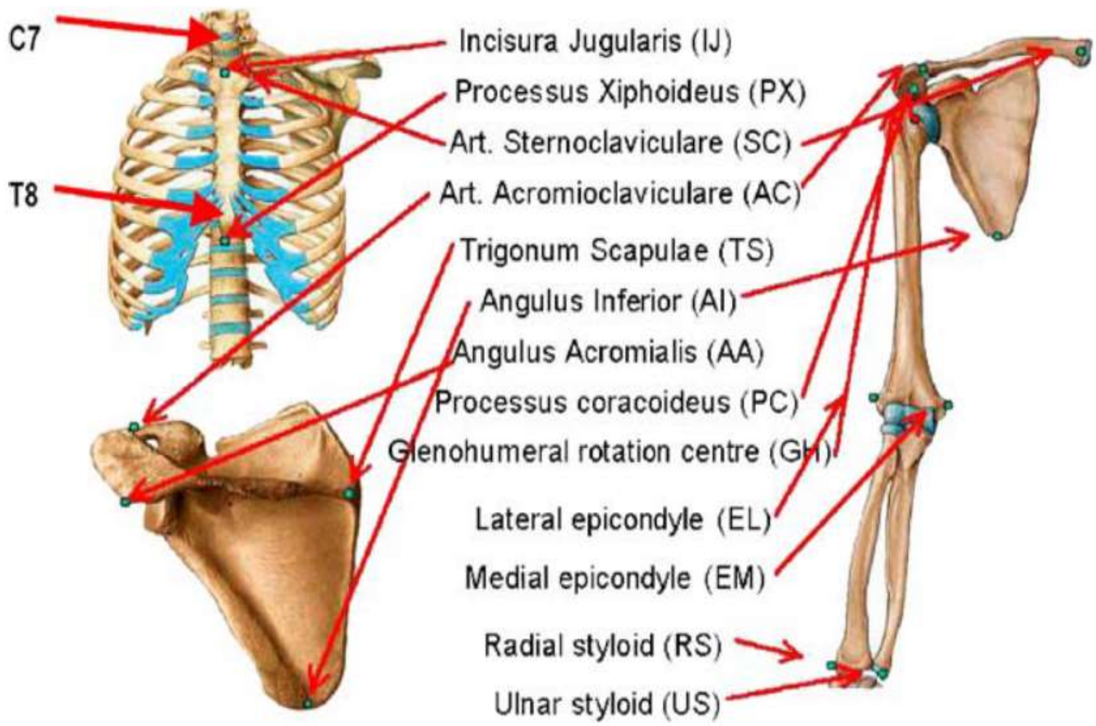

Figure 2.10: Location of Boney Landmarks for Kinematics Rigid Bodies (Wu, van der Helm et al. 2005)

For the purposes of this thesis, thorax, humerus, forearm, and hand coordinate systems are all pertinent. However, the scapula and clavicle were not defined or captured for kinematics analysis, and as such, are not included in this review. It should be noted that whenever left sided limbs are measured with respect to the sagittal plane, raw positions are mirrored (eg. Z = -Z) thereby all definitions for right sided limbs can be applicable. The coordinate systems referenced for the thorax, humerus, and forearm of the right upper limb are as follows:

<u>Thorax coordinate system:</u> Where origin is set at the point IJ

Y axis: the line connecting the midpoints of the perpendicular axes between PX:T8 and IJ:C7, where positive points upward.

Z axis: The line perpendicular to the Y axis set at the origin, where positive is denoted in the right direction.

X axis: The common 3rd dimensional line perpendicular to the Z and Y axes set at the origin and positive set to forward.

Humerus coordinate system: where the origin is set at the point GH.

Y axis: The line connecting GH to the midpoint of EL and EM where positive is proximal towards GH.

Z axis: The line perpendicular to the Y axis and set at the origin, where right is denoted positive.

X axis: The common 3rd dimensional line perpendicular to the Z and Y axes set at the origin, where forward is denoted positive.

- Note that due to suggestions by Wu et al. the second option for humerus JCS was selected for this review as the forearm was also included during recording.

Forearm coordinate system: where the origin is set at the point US.

Y axis: The line connecting US and the midpoint between EL and EM, where the proximal is denoted positive.

X axis: The line perpendicular to the Y axis, where forward is denoted positive.

Z axis: The common 3rd dimensional line incident with the X and Y axis at the origin, where right is denoted positive.

JCS and motion of the humerus relative to the thorax followed a Y-X-Y Cardan translation sequence as follows:

X = **plane of humeral elevation** (γ) where adduction is positive (+)

Y = **negative elevation of the humerus** (β) where depression is positive (+)

Z = **axial rotation about the humerus** (γ_2) where internal rotation is positive (+)

Cardan translation sequences (X,Y,Z vs Z,X,Y, vs Y,X,Y) denote the order in which the chosen segments translation is ordered. This is done with respect to the local coordinate system (LCS) of the referential segment in anatomical position. Therefore, an example of a negative X translation of the humerus about the thorax would be elevation of the humerus from anatomical position, as this is denoted negative about the X axis of the referential segment to the humerus; the thorax. As such, axial rotation of the shoulder is recommended as a "Y" translation by Wu et al. (2005): because in anatomical position, axial rotation occurs about the Y axis of the thorax. However, this is also why axial rotation of the shoulder is denoted as a "γ_2" rotation: because the plane of axial rotation of the shoulder is derivative of the preceding "γ" and "β" translations in the Shoulder POE and elevation respectively.

2.5.3. Unique Methods to Quantify the Shoulder in Literature

Being foundational to most, if not all gross upper limb movements, there have been many attempts at designing devices and protocols to aid laboratory quantification of shoulder mechanics. The most common method of shoulder motion analysis is 3-dimensional optical measurement. Optical motion capture represents the gold standard in kinematics assessment, although the 3D modelling accuracy of this system is based on an assumption that markers attached to the skin represent 3D motion of the underlying skeletal structure. This can potentially present an issue when attempting to directly model the scapula, for which its unique joint mechanics allow it to glide under the skin and have very few reliable bony prominences that will not shift (Karduna, McClure et al. 2001). Additionally, due to the broad DOF and range of variability in preferred movement patterns, kinematic markers can easily lose contact with infrared cameras behind limb segments, disrupting data sets. There have been many attempts to set a standard practice for quantifying the scapula via kinematics such as Karduna and colleagues (2001) who found a reliable correlation between an invasive technique and their novel non-invasive approach (Karduna, McClure et al. 2001). Bourne et al. (2011) attempted to refine Karduna's methodology in their 2011 paper and found relative success in achieving significant improvements, however they stipulated that the most accurate kinematics results come from marker placements specifically designed for a single plane of movement (Bourne, Choo et al. 2011).

2.5.4. Non-Optical Tracking Methods for Measuring Joint Position

While kinematics represents the gold standard method for quantifying human motion, there have also been several attempts to quantify shoulder motion using non-optical methodologies (Dickerson, Chaffin et al. 2007, Quental, Folgado et al. 2012). Lee and colleagues (2003) attempted to utilize a custom built motor-driven passive shoulder rotation measurement device which mobilized participants passive shoulder joint through comfortable internal and external rotation while the participant attempted to estimate their shoulder angles in the absence of vision (Lee, Liau et al. 2003). Mechanisms such as this can be effective for simple shoulder modelling, as by isolating the elevation of the scapula, they can effectively minimize its contribution to shoulder biomechanics and focus solely on the movement of the humerus relative to the thorax (Lee, Liau et al. 2003). However a limitation of such designs, with respect to studying proprioception as an outcome measure, is that their constraint of the upper limb to isolate the humerus also provides many articular surfaces with the skin which may provide additional feedback through cutaneous sensory afference (Voisin, Lamarre et al. 2002, Collins, Refshauge et al. 2005, Proske 2015). Due to this limitation there have also been attempts in literature to explicitly constrain the shoulder as minimally as possible while still providing accurate movement analysis. One such novel apparatus that has been used in a series of studies at the University of Canberra (Waddington and Adams 1999, Waddington, Seward et al. 2000, Naughton, Adams et al. 2005, Han, Waddington et al. 2011, Han, Anson et al. 2013) involves queuing participants through their active range of motion without visual occlusion. This methodology utilizes the laboratories novel AMEDA to quantify relative active joint angles. However, tasks such as these may be prone to the inherent limitation of variability in individual’s preferred movement pattern when they are not constrained to a single plane of motion. This variation can make it more challenging to make statistically significant comparisons between individuals.

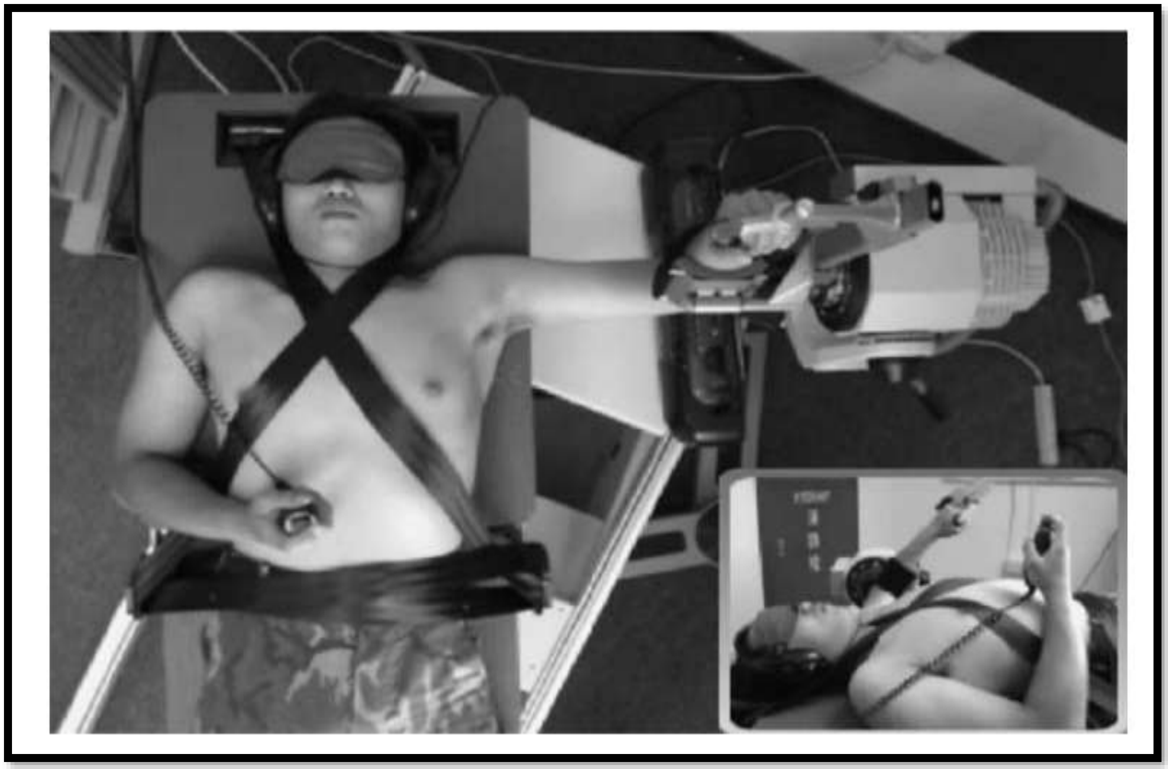

Figure 2.11: Active Repositioning Using the Novel Proprioception Testing Device Design by Lee (1998) and Lee et al. (2003) – Adapted from (Lee, Liau et al. 2003)

Figure 2.12: Passive Repositioning Using the Novel Proprioception Testing Device Design by Lee (1998) and Lee et al. (2003) – Adapted from (Lee, Liau et al. 2003)

2.5.5. A Novel Shoulder Joint Position Sense Measurement Device

Based upon the many diverse previous approaches to quantifying shoulder motion found in literature, the Neuromechanics group at Ontario Tech University partnered with the Faculty of Engineering to design and develop a novel Shoulder JPS Measurement Device. In researching the many previous attempts to quantify shoulder motion found in the literature, the device constructed by Ontario Tech took design concepts from the mechanized iteration proposed by Lee and colleagues (2003). However, our lab also wanted to address the issue of excessive cutaneous feedback that may confound the reliability of their approach. Therefore, the device design also tried to incorporate concepts from the AMEDA protocol, namely keeping the limb relatively weight bearing and minimizing constraint. This harmonized approach led to a design where participants interacting with the Shoulder JPS Measurement Device were implicitly locked into a posture of 90° humeral elevation in a neutral abduction about their plane of elevation, such that participants could only explicitly perform humeral rotation.

2.6. RELEVANT ANATOMY

2.6.1. Anatomy of the Neck

The key anatomical landmarks of the neck include the anteriorly protruding trachea and its cartilaginous rings, the 7 posterior cervical vertebral bodies perceptible by their distinct spinous processes and the supporting paraspinal musculature primarily along the lateral and posterior surface. The anterior surface of the neck is anatomically subdivided by the different layers of cartilage that vitally protect the Larynx and Trachea. These various layers of cartilaginous overlaps terminate superiorly at the Hyoid bone which lies immediately superior to the Larynx.

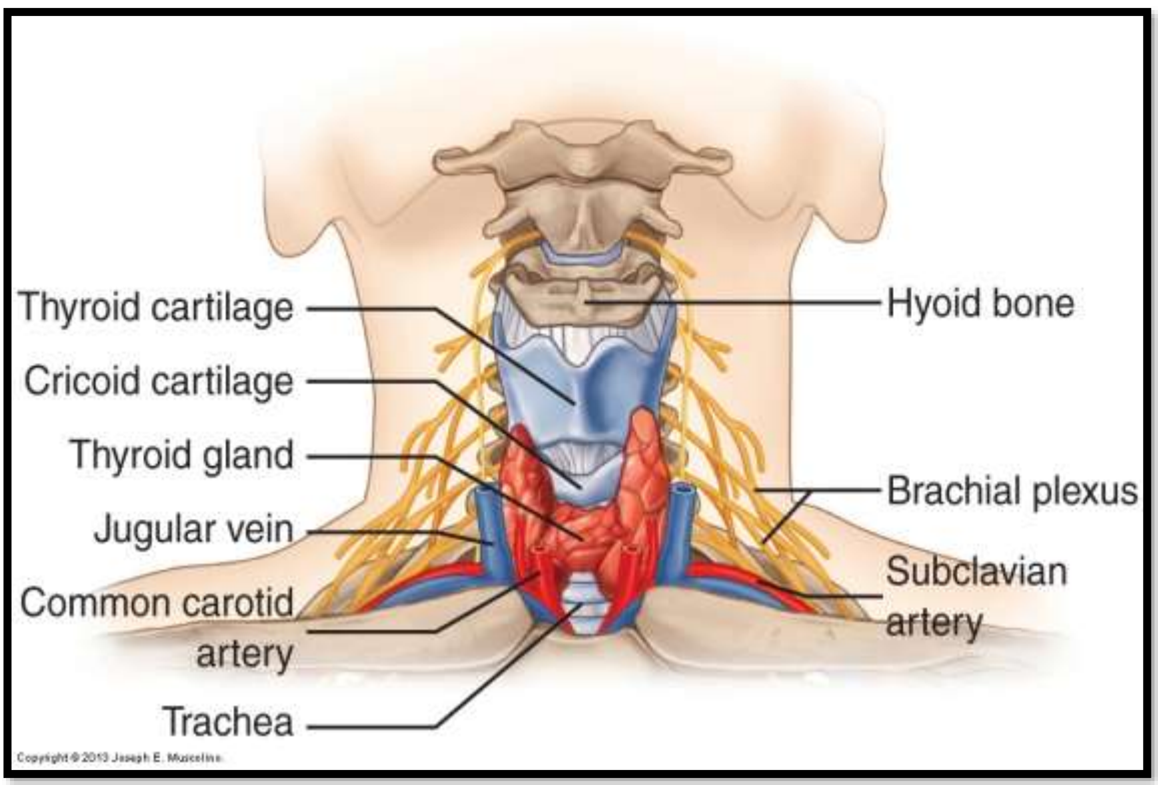

Figure 2.13: Anatomical Structures and Tissues of the Anterior Aspect of the Neck. Adapted from: *Structures of the anterior neck*, by J. E. Muscolino. 2017. Retrieved from: https://learnmuscles.com/blog/2017/08/03/manual-therapy-precautions-working-neck/

The most prominent landmark of the lateral aspect of the neck is the Sternocleidomastoid, an oblique band of surface muscle that connects the Sternum and Clavicle to the Mastoid process of the posterior skull. No muscles located on the lateral surface of the neck directly insert on the scapula.

Two groups of neck muscles can be considered most relevant to the protocols contained within this thesis: neck muscles involved in neck extension, and neck muscles which contribute to function of the scapula.

The first group of relevant musculature, muscles which are involved in neck extension, particularly relates to the function of isometric spino-neutral aligned extension of the neck as is outlined by Edmonston and colleagues in their 2008 neck extensor fatigue protocol (Harms-Ringdahl, Ekholm et al. 1991, Ljungquist, Harms-Ringdahl et al. 1999, Edmondston, Wallumrød et al. 2008). The authors of this publication and its preceding publications do not comment on the specific muscles being fatigued for the purpose of neck extension, rather they base their assertion of neck fatigue on post test decrements in maximal neck extension force (Alricsson, Harms-Ringdahl et al. 2001). However, based on the Biering-Sørensen lumbar extension fatigue protocol, their fatigue protocol was highly reliable for inducing fatigue of the neck extensors (Biering-Sørensen 1984). Contributing to neck extension is a multilayered synergy of three muscle stratums. The deepest of these layers constitutes the cervical transversospinalis group or deep cervical extensors. The deep cervical extensors include the Semispinalis cervicis and the Multifidus. Both muscles directly connect superior vertebrae to inferior vertebrae and act to reduce kyphosis of the spine (Beer, Treleaven et al. 2012). Superficial to the deep cervical extensor group is the Splenius capitus and Splenius cervicis, which when engaged bilaterally contribute to extension: Splenius capitus extends the upper cervical segments and skull, and Splenius cervicis extends the lower cervical segments in relation to the thorax (Cleland, Childs et al. 2005). The final and most superficial layer of neck extensors are truly muscles of the shoulder girdle which originate on the cervical vertebrae. These muscles include the Levator scapulae and Trapezius descendens. Additionally, Rhomdoideous minor may be implicated as it originates at the 7th cervical and 1st thoracic vertebrae.

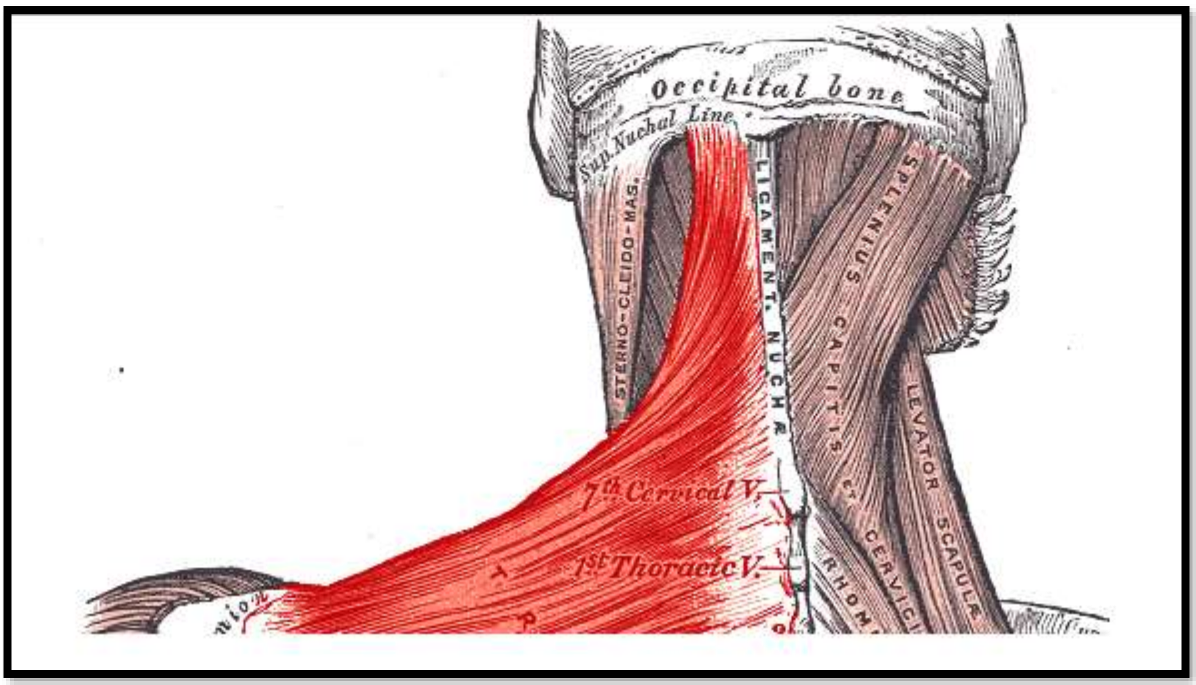

Figure 2.14: Surface and Deep Extensor Muscles of the Posterior Neck. Bright Red Muscle Indicates Superficial Tissue. Adapted from: *Cervical Motor Control Part 1 - Clinical Anatomy of Cervical Spine*, S. Smale. 2016. Retrieved from: https://www.raynersmale.com/blog/2016/7/26/cervical-motor-control-part-1-clinical-anatomy

What neck muscles that are notable to scapular function originate on various cervical and thoracic processes in the posterior vertebral column. There are in fact four muscles which contribute to such axioscapular function and they are Trapezius (descendens, transversus, and ascendens), Levator Scapulae, Rhomboideus minor, and

Rhomboideus major. Of these muscles, three originate at the cervical spine: Trapezius descendens (upper fibers), Levator Scpaulae, and Rhomboideus minor. These muscles are important in guiding axioscapular rhythm in relation to the vertebral column.

