Effect of Degrees of Freedom on Effort and Rate of Fatigue Accumulation During the Supine Chest Press

by

Edvin Ghahramanyan

A thesis

presented to the University of Waterloo

in fulfillment of the

thesis requirement for the degree of

Master of Science

in

Kinesiology

Waterloo, Ontario, Canada, 2017

Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

INTRODUCTION

For workplace tool design and exercise equipment design alike, the number of the degrees of freedom to be controlled in a tool or machine has been shown to affect the biological response of the user. Studies show more muscle activation when either the load, or the supporting surface is less stable (i.e. has more degrees of freedom). Despite this, there has been little research on the effect of added degrees of freedom on neuromuscular fatigue, although it may be an intervening variable of interest as fatigue has been shown to increase the incidence of labour accidents leading to injuries at the workplace due to diminished motor control, increased force variability and reduced maximal strength. The purpose of this study is twofold. Firstly, it is to analyze the effect which the demand of controlling additional degrees of freedom has on effort and rate of fatigue accumulation in a strength trained population. Secondly, it is to observe how activation of non prime mover muscles changes with fatigue with different stability requirements, and how prime mover muscle activation changes in response.

METHODS

In this study, the supine chest press exercise was utilized to demonstrate the effect of allowing more degrees of freedom at the hands, on effort and fatigue. A Smith machine was modified to allow uncoupled side-side and coupled back-forward degrees of freedom. Six bench press "modes" were tested, each with varying number of degrees of freedom; four Smith machine bench press modes, barbell bench press and dumbbell bench press.

19 strength trained participants were recruited. 1RM barbell bench press was tested and each participant performed 50%1RM bench press to fatigue on every bench press mode over two sessions. Mean EMG and mean power frequency from every repetition, mean hand and elbow action tremor, load path deviation, mean thoraco-humeral angle were collected.

A mixed effect linear model was used to obtain initial values and rate of change with fatigue. Initial value differences were used to compare effort between modes and rate of change between modes was used to compare rate of fatigue accumulation between modes. Initial and final rate of perceived difficulty were collected and analyzed using a repeated measures ANOVA.

RESULTS

There was a significant main effect on all prime mover muscles' activity for condition ($p<0.01$). Overall, there was greater overall initial muscle activity in modes with more unrestrained degrees of freedom. Most notably, there was a redistribution of muscle stress from elbow extensors to shoulder (horizontal) flexors during modes which had uncoupled side-side degree of freedom unrestrained. For instance, during the Smith machine mode with all degrees of freedom unlocked, there was a 17% increase in pectoralis major activity and a 5% decrease in triceps' activity. This muscle redistribution, which corresponded with the mechanical nature of the task was correlated with perceived difficulty of control.

There was a significant main effect on the number of repetitions completed to failure ($p<0.01$). Only modes which had uncoupled side-side degree of freedom unlocked had a significant effect on fatigue. These modes produced, on average, 5 less repetitions.

Out of initial non-prime mover muscle activation, only biceps showed a general trend towards increasing with more unrestricted degrees of freedom, while shoulder musculature was unchanged in the absence of fatigue. However, with fatigue accumulation, modes which had more degrees of freedom unrestricted generally had greater rate of non-prime mover musculature activation increase, which was also correlated with prime mover muscle activation and prime mover mean power frequency decrease. Additionally, the results showed a trend towards individuals with more strength training experience being able to perform better with the more unstable dumbbell and barbell bench press as compared to less strength trained individuals.

CONCLUSION

Although each degree of freedom altered did not have the same effect, general findings included: as the unrestrained degrees of freedom increased, effort required increased and participants fatigued more rapidly. The findings give insights into the effects of people exerting forces against unstable loads in strength training and occupational settings.

Acknowledgements

First and foremost, I'd like to thank my supervisor Dr. Richard Wells for the opportunity to learn, grow, and become a better problem solver. For his wisdom and guidance through-out this process. My committee members, Dr. Clark Dickerson and Dr. Steven Fischer for challenging me as a student and helping me think more critically about my project. My thinking both as a researcher and as an individual has completely changed by the challenges offered in this program.

I'd like to thank my lab mates and exemplary graduate students Binh Ngo and Marcus Yung for their friendship and academic guidance throughout my graduate experience. Many thanks to Betina Butler and Denise Hay for their outstanding administrative support.

Last but not least, I'd like to show my gratitude to my parents for their endless love, support and encouragement. Thank you.

Table of Contents

List of Figures

List of Tables

I. Introduction

For workplace tool design and exercise equipment design alike, the number of the degrees of freedom to be controlled in a tool or machine has been shown to affect the biological response of the user. This has been shown in both maximal effort tasks and submaximal effort tasks (Kornecki, Kebel, & Siemienski, 2001) (Fischer, Wells, & Dickerson, 2009) (Welsch, Bird, & Mayhew, 2005) (Saeterbakken, Van Den Tillaar, & Fimland, 2011). Degrees of freedom is defined, in this study, as a rigid object's unrestricted translations along X, Y and Z axis and unrestricted rotations about X, Y and Z axis. An "open", "externally unrestricted" or "available" degree of freedom refers to a movement along, or a rotation about any of three axes which is not mechanically restrained externally. Controlling a degree of freedom refers to a person putting forth muscular effort to restrain an externally unrestricted degree of freedom via the neuromuscular system.

In the recent years, a great emphasis has been made in the fitness world on switching from exercise machines to free weight training. The term "real world resistance" has been coined to describe this exercise approach. Many fitness experts are now adapting this approach, promising more "carry-over strength" from the gym to everyday activities such as carrying groceries, picking things off the ground and playing recreational sports. In fact, many fitness magazines now promote exercising with equipment with unstable surfaces and unstable resistance, such as the BOSU ball, TRX suspension trainer, stability cushions, etc. The general belief is that the limbs which make contact with the floor and/or resistance have less "stability" (more unrestricted degrees of freedom) at the point of contact, forcing the musculoskeletal system to create stiffness at all affected joints by co-contracting musculature acting on those joints.

Numerous studies, however, conclude that although there is greater co-activation of the abdominal musculature with training on unstable surfaces (more degrees of freedom of the supporting surface), the great decrease in the amount of maximal force production is not an optimal trade-off for strength athletes (Kohler, Flanagan, & Whiting, 2010) (Anderson & Behm, 2004). On the contrary, unstable surface training may be good for rehabilitative purposes. Bench pressing with dumbbells, which allows six uncoupled degrees of freedom at each hand greatly reduces maximal force production relative to the barbell bench press, while slightly increasing pectoralis major activation (Saeterbakken, Van Den Tillaar, & Fimland, 2011). From a rehabilitative standpoint, this mode of training has merit as lower external loads may be applied on to the joint while the prime mover activation remains sufficient for strength gains. Further, during unstable surface training, abdominal musculature activation has shown to be increased. Spine stiffness, achieved by balanced co-activation of musculature on either side of the spine, has a protective mechanism on the spine by allowing it to bear greater perturbations (McGill, 2007), making unstable surface training great for rehabilitative purposes of the back. This is also evidenced in Sandler et al. (2014), who found that amongst 4610 individuals between the ages 20 and 81, those who exercised with free weights (i.e. dumbbells and barbells) and calisthenics (i.e. body weight exercises), had a lower rate of low back pain than individuals who exercised with exercise machines.

In a workplace setting, estimation of the capability of a specific percentage of a population capable of exerting forces requires comparison to population strength norms. However, if the strength data were determined in situations where a small number of degrees of freedom had to be controlled, their use in “real world” settings where more degrees of freedom need to be controlled, may produce estimates that are misleading.

There has been little research on the effect of added degrees of freedom on neuromuscular fatigue, although it may be an intervening variable of interest. Fatigue has been shown to increase the incidence of labour accidents leading to injuries at the workplace (Kajimoto, 2007) due to diminished motor control, increased force variability and reduced maximal strength (Hammarskjold & Harms-Ringdahl, 1992). As the degrees of freedom increase, the possible mechanisms for injury may be twofold; a) directly, due to a decrease of control because of a lack of external constraints and b) indirectly, due to an accelerated decrease of control due to fatigue caused by the additional demands on the neuromuscular system to control the unwanted degrees of freedom. **Figure 1** shows how the change of the number of unrestricted degrees of freedom and difficulty of control of each unrestricted degree of freedom affects "effort" both directly, and indirectly through increased rate of fatigue.

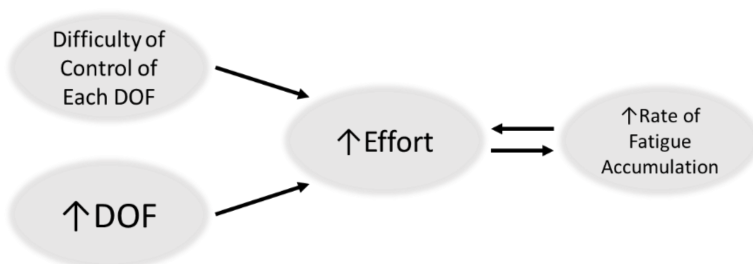


Figure 1. Effect of increased DOF on effort.

Neuromuscular strategy, measured by muscular co-contraction, may also potentially be altered by both increased degrees of freedom, and accelerated fatigue caused by increased degrees of freedom.

There is a mathematically redundant amount of muscle activation patterns to balance joint loads (Cashaback & Cluff, 2014). These activation patterns change based on the neuromuscular system's goals, which have been postulated to include energy efficiency (Anderson & Pandy, 2001), joint stability (Stokes & Gardner-Morse, 2001) (Brown & Potvin,

2005) or muscle stress (Crowninshield & Brand, 1981) (An, Kwak, Chao, & Morrey, 1984) (Hughes, Bean, & Chaffin, 1995). For example, Fischer et al. (2009) have demonstrated a large increase in wrist musculature co-contraction with more degrees of freedom during a screwdriver pushing task, demonstrating that when mechanical constraints are removed, the musculoskeletal system has the added task of creating mechanical stiffness by co-contracting the musculature acting on the joint, therefore giving joint stiffness a higher importance. Kornecki et al. (2001) have shown that as more mechanical degrees of freedom are unrestricted, maximal force production and velocity decreases, further showing that maximal force production is compromised when the load has to be stabilized. Cashaback & Cluff (2014) show that both fatigue, and percentage of maximal force production (intensity) affect these activation patterns, consistent with shifting the system's objective from energy efficiency during a low intensity and/or low fatigue state task, to maximal force production during a high intensity and/or high fatigue state task.

In this study, the supine chest press exercise was utilized to demonstrate the effect of allowing more degrees of freedom at the hands, on effort and fatigue. The Smith machine is a common piece of exercise equipment found in most commercial gyms. It allows for one vertical translational degree of freedom. Typically used for the bench press and shoulder press, this machine takes the task of stabilizing the weight away from the user, and is generally considered a safer alternative for individuals with pre-existing injuries. In this study, the Smith machine was modified to have additional uncoupled (hands able to move independently) medial-lateral and coupled (hands moving together) anterior-posterior translational degrees of freedom which can be locked and unlocked. The medial-lateral translational degree of freedom refers to movement along the horizontal axis orthogonal to the torso's longitudinal axis when a person is lying down.

The anterior-posterior translational degree of freedom refers to movement along the horizontal axis parallel to torso's longitudinal axis a person is lying down.

The independent variable is the degrees of freedom (DOF) of the load which are controlled during the bench press. The machine allows three degrees of freedom and four combinations are used on it. The combinations of unrestricted degrees of freedom on the Smith machine are organized into 4 "modes" of bench press. The combinations of the 4 modes which the Smith machine presents are as followed: 1) 1 DOF: default coupled vertical translational degree of freedom, 2) 2 DOF; coupled vertical + uncoupled medial-lateral translational degrees of freedom, 3) 2 DOF: coupled vertical + coupled anterior-posterior translational degrees of freedom, 4) 3 DOF: coupled vertical + uncoupled medial-lateral + coupled anterior-posterior translational degrees of freedom. It is important to note that the uncoupled rotational degree of freedom about the longitudinal axis of the bar is present when uncoupled translational medial-lateral degree of freedom is unrestricted, but due to the low difficulty of control of this degree of freedom, its effect is not discussed in detail.

In addition, two free weight conditions are used. The combinations which free weights present are as follows 5) 5 DOF: Olympic barbell (coupled vertical + coupled medial-lateral + coupled anterior-posterior translations + coupled rotation about the vertical axis + coupled rotation about the axis parallel to the floor and orthogonal to the barbell) and 6) 5 DOF; dumbbells (same as barbell but all degrees of freedom are uncoupled between the hands).

The purpose of this study is twofold. Firstly, it is to analyze the effect which the demand of controlling additional degrees of freedom has on effort and rate of fatigue accumulation in a strength trained population. Effort is measured with a) initial activation of prime mover and non-prime mover muscles, b) elbow and hand action tremor amplitude and c) initial rate of perceived difficulty score. Fatigue is measured using a) number of repetitions to failure b) rate of mean power frequency decrease of prime mover muscles, c) final rate of perceived difficulty scores, d) rate of change of action tremor e) rate of change of deviation of the bar from a straight path between only bench press modes which allow translation along the anterior-posterior axis (mode 3, 4, 5, 6) and f) rate of change of the thoraco-humeral angle. Secondly, it is to observe how activation of non prime mover muscles changes with fatigue on bench press modes with different stability requirements, and how prime mover muscle activation changes in accordance. **Figure 2** shows how "effort" may be affected by fatigue during tasks with varying degrees of freedom fatigue.

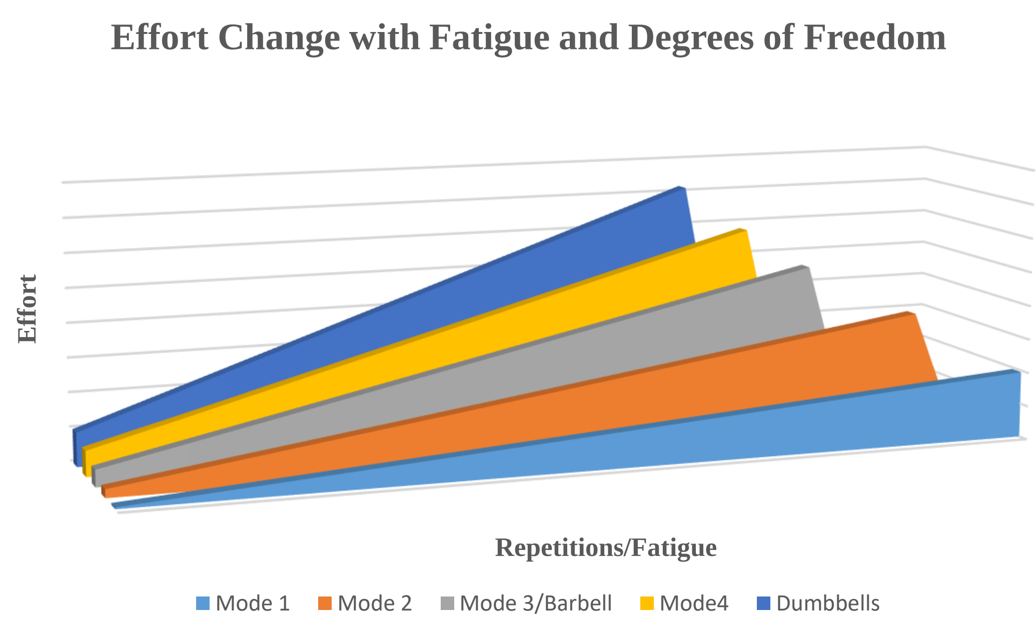


Figure 2. Change in effort with fatigue and degrees of freedom.

1.1 Hypotheses

1. Effort

We hypothesize that as the number of unrestricted degrees of freedom increases with different bench press modes, the effort, as measured by the changes in muscle activation, action tremor and initial ratings of perceived difficulty, will increase.

Rationale

When testing 1RM bench press with dumbbells, barbell and Smith machine, Saeterbakken et al. (2011) found a higher pectoralis major activation during the dumbbell bench press than barbell bench press despite barbell bench press' 1RM being significantly greater than dumbbell bench press' 1RM. They also found a significantly higher pectoralis major activation during the dumbbell bench press than Smith machine bench press, despite the Smith machine bench press' 1RM being significantly higher. This indicates that as degrees of freedom increase, maximal force production decreases possibly due to antagonistic co-contraction, which may produce moments contributing negatively to the required net moment (Brookham, Middlebrook, Grewal, & Dickerson, 2011). This agrees with findings of Kornecki et al. (2001) who found reduced force, velocity and power with increased instability. As more degrees of freedom are unrestricted during the bench press, the counter moment created by antagonistic (i.e. latissimus dorsi) co-contraction may require agonist muscles to produce greater internal joint moment to produce same external force.

2. Fatigue and Muscle Activation:

We hypothesize that as the number of unrestricted degrees of freedom increases with every bench press mode, accumulation of fatigue, as measured by the number of repetitions completed to failure, rate of change of mean power frequency of prime mover muscles, final rating of perceived difficulty, rate of action tremor amplitude increase per repetition, bar path deviation and thoraco-humeral angle, will increase. In addition, we hypothesize that co-contraction of antagonistic musculature will increase at a faster rate with fatigue, resulting in greater activation, and quicker fatigue of prime mover muscles.

Rationale

As previously discussed, unrestricting more degrees of freedom may cause greater prime mover activation, greater non-prime mover co-contraction and reduced maximal force production. This may be due to the fact that when the load is less stable (i.e. has more unrestricted degrees of freedom), joint stiffness, created via agonist-antagonist co-contraction, takes priority over maximal force production. Increased muscular activity over time may cause greater fatigue accumulation in bench press modes with more unrestricted degrees of freedom.

Duffey & Challis (2007) show a very similar bar path between the last repetition of a fatigue protocol and 1RM test, both of which, differ from the first repetition of the fatigue protocol. This may be indication that both high intensity and high fatigue conditions cause the same muscle activation patterns, as reflected by the bar path. If more unrestricted degrees of freedom will cause more overall fatigue, bar path may change accordingly.

To the extent of our literature review, no studies have looked at the torso-humerus angle, but based on observation, “flaring elbows” is a common phenomenon seen in the barbell bench press when fatigue accumulates. Further, Duffey & Challis (2007) have observed a change in bar path with accumulation of fatigue, where the bar starts at nipple level at the bottom of the movement and ends up at shoulder level, mimicking a reverse C shape in the sagittal plane. It may be that when the elbows “flare out” during the supine bench press, the moment arm in torso’s longitudinal axis between the point of load (elbow) and pivot point (shoulder) decreases, turning the exercise from a shoulder flexion task to horizontal shoulder flexion task, and redistributing stress between the pectoralis major, anterior deltoid, and triceps.

II. Review of Relevant Literature

2.1 The concept of degrees of freedom

In mechanics in general, the six degrees of freedom refer to positional change of a rigid body in three dimensional space. More specifically, the degrees of freedom are divided into three translational degrees of freedom along the three axes and three rotational degrees of freedom about the three axes. In mechanics, anterior-posterior, lateral-lateral and superior-inferior degrees of freedom are referred to as surge, sway, and heave, respectively. Rotation about those axes are referred to as roll, pitch and yaw (McCormick, 2010).

In biomechanics, this concept has been used extensively to describe movement of joints within the body (Siegler, Lapointe, Nobilinit, & Berman, 1996) (Cheze, Dumas, Comtet, & Rumelhart, 2011) (Mokhtarzadeha, et al., 2014). In strength training, however, this terminology isn't typically used to indicate how much freedom of movement the resistance allows. Many exercise machines are made to replicate free weight dumbbell and barbell exercises, but they don't require the user to restrict all unwanted translations and rotations of the resistance. Usually, these machines allow for only one degree of freedom. Although strength gains and muscle hypertrophy will occur with both machines and free weights when the principle of progressive overload is applied (Beachle & Earle, 2008), many argue that this training isn't optimal for sports, as most sports require a high demand on dynamic stability (Behm & Anderson, 2006) because when movement isn't constrained by the external environment, as is the case in most sports, the neuromuscular system has the added task of providing stability at all joints (Perreault, Chen, Trumbower, & Lewis, 2008) through co-contraction during movement.

The Smith machine is a common piece of exercise equipment which replicates many compound barbell exercises. It takes the task of stabilizing the weight away from the user by allowing the bar to only have one translational (vertical) degree of freedom. Although studies have assessed how maximal and submaximal strength compares between tasks which allow for 1 degree of freedom (e.g. linear motion in a chest press machine) and tasks which allow 6 coupled or uncoupled unrestricted degrees of freedom (i.e. dumbbell or barbell bench press) (Lyons, McLester, Arnett, & Thoma, 2010), (Cotterman, Darby, & Skelly, 2005), (Simpson, Rozeneck, Garhammer, Lacourse, & Storer, 1997), (Willardson & Bressel, 2004), there are still conflicting findings in the literature regarding the advantages of either (Arandjelovic, 2012). Despite extensive research on the differences between free weights and machines, there remains a gap in the literature in understanding how unrestricting more degrees of freedom affects the rate of fatigue accumulation.

2.2 Force Production

It is speculated that machine assisted exercises allow for more weight to be lifted due to the lack of need to stabilize (i.e. control unwanted degrees of freedom) the weight, and thus, allow for more force to be applied along the linear degree of freedom which the machine allows (Lyons, McLester, Arnett, & Thoma, 2010). Conversely, it may be that when the resistance's degrees of freedom are restricted, an individual may be forced to push the resistance through a path which is non-optimal for the task.

The Smith machine's bench press 1RM is reported to be anywhere between 63% and 83% of barbell bench press' 1RM (Welsch, Bird, & Mayhew, 2005) (Saeterbakken, Van Den Tillaar, & Fimland, 2011). Cutterman et al. (2005) showed a 16% greater 1RM with the barbell bench press than with the one translational degree of freedom Smith machine bench press. In accordance to their findings, the natural S or reverse C shaped sagittal bar path described in Stone (1986) is restricted to a straight line during a Smith machine bench press and thus, they conclude that muscle use and force production is limited (Cotterman, Darby, & Skelly, 2005). Although optimal bar path may allow for most force production, unfamiliarity with the exercise may have an effect on maximal force production with barbell and Smith machine. In the same study, individuals with little squatting experience have shown a higher 1RM during Smith machine squat than during the barbell squat. On the contrary, individuals who have extensive experience with the free weight squat were able to squat more weight with a barbell as compared to a Smith machine. Further, these individuals reported "awkwardness" and felt unease during the Smith machine squat. This may be attributed to restricted motion in the transverse plane during the Smith machine squat, as experienced subjects were unable to recall a familiar motor skill and were forced to alter their movement pattern and muscle recruitment. Madsen and McLaughlin (1984) observed a much accentuated reverse C shape bar path strategy among top level national and international athletes, as opposed to beginners. **This is indicative that although the optimal bath path allows for more optimal force production in trained athletes, beginners may be able to produce more force along a fixed path due to unfamiliarity with the optimal bar path and lack of need to stabilize the load.**

2.3 Activation pattern

It is important to note that there are a wide variety of activation patterns which may be used to complete a task (Stokes & Gardner-Morse, 2001). The activation pattern and degree of co-contraction for task completion will vary depending on the strategy employed by the neuromuscular system (Cashaback & Cluff, 2014), which may include energy efficiency (Anderson & Pandy, 2001), joint stability (Stokes & Gardner-Morse, 2001) (Brown & Potvin,

2005) or muscle stress (Crowninshield & Brand, 1981) (An, Kwak, Chao, & Morrey, 1984) (Hughes, Bean, & Chaffin, 1995). The bar path, whether it is straight or deviated from a straight line, may be an indicator of neuromuscular strategy. Duffey & Challis (2007) found that during the bench press repetitions-to-failure test with 75%1RM, the difference of maximal and mean deviation of the bar from a straight line was much greater between the first repetition and last repetition and between the first repetition and the 1RM test for all subjects, with no significant difference between the last repetition and the 1RM test. Tillaar and Seaterbakken (2014) found that the bar mean bar velocity change from repetition 1 to repetition 6 was identical to the mean bar velocity change from 90% to 100% of 1RM (Gonzalez-Badillo & Sanchez-Medina, 2010). These findings indicate that the last repetition during a fatigue test resembles a 1RM test, which may be indicative that **when maximal force production objective (during the last repetition and 1RM test) takes priority over energy efficiency objective (during the first repetition), a different neuromuscular strategy, as reflected by bath path and mean bar velocity, is adopted.**

Although the movements may be similar, the number of the unrestricted degrees of freedom may also alter the activation patterns (Lyons, McLester, Arnett, & Thoma, 2010), possibly due to the joint stiffness objective having more or less weight. Schick et al. (2010) and McGaw and Friday (1994), found a significantly greater medial deltoid activation during a barbell bench press, as opposed to a Smith machine bench press, despite Smith machine's 1RM being significantly lower (Welsch, Bird, & Mayhew, 2005) (Saeterbakken, Van Den Tillaar, & Fimland, 2011) (Cotterman, Darby, & Skelly, 2005). Medial deltoid may act as a shoulder stabilizer, which is activated when joint stiffness becomes a priority. Further, the difference between medial and anterior deltoid activation was shown to be greater at lower intensities

(60%1RM) and lower at higher intensities (80%1RM), indicating that **during lighter loads, less co-contraction is present** (McCaw & Friday, 1994). Saeterbakken et al. (2011) found a slightly higher pectoralis major activation during the dumbbell bench press than during the barbell bench press despite the 1RM on the dumbbell bench press being significantly lower than the barbell bench press. There was also a significantly higher pectoralis major activation in dumbbell bench press than Smith machine bench press during the eccentric phase, even though Smith machine's 1RM bench press was significantly higher. Despite significantly less force production, prime mover activation may be greater when more degrees of freedom are introduced during the supine bench press. **This may be indicative that as more unrestricted degrees of freedom are introduced, joint stability takes priority over maximal force production. This may be achieved through co-contraction of muscles which create a counter moment to the desired net moment and thus, decrease maximal force production.**

This co-contraction is also seen in Fischer et al. (2009). During a 50N static isometric pushing task, the effect of increasing unrestricted degrees of freedom at the endpoint of a screwdriver and D-shaped handle on activation of the arm and shoulder girdle musculature and rate of perceived control score was examined. The participants were standing upright with the elbow bent at 90°. The pushing tasks involved a completely stable pushing task on rigidly fixed handle (0 DOF), a pushing task with unrestricted horizontal and/or vertical translation at the screwdriver/D-handle tip (1 and 2 translation DOF), a pushing task with unrestricted rotation about horizontal and vertical axis at the screwdriver/D-handle tip (2 rotation DOF), and a pushing task with unrestricted translation and rotation about horizontal and vertical axis at the screwdriver/D-handle tip (4 DOF). Subjectively, the rate of perceived exertion scores showed a 6 fold increased difficulty perception with 4 unrestricted degrees of freedom as compared to the stable pushing task. When the rotational degrees of freedom were introduced, seven of eight measured forearm muscles showed 0-37% increase in activation from baseline (i.e. stable pushing task). Similar results were seen for all pushing tasks; **as the allowed degrees of freedom increased, forearm muscle activation increased relative to baseline. It may be concluded that as the mechanical constraints are removed, the musculoskeletal system has to control the motion which was previously constrained externally.** This may be accomplished by creating stiffness at the arm and the wrist by co-contracting wrist flexor and extensor musculature. This concept is demonstrated Hickok et al. (2014), who looked at the "effort ratio", which is a value of the tangential force applied by the hand to the screwdriver divided by the axial force applied to the screwdriver, between a Phillips screwdriver bit, a straight screwdriver bit and an ECX^{TM} screwdriver, which has components of both screwdriver bits. A higher "effort ratio" indicates more efficiency. Their findings show that the Philips screwdriver bit, which has more constraints had a greater "effort ratio", which was highly influenced by a reduced axial force, as opposed to a greater tangential force. In the case of screwdriver heads, these results indicate more mechanical constraints at the point of contact between the screwdriver and screw head reduce the requirement to create stiffness.

