Investigation of Hand Forces, Shoulder and Trunk Muscle Activation Patterns and EMG/force Ratios in Push and Pull Exertions

by

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Authors 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.

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Abstract

When designing work tasks, one goal should be to enable postures that maximize the force capabilities of the workers while minimizing the overall muscular demands; however, little is known regarding specific shoulder tissue loads during pushing and pulling. This study quantitatively evaluated the effects of direction (anterior-posterior pushing and pulling), handle height (100 cm and 150 cm), handle orientation (vertical and horizontal), included elbow angle (extended and flexed) as well as personal factors (gender, mass and stature) on hand force magnitudes, shoulder and L5/S1 joint moments, normalized mean muscle activation and electromyography (EMG)/force ratios during two-handed maximal push and pull exertions. Twelve female and twelve male volunteers performed maximal voluntary isometric contractions under 10 push and pull experimental conditions that emulated industrial tasks. Hand force magnitudes, kinematic data and bilateral EMG of seven superficial shoulder and trunk muscles were collected. Results showed that direction had the greatest influence on dependent measures. Push exertions produced the greatest forces while also reducing L5/S1 extensor moments, shoulder moments with the 150 cm height and overall muscular demands ($p < 0.0001$). The 100 cm handle height generated the greatest forces ($p < 0.0001$) and reduced muscular demands ($p < 0.05$), but were associated with greater sagittal plane moments ($p < 0.05$). Females generated, on average, 67% of male forces in addition to incurring greater muscular demands ($p < 0.05$). The flexed elbows condition in conjunction with pushing produced greater forces with reduced overall muscular demands ($p < 0.0001$). Furthermore, horizontal handle orientation caused greater resultant moments at all joints (p <. 0.05) The results have important ergonomics implications for evaluating, designing or modifying workstations, tasks or equipment towards improved task performance and the prevention of musculoskeletal injuries and associated health care costs.

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Table of Contents

List of Figures

List of Tables

List of Equations

1.0 Introduction

1.1 Pushing and Pulling

Ergonomics researchers have studied horizontal force generation, or pushing and pulling, extensively, but unanswered questions remain. Pushing and pulling tasks pervade the workplace, accounting for nearly half of all manual materials handling tasks performed (Hoozemans, van der Beek, Frings-Dresen, van der Woude, & van Dijk, 2002). Pushing and pulling may be defined as an exertion of a (hand) force by a person onto another object, in which the direction of the largest component of the resultant force is anterior or posterior to the person (Hoozemans, van der Beek, Frings-Dresen, van Dijk, & van der Woude, 1998). Pushing occurs when the force is anterior and directed away from the body and pulling occurs when the force is posterior and directed towards the body. Common industries involving pushing and pulling tasks include agriculture, nursing, mining, shipping and moving, warehousing, manufacturing and firefighting (Kumar, 1995). A number of variables are known to affect push and pull strengths aside from the exertion direction. These include gender, handle height, handle orientation and reach distance, among others. As a result of the multifactorial nature of pushing and pulling exposures, reported differences persist concerning whether pushing or pulling is more harmful to the human body and which of these actions produces greater force (Hoozemans et al., 1998; Todd, 2005). In particular, little is known regarding specific tissue loads in the shoulder, and even less about female physical exposures during pushing and pulling exertions.

1.2 Overexertion

Strength varies considerably between individuals and tasks. When pushing or pulling tasks require operators to approach or exceed their force generating capabilities, the musculoskeletal system is susceptible to overexertion injuries (Chaffin, 1975; Hoozemans et al., 1998; Todd, 2005). Common overexertion injuries include torn tendons, muscles and ligaments, which can be debilitating and lead to further injuries if appropriate administrative and/or engineering changes are not made (Chengalur, Rodgers, & Bernard, 2004). Thus, establishing strength profiles is important for human resources management and inclusive design. This enables superior selection of workers for specific work tasks, suitable job placements for incumbent workers, and recognition of force thresholds in the design of equipment and machines where strength exertion for task performance is necessary (Das & Wang, 2004). It is clear that human strength assessment is important for establishing clear guidelines towards redesigning jobs and tools in order to reduce or eliminate overexertion injuries.

1.3 Muscle loading

Several researchers have proposed that a positive relationship exists between local muscular strain and musculoskeletal disorders (Yates & Karwowski, 1992). There is also evidence that reducing muscular loading can have positive effects on the incidence of reported musculoskeletal complaints (Grant & Habes, 1997). It would thus seem practical to minimize the muscular strain by reducing the level of muscle activation; however, achieving this is not straightforward (Wesgaard & Aarås, 1985; Yates & Karwowski, 1992; Aarås, 1994). A quantitative relationship between muscle force and muscle activation is elusive as it can only be qualitatively stated that increases in muscle force are accompanied by increases in electromyography (EMG) signal amplitude (De Luca, 1997). Similarly, although decreases in the external force requirement may decrease the EMG signal amplitude, most exertions require the coordinated contraction of multiple muscles acting at multiple joints to achieve both force production and joint stabilization (Grant & Habes, 1997). Despite these known limitations, insight into how specific muscle groups interact and change with different task situations may assist engineers and ergonomists in identifying susceptible injury sites and make design recommendations to either reduce muscular stress and strain or make the most productive use of the participating muscle groups (Grant & Habes, 1997; Jongkol, 2001). One method to measure muscular efficiency is with the use of an EMG/force output ratio (Grant & Habes, 1997). However, the main difficulty in applying this technique to pushing and pulling exertions is that there is a paucity of EMG information for these tasks (Jongkol, 2006), particularly for the upper limb. Therefore, although EMG data can be useful in designing work tasks and tools, push and pull strengths and their interactions with associated muscle activation strategies remain largely undefined.

1.4 Limitations of Past Studies

Historically, much attention in the area of pushing and pulling research has focused on the low back, resulting in inadequate knowledge of the upper extremities. This raises concerns as the shoulder musculature has been suggested to operate at a higher percentage of muscle activity capacity than the trunk musculature during pulling tasks (MacKinnon & Vaughan, 2005). Additional evidence showed that participants perceive the shoulder to be the most stressed compared to the elbow and the back for these tasks (Garg & Beller, 1990). Several authors of longitudinal studies concur that pushing and pulling is associated with a greater risk of shoulder than low back complaints (Hoozemans et al., 2002; Smedley et al., 2003). These findings justify and mandate further study of the shoulder musculature when examining pushing and pulling.

In addition, a majority of the studies regarding pushing and pulling strengths have used male participants. However, due to the lower average absolute strength capabilities observed in females, female strength values should be determined and used as the reference population for work tasks (Schaefer, Boocock, Rosenberg, Jäger, & Schaub, 2007). This coincides with the frequent use of 5^{th} percentile female strength values as design thresholds (Kroemer, 1974).

In terms of external exposures resulting from pushing and pulling, many investigators have focused on hand forces measured in psychophysical studies and strength testing. To assess internal exposures of the shoulder and low back resulting from pushing and pulling, researchers have calculated net reactive forces and moments and/or compressive forces at the glenohumeral (shoulder) joint and L5/S1 intervertebral disc. However, few studies have directly examined the associated muscle activation strategies and patterns during pushing and pulling with EMG. Most studies either made use of EMG-assisted biomechanical models or examined back muscles. A major issue persists: insufficient EMG data for the shoulder muscles exists to corroborate the strength data, especially related to typical populations used in inclusive design processes.

1.5 Economic Importance of Investigating Push and Pull Tasks

Worldwide, overexertion injuries caused by pushing and pulling are widespread and expensive. In the United States alone, approximately 20 percent of overexertion injuries are attributed to pushing and pulling tasks (NIOSH, 1981). This represents five percent of all compensable occupational injuries (Kumar, 1995a). Due to the lack of automation in industrially developing countries, injury rates and associated costs to industry will be significantly greater (Todd, 2005). This is of significance as worker absences have been estimated to be 170 to 240 million work days leading to 4.6 billion dollars per year in compensation claims within the United States (Resnick and Chaffin, 1996). More specifically, shoulder injuries also affect Canadians. Throughout the last decade (1999 to 2008), there were over 5000 work-related lost time claims per year due to shoulder injuries in Ontario alone (WSIB, 2009). Further, between 1996 and 2004, MSD-related compensation claims resulted in over 3.3 billion dollars in benefit costs in the province (WSIB, 2005). Injuries resulting from overexertion due to inefficient use of muscular capabilities while pushing and pulling may be preventable if task demands are matched to worker capabilities. This will result in reduced compensation claims and associated costs towards improved worker safety, increased morale, and financial savings to the economy.

1.6 Purposes

The purposes of this research were to:

- Quantitatively evaluate the influence of gender (females and males) handle height (100 cm and 150 cm), handle orientation (vertical and horizontal) and elbow angle (fully extended and $\leq 90^{o}$) on force-, moment- and EMG-based exposure estimates during two-handed maximal voluntary isometric push and pull exertions
- Determine which conditions maximize the force producing capacity while

minimizing the demands on the shoulder and trunk musculature, as estimated by EMG/force ratios

- Extend the existing database of female and male isometric shoulder force capabilities for two-handed anterior-posterior pushing and pulling

Few studies have directly examined the muscle activation strategies of the shoulder musculature associated with two-handed maximal push and pull tasks. Measuring bilateral hand forces, kinematic data and the activity levels of selected shoulder and trunk muscles while performing exertions in postures, similar to those observed in industrial push and pull tasks, enabled the determination of hand force capability, specific mechanical loading of the shoulder and trunk muscles and EMG/force ratios. Identifying conditions that enable greater hand force magnitudes with reduced muscle activity levels will improve the ability of engineers and ergonomists to assess and quantify push and pull tasks and determine which factors will have the greatest positive impact when designing or modifying workstations, tasks or equipment. The results will also assist in the proper selection and placement of healthy and injured workers where push and pull strengths may be limiting factors. The objective was to provide guidance to enable evidence-based recommendations and preventative measures to reduce the incidence of shoulder injuries, thereby decreasing associated worker absences and health care costs.

1.7 Hypotheses

The following five hypotheses guided the experimental design:

1.7.1 Hypothesis 1

There will be statistically significant differences in hand force magnitudes, shoulder and

L5/S1 joint moments, normalized EMG values and EMG/force ratios between measured male and female data.

Rationale: Increases in body weight have been suggested to increase maximum producible push and pull forces by decreasing the required coefficient of friction to prevent slipping during pushing and pulling (Lee, Chaffin, & Parks, 1992; Kumar, Narayan, & Bacchus, 1995). Further, taller individuals may be able to exert greater maximum forces (van der Beek, Kluver, Frings-Dresen, & Hoozemans, 2000). The male population is generally heavier and taller (Pheasant & Haslegrave, 2006). Thus, it would seem likely that males would produce greater push and pull forces. Reported female strength values during push and pull tasks have been reported to be as low as 33 percent (Chengalur et al., 2004) to as much as 92.5 percent (Al-Eisawi et al., 1999) of male strength values. As a result of this difference, to complete a task of specified force generation requirements, females were expected to activate their muscles to a greater proportional extent than males (Schaefer et al., 2007).

1.7.2 Hypothesis 2

There will be statistically significant differences in hand force magnitudes, shoulder and L5/S1 joint moments normalized EMG values and EMG/force ratios during push and pull exertions.

Rationale: In psychophysical studies to determine the maximum acceptable push and pull forces, higher values for pushing than for pulling have been reported (Snook, 1978; Ciriello, Snook, & Hughes, 1993). Conversely, studies by Kumar (1995) and Kumar et al. (1995) found that maximum pull forces were up to 30 percent higher than push forces. Regarding normalized muscle activities, the erector spinae muscles during pull strength exertions were greater than those during push strength exertions. However, the normalized EMG values for the middle deltoid and trapezius muscles during push strength exertions were higher than those during pull strength exertions (Jongkol, 2006). These studies suggest that differences will exist, although it is unclear how task conditions will modify these differences.

1.7.3 Hypothesis 3

There will be statistically significant differences in hand force magnitudes, shoulder and L5/S2 joint moments, normalized EMG values and EMG/force ratios as the handle height increases from 100 cm to 150 cm.

Rationale: Based on a psychophysical study by Snook and Ciriello (1991) looking at maximum acceptable push and pull forces, participants were able to produce greater push forces at higher handle heights and greater pull forces at lower handle heights. Although maximal force occurred at the 100 cm handle height for both pushing and pulling, the 50 cm height produced greater forces than the 150 cm height in pushing, whereas in pulling, the 150 cm height generated greater forces than those at the 50 cm height (Kumar et al., 1995). For horizontal pull forces, posterior deltoid and biceps brachii EMG/force ratios reportedly increased with increasing work heights, while the triceps behaved oppositely (Grant & Habes, 1997). During pushing tasks, the EMG/force ratios of the triceps, anterior deltoid and erector spinae muscles decreased as the shoulder angle increased at maximum reach, whereas the opposite occurred for the biceps brachii muscle (Jongkol, 2001). Another study investigating push and pull tasks found that the muscle activities of the middle deltoid, trapezius and erector spinae muscles decreased with an increase in height (from knuckle to head heights) when pushing, but typically increased with an increase in height when pulling (Jongkol, 2006). Integrating these diverse results suggests that important differences may exist in the biomechanical output measures with respect to handle height.

1.7.4 Hypothesis 4

There will be statistically significant differences in hand force magnitudes, shoulder and L5/S1 joint moments, normalized EMG values and EMG/force ratios between vertical and horizontal handle orientations.

Rationale: Vertical handles enable users to adopt a neutral forearm posture, which may explain why two vertical handles are recommended for hospital meal carts and why they have been shown to reduce the pushing force when making turns (Das, Wimpee, & Das, 2002; Jensen, Nilsen, Hansen, & Westgaard, 2002). It is also thought that enabling a neutral wrist posture with the use of a vertical handle can double force exertion capacity while reducing muscle activation by as much as 80 percent (Grant & Habes, 1997). Thus, differences are expected in biomechanical outputs when performing exertions under these two conditions.