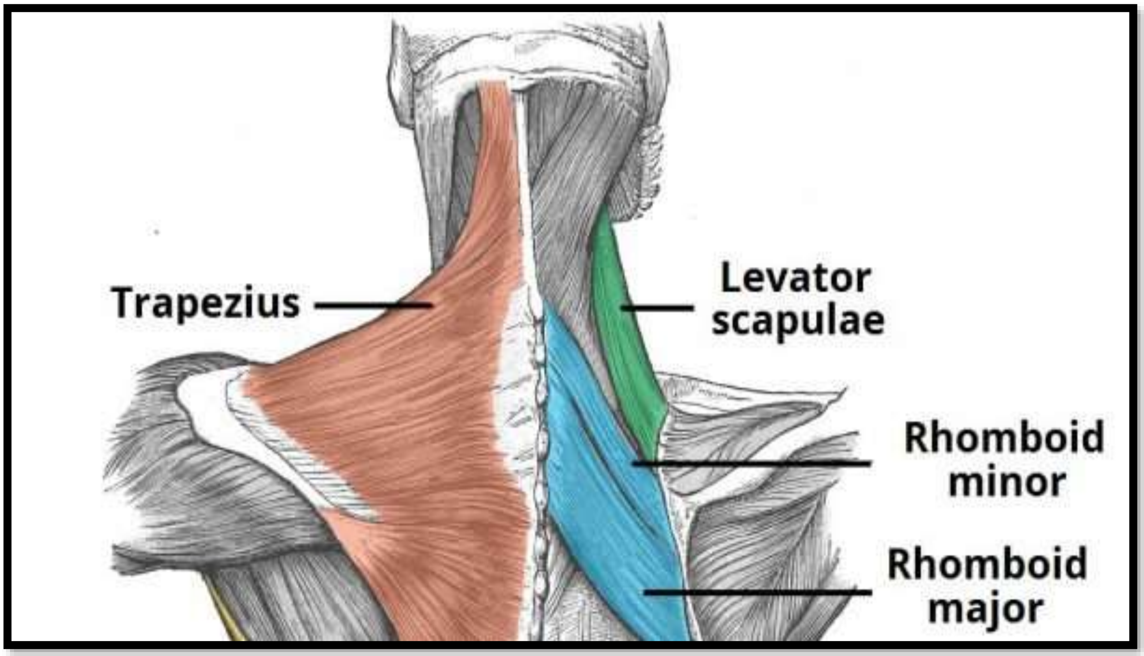

Figure 2.15: Axioscapular Musculature. Adapted from: *The Superficial Back Muscles*, O. Jones. 2019. Retrieved from: https://teachmeanatomy.info/back/muscles/superficial/

<u>2.6.2. Anatomy of the Scapula</u>

The scapula connects the head of the humerus to the clavicle at the acromioclavicular joint, and also connects the humerus to the thorax indirectly via connective tissue and muscular approximations. However the scapula does not form a rigid girdle. In fact as much as one third of the range of motion at the shoulder is contributed from rotation of the scapula along the posterior thorax. The scapula is only viewable from the posterosuperior aspect, notably where the spine of the scapula forms a boney ridge that is palpable bilaterally in the upper region of the thorax. The spine of the scapula runs superior and laterally to form the acromion process which is important in locking the superior aspect of the glenohumeral joint and resisting humeral dislocation. The scapula can perform six general movements: elevation and depression, retraction and protraction, and upward rotation and downward rotation. Of these functions, three involve axioscapular musculature; one movement of which is in each of the planes of movement. These are elevation, retraction, and upward rotation. To this end, axioscapular musculature contributes to each cardinal plane of motion of the scapula, and lends to limit shoulder functions which are directly impacted by muscles originating from the neck.

Additionally, not only does axioscapular musculature contribute to active movement control of the scapula, but they are also crucial in the stability and postural control of the scapula. Namely Trapezius descendens and Levator scapulae are integral, as they work synergistically with Serratus anterior and Trapezius ascendens to guide scapulohumeral rhythm. For the purpose of this thesis, scapulohumeral rhythm may be an important function of Trapezius descendens to consider, namely its contributions to scapular stability during humeral rotation. While it is preferential to minimize active contributions from the scapula during shoulder motion, isometrically abducted posture utilizes the scapular stabilizers to maintain limb posture in this position. However, the main muscular contributions to our task of glenohumeral rotation come from the muscles of the rotator cuff. While this musculature inserts on the humerus to provide moments of internal and external rotation it should be noted that these muscles originate on the scapula, and therefore could potentially be impacted by dyskinesis of the scapular stabilizers. However, this may be desirable to the other planes of shoulder motion which more actively engage axioscapular contributions.

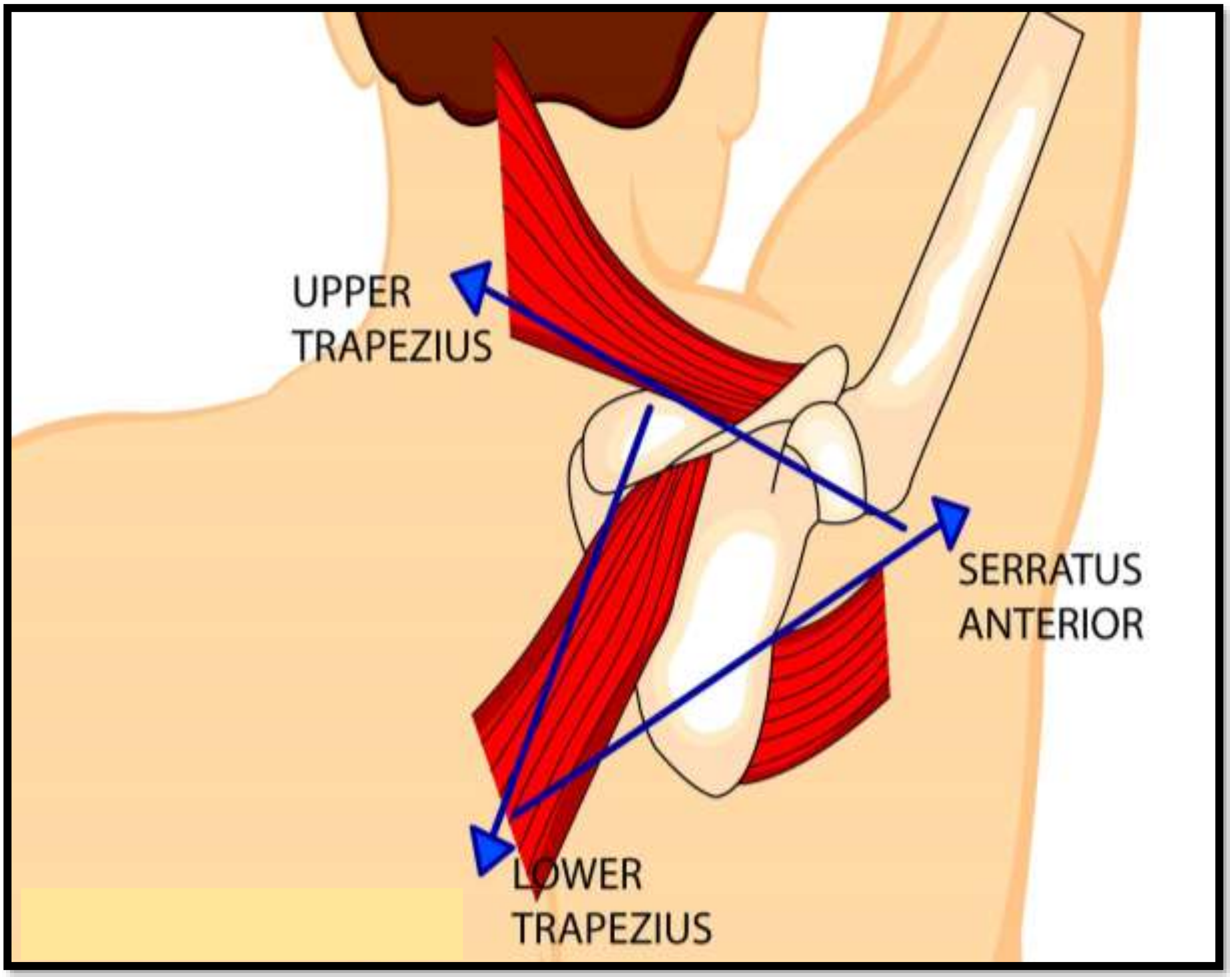

Figure 2.16: Scapular Stabilizers. Adapted from: *How to Prevent Rotator Cuff Injuries Through Corrective Exercise Programming (Part 1)*, D. Cruz. 2018. Retrieved from: https://www.postureanalysis.com/how-to-prevent-rotator-cuff-injuries-through-corrective-exercise-programming-part-1/

2.6.3. Anatomy of the Humerus

The humerus is a long bone which is located in the upper limb and connects the glenohumeral joint at the scapula to the distal elbow. The humerus serves as the attachment sight for various muscles of the trunk acting on the upper limb, including the

Pectorlis muscles, Latissimus Dorsi which contribute to protraction and retraction of the humerus. In addition to this, the muscles of the scapula and clavicle insert on the humerus as well to provide motions of elevation and depression, and internal and external rotation. Elevation of the humerus is guided by the Deltoid muscles and depression of the humerus is guided by the Coracobrachialis and Teres major muscles. Primary motion of internal and external rotation of the humerus is contributed by the rotator cuff complex: four muscles originating at the scapula which insert on the proximal humerus to assist in rotation. These four muscles are the Infraspinatus, Subscapularis, Supraspinatus, and Teres minor. All fours of these muscles are innervated by projections of the C5-C6 nerve branch. The individual tendons of each of the muscles of the rotator cuff blend into a general confluence of articular tissue before inserting on the greater and lesser tubercle of the proximal humerus.

Muscles responsible for humeral elevation and depression (Deltoid and Coracobrachialis), and humeral protraction and retraction (Pectoralis major and Lattisimus Dorsi) do not originate on the scapula, however scapula rhythm is directly adjunct in these motions, in order to unlock the degrees of freedom of the glenohumeral joint for greater range of motion. In contrast, the muscles of the rotator cuff do directly insert on the scapula but motion of the scapula itself is minimized during humeral rotation. For this reason, anticipate that scapular contributions to shoulder function will be minimized during a glenohumeral rotation task, and thus will minimize any impact of axioscapular muscle fatigue on local JPS.

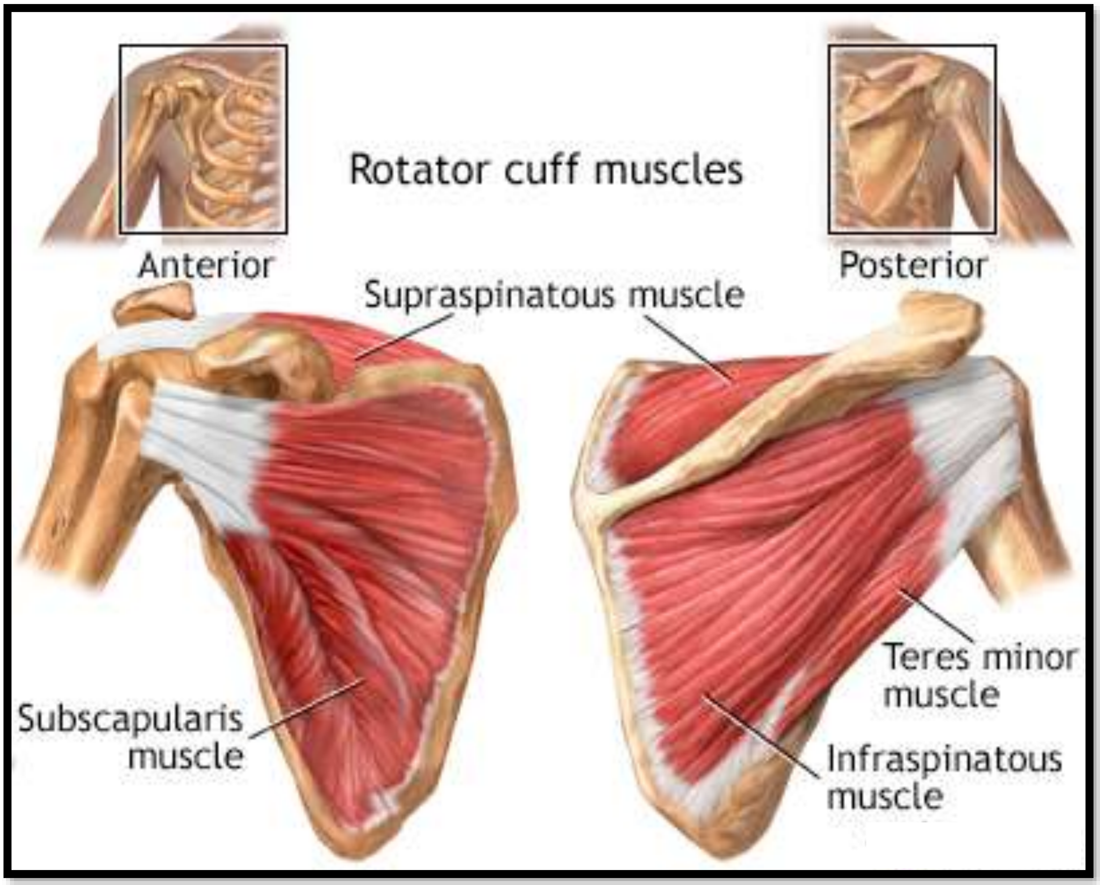

Figure 2.17: Muscles of the Rotator Cuff. Adapted from: (Caceres 2018).

2.7. Conclusion of Literature Review

Fatigue is a neuromuscular process that has been shown to negatively influence proprioceptive accuracy. However, emergent research suggests that body schema is heavily reliant on a sensory projection of cervical posture to determine body orientation. When neck muscles are fatigued, their sensory afference is disrupted, but the implications of neck fatigue on constructing body schema can provoke widespread decrements in limb proprioception, balance and gait, motor learning, and spatial orientation.

While the gold standard, camera-based optical tracking systems can be challenging due to marker loss and challenges in certain camera systems being able to visualize the joint as it moves away from the field of view. Thus, it would be desirable to have a way to reliably measure shoulder JPE. There is evidently a gap in the literature surrounding quantification of the many diverse ranges of motion that the shoulder joint is capable of, and this is evident in research pertaining to altered sensory input to the neck affecting body schema. Findings from research at the elbow and wrist, and limited insights at the shoulder present that altered sensory input to the neck likely impairs shoulder proprioceptive accuracy. However, before the shoulder complex can be quantified, it first requires a reliable tool for analysis. There have been many attempts at devices and protocols aimed at setting the standard for shoulder motion analysis through kinematics, digital human modelling, and unique mechanical interfaces. In an effort to meet the specific needs of assessing shoulder proprioception, our lab has pioneered a new Shoulder JPS Measurement Device, in an attempt to quantify shoulder JPS matching accuracy in healthy populations and in individuals with aberrant shoulder feedback due to acute neck fatigue.

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CHAPTER 3.

STUDY 1: THE EFFECT OF NECK MUSCLE FATIGUE ON SHOULDER HUMERAL ROTATION JOINT POSITION SENSE

3.1. INTRODUCTION

Previous research has suggested that altered sensory input to the neck can lead to generalized decreases in upper limb proprioceptive accuracy (Barden, Balyk et al. 2004, Goble and Brown 2009). Neck muscle fatigue which causes altered sensory feedback has been linked to specific proprioceptive disturbances at the wrist (Reece 2019) and elbow (Knox and Hodges 2005, Haavik and Murphy 2011, Zabihhosseinian, Holmes et al. 2015) as well as mechanical adaptations at the shoulder (Zabihhosseinian, Holmes et al. 2017). A number of common workplace tasks require either sustained neck flexion (keyboard work) or extension (assembly line work) with the potential to lead to neck muscle fatigue (Hagberg and Wegman 1987). Given the large number of workplace and recreational tasks that require accurate shoulder movements with the neck in sustained awkward postures, it is important to understand if neck muscle fatigue also impacts shoulder proprioception. However, the impact of neck fatigue on shoulder proprioception has yet to be investigated. This may be partly because the shoulder joint has appreciable DOF, making it very difficult to constrain kinematic performance during movement tasks, so that change in proprioceptive acuity can be accurately measured. Given the shoulder's role in the majority of upper limb tasks, it is critical to further investigate the implications of altered sensory input to the neck on shoulder proprioception.

Position sense refers to the awareness of the location of limbs and body segments in three-dimensional space, and it is essential for movement and proper postural control (Strimpakos, Sakellari et al. 2006, Ribeiro and Oliveira 2011). There are two mechanisms in the human CNS which contribute to position sense: exteroception and proprioception (Proske 2015). Exteroception involves the coordination of our non-tactile senses such as vision and hearing to integrate a schematic of the body's extrinsic position within its environment (Stillman 2002, Proske 2015). Proprioception involves the interpolation of tactile sensations such as joint angle perceptions and muscle tension, in order to consolidate an intrinsic schematic of our body segment positions relative to each other (Riemann and Lephart 2002).

When performing limb movement, our senses of exteroception and proprioception are continually referenced in tandem, which allows the brain to match kinesthetic and visual afference to predict future limb position (Proske and Gandevia 2009). This multisensory representation of our bodies assumed position in its environment is called our body schema (Head and Holmes 1911, Holmes and Spence 2004). However, if one of these mechanisms of perception becomes impaired, it can compromise the accuracy of our body schema (Granit 1972, Lephart, Warner et al. 1994, Collins and Prochazka 1996, Jerosch and Prymka 1996, Haggard, Newman et al. 2000, Lee, Liau et al. 2003, Falla, Bilenkij et al. 2004, Halseth, McChesney et al. 2004, Knox and Hodges 2005, Goble and Brown 2010, Haavik and Murphy 2011, Emery and Cote 2012, Baarbe 2015, Zabihhosseinian, Holmes et al. 2015, Han, Waddington et al. 2016, Zabihhosseinian, Holmes et al. 2017, Zabihhosseinian, Yielder et al. 2019).

The provision of exteroceptive input in the consolidation of body schema is heavily dependent on one's attentional foci and gaze (Holmes and Spence 2004). By contrast, the proprioceptive feedback from muscle, joint, and cutaneous afferents is essentially constant. This means that exteroceptive input to body schema can often perform a supplemental role by providing additional information towards refining movements within the visual scope (Proske 2015). A body of work indicates that proprioception is heavily relied on to construct the brain's representation of body schema, especially in the absence of vision (Jerosch and Prymka 1996, Goble and Brown 2008, Proske and Gandevia 2009). For this reason, perturbations which impact the accuracy of proprioception have the potential to compromise the accuracy of our sense of body schema.

One of the densest arrays of intramuscular mechanoreceptors found throughout the entire body is located in the posterior supporting musculature of the cervical spine (McLain and Raiszadeh 1995). It is largely hypothesized that this is a purposeful design so that the mechanoreceptors of the neck can contribute a constant and highly accurate sensation of cervical posture to the brain. This contributes to awareness of the body's orientation relative to the head and forms the foundation of our body schema (Holmes and Spence 2004, Knox and Hodges 2005, Baarbe 2015).

The sensations which comprise proprioception can be divided into 3 sub references: tensile sense (sense of resistance), sense of movement, and joint position sense (JPS) (Ribeiro and Oliveira 2011). Sense of movement is generated by the afferent feedback from muscle spindles which serve to signal changes in muscle length (Proske 2015), while GTOs signal change in muscle tension (sense of resistance) (Jami 1992).

Sense of movement is generated by Ruffini and Ruffini-like receptors in the joint capsules and skin which detect changes in motion, direction, velocity, and acceleration. JPS integrates the sensations of tension and movement to determine joint position under resistance, whether it be due to the weight of the arm or an additional load, as these stimuli provoke responses from our tension-specific sensory nerve endings (Riemann and Lephart 2002). Since JPS is established on the relationship between sense of tension and sense of movement, if one of these sensations is manipulated and the relationship between tension and movement becomes unbalanced, it can lead to errors in JPS judgement (Riemann and Lephart 2002). One condition under which the tension-movement relationship can be disturbed to cause disruptions in JPS is when muscle tissues are locally fatigued.

When muscle fibers are continually recruited to the point of endemic local failure, exercise induced muscle fatigue, metabolite buildup, and DOMS collectively occur (Ribeiro and Oliveira 2011). The limit to which one can compulsively elicit muscular fatigue is known as volitional fatigue. It is the point at which movements can no longer be optimally performed (Enoka and Duchateau 2008, Emery and Cote 2012). The metabolic byproducts that accumulate during muscle fatigue are purposeful in pursuing adaptations to neuromuscular potential, however in the incubation time until recovery, said tissue is damaged and will underperform. This transient decrease in the capacity to perform physical actions affects muscle spindle firing frequency (Ashton-Miller, Wojtys et al. 2001) and decreases joint accuracy (Torres, Vasques et al. 2010). At the nervous level, fatigue occurs due to the many neurons in a motor pathway slowing their release of neurotransmitters, reducing LMN excitation and central drive (Enoka and Duchateau

2008). Therefore, it stands to reason that one mechanism by which the tension-movement relationship of cervical sensory afference can be manipulated is through the induction of muscular fatigue.

Therefore, the purpose of this study was to examine the differences in shoulder joint proprioceptive accuracy between a group of participants with acute neck fatigue and controls. In order to streamline the parameters of a shoulder JPS task, a novel device was utilized to lock the shoulder into a plane of internal and external rotation, an action which is commonly used to perform a number of workplace and re-creational tasks.

3.2. METHODS

3.2.1. Subjects

Fifty participants were recruited from the local student population at Ontario Tech University. Thirty participants were selected for the control group (15 male, 15 female) (23.0± 3.6 years), and the remaining twenty participants were selected for the experimental group (10 male, 10 female) (21.8 ± 2.8 years). All participants were right hand dominant and free of neck and shoulder pain for the last 6 months. Participants were excluded if they reported being involved in an occupation which required exertion of the neck or upper arm such as heavy machinery operation or carpentry. Participants who disclosed that they had undergone shoulder or spine surgery were also excluded. Upon arriving to the lab, participants were asked to read and sign the informed consent forms. Inclusion criteria were verified using the Edinburgh Handedness Inventory (EDH), Neck Disability Index (NDI) and Chronic Pain Grade Scale (CPGS) to determine handedness, neck pain intensity, and neck pain effects, respectively. All portions of this study were approved by the University’s research ethics board.

Table 3.1: Participant Anthropometrics – Fatigue Group

Gender	Age	Stature (cm)	Mass (kg)
Male	21.4 (SD+/- 2.7)	175.6 (SD+/- 6.4)	76.7 (SD+/- 9.0)
Female	23.0 (SD+/- 2.4)	166.6 (SD+/- 8.3)	56.3 (SD+/- 3.2)

Table 3.2: Participant Anthropometrics – Control Group

Gender	Age	Stature (cm)	Mass (kg)
Male	23.5 (SD± 3.4)	179.3 (SD± 6.4)	79.5 (SD± 11.8)
Female	22.3 (SD± 3.5)	164.3 (SD± 9.7)	60.4 (SD± 8.8)

3.2.2. Neck Extension Dynamometry

Throughout the fatigue group protocol, maximum isometric neck extension forces were measured using a Series 5i Force Transducer (Mark 10, New York, USA). The force transducer was affixed the floor and oriented perpendicular to, and facing the participants who were lying prone and raised on a massage table. A bracing strap designed to secure cervical loads to the head during neck extension tasks was used to comfortably connect a tensile cable between the force transducer and the participant. An illustration of the experimental set up during neck extensor MVC's as such described can be seen in figure 3.1.