During a seated horizontal maximal effort static push with unrestricted degrees of freedom varying from 0 (stable), to 1 (horizontal translation) to 2 (horizontal and vertical translations) to 3 (horizontal, vertical translations, rotation about the longitudinal axis of forearm), Kornecki et al. (2001) found, amongst other things, that maximal force, velocity and power produced against an external static object is greatly reduced when the muscular system has to restrict more degrees of freedom (Kornecki, Kebel, & Siemienski, Muscular co-operation during joint stabilization, as reflected by EMG, 2001).

As mechanical constrains are removed, the musculoskeletal system has to control the motion previously constrained externally. As the unrestricted degrees of freedom increase, co-contraction of musculature acting on the working joints is increased and maximal force production is reduced.

2.4 Effects of Fatigue

When a limb faced with a task of force production against resistance, the nervous system is faced with a complex task of choosing specific muscle activation patterns (Cashaback & Cluff, 2014). This task of optimizing muscle forces and muscle activation patterns is mapped based on one or a combination of several criteria and objectives, whether it be joint stiffness, caused by antagonistic co-contraction, or energy conservation (Anderson & Pandy, 2001), achieved by minimizing muscle stress (Cashaback & Cluff, 2014). There is strong evidence that muscular activation used to perform a movement may differ based on various factors. Fatigue is amongst the main contributors for altering muscle activation. Psek & Cafarelli, (1993), and Reeves et al., (2008) show a strong positive correlation between a) muscle fatigue and antagonistic co-contraction, and between b) muscular co-contraction and force variability. Cashaback & Cluff, (2014), tested to confirm how the musculature acting on the elbow joint changes the main objective between energy efficiency (minimizing muscle stress) and joint stability (maximizing joint stiffness) throughout a 40%, 70% and 100% of maximum elbow flexion moment fatigue task by formulating a cost function with 2 previously mentioned competing objectives. Their calculated weighed objective (w), on a scale of 0 to 1, with 0 representing complete energy conservation objective and 1 representing complete joint stiffness objective, was strongly correlated with co-contraction. Their results show a strong positive correlation between a) fatigue and objective weighing in 70% and 100% trials and b) between contraction intensity and objective weighing. **Both, increased contraction intensity and fatigue, cause co-contraction of the antagonistic musculature. This is commonly attributed to objective weighing change from energy conservation in the initial stages to maximizing joint stiffness towards the end of a fatigue trial.**

Along with antagonistic co-contraction and altered activation observed through EMG data, this change in muscle recruitment optimization objective may also alter kinematic movement. It is evident from observation that as an individual becomes fatigued, their movement patterns change. It may then be hypothesized optimization objective changes not only muscle activation, but also kinematic movement as the nervous system seeks alternative strategies to complete the task. During a repetitions-to-failure set of bench press with 75%1RM, Duffey and Challis (2007) compared how the bar kinematics compare between the first repetition of the set, the last repetition of the set and the 1 RM test. a) Mean bar position, which is the mean distance of the bar from the shoulder joint through-out one repetition, b) path length ratio (PLR), which is the ratio of the straight vertical path to actual bar path, and c) bar path deviation (BPD), which is the ratio of straight vertical path to the distance of greatest orthogonal deviation, were measured during the first repetition, last repetition and 1RM test. Their findings show a very significant trend. For all three measurements, there was a significant difference between the first and last repetition and between the first repetition and the 1RM test. There is, however, not a significant difference between the last repetition and the 1RM test for all three measures. This is indicative of several things; firstly, **as fatigue increases, the subjects decreased the mean bar position, decreasing the moment arm between the bar and the shoulder in the sagittal plane.** This was a similar strategy adapted in both the subjects in this study during the 1RM testing, and by top national and world level athletes (Madsen & McLaughlin, 1984). Secondly, the apparently likely strategy of moving the bar in a straight line was not utilized as the bar path had a much greater deviation from a straight line in both the last repetition and 1RM test alike. This is indicative that **when the strategy changes from maximizing energy efficiency during the first repetition with 75%1RM, to maximizing force production during the last repetition and the 1RM test, similar strategies, which are reflected by the kinematic data, are adapted in both cases.**

2.5 Tremor

Tremor, often measured at the hands and arms, is involuntary shaking, most commonly involving the upper limbs and head. As defined by Deuschl et al. (1998), it is a visible and persistent bilateral, largely symmetric postural or kinetic tremor involving the hands and forearms which may or may not include head tremor in the absence of abnormal postures. Most commonly measured “postural tremor”, which is a sub-category of action tremor, is a term coined to describe postural wobble when no external resistance is applied to the segment of interest (Rehman, 2009). “Essential tremor”, which is clinical in nature, has been studied as far back as 1930s. In 1981, David Mardsen first established tremor to not be a single entity, and classified the severity of tremor into four types, varying from non-pathological, enhanced physiological tremor, to severe pathological essential tremor. The condition may vary from mild postural tremor to disabling tremor (Marsden, Obeso, & Rothwell, 1983). In healthy individuals, change in tremor amplitude at specific bandwidths has been shown to have a correlation with fatigue (Furness & Jessop, 1977) (Yung, Bigelow, Hastings, & Wells, 2014) (Lippold, 2008). Generally, non-clinical, “physiological tremor” has a power spike in the 6-12Hz bandwidth (Lakie, et al., 2015) (Furness & Jessop, 1977) (Yung, Bigelow, Hastings, & Wells, 2014), with nearly 10% of clinically healthy individuals exhibiting a visible 8-12Hz tremor in a non-fatigued state (Elble R. , 2003) (Pitman Medical Ltd., 1981), which may be a function of mechanical joint stiffness created by muscles, and inertia at the joint (Elble R. , What is essential tremor?, 2013).

Tremor, observed at the limbs, has mechanical and muscular mechanisms which include: 1) oscillations in muscular force, 2) mechanical resonance of a limb caused by limb inertia and elasticity of the muscles acting on the joint, and 3) cardiac pulse (Pitman Medical Ltd., 1981). To test the effect of the first mechanism strain gauges are used, as they eliminate the mechanical resonance and nullify cardiac impulse tremor, isolating the tremor caused by high frequency muscular force variance. Findings from Furness et al. (1977), show that tremor, resulting from a short bout of intense effort (2 min until complete exhaustion) is governed by central nervous fatigue as opposed to cellular muscle fatigue (lactic acid accumulation, ATP depletion, etc.) or peripheral neuromuscular junction fatigue. To test this, they measured the variance of force applied superiorly against a strain gauge by the third distal phalange before a 2 minute test to complete exhaustion and after 2 minute test to complete exhaustion. Participants were asked to apply a constant 0.5N of force. All other joints in the body were fixated to eliminate contamination of signal. The tremor in the 1HZ – 15Hz had a large increase post trial, with no sign of decline 40 minutes after the test. To test whether tremor may be a result of global hormonal changes, such as adrenalin secretion or intersegmental irradiation of neural activity (Furness & Jessop, 1977), the tremor of the same phalange of the non-fatigued hand was measured post-fatigue trial, with no observed changes.

To test whether the tremor is a result of neurological fatigue as opposed to cellular muscle fatigue, the nerves innervating the third phalange were electrically stimulated with a 30Hz stimulus with fitting magnitude to reproduce the same force output as in the fatigue test.

No change in tremor was observed post trial, indicating that **tremor, induced by a short burst of intense activity, is a measure of neural fatigue as opposed to cellular muscle fatigue.**

To test whether this tremor is a result of central nervous system fatigue, or peripheral nervous system fatigue or neuromuscular junction fatigue, two sphygmomanometers were inflated and cuffed around both of the subjects' forearms for 20 minutes, until no pain and temperature was felt in either hands, which was indicative that the nerves distal to the sphygmomanometers were blocked. Thereafter, the subject was asked to apply maximal effort at abducting one of the hands' third phalanges against resistance. Although effort was applied, no movement happened at the finger. After the sphygmomanometers were removed, tremor from both hands was observed using a strain gauge. The finger which the subject put effort into showed a significant increase in tremor in the reported 1Hz and 12Hz frequencies. No increase in tremor in the resting hand was observed, indicating that the effort to contract the muscle, not the nerve blocking alone, induced tremor. Although very little neural activity happened at the hand, the "effort" from the central nervous system caused an increase in tremor. **These findings indicate that during a 2 minute bout of intense effort, central fatigue, as opposed to peripheral neuromuscular junction fatigue is the causation tremor.**

When measuring tremor, graphic activity of tremor can be observed and evaluated subjectively by examining hand writing and pattern drawing. These methods, however, are not easily standardized across subjects (Hess & Pullman, 2012). There is a variety of published scales in the literature, including the most commonly used 5 point Fahn-Tolosa-Marin Tremor Rating Scale (Fahn & Tolosa, 1993) and The Essential Tremor Rating Assessment Scale (TETRAS) (Elble, et al., 2014). Although inter-rater reliability, measured by Kappa statistic, is high in various rating scales (Stacy, Elble, Ondo, Wu, & Hulihan, 2007) (Louis, Ford, &

Bismuth, 1998) (Elble R. , et al., 2012), they may not capture minor abnormalities and subtle changes in tremor (Christopher & Seth, 2012), which may be caused by fatigue. To acquire more precise tremor data, accelerometry, electromyography and other signals such as gyroscopy and force are used (Christopher & Seth, 2012). For measurement of muscular tremor, strain gauges, such as hand dynamometers are used, and for measurement of mechanical tremor (i.e. postural wobble), accelerometers, firmly attached to a limb are used (Pitman Medical Ltd., 1981). When using accelerometers, the tremor is assessed in frequency and amplitude domains. In the frequency domain, the Nyquist theorem may be followed to determine the lowest frequency which would allow the entire signal to be captured. In the amplitude domain, changes from pre-task to post-task may be indicative of neurological fatigue accumulation (Yung, Bigelow, Hastings, & Wells, 2014) (Furness & Jessop, 1977) (Lippold, 2008). To date, there are very few publications assessing changes in non-clinical, physiological tremor amplitude as a result of fatigue. Further, to the extent of our literature review, no studies look at action tremor in a healthy population during performance of a dynamic task. Most studies looking at action tremor deal with subjects with neurological pathologies such as Parkinson's disease

When developing a battery of usable field tests for measurement of accumulated neuromuscular fatigue of plumbers within a workday and recovery between workdays, Yung, et al. (2014) have implemented, among other tests, a measure of physiological and postural tremor (i.e. action tremor). To measure physiological tremor, a tri-axial accelerometer was firmly mounted on the dorsal side of the third metatarsal bone, with the arm supported and the hand hanging freely. To measure postural tremor, a tri-axial accelerometer was attached to a dowel and the subjects were asked to point at target at approximately elbow height. Accelerometer data was recorded for 15 seconds. As hypothesized, there was a general increase in tremor, both within each workday, and throughout the week, with significant increase between day 1 and 3, 4 in physiological tremor and day 1 and 4, 5 in postural tremor (Yung, Bigelow, Hastings, & Wells, 2014).

In addition to neuromuscular fatigue, cognitive fatigue may also contribute to increase in tremor amplitude. Slack, et al. (2009) compared pre-operative tremor to post-operative tremor in 10 head and neck surgery consultants. They found that compared to normal day's desk work, days where surgeons completed surgery increased tremor 8.4 fold, with a strong correlation between surgery time and tremor amplitude increase. Findings from Yung, et at. (2014) and Slack, et al. (2009) are indicative that **tremor increase may be a result of short burst of intense activity, long periods of sustained work and cognitive fatigue. In the literature published to date, there is strong evidence that tremor may be a reliable tool for measurement of fatigue.**

2.6 EMG Measures

Electromyography (EMG) is a common method used to measure muscle activation by detecting neural drive into the muscle. Numerous papers implement this tool to examine the extent to which certain muscles are activated and how their activation and fatigue levels change with treatment (Jung & Cho, 2015) (Saito & Akima, 2015) (Castelein, Cools, Parlevliet, & Cagnie, 2015) (Petrofsky, 1979).

EMG signal is examined in the frequency and amplitude domains. Amplitude may be used to assess both muscle fatigue and muscle activity. Milner-Brown and Stein (1975), Edwards and Lippold (1956) and Bigland and Lippold (1954) have all shown a linear relationship between muscle tension during brief isometric contractions and, both, integrated amplitude of EMG and root mean square amplitude of EMG. On the contrary, Vredenbregt and Rau (1973), and Zuniga and Simons (1969) have found non-linear relationship between EMG amplitude and isometric tension, with more amplitude required per unit of force as the muscle approaches maximal tension capacity. Caution must be taken, however, when comparing activation during a fatigue trial, as fatigue also increases magnitude of EMG (Lippold O. , 1960) (DeVries, 1968) due to increased neural drive and additional motor unit recruitment acting to compensate for cellular muscle fatigue.

Frequency spectrum analysis may provide a reliable measure of muscular fatigue. Fatigue induced power shift to lower frequencies was first observed by Piper, (1912) (Petrofsky J. , 1979). This downshift has since been well documented, confirmed and used in literature as a fairly reliable method of detecting fatigue during isometric contractions (Winter, 2009). Petrofsky, Dahms (1975), have shown that center frequency (the frequency which divides all frequencies into two equal parts when power is arranged from lowest to highest), also referred to as median power frequency, is not affected by the isometric tension developed by the muscle during brief isometric contractions as decreased propagation velocity of action potentials along muscle fibers (Lindstrom, Magnusson, & Petersén, 1970), due to metabolite accumulation, may be amongst the main contributors to the downshift of median power frequency.

Co-activation index has been used in numerous papers to evaluate the ratio of activation of agonist to antagonist muscles (Busse, Wiles, & van Deursen, 2006) (Ervilha, Arendt-Nielsen, Duarte, & Graven-Nielsen, 2004) (Benjuya, Melzer, & Kaplanski, 2004) (Cashaback & Cluff, 2014) (Brookham & Dickerson, 2014) (Brookham, Middlebrook, Grewal, & Dickerson, 2011). There are numerous methods of calculating the co-activation index. Cashaback & Cluff (2014) used the following formula to quantify the ratio of activation between elbow flexor and extensor musculature;

$$CI = \frac{\int_{t2}^{t1} sEMG_{ext}(t)dt}{\int_{t2}^{t1} [sEMG_{flex} + sEMG_{ext}](t)dt} * 100 \qquad (1)$$

where $sEMG_{ext}$ and $sEMG_{flex}$ represent the processed, normalized activity of elbow extensor and flexors respectively and t_1 and t_2 are the start and end of each time window respectively.

There are more simplified way to measure co-activation index. Formulas presented in Ervilha, et al. (2012) were compared to test the reliability of the estimation of muscle co-activation;

$$CI = 2 * \frac{EMG_{ANT}}{EMG_{AG} + EMG_{ANT}} * 100 \qquad (2) \qquad\qquad CI = \frac{EMG_{ANT}}{EMG_{AG}} * 100 \qquad (3)$$

where EMG_{ant} is the normalized amplitude of the muscle which contributes positively to the desired net moment at the joint, and EMG_{ag} is the normalized amplitude which contributes negatively to the desired net torque at the joint.

When measuring the accuracy of two co-activation index methods, Ervilha et al. (2012) found during a 25%, 50%, 75% and 100% maximal effort isometric elbow flexion, with no external resistance (maximum simultaneous bicep and tricep tension, formula (2), yielded a number closest to 100, indicating this to be the more accurate method of co-activation evaluation. Further, they found that using normalized percentage of co-activation yielded more accurate results than using absolute voltage of EMG. **Co-activation may be reliably measured using the formulas presented above. During the bench press task, the co-activation index may be used to show the degree of antagonist activation as degrees of freedom increase. Further, the bench press exercise implements numerous "prime mover" muscles. The co-**

activation index may be used to measure the degree to which the muscle activation between three prime movers, mainly the pectoralis major, anterior deltoid and triceps brachii, changes with fatigue with various combination of degrees of freedom.

III. Methodology

3.1 Modified Smith machine

A Smith machine, commonly found in most commercial gyms, is an exercise piece of equipment which allows the bar to have translational movement along the vertical axis. The Smith machine bar is attached to two tubes with bearings on both ends, which slide along two slightly angled (9°) vertical bars. For our study, a Smith machine has been modified to allow coupled anterior-posterior translation and uncoupled medial-lateral translations, both of which, both of which can be locked and unlocked.

Figure 3. Modified Smith machine allowing for coupled anterior-posterior translation and uncoupled medial-lateral translation.

To allow for uncoupled medial-lateral translation, the Smith machine bar has been replaced with a polished, hardened steel rod. Two bearing tubes, with a 1.25” diameter, have then been welded together and slid on to the steel rod to act as handles. To lock and unlock this translation, two plastic sleeves, with a set screw going through them, have been placed on either side of both handles. When tightened, the plastic sleeves immobilize the handles, restricting medial-lateral translation. The steel rod, on which the bearing handles slide, has been fit into the Smith machine's plate sleeves, where the weight plates are loaded. **Figure 4** demonstrates the handles. Plate sleeves can be seen on either side of the bar in **Figure 3**.

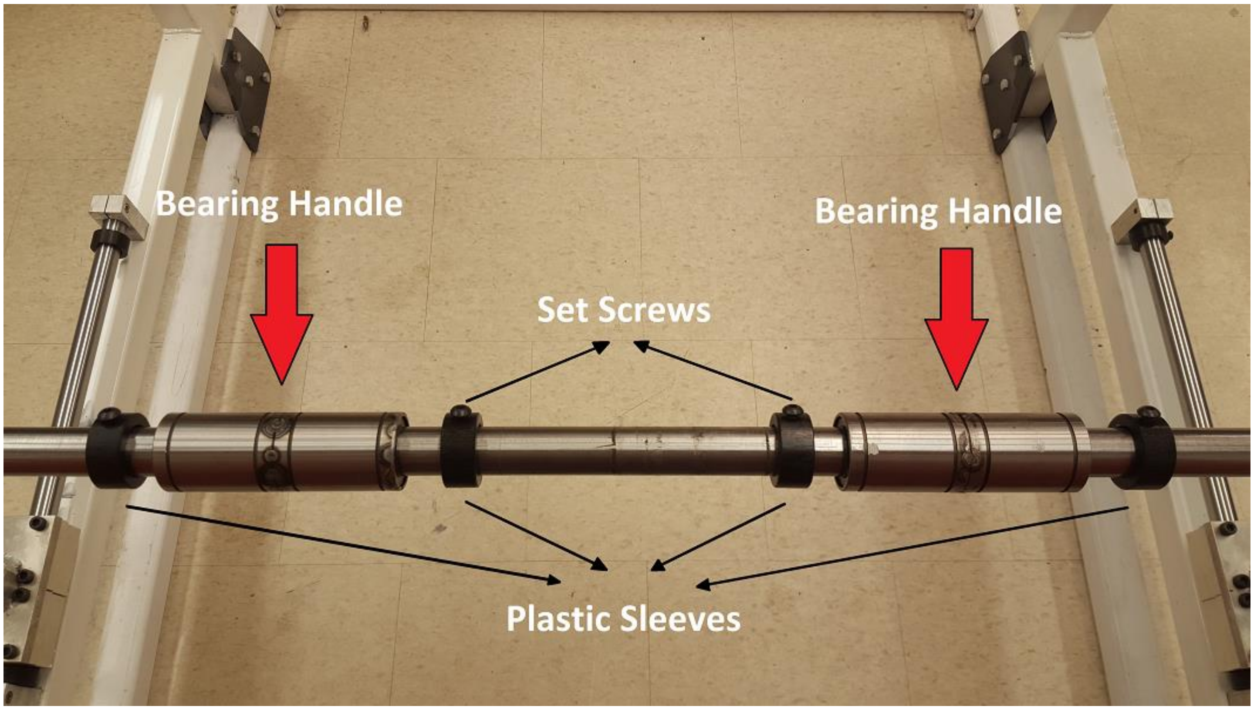


Figure 4. Uncoupled medial-lateral translation is allowed by bearing handles, which slide freely along the horizontal bar. This degree of freedom can be restricted by tightening the set screws on either side each handle, immobilizing the handles.

To allow translation in the anterior-posterior axis, vertical bars have been attached on to bearing blocks on both superior and inferior edges. The bearing blocks slide along hardened steel rods which are attached at the top and the bottom of the Smith machine on both sides. This translation can be restricted by tightening the set screws on the plastic sleeves attached behind the bearing blocks on the horizontal rods.

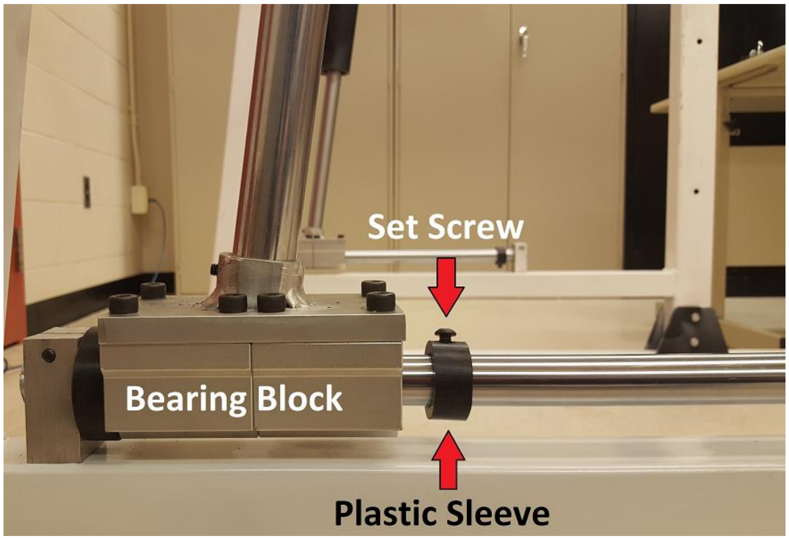


Figure 5. Bottom bearing block slides along the horizontal rod, giving the bar anterior-posterior translation. This degree of freedom can be restricted by tightening the set screws on the plastic sleeves.