1.7.5 Hypothesis 5

There will be statistically significant differences in hand force magnitudes, shoulder and L5/S1 joint moments, normalized EMG values and EMG/force ratios between fully extended elbows and elbow angles of less than or equal to 90 degrees.

Rationale: Individuals with a larger reach reportedly exert higher push and pull forces (Chaffin & Andres, 1983; Jongkol & Das, 2004). Pushing with fully extended elbows has also been found to increase pushing capability (Ayoub & McDaniel, 1974; Chaffin & Andres, 1983; Grant & Habes, 1997). Concerning EMG/force ratios, the deltoid EMG/force ratio increased as the reach distance increased (Grant & Habes, 1997; MacKinnon & Vaughan, 2005). During one-handed submaximal pull forces, MacKinnon and Vaughan (2005) found that the trapezius muscle decreased its activity as the handle moved further away from the body, however, the erector spinae muscle activities increased with increasing reach distance. Although the directionality of the response is unclear, these studies suggest that elbow angle, and its influence on flexors and extensors of the elbow and shoulder likely modifies biomechanical exposures and demands.

2.0 Literature Review

2.1 Past Push and Pull Studies

The effects of pushing and pulling on the physiological loads on the human body have been investigated by several methods. Based on intra-abdominal pressure, maximum acceptable forces were found to be consistently greater for pulling than for pushing, indicating more mechanical stress on the low back for pushing (Davis & Stubbs, 1977). It was further suggested that tasks causing intra-truncal pressures to exceed 100 mm Hg were more likely to be associated with back injuries (Davis & Stubbs, 1976). Oxygen consumption and heart rate have also been found to be higher for pushing than for pulling (Ciriello et al., 1993). Based on these studies, it would seem that pushing is more demanding than pulling.

Several investigators have also examined the mechanical loading at the low back and shoulders during pushing and pulling. With respect to net reactive forces and moments and/or compressive forces at the shoulder and low back, Abel and Frank (1991) found low moments at both joints for pushing at low and high handle heights. van der Woude, van Koningsbruggen, Kroes, & Kingma (1995), de Looze, van Greuningen, Rebel, Kingma and Kuijer (2000) and Hoozemans et al. (2004) found that handle height significantly affected net shoulder and L5/S1 moments in pushing and pulling. Net L5/S1 moments were found to be lower for higher handle heights and for pushing. For the shoulder joint, maintaining the joints in line with the direction of the exerted force was found to minimize moments. More specific to the low back, maximal push and pull force exertions at the low handle height caused the largest L5/S1 spinal compression values (Chaffin & Andres, 1983). Conversely, Lee et al. (1991) showed that pulling resulted in greater compressive forces on the L5/S1 disc than pushing, regardless of the handle height. Furthermore, Kuijer, Hoozemans, and Frings-Dresen (2007) found that pushing at shoulder height compared to pulling at hip height led to a 52% decrease in the compression force at the low back and a 23% decrease in the shoulder moment. From the current literature, it appears that much of the attention has been on net moments. It is also clear that maximal pushing and pulling have significant effects on the mechanical loads on the low back and shoulders, which are largely dependent on force direction and handle height.

Further studies on pushing and pulling have focused on the external exposures such as hand forces. Psychophysical studies have shown maximum acceptable hand forces to be greater for pushing than for pulling (Haslam, Boocock, Lemon, & Thorpe, 2002; Resnick & Chaffin, 1996; Snook, 1978; Snook & Ciriello, 1991). Other psychophysical studies concluded no significant differences between maximum acceptable push and pull forces (Ciriello et al., 1993; Ciriello, Snook, Blick, & Wilkinson, 1990). Studies evaluating strength have shown that pull strength was greater than push strength (Das & Wang, 2004; Kumar, 1995; Kumar et al., 1995) MacKinnon (1998) examined maximum pull strengths in sitting, fixed standing and free standing. It was found that pull forces increased with decreasing vertical height and when lateral deviation from the mid-line of the body was minimal (0-20 cm). Furthermore, of the postures examined, free standing led to the largest forces while fixed standing led to the smallest forces. These studies demonstrate the inconsistencies in push and pull literature and underscore the need for further investigation.

Few studies have directly examined the associated muscle activation strategies/patterns during pushing and pulling at the shoulder level with the use of EMG. An early study examined muscle activation strategies via muscle palpations (Gaughran &

Dempster, 1956). Jongkol (2001) examined muscle activities of the biceps brachii, triceps brachii, anterior deltoid and erector spinae on the right side of the body for seated participants during push tasks alone. A more recent study examining pushing and pulling found that normalized low back muscle activities were greater during pull exertions than push exertions, while the opposite trend was observed for shoulder muscle activities (Jonkol, 2006). MacKinnon and Vaughan (2005) examined dynamic submaximal pulling and found that with increasing reach distance the erector spinae activity increased while the deltoid and trapezius muscle activities decreased. Another submaximal study by Laursen, Jensen, Németh and Sjøgaard (1998) examined pushing and pulling up to 20% of maximal voluntary force with seated participants. Other studies used indirect methods including EMG-assisted biomechanical models and focused on the back musculature (Marras, Knapik, & Ferguson, 2009; Theado, Knapik, & Marras, 2007). Although pushing and pulling have been studied for decades, it is evident that little EMG data pertaining to shoulder muscles exists to elucidate the published strength data.

Furthermore, a few questionnaire studies to determine the risk factors of shoulder and low back pain during pushing and pulling tasks have been conducted. Hoozemans et al. (2002) and Smedley et al. (2003) found that the prevalence rate ratios and incidence, respectively, were higher for shoulder complaints than back complaints. van der Beek, Frings-Dresen, van Dijk, Kemper, and Meijman (1993) also found higher odds ratios for the shoulder than the low back among lorry drivers. These studies suggest a stronger association between pushing and pulling and shoulder complaints than low back complaints. Table 1 summarizes some key outcome variables examined by several prominent investigators of push and pull exertions.

Table 1: Outcome parameters examined by leading authors of push and pull studies.

Author(s)	Research Focus	Hand Force	Direct or Indirect Measurements	Muscles (EMG)	Moments	Net Reactive Forces	Compressive or Shear Forces
Chaffin and Andres 1983	SB	Yes	Model	-	B, sagittal	-	B (compression)
Kumar, 1995	SB	Yes	Direct	-	-	-	-
Kumar et al., 1995	SB	Yes	Direct	-	-	-	-
Grant and Habes, 1997	S	Yes	Direct	Posterior deltoid; Long head of the biceps brachii; Triceps brachii; Flexor digitorum superficialis, Extensor digitorum	-	-	-
de Looze et al., 2000	SB	Yes	Model	-	SB, sagittal	-	-
Jongkol, 2001	SB	Yes	Direct	Biceps; Triceps; Anterior deltoid; Erector spinae	-	-	-
Hoozemans et al., 2004	SB	Yes	Direct, model	Longissimus; Multifidus; Iliocostalis; Lateral and anterior internal oblique; Lateral and anterior external oblique; Rectus abdominis	SB, around three axes	SB	B (compression and shear); S (compression)
Jongkol and Das, 2004	SB	Yes	Direct	-	-	-	-
MacKinnon and Vaughan, 2005	SB	No, set	Direct	Left, right erector spinae; Left and right external obliques; Latissimus dorsi; Deltoid; Trapezius; Biceps brachii	-	-	-
Jongkol, 2006	SB	Yes	Direct	Left and right erector spinae; Middle deltoid, Trapezius	-	-	-
Theado, 2007	B	Yes	Model	Left and right (Erector spinae; Latissimus dorsi; Internal obliques; External obliques; Rectus abdominis)	B, sagittal	B	-
Marras, 2009	B	No, set	Model	Left and right (Erector spinae; Latissimus dorsi; Internal obliques; External obliques; Rectus abdominis)	-	-	B (compression and shear)

***Shoulder = S, Back = B.**

2.2 Importance of Strength

Adequate strength is important for both performance and the prevention of musculoskeletal injury. Using subjective physical ability task ratings to evaluate the physical requirements across a variety of industrial occupations, it was found that muscular strength was fundamental to physical performance (Hogan, 1991). Generally, there is an increased risk of developing MSDs when exerted forces approach the maximum strength or when maximum acceptable forces are exceeded (Chaffin, 1975; Hoozemans et al., 1998). For example, when task requirements exceed the strength of the muscle groups involved, overexertion injuries such as a torn tendon, muscle, or ligament may result (Chengalur et al., 2004). Thus, human strength assessment may be useful in establishing population norms such that the design thresholds of equipment, machines and other tools requiring force exertion may accommodate the strength limitations of the intended user group.

Specific strength guidelines have been developed to reduce the risk of musculoskeletal complaints and injury. In quantifying excessive force, Putz-Anderson (1988) stated that muscles of the upper extremity are capable of maintaining a contraction level above 20 percent of their strength for only a few seconds before fatiguing. It was further recommended that workers should not be required to exert more than 30 percent of his or her maximum force for a particular muscle, in a prolonged or repetitive way (Putz-Anderson, 1988). Additionally, all muscular contractions exceeding 50 percent of their maximum force should be avoided (Putz-Anderson, 1988). In the absence of rest, prolonged and excessive static work will weaken joints, ligaments and tendons (Putz-Anderson, 1988), which could result in added compensation costs and workers being susceptible to further musculoskeletal injuries. From a biomechanical viewpoint, the assessment of static muscular strength under different task conditions is important for strength prediction and determining which conditions may be more conducive to staying within strength guidelines, towards reducing injuries related to the overexertion of force.

2.3 Importance of EMG Data

When designing work tasks, one objective should be to enable postures that maximize the force capabilities of the workers while minimizing the overall muscular demand. When there is a mismatch between the task requirements and the worker's force generating capabilities, the musculoskeletal system is vulnerable to overexertion injuries (Todd, 2005). Several investigators have suggested a relationship between local muscular strain and MSDs (Yates & Karwowski, 1992). One method of minimizing the muscular strain is to simply reduce the level of muscle activation (Aarås, 1994; Westgaard & Aarås, 1985; Yates & Karwowski, 1992). However, most exertions require the coordinated contraction of multiple muscles acting at multiple joints, where some muscles are used directly for force production and others are activated to stabilize joints or the entire body (Grant & Habes, 1997). Nevertheless, knowledge of how the contributions of specific muscle groups change with different task situations will enable engineers and ergonomists to identify susceptible injury sites and make design recommendations to alleviate muscular demands or make the most productive use of the participating muscle groups.

One approach that has been used to evaluate tasks and work situations is the EMG/force ratio. It has been suggested that the EMG/force ratio can identify conditions in which high levels of muscle activity are effectively translated into force production (Grant & Habes, 1997). Tasks are thought to be classified as acceptable when the EMG/force ratio is less than one, and unsafe and to be avoided when the ratio is greater than one. Jongkol (2001) also made use of EMG/force ratios in evaluating seated push and pull tasks under different working conditions. Although EMG/force ratios may be a useful approach to evaluating work tasks, the main concern with this method is that very little EMG information during pushing and pulling tasks are available.

EMG data may have several practical applications. One study evaluating meat cutting found that modifying the knife handle to allow the use of a stab grip with a vertical handle rather than a slice grip with a horizontal handle could potentially double force exertion capacity and reduce muscle activation by up to 80 percent (Grant & Habes, 1997). This was partially explained in that postures which cause the force vectors to be transferred from larger, stronger muscles to smaller, weaker muscles are more likely to result in overexertion injury, for a given force level (Grant & Habes, 1997). This is important because in most situations, it is impractical to reduce the force demands, however, changing the work posture or modifying the workstation is more feasible (Jongkol, 2001). In a study investigating maximal push and pull exertions, Chaffin and Andres (1983) found that one-handed push and pull strengths were about 73 percent of the two-handed strengths. As the ratios were not very close to 50 percent, these strength values are only partially dependent on arm strengths. If muscle activation was recorded, it might have provided greater insight into the muscle strategies used to generate maximal force. Although it is evident that EMG data could be useful in designing work tasks, it is apparent that the relationship between pushing and pulling and local muscle activity is still largely unclear (Hoozemans et al., 1998).

2.4 Mechanical Loading at the Shoulders

Regarding the mechanical loads associated with pushing and pulling tasks, the shoulders are arguably more affected than the low back. In a recent study, Hoozemans et al. (2004) found that cart weight and handle height significantly affected the net moment at the back, while net moments at the shoulder were also affected by the use of one or two hands. In another study examining net moments related to cart pushing and pulling, the absolute shoulder torque was significantly affected by handle height and horizontal force level, however, only the effect of horizontal force level on the absolute L5/S1 torque was significant and the effect was small (de Looze et al., 2000). Concerning muscle activity, it appears that a shoulder strategy is employed for near pulls, as greater muscle activity existed in the shoulder than the trunk. For pulls with a further handle distance, the participant might take advantage of the inertial properties of the trunk by rotating backwards at the hip or back, increasing the muscle activity of the trunk muscles such as the erector spinae, while decreasing the use of the shoulder muscles to achieve a given force level (MacKinnon & Vaughan, 2005). However, the shoulder musculature generally exhibited a higher percentage of muscle activity than the trunk musculature, regardless of reach distance (MacKinnon & Vaughan, 2005). Further, Laursen and Schibye (2002) found that when loads on the shoulder muscles were very high during pushing and pulling, loads on the low back muscles were small. These studies suggest that the mechanical load on the shoulders may be more dependent on task factors than the low back, and thus a shift of emphasis on the mechanical loads at the shoulder may be fruitful.