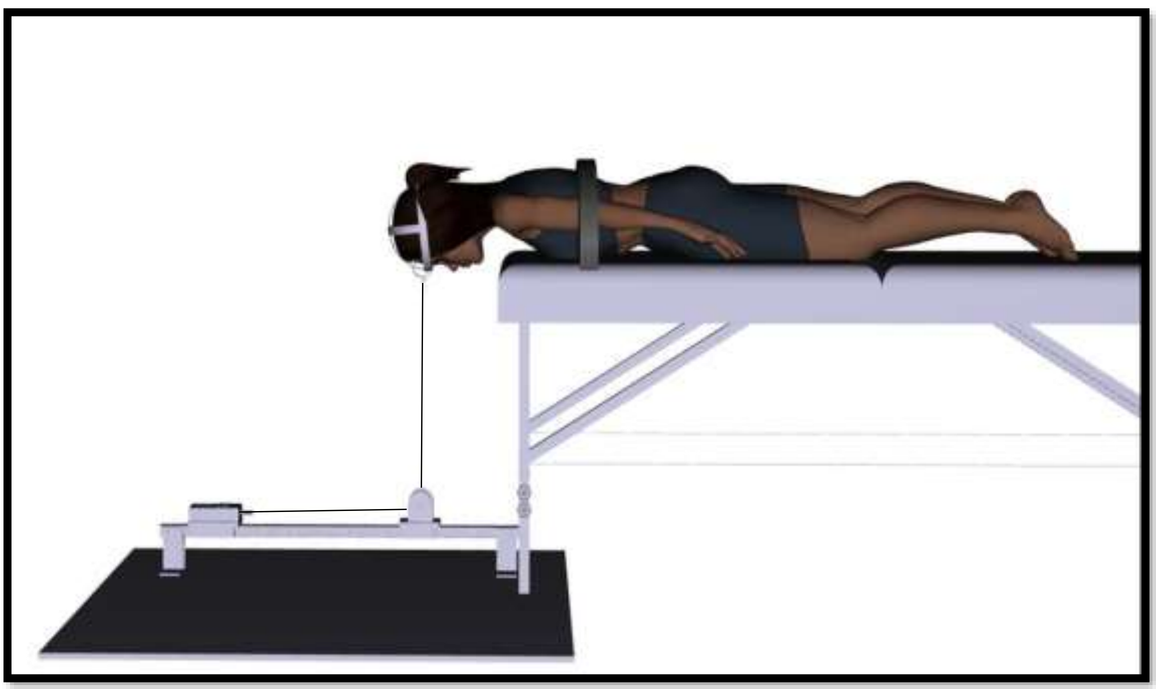

Figure 3.1: Illustration of Isometric Neck Extension MVC Setup

3.2.3. Shoulder Joint Position Sense Measurement Device

A novel Shoulder JPS Measurement Device was designed for this evaluation. The purpose of this device was to measure the rotation of its mechanical arm about the central axis, with the intention of isolating shoulder humeral rotation. The length of the handle on this arm was adjusted to match each participant's forearm length, such that rotation of the machines central axis allowed for humeral rotation when the users forearm was aligned parallel to the machine's rotating arm. The device's adjustable handle also has a button on its distal end, which when pressed, will record the rotation of the arm about the central axis. The device's E6B2-C incremental rotary encoder (OMRON Corporation, Kyoto, Japan) output the axial rotation and angular velocity of rotation at 1000 Hz.

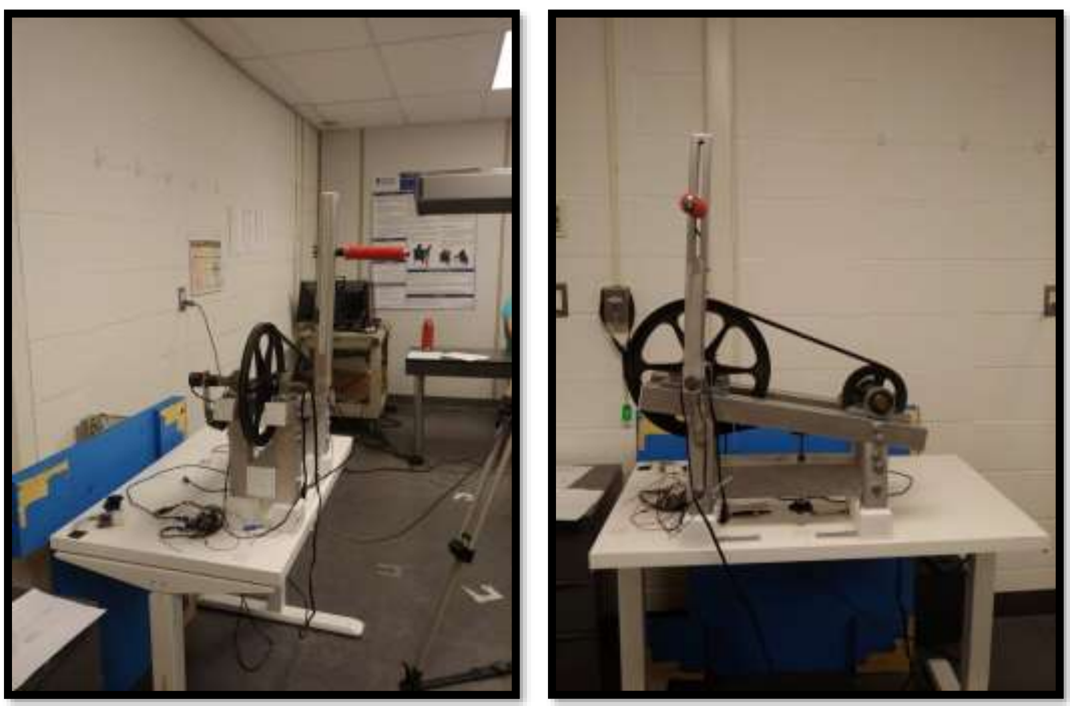

Figure 3.2: Images of Shoulder Joint Position Sense Measurement Device

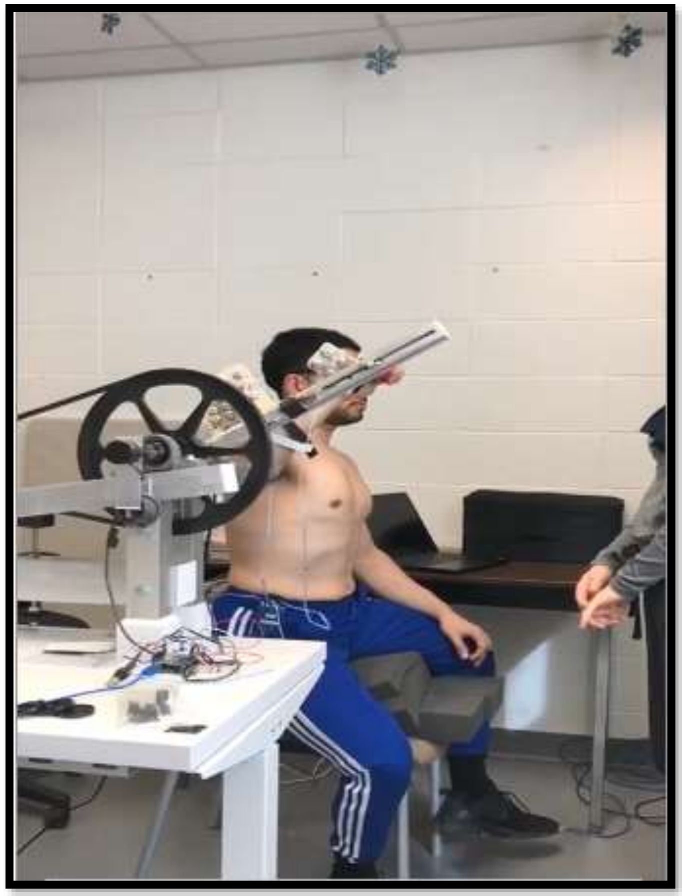

Figure 3.3: Participant Interaction with the Shoulder Joint Position Sense Measurement Device

3.2.4. Task Description

3.2.4.1. Neck Extension MVC Task

Measures of isometric neck extension MVC were used to determine a decline in neck extensor strength following a neck fatigue intervention. Neck extension MVC's were only measured for fatigue group participants. During the pre-fatigue neck extension MVC trials, three 3-second isometric neck extension MVC's were measured from a neutral cervical posture, each separated by 1 minutes rest, of which the strongest was considered as the participants maximum (Edmondston, Wallumrød et al. 2008, Zabihhosseinian, Yielder et al. 2019) . Immediately following the fatigue protocol, two more bouts of 3-second neck extension MVC's were collected to observe the immediate decrement in neck extensor performance. Following completion of the experimental protocol, two last bouts of 3-second neck extension MVC's were again collected to confirm that the neck extensors were still significantly reduced from baseline capacity.

3.2.4.2. Joint Angle Matching Task

The basis of our experimental protocol involved a joint angle matching task for which participants were passively shown shoulder joint positions and then asked to actively recreate them. Before beginning this task, the participant was instructed to hold their right arm in a position of 90° of elbow flexion, 90° external rotation and 90 degrees of abduction. This position was termed the 'Home' position for the purposes of our protocol, and represented 0° of humeral rotation. Once participants became familiar with the Home position, they were blindfolded and instructed to begin the joint angle matching task. This task consisted of 4 sets of 3 joint angle matching trials each. For each trial, the participant began with their arm in the Home position, from which they had their arm passively guided to a new arm posture between 30 and 60 degrees of internal rotation from the Home position. For the purpose of the protocol, this new posture was termed the 'Target' position. The participants arm was supported in the Target position for 5 seconds before being returned to the Home position. The blindfolded participant was then instructed to attempt to recreate the Target shoulder posture to the best of their ability. To do this, the participant would press a button on the handle of the Shoulder JPS Measurement Device, rotate their arm and the device arm to the position they believed best represented the Target shoulder posture, and then press the button on the device handle when they were confident in their re-creation of the Target posture. This process of being passively shown a target shoulder posture and actively attempting to recreate this posture accounted for the completion of one trial. Each set consisted of 3 trials each, and each trial required the participant to approximate a new, randomly generated shoulder position within 30-60 degrees of internal rotation from the Home position. Post-Hoc analysis confirmed a balanced proportion of angles between 30-60 degrees was randomly generated across all trials. Between each set, participants were instructed to stop interacting for the device and rest their arm for 5 minutes; they were also allowed to remove their blindfold at his time and were not privy to any information indicating their proprioceptive accuracy thus far. These 5 minute breaks were intended to mitigate the potential buildup of local muscle fatigue in the muscles performing the joint angle re-creation task, to remove the influence of fatigue on shoulder joint proprioception. For the purposes of this protocol, all passive shoulder joint angle rotations were performed at an approximate speed of 10 degrees/second.

3.2.4.3. Isometric Submaximal Neck Extension Fatiguing Task

Following the completion of their second set of shoulder joint angle matching trials, fatigue group participants were instructed to perform the isometric neck extension fatiguing task (control group participants were given a 5 minute break). To perform this task, participants were asked to wear a head brace designed specifically to support the head, which could also be clipped to a load hanging from the forehead. Participants were then instructed to lay prone on a massage table with their acromion processes aligned to the edge of the table and their neck and head overhanging. In this overhanging prone position, the participants were instructed to maintain a neutral cervical posture as indicated by a level. A standard load of two kilograms (2 kg) was then affixed to the participants head brace, creating a significant neck flexion moment to be counteracted by the neck extensors (Fig. 3.4). The caudal load and table height were set so that the weight would rest on a raised platform when the participant was in approximately 5 degrees of neck flexion. This made an easy setup to quickly get the participant into position for the neck fatigue protocol, and incidentally this also created a simple objective stopping criteria: When the weight touched the platform, the protocol was terminated. The participants were instructed to maintain a spinal-erect posture under load until they broke sagittal neck posture by 5 degrees or verbally indicate volitional fatigue. When either of these termination criteria were met, the neck extension fatiguing protocol was stopped.

Figure 3.4: Isometric Submaximal CEM Fatiguing Task Setup

3.2.4.4. Isometric Neck Extension Maximum Voluntary Contractions (MVCs)

In total, three sets of isometric CEM MVC's were conducted throughout the fatigue group protocol. These sets were, in order: (1st) three baseline CEM MVC's, (2nd) two post-fatigue CEM MVC's, and (3rd) 2 recovery CEM MVC's. The CEM MVC trials were laid out according to figure 3.5, seen below.

To perform isometric CEM MVC's, participants lied prone on the massage table, with the load brace strap secured to their head – identically to fatiguing protocol. However, for the MVC task the load bracing strap was instead affixed to a perpendicular cable secured to a force transducer. For the CEM MVC protocol, participants were instructed to maximally extend their neck for 3 seconds in an effort to maximally activate the CEM and not the deep posterior transversospinals muscles. The set of baseline CEM MVC's included a third set as this was determined to be necessary for participants to familiarize themselves with the task to give a true maximum contraction (Zabihhosseinian, Yielder et al. 2019).

3.2.5. Experimental Procedure

Participants were first familiarized with the Shoulder JPS Measurement Device by performing practice shoulder joint rotations with their non-dominant arm. The practice session was allowed to continue until the participant demonstrated and gave verbal confirmation that they understood how to properly interact with the Shoulder JPS Measurement Device for the purposes of the experimental protocol. If participants were in the fatigue group they began the experimental protocol by performing three isometric neck extension MVC's – the greatest of which was considered their true CEM MVC. Following completion of their third MVC attempt, fatigue group participants began the shoulder joint angle matching protocol. Control group participants were able to skip this step and immediately begin the joint angle matching task. Participants in both groups were blinded with visual occlusion goggles and tasked to recreate shoulder joint angles immediately after they were shown to them. Shoulder joint angles were presented with 4 sets of 3 joint angle matching trials to random joint angles between 30 and 60 degrees of internal rotation. The first of these 4 sets was considered as a task familiarization set as this has been seen necessary in previous research, and the data for this set was not considered towards the participants true baseline JPS ability (Barden, Balyk et al. 2004). The participant was not informed that their first set was a practice set, to dissuade them from potentially putting less focus into their performance. Following the familiarization set, the second JPS matching set was considered the participants true JPS baseline (Fig. 3.5). Following the CEM fatigue protocol in the fatigue group protocol, sets three and four were considered the participants "post-fatigue" and "5-minutes post-fatigue" sets respectively. Five minute breaks were provided between each set of 3 joint angle matching trials to prevent local muscle fatigue. Between set 2 and set 3 participants were instructed to perform an isometric CEM fatiguing task designed by Edmonston (Edmondston, Wallumrød et al. 2008) during their break period. Following completion of the CEM fatiguing task, participants performed another bout of neck extension MVC's to confirm the presence of neck fatigue following the neck fatiguing task. Control group participants were tasked to take an additional 5 minute break instead of performing the

CEM fatigue protocol. Participants then completed the remainder of the shoulder joint angle matching task which consisted of sets 3 and 4, separated by another 5 minute break. Upon completion of set 4, participants performed one last bout of neck extension MVC's to observe any potential increase in neck performance from their previously fatigued state. Participants were then debriefed, thanked for their time, and allowed to leave.

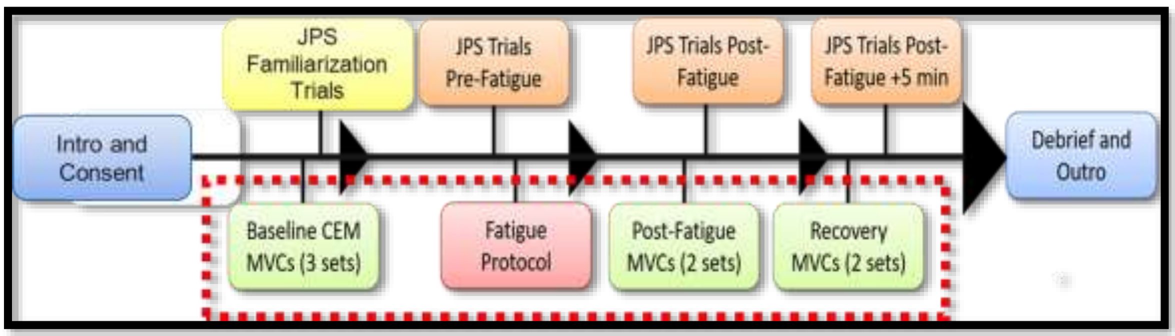

Figure 3.5: Timeline of Experimental Protocol. Red Outline Indicates Fatigue-Group Specific Protocol.

<u>3.2.6. Data Analysis</u>

3.2.6.1. Quantification of Cervical Extensor Muscle Fatigue

To evaluate the fatigue protocol, fatigue group participants neck extensor MVC measures taken after completing the neck fatiguing protocol were compared to their baseline neck extensor MVC's. This was done twice: once immediately following the participants termination of the fatigue protocol, and again immediately after they completed their fourth and final shoulder joint angle matching set.

3.2.6.2. Shoulder Joint Position Error

The accuracy of participant's active joint angle re-creation was measured similarly to Zabihhosseinian (2015) and Knox and Hodges (2005). JPS accuracy was measured by calculating absolute, constant, and variable error to measure differences between the target angle and the produced angle. As presented below, absolute error (Eq. 2.1) measures the deviation between the target angle and the reproduced angle, irrespective of the direction of error. Constant error (Eq. 2.2) measures the deviation between the target angle and the reproduced angle, the difference being that constant error is sensitive to the direction of error. Variable error (Eq. 2.3) measures the consistency of the variability, or standard deviation, between the target and reproduced angles.

Equation 3.1: Absolute Error

$$AE = \sum \left| \frac{[X - X0\ ]}{N} \right|$$

Equation 3.2: Variable Error

$$VE = \sqrt{\left[\sum \left[\frac{[X0 - M^2\ ]}{N} \right] \right]}$$

Equation 3.3: Constant Error

$$CE = \sum \frac{[X - X0\ ]}{N}$$

*Where X represents raw score, X_0 represents the desired criterion score, N represents number of trials, and M represents mean of the values.

3.2.6.3. Joint Angle Re-Creation Error Trend

In addition to our measure of constant error across sets, the trend of central tendency across all joint angle re-creation trials in all three sets as examined as well. The purpose of this analysis was to identify if participants were subject to a greater degree of error at larger target angles, as this has been documented in previous literature (Mountcastle, Poggio et al. 1963, Lephart, Warner et al. 1994).

3.2.7. Statistical Analysis

A repeated measures ANOVA was used to determine significant differences between baseline, post-fatigue, and 5 minute post-fatigue neck extension MVC to assess the level of neck fatigue. Additionally, a 3 (set) x 2 (sex) mixed ANOVA was used to determine sex differences for MVC and a dependent-sample t-test was used to determine sex differences in time-to-fatigue. Changes in Absolute, Constant, and Variable error were each analyzed with a mixed-model 2 (group) x 3 (set) repeated measures ANOVA. Raw error scores for all three measures were all transformed with a square root function to ensure a normally distributed sample for ANOVA.

3.3. RESULTS

3.3.1. Questionnaires

The mean participant score for the fatigue group EHI was 57.5 (SD± 26.5) and the mean participant score for the control group EHI was 67.2 (SD ±26.0). This indicates participants self-reported as being moderate to strong right hand/right arm dominance.

The mean participant baseline score for the fatigue group NDI was 4 (SD± 3.5) and the mean participant baseline score for the control group NDI was (6 ± 6.5), indicating that participants self-reported neck pain disability ranged from non-existent to very mild.

The CPGS reports two measures of chronic pain: characteristic pain intensity and pain related disability. The average participant score for the characteristic pain intensity section on the fatigue group CPGS was 19 (SD± 14.5) and for the control group was 19.5 (SD+/- 14.5), indicating participants self-measured themselves as having the lowest intensity of chronic pain categorized by the CGPS. The average participant score for the pain related disability section on the fatigue group and control group CPGS was 0 (SD± 0) indicating that participants self-reported themselves as having no physical or lifestyle disability due to chronic pain.

3.3.2. Cervical Extensor Muscle Fatigue

3.3.2.1. Neck Extension Maximum Voluntary Contractions

The mean participant neck extension maximum voluntary contraction (MVC) force at baseline was 121.50 ± 38.67 N (Fig. 3.6). Average participant submaximal neck extension endurance time during the fatigue protocol was 8.16 ± 4.45 minutes. There was a significant difference between males and females for force at baseline MVC ($p < 0.01$), immediately post fatigue MVC ($p < 0.05$), and 5 minutes post fatigue MVC ($p < 0.05$) (Fig. 3.7), and there was also a significant difference between males & females for time to fatigue ($P < 0.05$) (Fig. 3.8). The average maximum neck extension force at baseline for males was 144.40 ± 35.62 N and the average maximum neck extension force at baseline for females was 100.82 ± 28.93 N. The mean time-to-fatigue contraction time for males and females was 10.60 ± 5.06 minutes and 5.74 ± 1.76 minutes respectively. A significant ($P<0.001$) 16% drop in maximum voluntary neck extensor force was observed between the Baseline MVC and Post Fatigue MVC trials (Fig. 3.6). This drop in maximum neck extensor capacity persisted to the 5 Minutes Post Fatigue MVC trials as a significant ($P<0.001$) 11% drop from baseline (Fig. 3.6).

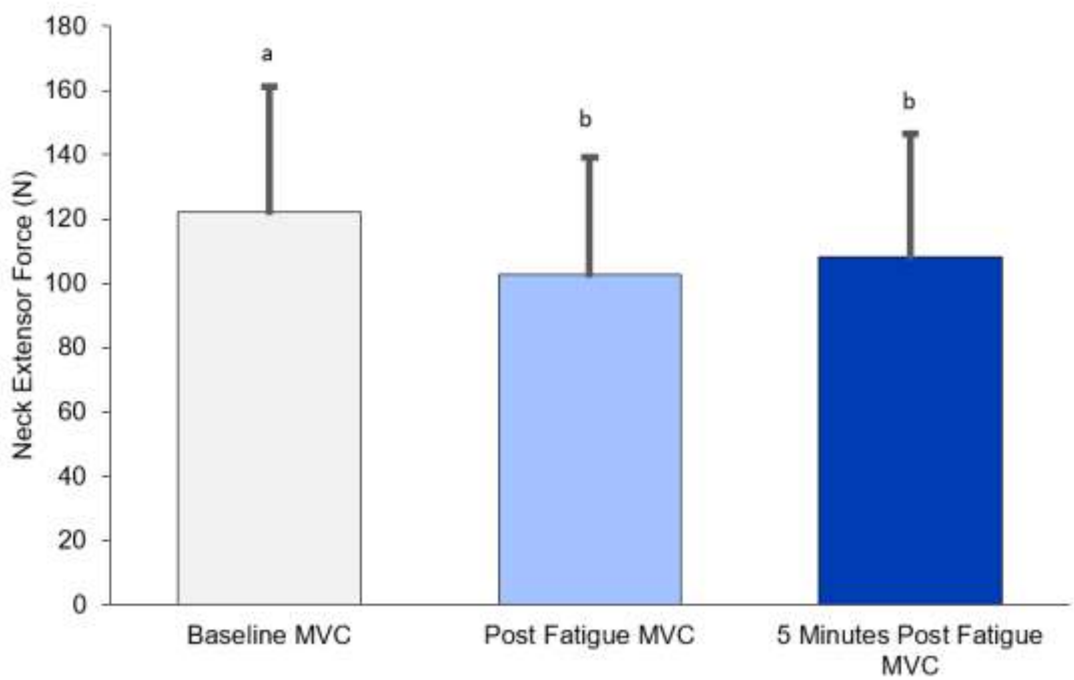

Figure 3.6: Mean Neck Extension MVCs. Bars with differing letters indicate significant difference ($p<0.001$) between means.