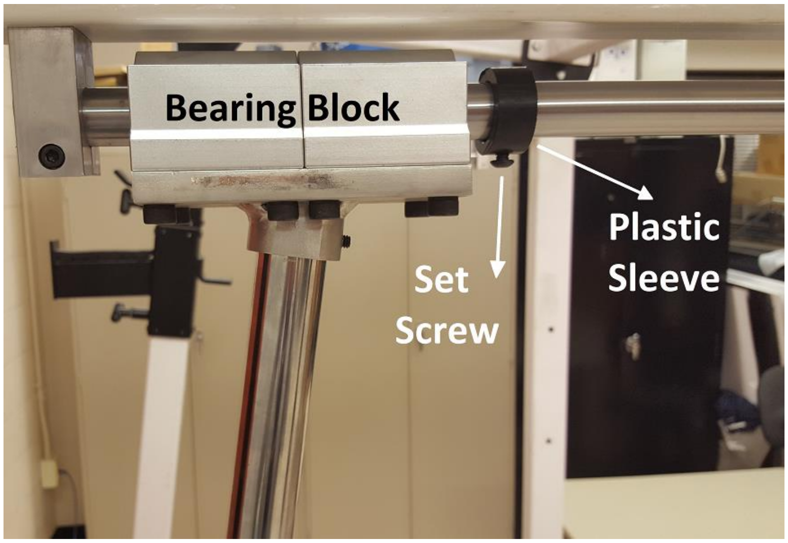


Figure 6. Bottom bearing block slides along the horizontal rod, giving the bar anterior-posterior translation. This degree of freedom can be restricted by tightening the set screws on the plastic sleeves.

Two locking hooks are attached on either ends of the bar. When the hardened steel rod is twisted, the hooks can slide in and out of the holes made in the Smith machine, safely racking and unracking the bar. The subject cannot rack or unrack the bar by twisting the bearing handles as the handles will twist independently of the rod. Once the subject and the spotters on either side of the bar lift the weight, the middle spotter will twist the hardened steel rod to unrack the barbell. The locking mechanism can be seen in Figure 7.

Figure 7. Locking hooks installed on either side of the barbell.

3.2 Chest Press Modes

Six combinations of allowed degrees of freedom (modes) were tested. 1RM was completed on a 1 translational degree of freedom (DOF) Smith machine, as the previous literature has shown the barbell 1RM bench press to be significantly greater than Smith machine's (Cotterman, Darby, & Skelly, 2005) and Smith machine's 1RM bench press to be significantly greater than dumbbell's (Saeterbakken, Van Den Tillaar, & Fimland, 2011) ; barbell > Smith machine > dumbbell (Van Der Tillaar & Saeterbakken, 2012). 50% of Smith machine's 1RM weight was then used for completion of fatigue protocols with every bench press mode.

Mode	# of DOF	DOF
Modified Smith machine 1 Mode 1	1	**Translations**: Coupled vertical
Modified Smith machine 2 Mode 2	2	**Translations**: Coupled vertical, uncoupled med-lat **Rotations**: Uncoupled med-lat
Modified Smith machine 3 Mode 3	2	**Translations**: Coupled vertical, ant-post
Modified Smith machine 4 Mode 4	3	**Translations**: Coupled vertical, ant-post, uncoupled med-lat **Rotations**: Uncoupled med-lat
Barbell Mode 5	5	**Translations**: Coupled vertical, med-lat, ant-post **Rotations**: Coupled vertical, med-lat

Dumbbells Mode 6	5 (each arm)	**Translations**: Uncoupled vertical, med-lat, ant-post **Rotations**: Uncoupled vertical and med-lat

Table 1. All modes of bench pressing, with according degrees of freedom.

3.3 Study Design

The study collected repeated measures of each bench press mode from all participants. Each subject attended three lab sessions; one 1RM testing session and two data collection sessions. Three bench press mode fatigue protocols were performed on collection day. Latin square design was used to mitigate the sequential bias caused by fatigue during the second and third fatigue protocol. **Table 2** shows each participant's schedule for two collection days.

Participant	Bench Press Mode					
	Lab Session 1			Lab Session 2		
1	Mode 1	Mode 2	Mode 3	Mode 4	Mode 5	Mode 6
2	Mode 2	Mode 3	Mode 4	Mode 5	Mode 6	Mode 1
3	Mode 3	Mode 4	Mode 5	Mode 6	Mode 1	Mode 2
4	Mode 4	Mode 5	Mode 6	Mode 1	Mode 2	Mode 3
5	Mode 5	Mode 6	Mode 1	Mode 2	Mode 3	Mode 4
6	Mode 6	Mode 1	Mode 2	Mode 3	Mode 4	Mode 5
…19	Mode 1	Mode 2	Mode 3	Mode 4	Mode 5	Mode 6

Table 2. Latin square study design.

Upon first visitation, 1RM on barbell bench press (mode 4) was tested. Subjects were given instructions on the performance of the tests, and biacromial width was collected so grip width for all bench press modes could be set and maintained at 165% of biacromial width (Schick, et al., 2010), as it has been previously shown to provide highest force production of all grips during the supine bench press (LL, Evans, Wier, Housh, & Johnson, 1992). On the second and third visits, the six bench press mode fatigue tasks were performed. The fatigue protocols in the same session were spaced 20 minutes apart (Brennecke, et al., 2009). Lab sessions were spaced a minimum of 72 hours apart as this has previously been shown to be sufficient time for complete recovery from a moderate volume strength training session (Beachle & Earle, 2008).

3.4 Modified Smith machine bench press/Barbell bench press set up

A standard bench press setup was used; during each test, the spotter was standing behind participant's head, gripping the bar with an alternated over-under grip (Beachle & Earle, 2008). The bar was racked at a height which allowed the participant to grip the bar while maintaining approximately 45° elbow flexion. Once the subject is ready, the bar was lifted with the help of the spotter until the participant's elbows are fully extended. Once the elbows are fully extended, the spotter removed their hands from the bar. When the participant was ready and had full control of the weight in their hands, they were free to start the test at any time of their choosing, abiding by the pre-set metronome with the cadence of 60 beats/minute. Once the participant missed two consecutive metronome beats or failed to lift the weight, the spotter aided in lifting and re-racking the barbell.

3.5 Dumbbell Bench Press Set-up

The traditional self-setup for a dumbbell bench press (which involves the trainee rolling from a seated to supine position by kicking each dumbbell from the knee to chest) was avoided to conserve the participants' energy for the fatigue protocol. Once the participant is supine on the bench and ready, the spotter carefully handed the dumbbells onto each of participant's hands. When the participant was ready and had full control of the weight in their hands, they were free to start the test at any time of their choosing, abiding by the pre-set metronome with the cadence of 60 beats/minute. Once the participant missed two consecutive metronome beats or failed to lift the weight, the dumbbells were either removed from the participants' hands by the spotter or dropped on the floor.

During this mode, participants were instructed to maintain the pre-calculated distance of 165% of their biacromial width throughout the entire movement, as opposed to having the dumbbells making contact at the top of the movement, as is often the case with the traditional dumbbell bench press exercise. The rationale behind this is that as the dumbbells come closer together at the top of the movement, the horizontal moment arm between the load (dumbbell) and pivot point (shoulder joint) gradually decreases, eventually nullifying when the dumbbells are directly above the shoulder joint. This movement was practiced without weights prior to starting the test.

3.6 Load intensity

1RM testing procedure followed the protocol outlined in NSCA's Essentials of Personal Training Symposium Workbook (National Strength & Conditioning Association, 2006). The protocol is as follows;

- Before beginning the test, each participant was asked to give an approximate 1RM bench press.

1. Participant's 15RM was calculated according to Brzycki's formula and 5-10 repetitions were performed, followed by a 1-minute rest.
2. 10-20lbs were added and 3-5 repetitions were performed, followed by a 2 minute rest.
3. 10-20lbs were added and 1-3 repetitions were performed, followed by a 2-4 minute rest.
4. 10-20lbs were added and 1 repetition was performed.

5. 10-20lbs were added with each successful attempt with 2-4 minute rest periods until true 1RM has been reached.
6. If the subject failed an attempt and felt like the increment in weight was too great, 5-10lbs was decreased and the test was re-attempted until true 1RM has been reached.

According to Brzycki's 1 repetition maximum formula, trainees should be able to complete 20 repetitions with 47% of their 1RM (Brzycki, 1993) on any "compound" exercise (i.e. exercises which work the musculature of multiple joints, ex: deadlift, squat, bench press);

$$(1RM = \frac{weight}{1.0278*(0.0278*reps)}) \tag{4}$$

where, *weight* is the weight used for the exercise, *reps* is the maximum number of repetitions performed with that weight and *1RM* is the predicted 1 repetition maximum.

During the fatigue protocol, 50% participants' barbell 1RM bench press was used to perform maximum number of repetitions on every bench press mode.

3.7 Subjects

Number of Subjects

A sample size formula proposed Kadam & Bhalerao (2010) was used to calculate our sample size;

$$n = \frac{2\left(Z_a+Z_{1-\beta}\right)^2\sigma^2}{\Delta^2} \tag{5}$$

Where, Z in Z_a is a constant set by convention according to values found in table 3, Z in $Z_{1-\beta}$ is a constant set by convention according to the power of the study (table 4), σ is the estimated standard deviation of the treatment and Δ is the estimated effect size of the treatment, compared to the control (Kadam & Bhalerao, 2010).

α-error	0.05	0.01	0.001
2-sided 1-sided	1.96 1.65	2.5758 2.33	3.2905

Table 3. Constant Z_a values.

Power	0.8	0.85	0.9	0.95
Value	0.8416	1.0364	1.2816	1.6449

Table 4. Constant $Z_{1-\beta}$ values.

A significance level of p>0.05 and a power of 0.8 was chosen. According to the tables shown above, our Z_a and $Z_{1-\beta}$ values are 1.65 and 0.8416, respectively. Findings from Van Den Tillaar, Saeterbakken (2012), show men's mean 1RM dumbbell bench press and 1RM Smith machine bench press to be 89.5kg±13.7kg and 103.6kg±14.8kg respectively. Changing the unrestricted degrees of freedom from 5 uncoupled (dumbbells) to 1 (Smith machine), caused approximately a 16% effect size for men (((103.6-89.5)/89.5)*100), giving us a Δ value of 0.16. Standard deviation of the Smith machine bench press' 1RM, which in our case, is the treatment, is approximately 14% for men ((14.8/103.6)*100), giving us σ values of 0.14. Given our values, our formula is as follows;

$$n = \frac{2(1.65 + 0.8416)^2 0.14^2}{0.16^2} = 9.5 \approx 10$$

According to Kadam & Bhalerao's sample size calculation formula, when effect size and standard deviations of treatment values found in Van Den Tillaar, Saeterbakken (2012) are used, a sample size of 10 males and 10 females was needed. 19 participants were recruited; 10 males and 9 females.

Experience

Specific movement motor patterns become “ingrained” in the central nervous system with repetition and practice. Further, it has been documented that the nervous system adapts to an exercise program much more rapidly during the initial stages of training than during the later stages (Rippetoe, 2011). To ensure proper execution of the bench press during the fatigued stage and to control for variation caused by lack of good motor control, only individuals with 6 or more months of strength training experience were recruited.

Sex

Males and females were recruited.

Age

It is unclear at which rate the muscular system’s susceptibility to fatigue increases with age. It is, however, evident from observation that coordination and kinematic awareness decreases with age. To control for variation which may be caused by age of subjects, participants in the age group of 17-35 were recruited.

Exclusion Criteria

Exclusion criteria for the study included:

1. Diagnosed chronic and/or acute injuries to the shoulder, upper arm, elbow, forearm, wrist, and/or hand (i.e. joint sprains, tendonitis, arthritis, etc.).
2. Pain or discomfort of the shoulder, upper arm, elbow, forearm, wrist, and/or hand.

3. Diagnosed neurological disorders which may affect motor function (i.e. multiple sclerosis, Parkinson's disease, etc.).

3.8 Protocol

For all chest press fatigue protocols, proper exercise technique was followed as described in NSCA's Essentials of Strength Training and Conditioning textbook (Beachle & Earle, 2008). The subjects lied supine on the bench with 5 points of contact with the bench and floor; scapulae, buttocks and feet. Excessive lumbar arch (i.e. excessive lordosis) was avoided, neutral spine was maintained and scapulae were comfortably retracted. Feet were always in full contact with the ground. The investigators monitored the subjects to ensure that compensatory movement strategies didn't occur anywhere besides the shoulder and the arm as the participants became fatigued. In few cases, spine extension and neck lateral flexion occurred as participants became fatigued, but they were cued to return to proper positioning. Participants were also monitored to ensure the set-up technique is similar during both the 1RM test and the fatigue protocol. During the modified Smith machine bench press modes and barbell bench press modes, each repetition entailed the participant descending the bar until it touches the chest and ascending the bar until the elbows are fully extended. During the dumbbell bench press mode, each repetition entailed the participant descending the dumbbells until the hypothetical straight line between their arms was at their chest height, and ascending the dumbbells until the elbows are fully extended. Participants were instructed to maintain their pre-measured 165% biacromial throughout the duration of the entire test (Schick, et al., 2010), although as shown in our data, this was, on average, violated with fatigue accumulation. For consistency, all fatigue protocols followed a 1-0-1-0 repetition tempo; 1 second eccentric phase, 0 second pause at the bottom of the bench press, 1 second concentric phase, and 0 second pause at the top of the concentric phase. This tempo was maintained as increased repetition tempo has been shown to help subjects produce more repetitions to failure (Pryor, Sforzo, & King, 2011), which may be partly due to the fact that during a faster repetition tempo, the total time under tension (TUT), which is total time the muscle performs work (Hatfield, et al., 2006) per set number of repetitions, is decreased.

During all fatigue protocols, the test began when the subject had complete control over the weight in their hands. The metronome was pre-started as to not create sudden disturbances, and participants were free to start at any time. The subjects were given a verbal cue if the tempo was not maintained. Once the subject missed two consecutive metronome beats or failed to perform another repetition, the trial was terminated and the spotter safely removed the load from the subject's hands as previously described.

3.9 Variables Measured

3.9.1 EMG

All EMG data was collected unilaterally from the right shoulder and arm. 1) Pectoralis major (sternal head), 2) bicep brachii, 3) anterior deltoid, 4) lateral deltoid, 5) posterior deltoid, 6) upper trapezius, 7) latissimus dorsi and 8) triceps brachii's EMG were collected at 2160Hz and band pass filtered between 10–500 Hz and differentially amplified to generate maximum signal amplification (Fischer, Wells, & Dickerson, 2009). Prior to electrode placement the skin was shaved, abraded with skin prepping gel and then wiped with ethanol to minimise skin impedance. Ag-AgCl surface electrodes were placed with an inter-electrode distance of 2.5 cm and oriented in parallel to the direction of the muscle fibers.

Posterior, middle and anterior deltoid electrodes were placed lengthwise along the muscle fibres as described in Konrad (2006). Triceps brachii electrodes were placed on the long head of the tricep, 1/3 of the way between acromion and olecranon processes (Criswell, 2011). Pectoralis major electrodes were placed 3cm medial to the anterior axillary line (Boettcher, Ginn, & Cathers, 2008). Upper trapezius and latissimus dorsi electrode placements were slightly modified. Because the participants are lying supine on a 12" wide bench, most of the trapezius and latissimus dorsi muscles are in contact with the bench. Because of this, trapezius electrodes were placed on the superior surface of the trapezius muscle and latissimus dorsi electrodes were placed on the very lateral surface of the latissimus dorsi muscle. Figure 8 shows the electrode placements.

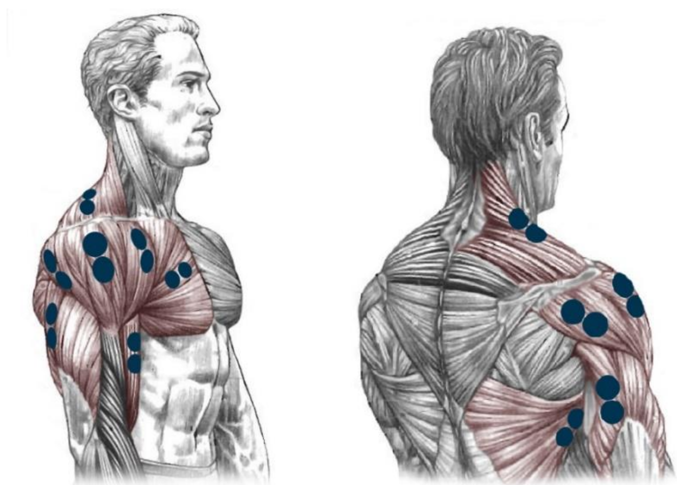


Figure 8: Electrode placement, as indicated by blue markers. Muscles measured are shown in red.

All EMG procedures and electrode placements as described in Boettcher (2008), Criswel (2011) and Konrad (2006), were be followed. Prior to collections, maximum voluntary contractions were performed so EMG signals can be normalized to percentage of maximal voltage and compared between subjects (Attebran, Mathiassen, & Winkel, 1995).

For pectoralis major MVC, the participant lied supine on a therapist's table, with shoulder abducted at approximately 80° and elbow slightly flexed. Shoulder was horizontally flexed against manual resistance provided by a researcher. For anterior deltoid MVC, the participant was standing with their shoulder is flexed at 45° with the arm internally rotated and elbow extended. Shoulder was flexed against manual resistance provided at the wrist by the researcher. For lateral deltoid MVC, the participant was standing with their shoulder abducted at 45° and elbow is extended. Shoulder was then abducted against manual resistance provided at the wrist by the researcher. For upper trapezius MVC, the participant was standing with their shoulder abducted at 90°, and the neck laterally flexed to the same side and rotated to the opposite side. Neck was extended and elbow was abducted against manual resistance provided by the researcher. For posterior deltoid MVC, the subject was prone on a therapist's table. Arm was elevated above head in line with lower trapezius muscle fibres and shoulder was flexed against manual resistance provided by the researcher. For latissimus dorsi MVC, participant was standing with their shoulder abducted at 90° with the scapula retracted elbow flexed. Shoulder was then adducted against manual resistance applied at the elbow by the researcher. For biceps brachii and triceps brachii MVC, the participants were standing. For biceps brachii, the participant applied maximal elbow flexion force against manual resistance applied at the wrist by the researcher. For triceps brachii, the participant applied maximal elbow extension force against manual resistance applied at the wrist by the researcher.

3.9.2 Tremor

Action tremor data was collected at 2160Hz with VICON Nexus throughout the duration of the fatigue protocol. Tremor data was band-pass filtered between 1-20Hz. Lower bound was chosen to eliminate movement caused by the ascend and descend of the arm, which occurred at a cadence of 1-0-1-0 as previously mentioned, and the upper bound was chosen according to the upper bound of physiological tremor. ADXL335 tri-axial accelerometers were attached on the participants' right olecranon process and distal, dorsal surface of the third metatarsal bone for measurement of dynamic action tremor. The hand accelerometer was placed on the finger for measurement of postural tremor pre and post trials which was not used in this thesis. Action tremor was measured throughout the duration of the fatigue protocol.

3.9.3 Kinematics

Kinematic data was collected at 60Hz during all bench press modes with VICON Motion Capture System. All missing data points less than 200ms were interpolated using cubic spline and all missing data points more than 200ms were interpolated by matching it to the marker which most closely resembles missing marker's motion in VICON Nexus. Kinematic data was low pass filtered at a frequency of 6Hz with a third order dual pass Butterworth filter in MATLAB. 19mm markers were attached on to the ulnar and radial styloid processes, lateral and medial elbow epicondyles, acromion process, suprasternal notch, xiphoid process, and the edges of the barbell and dumbbells. However, due to the reflective nature of the metal barbell and dumbbells, the markers were unrecognizable by the Vicon system and indication of barbell and dumbbell markers was not possible on most trials due to marker dropout. Because of this, the joint centre of the wrist, which was calculated using the ulnar and radial styloid process markers, was used to represent the path of the load, as it followed the load very closely.

The angle between the torso and humerus in the fontal plane was calculated for every frame, and averaged for every repetition. The torso vector was formed by the suprasternal notch and xiphoid process markers. The humerus vector was formed by the glenohumeral joint and elbow joint markers. Glenohumeral joint center was estimated by subtracting 60mm from the shoulder joint marker in the direction of torso's vector (Nussbaum & Zhang, 2000). Elbow joint centre was estimated by calculating the mid-way point between the medial and lateral elbow epicondyles. Thoraco-humeral angle was calculated only in the frontal plane.

3.9.4 Rating of Perceived Difficulty (RPDS)

Participants were asked to rate the level of difficulty of stabilizing the weight (i.e. controlling degrees of freedom to maintain movement along the desired path) on the first and last completed repetition immediately after completion of the fatigue protocol. Although the perceived exertion must be near 10 on the last repetition during a fatigue protocol, it was explained to participants that the RPDS scale will focus on the difficulty of stabilizing the weight, as opposed to the level of muscular exertion. An example of bench press mode 1 was presented; although perceived exertion should be near maximal on the last repetition, the weight is fairly easy to stabilize, potentially giving it a low RPDS.

In addition to RPDS scores, participants were also asked to give verbal comments and justification for their RPDS scores after completion of fatigue protocols. These comments were used for qualitative analysis and interpretation of results.

3.10 Analysis

3.10.1 Data Cropping and Analysis

All data was collected simultaneously using Vicon NEXUS and it was temporally aligned. Kinematic data was collected at 60Hz while tremor and EMG data was collected at 2160Hz. During post processing, all data was resampled to 2048Hz and beginning and end of each ascend and descend phase was marked by using the highest and lowest points in the wrist joint centre’s vertical axis kinematic data in MATLAB. All data was then cropped according to these markers to isolate the ascend phase of every repetition for analysis.

A mixed-effect linear model was used to find significant differences in variables between modes (mode effect) and significant differences in the rate of change of each variable with repetitions between modes (repetition/mode interaction). Preliminary analysis was conducted to check for linearity of relationships. Overall, the linearity assumption was not violated. No exponential, or other non-linear patterns in the relationship were observed. All data was inputted into a mixed-effect linear model, and effects of mode and rate of change per each additional repetition (i.e. fatigue) for every mode were tested. The mixed-effect linear model output's effect of mode and mode/repetition interaction was used to create linear trendline for every variable for every mode.

Variability from one movement cycle to the next is a well observed phenomenon both anecdotally in strength training in the beginner phase, and scientifically (Churchland, Afshar, & Shenoy, 2006) (Beers, Haggard, & Wolpert, 2004) (Harris & Wolpert, 1998), especially when using very light loads in the absence of fatigue, as there are a redundant number of muscle synergies which may be used accomplish a task. Movement variability is a normal variation of both the movement kinematics and muscle activation across repetitive cycles of motor tasks (Stergiou & Decker, 2011). In addition, repetitive practice may make the movement less variable. This may be observed in athletes like basketball players who practice skills like free throws to make them as consistent and reproducible as possible (Lametti, Houle, & Ostry, 2007). When introduced to a new task, as is the case in our study on modified Smith machines bench press modes, the movement variability from one repetition to the next may be greater. Due to this movement variability, comparing the effect of modes using trendlines was more advantageous because if we compared the first repetitions between modes, within trial variability may add random variation which would inflate our p-values.

3.10.2 Effort

a) Initial EMG Amplitude

Raw EMG data from all muscles was collected as described previously. Signals were band-pass filtered between 10-500Hz with a 3rd order dual pass Butterworth filter, bias was removed and all values were converted to absolute values. Data was cropped using temporal markers indicating the beginning and end of each ascend and descend phase from wrist joint centre's kinematic data. All data was normalized to MVC values and low pass filtered at 6Hz (Winter, 2009) with 3rd order dual pass Butterworth filter. Normalized, filtered EMG signals were saved for qualitative analysis and then mean activation was calculated for every ascend phase of every repetition.

Each repetition was represented with one value; the mean magnitude of the 6Hz low pass filtered muscle activation amplitude. These values were input to the mixed-effect linear model as previously described. The model output was an average linear trendline of every muscle for every mode among 19 participants (n(male) = 10, n(female) = 9). Significances between the intercepts were used to compare significances between modes in the absence of fatigue. Significance level was set at 5% (α=0.05).

b) Action Tremor

Raw EMG data from two tri-axial accelerometers was collected in 3 axes as described previously. The total tremor amplitude was calculated from the three axes using Pythagorean quadruple ($a^2+b^2+c^2=d^2$). Signals were band-pass filtered between 1-20Hz with a dual pass Butterworth filter. Tremor data was then converted to absolute numbers and cropped using temporal markers indicating the beginning and end of each ascend and descend phase from wrist joint centre's kinematic data. Mean tremor amplitude was calculated for every repetition.

Each repetition was represented with one value; the mean magnitude of the absoluted tremor amplitude. These values were inputted into the mixed-effect linear model as previously described, but without comparing the effect of each additional repetition (i.e. intercept and slope were combined). The model output was an average tremor magnitude which included the effect of the intercept and fatigue, among 19 participants (n(male) = 10, n(female) = 9). Significances between every mode's mixed-effect linear model's magnitudes were used to compare significances between modes. Significance level was set at 5% (α=0.05).

c) Initial Rating of Perceived Difficulty (RPDS)

Initial RPDS scores were input to a repeated measures ANOVA to test if there are significant differences between modes. Sample size was 19 participants (n(male) = 10, n(female) = 9), significance level was set at 5% (α=0.05).