2.5 Shoulder Complaints

Shoulder discomfort reports often accompany high shoulder mechanical loads. A study examining the effects of pulling speed, handle height and angle of pull from the horizontal plane on one-handed dynamic pulling strength found that the shoulders were perceived as most stressed, based on ratings of perceived exertion for the elbow, shoulder and back (Garg & Beller, 1990). Abel and Frank (1991) also noted that subjects experienced discomfort in the shoulder area while pushing against a high handle. In examining physiological strains while pushing and hauling, Garcin, Cravic, Vandewalle and Monod (1996) reported that a majority of subjects complained of muscle pains in the arms and back and of articular pains in the shoulders and wrists while pushing.

Shoulder complaints are related to the degree of exposure to pushing and pulling tasks. From a review of pushing and pulling, shoulder complaints were associated with working above acromion height, twisted trunk postures, and isometric load of the shoulder muscles (Hoozemans et al., 1998). van der Beek and colleagues (1993) reported increased shoulder complaints in lorry drivers responsible for pushing and pulling wheeled cages compared to those with only a driving task. Two out of three lorry drivers in the study had low back complaints while shoulder, neck and knee complaints were reported by one third of lorry drivers (van der Beek et al., 1993). Although 45 percent of lorry drivers suffer from regular pain or stiffness in the back (24 percent middle region and 27 percent lumbar region), 26 percent of the drivers suffer from regular pain or stiffness primarily in the shoulder (Van der Beek et al., 1993). Hoozemans et al. (2002) also found similar, but substantially higher ratios, with a significant odds ratio of 2.0 (with 90 percent confidence interval 1.1 to 3.7) for regular pain or stiffness in the shoulder. Their crude odds ratio in the highly exposed group was 4.22 for high shoulder pain intensity while the corresponding back odds ratio was only 1.35. This led to the conclusion that pushing and pulling are specific risk factors for shoulder complaints and that there is a stronger association between pushing and pulling and shoulder complaints than low back complaints. Finally, they identified a trend towards a dose-response relationship for pushing or pulling and shoulder complaints.

In a collection of longitudinal data on the occurrence of neck and shoulder pain in a cohort of nurses, up to 22 percent of nurses' neck and shoulder pain was calculated to be preventable by controlling exposure to pushing and pulling at work (Smedley et al., 2003). The authors agreed with Hoozemans et al. (2002) that pushing and pulling is associated with a higher risk of shoulder than low back complaints. Ratings of perceived exertion of nurses were also higher for the shoulder region than the low back when using different types of sliding aids requiring pushing and pulling. Thus, the use of equipment requiring pushing and pulling intended to eliminate lifting and carrying may have inadvertently transferred the risk of low back problems to neck and shoulder problems. All these research results point towards the need to examine the mechanical load at the shoulders during pushing and pulling tasks as a first step towards determining how to reduce the incidence of shoulder complaints associated with their performance.

2.6 Factors Affecting Horizontal Push and Pull Strengths

Push and pull strengths are dependent on several personal and work factors including gender handle height, handle orientation, postural asymmetry, static or dynamic modes of action, reach distance, and more. This may explain the lack of consensus regarding which of these two actions is associated with the greatest horizontal force production capability. A comparison table summarizing the factors examined by several prominent studies that have investigated push and pull exertions (Table 2) provides insight into some potential causes for disagreement between studies, where they exist.

Table 2: Summary of factors examined by leading authors of push and pull studies.

Author(s)	Gender	Push or Pull	Strength or Submaximal	Hand(s)	Handle Height	Handle Orientation
Chaffin and Andres, 1983	MF	Both	Strength	Right, both	68, 109, and152 cm	Vertical
Kumar, 1995	MF	Both	Strength	Both	35, 100, and 150 cm	Vertical
Kumar et al., 1995	MF	Both	Strength	Both	50, 100, and 150 cm	Vertical
Grant and Habes, 1997	M	Pull	100, 75, and 50% of maximum effort	Right	Shoulder and elbow height	Horizontal, sagittal plane
de Looze et al., 2000	M	Both	15, 30, and 45% of total body mass	Both	Pushing (60, 70, and 80% of shoulder height); Pulling (50, 60, and 70% of shoulder height)	Horizontal, frontal plane
Jongkol, 2001	M	Push	Strength	Right	Vertical angle	Vertical
Hoozemans et al., 2004	M	Both	Initial and sustained forces	Right, both	Hip or shoulder	Vertical
Jongkol and Das, 2004	MF	Both	Strength	Right	Vertical angle	Vertical
MacKinnon and Vaughan, 2005	M	Pull	Submaximal (12% lean body mass)	Right	Elbow	Vertical
Jongkol, 2006	M	Both	Strength	Both	Knuckle, elbow, shoulder, head	Not specified
Theado, 2007	MF	Both	20, 30, and 40% of body weight	Both	Between 50 and 80% of height	Horizontal, frontal plane
Marras, 2009	MF	Push	54.5 and 145.5 kg	Both	50, 65, and 80% of height	Vertical

*M = males, F = females.

continued

Table 2 continued

Author(s)	**Reach**	**Horizontal Deviation**	**Mode**	**Stance**	**Sitting or Standing**
Chaffin and Andres, 1983	Free posture	Free posture, sagittal plane analysis	Isometric	Feet side-by-side or one foot forward	Standing
Kumar, 1995	53 cm between subject ankle and handle	Sagittal	Isometric, isokinetic	Lower extremities stabilized at hip, knees and ankle, symmetrical	Standing
Kumar et al., 1995	Pulling with arms fully extended, pushing with hands close to the body	Sagittal, 30^o and 60^o lateral deviation	Isometric, isokinetic	Symmetrical, lower extremities stabilized at hip, knees and ankle	Standing
Grant and Habes, 1997	50 and 100% of the distance from shoulder to the hand when the shoulder is flexed 90^o and the elbow is fully extended	Sagittal	Isometric	Symmetrical, feet side-by-side	Standing
de Looze et al., 2000	Varied: "most comfortable way"	Sagittal	Dynamic	Walking	Standing
Jongkol, 2001	Normal, maximum, extreme	Radial push, horizontal angle	Isometric	Symmetrical, feet side-by-side	Standing
Hoozemans et al., 2004	Not specified	No	Dynamic	Walking	Standing
Jongkol and Das, 2004	Normal, maximum, extreme	Radial push, horizontal angle	Isometric	Not specified	Both
MacKinnon and Vaughan, 2005	10, 15, 20, 25, 30, 35, and 40% of stature	Standardized asymmetrical foot orientations	Dynamic, Isokinetic	Asymmetrical, left foot forward	Standing
Jongkol, 2006	Extended arms	Horizontal angle	Isometric	Not specified	Standing
Theado, 2007	Free stride length	Sagittal	Dynamic	Free stride length	Standing
Marras, 2009	Not specified	Two conditions: No constraint; Push load through a target 15% larger than load width on each lateral edge of the load	Isokinetic	Walking	Standing

2.6.1 Gender

Most gender-related differences in pushing and pulling performance appear to be related to anthropometrics. For instance, increased body weight has been linked to increased maximum push and pull forces (Ayoub & McDaniel, 1974). Further, it was thought that taller individuals may exert greater maximum forces (van der Beek et al., 2000). Males are heavier and taller (Pheasant & Haslegrave, 2006) and produce greater push and pull forces. The supposed mechanism relates to the higher body weights, which in turn decrease the required coefficient of friction to prevent slipping during pushing and pulling (Lee et al., 1992), and has been confirmed in more recent work (Kumar et al., 1995). Stobbe (1982) believed that muscular exertions involving flexion, abduction, and rotation of the arm about the shoulder were comparatively more difficult for females due to the smaller muscle moment arms associated with the smaller average female shoulder and thoracic skeletal frame. These studies provide some insight into why gender differences exist, but it should be noted that these differences, albeit to a less extent, also exist within a given gender.

Several studies document significant differences in push and pull strength values and musculoskeletal complaints between genders. Female strength values have been reported to be as low as 33 percent (Chengalur et al., 2004) to as high as 92.5 percent (Al-Eisawi et al., 1999) of male strength values. Kumar et al. (1995) found ranges between 71 percent and 99 percent. The same trend was evident in psychophysical studies examining maximum acceptable forces where males were capable of exerting larger push or pull forces than females (Snook, 1978; Snook & Ciriello, 1974, 1991). As a result, it is expected that for a given task of specified intensity, females must activate their muscles to a greater proportion of capacity than males (Schaefer et al., 2007). With the use of a cross-sectional questionnaire survey to examine the association between exposure to pushing and pulling at work and low back and shoulder complaints, women had significantly increased prevalence rate ratios when compared to men (Hoozemans et al., 2002). Due to the lower strength capabilities inherent of females, it can be argued that female strength values should thus be used as the reference population for work tasks (Schaefer et al., 2007). This will bring about a more conservative approach that is more inclusive to the broad strengths of a population. Accordingly, for universal design, 5th percentile female strength values are often accepted as the threshold (Kroemer, 1974). Due to the inherent strength differences and resulting biomechanical loading disparities between males and females, it emphasizes the importance of investigating female populations such that they can be adequately accommodated for in design and in the workplace.

2.6.2 Force Direction

While pushing and pulling are frequently performed activities, there is much debate about which of these two tasks are more detrimental to the human body and for which action humans can generate greater force (Hoozemans et al., 1998; Todd, 2005). In a psychophysical study to determine the maximum acceptable initial and sustained forces that can be pushed or pulled 15.2 m, the force difference between pushing and pulling were statistically insignificant, although pull forces were 13 percent and 20 percent lower (Ciriello et al., 1993). Higher values for pushing than for pulling have also been reported by Snook (1978), Warwick et al. (1980) and van der Beek et al. (2000). van der Beek et al. (2000) suggested that postural constraints due to space restrictions may partly explain why the push strength was higher than the pull strength. Other studies have found push strength to be less than pull strength. Kumar (1995) and Kumar et al. (1995) found that maximum pull forces were up to 30 percent higher than push forces, but it should be noted that in these studies, the lower extremities were stabilized. Thus, the results may be more representative of upperbody push-pull strength instead of whole-body push-pull strength. However, Das and Wang (2004) also found that push strength was on average 71 percent of pull strength while Al-Eisawi et al. (1999) found values of about 93.5 percent.

Aside from strength values, other outcome variables lack consensus when considering differences between pushing and pulling. With the use of a single muscle equivalent (either grouped as the back muscles or the abdominal muscles) biomechanical model, pulling tasks caused about twice as much low back compressive forces as pushing tasks (Lee, 1982). Further, the rate of increase in compressive force with increase in body weight was greater in pulling than in pushing (Lee et al., 1991). The results of de Looze et al. (1995) verified that peak compression and shear forces were higher during pulling than pushing. However, based on IAP measurements, acceptable forces were always greater for pulling than for pushing, indicating less mechanical stress on the low back during pulling (Davis & Stubbs, 1977). In addition, when examining oxygen consumption and heart rate, these measures were significantly higher for pushing than for pulling (Ciriello et al., 1993). Furthermore, normalized muscle activities in the left and right erector spinae during maximum pull exertions were greater than those during maximum push exertions, while the middle deltoid and trapezius showed the opposite trend (Jongkol, 2006). Thus, when pushing and pulling tasks are compared in terms of strength values and effects to the human body, the results are inconsistent and limited, particularly for the arms and shoulders.

2.6.3 Hand Usage

It is well established that two-handed strength is greater than one-handed strength. Chaffin and Andres (1983) found that the one-handed push and pull strengths averaged about 73 percent of the two-handed strength values. Since the one-handed values were significantly more than 50 percent of the two-handed values, it was concluded that push and pull force capability is only partially dependent on arm strength. It was suggested that handle height, and thus posture, was also an important factor as the one-hand push forces were approximately 65 percent (closer to 50 percent) of the two-hand push forces when pushing above shoulder height (Chaffin & Andres, 1983; Pinder et al., 1995). Placing a cart in front of the operator and pushing forwards with both hands resulted in the least reactive forces and moments at the shoulder and L5/S1 joints or compressive forces at the L5/S1 disc in a recent study (Jung et al., 2005). Increasing levels of asymmetry such as the use of only one hand, horizontal deviation or staggered foot positions, show increased levels of co-activity across several pushing and pulling conditions (MacKinnon & Vaughan, 2005). Asymmetric postures, specifically one-handed pulling, have also led to higher compression forces on the L5/S1 disc (Hoozemans et al., 1998).

When determining push and pull strength values, it is important to determine how these tasks are performed in industry. Through observation, it was found that postal workers used two hands for initial and sustained pushing as well as initial pulling, while one-handed exertions were only performed for sustained pulling (Hoozemans, Slaghuis, Faber, & van Dieën, 2007; van der Beek et al., 2000). Flight attendants also use both hands when pushing and pulling trolleys aboard aircraft (Jäger et al., 2007). Thus, it is evident that two-handed push and pull exertions are more common and critical for strength-limiting tasks and should be the research area of interest.

2.6.4 Handle Height

Handle height significantly affects push and pull capability. It was found that the height at which push and pull forces were applied had a considerable influence on maximal force output (Ayoub & McDaniel, 1974; Chaffin & Andres, 1983; Kroemer, 1974; Martin & Chaffin, 1972). Another study indicated that handle height was responsible for up to 22 percent of the strength variation among men and up to 25 percent among women (Kumar et al., 1995). There appears to be a general consensus regarding the height at which push and pull capability is highest with several past studies having shown push and pull capability to be the greatest when the point of force application is between shoulder and hip height, or between 91 cm and 115 cm (Al-Eisawi et al., 1999; Ayoub & McDaniel, 1974; Chaffin & Andres, 1983; Jung, Haight, & Freivalds, 2005; Kumar, 1995; Kumar et al., 1995; Lee, Chaffin, Herrin, & Waikar, 1991; MacKinnon, 1998; Snook & Ciriello, 1991; Warwick, Novak, & Schultz, 1980). It is believed that as the handle height approaches the height of the individual's centre of mass, the stability of the human-machine interface improves, allowing for greater horizontal pull force production (MacKinnon, 1998). Preferred height for pushing was also found to be around 0.90 m and 1.10 m for the middle 90th percentile of British male and female populations (Abel & Frank, 1991). Subjects reported shoulder discomfort at higher handle heights (Abel & Frank, 1991). Thus, there may be optimal work heights that enable maximal push and pull force exertion, which should be considered when conducting push and pull investigations as well as in work design.