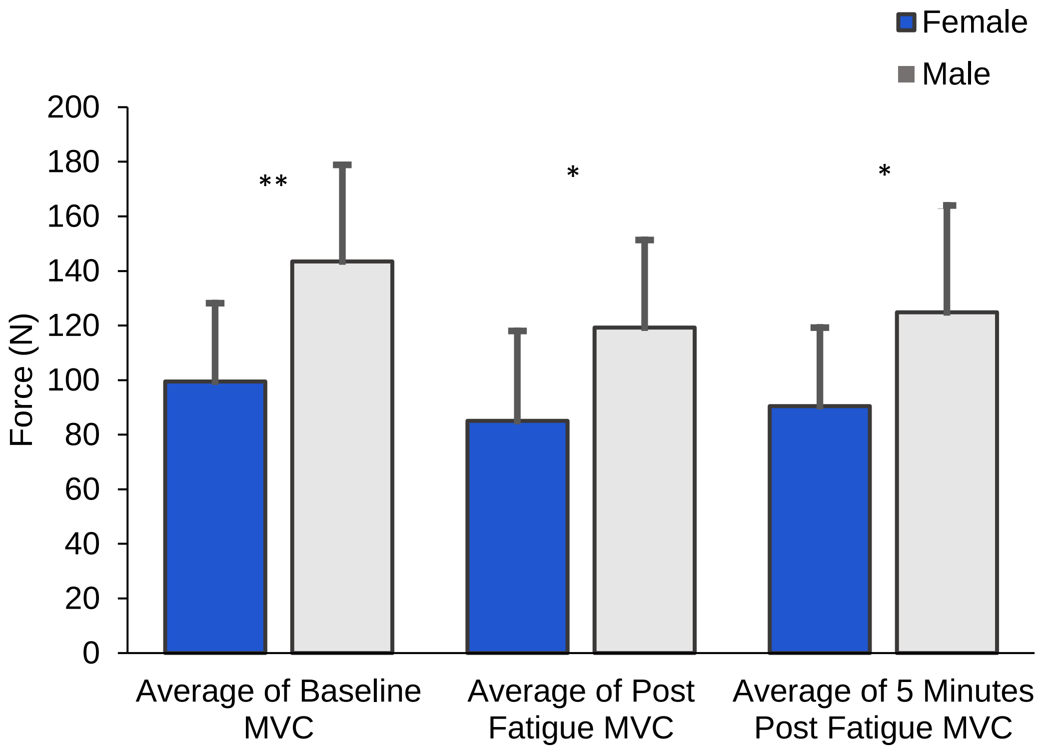

Figure 3.7: Mean Neck Extension MVCs by Sex. Sets with "**" denote p ≤0.01. Sets with "*" denote p ≤0.05.

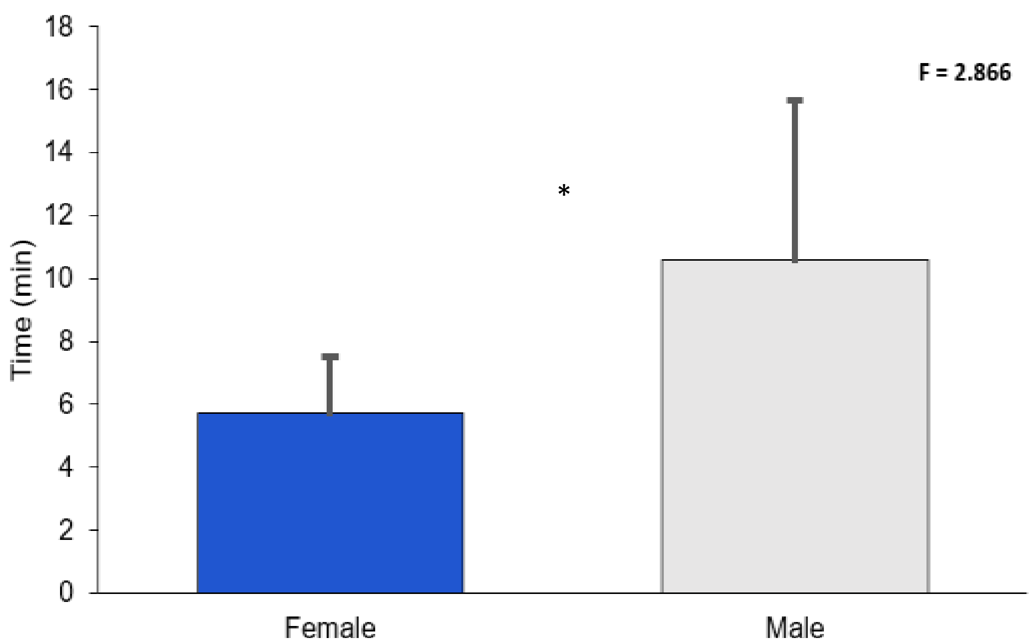

Figure 3.8: Mean Fatigue Protocol Time to Fatigue by Sex. Sets with "*" denote p ≤0.05.

3.3.3. Shoulder Joint Position Sense Error

3.3.3.1. Absolute Error

The average relative absolute error values for control group and fatigue group at baseline were 1.00 ± 0.68 and 1.00 ± 0.81 respectively. For the control group, error held constant for the post control measurement and 5 minutes post control at 1.00 ± 0.70 and 0.99 ± 0.73 respectively ($p \leq 0.672$). The fatigue group showed a non-significant ($p \leq 0.586$) increase in absolute error post fatigue and 5 minutes post fatigue at 1.04 ± 0.88 and 1.21 ± 0.83 respectively. The control and fatigue groups showed almost no difference in task accuracy for joint angle matching immediately following the onset of neck muscle fatigue. Fatigue group error performance increased by 4% from baseline to 5 minutes after the onset of neck muscle fatigue, however this interaction was not significant (Fig. 3.9). Control group joint angle matching error performance did not change by more than 1% between trials.

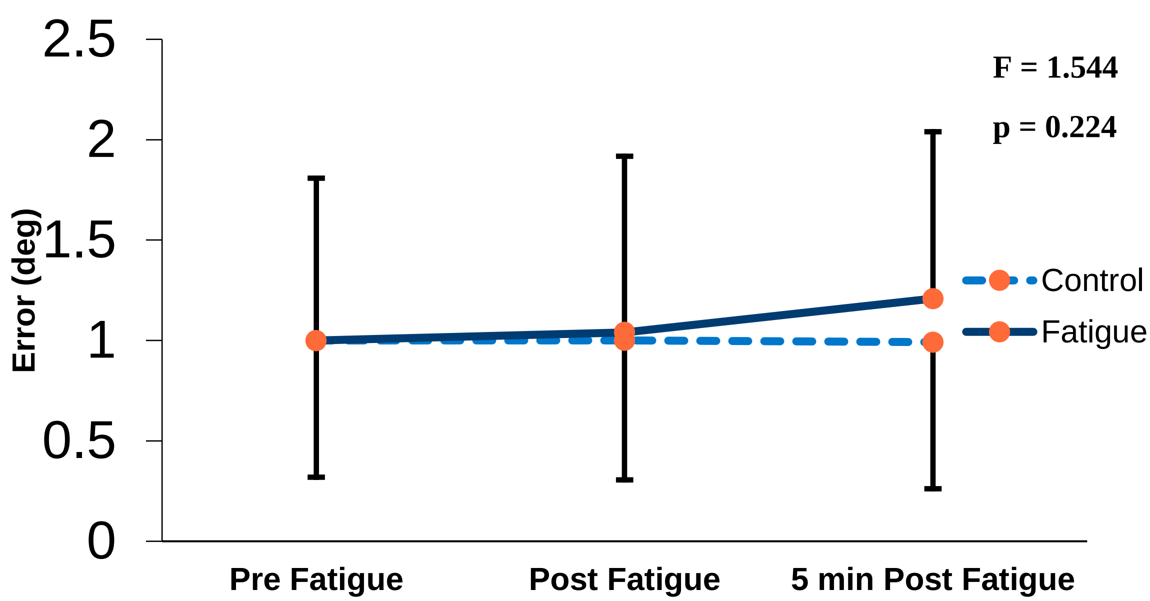

Figure 3.9: Comparison of Relative Shoulder Rotation Joint Angle Matching Task Absolute Error between Control and Neck Fatigue groups.

3.3.3.2. Constant Error

The average relative constant error values for control group and fatigue group at baseline were 1.00 ± 2.11 and 1.00 ± 2.30 respectively. For the control group, error post fatigue and 5 minutes post fatigue at 1.28 ± 1.84 and 1.22 ± 1.83 respectively, but this was not significant ($p \leq 0.706$). The fatigue group also showed an increase in constant error post fatigue and 5 minutes post fatigue at 1.10 ± 2.41 and 1.49 ± 2.34 respectively, however this was not significant ($p \leq 0.664$). The control and fatigue groups showed a non-significant ($p \leq 0.735$) 18% difference in task accuracy for joint angle matching immediately following the onset of neck muscle fatigue. Fatigue group error performance then sharply rises 39%, 27% above the control baseline 5 minutes after the onset of neck muscle fatigue. However, this interaction was not significant ($p \leq 0.877$) (Fig. 3.10).

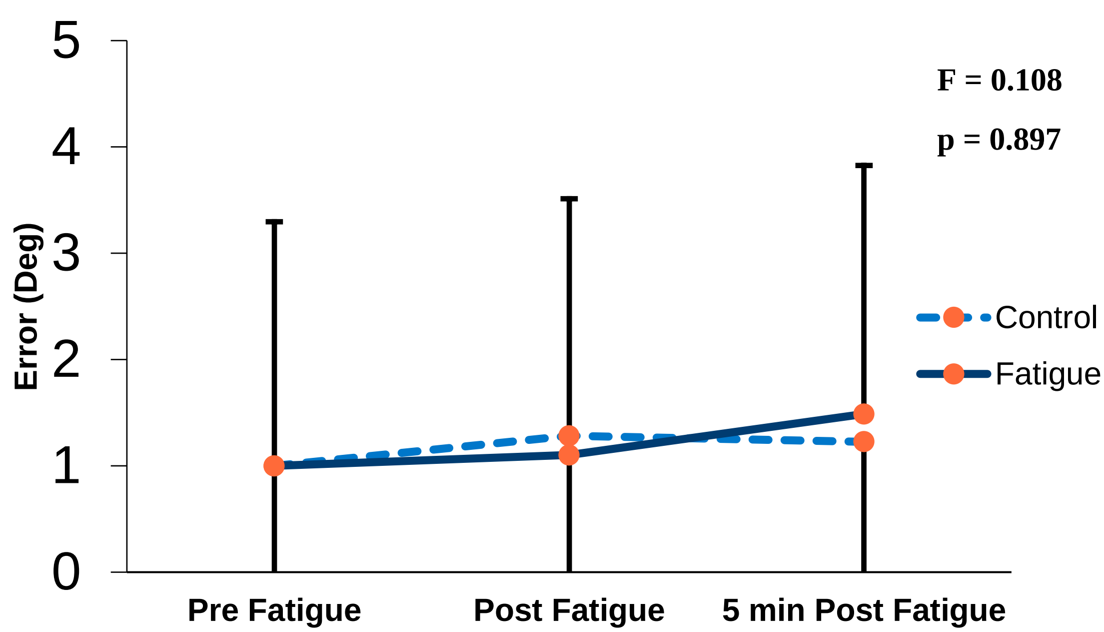

Figure 3.10: Comparison of Relative Shoulder Rotation Joint Angle Matching Task Constant Error between Control and Neck Fatigue Groups.

3.3.3.3. Variable Error

The average relative variable error values for control group and fatigue group at baseline were 1.00 and 1.00 respectively. For the control group, error decreased non-significantly to 0.87 post fatigue and held consistent 5 minutes post fatigue ($p \leq 0.461$). The fatigue group showed a non-significant ($p \leq 0.927$) increase in variable error post fatigue and 5 minutes post fatigue at 1.05 and 1.02 respectively. The control and fatigue groups showed a non-significant ($p \leq 0.50$) difference in task accuracy for joint angle matching immediately following the onset of neck muscle fatigue. This is primarily due to the control group improving their variable error, while the fatigue group's variable error did not change. This trend persisted 5 minutes post neck muscle fatigue as where the variable error for both groups stayed consistent, and the interaction between groups was not significant ($p \leq 0.358$) (Fig. 3.11). Standard deviations were not calculated for variable error, nor presented in figure 3.11, as variable error is effectively the standard deviation of constant error.

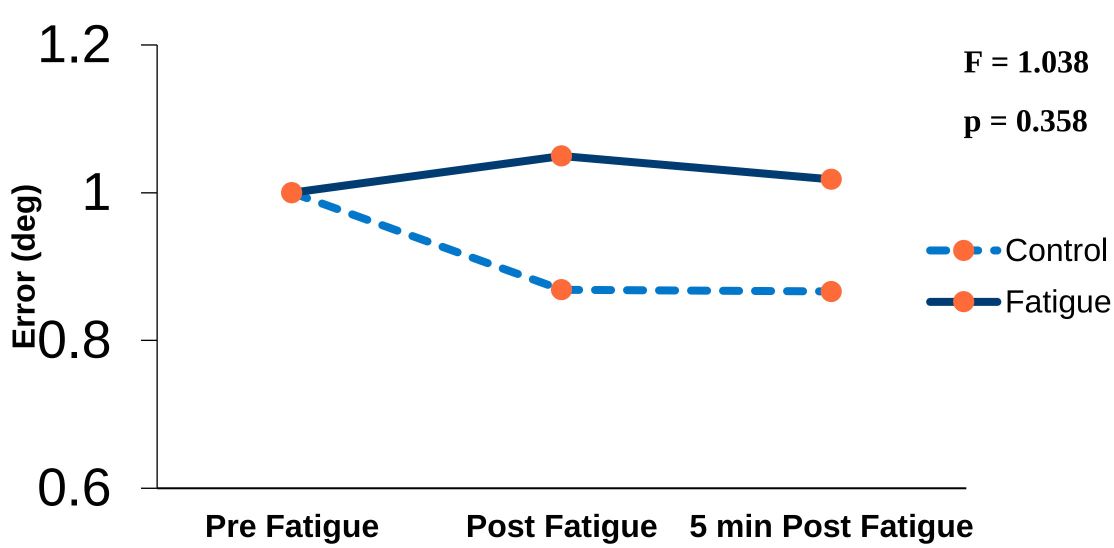

Figure 3.11: Comparison of Relative Shoulder Rotation Joint Angle Matching Task Variable Error between Control and Neck Fatigue Groups.

3.3.4. Error Trend

Figure 3.12 illustrates the bivariate plotting of all given target angles to their reproduced angles. Given the regression of this data is y = 1.0297x + 1.0829, one can see that the interaction is essentially a 1:1 relationship, with a ≈3% increase in over approximation at greater angles. The average overshoot of any actively reproduced angle is given here as the intercept b = 1.0829.

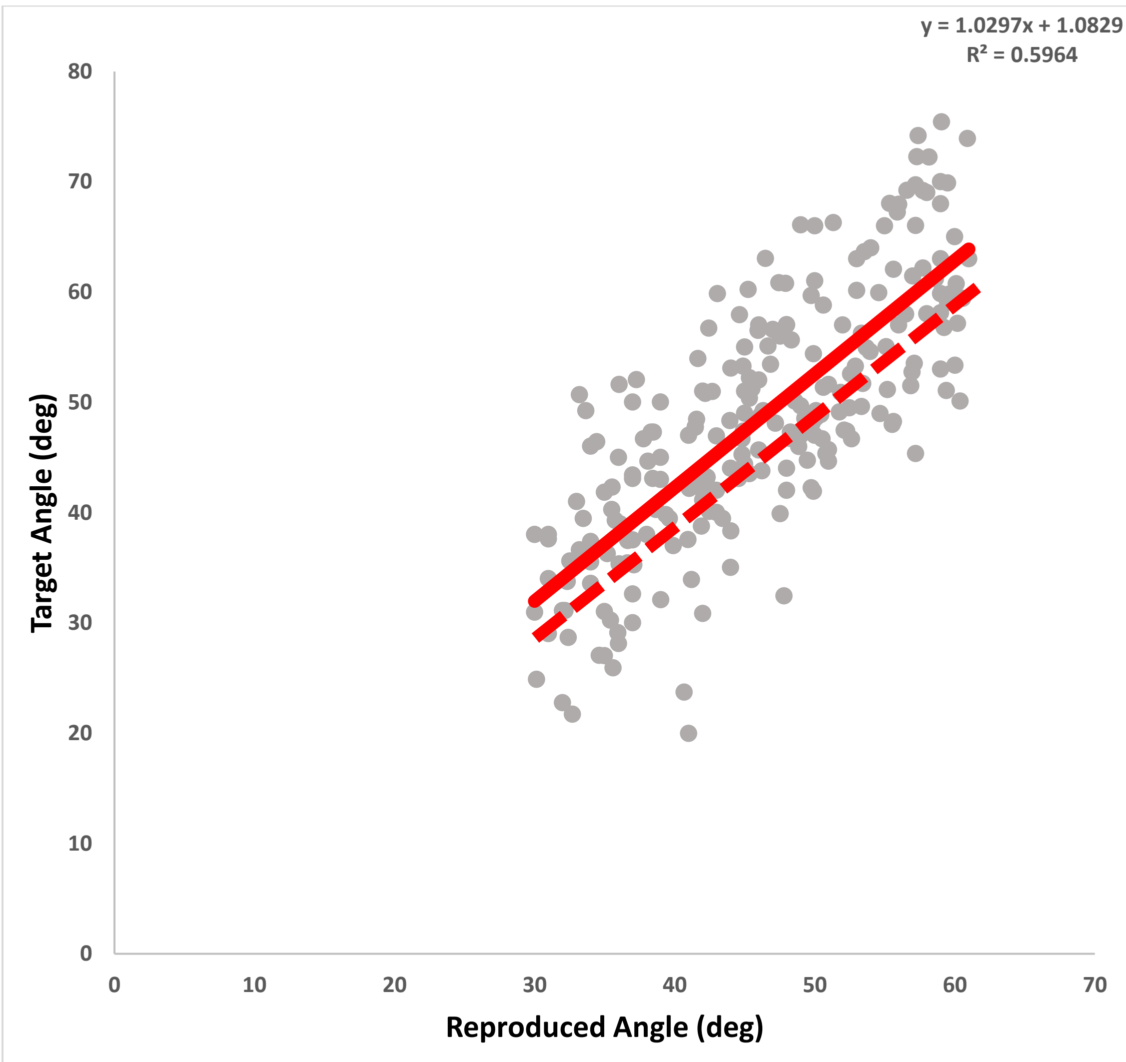

Figure 3.12: Bivariate Plotting and Regression of Target Angle to Actively Reproduced Angle.
*The linear regression fitted to the data above is illustrated as an unbroken line.
** A perfect y=x regression example is illustrated above as a dashed line.

3.4. Discussion

The purpose of our study was compare shoulder proprioceptive differences between individuals with acute neck muscle fatigue in comparison to healthy controls. Despite the induction of significant neck muscle fatigue, no significant changes in shoulder JPS proprioception were observed between the fatigue and control groups. This is contrary to previous research at the elbow (Knox and Hodges 2005, Haavik and Murphy 2011, Zabihhosseinian, Holmes et al. 2015) and wrist (Reece 2019) which showed that neck muscle fatigue increased error within upper limb joint position matching tasks. Many of these studies also utilized the Edmonston et. al neck fatigue protocol as we also chose to do to keep consistency with the fatiguing stimulus (Edmondston, Wallumrød et al. 2008). Our method of error measurement via absolute, constant, and variable error is also consistent with previous literature (Granit 1972, Lephart, Warner et al. 1994, Goble 2010, Haavik and Murphy 2011, Han, Anson et al. 2013, Reece 2019, Zabihhosseinian, Yielder et al. 2019).

We postulate that this is most likely due to our novel device allowing too much freedom across the joints of the upper limb and potentially torso. As this was our first research study with the device, we purposefully chose to opt for a less constrained task design to mirror the movement freedom in a work environment. While this opened our task up to significantly more DOF, the intent was that simulating a work environment would further the power of our findings in ergonomics research.

However, we can see now that any potential for the emergent trends between our fatigue and control groups (Fig. 3.9 & 3.10) struggle to overcome the large degree of variability (Fig. 3.11) that is an inherent product of our task being unconstrained. This is best illustrated by our findings of absolute error in figure 3.9, as absolute error is the best predictor of error between variables because it presents the overall deviation between them without considering the direction of error. In our research, this removes the potential for negative and positive errors to average out and be considered as a smaller error than there truly is. In figure 3.9 we see that absolute error is trending towards a significant difference, however this interaction is unfortunately dwarfed by a large F critical value which is a result of the large standard deviations in this data set.

Variability and standard deviation size are further augmented in our findings for constant error (Fig. 3.10). Constant error is an important measure in joint angle re-creation tasks to observe if there is a directional tendency. Many previous studies examining the efficacy of joint angle matching/re-creation tasks observe a tendency to over approximate joint angles determined solely by proprioception by roughly 8% (Jerosch and Prymka 1996, Jerosch, Thorwesten et al. 1997, Knox and Hodges 2005, Goble and Brown 2008, Zabihhosseinian 2014, Zabihhosseinian, Holmes et al. 2015), while targets determined with vision are likely to undershoot (Worringham, Stelmach et al. 1987, Goble and Brown 2008, Han, Anson et al. 2013). The tendency to overshoot in the absence of vision was observed to the same magnitude in our raw data for constant error where our fatigue and control groups collectively over approximated joint angles by 3 degrees (7%) at baseline (Fig. 3.12).

We see an increase in constant error from baseline for both our fatigue and control groups across subsequent sets, indicating that the tendency to over approximate joint angles worsened. Studies in literature typically only present average constant error, focusing on inter-set changes in absolute error instead. Few studies investigate changes in constant error across set in the absence of an intervention such as fatigue (Goble and Brown 2008, Zabihhosseinian, Holmes et al. 2015), and therefore there is minimal data on normal changes in inter-trial central tendency among healthy participants. One might expect to see the directional tendency of constant error approach neutral as participants refine their proprioceptive acuity over multiple trials. However, in the absence of empirical feedback, participants cannot base future joint angle re-creation attempts on their previous performance errors. Granit (1972) may provide one suggestion as to why participant's constant error trends towards greater over estimations across sets (Granit 1972). Granit found that participants were more accurate at joint angle re-creation tasks when they actively selected target angles themselves, versus having target angles passively presented to them. He hypothesized that this interaction was a product of gamma motor signaling from the muscle spindle increasing the sensitivity of alpha motor neurons through alpha-gamma co-activation during subsequent contractions. This progressive increase in perceived proprioceptive afference associated with active joint position re-creation attempts, combined with the observed trend to over approximate joint angles in the absence of vision, may be one possible hypothesis towards why we observed both our fatigue and control participants drift away from central tendency.