3.10.3 Fatigue

a) Repetitions to Failure

The total completed number of repetitions to failure were inputted into a repeated measures ANOVA to test if there are significant differences between modes. Sample size was 19 participants (n(male) = 10, n(female) = 9), significance level was set at 5% (α=0.05).

b) Rate of EMG Amplitude Change

Same processing steps were done as during initial EMG amplitude analysis. Each repetition was represented with one value; the mean magnitude of the 6Hz low pass filtered muscle activation amplitude. These values were inputted into the mixed-effect linear model as previously described. The model output was an average linear trendline of every muscle for every mode among 19 participants (n(male) = 10, n(female) = 9). EMG amplitude average change per repetition values among 19 participants were used for comparative analysis. Significance level was set at 5% (α=0.05).

c) Rate of EMG MPF Change

Raw EMG data from all muscles was collected as described previously. Signals were band-pass filtered between 10-500Hz with a 3rd order dual pass Butterworth filter. Data was cropped using temporal markers indicating the beginning and end of each ascend and descend phase from wrist joint centre's kinematic data. From the beginning of each ascend phase, data was cropped into three 512 frame bits (0.25s) and MPF analyses were done on each window. The average MPF of the three bits were calculated as the repetition's mean MPF.

Each repetition was represented with one value; the mean MPF of the muscle's EMG signal. These values were inputted into the mixed-effect linear model as previously described. The model output was an average linear trendline of every muscle for every mode among 19 participants (n(male) = 10, n(female) = 9). MPF average change per repetition values among 19 participants were used for comparative analysis. Significance level was set at 5% (α=0.05).

d) Rate of Action Tremor Change

Same processing steps were done as during action tremor amplitude analysis. Each repetition was represented with one value; the mean magnitude of each repetition's action tremor. These values were inputted into the mixed-effect linear model as previously described. The model output was an average linear trendline of every muscle for every mode among 19 participants (n(male) = 10, n(female) = 9). Tremor amplitude average change per repetition values among 19 participants were used for comparative analysis. Significance level was set at 5% (α=0.05).

e) Kinematics

Raw kinematics data was collected as previously described. Signal was low pass filtered with a 3^{rd} order dual-pass Butterworth filter with a cut-off frequency of 6Hz. Data was cropped using temporal markers indicating the beginning and end of each ascend and descend phase from wrist joint centre's kinematic data and ascend phases were isolated. From this data, four measures were used to analyze kinematics change with fatigue.

First measure, path length ratio (PLR), is the ratio of the actual bar ascend distance travelled (A) to the theoretical bar ascend distance travelled (S) in the sagittal plane. This variable was calculated using the formula; $PLR = (\frac{A}{S})$. A number closer to 1.0 is indicative of a straight bar path (Duffey & Challis, 2007). Modes 1 and 2 were excluded from the analysis because no movement in the anterior-posterior degree of freedom was allowed. Mode 4 was excluded from the analysis because our joint markers were attached on the wrist as opposed to the bar, and because the rotational degree of freedom about the medial-lateral axis was unrestrained and had very little inertia during modes 2 and 4, potentially inflating the magnitude of motion of the load path due to movement of the wrist.

The second and third measures, mean shoulder distance in the lateral-medial axis and in the anterior-posterior axis, are the measures of each concentric repetition's mean distance between the shoulder and wrist in said axes in the frontal plane. Modes 1, 3 and 5 were excluded from the analysis of mean shoulder distance in the lateral-medial axis as lateral-medial degree of freedom was restrained during these modes. Modes 1 and 2 were excluded from the analysis of mean shoulder distance in the anterior-posterior axis because anterior-posterior degree of freedom was restrained during these modes.

The fourth measure, the press angle, is the angle formed between position of the first frame of the ascend phase and the last frame of the ascend frame in the sagittal plane. The press angle of modes 1 and 2 was not included in the results sections as anterior-posterior degree of freedom is restrained during these modes and the press angle remained at a constant 9°.

The fifth measure, the thoraco-humeral angle, is the angle formed between the thorax and the humerus in the frontal plane. This variable is each concentric repetition's mean thoraco-humeral angle.

Each repetition was represented by one of each values described above. These values were input to the mixed-effect linear model as previously described. The model output was an average linear trendline of every muscle for every mode among 17 participants (n(male) = 10, n(female) = 9). The difference between the intercepts were used to determine whether there were significant differences between modes in the absence of fatigue. The average change per repetition were used to determine whether there were significant differences between modes in the rate of fatigue development. Significance level was set at 5% (α=0.05).

f) Final RPDS Scores

Final RPDS scores were input to a repeated measures ANOVA to test if there are significant differences between modes. Sample size was 19 participants (n(male) = 10, n(female) = 9), significance level was set at 5% (α=0.05).

IV. Results

4.1 Repetitions to Failure

There was a significant main effect on the number of repetitions to failure between bench press modes. Modes 1, 3 and 5 had a significantly higher number of repetitions to failure than modes 2, 4 and 6. The number of repetitions did not differ significantly between modes 1, 3 and 5 and between modes 2, 4 and 6.

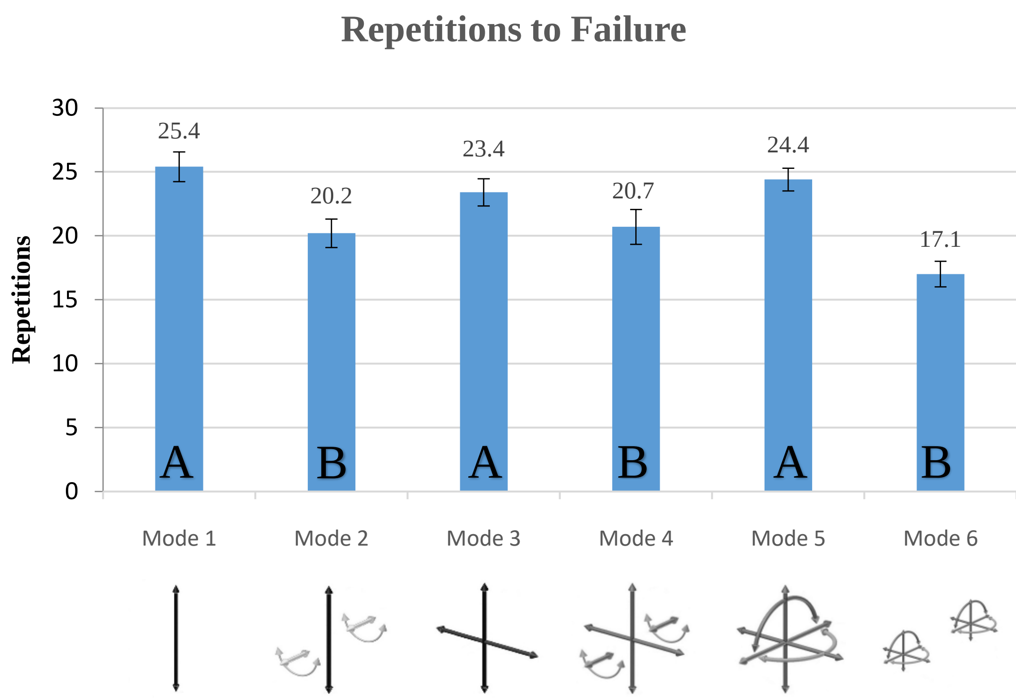


Figure 9. Number of repetitions to failure during each mode of bench press. Numbers above bars represent the average number of repetitions among 19 participants. Error bars represent the standard error for each mode. Letters inside bars group modes together based on significances.

4.2 Initial Activation

There was a significant main effect on all prime mover muscles' activity for condition. Modes 2 and 4 had the greatest overall prime mover muscle activity. During these modes, there was a redistribution of muscle stress from elbow extenders to shoulder (horizontal) flexors as they resulted in significantly lower triceps activation than mode 1 and significantly greater anterior deltoid and pectoralis major activation than mode 1. Mode 3, despite having less unrestricted degrees of freedom than mode 5, resulted in greater activation of pectoralis major and anterior deltoid, with no significant difference in triceps activation from modes 1 or 5. Modes 5 and 6 showed no significant differences from mode 1 except significantly greater pectoralis major activation during mode 6 and pectoralis major activation approaching significance at p=0.098 during mode 5.

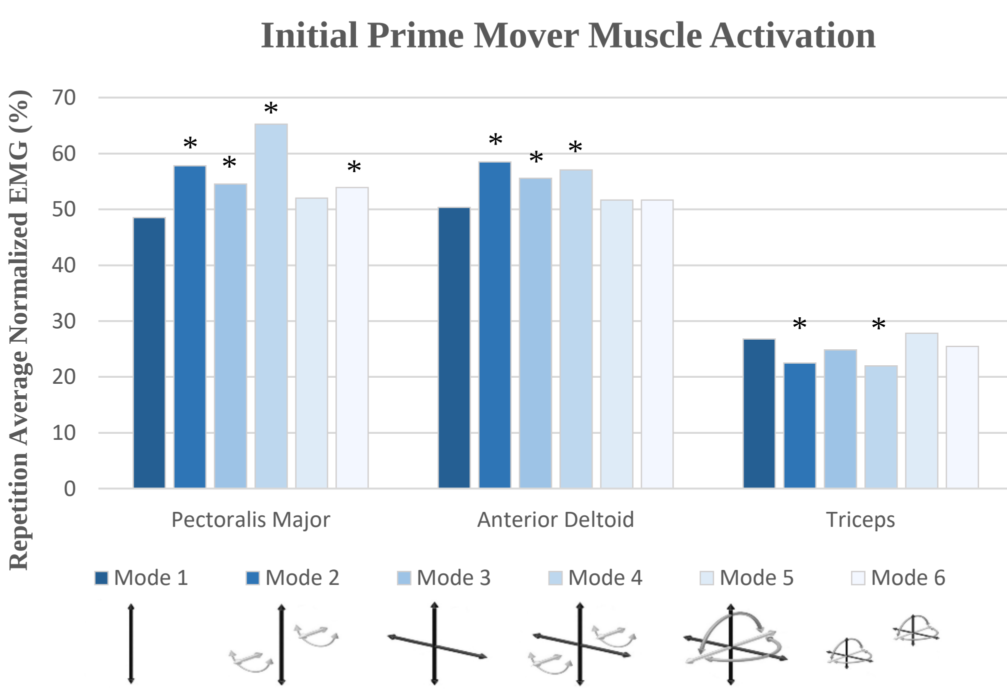


Figure 10. Mixed effect linear model's estimated initial average prime mover activation among 19 participants. Asterisks above each bar represent significance differences from mode 1.

Biceps' initial activation showed a significant increase from mode 1 to all modes except mode 3, which was approaching significance at p = 0.09. There was a clear trend towards biceps' activation being greater in modes which had uncoupled medial-lateral degree of freedom unrestrained, with mode 6 having significantly greater biceps activation than all other modes. Other non prime mover muscles which act on the shoulder did not increase overall activation with less stable modes. With the exception of posterior deltoid and latissimus dorsi during mode 6 and trapezius during mode 4, there were no significant differences in initial muscle activation from mode 1.

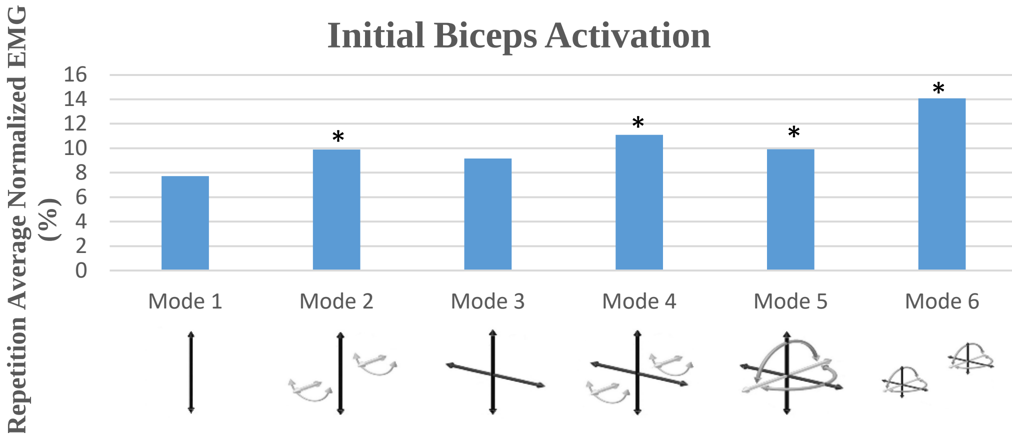


Figure 11. Mixed effect linear model's estimated initial average biceps activation among 19 participants. Asterisks above each bar represent significant differences from mode 1.

	Mode 1	**Mode 2**	**Mode 3**	**Mode 4**	**Mode 5**	**Mode 6**
Latissimus Dorsi	13.8	13.4	**11.2***	**11.1***	13.7	**15.5***
Lateral Deltoid	12.4	12.3	13.1	12.3	12.2	**14.8***
Posterior Deltoid	6.4	**4.4***	**3.6***	**3.9***	6.5	7
Trapezius	9.9	9	11	**12***	9.2	10.4

Table 5. Mixed effect linear model's estimated initial average non prime mover muscle activation among 19 participants. Asterisks beside each bolded number represent significant differences from mode 1.

4.3 Rate of EMG Change

Figures 4 through 6 show the mixed effect linear model's average rate of normalized EMG amplitude increase of each muscle for each mode among all participants, for the average number of repetitions on each mode among all participants. There was a significant main effect on mode and mode-repetition interaction for all muscles. Pectoralis major had the greatest rate of EMG increase during modes 4 and 6, which had the most degree of freedom unrestricted. Mode 4, due to muscle stress redistribution from the triceps to pectoralis major and number of repetitions completed, achieved the greatest peak pectoralis major EMG amplitude. Mode 1, despite creating the greatest number of repetitions to failure, achieved the lowest final EMG amplitude.

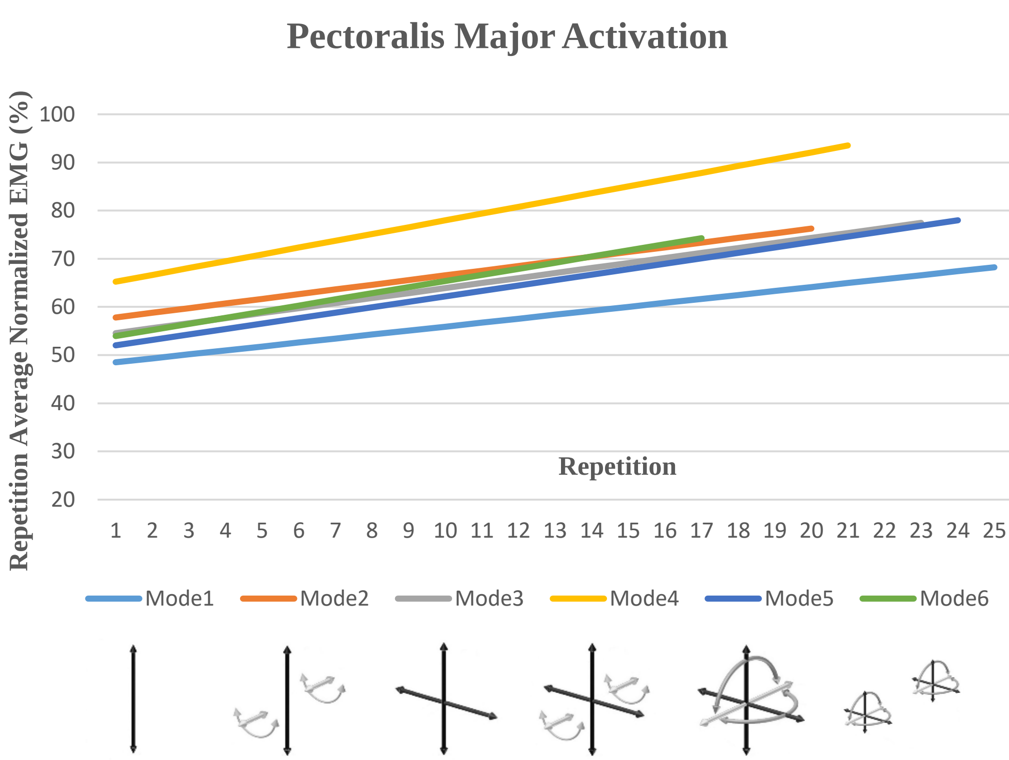


Figure 12. Mixed-effects linear model's estimated average rate of normalized pectoralis major EMG increase during each mode.

Modes 5 and 6 resulted in a significantly greater rate of anterior deltoid EMG amplitude increase with no significant difference between them. However, the barbell bench press achieved the greatest average final recruitment of the anterior deltoid due to a greater number of repetitions completed. Mode 1 achieved the lowest average peak EMG amplitude despite having the greatest number of repetitions to failure. Triceps had a significantly greater rate of increase during modes 3 and 5, where medial-lateral degree of freedom was unrestrained, than during all other modes.

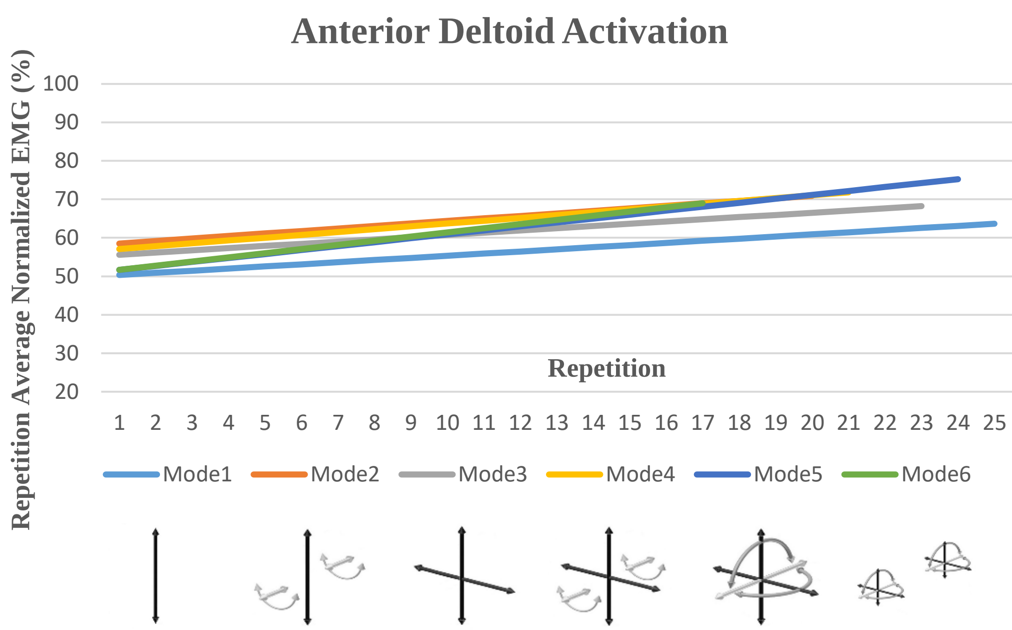


Figure 13. Mixed-effects linear model's estimated average rate of normalized anterior deltoid EMG increase during each mode.

There was a general increase in non-prime mover muscle activation with fatigue. Modes which had the greatest initial biceps activation had the lowest rate of biceps EMG increase. Lateral deltoid, posterior deltoid, and trapezius had the highest rate of increase on modes which required control anterior-posterior degree of freedom. Latissimus dorsi had the highest rate of increase on modes which required control of uncoupled lateral-medial degree of freedom.

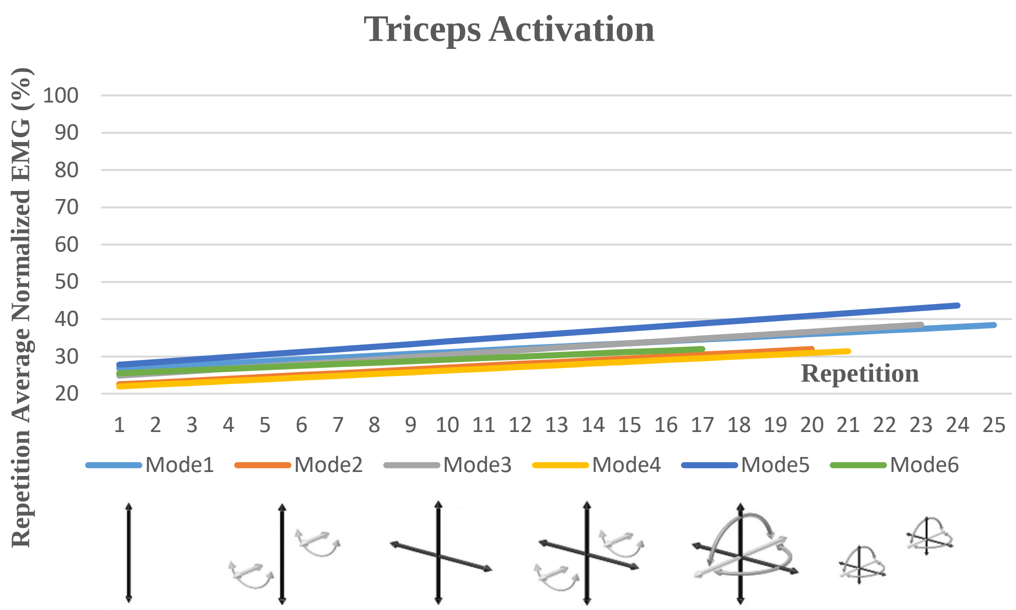


Figure 14. Mixed-effects linear model's estimated average rate of normalized triceps EMG increase during each mode.

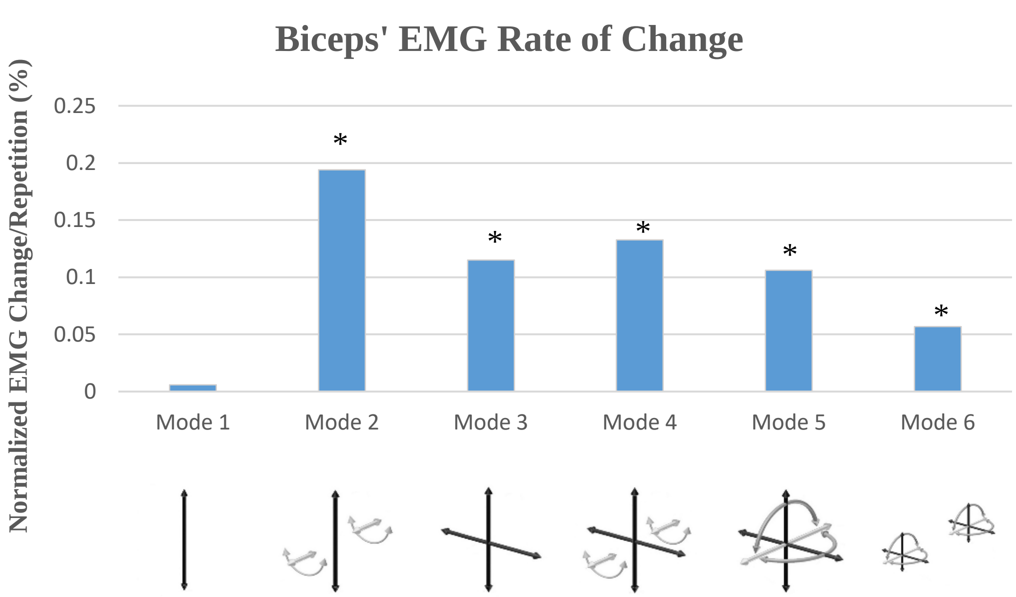


Figure 15. Mixed-effects linear model's estimated average rate of normalized biceps EMG increase during each mode. Asterisks above each bar represent significant differences from mode 1.

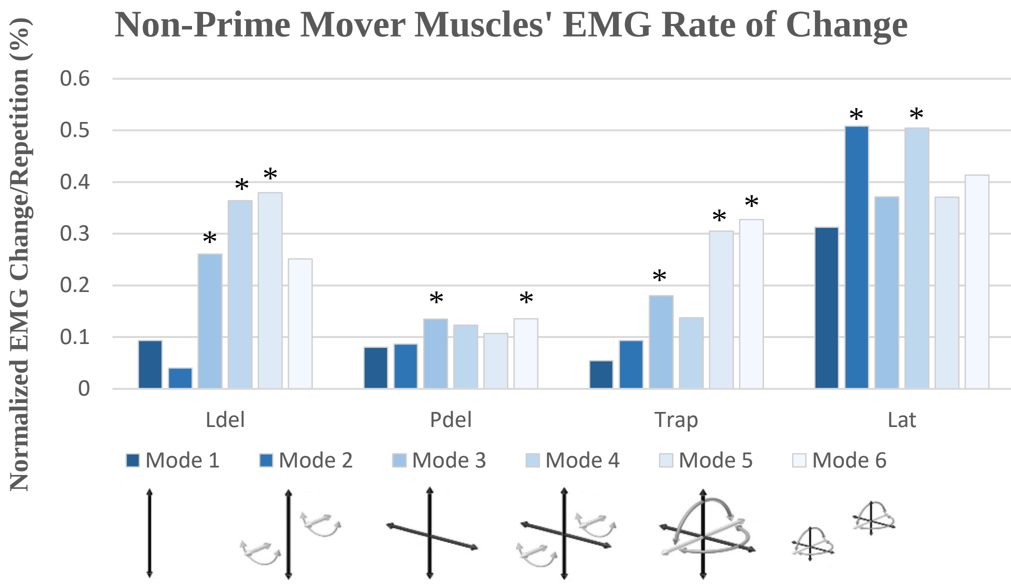


Figure 16. Mixed-effects linear model's estimated average rate of normalized EMG increase for non-prime mover muscles during each mode. Asterisks above each bar represent significant differences from mode 1.