Specific trends in strength with height changes have also been identified. Snook and

Ciriello (1991) carried out a psychophysical study and found that individuals produce greater push forces at higher handle heights, whereas greater pull forces are generated at lower handle heights. Generally, lower handle height locations are associated with greater pull forces, regardless of whether one or two hands are utilized (Ayoub & McDaniel, 1974; Chaffin & Andres, 1983; MacKinnon, 1998; Pheasant, Grieve, Rubin, & Thompson, 1982; Snook, 1978; Warwick et al., 1980). In investigating 50 cm, 100 cm and 150 cm handle heights, maximal force occurred at the 100 cm handle height for both pushing and pulling (Kumar et al., 1995). The lowest push forces resulted at the 150 cm height while the lowest pull forces were recorded at the 50 cm height (Kumar et al., 1995). Al-Eisawi et al. (1999) also showed that there is an interaction effect between cart load and handle height where lower initial forces are applied at higher handle heights for heavier cart loads (181 kg) in pushing and pulling. When recommending handle heights, significant handle height and direction interaction effects must also be considered.

In addition to its effect on strength capability, handle height appears to have a significant effect on muscle activity patterns. At higher handle heights, elbow extension strength may be the limiting factor in determining push strength (Chaffin & Andres, 1983). At lower handle heights, the elbow strength requirement was not as demanding. It was shown that elevation of the arm can increase levels of co-contraction within the musculature of the shoulder complex, which would decrease the ability of the individual to apply force in a given direction (MacKinnon, 1998). It was also suggested by Ayoub and McDaniel (1974) that the optimum handle height should be as low as possible, and the distances between the feet as large as possible, to delay the onset of fatigue. As the handle height increased for horizontal exertions, the posterior deltoid and biceps brachii muscle activities increased relative to force production, while the triceps muscle showed the reverse trend (Grant & Habes, 1997). In another study examining muscle activation patterns, the EMG/force ratios of the triceps, anterior deltoid and erector spinae muscles decreased as the vertical angle increased from the horizontal plane at elbow height for normal reaches and shoulder height for maximum and extreme reaches (Jongkol, 2001). The biceps brachii muscle showed the opposite pattern (Jongkol, 2001). Normalized EMG of the left and right erector spinae, middle deltoid and trapezius muscles were shown to decrease with an increase in height (from knuckle to head height) when pushing, but generally increase with an increase in height when pulling (Jongkol, 2006). It was also found that the effect of exertion height on muscle activities in these four muscles was greater than that of horizontal angle, which was defined as the vertical plane at the right side of the body, rotating medially (Jongkol, 2006). Finally, it has been reported that the more horizontally directed force direction at increasing handle heights results in a lower resultant force (reduced off-axis forces) and associated co-contraction (de Looze et al., 2000).

Safety must also be taken into account when recommending handle heights. For example, it was expected that a handle height lower than 91 cm could achieve greater pull force capability, however, a low squatting posture could cause the person to be thrown backwards or forwards if the person were to slip or if the object suddenly moved (Chaffin & Andres, 1983). In addition, a lower handle height in pushing and pulling has been found to create the largest L5/S1 spinal compression forces, averaging 3600 N, which is slightly above the NIOSH action limit of 3400 N (Chaffin & Andres, 1983). Lee et al. (1991) also showed that the compressive force on the L5/S1 disc increased as the handle height decreased when pulling, but that the compressive forces were not affected by handle height when pushing. The authors further suggested that the handle height be 150 cm for pulling and 100 cm for pushing with respect to the compression force at L5/S1 and reducing the slip potential (Lee et al., 1991). Furthermore, a handle height at waist level resulted in the highest push and pull forces while allowing the intra-abdominal pressure (IAP) to remain within a safe level (12.0 kPa) (Hoozemans et al., 1998). While most investigators examining the effects of handle height on strength acknowledge that variation in handle height affects posture, which in turn affects strength, the decision to examine fixed handle heights, rather than heights defined by body landmarks, is justified in that the measured forces are more applicable to tasks than postures (Pinder, Wilkinson, & Grieve, 1995).

2.6.5 Handle Orientation

The orientation of handles in pushing and pulling is not standardized, but may have important strength and muscle demand implications. Although industrial activities often involve horizontal handles with pronated forearms (Kumar et al., 1995), two vertical handles for four-wheeled carts (Das et al., 2002; Jensen et al., 2002) and vertical handles with horizontal handles for two-wheeled hand trucks (Mack, Haslegrave, & Gray, 1995) have shown to provide better maneuverability. Vertical handles have also shown to reduce the pushing force when turning at a corner (Jensen et al., 2002). In addition, when the arms and hands maintain a neutral posture, it has been shown to decrease steering errors by 44 percent (Wissenden & Evans, 2000). In examining meat cutting, it was shown that the biceps brachii EMG/force ratio increased with handle height, regardless of handle orientation (Grant & Habes, 1997). However, it was estimated that modifying the knife handle to allow the use of a stab grip to promote a neutral wrist posture (vertical handle) rather than a slice grip, which causes ulnar deviation (hoizontal handle), could double force exertion capacity and reduce muscle activation by as much as 80 percent (Grant & Habes, 1997). The ability to adopt a neutral forearm posture may explain why two vertical handles were recommended for hospital meal carts (Das et al., 2002). Thus, further investigation of the differences in force producing capacity and muscle activities with different handle orientations is warranted.

2.6.6 Reach Distance

Reach distance plays a significant role in defining push and pull force capability. Although, the choice to maintain a flexed or extended elbow is often given to the operator, it has been found that individuals with a larger reach can achieve higher push or pull force capability (Chaffin & Andres, 1983). Greater reach distances extending beyond the maximum reach envelope such that trunk movement is necessary have also been found to increase push and pull strength (Jongkol & Das, 2004). Locked elbows have also been shown to enhance pushing capability and it is thought that maintaining extended elbows prevents the exertion from being limited by arm strength (Ayoub & McDaniel, 1974; Chaffin & Andres, 1983). One study found the opposite trend in that pulling forces of standing participants decreased with increasing reach distance, however, pulling forces did increase with increasing reach distance when seated (Das & Wang, 2004). Grant and Habes (1997) also found that the full reach posture in simulated meat cutting did not maximize the strength of the elbow flexors or extensors, but found that the posture stabilized the elbow, which could allow the transmission of higher forces to the hand from stronger shoulder and/or back muscles.

Reach distance has also been examined to determine its effects on the resulting muscle activities of the shoulder and trunk muscles. For the posterior deltoid muscle, muscle activity increased as the handle height increased and the reach distance decreased (Grant & Habes, 1997; MacKinnon & Vaughan, 2005). MacKinnon and Vaughan (2005) also found that the deltoid and right trapezius muscle activities increased as the handle moved closer towards the body. On the other hand, the left and right erector spinae muscle activities increased significantly as the reach distance increased (MacKinnon & Vaughan, 2005). This is related to the greater flexion moment and increased range of motion the trunk experiences with greater forward reaching (MacKinnon & Vaughan, 2005). From the data on muscle activation patterns during pulling, it appears that a shoulder strategy is employed for near pulls as the muscles surrounding the shoulder joint are very active, while limited erector spinae muscle activity is observed. This limited activation may result since these muscles are not in a desirable orientation to create a sufficient extensor moment and contribute to pull forces to a large extent. Finally, as the reach distance increases, a trunk motion strategy appears to prevail, reducing the relative activity of the muscles around the shoulder joint (MacKinnon & Vaughan, 2005). There is good evidence that extended elbows will not only increase the push or pull force capability, but also reduce the level of activation of the shoulder muscles.

2.6.7 Horizontal Deviation

Several investigations have found negative implications of horizontal deviations while pushing and pulling. The sagittal plane in front of the active shoulder has been shown to be biomechanically favourable for the extensor and flexor muscles around the shoulder in generating maximal horizontal forces (Das & Wang, 2004; MacKinnon, 1998). Moreover, several researchers have reported that horizontal deviation, or postural asymmetry, has been associated with reduced strength. Jongkol and Das (2004) found that increasing external rotation from the sagittal plane resulted in significantly decreased push and pull strengths. The same finding was evident as the horizontal distance of the handle increased away from the mid-line of the body (MacKinnon, 1998). When measuring one-handed isometric push strength and the corresponding muscle activities in standing, the EMG/force ratio of the anterior deltoid and erector spinae muscles tended to increase with increasing horizontal angle (Jongkol, 2001). Further, shoulder complaints have been associated with twisted working postures, which may be characteristic of some dynamic pushing and pulling tasks (Hoozemans et al., 1998). Postural asymmetry has been shown to be responsible for up to 40 percent of push and pull strength variation among men and women (Kumar et al., 1995). Therefore, if the goal is to determine maximum push and pull force, postural asymmetry of the upper extremity should be minimized.

2.6.8 Mode (Static or Dynamic)

Strength is dependent on the mode of pushing or pulling. There are two common modes of pushing and pulling; one where the object is not moved (static) and one where the object is moved (dynamic) (Todd, 2005). Isometric strength is often evaluated during the static condition, while isokinetic strength is often examined in the dynamic condition. It has been shown that isokinetic push and pull strengths were significantly lower than isometric push and pull strengths (Garg & Beller, 1990; Kumar, 1991; Kumar, 1995) with the difference in strength values between the isometric and isokinetic modes of exertion ranging between 10 percent and 20 percent (Kumar et al., 1995). This may explain why force exertion guidelines are often lower than reported isometric strengths, as dynamic force exertions require more complex balance maintenance strategies in addition to strength. However, isometric muscle strengths of the muscles involved in pushing and pulling can provide an indication of the capabilities available for these tasks (Chengalur et al., 2004). Thus, upon evaluating strength, it is important to determine the mode in which push and pull tasks will be performed.

There has been some debate whether the use of static strength data can be applied to dynamic task situations. Todd (2005) stated that workers in all industries are seldom required to exert static push and pull forces in a single plane, however, Hoozemans et al. (1998) cited several studies that found that shoulder complaints are associated with isometric loading of the shoulder muscles, which frequently occurs in pushing and pulling tasks. In a study where subjects pushed and pulled against a stationary bar or movable cart at various handle heights while walking on a treadmill, the results obtained using the stationary bar and moveable cart were comparable in terms of force exertion and body posture (de Looze et al., 2000). In addition, a study examining the dynamic task of meat cutting simulated the task using static forces (Grant & Habes, 1997). It was assumed that the cuts were relatively short (posture does not change dramatically) and are performed slowly (i.e. the meat would resist the worker's motion). Furthermore, Resnick and Chaffin (1995) indicated that for movements where high accelerations are not present, static approximations may be useful predictors of peak pushing forces. From several studies, it is evident that pushing and pulling tasks can cause isometric loading of the shoulder muscles and that push and pull strength values may be estimated using static forces.

2.6.9 Foot Stance

Multiple studies have suggested that foot stance significantly affects push and pull strength.

Several foot placements are possible, each with differing effects on push and pull strength. In general, dictating foot positions decreased pull strength by about 36 percent when compared to freely chosen foot positions (MacKinnon, 2002). Another study investigating foot postures found that free standing postures yielded the largest horizontal forces while the fixed posture generated the smallest horizontal forces (MacKinnon, 1998). MacKinnon (1998) also stated that enabling the individual to freely select their base of support facilitates the best coordinated muscular synergism to create a pull force while standing. When given this choice, it was observed that when large horizontal forces were required, participants chose a large base of support, straddled the legs and centred the body relative to the direction of the applied force. Regardless of the foot positions, the choice to dictate foot posture should be decided based on the purpose of the experiment. If the purpose of the study is to determine maximal forces, then subjects should be able to freely choose foot positions. However, if the purpose is to compare spinal kinematics between subjects or precise kinematic measurements are required to resolve internal forces and moments, then measurement of foot positions must be strictly controlled (MacKinnon, 2002).

Staggered foot positions appear to enable greater push and pull force generation. Staggering the feet often results in increased relative handle height, though Pheasant et al. (1982) showed that at a given height, strength in the sagittal plane is strongly related to foot positions. Kroemer (1969) showed that bracing of one foot or use of the back (rather than the hands) to exert force further increased the push force capabilities. Chaffin and Andres (1983) found that when the feet are required to be positioned directly beside each other, push and pull values were not significantly different, however, when the feet were separated, males in particular were capable of significantly greater pushing strength than pulling strength. When the feet were separated during pushing trials, participants chose a rearward foot position moving further away from the handle. This allowed subjects to lean forward more, pivoting about the rearward foot and enabling the forward leg as both an additional weight to increase the forward turning moment, and to "catch themselves" if a slip occurred. Likewise, during the pulling trials, the participants moved their forward foot closer to the handle, which allowed subjects to lean backward, pivoting about their forward foot and to increase the rearward turning moment. From this viewpoint, it could be argued that staggering the feet when pushing or pulling is safer for the operator. Ayoub and McDaniel (1974) found that the two-handed pull force increased as the subject positioned the leading foot closer to the frontal plane in line with the handle, implying that foot position was critical for biomechanical efficiency. The authors also stated that foot distances should be as large as possible to delay the onset of fatigue. Thus, these examples confirm that the recommendation to position the feet side by side as a means to increase pushing or pulling strength appears to be incorrect.