One study by Barden et al. (2004) collected 10 trials of upper limb repositioning tasks in 12 healthy subjects as well as 12 subjects with multidirectional shoulder instability (Barden, Balyk et al. 2004). The upper limb repositioning tasks involved unconstrained active joint angle re-creations across all three planes of shoulder motion: humeral abduction/adduction in the plane of elevation, humeral elevation, and humeral rotation. Their study found that average absolute error in these three planes of movement improved significantly across sets 1 and 2, and stabilized through sets 3 to 10. These findings would further validate that the increases in absolute error and decreases in variable error we observe in the fatigue group may be due to more than random chance.

It is likely that the large amount of variability in our findings was a result of our joint angle re-creation task unlocking too many DOF. Previous studies have almost exclusively focused on isolated single-joint performance changes following altered sensory input to the neck (Ribeiro and Oliveira 2007, Haavik and Murphy 2011, Baarbe 2015, Zabihhosseinian, Holmes et al. 2015, Reece 2019, Zabihhosseinian, Yielder et al. 2019). In order to improve the reliability of the Shoulder JPS Measurement Device in our analysis of JPS and proprioception, we will need to take a more in depth look at how the unconstrained joints of the upper limb integrate to perform the task of axial rotation about the shoulder. Prior to constraining shoulder movements, future research should the first validate the Shoulder JPS Measurement Device to gold standard kinematics of the upper limb during shoulder axial rotation, to better understand what planes of movement are contributing to this task, and ensure that the device software is accurately recording shoulder joint angle.

Future protocols involving this device may need to find an appropriate method of constraining participants to lock more DOF. However, researchers should take care in minimizing the contribution of additional cutaneous feedback as they look to find ways of constraining the upper limb. This can be a very difficult trade off when designing protocols to constrain and test JPS, as the CNS will readily incorporate available cutaneous afference to refine its subconscious body schema (Laszlo 1992, Han 2013).

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LINKING STATEMENT TO MANUSCRIPT 2

The results from our study investigating the effect of neck fatigue on thoracohumeral rotation of the upper limb determined that, despite the induction of significant neck fatigue, no significant differences in error occurred between the fatigue and control groups. These findings were antithetical to the dominant trend reported in literature which suggests the neck fatigue results in proprioceptive decrements to all joints of the upper limb, including the shoulder. In our initial piloting of the Shoulder JPS Measurement Device, it was assumed that while rotating the shoulder the device would primarily be measuring glenohumeral rotation of the unconstrained joint, and that other DOF of the arm would be negligible. However, the results of our study have found a large degree of variability in error scores; such that differences in variability between neck fatigue and control groups were dwarfed by very large standard deviations. We hypothesized that this was likely due to the device allowing too much freedom of movement across the other unconstrained joints of the upper limb, and potentially even contributions from the torso. This suggested a need to further evaluate mechanics across all upper limb joints when using this novel device, in order to validate its ability to measure glenohumeral rotation. Therefore, the second study in this thesis aimed to examine the motion at all DOF of the upper limb including and distal to the glenohumeral joint when interacting with Shoulder JPS Measurement Device during a joint position reconstruction task identical to the one employed for study one. The goal of this study was to evaluate all upper limb DOF in a multiple regression to predict device arm rotation measurement and subsequently assess the statistical coefficients associated with each DOF to develop a plan for how future research studies can potentially improve the relevance of this device in measuring a humeral rotation task.

CHPATER 4.

STUDY 2: ASSESSING THE CONTRIBUTION OF DIFFERENT UPPER LIMB DEGREES OF FREEDOM TO AN UNCONSTRAINED GLENOHUMERAL PROPRIOCEPTION TASK

4.1. INTRODUCTION

The past decade has seen major contributions to the literature surrounding the topic of body schema. Research has demonstrated that interpreted head and neck posture can influence the cortical representation of posture and JPS, termed "body schema" (Knox and Hodges 2005, Knox, Cordo et al. 2006, Knox, Coppieters et al. 2006). Deviations in head and neck posture could significantly impair the ability of healthy participants to recreate elbow posture (Knox and Hodges 2005). However, other research suggests that afferently disruptive stimuli can have the same detrimental effects to proprioception as manipulating head posture (Schieppati, Nardone et al. 2003, Falla, Bilenkij et al. 2004, Falla, Jull et al. 2004, Falla and Farina 2005, Schmid and Schieppati 2005, Knox, Cordo et al. 2006, Knox, Coppieters et al. 2006, Barker 2011, Haavik and Murphy 2011, Schomacher and Falla 2013, Baarbé, Murphy et al. 2015, Baarbé, Yielder et al. 2015, Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Holmes et al. 2017, Reece 2019, Zabihhosseinian, Yielder et al. 2019).

Some of the more common stimuli used to disrupt sensory afference in literature include pain (Lewis, Kersten et al. 2010), tendon vibration (Knox, Cordo et al. 2006), and neck fatigue (Zabihhosseinian, Holmes et al. 2015). Of these factors, neck fatigue may be the most endemic disruptive stimuli to the neck in healthy populations, where workplace stresses on the cervical spine clinically impact one in three people within their lifetime (Huisstede, Bierma-Zeinstra et al. 2006, Hogg-Johnson, Van Der Velde et al. 2008). Disruptive stimuli, such as neuromuscular fatigue, impair muscle spindles and mechanoreceptors, for which there is an abundance in the neck (Jull, Falla et al. 2007). This is because the CNS readily references neck and head posture to update body schema (Gallagher 1995, Holmes and Spence 2004, Proske 2015). While implications of altered sensory afference at the neck to disturb body schema and proprioceptive efferents in the upper limb have been established at the elbow and wrist (Zabihhosseinian, Holmes et al. 2015, Reece 2019), the shoulder has yet to be examined.

In recent literature, two key studies have focused on different mechanical analyses for quantifying the shoulder joint and used said approaches to study the effect of neck fatigue on shoulder motion. Zabihhosseinan et al. (2017) utilized an experimental scapular kinematics approach (Zabihhosseinian, Holmes et al. 2017) originally pioneered by Karduna and colleagues (Karduna, McClure et al. 2001) and later refined by Bourne (Bourne, Choo et al. 2011) to optimize uni-directional planes of motion. Zabihhosseinain et al. (2017) found a significant difference in compensatory mechanics between healthy individuals and those with chronic low-level neck pain when adapting to acute neck muscle fatigue to complete an arm elevation task. The second study was a follow-up by Zabihhosseinian et al (2019), who quantified shoulder proprioceptive performance using a hand distance-to-target error measurement for an eye-hand tracking task (Zabihhosseinian, Yielder et al. 2019). For this study, participants were tasked to move an object to a target during trials with both vision allowed and occluded. This task was performed by isolating the limb such that end effector target performance was indicative of shoulder rotation. This study found a significant decrease in visually occluded accuracy following induction of neck fatigue, suggesting that the neck muscle fatigue impacted the neck shoulder body schema relationship, resulting in decreased upper limb proprioceptive accuracy.

While the recent work by Zabihhosseinian et al. (2017 & 2019) has been an invaluable foray for beginning to address the minimal research in shoulder proprioception, one limitation on findings thus far is the incorporation of axioscapular musculature in unconstrained shoulder mechanics (Zabihhosseinian, Holmes et al. 2017). This occurs following the onset of acute neck muscle fatigue as prime movers of the scapula that originate on the cervical spine may also become fatigued, contributing to performance decrements and confounding the true effect of altered sensory feedback. One possibility step to mitigate this interaction may be in the design of a task which serves to isolate the humeral function of the shoulder from the scapular function. However, it is vital to find a method of isolating glenohumeral motion from scapulohumeral motion without providing excess cutaneous afference through excessive constraint of the arm (Collins, Refshauge et al. 2005, Proske 2015). A potential approach would be the design of a humeral rotation task, whereby the scapula will be inclined to contribute very little as it is primarily involved in unlocking the glenohumeral joint for the purpose of clearing the acromial process during elevation and abduction (Prescher 2000).

In 2003, Lee and colleagues attempted such a design to quantify decrements in humeral rotation proprioception due to local muscle fatigue (Lee, Liau et al. 2003). Their experimental approach combined a Con-Trex MJ dynamometer (Con-Trex, Zurich, Switzerland) and a self-designed proprioception testing apparatus that had been previously validated (Lee 1998). This mechanism proved effective at isolating and measuring humeral rotation, however a key limitation of this design was the many surfaces of cutaneous articulation which promote additional sensory feedback. There have also been attempts to minimally constrain the shoulder in an attempt to minimize cutaneous feedback. One such approach involved the experimental design and application of the active movement extent discrimination apparatus (AMEDA), which has been extensively utilized in joint angle re-creation tests at the University of Canberra (Waddington and Adams 1999, Waddington, Seward et al. 2000, Naughton, Adams et al. 2005, Han, Waddington et al. 2011, Han 2013). However, a trade-off exists, as attempts to minimize cutaneous feedback may increase movement variability, making it more challenging to find differences when comparing between individuals.

Therefore, for the purposes of quantifying shoulder biomechanics performance during joint angle repositioning tasks, similar to those previously conducted at the elbow (Zabihhosseinian, Holmes et al. 2015), our lab collaborated with our institutions engineering department to design a novel Shoulder JPS Measurement Device. This device was used to quantify the effect of neck muscle fatigue on shoulder rotation joint accuracy (study one of this thesis). However, this research found a large degree of variability, potentially due to the wide variety of movement strategies adopted when the limb was unconstrained. Therefore, the purpose of this chapter was to evaluate upper extremity kinematics using our custom-built humeral rotation proprioception device. The goal was to determine how much of the motion in the humeral rotation angles recorded by the proprioception device was explained by optically-obtained shoulder humeral rotation in comparison to the other DOFs in the arm.

4.2. METHODS

4.2.1. Participants

32 participants (16 M, 16 F) were recruited from the local student population at Ontario Tech University. Participants had a mean age of 22.96 ± 3.64 years (table 4.1). Eligibility required that all participants be right hand dominant, and free of neck and shoulder pain for the last 6 months. Participants were excluded if they reported being involved in an occupation which required exertion of the neck or upper arm such as heavy machinery operation or carpentry. Participants who disclosed that they had undergone shoulder or spine surgery were also excluded. Written informed consent was obtained and participants completed the Edinburgh Handedness Inventory (EDH), Neck Disability Index (NDI) and Chronic Pain Grade Scale (CPGS) to determine handedness, neck pain intensity, and neck pain effects respectively.

Table 4.1: Participant Anthropometrics

Gender	Age	Stature (cm)	Mass (kg)
Male	23.5 (SD± 3.4)	179.3 (SD± 6.4)	79.5 (SD± 11.8)
Female	22.3 (SD± 3.5)	164.3 (SD± 9.7)	60.4 (SD± 8.8)

4.2.2. Instrumentation and Data Acquisition

4.2.2.1. Kinematics

Participants were instrumented with 22 infrared emitting diodes (IREDs) (NDI, Waterloo, Ontario) across their posterior thorax and right upper limb. Two banks of OptoTrak 3D Investigator cameras (NDI Instruments, Waterloo) tracked the location of these IREDs at a sampling rate of 64 Hz. IREDs were organized into four rigid bodies to track the 3D orientation and position of the following segments: 1) 8th thoracic vertebrae, 2) midpoint of the lateral humerus at the deltoid tuberosity, 3) midpoint of the dorsal forearm, and 4.) across the dorsum of the third metacarpal. These rigid bodies were used to track virtual digital anatomical landmarks (Table 4.2, Fig. 4.1). All kinematics procedures were in accordance with International Society of Biomechanics (ISB) standards (Wu et al. 2005).

Table 4.2: Rigid Bodies and their Resultant Digitized Anatomical Structures

Referential Rigid Body	Digitized Anatomical Landmark
8th Thoracic Vertebra	8th Thoracic Vertebra, Left Acromion, Right Acromion, Incisura Jugularis, Xiphoid Process
Right Humerus	Acromion Process, Lateral Epicondyle, Medial Epicondyle
Right Forearm	Lateral Epicondyle, Medial Epicondyle, Radial Styloid, Ulnar Styloid
Right Hand	Radial Styloid, Ulnar Styloid, Base of 2nd Phalange, Base of 3rd Phalange, Base of 5th Phalange

Figure 4.1: Kinematic Marker Layout.

Once fully digitized, participants sat with their dominant right arm abducted to 90 degrees in the frontal plane, with their elbow flexed to 90 degrees and their hand pointing straight up. In this position, participants were matched with the Shoulder JPS Measurement Device.

4.2.2.2. Shoulder Joint Position Sense Measurement Device

A custom-made Shoulder JPS Measurement Device was engineered to isolate and measure humeral rotation shoulder motion, about an axis defined by the humerus longitudinally (Fig. 4.1). The device included an adjustable arm to match an individual's forearm length, such that rotation about the device's central axis paralleled humeral rotation of the participant. The devices adjustable handle also had a button on it's distal end, which when pressed, would record the rotation of the central axis at 1000 Hz until the button was pressed a second time. The device would output the axial rotation and rate of axial rotation to the hundredth of a degree and hundredth of a degree per second, respectively.

The Shoulder JPS Measurement Device consists of a central axel, which on one end was connected to a perpendicular arm with a length-adjustable hand grip (Fig. 4.2). On the other end, the central axel was attached an E6B2-C incremental rotary encoder (OMRON Corporation, Kyoto, Japan) which converts rotation along the central axis into analog signal. The rotary encoder was wired to an Arduino dual-board microcontroller to convert analog signal to digital. The Arduino board also integrates with a button located on the adjustable hand grip. Toggle interaction with this button was used to turn recording of the encoder on and off. The Arduino circuit board then outputs to the host computer via universal serial bus (USB) cable.

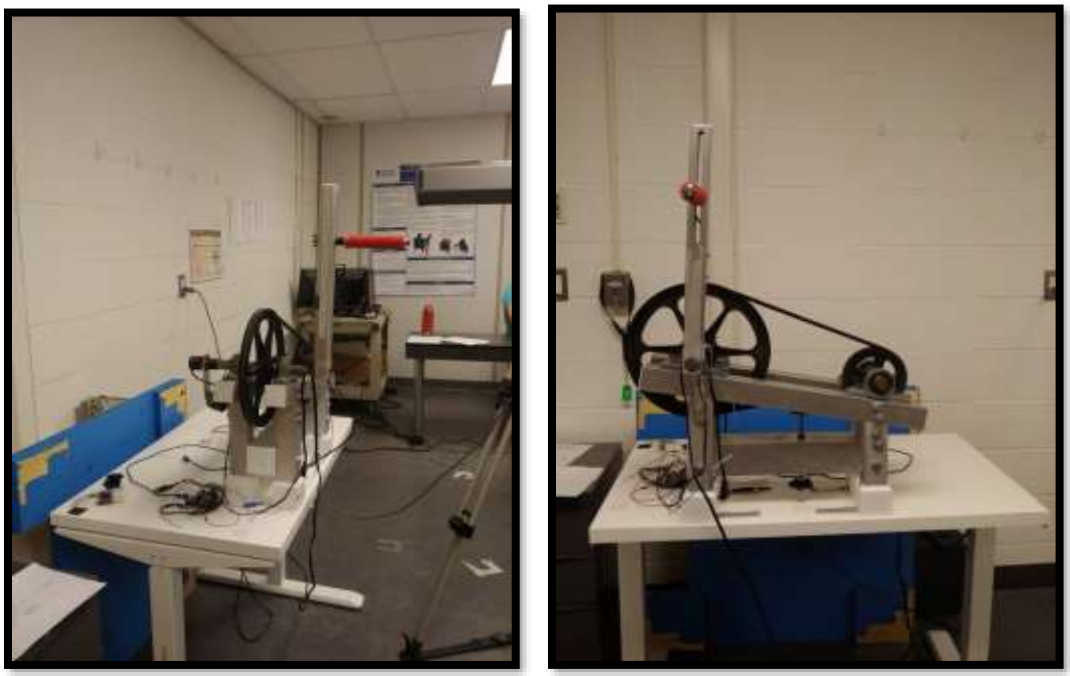


Figure 4.2: Images of Shoulder Joint Position Sense Measurement Device.

Figure 4.3: Participant Interaction with the Shoulder Joint Position Sense Measurement Device.

4.2.3. Task Description

4.2.3.1. Joint Angle Matching Task

Before beginning the joint angle re-creation task, participants were first familiarized to both the task and device. This was done by instructing participants in the proper interaction with the device while they were presented with live visual feedback. Once participants verbally acknowledged their comfort and understanding of the device parameters, the familiarization period was terminated.

Participants then began the joint angle matching task. This task consisted of four sets of 3 unique joint angel matching trials. At the beginning of each trial, participants were correctly aligned with the device and asked to put on their visual occlusion goggles when they were ready to begin. Each trial began with the participant holding the handle on the device in the straight up position (indicated via level) which was termed the "home" position. The participant's arm was then guided passively to a randomly generated angle between 30-60 degrees of internal humeral rotation, termed the "target" position. Passive target joint approximation was guided by the researcher at a speed of ~10 deg. /sec. The target arm posture was maintained for five seconds and then passively returned to the home position. At this point, the participant was given active control over the manipulation of the device arm and asked to recreate the target joint angle, pressing a button on the device's arm when they believe their arm posture to be accurate.

This process was repeated 2 more times (3 trials total) to constitute a single set. For each trial, a different target angle between the parameters of 30-60 degrees was randomly generated. Home positon was always maintained at the same posture. Between each set, participants were provided five-minute breaks to take off their occlusion goggles and rest their arm to mitigate local muscle fatigue. At no point was the participant privy to data regarding their performance. Upon completion of the fourth set, participants were unequipped from their kinematics gear, debriefed, thanked for their participation, and allowed to leave.

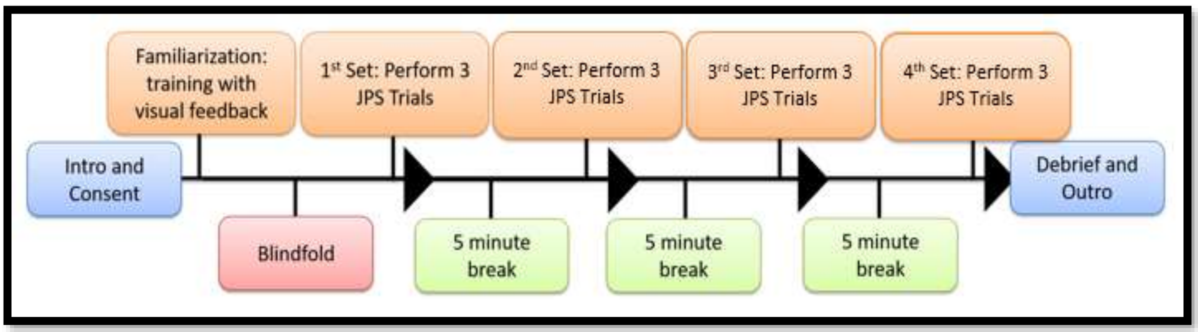


Figure 4.4: Study Timeline.

4.2.4. Experimental Procedure

After setting up participants with kinematic rigid bodies and familiarizing them with the Shoulder JPS Measurement Device, participants began the experimental joint angle matching protocol. Participants were blinded with visual occlusion goggles and tasked to recreate shoulder joint angles immediately after they were shown to them (as described in section 2.3.1 above). Trials were blocked into 4 sets, with 3 joint angle matching trials per set. 5 minute breaks were provided between each set of 3 joint angle matching trials to prevent local muscle fatigue.

Torso and right upper limb kinematics were captured using NDI First Principles motion capture software (Northern Digital Inc., Waterloo, Canada). The participant was instructed to hold onto the Shoulder JPS Measurement Device handle, to restrict the motion of the joint angle matching task to the same space for each trial. Holding on to the Shoulder JPS Measurement Device also served to support the arm and reduce local muscle fatigue in the shoulder and rotator muscle complex. The self-supporting structure of the task also served to relay minimal cutaneous feedback throughout both passive and active shoulder joint rotations; only the right hand touched the machine during trials. Each trial consisted of two phases: one for the passive joint approximation, where the experimenter moved the participants arm to a certain angle, and one for the active angle re-creation by the participant. This created a combined total of 24 kinematics files for each participant. This data was saved and exported in Visual 3D v6 Professional Biomechanics Modelling software.

4.2.5. Data Analysis

Kinematics trial data weas imported to Visual 3D for joint angle processing. Joint angles for the shoulder, elbow and wrist were computed using an Euler decomposition sequence as per standard International of Society Biomechanics (ISB) conventions (Wu et al. 2005) (table 4.3). For each trial the first 50 frames and last 50 frames (100 Hz) of data for each joint DOF were averaged to constitute the start and end upper limb joint positions respectively for each trial.

Table 4.3: Rotation Sequences used to Calculate Upper Limb joint Angles in Visual 3D

Joint Angle	Rotation Sequence	α	β	γ
Shoulder (Thoracohumeral)	Y – X - Y	Plane of Elevation (γ) Horizontal Flexion (+) Horizontal Extension (-)	Negative Elevation Depression (+) Elevation (-)	Axial Rotation (γ2) Internal Rotation (+) External Rotation (-)
Elbow	Z – X –Y	Flexion Flexion (+) Extension (-)	Carrying Angle	Pronation Pronation (+) Supination (-)
Wrist (Radiocarpal)	Z – X – Y	Flexion Flexion (+) Extension (-)	Pronation Pronation (+) Supination (-)	Deviation Radial Deviation (+) Ulnar Deviation (-)

*Rotation Sequences are ISB Suggested Rotation Sequences (Wu et al., 2005)

To compare shoulder JPS device output to kinematics output for each trial, rotation of the devices central axis was obtained from the integrated rotary encoder. Internal rotation of the humerus was expressed by both the Shoulder JPS Measurement Device and Visual 3D as a positive joint rotation from a starting point of 0 degrees.

4.2.6. Statistical Analysis

The angles output by the shoulder rotation device were assumed to be the dependent variable (DV) in all analyses. Pearson correlations were computed between the device angles and each of the corresponding joint angles derived from motion capture.

A multivariate regression approach was utilized with Shoulder JPS Measurement Device axial rotation angle as the DV. Plane of humeral elevation [HumPOE], negative humeral elevation [HumNegElv], thoracohumeral rotation [HumRot], elbow flexion [ElbFlex], elbow pronation [ElbPro], wrist flexion [WrstFlex], wrist pronation [WrstPro], and wrist deviation [WrstDev], were all input into the model as predictor variables (PV).

Separate multiple regression models were computed using forward standard and step-wise approaches. In the standard regression model, all PVs were included. In the stepwise model, PVs were included based on their beta-weighting (β-weighting) towards the DV, their variance inflation factor, and their significance towards predicting the DV.

4.3. RESULTS

4.3.1. Questionnaires

The mean participant score for the EHI was 67.2 ±26.0. This indicates that our participants were dominantly right handed/right armed. Additionally, no single participant scored below 20, indicating that every participant was right-handed, according to the EHI.