4.4 Co-contraction Index

For initial C.I., modes 2 and 6 showed a significant difference from mode 1 despite mode 2 being only marginally lower. During mode 1, the rate of C.I. change was negative, which was significantly different from all other modes. During this mode, the prime mover muscles increased in EMG amplitude at a significantly faster rate than non-prime mover muscles when compared to other modes, indicating that non-prime mover muscles increase at a faster rate with more instability. During mode 6, the rate of C.I. change was negative, which was significantly different from all other modes. However, the initial C.I. during mode 6 was significantly greater than during all other modes, indicating that even in the absence of fatigue, co-contraction of non-prime mover muscles was greater than during mode 6 than during all other modes.

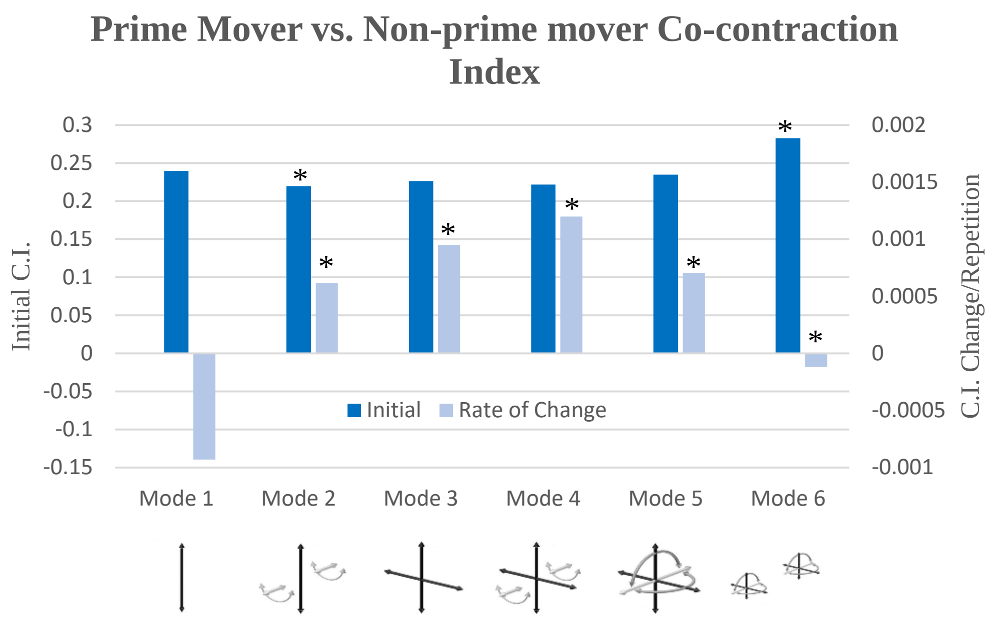


Figure 17. Mixed-effects linear model's co-contraction indexes among 19 participants. Asterisks above each bar represents significant difference from mode 1.

4.5 Rating of MPF Change

There was a significant main effect on all prime mover muscles' MPF's mode-repetition interaction. Prime movers' MPF rate of decrease followed a similar trend as their initial activation. Overall, muscles which had higher initial activation had a greater rate of fatigue. Generally, all modes which had uncoupled lateral-medial degree of freedom unlocked induced greater pectoralis major and anterior deltoid fatigue and lesser triceps fatigue, with the exception of mode 6, which had amongst the greatest rates of pectoralis major fatigue, greatest rate of triceps fatigue and no significant difference in the rate of anterior deltoid fatigue from mode 1.

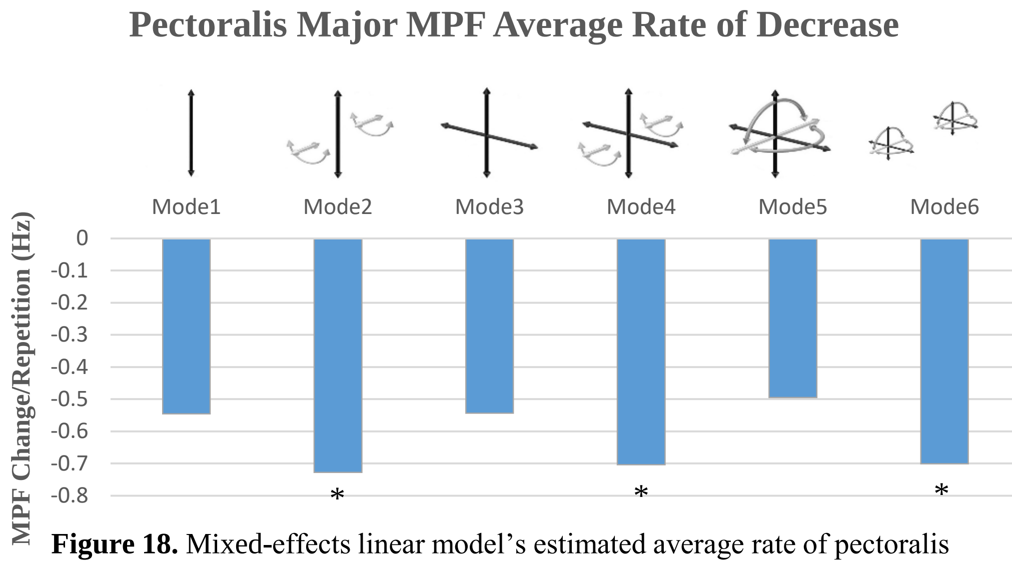


Figure 18. Mixed-effects linear model's estimated average rate of pectoralis major's mean power frequency change per repetition among 19 participants. Asterisks below each bar represent significant differences from mode 1.

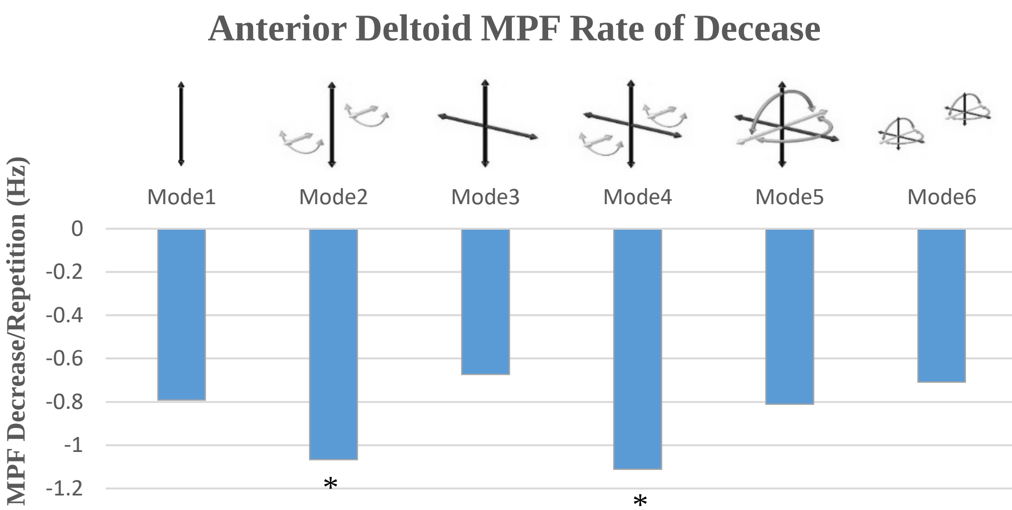


Figure 19. Mixed-effects linear model's estimated average rate of anterior deltoid's mean power frequency change per repetition among 19 participants. Asterisks below each bar represent significant differences from mode 1.

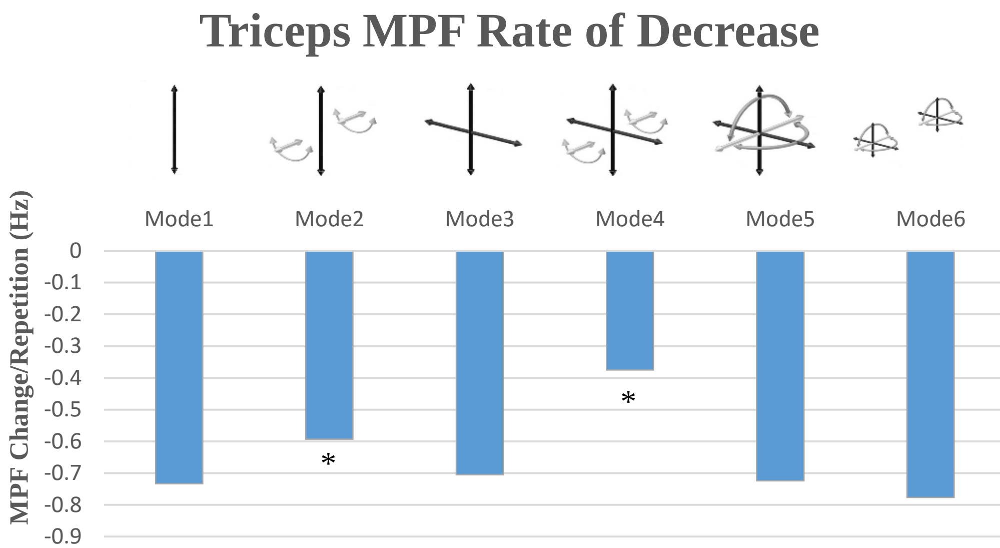


Figure 20. Mixed-effects linear model's estimated average rate of triceps' mean power frequency change per repetition among 19 participants. Asterisks below each bar represent significant differences from mode 1.

4.6 Rate of Perceived Difficulty Score

There was a significant main effect on both initial and final rate of perceived difficulty. Unstable Smith machine modes resulted in greater RPDS scores than free weight modes. Interestingly, mode 4 induced a significantly greater initial and final RPDS score than mode 6 and modes 2 and 4 induced significantly greater final RPDS scores than modes 6 despite having less unrestrained degree of freedom. Mode 3 produced significantly greater initial and final RPDS scores than mode 5 despite having less unrestrained degrees of freedom.

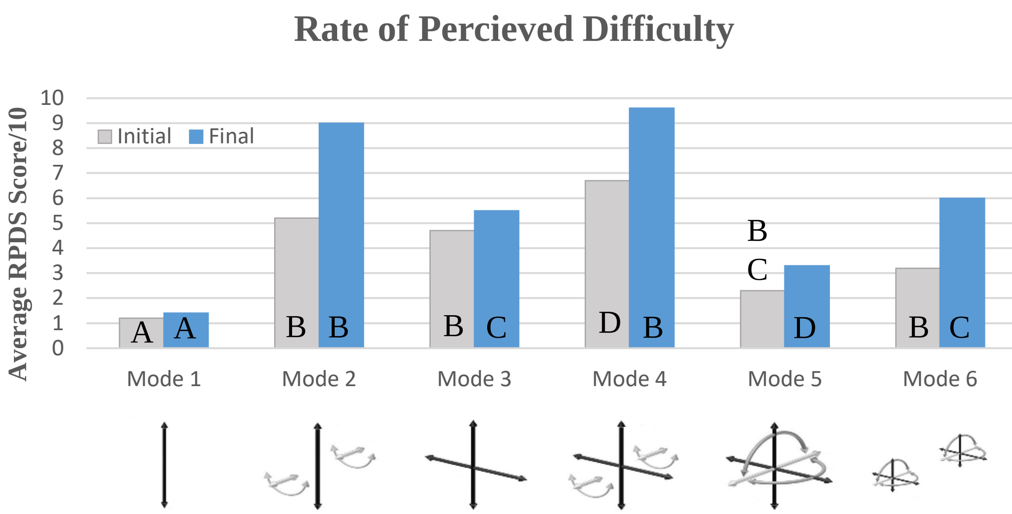
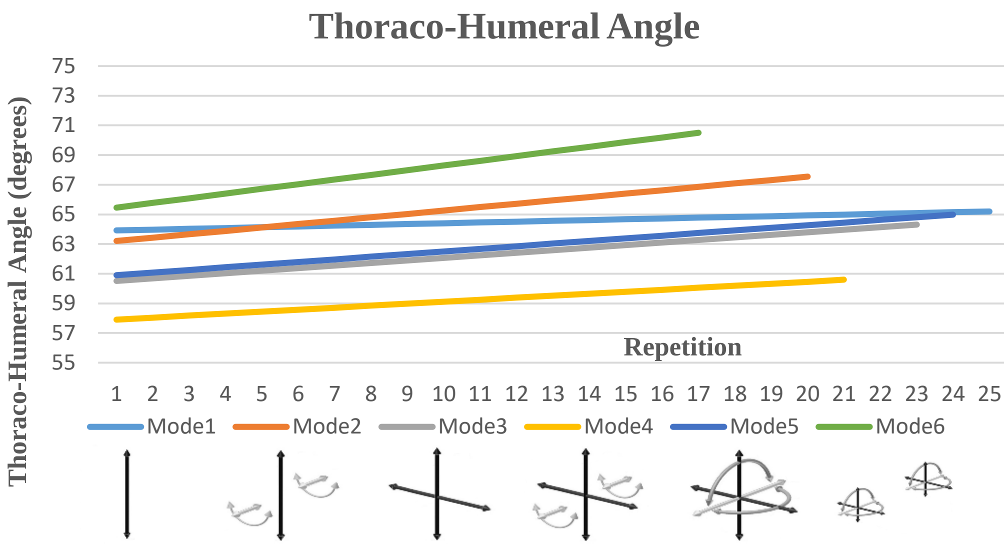


Figure 21. Average initial and final rate of perceived difficulty (RPDS) among 19 participants for every mode. Letters inside bars group modes together based on significances.

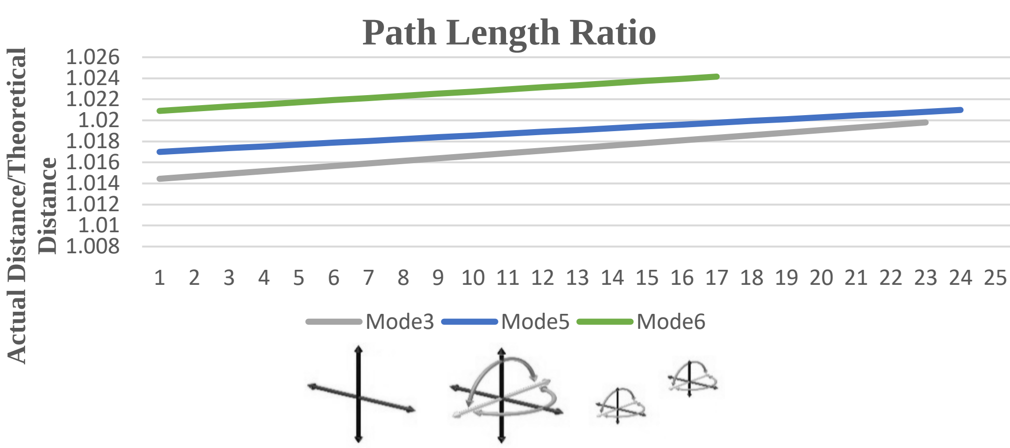
4.7 Kinematics

There was a significant main effect for all measured kinematic variables for both mode and mode-repetition interaction. All modes had a significantly greater rate of thoraco-humeral angle increase from mode 1 with fatigue, with mode 6 having the greatest initial angle and greatest rate of increase. Surprisingly, mode 4, which was considered the most difficult, had the lowest initial and overall angle. Mode 6 had the most vertical pressing angle (i.e. greatest pressing angle) which potentially resulted in a smallest average mean wrist-shoulder distance in anterior-posterior axis as the load was held more directly over the shoulder in both the anterior-posterior and medial-lateral axes through-out the entirety of the lift. Between modes which had anterior-posterior degree of freedom unlocked, machine modes 3 and 4 had the smallest rate of press angle change with fatigue, significantly lower than modes 5 and 6 despite mode 3 having the least unrestrained degrees of freedom out of those modes. The overall PLR is lower during mode 3 than during mode 5, despite mode 3 having less unrestrained degrees of freedom than mode 5. Mode 6 had the greatest PLR, consistent with having greatest difficulty of control.



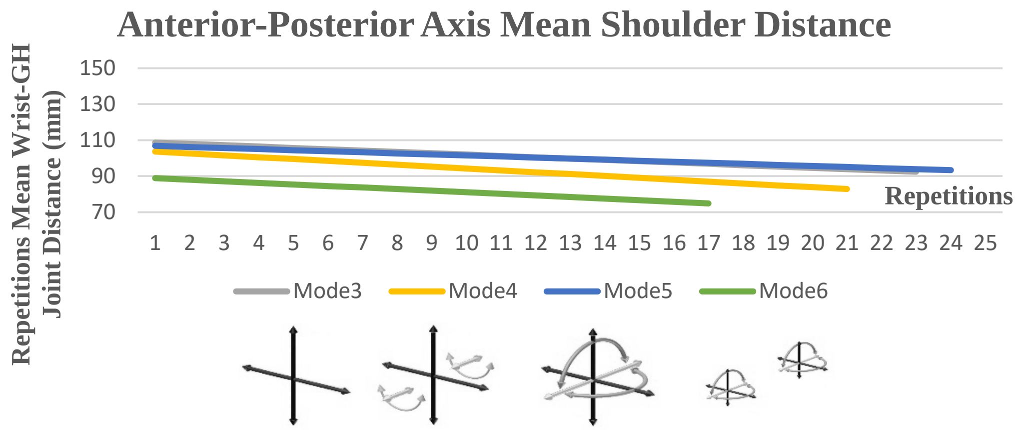
	Mode 2	Mode 3	Mode 4	Mode 5	Mode 6
Initial		*	*	*	
Rate of Change	*	*		*	*

Figure 22. Mixed-effect linear model's average thoraco-humeral angle among 19 participants for every mode. Asterisks in each box represent significant differences from mode 1.



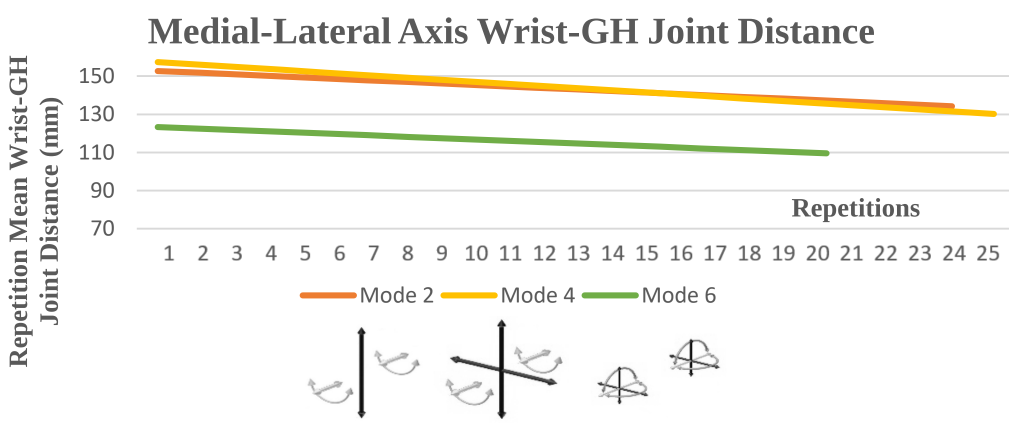
	Mode 3	Mode 5	Mode 6
Bias	*	*	*
Rate of Change	*	*	*

Figure 23. Mixed-effect linear model’s average path length ratio among 19 participants for modes which did not use the Smith machine moving handles. Asterisks in each box represent significant differences from mode 1.



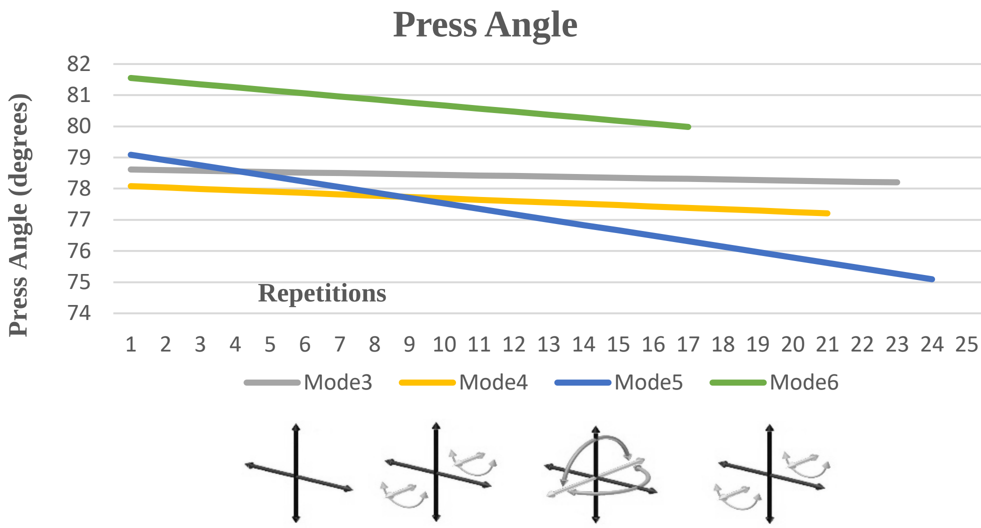
	Mode 3	Mode 4	Mode 5	Mode 6
Initial	*	*	*	
Rate of Change	*		*	*

Figure 24. Mixed-effect linear model's average back-forward axis distance between the GH joint and the wrist among 19 participants for modes which had back-forward degree of freedom unrestricted. Asterisks in each box represent significant differences from mode 1.



	Mode 2	Mode 5	Mode 6
Initial	*	*	
Rate of Change	*	*	*

Figure 25. Mixed-effect linear model's average side-side axis distance between the GH joint and the wrist for modes which had side-side degree of freedom unrestricted. Asterisks in each box represent significant differences from mode 1.



	Mode 3	Mode 4	Mode 5	Mode 6
Initial	*	*	*	
Rate of Change			*	*

Figure 26. Mixed-effect linear model's average press angle among 19 participants for modes which had back-forward degree of freedom unrestricted. Asterisks in each box represent significant differences from mode 1.

4.8 Action Tremor

There was a significant main effect for both hand and elbow action tremor for mode. Overall, hand tremor was more sensitive to mode than elbow tremor. Most notably, hand tremor was significantly greater on modes 2 and 4 than all other modes and elbow tremor during mode 4 was significantly greater than during all other modes.

There was a significant main effect for both hand and elbow action tremor rate of change. Most notably, modes 2 and 4, which had the lowest inertia associated with the mass which moves with the hands, had the greatest rate of hand action tremor increase with fatigue. Mode 6, which has the greatest number of unrestricted degrees of freedom, had the greatest rate of elbow action tremor increase with fatigue.

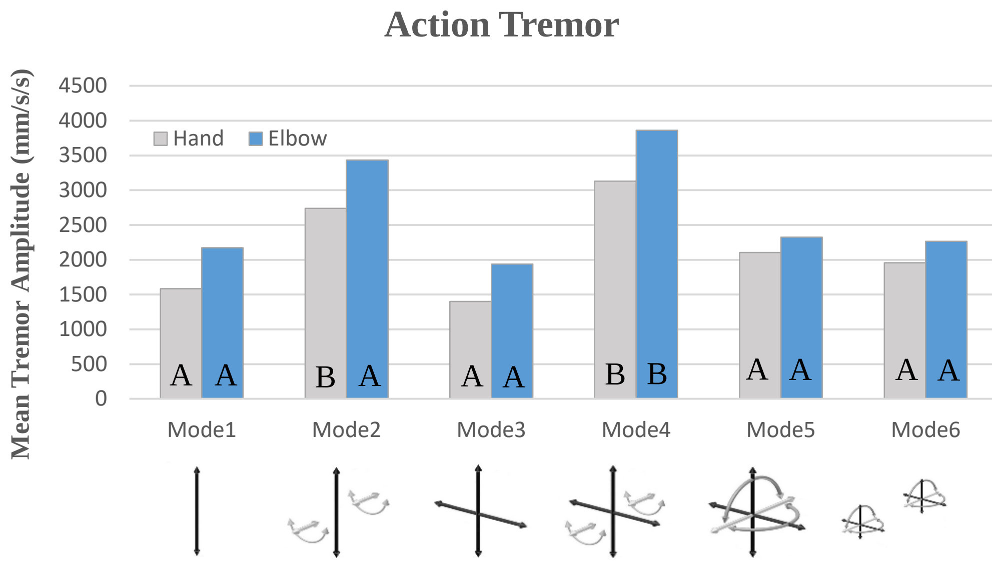


Figure 27. Mixed-effect linear model's average hand and elbow action tremor among all modes among all participants. Letters inside bars group modes together based on significances.