The decision to select or control foot positions has a significant effect on the experimental protocol adapted. In a study by MacKinnon (2002), foot positions were critical to the point where subjects in a study investigating single-handed pulling were screened for right-hand dominance and a preference for placing the left foot in front of the right foot when exerting the force. This decision was based on the finding that maximal forces are generated when the left foot is in front for right-handed pull exertions (Daams, 1993). In other studies, structures were built for bracing body parts and providing anchorage and support for the feet and legs in order to maximize force output, with the objective being to determine the maximal force that could be generated in push and pull tasks using the entire body without chance of slippage (Kroemer 1974; Kroemer and Robinson, 1971). However, these conditions of testing are often found only in the laboratory. As a result, tolerance guidelines developed from strength values determined in this manner may increase the possibility of overexertion and/or accelerate precipitation of injuries (Kumar et al., 1995). From the findings above, it is evident that foot positions significantly affects push and pull strength, however, defining foot positions is only warranted when the standardization procedure does not significantly affect the manner in which the task is normally performed outside the laboratory (MacKinnon, 2002).

2.6.10 Whole Body Posture

A standing work posture may be more effective in generating push and pull forces compared to sitting. Many push and pull tasks are carried out while standing as most tasks are dynamic in nature and require an object to be moved. Standing not only increases the reach envelope, but in changing from a seated to a standing posture, it could change the functional muscular synergies that produce the push and pull forces. When adopting a standing posture, larger muscle groups, foot-floor friction or inertial properties may be used to assist in generating the horizontal force. However, smaller muscle groups may have to be recruited to a greater extent to generate the same force level when exerting force from a seated position (MacKinnon, 1998). However, minimal differences in female shoulder strength were found between standing and seated positions for push forces at or above shoulder height (Chow & Dickerson, 2009). Furthermore, from a safety perspective, standing (especially with staggered feet) while pushing or pulling provides the operator the increased ability to "catch themselves," should a slip occur. It is apparent that a standing posture is able to generate greater push or pull forces and may be a safer alternative to seated work postures.

3.0 Methodology

3.1 Participants

Twelve female and twelve male right-hand dominant university-aged volunteers participated (Table 3). Participant exclusion criteria included a history of shoulder or back surgery, upper extremity or low back disorder within the past year, chronic musculoskeletal or cardiovascular disorders, or an allergy to isopropyl alcohol, electrode gel, or tape adhesive. The Office of Research Ethics approved the study and participants gave informed consent. All participants wore rubber-soled shoes so that maximal push and pull forces were not limited by foot slip. To meet design thresholds for hand strength data and to enable results to be comparable to the general population, participants were recruited to cover the male and female North American 5^{th} to 95^{th} percentiles with respect to stature, where percentiles were calculated using Equation 1 and human data by Robinette et al. (2002). It was assumed that anthropometric data were normally distributed. Calculated male and female North American (Canada and U.S.A.) 5^{th} to 95^{th} percentile ranges were determined to be 165 cm to 191 cm and 152 cm to 176 cm, respectively. Actual male and female statures reported for this study ranged from about the 4th to the 97.5^{th} percentiles and the 0.5^{th} to the 99^{th} percentiles, respectively.

Table 3: Participant data (n = 24).

	Mean		Range	
	Males	Females	Males	Females
Age (years)	22 (2.6)	23 (2.2)	18-26	21-28
Stature (cm)	180.1 (9.4)	165.8 (12.2)	164-193	145.5-182
Body mass (kg)	76.8 (11.3)	62.8 (8.8)	55-92	47.1-73.4
Functional arm reach (cm)	65.0 (5.2)	60.1 (5.0)	57-71	54-71

Standard deviations are reported in brackets.

$$p = m + kS \tag{1}$$

where p = percentile
m = mean
k = factor for desired percentile
S = standard deviation

3.2 Experimental Conditions

Force, EMG and kinematic data were collected for all trials. The study incorporated five independent variables, each with two levels:

1. Gender: male and female
2. Direction: anterior-posterior pushing and pulling
3. Handle height: 100 cm and 150 cm, measuring from the floor
4. Handle orientation: vertical and horizontal
5. Elbow angle: fully extended and $\leq 90^{o}$

Elbow angles were only examined for a subset of conditions. This decision was made to limit the number of maximal contractions performed by the participants in an effort to reduce fatigue likelihood. A total of 10 two-handed experimental conditions or maximal voluntary isometric contractions (MVICs) were investigated to examine maximal forces (Table 4). MVIC is used to refer to the biomechanical domain in terms of force or torque performance during a maximal voluntary contraction (Mathiassen, Winkel, & Hägg, 1995). Each MVIC was performed twice generating a total of 20 MVICs. Dependent variables included hand force magnitudes, moments, normalized EMG values,EMG/force ratios and foot stance calculated for each MVIC.

Table 4: Experimental conditions.

	Direction		Handle Height		Handle Orientation		Elbow Angle	
Condition	Push	Pull	100 cm	150 cm	Horizontal	Vertical	Fully Extended	≤ 90^{o}
1	✓			✓	✓		✓	
2		✓		✓	✓		✓	
3	✓		✓		✓		✓	
4		✓	✓		✓		✓	
5	✓			✓		✓	✓	
6		✓		✓		✓	✓	
7	✓		✓			✓	✓	
8		✓	✓			✓	✓	
9	✓		✓			✓		✓
10		✓	✓			✓		✓

3.3 Equipment

3.3.1 Surface Electrodes

Fourteen pairs of bipolar Ag-AgCl surface electrodes with a fixed inter-electrode spacing of 2 cm (Noraxon, USA Inc., Arizona, USA) were placed bilaterally on the skin over seven superficial shoulder and trunk muscles (i.e., a total of 14 muscles) as per Zipp (1982), McGill (1992) and Boettcher et al. (2008) (Table 5). One additional electrode was placed on the clavicle as a ground electrode (Zipp, 1982). Prior to electrode placement, the skin over the muscle was prepared by shaving any hair and/or dead skin and wiping with isopropyl alcohol. This was done to enhance the signal by minimizing the impedance. A new disposable razor was used for each participant. Each pair of bipolar electrodes was connected to a 16-channel Noraxon Telemyo 2400R T2 EMG wireless transmitter (Noraxon, USA Inc., Arizona, USA). All EMG signals were A/D converted using a 16-bit A/D card with a ± 2.5 V range. Raw EMG signals had analog band pass filters set at 10–500 Hz and were differentially amplified (common-mode rejection ratio > 90 dB at 60 Hz, input impedance of 2 MΩ, hard gain set at 500) to produce maximum signal amplification. Signals were sampled at 1500 Hz to satisfy the Nyquist theorem (Smith, 2003).

Table 5: Location of surface electrode placements.

Muscle	Electrode Placement
Biceps Brachii	About 1/3 proximal to the cubital fossa, on the lead line connecting the acromion to the cubital fossa.
Long Head of Triceps Brachii	About 4 fingerbreadths distal to the posterior axillary fold, with arm abducted 90^{o}, parallel to the line connecting the acromion to the olecranon.
Pectoralis Major (clavicular insertion)	3.5 cm medial to the anterior axillary line, on the lead line connecting the anterior surface of the medial half of the clavicle to the lateral lip of the intertubercular groove of the humerus.
Middle Deltoid	Intersection of the midpoint between the anterior and posterior deltoid muscles, midway on the lead line between the acromion and deltoid tuberosity.
Middle Trapezius	2 cm medial to and above the trigonum spinae scapulae, parallel to muscle fibres.
Erector Spinae	3 cm lateral to L3 spinous process, parallel to muscle fibres.
Rectus Abdominis	3 cm lateral to the umbilicus, parallel to muscle fibres.

3.3.2 Force Transducers

Forces at the hand were measured using two AMTI six degrees of freedom transducers (MC3-A-500, AMTI, MA, USA), with the sampling rate set at 1500 Hz in synchrony with the EMG, through VICON Nexus 1.4 software (Oxford Metrics Group Plc, Oxford, UK). A custom-built cylindrical handle was instrumented to each transducer to allow participants to exert forces simultaneously and independently. A photograph of the handle setup is shown in Figure 1. The transducers were mounted 40 cm apart onto a metal pole with height-adjustable capabilities (Das et al., 2002). The handles were made of steel and were further wrapped with hockey tape. Each handle had a diameter of 2.8 cm which permitted participants to completely wrap their fingers around the handles using a power grip (Chaffin, 1975).

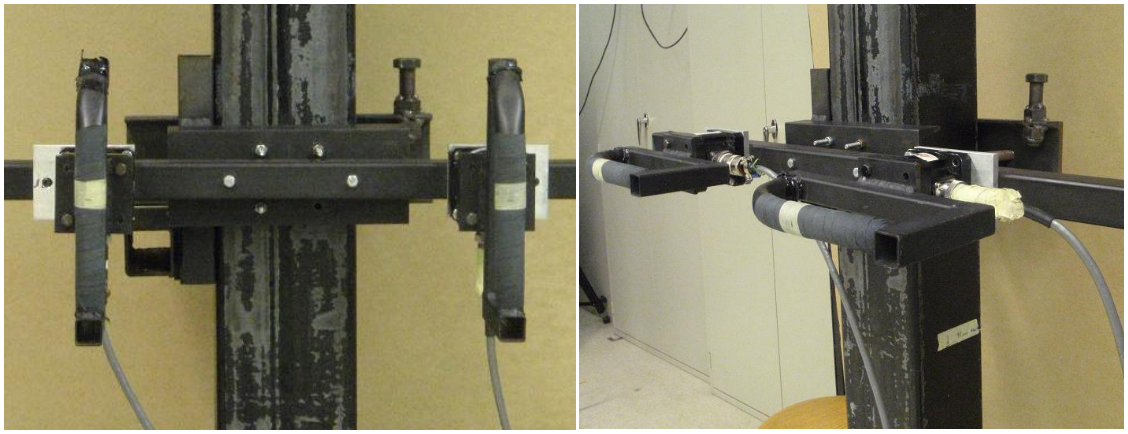

Figure 1: Handle setup shown with handles oriented vertically (left) and horizontally (right).

3.3.3 Motion Tracking

During the MVIC trials, participant kinematics were captured and recorded with the VICON Nexus 1.4 software (Oxford Metrics Group Plc, Oxford, UK). Eighty-nine reflective markers were placed on each participant over bony landmarks and in clusters on specific segments of the body (Table 6, Figure 2). Eight VICON MX20+ (2.0 MP) cameras encircled the collection space and tracked movements of the reflective markers. This data was used to identify full body positions and orientations adopted during the MVIC trials to estimate shoulder and L5/S1 joint moments and foot stance. Motion analysis was also done to detect any movements made during the trials indicating that trials needed to be repeated. Motion tracking was collected at 50 Hz.

Table 6: VICON reflective marker locations

Marker(s)	**Location**
1, 2	Left and right ear
3	C7/T1
4	Xiphoid process
5	Suprasternal notch
6	L5/S1
7, 8	Left and right anterior superior iliac spine
9, 10	Left and right acromion
11, 12	Left and right medial epicondyle
13, 14	Left and right lateral epicondyle
15, 16	Left and right ulnar styloid
17, 18	Left and right radial styloid
19, 20	Left and right 2nd metacarpal head
21, 22	Left and right 5th metacarpal head
23, 24	Left and right greater trochanter
25, 26	Left and right medial condyle
27, 28	Left and right lateral condyle
29, 30	Left and right lateral malleolus
31, 32	Left and right medial malleolus
33, 34	Left and right 5th metatarsal head
35, 36	Left and right tip of the big toe
37, 38	Left and right distal bisection of the calcaneous
39-43	Trunk (chest)
44-47, 48-51	Left and right upper arm
52-55, 56-59	Left and right forearm
60-64, 65-69	Left and right thigh
70-74, 75-79	Left and right shank
80-84, 85-89	Left and right foot

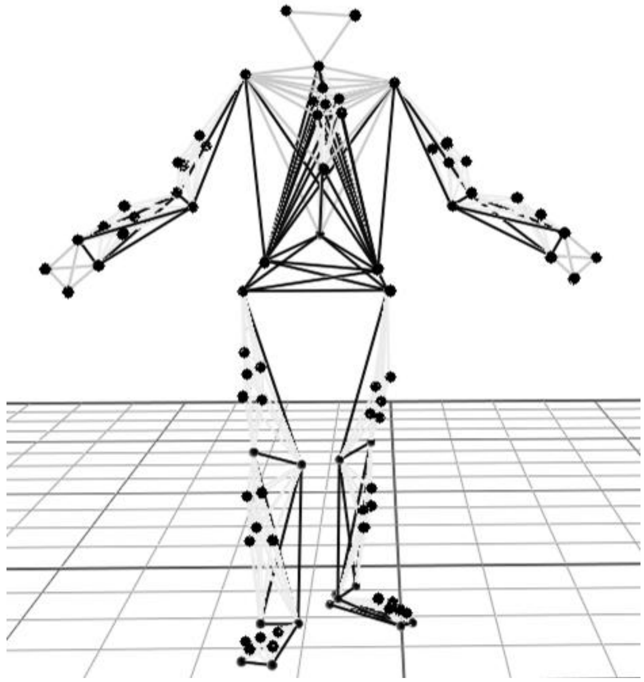

Figure 2: Reflective marker placements indicated in VICON software.

3.4 Experimental Protocol

The experimental protocol consisted of participant preparation and experimental data collection phases (Table 7), which were identical for all participants. The collection period took between 2.5 and 3 hours. Each participant completed all experimental trials in one session on a single day.

Table 7: Experimental protocol overview.