The mean participant baseline score for the NDI was 6 ± 6.5 indicating that participants self-reported neck pain disability ranged from non-existent to very mild.

The CPGS reports two measures of chronic pain: characteristic pain intensity and pain related disability. The average participant score for the characteristic pain intensity section on the CPGS was 19.5 (SD+/- 14.5), indicating participants self-measured themselves as having the lowest intensity of chronic pain categorized by the CGPS. The average participant score for the pain related disability section on the CPGS was 0 (SD+/- 0) indicating that participants self-reported themselves as having no physical or lifestyle disability due to chronic pain.

4.3.2. Joint Angle Matching Task

The randomly generated passive target angles for the joint angle matching task averaged to 45.11° across all trials. This is broken down by set in figure 4.5. The actively reproduced target joint angles for the joint angle matching task averaged to 48.57° across all trials. This is broken down by set in figure 4.5.

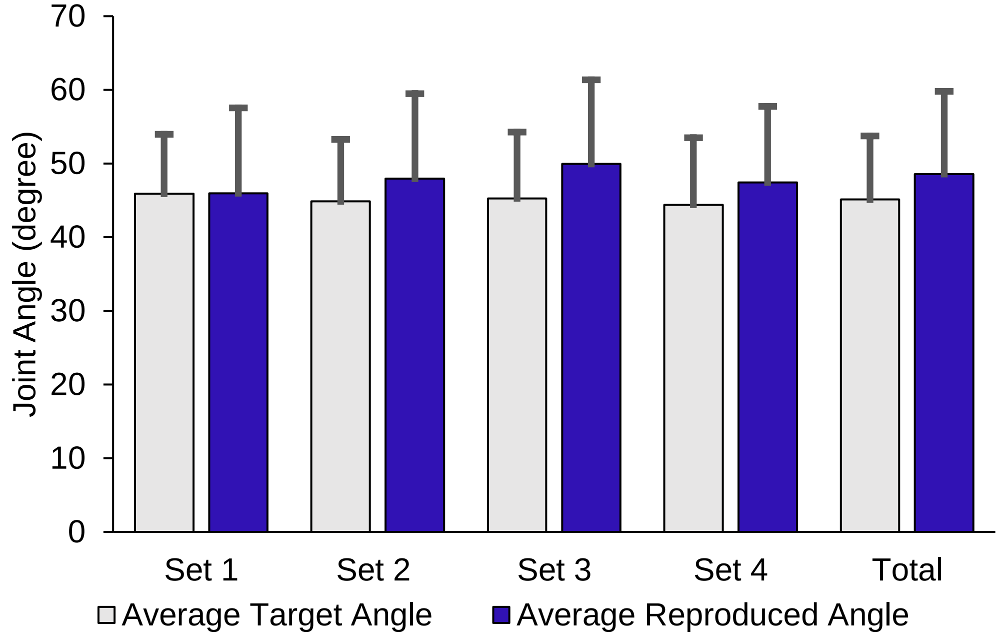


Figure 4.5: Changes in Average Target Angle and Average Reproduced Angle.

<u>4.3.3. Multiple regression</u>

4.3.3.1 Standard Multivariate Regression

The standard multiple regression entered all predictor's from the upper limb DOF into the regression model and returned a correlation of R=**0.45** (Fig. 4.6) with the device arm rotation. The β-weights of all predictor's entered into the standard multiple regression are displayed in figure 4.7.

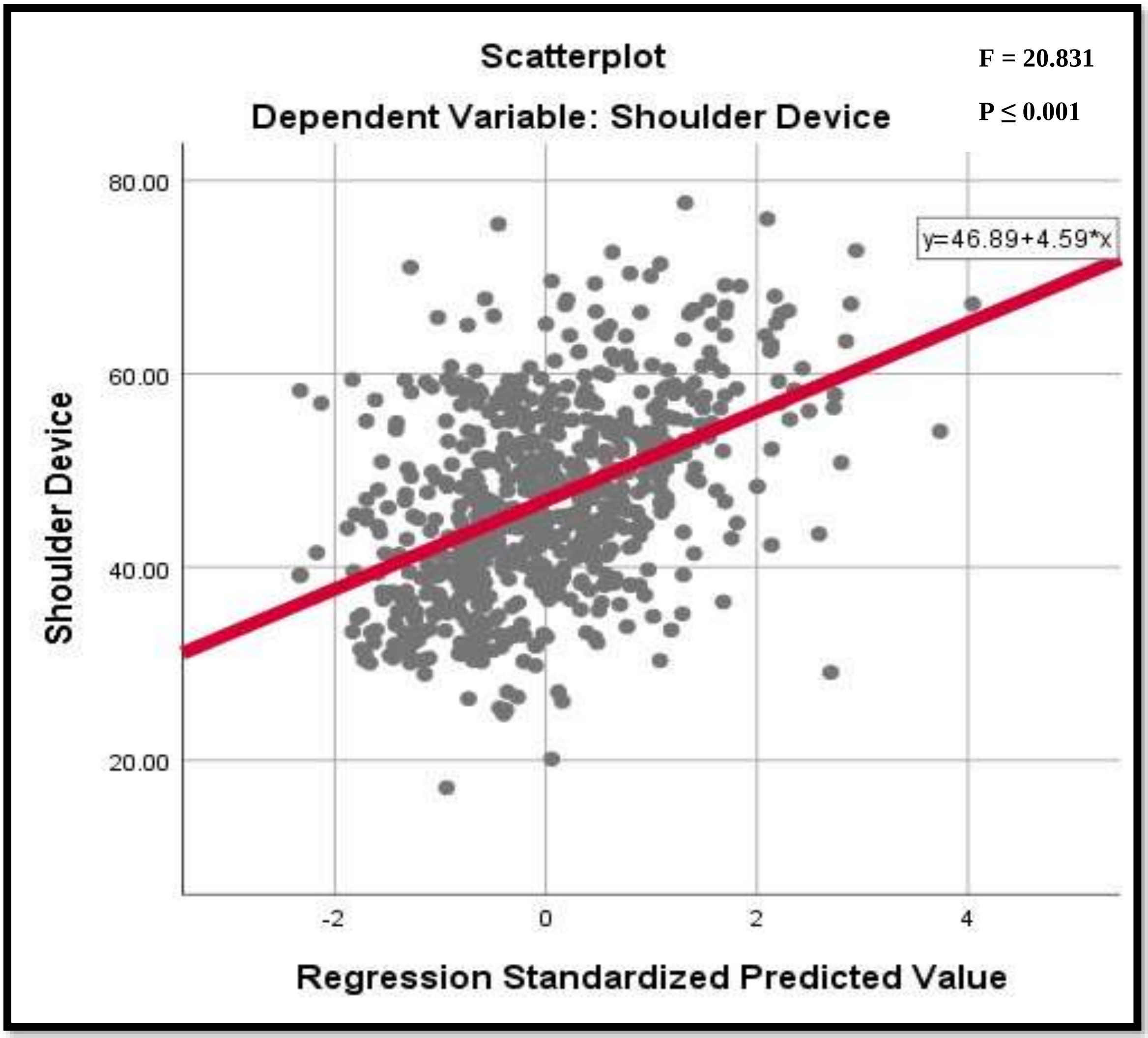


Figure 4.6: Standard Multivariate Regression of all Upper Limb Degrees of Freedom Predictor Variables Plotted to Shoulder Joint Position Sense Measurement Device Axial Rotation.

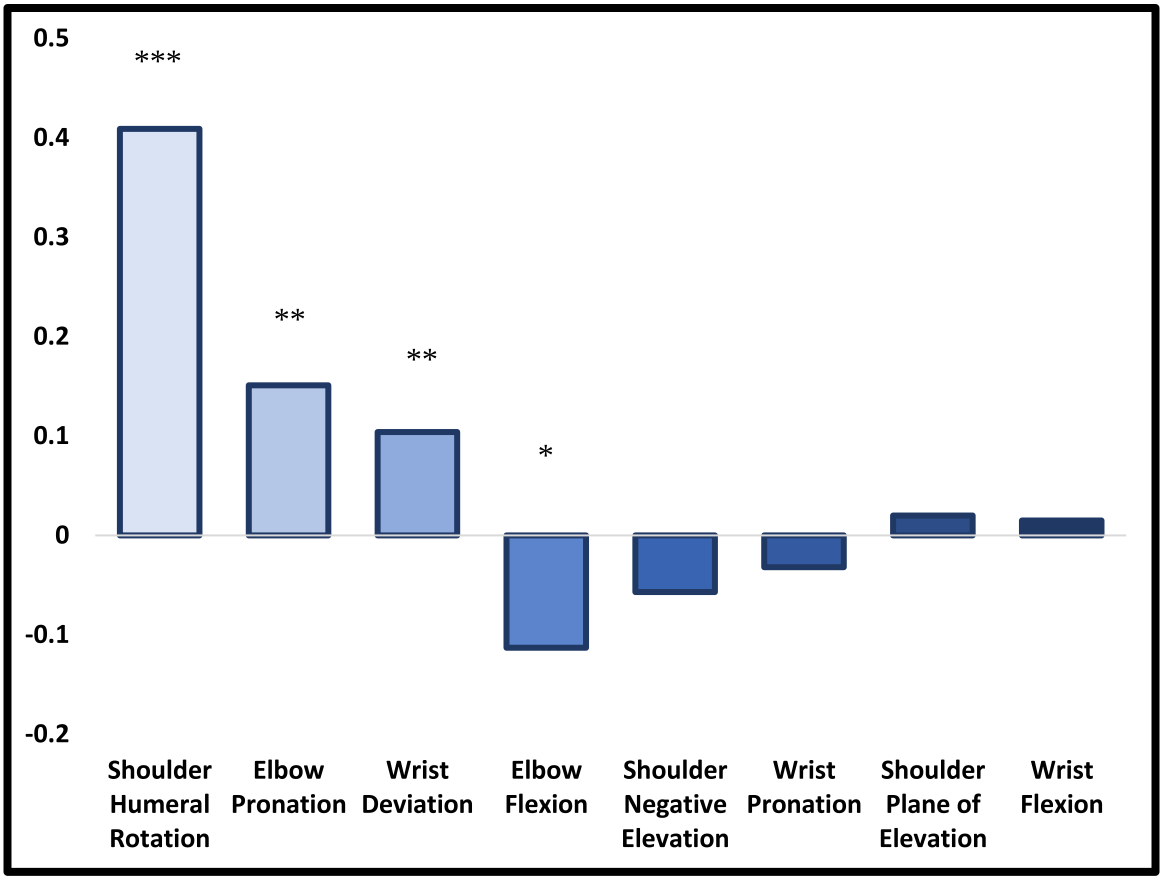


Figure 4.7: β-Weights of all Upper Limb Predictor Variables in Standardized Multivariate Regression to Shoulder Joint Position Sense Measurement Device Axial Rotation. Where * denotes $p \leq 0.05$, ** denotes $p \leq 0.01$, and *** denotes $p \leq 0.001$.

4.3.3.2. Stepwise Multivariate Regression

The stepwise multivariate regression model begins with a bivariate regression between the DV and most significant PV and then progressively enters subsequent predictors into the regression formula. This is performed iteratively until correlation significance is no longer impacted by the inclusion of additional variables.

The stepwise multiple regression model started with a bivariate regression formula between the Shoulder JPS Measurement Device axial rotation and Shoulder

Humeral Rotation [HumRot]. This model returned a correlation of R=**0.41** as seen by the predictor's zero-order correlation (table 6.1 appendices).

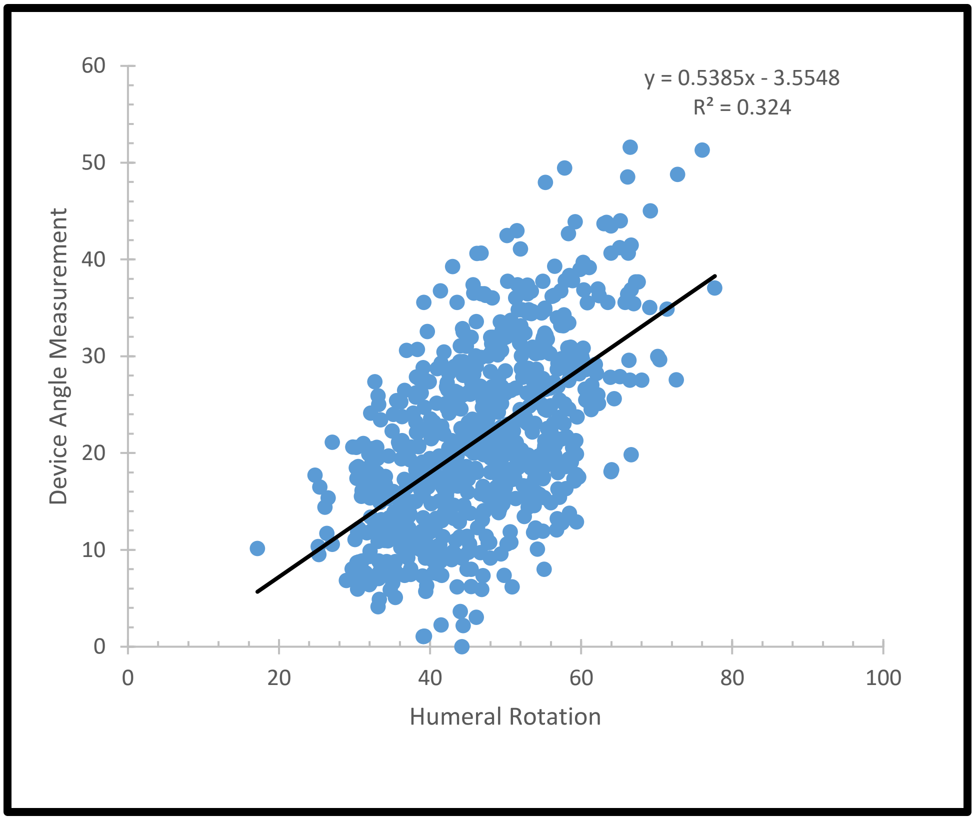
Based on the values presented in tables 6.2 and 6.3 (in appendices), subsequent predictors of the shoulder – [HumNegElv] and [HumPOE], as well as subsequent predictors of the elbow – [ElbFlex] and [ElbPro], all presented a significant partial correlation with humeral rotation. Only the predictors of the wrist showed significant variance in further predicting device rotation. Of these predictors, wrist deviation had the strongest contribution to device rotation correlation. Wrist deviation was therefore the next DOF PV to be entered in the stepwise model.

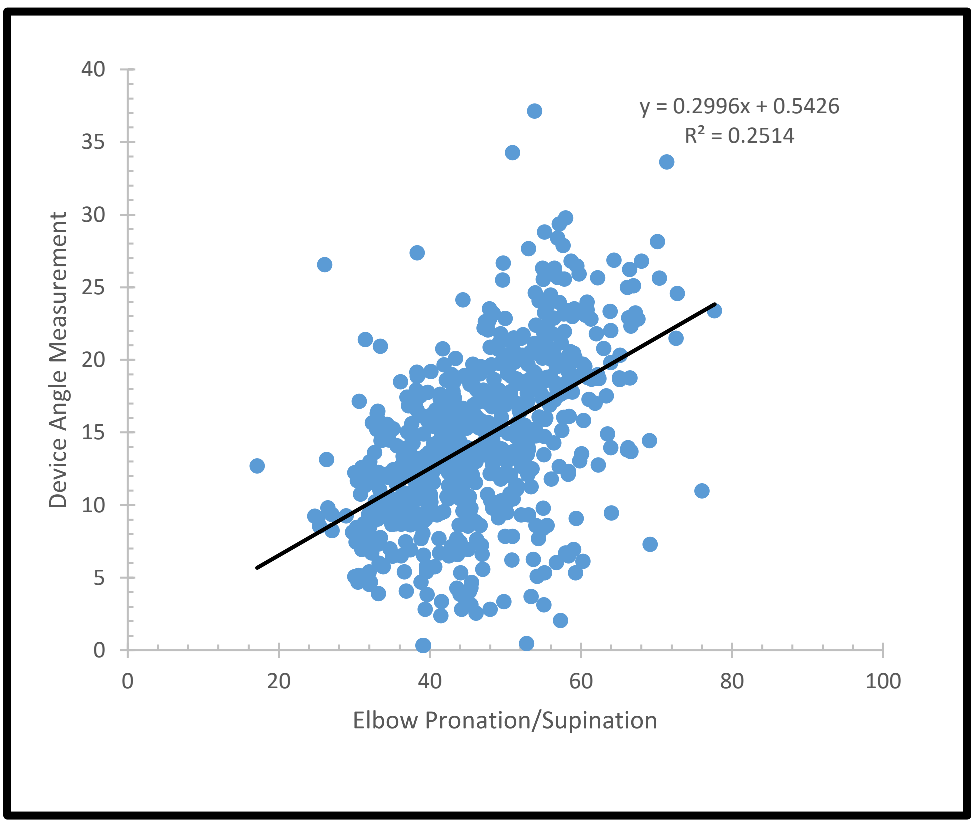
Due to wrist deviations' significant common variance with the other wrist PVs', and humeral rotation's significant common variance with the shoulder and elbow PVs' as well as shoulder plane of elevation, the stepwise models PV input peaked at the inclusion of [HumRot] + [WrstDev]. The model returned a correlation of R=**0.433**.

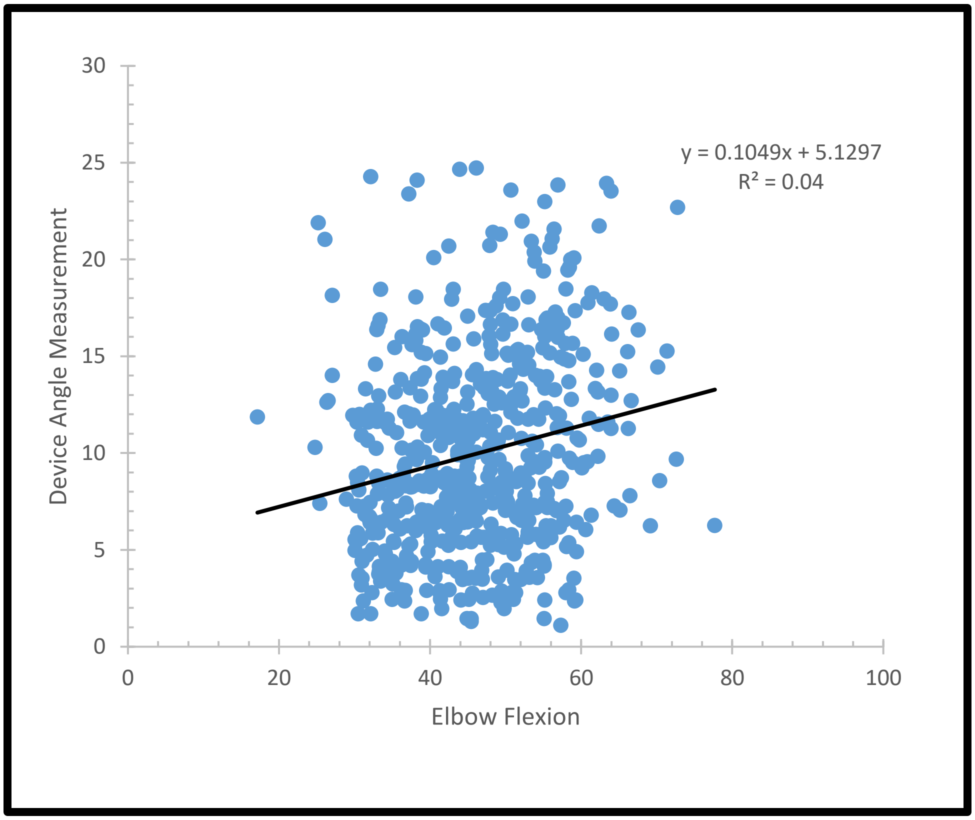
The final equation for the stepwise multivariate regression is presented in equation 4.1 and figure 4.9.

Equation 4.1: Stepwise Regression Formula

$$\boldsymbol{Y} = 36.504 + 0.390\,[HumRot] + 0.232\,[WrstDev]$$


y = 0.5385x - 3.5548
R² = 0.324
Device Angle Measurement
Humeral Rotation
0
10
20
30
40
50
60
0
20
40
60
80
100


y = 0.2996x + 0.5426
R² = 0.2514
Device Angle Measurement
Elbow Pronation/Supination
0
5
10
15
20
25
30
35
40
0
20
40
60
80
100



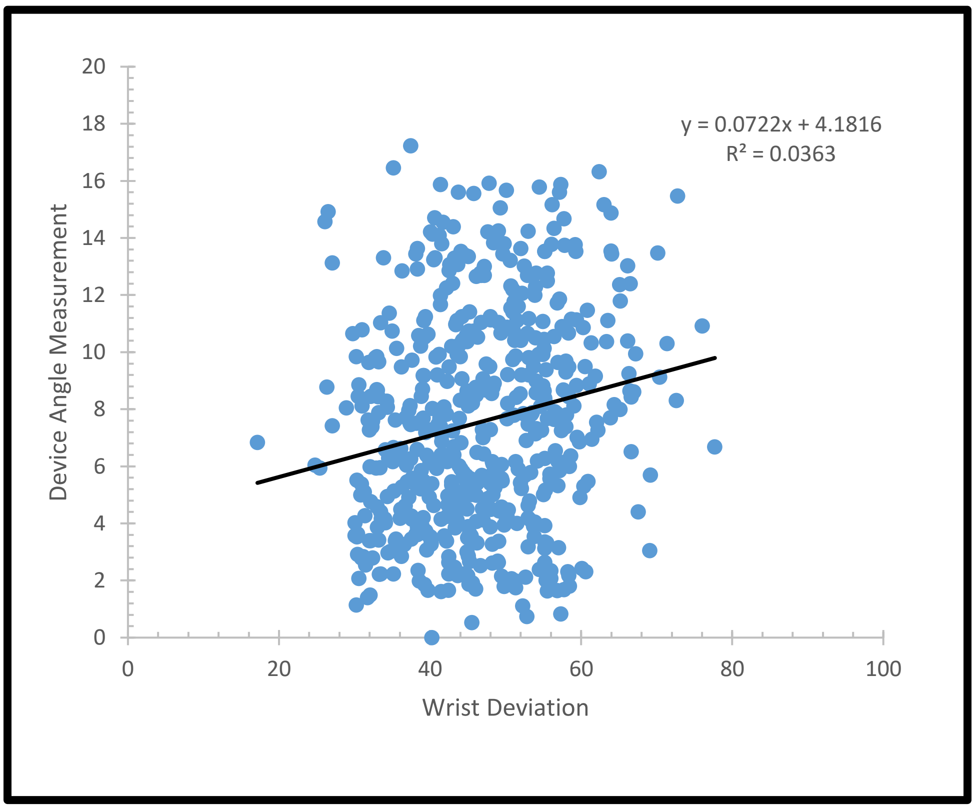


Figure 4.8: Bivariate Correlations of Significant Upper Limb Predictor Variables.