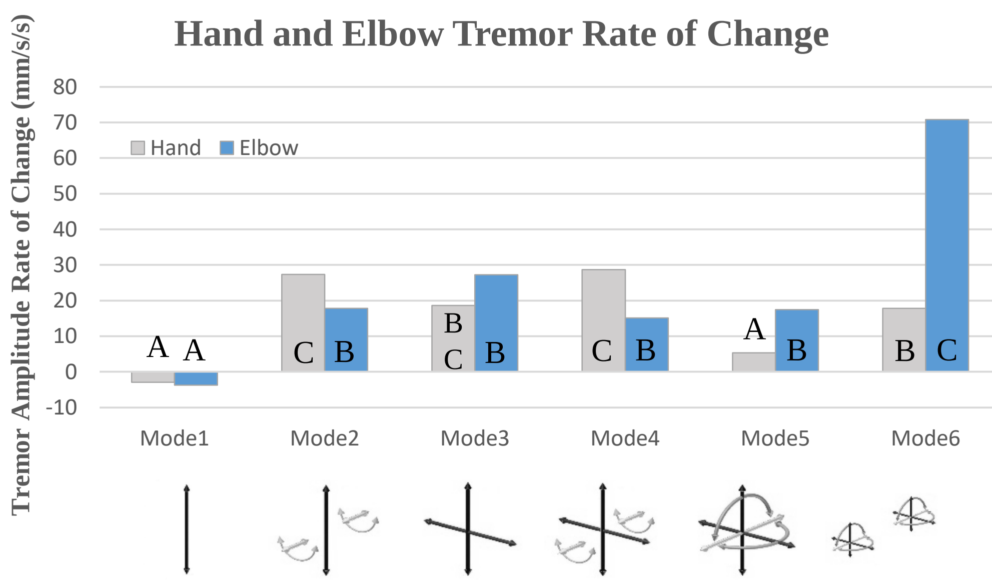


Figure 28. Mixed-effect linear model's average hand and elbow action tremor rate of change per repetition among all modes among all participants. Letters inside and above bars group modes together based on significances.

V. Discussion

The purpose of this study was to examine how changing the unrestricted degrees of freedom during a 50%1RM chest pressing task to fatigue would affect upper body muscle activation, rate of muscle fatigue, and neuromuscular and kinematic strategy changes. It is evident from our findings that effort and rate of fatigue accumulation is increased as the load becomes more unstable. Furthermore, the findings of this study, along with evidence in the literature on this subject indicate that all degrees of freedom don't add an equal degree of difficulty to all tasks. More over, the effect of unrestricted degrees of freedom is very task specific, which will be the initial focus of this discussion.

5.1 The Question of Degrees of Freedom

It is important to note that one cannot generalize additional unrestricted degrees of freedom to be equally beneficial or detrimental during all tasks. There are several factors which determine whether more freedom of movement adds a positive or a negative contribution to the desired outcome, whether the desired outcome is more force production, or decreased rate of fatigue accumulation. In this discussion, the degrees of freedom refer to the six mechanical degrees of freedom of the load against which one applies force. An "open", "externally unrestricted" or "available" degree of freedom refers to a movement along, or a rotation about any of three axes which is not mechanically restrained externally. Controlling a degree of freedom refers to a person putting forth muscular effort to restrain an externally unrestricted degree of freedom via the neuromuscular system.

Generally, adding additional unrestricted degrees of freedom to a task has both benefits and detriments to performance, the magnitudes of which are specific to the nature of the task. When the objective of the task is maximal force production or minimization of neuromuscular fatigue, the detrimental effect with more unrestrained degrees of freedom comes in the form of co-contraction of muscles which may or may not create antagonist moments to desired moments about the joints of interest. When applying force against an object while attempting to prevent motion in certain degrees of freedom, every degree of freedom which is unrestrained externally must be controlled via additional muscular contraction internally. Further, the degree of difficulty of controlling each degree of freedom varies depending on factors discussed later in this chapter. This controlling of unwanted degrees of freedom to ensure load movement only along the desired path may be referred to as “stabilizing” the load. This was demonstrated in findings in Kornecki et al, (2001) and Fischer et al, (2009), where during an upper limb static pushing tasks, increasing the number of unrestrained degrees of freedom at the point of load application was shown to decrease maximal force and power production (Kornecki, Kebel, & Siemienski, 2001), and increase activation of muscles acting on the joint articulation most proximal to the load during a submaximal task (Fischer, Wells, & Dickerson, 2009). This was consistent with our findings during a submaximal dynamic task as well, as initial activation of the biceps, which is an antagonist muscle acting on the examined joint most proximal to the point of load application, increased significantly with all modes which had uncoupled medial-lateral degree of freedom unrestrained (**Figure 11**), while initial activation of the non-prime mover musculature acting on the joint more distal from the load, with the exception of certain muscles, did not increase overall with instability in the absence of fatigue (**Table 5**). Fischer et al (2009) proposed that with unstable loads, greater muscle activation may lead to more accumulated fatigue over time, which was shown to be true during a dynamic task in our findings as bench press modes with more unrestrained degrees of freedom generally produced less repetitions to failure than bench press modes with less unrestrained degrees of freedom (**Figure 9**).

There is, however, evidence that during dynamic tasks, in some instances, some freedom of movement may be beneficial for maximal force production and perhaps to less fatigue accumulation. Saeterbakken (2011) have demonstrated that during a maximal effort chest pressing task, barbell bench press' 1RM was significantly higher than Smith Machine bench press' 1RM. As shown in Duffey & Challis (2007), an exaggerated C shaped bar path, that is to say, a bar path which utilizes the additional degree of freedom along the longitudinal axis of the torso, is more prevalent during both maximal effort barbell bench press and during submaximal effort barbell bench press in a fatigued state. This is in agreement with our findings as participants' wrist path during the dumbbell and barbell bench press moved along an arc, causing a deviation from a straight path in the sagittal plane (**Figure 30**). Further, this deviation was increased with accumulated fatigue (**Figure 23**). However, the conclusion that additional unrestrained degrees of freedom will always yield an increase in performance when compared to more stable tasks may be invalid because our findings show that although unrestricted degrees of freedom were used to a greater extent in a fatigued state, more unrestricted degrees of freedom resulted in an insignificant decrease in performance (i.e. less repetitions to failure) when comparing the barbell bench press to the Smith machine bench press (**Figure 9**). Given the assumption that the effect of instability on performance is equal during both maximal force production tasks and fatigue tasks, this discrepancy in performance between a task with 1 degree of freedom and a task with 6 degrees of freedom between a maximal force production task in Saeterbakken (2011) and a fatigue task in our study may be caused by the difference in the allowed bar path during the 1 degree of freedom Smith machine bench press. We make this assumption based on the fact that the relation between the number of repetitions to failure with dumbbell and barbell chest press was proportional to the degree of 1RM difference between the two. Welsch et al (2005) reported the dumbbell bench press' 1RM to be 63% of barbell bench press' 1RM. Saeterbakken et al. (2011) found it to be 83%. In our study, the number of repetitions to failure with the dumbbells was 70% that of barbell's.

In Saeterbakken (2011), the slope of the vertical bars which guide the barbell on the Smith machine wasn't specified, although it may be an important variable as a more sloped angle of push resembles the barbell's bar path more closely. Most standard Smith machine manufacturers angle the bars which guide the horizontal bar vertically, while few newer models angle them at a slight slope. Our Smith machine's vertical bars were angled at a more optimal 8.5° from vertical, and we presume that if our Smith machine restrained participants to using a less optimal vertical bar path, as may have been the case in Saeterbakken (2011), the Smith machine bench press may have allowed for fewer repetitions to failure. If this is the case, we may presume that at least in the case of the bench press, when anterior-posterior motion is restricted, it is not the lack of ability to self select and change the movement path with fatigue which may be detrimental to performance, but rather, it is a restriction to a movement pattern which is non optimal for performance. It may be that during tasks which allow movement along only 1 degree of freedom, when the movement path deviates from the most optimal path, which may be observed when all motion is unrestricted, the magnitude of the detriment created by a non-optimal load path outweighs the benefit of reduced complexity of the task and reduced co-contraction of musculature which acts to control unrestricted degrees of freedom. This builds a case for dynamic tasks with more freedom of movement to be more beneficial only when compared to tasks which restrict the most optimal movement path of the load. When the more stable dynamic task’s load path resembles the self selected load path more closely, as was the case in our study, having to control less unrestrained degrees of freedom has shown to be beneficial to performance, although the difference was non-significant (**Figure 9**).

Ultimately, increasing unrestricted degrees of freedom of the load may hinder performance by increasing the complexity of the task and creating a need for co-contraction as joint stiffness begins to have a greater weighing in the objective of the neuromuscular system. On the contrary, it may be beneficial to performance as the most optimal path of the load may be self selected and changed according to the change in the objective of the neuromuscular system. This is crucially important to consider when examining the effect of additional unrestricted degrees of freedom during a static task, as the detriment of controlling unwanted movement remains, while the benefit of a more optimal self selected load path, and the ability to alter the load path with fatigue, isn’t present due to the static nature of the task. In this case, when the effect of the ability to optimize the load path isn’t in the equation, the effect of additional demand for stabilization of the load may be isolated. Similarly, when comparing the effect of additional degrees of freedom during dynamic tasks, in order to isolate the detrimental, or perhaps beneficial effect which stabilization of the weight provides, the load path of the stable task must resemble the self selected path closely, as the self selected load path may be more optimal for performance. In our study, the self selected path deviated from vertical in both sagittal and transverse planes when anterior-posterior and medial-lateral degrees of freedom were unrestrained, which will be discussed in greater detail in the next chapter. Further, these deviations were exaggerated with fatigue. This change in load path with fatigue raises a question about the magnitude of the benefit to performance granted by the ability to alter the load path with fatigue during an unstable task. While not having a definitive answer based on findings of our study, the benefit to performance which is granted by having the option to alter the load path with fatigue may be better understood when compared to a stable task which, although allows an optimal load path, does not give the option to change the path.

Through-out this discussion, the term instability is ultimately used to quantify the degree of perceived difficulty of controlling unwanted motion of the load. However, in addition to the number of unrestrained degrees of freedom to be controlled, directional bias created by a moment arm and the inertial properties associated with each degree of freedom will also contribute to perceived instability and its effect on the desired outcome. In our study, modes 2, 4 and 6 had the medial-lateral degree of freedom unlocked. However, the medial-lateral degree of freedom was more difficult to control during modes 2 and 4 than during mode 6, as reflected by final RPDS scores (**Figure 21**). In addition, six participants verbally noted immediately after the completion of bench press modes 2 and/or 4 that majority of their effort was directed towards controlling this degree of freedom to keep the load path vertical. Those participants also noted that their decision to complete the test was mainly due to inability to control this degree of freedom and not exclusively due to muscular fatigue preventing their force production. This may be due to the inertial properties of this degree of freedom. Objects with more mass have larger inertia, that is to say, they require a greater force to cause a given acceleration. In the case of our medial-lateral degree of freedom on modes 2 and 4 on the modified Smith machine, the object against which participants applied force, which are our handles sliding on a bar, moved independently of the load and had very little mass and thus, required a much lesser force to accelerate and decelerate. In mode 6, medial-lateral degree of freedom was coupled with a larger mass and inertia. In order to control a degree of freedom, a constant feedback loop must be created to assess the limb's position in space and readjust it to keep the motion along the desired path. The threshold for joint angular change detection is velocity dependant (Ashton-Miller, Wojtys, Huston, & Dry-Welch, 2001), and less rapid change of velocity of the weight may make positional change of all joints involved in the kinetic chain have lower velocity, aiding in quicker detection and readjustment and in turn, less perceived instability. Fatigue may have detrimental effects on this detection threshold, which increased perceived difficulty of control to a greater extent during modes 2 and 4. This will be discussed in the last chapter.

Similar results were seen in our machine replicated degree of freedom along the longitudinal axis of the torso. Same participants reported more difficulty controlling this degree of freedom during Smith machine mode 3 than during barbell mode 5, although the barbell mode had more unrestricted degrees of freedom which theoretically would make it more difficult to control. Although participants did press at a significantly steeper angle during mode 3 than mode 1, this pressing angle change with fatigue did not differ significantly from mode 1, whereas barbell bench press mode 5 had a significant change in pressing angle with fatigue (**Figure 26**). This also may be due to the inertial properties of this degree of freedom on the Smith machine. With the barbell bench press, all degrees of freedom had inertia associated with the mass of 50%1RM. In the case of Smith machine mode 3, the anterior-posterior degree of freedom had inertia associated with the mass of 50%1RM, plus four 8lbs bearing sets which moved the vertical bars along this degree of freedom. It is plausible that due to the increased difficulty of accelerating and decelerating the added mass from the bearings in the degree of freedom along torso's longitudinal axis, the neuromuscular system opted for utilizing a straighter bar path even in presence of fatigue. Further, it may be that the disparity in the difficulty of control of different degrees of freedom placed an unfamiliar demand on the neuromuscular system, causing an insignificant decrease in performance (i.e. less repetitions to failure) on mode 3 vs. mode 5, despite mode 3 having less unrestricted degrees of freedom. This will be discussed in further detail later.

Directional bias may also play a role in difficulty of control of degrees of freedom. In our study, participants were instructed to maintain a consistent grip width which is 165-200% of their biacromial length. At the top of the movement, this creates a moment arm between the load and the pivoting shoulder joint in the medial-lateral axis. During the chest press, when medial-lateral motion is locked, given that the musculature acting on the scapulae are statically contracted to restrict scapular movement, the subjects can theoretically maintain the bar at the top of the concentric phase by only creating an extension torque about the elbow. When medial-lateral motion is unlocked, however, horizontal adductor muscle activation is necessary at the top of each repetition in order to control this degree of freedom. **Figure 29** shows the rubber banded, averaged pectoralis major activation throughout the concentric phase of every repetition of the participant with the median number of repetitions on modes 1 and 6. Despite the same load, pectoralis major activation is greater when the arm is fully extended at the top of the repetition during mode 6. This directional bias may be the reason for why the wrist-shoulder distance decreased with fatigue (**Figure 25**), despite participants having been given the instruction to maintain distance between their hands.

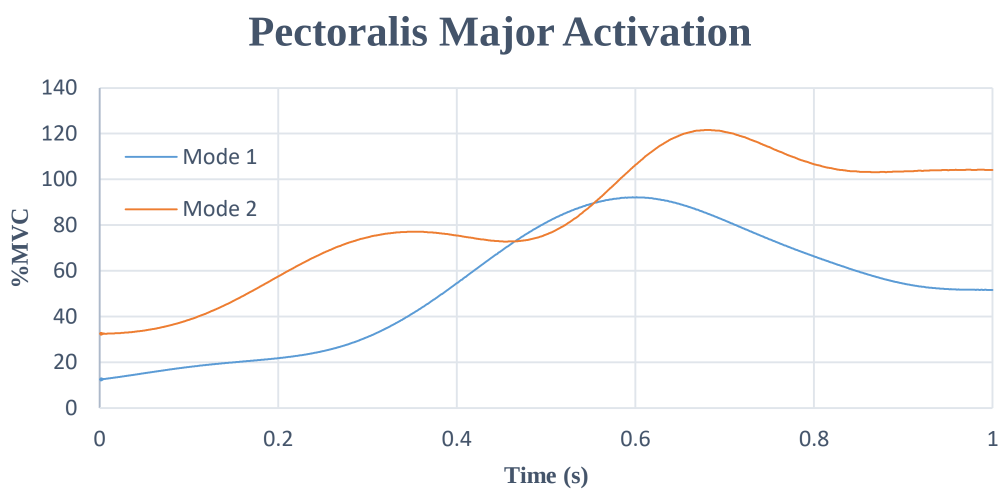


Figure 29. Rubber banded, averaged pectoralis major activation of each repetition of the participant with median number of repetitions on mode 1.

5.2 Hypothesis Revisited

5.2.1 Effort

Our first hypothesis stated that as the degrees of freedom of the load increased, the effort, as measured through initial prime mover and non-prime mover muscle activation, rate of perceived difficulty, and action tremor would also increase. However, more variables than the number of unrestricted degrees of freedom affected the overall perceived difficulty and muscle activation response. According to our findings, although there is greater antagonistic co-contraction during modes with more unrestrained degrees of freedom which creates greater fatigue, the perceived difficulty was affected more by the difficulty of control of certain degrees freedom as opposed to the total number of unrestricted degrees of freedom. Further, it appears that the perceived difficulty of control does not correlate with the number of repetitions to failure, but rather, it correlates to more altered muscle stress distribution.

When comparing modes which had uncoupled medial-lateral degree of freedom unlocked, mode 4 resulted in significantly higher initial RPDS scores than mode 6, and modes 2 and 4 resulted in significantly higher final RPDS scores than mode 6 (**Figure 21**). This is inconsistent with our hypothesis because mode 6 has more unrestrained degrees of freedom. As discussed in the previous chapter, this may be due to the medial-lateral degree of freedom having very little inertia due to the low mass of the moving handles. During these modes, the contribution of the triceps was limited due to the mechanical nature of the task; excessive elbow extension would create a moment arm between the load and the elbow joint in the medial-lateral axis. Due to low resistance to horizontal acceleration along this degree of freedom, extension at the elbow and horizontal flexion at the shoulder have to be coupled very precisely, that is to say, their angle has to change in unison to keep the load directly above the elbow joint until full elbow extension is required to “lock out” the elbows at the top of the repetition. This was reflected in EMG data as triceps’ activation was significantly lower during modes 2 and 4 than during modes where medial-lateral movement is restricted (**Figure 10**). Along with this, the pectoralis major and anterior deltoid activation were significantly higher (**Figure 10**) as moment production about the shoulder may have been the primary contributor to force production, while elbow flexors and extenders acted as “stabilizers”, contracting to control medial-lateral motion. In contrast, in mode 6, where medial-lateral motion was also unlocked, there was no significant difference in triceps’ activation from mode 1 (**Figure 10**), possibly due to the high inertia associated with the mass of the dumbbells in the medial-lateral degree of freedom. Precise coupling of elbow flexion and shoulder horizontal adduction moments isn’t required as the mass of the load isn’t very sensitive to small changes in force, which may have been a contributor to lower perceived difficulty of controlling the load. It’s important to note that during this mode, activation of the biceps was significantly higher than in all other modes (**Figure 11**). This may also be a result of controlling a degree of freedom with high inertia, where minor fluctuations of elbow moments don't result in kinematic motion. Accordingly, the final RPDS scores of mode 6 were significantly lower than on modes 2 and 4, despite mode 6 having more unrestricted degrees of freedom. According to our RPDS scores, we propose that it is not solely the number of unrestrained degrees of freedom, which creates greater antagonistic co-contraction, that makes the supine chest press task be perceived to have more or less stability, but rather, the difficulty of control of each degree of freedom, which in our study, was contributed to by a directional bias combined with low inertia**.** This difficulty of control is also evident in our action tremor data, as modes where the uncoupled medial-lateral degree of freedom is unrestrained show correlation between the action tremor amplitude of the elbow and the hand, and perceived difficulty. Interestingly, despite the tremor amplitude and perceived difficulty of control being higher in modes 2 and 4 than in mode 6, these modes still produced greater number of repetitions than mode 6. We presume that increased joint stiffness objective which increases co-contraction of potentially antagonistic muscles, which we presume limits performance, does not correlate with perceived difficulty of control.

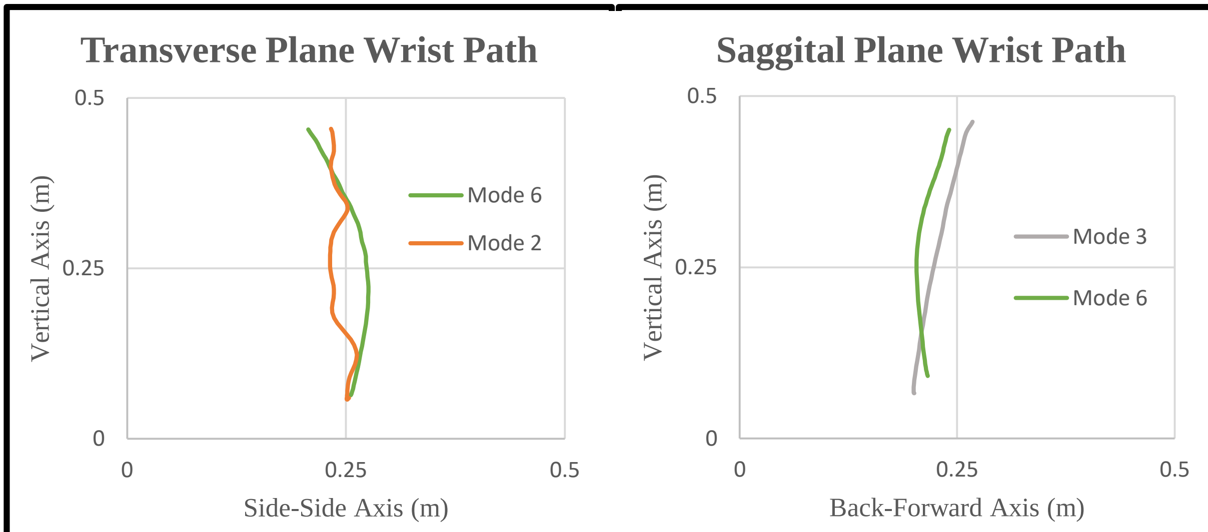
In the case of anterior-posterior degree of freedom, our RPDS scores showed significantly greater initial and final perceived instability during mode 3 than during mode 5 (**Figure 21**), despite mode 3 having less unrestrained motion. As previously discussed, this may be due to the discrepancy in inertia between the vertical and anterior-posterior degrees of freedom during mode 3. When moving along this axis in mode 3, the mass of four 8lbs bearing sets are added to the mass of the bar, creating greater inertia in this axis. Although the linear bearing had very low rolling friction, higher friction to initiate movement may have reduced the horizontal motion. As mentioned, it has been shown in the literature that a bar-path which is deviated from a straight and vertical line may be more optimal to maximal force production during the bench press. This deviation may be quantified as both deviation from a straight path and deviation in pressing angle, which is the angle formed by the initial and final positions of the barbell. In mode 3, in order to utilize this bar path, movement along the anterior-posterior degree of freedom must happen. However, it may be that because it takes greater force to accelerate and decelerate the bar along this path, it is limited, as it adds a degree of difficulty to the task. According to our findings, in contrast to mode 5, in which participants' pressing angle decreased with fatigue, in mode 3, it stays consistent (**Figure 26**). During mode 3, the neuromuscular system may need to optimize to find the most efficient amount of deviation from a straight vertical path as more deviation, although it may be beneficial, will require greater velocity along this axis and thus, will create greater momentum and more resistance to deceleration. To an individual who is well trained to perform the barbell bench press, this may create a demand to alter the muscle synergies to create a new movement path which is more optimal for the newly altered task. This resistance to acceleration is also observed in the tremor amplitude of the hand, which is the lowest among all modes. However, despite this, the elbow tremor was not significantly different from mode 5. During this mode, there was no correlation between action tremor of the load and perceived difficulty of control. It may be presumed that the perceived difficulty increases when inertial properties along different degrees of freedom change, causing an alteration in muscle stress distribution and kinematics. During modes 2, 3 and 4, EMG data shows a redistribution of the muscle stresses, as there is greater anterior deltoid activation and lower tricep activation (**Figure 10**). However, the mechanical nature of the task during mode 3 does not create a need for muscle activation redistribution. However, this may have been the result of altered difficulty of control of the anterior-posterior degree of freedom. Overall, the muscle stress redistribution appeared to be the main driver for initial RPDS scores.

Part of our first hypothesis was that antagonistic musculature activation is increased during unstable tasks as joint stiffness objective becomes a higher priority. This hypothesis was based on findings from Fischer et al (2009), who showed a positive correlation between instability at the point of force application and overall forearm musculature activation during a screwdriver pushing task. However, in their study no overall increase in muscle activation of the more distal shoulder joint and trapezius was found, suggesting that proximity of the joint to the unstable load may be a factor in the objective of the musculature acting on that joint during low intensity tasks in a non-fatigued state. Our findings are similar; our first hypothesis proved true for antagonist muscle acting on the joint closer to the unstable load. There was a significant increase in initial biceps' activation from mode 1 to all modes except mode 3 (**Figure 11**), which may be due to the fact that all medial-lateral motion was externally restricted. However, with the exception of few muscles, there was no overall increase in the antagonist muscle activation at the shoulder with increased instability (**Table 5**). This may be because during lower intensity tasks (50% 1RM) in a non-fatigued state, overall objective of the neuromuscular system is energy efficiency, and because joints which are more proximal to the unstable load are able to cancel out the micro oscillations of the load via increased neural drive to the muscles surrounding the joint, there's a reduction of need to create stability at joints more distal to the load. However, stability objective may extend to more distal joints during maximal effort tasks or during submaximal effort tasks in a fatigued state.