Subject Preparation
1. Shaved hair/dead skin at electrode site and cleansed with isopropyl alcohol 2. Placed 14 bipolar surface electrodes over mid-belly of selected muscles 3. Confirmed electrode placements and secured electrodes to skin with adhesive tape
Collection Procedures
1. Collected six-second quiet and shunt trials 2. Participant performed three sets of eight six-second maximal voluntary electrical trials (two-minute rest period between trials) 3. Participant was given a minimum of 15 minutes of rest 4. Participant performed 20 randomized six-second maximal voluntary isometric contraction trials (two-minute rest period between trials)

3.4.1 Participant Preparation

Following electromyographic setup, the placement of each electrode was confirmed by monitoring the EMG signal elicited when the participant performed each muscle's primary action under moderate resistance. Electrodes were then secured to the participant's skin using adhesive tape. No further electrode manipulation occurred for the remainder of the data collection. With the participant lying supine on the test bench, relaxed and with nothing touching the handles attached to the transducers, six-second quiet and shunt trials (a form of calibration in which a known resistance proportional to a specific force value is applied to determine the relationship between force and the resulting output voltage) were collected. This data was subsequently used as the baseline (direct current (DC) bias) for raw EMG and force, as well as to calibrate voltage into Newtons.

Maximal voluntary electrical activation or MVE is used to refer to the bioelectrical domain in terms of the normalizing to the maximal EMG amplitude obtained during standardized contractions (Mathiassen et al., 1995). The use of MVEs facilitates comparisons of muscle activity levels between muscles and participants. The EMG signal obtained from each muscle during the MVIC trials were normalized in terms of the corresponding peak EMG amplitude of each muscle obtained during standardized MVE trials. MVE trials were chosen based on recommendations by Delagi and Perotto (1980), McGill (1992), and Boettcher et al. (2008) for optimal normalization tests to elicit maximum muscle activity for the selected muscles. Descriptions of the MVE trials are found in Table 8. MVE trials were demonstrated to the participants and practice time was provided using negligible force until they felt comfortable with the exertion. Each participant performed three rounds of eight MVEs, for a total of 24 MVEs for the seven bilateral muscles being examined (Mathiassen et al., 1995). The participants gradually built up to their MVE over 2 s and maintained it for the remainder of a six-second collection period, which should have provided adequate time to reach and maintain a steady-state or constant level of muscle activation (Chaffin, 1975). A minimum rest period of 2 minutes was given between trials to avoid muscular fatigue (Chaffin, 1975; Mathiassen et al., 1995). During all MVE trials, the participants were sitting or lying on a clinical test bench. For all MVIC trials, participants stood upright. The participant's head was held neutral (untwisted) in all trials unless otherwise specified (Mathiassen et al., 1995).

Table 8: Description of the MVE trials.

Muscle	**MVE Condition**
Biceps Brachii	Participant is seated, flexing the elbows against resistance with 0^{o} of shoulder flexion and abduction, forearms supinated, and elbows flexed to 90^{o}
Long head of Triceps	Participant is lying supine, extending the elbows against resistance with shoulders and elbows flexed to 90^{o}, and forearms pronated
Pectoralis Major (clavicular insertion)	Participant is seated, horizontally adducting arms while shoulders are flexed 90^{o} bilaterally with the heels of the hands together and elbows flexed 20^{o} from full elbow extension
Middle Deltoid (Performed on each side of body)	Participant is seated, abducting the shoulders against resistance with shoulders abducted 90^{o}
Middle Trapezius	Participant is lying prone with shoulders abducted in line with the middle trapezius muscle fibres, retracting the scapulae against resistance applied above the elbows
Erector Spinae	Participant is lying prone leaning over the edge of the test bench at the hips with the legs restrained, hands placed on opposite shoulders, extending the L5/S1 joint against resistance
Rectus Abdominis	Participant is sitting on the test bench in a bent-knee sit-up posture, feet restrained by a strap, hands placed on opposite shoulders and the trunk forming an angle of approximately 30^{o} with the horizontal, flexing the hips against resistance applied to the shoulders

3.4.2 Experimental Data Collection Procedures

Fifteen minutes of rest was provided between the MVE and MVIC trials to minimize the effects of any localized muscle fatigue. The order of MVICs was randomized to minimize order effects. Participants were told the direction of force (push or pull) prior to each MVIC. The participant was asked to place their feet side by side and shoulder width apart, standing centred in front of the test handles, on a rubber exercise mat. It was assumed that a surface with a high coefficient of friction would reduce slip potential. The reach distance corresponded to the participant's functional arm reach for the fully extended elbow conditions, while the reach distance was adjusted such that the participant could maintain an elbow angle of $\leq 90^{o}$ for the flexed elbow conditions. Two sample MVICs are shown in Figure 3. The right foot positions were marked on the mat with tape to ensure consistent placement. They were free to position their left foot in any position they felt would enable them to exert their greatest force. Participants were instructed to attempt to maintain their body postures for the duration of each trial. Initially, the participants stood with arms at their sides. Then, using a power grip on each of the handles, participants gradually built up their maximum force over 2 seconds and maintained it for the remainder of a six-second collection period. Again, a minimum rest period of 2 minutes was provided between trials to avoid muscular fatigue (Chaffin, 1975; Mathiassen et al., 1995).

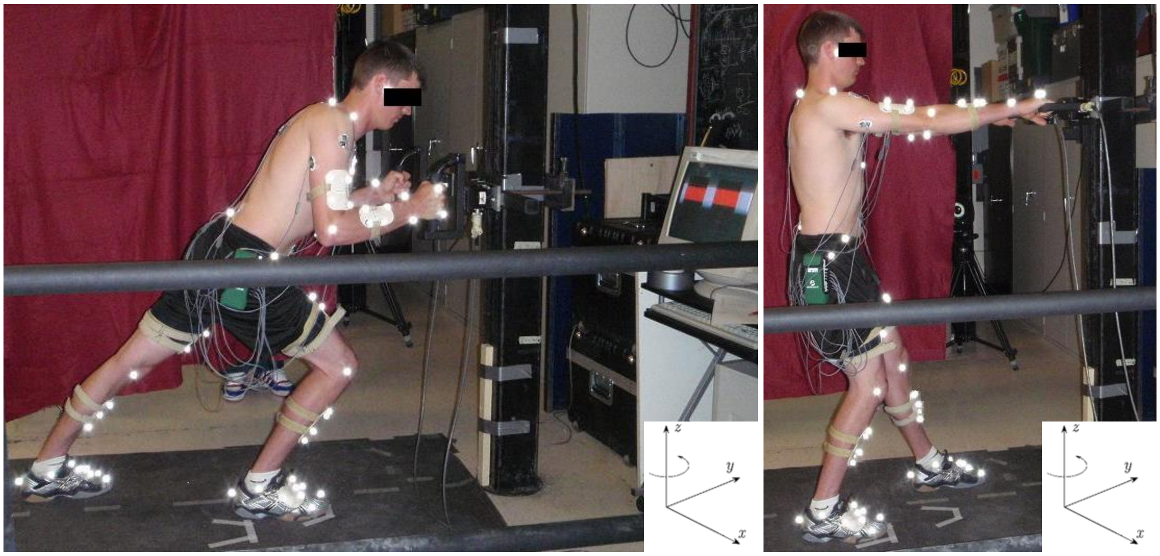

Figure 3: MVICs performed by a North American 95th percentile male participant. The posture on the left demonstrates a push exertion at 100 cm with vertical handles and elbows flexed while the posture on the right shows a pull exertion at 150 cm with horizontal handles and elbows extended.

As force feedback was shown to assist participants in increasing hand force during upper extremity exertions, as well as improve within-participant reproducibility of subsequent maximal exertions, visual feedback was provided during all MVE and MVIC trials (Fischer, Belbeck, & Dickerson, 2010). Muscle activation feedback was provided during the MVEs through VICON Nexus 1.4 software (Oxford Metrics Group Plc, Oxford, UK) while force feedback was provided during the MVICs via a custom computer program (Labview 6.1, National Instruments, Texas, USA) showing the instantaneous force the participant was exerting in real-time by means of a vertical bar. A horizontal reference line across the bar indicated the highest force they achieved for that MVIC during the previous trial, where applicable. Visual feedback was unitless and of undefined magnitude. Participants were not told the numerical results. In addition, verbal encouragement was given to the participants as this has been shown to increase participant effort, and thus muscle activation and force (Meskers, de Groot, Arwert, Rozendaal, & Rozing, 2004).

3.5 Data Analysis

The primary data analyzed were hand forces and muscle activity. These data were processed to yield derivative variables to develop specific answers to the hypotheses. A total of 50 dependent variables were examined. All data processing was done with MATLAB 7.0.1 (Mathworks Inc., MA, USA) software.

3.5.1 Force Data

Raw force recorded in volts was processed and calibrated into force in Newtons. Force data was only considered for the MVIC trials. In order to remove any DC bias, the average of the quiet trial was subtracted from all trials. The shunt trial was then used to convert the raw force trials recorded in volts into Newtons. Force data was then dual pass filtered using a 2nd order low pass Butterworth filter set at a cutoff frequency of 3.5 Hz based on residual analysis. Residual analysis was performed on 6 participants (3 males and 3 females) at random channels for 5 random MVICs including 0.1 Hz and from 0.5-8 Hz at 0.5 Hz intervals, using Equation 2.

$$\text{Residual } (f_c) = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (X_i - \hat{X}_i)^2} \qquad (2)$$

where X_i = raw data at the *i*th sample

$\hat{X}_i$ = filtered data at the *i*th sample

Each trial was visually inspected for data artifacts. If artifacts existed, those frames containing artifact were not included in the analysis. Only the steady-state part of all trials was analyzed (constant level of force production), expected to occur between the 2nd and 5th second, to minimize tremor, motion dynamics or inertial effects (Chaffin, 1975). A three-second time-averaged mean of the steady state part of all trials was used, corresponding in time to the three-second average of normalized EMG, described below. Only forces from the primary push-pull axis were analyzed, although all forces were measured and recorded. A second average was calculated across the two repetitions for each participant for each MVIC, resulting in a left hand force (LHF) and a right hand force (RHF). A total force value was calculated by summing together the forces from both hands in each MVIC (THF).

3.5.2 EMG Data

To remove the DC bias and bring the baseline of raw EMG about zero, the average of the raw EMG obtained from the quiet trial was subtracted from all MVE and MVIC trials. The EMG was then high pass filtered using a 2nd order dual pass Butterworth filter with a cutoff frequency of 30 Hz to remove heart muscle electrical activity (Drake & Callaghan, 2006). The raw surface EMG was then linear enveloped by full wave rectifying and digitally low pass filtering it at 3 Hz (based on residual analysis) using a 2nd order single pass Butterworth filter (Winter, 2005). The single pass of the filter created a phase lag to account for the electromechanical delay of muscle (De Luca, 1997). Each trial was visually inspected for data artifacts. If artifacts existed, the frames containing artifact were not considered in further analyses. Only the steady-state part of all trials was analyzed (constant level of muscle activation), expected to occur between the 2nd and 5th second, to minimize tremor, motion dynamics or inertial effects (Chaffin, 1975).

For the MVE trials, the peak surface EMG value was taken from the linear enveloped data using a 500 ms averaging moving window obtained during a single three-second time window between the 2nd and 5th second and used as the maximum muscle activity for the normalization reference. This method was intended to maximize within-participant reproducibility (Fischer et al., 2010). For each participant, the highest value among the three MVE trials for a given muscle is often chosen as the normalization reference (Mathiassen et al., 1995), however, the highest maximal muscle activation obtained for each muscle during any of the MVE trials was used as the normalization reference in this study. This decision was made based on the finding that many MVE trials may yield maximal muscle activations of multiple muscles (Boettcher et al., 2008; Chopp, Fischer, & Dickerson, 2010).

For the MVIC trials, a three-second time-averaged mean of the steady state part of the trials was used. For each participant, a final average was calculated across the two repetitions for each muscle for each MVIC. EMG signals obtained during the MVIC trials were then scaled to the maximum EMG signals obtained during the MVE trials for each muscle during each MVIC. The normalized values were calculated as in Equation 3. Measures of total muscle activation were computed based on the subset of muscles examined as in Equations 4 to 6 for each side of the body, as well as a total of all muscles, for each MVIC. Separate weighted averages of muscle activation were calculated based on muscle physiological cross-sectional areas (PCSAs) taken from McGill, Patt and Norman (1988) and Makhsous, Högfors, Siemien'ski and Peterson (1997) (Table 9), representative of muscle capacity. These were calculated for each side of the body, along with a total of all muscles, for each MVIC, as in Equations 7 to 9.

$$\text{Normalized Individual EMG (nEMG)} = \frac{\text{average EMG output in MVIC trial (mV)}}{\text{maximum EMG output in MVE trial (mV)}} \quad (3)$$

$$\text{Total Left EMG (nLEMG)} = \frac{\sum \text{nEMG of muscles on left side of body}}{7} \quad (4)$$

$$\text{Total Right EMG (nREMG)} = \frac{\sum \text{nEMG of muscles on right side of body}}{7} \quad (5)$$

$$\text{Total EMG (nTEMG)} = \frac{\sum \text{nEMG of all muscles}}{14} \quad (6)$$

$$\text{Weighted Average Left EMG (wLEMG)} = \frac{\sum_{i=1}^{7} \text{nLEMG}_i \ \text{PCSA}_i}{\sum_{i=1}^{7} \text{PCSA}_i} \quad (7)$$

$$\text{Weighted Average Right EMG (wREMG)} = \frac{\sum_{i=1}^{7} \text{nREMG}_i \ \text{PCSA}_i}{\sum_{i=1}^{7} \text{PCSA}_i} \quad (8)$$

$$\text{Weighted Average Total EMG (wTEMG)} = \frac{\sum_{i=1}^{14} \text{nEMG}_i \ \text{PCSA}_i}{\sum_{i=1}^{14} \text{PCSA}_i} \quad (9)$$

Table 9: PCSAs of muscles examined.