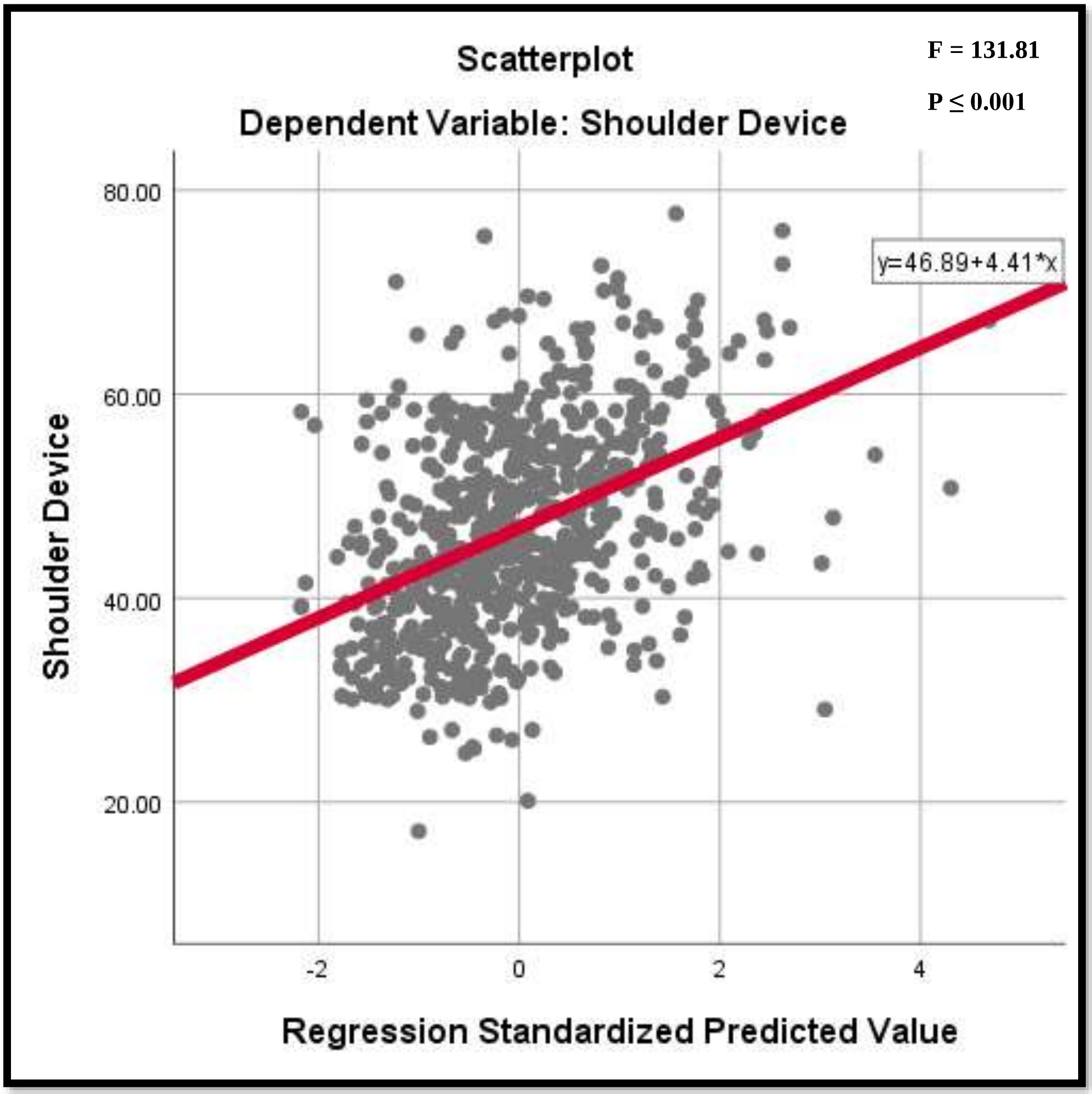


Figure 4.9: Stepwise Multivariate Regression of Shoulder Humeral Rotation and Wrist Deviation Plotted to Shoulder Joint Position Sense Measurement Device Axle Rotation.

4.4. DISCUSSION

Our standard multiple regression, involving all eight DOF of the upper limb, returned a correlation of R=0.45 (Fig. 4.6). However, when the same eight variables were entered into a forward stepwise model, all but 2 DOFs were dropped; Humeral Rotation and Wrist Deviation (Fig. 4.9). As explained within the methods section, these two variables were determined based on a number of criteria.

Firstly, the step-wise model built the regression formula based on the significantly most efficient contribution of PVs by either adding or removing variables depending on if it is a “forward” or “backward” model (Vincent and Weir 2012). In the case of our model, a forward stepwise approach was used, which means the model began with no PVs, and proceeded to add one PV at a time until the addition of further variables became non-significant to the refinement of the regression (Vincent and Weir 2012). The forward stepwise model then begins by first adding the PV of greatest significance (Vincent and Weir 2012). The coefficient results of our multiple regression displayed in figure 4.7 illustrate that humeral rotation was the DOF that explained the most variance in device arm rotation. The zero-order correlation of humeral rotation shown in table 6.1 (appendices) displays what is essentially the bivariate correlation between [HumRot] and device rotation (Johnson 2000). Therefore, based on the zero-order correlation between humeral rotation and device rotation we can identify that humeral rotation explains 41% of the rotation of the Shoulder JPS Measurement Device arm on its own.

The next variable to be entered into the stepwise model is then selected based upon a number of additional factors including the remaining variables β-weights’, their partial and semi-partial correlations, and their collinearity diagnostics. Significance is still taken into consideration as well, however it is not the only variable being considered at this stage. If we reference figure 4.7, we see that the PV of the next highest significance following humeral rotation is elbow pronation, however the model rejects the inclusion of elbow pronation. Again referencing table 6.1, we can determine that while elbow pronation had a high partial correlation, it also had a lower collinearity tolerance than any other variable, and subsequently had the highest variance inflation factor as well. If we want to break this down further, we can look at the correlations between PVs and their significances (covariances) presented in tables 6.2 and 6.3, respectively. Based on the coefficients presented, we see that humeral rotation had a significant correlation with all other PVs of the shoulder and elbow including elbow pronation. This significant correlation, combined with the poor collinearity coefficients for elbow pronation would indicate that the partial correlation and common variance between these PVs was likely very high and therefore elbow pronation provided very little unique variance in the prediction of device motion that could not already be predicted with humeral rotation.

Conversely, humeral rotation did not display a significant correlation with any of the variables of the wrist, making them a potential candidate for inclusion in the stepwise regression model. Of the DOF's of the wrist, wrist deviation supplied the largest β-weight by a significant margin, and also displayed the greatest zero-order correlation, a high collinearity tolerance, and a moderate variance inflation factor. Wrist deviation also reached significance, making it the next variable to be added to the stepwise regression model. At this point, the addition of subsequent PVs could not make a significant improvement in the prediction of device rotation as the combined variance of [HumRot] + [WrstDev] shared a significant correlation and collinearity with every other PV.

The average passive target joint angle of 45.11° (Fig. 4.5) for the joint angle matching task confirms that our random number generation of target angles did indeed produce an equal distribution of target joint angles between the parameters of 30° to 60°. This implies that we will have an equal distribution of target angles across our sample.

The average active joint target angle of 48.57 (Fig. 4.5) suggests that the participants tended to overshoot when recreating their passive target joint angle by approximately 3 degrees (8%). These findings are congruent with our previous research with the Shoulder JPS Measurement Device which found a tendency to over approximate by roughly 7% across a sample population or participants with acute neck fatigue and healthy controls. This is also supported by previous literature from other research studies involving JPR which support the observed tendency to over approximate joint angles by approximately 8-11% in the absence of vision (Jerosch, Thorwesten et al. 1995, Jerosch and Prymka 1996, Knox and Hodges 2005, Goble and Brown 2008, Zabihhosseinian, Holmes et al. 2015, Zabihhosseinian, Yielder et al. 2019).

Research in the function of multiple regressions suggests that one of the likely reasons for PVs to be highly collinear with each other is in a situation where change in one variable directly influences change in another (Mason and Perreault Jr 1991, Vincent and Weir 2012). This interaction is somewhat expected to influence PVs which describe DOF's of the same joint, as this is why in tables 6.2 and 6.3 we see that variables describing the shoulder and elbow have significant covariances with the other variables at that joint. However, it is interesting to note that this same phenomenon does not seem as apparent at the wrist where many of the correlations between these variables do not reach significance. Rather, the variables of wrist DOF seem to be more closely correlated with variables of the elbow and shoulder instead. This finding may suggest that while the movement at the wrist was still heavily derivative of the movement at the proximal arm joints, it may have had more variability than the shoulder or elbow.

The 0.41 zero-order correlation to the desired primary DOF of humeral rotation suggests there is room for improvement in refining the Shoulder JPS Measurement Device. The findings of our multiple regression suggest that too may DOF were unlocked for this protocol. At the very least, DOF at the shoulder and wrist are having significant effect on Shoulder JPS Measurement Device performance. This may mean that in its current state, the device is giving us more of a sense of full arm proprioception rather than just isolating humeral rotation as intended. This may begin to answer why so much variability was observed in our previous research (study one of this thesis) utilizing this device.

Future attempts at refining the Shoulder JPS Measurement Device should start by finding a method of constraining the wrist as it appeared to be the most variable joint based on partial correlations and collinearity coefficients. The challenge will be to find a method of stabilizing wrist posture that also minimizes the cutaneous feedback from the tissues so as not to provide unintended sensory cues during a joint angle matching task. At such a point that a viable method of wrist locking is implemented, a multiple regression may be performed with the data set involving the locked wrist to observe and reassess if any other DOF's become significantly unique in their collinearity statistics.

4.5. References

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CHAPTER 5.

SUMMARY OF THE THESIS

The objective of the first study of this thesis weas to explore changes to thoracohumeral rotation proprioception and mechanics following induction of a significantly fatiguing stimulus to the cervical extensors of the neck. In a follow up study, this thesis probed potential sources of variability in upper extremity kinematics when performing unconstrained rotation of the humeral using our proprietary shoulder proprioception device. Our hypothesis and findings are summarized as follows:

$H1_0$: Thoracohumeral JPR accuracy **did not** significantly diminish after becoming substantially fatigued.

This finding was inconsistent with the trend seen in the literature for upper limb proprioception to be negatively affected by neck muscle fatigue. We hypothesized that any potential for the emergent increase in absolute error between our fatigue group and our control group to reach significance may have been diminished by the overwhelming variability coincident with an unconstrained upper limb task. The standard deviations we observed for this task were observed to range from ≈75% of the mean score at baseline to ≈85% of the mean score post neck fatigue. Based on these statistics and the calculated ICC's for this data set, sample size calculations formulated by Dai et al. (2013) would suggest that in order to truly test the significance of the interaction between fatigue and control groups with significance set to $p \leq 0.05$ for two-tailed test, this data set would need to be appreciated to a sample size of at least 130 participants (Dai, Charnigo et al. 2013).

There was also no observed trend for variability to significantly increase following neck muscle fatigue, nor was there an observed trend for variability to decrease following repeated task attempts in the control group. In literature, changes in JPR variability are as common a finding from afferent feedback disruption as are changes in absolute error. It could even be theorized that due to the substantial DOFs of our unconstrained JPR task that there would be greater potential for observing increased variability as opposed to increased absolute error. This would coincide with previous research that found undesired changes to proximal limb joint accuracy could be supplanted by coordinated changes in distal limb joint accuracy to maintain accuracy of the end effector (Emery and Cote 2012). Ultimately, there could have been an emergent trend between the fatigue and control groups and increased variability, yet the initial variability in error scores before the induction of neck fatigue was so large compared to our mean error scores, that minute changes to variability failed to reach significance for the increases and decreases that were observed in our respective groups. The results from study two suggest that the significant interaction of wrist deviation on intended measurement of thoracohumeral rotation likely played a substantial part in both inflating our initial variability scores, and reducing the direct impact of altered shoulder proprioception on Shoulder JPS Measurement Device arm displacement. It was not the objective of this thesis, nor is there yet enough emperical published evidence in literature, to deduce the compound effects of multi joint proprioceptive decrements on end effector position. As such, the contributions of wrist ROM and altered sensory feedback to the collective upper limb joints could not be mediated to isolate error attributed separately to the wrist and shoulder.

$H2_{A1}$: The Shoulder JPS Measurement Device significantly reflected unconstrained humeral rotation about the thorax, however wrist deviation was revealed to significantly confound the devices measurement.

Due to the neuroanatomical model of body schema we can truly theorize that altered sensory input from the neck would instigate decrements in wrist proprioception as well as shoulder proprioception. However, since the wrist was determined to have a significant unique variance in determining device measurement, the unknown effects of altered wrist JPS on device rotation likely confounded the significance of our error measurement scores from our acute neck fatigue study. Therefore, while sample size calculations would suggest that an increased data pool might reduce the possibility of a type-2 error, a more pragmatic future direction in this research may be to experiment with methods for constraining the wrist in a way that also minimizes the contribution of additional passive sensory afference. Such research could advocate for beneficial changes to the way participants interact with the Shoulder JPS Measurement Device that would still maintain a minimally constrained approach to quantifying upper limb motion while also significantly reducing the multi-joint variability that is problematic with the current user interface design. With the successful implementation of such an upgrade to the device, future research studies could confidently recollect a more modest sample size such as our study with the assurance that the Shoulder JPS Measurement Device is more accurately representing thoracohumeral rotation.

5.1. CONCLUSION OF THE THESIS

Musculoskeletal demands on the erectors of the cervical spine are ever increasing as learning environments and office workplaces become more technologically integrated and manual labor jobs do little to limit repetitive static posture. With such a persistent endemic of chronic CEM fatigue on the rise, it is important to fully understand the detriments associated with altered body schema as a result of this postural neck exposure. Both studies of this thesis have collectively determined that unconstrained arm rotation is subject to a high degree of inter-trial variability which can make observations regarding sensorimotor changes difficult to detect. This has made it difficult to confirm if there is an interaction of neck muscle fatigue disrupting proprioception of the upper limb. However, kinematic quantification of the shoulder is not always a simple task, and it can be expected that novel approaches bear unique challenges. While our current findings may not suggest a significant interaction, we were able to confirm that unlocked DOFs contributed significantly to the variability observed in study one. This knowledge can be used to refine the proprioception device and improve its application as a reliable and accessible measurement tool to help in future studies investigating changes in shoulder mechanics. Ultimately this may provide a valid and reliable way to measure the effects of altered sensory input to the neck on shoulder joint proprioception.

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APPENDICES

APPENDIX A. MULTIPLE REGRESSION COEFFICIENTS

Table 6.1: Correlation and Collinearity Coefficients of off Upper Limb Predictor Variables in Standardized Multivariate Regression to Shoulder Joint Position Sense Measurement Device Arm Rotation.

Predictor Variable	Zero Order Correlation	Partial Correlation	Collinearity Tolerance	Variance Inflation Factor
Shoulder Humeral Rotation	0.407	0.366	0.739	1.354
Elbow Pronation	0.289	0.120	0.510	1.959
Wrist Deviation	0.170	0.103	0.794	1.259
Elbow Flexion	0.107	-0.096	0.578	1.731
Shoulder Negative Elevation	0.160	-0.049	0.611	1.635
Wrist Pronation	-0.001	-0.036	0.956	1.046
Shoulder Plane of Elevation	0.168	0.018	0.703	1.423
Wrist Flexion	0.048	0.017	0.989	1.011

Table 6.2: Correlations between all Upper Limb Predictor Variables.

Variables	HumRot	ElbPro	WrstDev	ElbFlex	HumNegElv	WrstPro	HumPOE	WrstFlex
HumRot	1.000	.256	-.011	.340	.401	.018	.329	.024
ElbPro	.256	1.000	.352	.578	.380	.161	.216	.033
WrstDev	-.011	.352	1.000	.051	.167	.071	.177	.071
ElbFlex	.340	.578	.051	1.000	.400	.025	.233	-.008
ShldNegElv	.401	.380	.167	.400	1.000	.089	.484	.012
WrstPro	.018	.161	.071	.025	.089	1.000	.117	-.017
HumPOE	.329	.216	.177	.233	.484	.117	1.000	.052
WrstFlex	.024	.033	.071	-.008	.012	-.017	.052	1.000

Table 6.3: Covariances between All Upper Limb Predictor Variables.

Variables	HumRot	ElbPro	WrstDev	ElbFlex	HumNegElv	WrstPro	HumPOE	WrstFlex
HumRot	.	.000	.769	.000	.000	.639	.000	.530
ElbPro	.000	.	.000	.000	.000	.000	.000	.389
WrstDev	.769	.000	.	.187	.000	.067	.000	.068
ElbFlex	.000	.000	.187	.	.000	.519	.000	.827
ShldNegElv	.000	.000	.000	.000	.	.022	.000	.751
WrstPro	.639	.000	.067	.519	.022	.	.002	.666
HumPOE	.000	.000	.000	.000	.000	.002	.	.177
WrstFlex	.530	.389	.068	.827	.751	.666	.177	.

Appendix B Informed Consent – Study 1

Title: The effect of neck muscle fatigue on shoulder joint position sense

Principal Investigator:

Dr. Bernadette Murphy	Associate Dean, Graduate Studies University of Ontario Institute of Technology (UOIT) 2000 Simcoe St. N., Oshawa, ON, L1H 7K4 Phone #: 905.721.8668 Email: Bernadette.Murphy@uoit.ca
Dr. Nicholas La Delfa	Assistant Professor, Faculty of Health Science University of Ontario Institute of Technology (UOIT) 2000 Simcoe St. N., Oshawa, ON, L1H 7K4 Phone #: 905.721.8668 Email: Nicholas.Ladelfa@uoit.ca

Student Lead:

Matthew Russell	Graduate Student, Faculty of Health Sciences University of Ontario Institute of Technology (UOIT) 2000 Simcoe St. N., Oshawa, ON, L1H 7K4 Phone #: 905.706.9446 Email: Matthew.Russell@uoit.net

Supervisory Committee:

Dr. Paul Yielder	Associate Professor, Faculty of Health Sciences University of Ontario Institute of Technology (UOIT) 2000 Simcoe St. N., Oshawa, ON, L1H 7K4 Phone #: 905.721.8668 ext. 2768 Email: paul.yielder@uoit.ca

Rationale:

Proprioception is a term which describes the body's ability to maintain awareness of its limbs and position in 3D space. Proprioception is signaled to the brain by receptors found in tendons and muscles in our body. These receptors measure the length and position of limbs and joints, making us aware of our body's position without having to consciously think about it. Our study will be looking at the ability of a new device to accurately and reliably measure shoulder proprioception.

There are many factors which are known to reduce proprioceptive accuracy. Emerging research suggests that fatigue of neck muscles can impair porprioception of the arm. However, no study as yet to measure exactly how much of an effect that neck muscle fatigue has on decreasing arm proprioception. With this study, we hope to better understand the relationship between neck muscle fatigue and arm proprioception, which can be important to improve many different rehabilitative and preventative measures. In order to effectivley study this relationship, the validity of a new system of shoulder kinematic measurement is necessary.

The **goal** of this study is to test the validity and reproducability of a new, custom built device designed to measure joint angle changes of the arm and compare this device to theh gold-standard kinematic equivalent. This device was designed and built by colleauges in the UOIT engineering faculty.

Information for Participants:

We are seeking healthy participants between 18 and 35 years of age. We are looking for participants who do not have a history of neck or shoulder pain severe enough to have sought medical intervention or taken more than 3 days off work in the past 6 months. Participants must not work in an occupation which requires prolonged reaching or upper limb strain (e.g. power tool use, machine operating, carpentering). Participants must also not have had shoulder reconstructive surgery and they must be able to comfortably sit for one hour consecutively.

We encourage you to read this form thoroughly and ask any questions that you may have. Your participation in this study is entirely voluntary (your choice), and you are free to decline taking part in this study. If you agree to participate, you may withdraw from the study at any time without giving a reason. This will in no way affect your academic progress.

This form outlines the procedures involved in this research, the risks and benefits associated with participation and what you can expect as a participant. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator – researchethics@uoit.ca or 905.721.8668 x. 3693.

Study Procedures:

Prior to the collection date, you will be emailed with the Informed Consent form as well as the Edinburgh Handedness Questionnaire, the Neck Disability Index, and the Chronic Pain Grade Scale which can be filled out ahead of participation or on your scheduled lab collection date. Doing this in advance will be able to confirm your eligibility or ineligibility criteria before requiring them to attend a lab collection date.

You will be required to attend one session, which will take place in the UAB Occupational Neuromechanics and Ergonomics research lab (UAB 357). This session will take approximately 1 hour to complete. You are expected to arrive to the laboratory dressed in, or prepared to change into, athletic bottoms (pants or shorts are both acceptable) and either no shirt (males only) or a loose-fitting shirt that can be rolled up to expose the shoulder area. The Edinburgh Handedness Questionnaire (EHQ), The Neck Disability Index (NDI), the Chronic Pain Scale, and demographic information forms will be filled out at this time unless they have already been completed and returned to the researchers via email prior to the study date.

Next, you will have your height and weight measured, along with your forearm length (Olecranon process to base of the carpals). You will then be instrumented with motion capture markers (1 cm x 1 cm) on their right shoulder, upper arm and forearm which are recorded by the gold-standard kinematic camera system. You will then become familiarized with the shoulder position sense measurement device. These instruments will measure your muscle activity, track your posture and measure your stability respectively.

When the study procedures begin, you will be asked to sit comfortably, with you right arm abducted 90 degrees while you have the shoulder rotation measurement device matched to a posture that is comfortable to your preference. Once you are comfortably matched with the shoulder position sense measurement device and ready to begin the study, you will be blindfolded for the testing process. Once blindfolded, you will have your right arm rotated to a comfortable position within your range of motion, held for 3 seconds, and returned to your original position. Your arm will be moved by the researcher. You will be asked to recreate that same arm position on your own, and without input from the researcher or your own vision, and then asked to return your arm to its original position. This trial will be repeated a total of four times, with five-minute breaks between each trial.

Potential Benefits:

If you decide to participate, you will get to learn more about the research process that occurs which is fundamental to the University. You will also gain a greater awareness of the accuracy of your own upper limb proprioception. You may also learn more about how their muscles and brain interact and how altered patterns of muscle use may perpetuate the chronic pain cycle.

Potential Risks:

Sitting for approximately 1 hour will be necessary for completion of this procedure. However, as a student, this is not outside of what would normally occur in a typical school day and therefore poses no significant risk outside of what may normally encounter.

You may experience anxiety or stress as a result of your responses to the preliminary questionnaires.

The physical risks are that you might get tired of holding your arm up, however all movement is voluntary and can be terminated when you so wishe. The most common adverse reactions would include the possibility of mild discomfort and fatigue of the joints of the upper limb that would not last longer than 24 hours.

Confidentiality:

Identifiers will be removed from all data to maintain confidentiality of the participants. The data will be stored in a locked cabinet at UOIT for seven years from the completion of the study, after which it will be destroyed in accordance with university protocol.