5.2.2 Fatigue

Part of our second hypothesis was that fatigue of the prime mover musculature, as measured with mean power frequency, will accumulate faster as the load becomes more unstable. However, given the factors discussed in the first chapter, all prime mover muscles did not get fatigued more with more instability. Rather, individual muscle fatigue was accumulated at different rates during different modes according to muscle stress redistribution. For instance, during modes 2 and 4, pectoralis major and anterior deltoid fatigue (**Figures 18, 19**) occurred at a significantly greater rate than all other modes, while triceps fatigue occurred at a significantly lower rate (**Figure 20**). This may be indicative that the muscle stresses changed so that the musculature acting on the shoulder joint may have had more load as it may have been a greater contributor to overall force production. On the contrary, during mode 6, despite medial-lateral degree of freedom being unrestrained, triceps' rate of MPF decrease did not differ significantly from mode 1 (**Figure 20**), while pectoralis major and anterior deltoid fatigue was significantly greater (**Figures 18, 19**). The disparity of triceps fatigue during modes 2 and 4 and mode 6 may be attributed to the inertial differences between the modes. It may be that when a degree of freedom has more mass and inertia associated with it, the neuromuscular system may control it with more antagonistic co-contraction, as was the case with greater biceps contraction during bench press mode 6 (**Figure 11**), which may have contributed to greater triceps fatigue as it was contracting against a greater flexion moment about the elbow. In contrast, during modes 2 and 4, when the resistance to acceleration (i.e. inertia) is very low, the neuromuscular system may control this degree of freedom via less antagonistic co-contraction.

Our second hypothesis also stated that with repetitions, modes which have more unrestrained degrees of freedom will deviate from a straight path more as muscle fatigue will be accumulated at a faster rate. This was based on findings from Duffey & Challis (2007), who showed that the barbell bar path which deviates from a straight line may be more optimal for maximal force production. This hypothesis was partly confirmed as there was a significantly greater rate of path length ratio increase from mode 1 to modes 3, 5 and 6, which indicates that deviation of the bar from a straight line does occur when motion is unrestricted, and it is amplified with fatigue. However, mode 6, which had the most unrestrained degrees of freedom, did not increase at a significantly greater rate than modes 5 and 3, although it had greater initial path length ratio (**Figure 23**). This increase in path length ratio is perceived to have resulted from a movement of the load along an arc as opposed to a straight line. This is well evidenced in the literature during the barbell bench press as the bar is described to follow a reverse C shape bar path in the sagittal plane, which is amplified with both increased load and increased fatigue. Our findings show that this reverse C shape bar path was not only present during the dumbbell bench press, but it was more exaggerated than during the barbell bench press. In addition, during the dumbbell chest press, when medial-lateral degree of freedom was unrestrained, and movement in the transverse plane was allowed, a similar arc was observed. **Figures 29 and 30**, below, show the mode 6 load path in transverse and sagittal planes during the 10th repetition of participant with the median number of repetitions on the dumbbell chest press. It may be that movement of the load along an arc is more favourable when it is permitted. Further, this arc is amplified in both maximal effort tasks (Duffey & Challis, 2007) during the barbell bench press and submaximal effort tasks in a fatigued state during the barbell and dumbbell bench press, which may be indicative that it is more optimal for force production not only with the barbell in the sagittal plane, but also with dumbbells in the transverse plane. Notably, when inertia in both planes is altered, it may alter the path of the load. Also shown in figures 22 and 23 is transverse plane wrist path of the same participant on the same repetition during mode 2, and sagittal plane wrist path during mode 3. The discrepancy in the path of the load is evident; the Smith machine replicated medial-lateral degree of freedom had little mass and inertia and thus, produced random movement variability which deviated the path from a straight line in the transverse plane. This path is very different from repetition to repetition, compared to the path of the dumbbell, which retains a similar shape with every repetition, but with an exaggerated arc. The Smith machine replicated anterior-posterior degree of freedom during mode 3 had greater inertia and produced less deviation from a straight path (**Figure 23**), although the angle of press was still significantly smaller (i.e. steeper) than mode 1 (**Figure 26**). The increase in path length ratio in the transverse plane in modes 2 and 4 may be attributed to greater random variation of the movement due to fatigue leading to decreased proprioception, where-as increase in path length ratio in the transverse plane in mode 6 may be attributed to a greater arc. This is also evidenced in **figure 28**, which shows action tremor of the hand, which attributes to a greater load path deviation, to increase at a significantly greater rate during modes 2 and 4 than during all other modes.



Figures 30 & 31: Transverse plane and sagittal plane wrist path during modes 2, 3 and 6.

When moving the wrist in space, a path along an arc may result by default as the glenohumeral and elbow joints rotate segments about a pivoting point, where-as a linear wrist path may result from a collaborative change in the angle of two joints. When selective attention isn't dedicated towards moving the endpoint of the upper limb in a linear path, movement along an arc will happen. This rationale may explain why an arc path is favoured over a linear path, and even more so during maximal effort tasks or when in a fatigued state. However, along with an increase in the path length ratio, the mean wrist-shoulder distance in both horizontal axes was decreased with fatigue for all modes which had unrestrained motion in their according axes (**Figures 24 & 25**). Mode 6, which had the most unrestricted degrees of freedom had the lowest mean wrist-shoulder distance. Although load path along an arc may be favoured for maximal force production, it may also be that reduction of the horizontal moment arm between the glenohumeral joint and the load at the top of each repetition may reduce the stress on shoulder flexor and horizontal flexor musculature and aid in less fatigue accumulation. Accordingly, the mean thoraco-humeral angle was the highest during mode 6 and was increased with fatigue on all modes except for mode 1. Even during mode 2, where back forward degree of freedom was restricted and anterior-posterior wrist-shoulder distance wasn't changed greatly with fatigue, the thoraco-humeral angle increased, possibly indicating that it doesn't happen as a consequence of decreasing the anterior-posterior wrist-shoulder distance but rather, it may be a compensatory movement strategy to redistribute the stress from fatigued muscles. Regardless, it appears that the thoraco-humeral angle change, as well as the bar path deviation, is a compensation strategy utilized with fatigue during the barbell and dumbbell bench press.

Our hypothesis that action tremor will increase at a faster rate in bench press modes with more unrestricted degrees of freedom was partially confirmed. When analyzing hand tremor, the inertia of the mass which moves with the hand must be considered as lower inertia may result in greater tremor due to a low threshold of force required to move the load. When looking at elbow tremor however, the effect of the inertia associated with the mass which moves with the hand doesn't affect the tremor amplitude increase to such a great extend, perhaps making elbow tremor a clearer representation of fatigue measured with action tremor. Generally, both hand and elbow tremor increased in amplitude with fatigue on all modes except for mode 1, although there weren't any clear trends. Mode 6 resulted in a significantly greater rate of elbow action tremor increase than all other modes (**Figure 28**), consistent with having the greatest overall rate of prime mover muscle fatigue (**Figures 18, 19, 20)**. Modes 2 and 4 had the greatest rate of hand action tremor increase (**Figure 28)**, possibly due to fatigue affecting joint displacement detection threshold, which will be discussed later in this discussion.

5.2.3 Antagonistic Co-contraction

Final part of our second hypothesis stated that antagonistic co-contraction will increase at a faster rate with fatigue during bench press modes with more degrees of freedom. Cashaback & Cluff, (2014) showed a positive correlation between intensity and antagonistic co-contraction, and between fatigue and antagonistic co-contraction during a static elbow flexion task. This was consistent with our findings as antagonistic muscle co-contraction increased with fatigue on every mode of the bench press (**Figure 16**). The co-contraction indexes, however, changed at different rates (**Figure 17**). During mode 1, the C.I. rate of change was significantly lower than during all other modes, indicating that the non-prime mover muscles, which contract as joint stiffness increases in priority, increased in activation at a smaller rate. Initially, only mode 6 showed a significant increase in C.I., despite having a lower rate of change with fatigue, which may indicate that even in the absence of fatigue, joint stiffness was a greater objective during mode 6 than during all other modes. It is well known that both peripheral muscle fatigue and demand for increased force production contribute to an increase in EMG amplitude. During our study, it may be fair to assume that peripheral muscle fatigue wasn't a great contributor to increased non-prime mover muscle neural drive as contraction intensities were not great enough to elicit a sufficient peripheral fatigue response. EMG increase, then, may be greatly contributed to by an increased neural drive to antagonistic musculature as joint stiffness objective takes priority in the presence of fatigue. This effect was amplified in modes with more unrestrained degrees of freedom as antagonistic muscle activation is increased at a greater rate with more unstable modes. In conclusion, with fatigue accumulation, more distal joints' musculature's objective may begin to shift and create more antagonistic co-contraction. Further, effects of instability on antagonistic muscle co-contraction of more distal joints during submaximal tasks in a fatigued state may be similar to that of maximal tasks' due to a similar decrease in performance with instability.

5.3 Implications

5.3.1 Experience Level and Bench Press Mode

When lifting weights with the objective of muscle hypertrophy, the total volume of work which the muscle performs is amongst the main drivers for a hypertrophic response (Klemp, et al., 2016). It is a common belief in the strength training world that free weight training is the best method to increase overall lean mass. In order to answer this question however, we must make a clear distinction between two separate training objectives; targeted muscle fibre hypertrophy vs. development of the skill associated with performing a movement pattern with the most efficiency. When the goal of training is hypertrophy, the objective is more internal; which method allows the trainee to safely create the greatest amount of work volume on a specific muscle. When the goal of training is performance (i.e. powerlifting, sport specific training), the objective is external; which method results in the ability to produce the most force against a free weight object via practice of a specific movement skill which optimizes force production while maintaining joint safety in the midst of external chaos (i.e. instability). It may be logical to presume that the two training methods are not exclusive in their benefits; free weight training with sufficient resistance and volume creates a hypertrophic response, and machine training with whole body exercises such as the Smith machine squat, despite the resistance only having one degree of freedom, still carry a great number of kinematic indeterminacies which must be controlled internally because of the great number of total available degrees of freedom at all joints in the kinetic chain. The question then becomes; which mode of training is most optimal when the objective is exclusively hypertrophy or performance.

During training cycles, the neuromuscular system adapts to imposed demands. The system which is stressed the most during a task will likely be the limiting factor, and will adapt to have better performance after recovery. This is known as the strength training principle of adaptation. Generally speaking, a higher demand on a specific system in a single bout of exercise will cause a greater adaptive response to an extent. For example, Schoenfeld and colleagues have shown that men who participated in high load strength training in the 8-12 rep range had a greater 1RM improvement on the squat and bench press than men who participated in low load strength training in the 25-35 rep range, and vice versa for muscular endurance improvements. Similar distinction may be made between training a movement skill, which may be a central function of creating muscle synergies which help to better perform a task given a specific objective, and training an individual muscle, which may be a peripheral function of increasing its cross sectional area and force generating capacity (Arabadzhiev et al, 2014). Our results, along with findings in Cotterman, Derby, & Skelly (2005) show that during unstable tasks, beginners' capacity to control unrestricted degrees of freedom with fatigue, as opposed to individual muscle fatigue, may be the limiting factor in their performance. Further, during extremely unstable tasks, which are created not only with more unrestricted degrees of freedom, but by degrees of freedom which are more difficult to control, even experienced trainees may be limited by a decrease in proprioception with fatigue, which becomes their limiting factor to maximal performance during a task.

Although the barbell bench press and squat have been shown to produce a greater 1RM than Smith machine variations of those exercises (Cotterman, Darby, & Skelly, 2005) (Saeterbakken et al. 2011), Cotterman et al, 2005 have demonstrated that in novice lifters, the Smith machine variation of the squat has been able to produce a greater 1RM than the barbell squat. In our study, we found a moderate negative correlation ($r^2 = -0.64$) between the level of experience of participants and the difference in the number of repetitions they were able to perform on the barbell bench press vs the Smith machine bench press and a weak negative correlation in the difference in the number of repetitions they able to perform on the dumbbell vs the Smith machine bench press ($r^2 = -0.24$); participants with more training experience, on average, had a greater number of repetitions to failure on the more unstable modes in relation to the most stable mode. These findings show that during both maximal effort tasks and submaximal effort tasks to fatigue, instability may be a greater detriment to novice lifters. We propose that this detriment in performance in both maximal force production tasks and fatigue tasks may be attributed to joint stiffness objective having greater weighing in the neuromuscular system's objective, causing co-contraction of muscles which create antagonist moments about our joints of interest. This may be exemplified when looking at the triceps' fatigue during mode

6; despite having the medial-lateral degree of freedom unlocked, inhibiting the contribution of triceps to force production, it still accumulated fatigue relatively fast, possibly due to greater contraction of the biceps.

Findings from Saeterbakken et al. (2011) also show that increased antagonistic co-contraction may limit force production. During a 1RM chest press on bench press modes with different stability requirements, Saeterbakken et al. (2011) found a significant difference in the maximal force production despite pectoralis major and anterior deltoid showing no significant difference in activation. In our study, the force production requirement was controlled, but we found a significant increase in pectoralis major activation from mode 1 to mode 6, with the difference to mode 5 approaching significance at p=0.098 (**Figure 10**). Also, the latissimus dorsi, which is a powerful shoulder extender, and posterior deltoid had a significant increase from mode 1 to mode 6, but not to mode 5 (**Table 5**). Although the actions of the latissimus dorsi and posterior deltoid vary based on the position of the humerus in relation to the torso, during our setup, contraction of both of these muscles creates a force vector which has a component which creates direct counter moment to our desired moments about the shoulder during a chest press, indicating that contraction of these muscles may be counterintuitive during a chest pressing task. To the extent of our literature review, latissimus dorsi and posterior deltoid activation disparity between different barbell and dumbbell bench press modes hasn't been measured during a 1RM test. During a submaximal task, we found the activation of these muscles to be lower when the load is more stable, perhaps due to joint stability objective having a lower weighing in the optimization criteria. Additionally, the rate of increase of these muscles changed at a greater rate. As mentioned, an increase in amplitude of non-prime mover muscles is unlikely to have resulted from fatigue, but rather, from an increase in activation as joint stability begins to have a greater weighing in the neuromuscular system's objective. If we extrapolate our findings to the 1RM bench press test, we can presume that during tasks which have maximum force production as the main objective, despite both of the main agonist muscles being contracted to their maximal capacity, the co-contraction of muscles which contribute counter moments may result in a lesser maximal force production with more unstable loads. Given that performance is inhibited to a greater extent in lifters with less experience, and antagonistic co-contraction is a contributor to inhibited performance, it may be that strength training experience may contribute to lesser co-contraction when handling unstable loads and when neuromuscular fatigue is present. The concept of reciprocal inhibition, which is a decrease in activation of muscles which oppose those which have the stretch reflex activated, has been well documented to improve with training (Geertsen, Lundbye-Jensen, & Nielsen, 2008) (Suzuki, et al., 2012). Further, reciprocal inhibition is a contributor to strength gains in the novice phase of strength training. However, the extent to which instability affects co-contraction may also be a variable which may be improved with training. It may be that in untrained individuals, the joint stiffness objective has a higher weighing than in trained individuals when handling unstable loads, and due to the trade-off between joint stiffness via co-contraction and force production, performance is inhibited to a larger extent in novice lifters.

During cases of extreme instability, our findings present an additional variable which may counter the previous argument. The machine replicated medial-lateral degree of freedom was harder to control than dumbbell chest press' medial-lateral degree of freedom due to lesser inertia and directional bias, as discussed previously. This was also reflected by rate of perceived difficulty scores during modes 2 and 4 vs mode 6 (**Figure 21**). There was no correlation of any great extent between the difference between the number of repetitions done on modes 2 or 4 and mode 1 to level of experience. In addition, 4 of the 6 subjects who chose to stop the test due to their inability to control this degree of freedom on modes 2 and 4, had four or more years or strength training experience. These were our most experienced subjects. Although these subjects may not have been familiar with a task of controlling extreme medial-lateral instability during a chest press, this may be indicative that during extreme instability, subjective assessment of safety may weigh into the perception of fatigue. These findings may indicate that during tasks with extremely unstable loads, perhaps the system which is being challenged to the greatest extent and thus, being the limiting factor in performance, is the neuromuscular system's ability to control unwanted degrees of freedom in the presence of fatigue. When controlling degrees of freedom which are externally unrestrained, a constant feedback loop must be created to assess the limb's position in space, and reposition it to assure movement only along the desired degrees of freedom. Proprioception, which is an awareness by the central nervous system (CNS) of the location of a limb in three-dimensional space (Zabihhosseinian, 2015), is decreased with local fatigue, as fatigue has been shown to affect shoulder's active repositioning into external rotation (Lee, Liau, Cheng, Tana, & Shiha, 2003). The shoulder joint has the lowest threshold (0.2°) for detecting joint rotations at a constant velocity of 0.3°/sec. This threshold, which is velocity dependant, is one critical factor in preventing injuries (Ahtom-Miller et al, 2009), perhaps by restraining unwanted movement. During modes 2 and 4, due to the low inertia of the handles in the medial-lateral degree of freedom, velocity of movement of elbows and hands was higher compared to other modes, as reflected by action tremor data (**Figure 27**). As fatigue accumulates, during modes 2 and 4, the accuracy of assessing limb's position in space may be lowered due to high velocity of movement. It is not completely clear, however, of the extent to which fatigue affects proprioception in trained vs. untrained individuals. Ashton-Miller and colleagues have shown that the probability of detecting a given joint rotation at the ankle is increased significantly with training. The threshold of detection, however, is not significantly changed. It is proposed that this change may not be due to an improved muscle spindle output, but rather, a change in selective attention to all joints involved in the movement. It can be said that adaptation to increase the probability of the joint positional change detection is task specific and happens centrally, as opposed to peripherally. In modes 2 and 4, trained individuals did not show a greater ability to control degrees of freedom compared to untrained individuals. During these modes, fatigue induced decrease in proprioception may have been the cause of cessation of the test due to high velocity of movement of the load which needs to be controlled.

5.3.2 Most Optimal Bench Press Mode

In application, the chest press exercise is utilized primarily to hypertrophy and strengthen the pectoralis major and/or develop a stronger chest press movement pattern. When the goal is pectoralis major hypertrophy via creating the greatest amount of work volume on the muscle, our EMG data presents a case for the barbell bench press being superior to Smith machine and dumbbell bench press. Our results showed that more unstable degrees of freedom caused greater prime mover muscle activation, but fewer repetitions to failure. When the purpose of training is creating the greatest volume of work on a muscle in a single set to elicit the greatest possible hypertrophic response, the former (greater muscle activation) may be thought of as a beneficial effect of having to control more degrees of freedom, while the latter (less repetitions to failure) may be considered detrimental. According to our data, although pectoralis major's fatigue, as measured with MPF, accumulated faster and rate of neural drive increase was greater during the dumbbell mode, during the barbell mode, the EMG amplitude reached a greater amplitude due to more repetitions completed. Similar results were seen with the anterior deltoid, where due to a greater number of repetitions completed with the barbell bench press, greater neural drive into the muscle was reached. During mode 6, triceps' activation was the lowest among the three modes due to the uncoupled nature of the task, while the barbell bench press mode was the highest due to the unrestricted coupled medial-lateral degree of freedom. Interestingly, pectoralis major and anterior deltoid fatigue had the lowest overall activation during mode 1, and reached a significantly lower final average activation.

These findings present a case for the barbell chest press being superior to the dumbbell and Smith machine chest press when the goal of exercise is muscular hypertrophy via creating a greater volume of stress on target muscles, namely the pectoralis major, anterior deltoid and triceps. The dumbbell bench press, despite creating greater pectoralis major and anterior deltoid activation at each repetition, was limited by the average number of repetitions participants were able to perform. Because untrained participants were able to perform less repetitions on the dumbbell bench press, this limitation may be amplified for them. When training with the goal of creating the greatest amount of work volume on the three prime mover muscle groups during a chest press, the bench press may be the most optimal exercise out of the three. However, our data also shows that the number of repetitions on the most unstable mode was greater in more experienced subjects. Although no data on experience on each mode of bench press was collected, it may be fair to assume that lifters with more lifting experience practiced the dumbbell bench press more. It may be that training on more unstable modes may help to elicit greater adaptations when dealing with less stable loads. Therefore, training with more unstable loads may be creating central adaptations as the skill of handing unstable loads is improved. Further, perhaps during very unstable modes of exercise, as was the case with modes 2 and 4, it is not individual muscle fatigue which is the limiting factor, but rather, an inhibited ability by the neuromuscular system to restrict unwanted motion and keep the movement along the desired path. If performance with more unstable loads is not an objective however, our findings show the barbell bench press to be optimal for targeted muscle strength and hypertrophy.

When the goal of training is isolation of the pectoralis major and anterior deltoid to the fullest possible extent, our data shows that modes which have low inertia in degrees of freedom orthogonal to the line of force application may be more optimal. When controlling degrees of freedom orthogonal to the line of force application which have very low resistance to acceleration, the muscle stress distribution was altered (**Figure 10**). Primarily, the majority of the stress was redistributed from elbow extensors to shoulder flexors. The contribution of triceps was the greatest during modes where medial-lateral degree of freedom was restricted, followed by dumbbell bench press, where resistance to acceleration was relatively high, followed by modes 2 and 4, where resistance to acceleration was very low. Accordingly, the pectoralis major and anterior deltoid activation was had the greatest activation during these modes. These modes mimic cable presses, where the only degree of freedom with substantia inertia is along the line of force application. Although the force vector will always be along the cable during a cable press, there is always a small force component pulling the end point of the cable back into the vector of force application, making degrees of freedom easier to control. However, similar to modes 2 and 4, contribution of the triceps may also be limited, making this exercise a better option if isolation of the shoulder horizontal flexors is the objective.

5.3.3 Occupational Implications

In a workplace setting, estimation of the capability of a specific percentage of a population capable of exerting forces requires comparison to population strength norms. However, if the strength data were determined in situations where a small number of degrees of freedom had to be controlled, their use in "real world" settings where larger degrees of freedom need to be controlled, may produce estimates that are misleading.

5.4 Limitations

5.4.1 Study Protocol

In order to assess how varying degrees of freedom at the point of load application affect muscle synergies and kinematic motion, we chose the supine chest press because the number of kinematic indeterminacies are mostly contributed to by the degrees of freedom at the glenohumeral and elbow joints as opposed to a full body exercise which may have a much greater number of kinematic indeterminacies contributed to by the degrees of freedom of all joints in the kinetic chain. In this study, the task may be referred to as a controlled chest pressing task, in which motion at the wrist, scapulae and torso was controlled. It is important to note that the bench press exercise, even in a competitive environment, varies in the desired or permitted amount of spine extension which may in turn dictate scapular motion. In addition, during the dumbbell bench press, subjects were instructed to keep the dumbbells parallel to each other and maintain the distance between their hands to mimic a barbell bench press but with additional unrestricted degrees of freedom. Because of the controlled nature of our task, we chose to exclude analysis of scapular, spinal and wrist motion from the scope of this study, although outside of a controlled environment, they may be a large contributor to altered movement and neuromuscular strategy when the objective of the neuromuscular system shifts. Accordingly, in this study, we analyzed movement of the upper limb in relation to the torso, the effect of this movement on the path the load travels, and the neuromuscular activity associated with the joints of the upper limb, excluding the wrist. Additionally, the bar speed, which is a variable which varies with fatigue (Duffey & Challis, 2007) and has effects on EMG data and possibly kinematics, has been controlled to diminish between subject variability caused by the difference in lifting speeds. However, similar to the movement along the medial-lateral degree of freedom occurring despite subjects having been given instructions to maintain same distance between their arms throughout the test, other minor compensatory strategies at the torso, scapulae, wrist, and tempo which have not been accounted for may have contaminated the results to an extent, although their extent is presumed to be negligible due to careful observation of completion of the task by investigators.

5.4.2 Study Design

Our study design included participants coming in for three sessions to complete a 1RM bench press test and 6 bench press fatigue protocols. The 1RM test was placed on a separate day in order to keep an equal number of fatigue protocols on the following two days. In addition, a maximal force production task may have greater effects on central fatigue than a 1-2 minute fatigue protocol. It is possible that during data collection lab sessions, residual fatigue has accumulated from one fatigue trial to the next, which may have created fatigue bias on second and third protocols. In our study design however, a Latin square sequence was used, where each sequence of modes was repeated an equal number of times with the exception of the sequence starting with mode 1, placing each one of the 6 modes 1st, 2nd, and 3rd in the collection sequence an equal number of times +/-1. This minimized the sequential bias created by having multiple fatigue protocols on the same day as every mode was done during a non fatigues state equal number of times as it was done during a potentially fatigued state. Additionally, 20-minute rest periods between trials were provided for all participants and all participants were asked about their perceived fatigue intermittently throughout the rest periods between trials and all participants noted to be fully recovered after 5-10 minutes of completion of tests.