Muscle (unilateral)	**Acronym**	**PCSA (cm^2)**
Biceps brachii (short head + long head)	BIC	6.65[2]
Long head of Triceps	TRIC	9.98[2]
Pectoralis Major (clavicular insertion)	PEC	4.52[2]
Middle Deltoid	DEL	9.27[2]
Middle Trapezius	TRAP	4.54[2]
Erector Spinae	ES	23.4[1] (left), 21.7[1] (right)
Rectus Abdominis	RA	7.5[1] (left), 8.3[1] (right)

Adapted from [1]McGill et al. (1988) and [2]Makhsous et al. (1997)

3.5.3 EMG/force ratios

EMG/force ratios were then calculated. The ratios of individual muscle activation (iEMG) to their respective left and right hand forces were calculated for each muscle. All EMG/force ratios for the muscles from the left and right hand sides of the body were calculated as in Equation 10 and Equation 11, respectively. Total muscle activation to total force and weighted average muscle activation to total force were also computed for each trial for the left, right and combined sides (Equations 12-17). Smaller ratios indicated greater hand force magnitudes for a given level of muscle activation.

$$\text{Left Individual Muscle/force Ratio} = \frac{\text{iEMG (\%)}}{\text{LHF (N)}} \tag{10}$$

$$\text{Right Individual Muscle/force Ratio} = \frac{\text{iEMG (\%)}}{\text{RHF (N)}} \tag{11}$$

$$\text{Total LEMG/force Ratio} = \frac{\text{nLEMG (\%)}}{\text{LHF (N)}} \tag{12}$$

$$\text{Total REMG/force Ratio} = \frac{\text{nREMG (\%)}}{\text{RHF (N)}} \tag{13}$$

$$\text{Total EMG/force Ratio} = \frac{\text{nTEMG (\%)}}{\text{THF (N)}} \tag{14}$$

$$\text{wLEMG/force Ratio} = \frac{\text{wLEMG (\%)}}{\text{LHF (N)}} \tag{15}$$

$$\text{wREMG/force Ratio} = \frac{\text{wREMG (\%)}}{\text{RHF (N)}} \tag{16}$$

$$\text{wTEMG/force Ratio} = \frac{\text{wTEMG (\%)}}{\text{THF (N)}} \tag{17}$$

3.5.4 Kinematic Data

Three-dimensional marker trajectories for all experimental trials were examined in VICON Nexus 1.4 software (Oxford Metrics Group Plc, Oxford, UK). Unlabelled or incorrectly labelled markers were identified and correctly relabelled according to the whole body marker template established within the software. Missing markers were recorded and gaps in marker trajectories were pattern filled. All marker coordinates were then exported in the form of comma separated values (CSV). Foot stance was calculated as the distance between right and left foot heel markers in the anterior-posterior plane.

3.5.5 Moments

For each participant, anthropometrics, postural data and 3D transducer forces recorded for all experimental conditions were used as the input for an upper body static 3D linked segment model created in MATLAB 7.0.1 (Mathworks Inc., MA, USA). Postural data was taken from a single frame representative of each condition. Due to data corruption, one participant was removed from moment analyses. An additional 10 trials were removed due to missing markers. The linked segment model consisted of the left and right hands, left and right forearms, left and right upper arms and the trunk-head segment adapted from Dickerson, Martin and Chaffin (2006). Sagittal plane moments represent the moments about the x-axis in the global coordinate system as calculated using standard linked segment mechanics. Resultant moments were calculated for the left and right shoulder joints as well as the L5/S1 joint as in Equation 18, where M represents the moment about each of the axes.

$$\text{Resultant moment} = \sqrt{M_x^2 + M_y^2 + M_z^2} \tag{18}$$

3.5.6 Statistical Analyses

All statistical analyses were performed in JMP® 8.0.2.2 (SAS Institute Inc., NC, USA). A total of 50 dependent variables were examined with repeated measures analysis of variance (ANOVA): left, right and total hand force capability (3), left and right shoulder and low back moments in the sagittal plane (3), resultant left and right shoulder and low back moments (3), normalized individual EMG (14), total and weighted average EMG (6), normalized individual EMG/force ratios (14), total and weighted average EMG/force ratios (6), and foot stance (1). In performing the analyses, three assumptions were made:

1. The samples were independent.
2. The samples were taken from a population in which the samples were normally distributed.
3. The variances of the samples in the population were equal.

Within a gender, mass and stature vary widely. Cross correlations were performed to determine the correlation between mass and stature. There was only a moderate association between the two variables (between 0.57 and 0.64). As a result, both variables were considered as additional independent variables and were included in further analyses.

Dependent variables were examined under two groups of independent variables due to the subset of test conditions examining elbow angle. For the first group, a set of six-way repeated measures ANOVA was used to identify statistically significant effects of gender, mass, stature, direction, handle height and handle orientation (GMSDHO) on all force-, moment- and EMG-based estimates and foot stance. In the second group, a set of five-way repeated measures ANOVA was used to determine statistically significant effects of gender, mass, stature, direction and elbow angle (GMSDE) on all force-, moment- and EMG-based estimates and foot stance. Factor effects were considered to be significant when $p < 0.05$. Post hoc analyses were then performed on any significant main effects using the Student's t test to compute individual pairwise comparisons ($\alpha < 0.05$, two-tailed). Significant interaction effects were further analyzed with the Tukey-Kramer honestly significant difference (Tukey HSD) test to compare all differences among the means ($\alpha < 0.05$, two-tailed). The ANOVA models helped to identify which of the investigated conditions had a significant effect on the measured outcomes while the post hoc comparisons indicated how each outcome changed in response to the different levels of each condition or interactions between conditions.

Two force transducers measured left and right hand forces simultaneously and independently. For each condition, left and right hand forces were compiled under a single hand force parameter. A similar process was carried out for left and right EMG estimates. A set of seven-way (gender, mass, stature, direction, handle height, handle orientation and hand (GMSDHOA)) and six-way (gender, mass, stature, direction, elbow angle and hand (GMSDEA)) repeated measures ANOVA models were then used to determine statistically significant differences in force- and EMG-based measures recorded for both sides of the body. This process helped to identify any left and right asymmetries.

4.0 Results

Results are presented for the two groups of independent variables, which include the GMSDHO model followed by the GMSDE model, in examining their effects on hand force magnitudes, shoulder and low back moments, mean %MVE, EMG/force ratios and foot stance. A final section examines left and right asymmetry based on the GMSDHOA and GMSDEA models. The overall significance for all models was $p < 0.0001$. In general, main effects of direction or interaction effects with direction had the greatest effect on most of the dependent variables examined.

4.1 Hand Force Magnitudes

Female total hand force capability (THF) for push trials ranged from 176.5 N to 514.9 N and THF for pull trials ranged from 85.0 N to 350.3 N. Male THF for push and pull trials ranged from 256.5 N to 741.5 N and from 105.8 N to 511.3 N, respectively. Mean hand force capabilities and their standard deviations (SD) are presented in Tables 10 and 11. On average, the smallest forces for both males and females occurred when pulling at 150 cm with the handles oriented horizontally and the elbows fully extended. On average, the greatest forces for both males and females resulted when pushing at 100 cm with the handles oriented vertically and the elbows flexed $\leq 90^{o}$. Hand force capability varied widely within the sample population as emphasized by the relatively large standard deviations. Direction accounted for the greatest proportion of the variance in hand force capability based on omega-square (ω^2) comparisons.

Table 10: Female hand force capabilities for all experimental conditions (n = 12).

		Push Forces (N)			Pull Forces (N)		
Condition		Left	Right	Total	Left	Right	Total
100-H-Ex	Mean	143.7	146.6	290.3	140.7	102.9	243.5
	SD	34.8	31.2	62.3	39.2	35.0	66.6
150-H-Ex	Mean	158.0	157.3	315.3	70.3	69.2	139.5
	SD	34.0	19.8	47.7	24.8	26.1	44.7
100-V-Ex	Mean	141.4	159.4	300.8	124.4	114.6	239.0
	SD	28.2	33.8	54.0	34.9	34.7	65.9
150-V-Ex	Mean	148.2	149.8	298.1	76.2	77.7	153.9
	SD	23.0	26.1	42.8	26.7	31.6	53.1
100-V-Fx	Mean	188.7	190.4	379.2	96.5	93.1	189.6
	SD	54.0	48.7	101.6	26.8	26.2	51.2

*100 = 100 cm, 150 = 150 cm, H = horizontal, V = vertical, Ex = fully extended elbows, Fx = elbow angles of $\leq 90^o$.

Table 11: Male hand force capabilities for all experimental conditions (n = 12).

		Push Forces (N)			Pull Forces (N)		
Condition		Left	Right	Total	Left	Right	Total
100-H-Ex	Mean	214.9	239.0	453.9	174.1	152.3	326.4
	SD	49.3	64.1	102.3	59.9	48.2	96.0
150-H-Ex	Mean	228.0	230.7	458.6	111.6	96.9	208.5
	SD	55.9	55.2	109.1	33.2	28.6	58.1
100-V-Ex	Mean	221.6	236.5	458.1	195.8	163.6	359.4
	SD	48.7	61.2	105.2	72.7	47.2	109.2
150-V-Ex	Mean	232.5	242.4	474.9	118.6	100.5	219.1
	SD	64.3	67.5	127.8	43.3	32.9	73.1
100-V-Fx	Mean	284.4	290.7	575.1	140.3	143.4	283.6
	SD	67.2	66.4	130.8	47.6	41.5	81.3

*100 = 100 cm, 150 = 150 cm, H = horizontal, V = vertical, Ex = fully extended elbows, Fx = elbow angles of $\leq 90^o$.

4.1.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Hand Force Magnitudes

There was a main effect direction, gender, mass and handle height on left, right and total hand force capability. The push direction resulted in significantly greater left, right and total hand force capability values than the pull direction ($p < 0.0001$). On average, the push direction generated 47%, 78%, and 61% more force than the pull direction in terms of left, right and total hand forces, respectively (Figure 4). Males had significantly greater left (p = 0.0474), right (p = 0.0304) and total (p = 0.0333) hand force capability values than females (Figure 5). Females, on average, had 67% of the maximal hand force of males. A larger body mass was also associated with significantly greater left (p = 0.0144), right (p = 0.0132) and total (p = 0.0115) hand force capability values. In regards to handle height, hand force capability was greater for the 100 cm height than the 150 cm height ($p < 0.0001$, Figure 6). There was no effect of handle orientation on hand force capability.

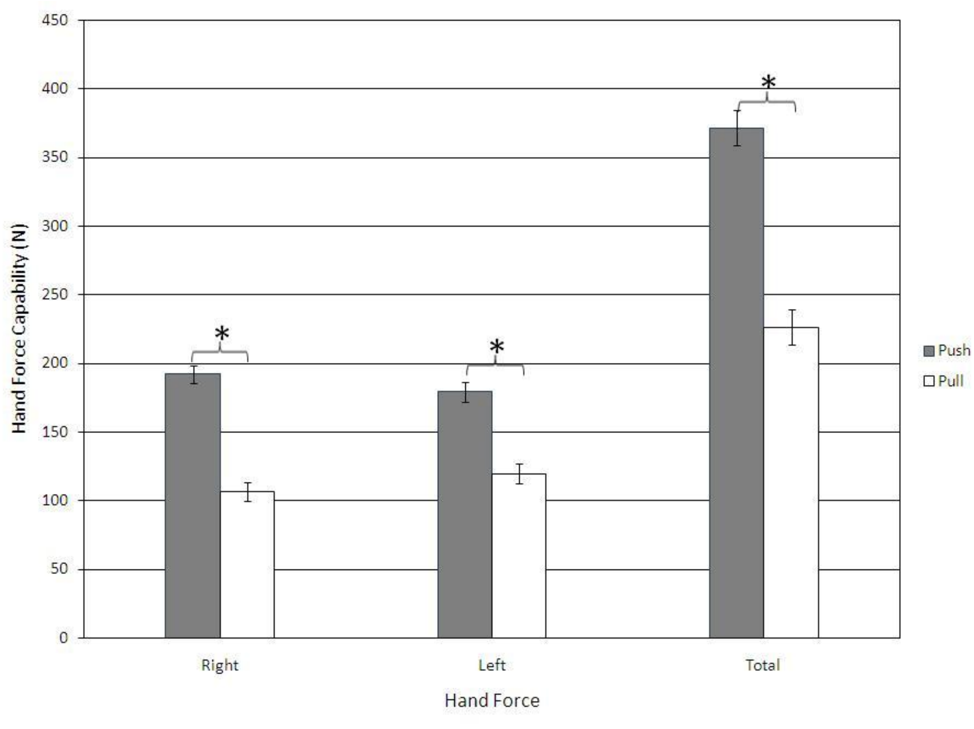

Figure 4: Effects of direction on least square mean (LSM) hand force capability collapsed across all conditions. * indicates significant differences.