Right to Withdraw:

Your participation in this study is voluntary and you are free to decline without providing a reason. Throughout the research process, you are free to withdraw from participation at any time without repercussion.

Debriefing and Dissemination of Results:

The intent of this research is to improve guidelines. As such, the data for this research may be submitted to scientific conferences and peer reviewed journals for publication. Published data will be coded and no personal identifiers will be included. If you wish to receive an aggregate of the research findings, please check the box at the bottom of this form and provide an email address to receive the results.

Thank You!

Thank you very much for your time and help in making this study possible. If you have any questions concerning the research study, please contact the researcher Matthew Russell at 905.706.9446 or Matthew.Russell@uoit.net. Alternatively, you can contact the principal investigators Dr. Bernadette Murphy at 905.721.8668 or Bernadette.Murphy@uoit.ca, or Dr. Nicholas La Delfa at 905.721.8668 or Nicholas.LaDelfa@uoit.ca.

This study has been approved by the UOIT Research Ethics Board REB 15034 on February 3rd, 2019. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

Sincerely,

Dr. Bernadette Murphy
Dean, Graduate Studies
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4

Dr. Nicholas La Delfa
Assistant Professor, Faculty of Health Science
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Matthew S. Russell
Masters of Health Sciences (M.H.Sc.) Student
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H

Please read the following carefully before signing. If you would like a copy of this consent form for your records, please ask the investigators. ***Received Copy:*** ***YES*** ☐ ***NO*** ☐

I understand that:

- Taking part in this study is voluntary and that I am free to withdraw from participation at any time without giving a reason and that withdrawal will in no way affect my academic process.
- This consent form will be kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- Data collected during the study will be coded, kept in a confidential form and kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- I may withdraw from participation at any time before, during or after the study up to two days following data collection. At which time my data will be included in the study.
- My participation in this study is confidential and that no material which could be used to identify me will be reported.

I have:

- Read and I understand the information provided within this consent form.
- Had the opportunity to ask questions and discuss the study with the investigators and am satisfied with the answers provided.
- Had time to consider whether or not to participate.
- Taken note of who to contact if I experience any adverse events.

I give consent for the data from this study to be used in future research as long as there is no way that I can be identified in this research. **YES** ☐ **NO** ☐

I would like to receive a short report about the outcomes of this study. (If you answer yes, please provide an email)___________________________ **YES** ☐ **NO** ☐

By signing this form, you consent to participate in the study and you indicate that you understand the information provided to you within this document.

______________________	______________________	______________________
Participants Name (Print)	Signature of Participant	Date
______________________	______________________	______________________
Witness' Name (Print)	Signature of Witness	Date

To be signed by the Primary Investigator and/or Student Lead:

I have fully explained the study to the participant to the best of my ability. I have provided ample opportunities for the participant to ask questions and I have provided clear answers. It is my opinion that the participant fully understands the requirements of the study, the potential risks and benefits of the study. The participant has provided voluntary consent and was not coerced into taking part in the study.

________________________________ ______________________

SIGNATURE OF THE INVESTIGATOR/STUDENT LEAD

Appendix C Informed Consent – Study 2

Title: Validation of a Novel Device and Protocol to Assess Uni-Planar Shoulder Proprioception

Principal Investigator:

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Student Lead:

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Supervisory Committee:

Dr. Paul Yielder	Associate Professor, Faculty of Health Sciences University of Ontario Institute of Technology (UOIT) 2000 Simcoe St. N., Oshawa, ON, L1H 7K4 Phone #: 905.721.8668 ext. 2768 Email: paul.yielder@uoit.ca

Rationale:

Proprioception is a term which describes the body's ability to maintain awareness of its limbs and position in 3D space. Proprioception is signaled to the brain by receptors found in tendons and muscles in our body. These receptors measure the length and position of limbs and joints, making us aware of our body's position without having to consciously think about it. Our study will be looking at the ability of a new device to accurately and reliably measure shoulder proprioception.

There are many factors which are known to reduce proprioceptive accuracy. Emerging research suggests that fatigue of neck muscles can impair proprioception of the arm. However, no study has yet measured exactly how much of an effect that neck muscle fatigue has on decreasing arm proprioception. With this study, we hope to better understand the relationship between neck muscle fatigue and arm proprioception, which can be important to improve many different rehabilitative and preventative measures. In order to effectivley study this relationship, the validity of a new system of shoulder kinematic measurement is necessary.

The **goal** of this study is to test the validity and reproducability of a new, custom built device designed to measure joint angle changes of the arm and compare this device to the current gold-standard kinematic equivalent, which is a motion capture camera system that uses infra-red markers. The new device was designed and built by colleagues in the UOIT engineering faculty.

Information for Participants:

We are seeking healthy participants aged between 18 and 35 years of age. We are looking for participants who do not have a history of neck or shoulder pain severe enough to have sought medical intervention or taken more than 3 days off work in the past 6 months. Participants must not work in an occupation which requires prolonged reaching or upper limb strain (e.g. power tool use, machine operating, carpentering). Participants must also not have had shoulder reconstructive surgery and they must be able to comfortably sit for one hour consecutively.

We encourage you to read this form thoroughly and ask any questions that you may have. Your participation in this study is entirely voluntary (your choice), and you are free to decline taking part in this study. If you agree to participate, you may withdraw from the study at any time without giving a reason. This will in no way affect your academic progress.

This form outlines the procedures involved in this research, the risks and benefits associated with participation and what you can expect as a participant. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator – researchethics@uoit.ca or 905.721.8668 x. 3693.

Study Procedures:

Prior to the collection date, you will be emailed with the Informed Consent form as well as the Edinburgh Handedness Questionnaire, the Neck Disability Index, and the Chronic Pain Grade Scale. These forms can be filled out ahead of participation or on your scheduled lab collection date. Doing this in advance will be able to confirm your eligibility or ineligibility criteria before you are required to attend the lab for a data collection date.

You will be required to attend one session, which will take place in the UAB Occupational Neuromechanics and Ergonomics research lab (UAB 357). This session will take approximately 1 hour to complete. You are expected to arrive to the laboratory dressed in, or prepared to change into, athletic bottoms (pants or shorts are both acceptable) and either no shirt (males only) or tank top or a loose-fitting shirt that can be rolled up to expose the shoulder area. The Edinburgh Handedness Questionnaire (EHQ), The Neck Disability Index (NDI), the Chronic Pain Scale, and demographic information forms will be filled out at this time unless they have already been completed and returned to the researchers via email prior to the study date.

Next, you will have your height and weight measured, along with your forearm length from your elbow to your wrist (Olecranon process to base of the carpals). You will then have motion capture markers (1 cm x 1 cm) attached to your right shoulder, upper arm and forearm with two side tape. The motion capture markers emit an infra-red signal which is recorded by the gold-standard kinematic camera system in the lab. You will then become familiarized with the shoulder position sense measurement device. These instruments will measure your muscle activity, track your arm posture and measure your arm stability respectively.

When the study procedures begin, you will be asked to sit comfortably, with your right arm abducted 90 degrees while you have the shoulder rotation measurement device matched to a posture that is comfortable to your preference. Once you are comfortably matched with the shoulder position sense measurement device and ready to begin the study, you will be blindfolded for the testing process. Once blindfolded, you will have your right arm rotated to a comfortable position within your range of motion, held for 3 seconds, and returned to your original position. Your arm will be moved by the researcher. You will be asked to recreate that same arm position on your own, and without input from the researcher or your own vision, and then asked to return your arm to its original position. This trial will be repeated a total of four times, with five-minute breaks between each trial.

Potential Benefits:

If you decide to participate, you will get to learn more about the research process that occurs which is fundamental to the University. You will also gain a greater awareness of the accuracy of your own upper limb proprioception. You may also learn more about how your muscles and brain interact and how altered patterns of muscle use may perpetuate the chronic pain cycle.

Potential Risks:

Sitting for approximately 1 hour will be necessary for completion of this procedure. However, as a student, this is not outside of what would normally occur in a typical school day and therefore poses no significant risk outside of what may normally encounter.

You may experience anxiety or stress as a result of your responses to the preliminary questionnaires.

The physical risks are that you might get tired of holding your arm up, however all movement is voluntary and can be terminated when you so wish. The most common adverse reactions would include the possibility of mild discomfort and fatigue of the joints of the upper limb that would not last longer than 24 hours.

Confidentiality:

Identifiers will be removed from all data to maintain confidentiality of the participants. The data will be stored in a locked cabinet at UOIT for seven years from the completion of the study, after which it will be destroyed in accordance with university protocol.

Right to Withdraw:

Your participation in this study is voluntary and you are free to decline without providing a reason. Throughout the research process, you are free to withdraw from participation at any time without repercussion.

Debriefing and Dissemination of Results:

The intent of this research is to improve guidelines. As such, the data for this research may be submitted to scientific conferences and peer reviewed journals for publication. Published data will be coded and no personal identifiers will be included. If you wish to receive an aggregate of the research findings, please check the box at the bottom of this form and provide an email address to receive the results.

Thank You!

Thank you very much for your time and help in making this study possible. If you have any questions concerning the research study, please contact the researcher Matthew Russell at 905.706.9446 or Matthew.Russell@uoit.net. Alternatively, you can contact the principal investigators Dr. Bernadette Murphy at 905.721.8668 or Bernadette.Murphy@uoit.ca, or Dr. Nicholas La Delfa at 905.721.8668 or Nicholas.LaDelfa@uoit.ca.

This study has been approved by the UOIT Research Ethics Board REB 15033 on December 8th, 2018. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

Sincerely,

Dr. Bernadette Murphy
Professor, Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4

Dr. Nicholas La Delfa
Assistant Professor, Faculty of Health Science
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H

Matthew S. Russell
Masters of Health Sciences (M.H.Sc.) Student
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H

Please read the following carefully before signing. If you would like a copy of this consent form for your records, please ask the investigators. ***Received Copy:*** ***YES*** ☐ ***NO*** ☐

I understand that:

- Taking part in this study is voluntary and that I am free to withdraw from participation at any time without giving a reason and that withdrawal will in no way affect my academic process.
- This consent form will be kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- Data collected during the study will be coded, kept in a confidential form and kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- I may withdraw from participation at any time before, during or after the study up to two days following data collection. At which time my data will be included in the study.
- My participation in this study is confidential and that no material which could be used to identify me will be reported.

I have:

- Read and I understand the information provided within this consent form.
- Had the opportunity to ask questions and discuss the study with the investigators and am satisfied with the answers provided.
- Had time to consider whether or not to participate.
- Taken note of who to contact if I experience any adverse events.

I give consent for the data from this study to be used in future research as long as there is no way that I can be identified in this research. **YES** ☐ **NO** ☐

I would like to receive a short report about the outcomes of this study.
(If you answer yes, please provide an email)___________________________ **YES** ☐ **NO** ☐

By signing this form, you consent to participate in the study and you indicate that you understand the information provided to you within this document.

____________________	____________________	____________________
Participants Name (Print)	Signature of Participant	Date
____________________	____________________	____________________
Witness’ Name (Print)	Signature of Witness	Date

To be signed by the Primary Investigator and/or Student Lead:

I have fully explained the study to the participant to the best of my ability. I have provided ample opportunities for the participant to ask questions and I have provided clear answers. It is my opinion that the participant fully understands the requirements of the study, the potential risks and benefits of the study. The participant has provided voluntary consent and was not coerced into taking part in the study.

____________________________________ ________________________

Signature of the Investigator/Student Lead

APPENDIX D PARTICIPANT INTAKE FORM – STUDY 1

Title: The effect of neck muscle fatigue on shoulder joint position sense.

If you would like a copy of this consent form for your records, please ask the investigators.

Received Copy: *YES* ☐ *NO* ☐

Name: ____ Gender (Circle one): Male Female

Date of Birth: ____________________________ Age: _________ Height: _____________

Email Address: ____

Have you experienced neck or shoulder pain in the last 12 months?	**YES**	☐	**NO**	☐
If you answered yes, has that neck/shoulder pain caused you to seek medical treatment or take more than 3 days off work?	**YES**	☐	**NO**	☐
Have you worked a job over the last 12 months that required you to bend your neck forward for longer than 1 hour?	**YES**	☐	**NO**	☐
Have you worked a job over the last 12 months that required you to constantly reach forward/overhead or operate machinery (eg. Forklift, power tools, desktop computer setup).	**YES**	☐	**NO**	☐
Have you ever had spine or shoulder surgery?	**YES**	☐	**NO**	☐
Are you able to sit for two hours without significant discomfort?	**YES**	☐	**NO**	☐
Would you like to be notified with the aggregate results of the study when they are released in early 2018 via email?	**YES**	☐	**NO**	☐
I hereby give consent for the information contained in this package to be used for the purposes of this study and in future research as long as there is no way that I can be identified**.**	**YES**	☐	**NO**	☐

If you have any questions concerning the research study, please contact the researcher Matthew Russell at 905.706.9446 or Matthew.Russell@uoit.ca. Alternatively, you can contact the principal investigators Dr. Bernadette Murphy at 905.721.8668 or Bernadette.Murphy@uoit.ca and Dr. Nicholas LaDelfa at 905.721.8668 or Nicholas.LaDelfa@uoit.ca.

Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

______________________________ ____________________________

Participant Signature **Date**

Appendix E Participant Intake Form – Study 2

Title: Validation of a Novel Device and Protocol to Assess Uni-Planar Shoulder Proprioception

If you would like a copy of this consent form for your records, please ask the investigators.

Received Copy: *YES* ☐ *NO* ☐

Name: ____ Gender (Circle one): Male Female

Date of Birth: ___________________________ Age: _________ Height: _____________

Email Address: _______________________________________

Have you experienced neck or shoulder pain in the last 12 months?	**YES**	☐	**NO**	☐
If you answered yes, has that neck/shoulder pain caused you to seek medical treatment or take more than 3 days off work?	**YES**	☐	**NO**	☐
Have you worked a job over the last 12 months that required you to bend your neck forward for longer than 1 hour?	**YES**	☐	**NO**	☐
Have you worked a job over the last 12 months that required you to constantly reach forward/overhead or operate machinery (eg. Forklift, power tools, desktop computer setup).	**YES**	☐	**NO**	☐
Have you ever had spine or shoulder surgery?	**YES**	☐	**NO**	☐
Are you able to sit for two hours without significant discomfort?	**YES**	☐	**NO**	☐
Would you like to be notified with the aggregate results of the study when they are released in early 2018 via email?	**YES**	☐	**NO**	☐
I hereby give consent for the information contained in this package to be used for the purposes of this study and in future research as long as there is no way that I can be identified**.**	**YES**	☐	**NO**	☐

If you have any questions concerning the research study, please contact the researcher Matthew Russell at 905.706.9446 or Matthew.Russell@uoit.ca. Alternatively, you can contact the principal investigators Dr. Bernadette Murphy at 905.721.8668 or Bernadette.Murphy@uoit.ca and Dr. Nicholas LaDelfa at 905.721.8668 or Nicholas.LaDelfa@uoit.ca.

Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

_______________________________   _____________________________

Participant Signature   **Date**

APPENDIX F NECK DISABILITY INDEX

Neck Disability Index

Office Use Only
Name ____________________
Date ____________________

This questionnaire has been designed to give us information as to how your neck pain has affected your ability to manage in everyday life. Please answer every section and **mark in each section only the one box that applies to you**. We realise you may consider that two or more statements in any one section relate to you, but please just mark the box that most closely describes your problem.

Section 1: Pain Intensity

☐ I have no pain at the moment
☐ The pain is very mild at the moment
☐ The pain is moderate at the moment
☐ The pain is fairly severe at the moment
☐ The pain is very severe at the moment
☐ The pain is the worst imaginable at the moment

Section 2: Personal Care (Washing, Dressing, etc.)

☐ I can look after myself normally without causing extra pain
☐ I can look after myself normally but it causes extra pain
☐ It is painful to look after myself and I am slow and careful
☐ I need some help but can manage most of my personal care
☐ I need help every day in most aspects of self care
☐ I do not get dressed, I wash with difficulty and stay in bed

Section 3: Lifting

☐ I can lift heavy weights without extra pain
☐ I can lift heavy weights but it gives extra pain
☐ Pain prevents me lifting heavy weights off the floor, but I can manage if they are conveniently placed, for example on a table
☐ Pain prevents me from lifting heavy weights but I can manage light to medium weights if they are conveniently positioned
☐ I can only lift very light weights
☐ I cannot lift or carry anything

Section 4: Reading

☐ I can read as much as I want to with no pain in my neck
☐ I can read as much as I want to with slight pain in my neck
☐ I can read as much as I want with moderate pain in my neck
☐ I can't read as much as I want because of moderate pain in my neck
☐ I can hardly read at all because of severe pain in my neck
☐ I cannot read at all

Section 5: Headaches

☐ I have no headaches at all
☐ I have slight headaches, which come infrequently
☐ I have moderate headaches, which come infrequently
☐ I have moderate headaches, which come frequently
☐ I have severe headaches, which come frequently
☐ I have headaches almost all the time

Section 6: Concentration

☐ I can concentrate fully when I want to with no difficulty
☐ I can concentrate fully when I want to with slight difficulty
☐ I have a fair degree of difficulty in concentrating when I want to
☐ I have a lot of difficulty in concentrating when I want to
☐ I have a great deal of difficulty in concentrating when I want to
☐ I cannot concentrate at all

Section 7: Work

☐ I can do as much work as I want to
☐ I can only do my usual work, but no more
☐ I can do most of my usual work, but no more
☐ I cannot do my usual work
☐ I can hardly do any work at all
☐ I can't do any work at all

Section 8: Driving

☐ I can drive my car without any neck pain
☐ I can drive my car as long as I want with slight pain in my neck
☐ I can drive my car as long as I want with moderate pain in my neck
☐ I can't drive my car as long as I want because of moderate pain in my neck
☐ I can hardly drive at all because of severe pain in my neck
☐ I can't drive my car at all

Section 9: Sleeping

☐ I have no trouble sleeping
☐ My sleep is slightly disturbed (less than 1 hr sleepless)
☐ My sleep is mildly disturbed (1-2 hrs sleepless)
☐ My sleep is moderately disturbed (2-3 hrs sleepless)
☐ My sleep is greatly disturbed (3-5 hrs sleepless)
☐ My sleep is completely disturbed (5-7 hrs sleepless)

Section 10: Recreation

☐ I am able to engage in all my recreation activities with no neck pain at all
☐ I am able to engage in all my recreation activities, with some pain in my neck
☐ I am able to engage in most, but not all of my usual recreation activities because of pain in my neck
☐ I am able to engage in a few of my usual recreation activities because of pain in my neck
☐ I can hardly do any recreation activities because of pain in my neck
☐ I can't do any recreation activities at all

Score: ___/50 Transform to percentage score x 100 = %points

Scoring: For each section the total possible score is 5: if the first statement is marked the section score = 0, if the last statement is marked it = 5. If all ten sections are completed the score is calculated as follows: Example: 16 (total scored)
50 (total possible score) x 100 = 32%

If one section is missed or not applicable the score is calculated: 16 (total scored)
45 (total possible score) x 100 = 35.5%

Minimum Detectable Change (90% confidence): 5 points or 10 %points

NDI developed by: Vernon, H. & Mior, S. (1991). The Neck Disability Index: A study of reliability and validity. Journal of Manipulative and Physiological Therapeutics. 14, 409-415

APPENDIX G EDINBURGH HANDEDNESS INVENTORY

Edinburgh Handedness Inventory

Surname______________________Given Name__________________________

Date of
Birth_________________________Sex________________________________

Please indicate your preferences in the use of hands in the following activities by *putting + in the appropriate column*. Where the preference is so strong that you would never try to use the other hand unless absolutely forces to, *put ++*. If any case you are really indifferent put + in both columns.

Some of the activities require both hands. In these cases the part of the task, or object, for which hand preference is wanted is indicated in brackets.

Please try to answer all the questions, and only leave a blank if you have no experience at all of the object or task.

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

L.Q.	Leave the spaces blank	DECLE

APPENDIX H CHRONIC PAIN GRADE SCALE

Graded Chronic Pain Scale (GCPS)

1. How would you rate your facial pain on a 0 to 10 scale at the present time, that is right now, where 0 is "no pain" and 10 is "pain as bad as could be"?

No pain										Pain as bad as could be
0	1	2	3	4	5	6	7	8	9	10

2. In the past six months, how intense was your worst pain, rated on a 0 to 10 scale where 0 is "no pain" and 10 is "pain as bad as could be"?

No pain										Pain as bad as could be
0	1	2	3	4	5	6	7	8	9	10

3. In the past six months, on the average, how intense was your pain rated on a 0-10 scale where 0 is "no pain" and 10 is "pain as bad as could be"? (That is you usual pain at times you were experiencing pain).)

No pain										Pain as bad as could be
0	1	2	3	4	5	6	7	8	9	10

4. In the past six months, how much has facial pain interfered with your daily activities rated on a 0 to 10 scale where 0 is "no interference' and 10 is "unable to carry on any activities"?

No interference										Unable to carry on any activities
0	1	2	3	4	5	6	7	8	9	10

5. In the past six months, how much has facial pain changed your ability to take part in recreational, social and family activities where 0 is "no change" and 10 is "extreme change"?

No change										Extreme change
0	1	2	3	4	5	6	7	8	9	10

6. In the past six months, how much has facial pain changed your ability to work (including housework) where 0 is "no change" and 10 is "extreme change"?

No change										Extreme change
0	1	2	3	4	5	6	7	8	9	10

7. About how many days in the last six months have you been kept from your usual activities (work, school or housework) because of facial pain?

________ Days

Scoring Criteria for Grading Chronic Pain Severity

Characteristic Pain Intensity is a 0 to 100 score derived from Questions 1 through 3:
Mean (Pain Right Now, Worst Pain, Average Pain) X 10

Disability Score is 0 to 100 score derived from Questions 4 through 6:
Mean (Daily Activities, Social Activities, Work Activities) X 10

Disability Points: Add the indicated points for Disability Days (Question 7) and for Disability Score.

Disability Points

Disability Days (0-180 Days)		Disability Score (0-100)	
0-6 Days	0 Points	0-29	0 Points
7-14 Days	1 Point	30-49	1 Point
15-30 Days	2 Points	50-69	2 Points
31+ Days	3 Points	70+	3 Points

Classification

Grade 0	No TMD pain in prior 6 months
Grade 1 Low Intensity Low Disability	 Characteristic Pain Intensity< 50 < 3 Disability Point
Grade II High Intensity Low Disability	 Characteristic Pain Intensity $\geq$ 50 < 3 Disability Points
Grade III High Disability Moderately Limiting	 3 to 4 Disability Points (Regardless of Characteristic Pain Intensity)
Grade IV High Disability Severely Limiting	 5 to 6 Disability Points (Regardless of Characteristic Pain Intensity)

Von Korff M, Ormel J, Keefe FJ et al. Grading the severity of chronic pain.

Pain 1992; 50:133-149.