5.4.3 Analysis

A shortcoming of our analysis method is our mixed-effect linear regression model's linearity assumption for all variables. In fact, it may have been the case that some variables changed at an exponential rate. However, this model, despite not catching potential non-linear trends, is not incorrect. Rather, it is a simplified model of the trends which occurred with fatigue. In counterargument, this model was more appropriate to use due to the within subject variability caused by variability of movement from repetition to repetition. When comparing first repetitions between modes, the difference may be contributed to by within subject variation. When using a regression line, every data point is considered and the within trial variation is minimized, giving us a clearer picture of the between mode differences. It also allows a similar analysis when the number of repetitions varies between participants and conditions.

5.5 Contributions and Future Research

The purpose of this study was to examine the effect of controlling additional degrees of freedom on effort and rate of fatigue accumulation using EMG, kinematic data, action tremor and perceived difficulty of control. In general, our hypotheses proved true; greater number of unrestricted degrees of freedom resulted in greater effort and fatigue accumulation. Due to our equipment design however, results from our study, combined with other findings in the literature on this subject, showed several trends which may have potential for further research.

Firstly, in this study, due to the similar degree of the performance disparity between the barbell and dumbbell bench press, during both a maximal force production task (Welsch, Bird, & Mayhew, 2005) (Saeterbakken, Van Den Tillaar, & Fimland, 2011), quantified in percent of maximal weight lifted, and a fatigue task, quantified in percent of number of maximal repetitions completed to failure, we made an assumption that the effect of instability on performance is similar during both maximal effort tasks and fatigue tasks. However, in our study, when comparing a stable fatigue task with 1 degree of freedom (i.e. Smith machine chest press) to a less stable fatigue task with six coupled degrees of freedom (i.e. barbell bench press), the 1 degree of freedom task showed to yield greater performance than the less stable task. This was inconsistent with findings from the literature, where a less stable barbell bench press 1RM task yielded better performance than the more stable Smith machine bench press 1RM task (Welsch, Bird, & Mayhew, 2005) (Saeterbakken, Van Den Tillaar, & Fimland, 2011). With our assumption that the difference in performance, in theory, should be similar, we attributed this difference in performance disparity to a difference in the allowed load path during the 1 degree of freedom task; our task resembled the self selected path, which may have been more optimal for performance, more closely. As mentioned previously, in order to isolate the effect which the demand for controlling additional degrees of freedom places on the neuromuscular system, the load path of the 1 degree of freedom task needs to replicate the self selected load path more closely. Because of this, we believe that more work needs to be done on comparing unstable dynamic tasks to stable dynamic tasks with similar load path in order to better understand the benefits gained from having the option to self select and change the load path, and compare it to the detriment of having to control additional degrees of freedom. We presume that during every task, both of the mentioned benefits and detriments exist, but the magnitude of each, which dictates overall performance, will vary depending on the task.

Secondly, the findings in this study, along with Fischer et al (2009), showed that at low intensities, in the absence of fatigue, the proximity of the joint to the unstable load which is being controlled may have a weighing in the objective of the neuromuscular system which acts on that joint. Further, neuromuscular fatigue of muscles acting on the joint more proximal to the load may cause the joints more distal to the load to increase stiffness via muscular co-contraction. In Fischer et al (2009), fatigue wasn't achieved and thus, elbow and shoulder musculature did not increase in activation overall. We believe this may be an area for further research on instability and muscular activation and fatigue.

Thirdly, findings from our study, along with findings in Saeterbakken et al (2011), showed a correlation between training experience and performance when dealing with loads which have more unrestrained degrees of freedom; individuals who have more strength training experience perform better with unstable loads when compared to individuals who don't have strength training experience. According to our EMG findings and EMG findings in Saeterbakken et al (2011), joint stiffness objective, which increases antagonistic co-contraction, may be a contributor to inhibited performance when the load is unstable. The weighing of this joint stiffness objective when handling unstable may vary with training experience; strength trained individuals may be able to control degrees of freedom via less antagonistic co-contraction than untrained individuals. However, this may also be a task specific skill, as there was no correlation between modes 2 and 4 (which our strength trained subjects didn't have experience with) and training experience. Regardless, the disparity in muscle activation between trained and untrained individuals when dealing with unstable loads may potentially be an area of further research.

VI. Conclusion

The purpose of this work was to investigate the effect of unrestrained degrees of freedom during a chest press activity on effort and fatigue. While some studies have compared free weight exercises to their machine counterparts, this appears to be the first work to systematically examine exercise from the perspective of task degrees of freedom. Although each degree of freedom altered did not have the same effect, general findings included: as the unrestrained degrees of freedom increased, effort required increased and participants fatigued more rapidly. The inertia associated with different degrees of freedom also contributed to difficulty of control and muscle stress redistribution. Additionally, the results showed a trend towards individuals with more strength training experience being able to perform better with unstable loads during both maximal force production tasks and fatigue tasks. The findings give insights into the effects of people exerting forces against unstable loads in strength training and occupational settings.

References

An, K., Kwak, B., Chao, E., & Morrey, B. (1984). Determination of muscle and joint forces; a new technique to solve the indeterminate problem. *Journal of Biomechanics*, 364-367.

Anderson, F. C., & Pandy, M. G. (2001). Dynamic optimization of human walking. *Journal of Biomechanical Engeneering*, 381-390.

Anderson, G., Bates, M., Cova, S., & Macdonald, R. (2008). *Foundations of Professional Personal Training.* Windsor: Human Kinetics.

Anderson, K., & Behm, D. (2004). Maintenance of EMG activity and loss of force output with instability. *Journal of Strength and Conditioning Research*, 637-640.

Arandjelovic, O. (2012). Common Variants of the Resistance Mechanism in the Smith Machine: Analysis of Mechanical Loading Characteristics and Application to Strength-Oriented and Hypertrophy-Oriented Training. *Journal of Strength and Conditioning Research*, 350-363.

Ashton-Miller, J. A., Wojtys, E. M., Huston, L. J., & Dry-Welch, D. (2001). Can proprioception really be improved by exercise? *Knee Surgery, Sports Traumatology, Arthroscopy*, 128-136.

Attebran, M., Mathiassen, S. E., & Winkel, J. (1995). Normalizing upper trapezius EMG amplitude: Comparison of ramp and constant force procedures. *Journal of Electromyography and Kinesiology*, 245-250.

Beachle, T. R., & Earle, R. W. (2008). *Essentials of Strength Training and Conditioning.* Windsor: Human Kinetics.

Beers, R. v., Haggard, P., & Wolpert, D. (2004). The role of execution noise in movement variability. *Journal of Neurophysiology*, 1050-1063.

Behm, D., & Anderson, K. (2006). he role of instability with resistance training. *Journal of Strength and Conditioning Research*, 716-722.

Benjuya, N., Melzer, I., & Kaplanski, J. (2004). Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing. *J Gerontol A Biol Sci Med Sci*, 166-171.

Bigland, B., & Lippold, O. (1954). The relation between force, velocity, and integrated electrical activity in human muscles. *Journal of Physiology*, 214-224.

Boettcher, C., Ginn, K., & Cathers, I. (2008). Standard Maximum Isometric Voluntary Contraction Tests for Normalizing Shoulder Muscle EMG. *Journal of Orthopaedic Research*, 1591-1597.

Brennecke, A., Guimaraes, T., Leone, R., Cadarci, M., Mochizuki, L., Simao, R., . . . Serrao, J. (2009). Neuromuscular activity during bench press exercise performed with and without the preexhaustion method. *Journal of Strength Training and Conditioning Research*, 1933-1940.

Brookham, R., & Dickerson, C. (2014). Empirical quantification of internal and external rotation muscular co-activation ratios in healthy shoulders. *Medical & biological engineering & computing*, 257-264.

Brookham, R., Middlebrook, E., Grewal, T., & Dickerson, C. (2011). The utility of an empirically derived co-activation ratio for muscle force prediction through optimization. *Journal of Biomechanics*, 1582-1587.

Brown, S., & Potvin, J. (2005). Constraining spine stability levels in an optimization model leads to the prediction of trunk muscle cocontraction and improved spine compression estimates. *Journal of Biomechanics*, 745-754.

Brzycki, M. (1993). Predicting a one-rep max from reps-to-failure. *Journal of Physical Education, Recreation and Dance*, 88-90.

Busse, M., Wiles, C., & van Deursen, R. (2006). Co-activation: its association with weakness and specific neurological pathology. *Journal of NeuroEngineering and Rehabilitation*, 26.

Cashaback, J., & Cluff, T. (2014). Increase in joint stability at the expense of energy efficiency correlates with force variability during fatiguing task. *Journal of Biomechanics*, 621-626.

Castelein, B., Cools, A., Parlevliet, T., & Cagnie, B. (2015). Modifying the shoulder joint position during shrugging and retraction exercises alters the activation of the medial scapular muscles. *Manual Therapy*.

Cheze, L., Dumas, R., Comtet, J., & Rumelhart, C. (2011). What is the number of independent degrees of freedom of the trapeziometacarpal joint? Preliminary in vitro results. *Computer Methods in Biomechanics and Biomedical Engineering*, 17-18.

Christopher, W. H., & Seth, L. P. (2012). Tremor: Clinical Phenomenology and Assessment Techniques. *Tremor and Other Hyperkinetic Movements*.

Churchland, M., Afshar, A., & Shenoy, K. (2006). A central source of movement variability. *Neuron*, 1085-1096.

Cotterman, M. L., Darby, L. A., & Skelly, W. A. (2005). Comparison of Muscle Force Production Using the Smith Machine and Free Weights for Bench Press and Squat Exercise. *Journal of Strength and Conditioning Research*, 169-176.

Cotterman, M. L., Darby, L. A., & Skelly, W. A. (2005). Comparison of Muscle Force Production Using the Smith Machine and Free Weights for Bench Press and Squat Exercises. *Journal of Strength Training and Conditioning Research*, 169-176.

Criswell, E. (2011). *Cram's Introduction To Surface Electromyography*. Mississauga: Jones and Barlett Publishers.

Crowninshield, R., & Brand, R. (1981). A physiologically based criterion of muscle force prediction in locomotion. *Journal of Biomechanics*, 793-801.

Deuschl, G., Bain, P., & Brin, M. (1998). Consensus statement of the Movement Disorder Society on Tremor. Ad Hoc Scientific Committee. *Movement Disorders*, 2-23.

DeVries, H. (1968). Method of evaluating of muscle fatigue and endurance from electromyographic curves. *American Journal of Physical Medicine*, 125-135.

Dimitrova, N., & Dimitrov, G. (2003). Interpretation of EMG changes with fatigue: facts, pitfalls, and fallacies. *Journal of Electromyography and Kinesiology*, 13-36.

Duffey, M., & Challis, J. (2007). Fatigue Effects on Bar Kinematics During the Bench Press. *Journal of Strength and Conditioning Research*, 556-560.

Edwards, R., & Lippold, O. (1956). The relation between force and the integrated electrical activity in fatigued muscle. *Journal of Physiology*, 677-681.

Elble, R. (2003). Characteristics of physiologic tremor in young and elderly adults. *Clinical Neurophysiology*, 624-635.

Elble, R. (2013). What is essential tremor? *Current Neurology and Neuroscience Reports*, 353.

Elble, R., Comella, C., Fahn, S., Hallett, M., Jankovic, J., Juncos, J. L., . . . Zesiewicz, T. (2014). Reliability of a new scale for essential tremor. *Movement Disorders*, 1567–1569.

Elble, R., Comella, C., Fahn, S., Hallett, M., Jankovic, J., Juncos, J., . . . Zesiewicz, T. (2012). Reliability of a new scale for essential tremor. *Movement Disorders*, 1567-1569.

Elliott, B., Wilson, G., & GK, K. (1989). A biomechanical analysis of the sticking region in the bench press. *Medicine & Science in Sports & Exercise*, 450-462.

Ervilha, U., Arendt-Nielsen, L., Duarte, M., & Graven-Nielsen, T. (2004). The effect of muscle pain on elbow flexion and coactivation tasks. *Experimental Brain Research*, 174-182.

Fahn, S., & Tolosa, E. (1993). Clinical rating scale of tremor. *Parkinson's Disease and Movement Disorders*, 271-280.

Fischer, S. L., Wells, R. P., & Dickerson, C. R. (2009). The effect of added degrees of freedom and handle type on upper limb muscle activity during simulated hand tool use. *Ergonomics*, 25-35.

Furness, P., & Jessop, J. (1977). Long-lasting increases in the tremor of human hand muscles following brief, strong effort. *Journal of Physiology*, 821-831.

Geertsen, S. S., Lundbye-Jensen, J., & Nielsen, J. B. (2008). Increased central facilitation of antagonist reciprocal inhibition at the onset of dorsiflexion following explosive strength training. *Journal of Applied Physiology*, 915-922.

Gonzalez-Badillo, J., & Sanchez-Medina, L. (2010). Movement velocity as a measure of loading intensity in resistance training. *International Journal of Sports Medicine*, 347-352.

Hammarskjold, E., & Harms-Ringdahl, K. (1992). Effect of Arm-Shoulder Fatigue on Carpenters at Work. *European Journal of Applied Physiology*, 402-409.

Harris, C., & Wolpert, D. (1998). Signal-dependent noise determines motor planning. *Nature*, 780-784.

Hatfield, D., Kraemer, W., Spiering, B., Häkkinen, K., Volek, J., Shimano, T., . . . Maresh, C. (2006). The Impact of Velocity of Movement on Performance Factors in Resistance Exercise. *The Journal of Strength and Conditioning Research*, 760-766.

Hess, C. W., & Pullman, S. L. (2012). Tremor: Clinical Phenomenology and Assessment Techniques. *Tremor and Other Hyperkinetic Movements*.

Hickok, M., Marklin, R., Nagurka, M., & Simoneau, G. (2014). Human Factors and Ergonomics Society 58th Annual Meeting. *Screwdriver Bit Head Design – Effect of Phillips, Straight, and a Hybrid Design on Torque, Axial Force, and Effort Ratio*, (pp. 1580-1584). Illinois.

Hughes, R., Bean, J., & Chaffin, D. (1995). Evaluating the effect of co-contraction in optimization models. *Journal of Biomechanics*, 875-878.

Jung, J., & Cho, W. (2015). Effects of push-up exercise on shoulder stabilizer muscle activation accoding to the grip thickness of the push-up bar. *Journal of Physical Therapy Science*, 2995-2997.

Kadam, P., & Bhalerao, S. (2010). Sample size calculation. *Int J Ayurveda Res*, 55–57.

Kajimoto, O. (2007). Development of a Method of Evaluation of Fatigue and its Economic Impacts. *Fatigue Science for Human Health*, 33-46.

Klemp, A., Dolan, C., Quiles, J. M., Blanco, R., Zoeller, R. F., Graves, B. S., & Zourdos, M. C. (2016). Volume-equated high- and low-repetition daily undulating programming strategies produce similar hypertrophy and strength adaptations. *Applied Physiology, Nutrition, and Metabolism*, 699-705.

Kohler, J., Flanagan, S., & Whiting, W. (2010). Muscle activation patterns while lifting stable and unstable loads on stable and unstable surfaces. *Journal of Strength Training and Conditioning Research*, 313-321.

Konrad, P. (2006, March). The ABC of EMG. *A Practical Introduction to Kinesiology Electromyography*. Scottsdale, Arizona, USA: Noraxon INC.

Kornecki, S., & Zschorlich, V. (1994). The nature of the stabilizing functions of skeletal muscles. *Journal of Biomechanics*, 215-225.

Kornecki, S., Kebel, A., & Siemienski, A. (2001). Muscular co-operation during joint stabilization, as reflected by EMG. *European Journal of Applied Physiology*, 453-561.

Lakie, M., Vernooij, C., Osler, C., S. A., JP, S., & RF, R. (2015). Increased gravitational force reveals the mechanical, resonant nature of physiological tremor. *Journal of Physiology*, 4411-4422.

Lametti, D., Houle, G., & Ostry, D. (2007). Control of Movement Variability and the Regulation of Limb Impedance. *Journal of Neurophysiology*, 3516-3524.

Lander, J., Bates, B., Shahill, J., & Hamill, J. (1985). A comparison between free-weight and isokinetic bench pressing. *Medicine and Science in Sports and Exercise*, 344-353.

Lee, H.-M., Liau, J.-J., Cheng, C.-K., Tana, C.-M., & Shiha, J.-T. (2003). Evaluation of shoulder proprioception following muscle fatigue. *Clinical Biomechanics*, 843-847.

Lindstrom, L., Magnusson, R., & Petersén, I. (1970). Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals. *Electromyography*, 341-356.

Lippold, O. (1960). The electromyography of fatigue. *Ergonomics*, 121-131.

Lippold, O. (2008). *Human Muscle Fatigue: Physiological Mechanisms: The tremor in fatigue.* London: Ciba Foundation Symposium.

LL, W., Evans, S., Wier, J., Housh, T., & Johnson, G. (1992). The effect of grip width on bench press performance. . *Int. J. Sport Biomechanics*, 8:1-10.

Louis, E., Ford, B., & Bismuth, B. (1998). Reliability between two observers using a protocol for diagnosing essential tremor. *Movement Disorders*, 287-293.

Lyons, T. S., McLester, J. R., Arnett, S. W., & Thoma, M. J. (2010). Specificity of Training Modalities on Upper-Body One Repetition Maximum Performance: Free Weights vs. Hammer Strength Equipment. *The Journal of Strength and Conditioning Research*, 2984-2988.

Madsen, N., & McLaughlin, T. (1984). Kinematic factors influencing performance and injury risk in the bench press. *Medical Science of Sport Exercise*, 376-381.

Marsden, C., Obeso, J., & Rothwell, J. (1983). Bening essential tremor is not a single entity In: Yahr MD, editor. Current Concepts of Parkinson's Disease and Related Disorders. *Excerpta Medica*, 31-46.

McCaw, S., & Friday, J. (1994). A comparison of muscle activity between a free weight and machine bench press. *Journal of Strength Training and Conditioning Research*, 259-264.

McCormick, E. M. (2010). *Ocean Engeneering Mechanics With Applications.* New York: Campbridge University Press.

McGill, S. (1991). Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics. *Journal of orthopaedic research : official publication of the Orthopaedic Research Society*, 91-103.

McGill, S. (2007). *Low Back Disorders; Second Edition.* Human Kinetics.

Milnet-Brown, H., & Stein, R. (1975). The relation between the surface electromyogram and muscular force. *Journal of Physiology*, 549-569.

MM, C., A, A., & KV., S. (2006). A central source of movement variability. *Neuron*, 1085–1096.

Mokhtarzadeha, H., Perraton, L., Foka, L., Muñoza, M. A., Clark, R., Pivonka, P., & Bryant, A. L. (2014). A comparison of optimisation methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings. *Journal of Biomechanics*, 2863–2868.

National Strength & Conditioning Association. (2006). Weight training exercise prescription. In E. R.W., *Essentials of Personal Training Symposium Workbook.* Lincoln: Human Kinetics.

Nussbaum, M. A., & Zhang, X. (2000). Heuristics for locating upper extremity joint centres from a reduced set of surface markers. *Human Movement Science*, 797-816.

Perreault, E., Chen, K., Trumbower, R., & Lewis, G. (2008). Interaction with compliant loads alter stretch reflex gains but not neuromuscular coordination. *Journal of Neurophysiology*, 2101-2113.

Petrofsky, J. (1979). Frequency and Amplitude Analysis of the EMG During Exercise on the Bicycle Ergometer. *European Journal of Applied Physiology*, 1-15.

Petrofsky, J., Dahms, T., & Lind, A. (1975). Power spectrum analysis of the EMG during static exercise.

Pitman Medical Ltd. (1981). *Human Muscle Fatigue; Physiological Mechanisms.* London: Pitman Medical Ltd.

Pryor, R. R., Sforzo, G. A., & King, D. L. (2011). Optiomizing Power Output by Varying Repetition Tempo. *The Journal of Strength and Conditioning Research*, 3029–3034.

Psek, J., & Cafarelli, E. (1993). Behavior of coactive muscles during fatigue. *Journal of Applied Physiology*, 170-175.

Reeves, N., Cholewicki, J., Milner, T., & Lee, A. (2008). Trunk antagonist co-activation is associated with impaired neuromuscular performance. *Experimental Brain Research*, 457-463.

Rehman, H. (2009). Diagnosis and Management of Tremor. *Archives of Internal Medicine*, 2438-2444.

Rippetoe, M. (2011). *Starting Strength.* The Aasgaard Company.

Saeterbakken, A., Van Den Tillaar, R., & Fimland, M. (2011). A comparison of muscle activity and 1-RM strength of three chest-press exercises with different stability requirements. *Journal of Sport Sciences*, 533-538.

Saito, A., & Akima, H. (2015). Neuromuscular Activation of the Vastus Intermedius Muscle during Isometric Hip Flexion. *PLoS One*.

Sandler, R., Sui, X., Church, T., Fritz, S., Beattie, P., & Blair, S. (2014). Are flexibility and muscle-strengthening activities associated with a higher risk of developing low back pain? *Journal of Science and Medicine in Sport*, 361-365.

Schick, E., Coburn, J., Brown, L., Judelson, D., Khamoui, A., Tran, T., & Uribe, B. (2010). A comparison of muscle activation between a Smith Machine and free weight bench press. *Journa of Strength and Conditioning Research*, 779-784.

Schoenfeld, B. J., Ogborn, D. I., & Krieger, J. W. (2015). Effect of Repetition Duration During Resistance Training on Muscle Hypertrophy: A Systematic Review and Meta-Analysis. *Sports Medicine*, 577–585.

Siegler, S., Lapointe, S., Nobilinit, R., & Berman, A. T. (1996). A six-degrees-of-freedom instrumented linkage for measuring the flexibility characteristics of the ankle joint complex. *Journal of Biomechanics*, 943-947.

Simpson, S., Rozeneck, R., Garhammer, J., Lacourse, M., & Storer, T. (1997). Comparison of one repetition maximums between free weight and universal machine exercises. *Journal of Strength Training and Conditioning Research*, 103-106.

Slack, P., Coulson, C., Ma, X., Pracy, P., Parmar, S., & Webster, K. (2009). The effect of operating time on surgeon's hand tremor. *European Archives of Oto-Rhino-Laryngology*, 137-141.

Stacy, M., Elble, R., Ondo, W., Wu, S., & Hulihan, J. (2007). Assessment of interrater and intrarater reliability of the Fahn-Tolosa-Marin Tremor Rating Scale in essential tremor. *Movement Disorders*, 833-838.

Stergiou, N., & Decker, L. (2011). Human movement variability, nonlinear dynamics, and pathology: is there a connection? *Human Movement Science*, 869-888.

Stokes, I. A., & Gardner-Morse, M. (2001). Lumbar spinal muscle activation synergies predicted by multi-criteria cost function. *Journal of Biomechanics*, 733-740.

Stone, M., & O'Bryant, H. (1986). *Weight Training, a Scientific Approach.* Burgess International Group.

Suzuki, M., Kirimoto, H., Onishi, H., Yamada, S., Tamaki, H., Maruyama, A., & Yamamoto, J.-i. (2012). Reciprocal changes in input–output curves of motor evoked potentials while learning motor skills. *Brain Research*, 114-123.

Tillaar, R. V., & Ettema, G. (2010). The "sticking period" in a maximum bench press. *Journal of Sports Sciences*, 529-535.

Tillaar, R. v., & Saeterbakken, A. (2014). Effect of fatigue upon performance and electromyographic activity in 6-RM bench press. *Journal of Human Kinetics*, 57-65.

Van Der Tillaar, R., & Saeterbakken, A. (2012). The Sticking Region in Three Chest-Press Exercises with Increasing Degrees of Freedom. *Journal of Strength Training and Conditioning*, 2962-2969.

Vredenbregt, J., & Raw, G. (1973). Surface electromyography in relation to force, muscle length, and endurance. In J. Desmedt, *New Developments in Electromyography and Clinical Neurophysiology, Vol 1* (pp. 607-622). Basel: Karger.

Welsch, E., Bird, M., & Mayhew, J. (2005). Electromyo-graphic activity of the pectoralis major and anterior deltoid muscles during three upper-body lifts. *Journal of Strength and Conditioning Research*, 449-452.

Willardson, J., & Bressel, E. (2004). Predicting a 10 repetition maximum for the free weight parallel squat using the 45* angled leg press. *Journal of Strength Training and Conditioning Research*, 561-567.

Willardson, J., & Bressel, E. (2004). Predicting a 10 repetition maximum for the free weight squat using the 45* angled leg press. *The Journal of Strength Training and Conditioning Research*, 561-567.

Winter, A. D. (2009). *Biomechanics and Motor Control of Human Movement.* New Jersey: John Wiley & Sons, Inc.

Wu, G., & al, e. (2005). ISB recommendation on defenition of joint coordinate systems of various joints for the reporting of human joint motion - Part II: shoulder, elbow, wrist and hand. *Journal of Biomechanics*, 981-992.

Yung, M., Bigelow, P., Hastings, D., & Wells, R. (2014). Detecting within- and between-day manifestations of neuromuscular fatigue at work; an exploratory study. *Ergonomics*, 1562-1573.

Zuniga, E., & Simons, D. (1969). Non-linear relationship between averaged electromyogram potential and muscle tension in normal subjects. *Archives of Physical Medicine and Rehabilitation*, 613-620.