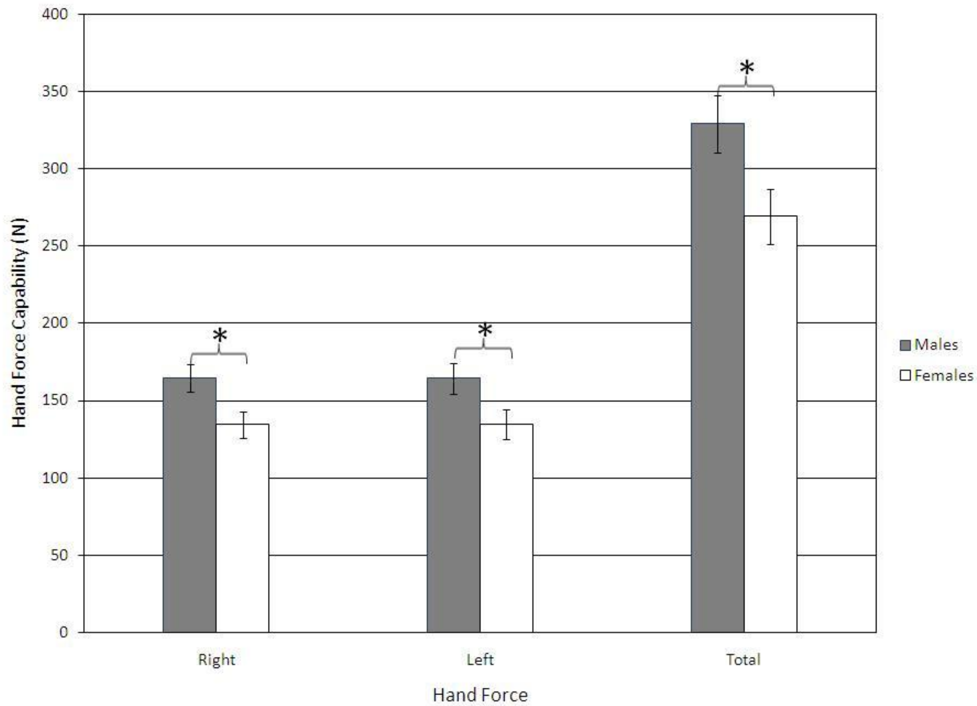

Figure 5: Effects of gender on LSM hand force capability collapsed across all conditions. * indicates significant differences.

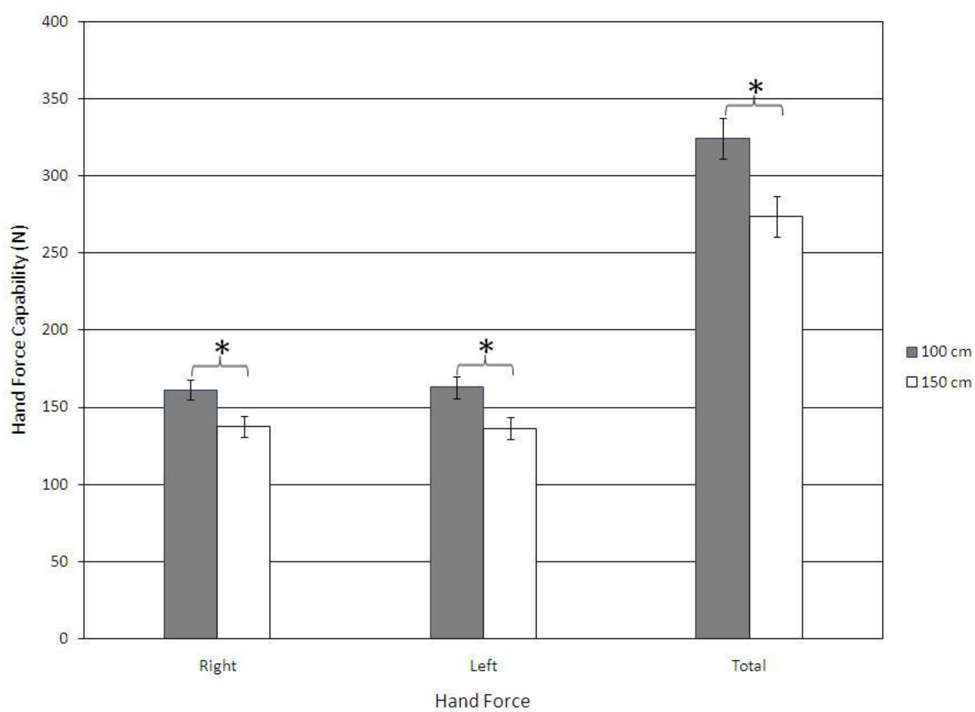

Figure 6: Effects of handle height on LSM hand force capability collapsed across all conditions. * indicates significant differences.

4.1.2 Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Hand Force Magnitudes

Interaction effects existed between gender and direction and between direction and height on left, right and total hand force capability. An ordinal interaction occurred between gender and direction, in which gender differences with respect to forces were larger for the push direction than the pull direction (Figures 7-9). This trend was the same and statistically significant for left, right and total hand force capability values ($p < 0.0001$). An ordinal interaction was also present between direction and height where differences in force due to direction were larger at the 150 cm than at the 100 cm height ($p < 0.0001$, Figures 10-12). Within each direction, the 100 cm height generated greater right hand forces than the 150 cm height, but for left and total hand force capability values, the greatest and least forces both occurred at the 150 cm height.

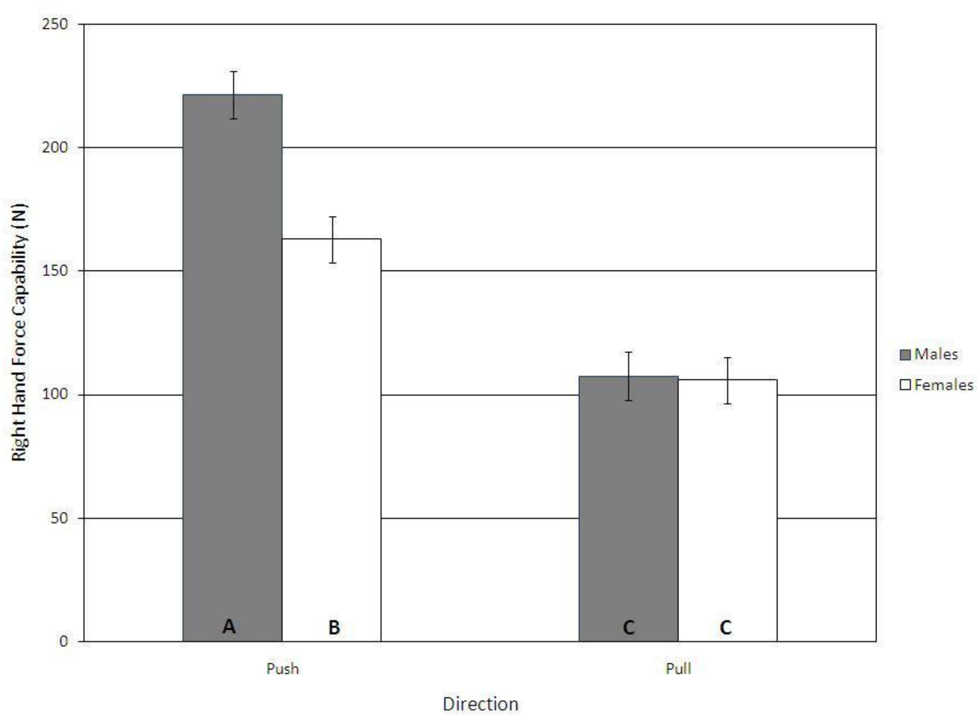

Figure 7: Effects of gender and direction on LSM right hand force capability. Letters indicate significantly different gender by direction interactions.

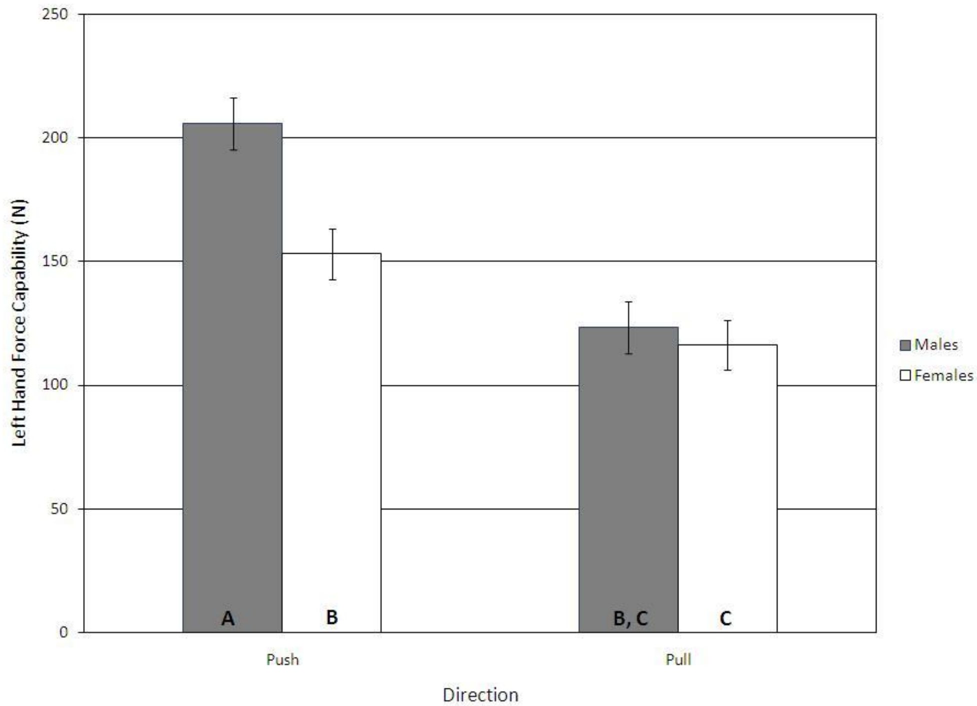

Figure 8: Effects of gender and direction on LSM left hand force capability. Letters indicate significantly different gender by direction interactions.

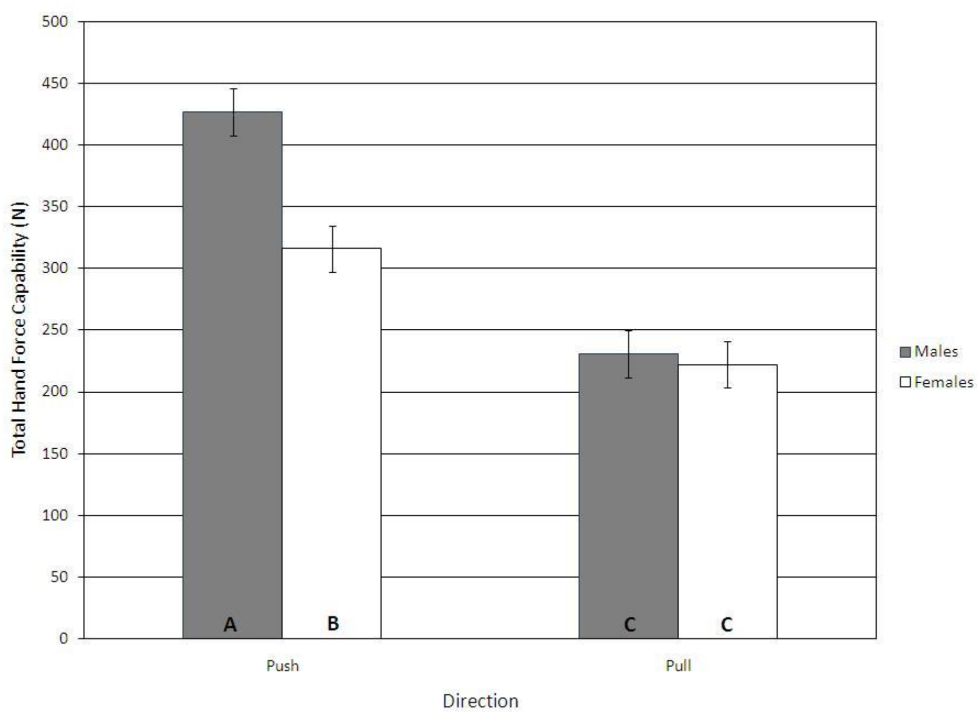

Figure 9: Effects of gender and direction on LSM total hand force capability. Letters indicate significantly different gender by direction interactions.

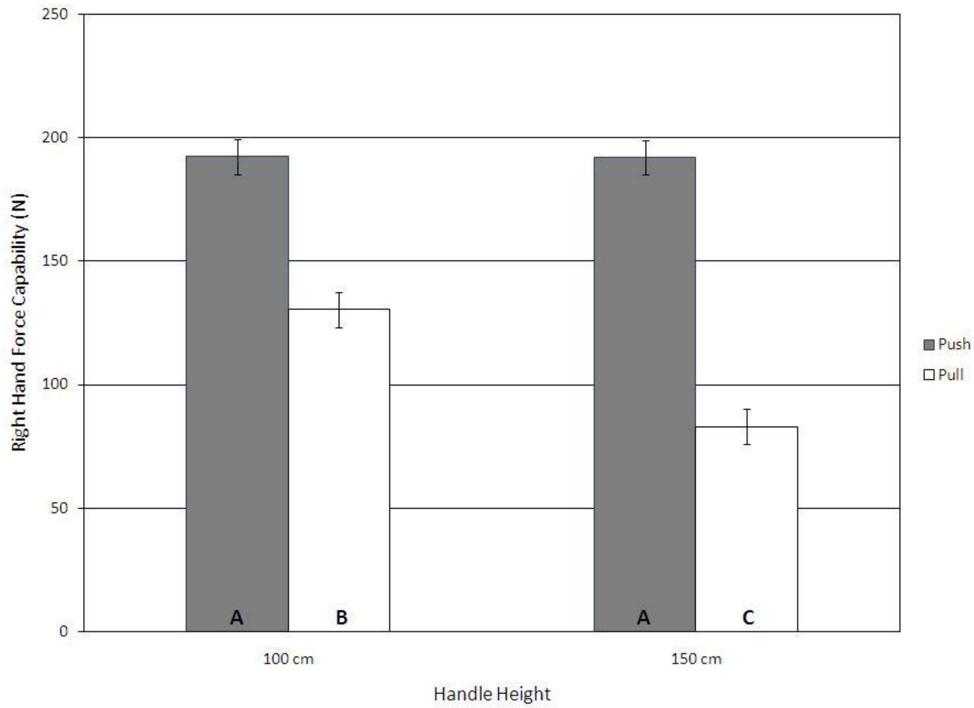

Figure 10: Effects of direction and handle height on LSM right hand force capability. Letters indicate significantly different direction by handle height interactions.

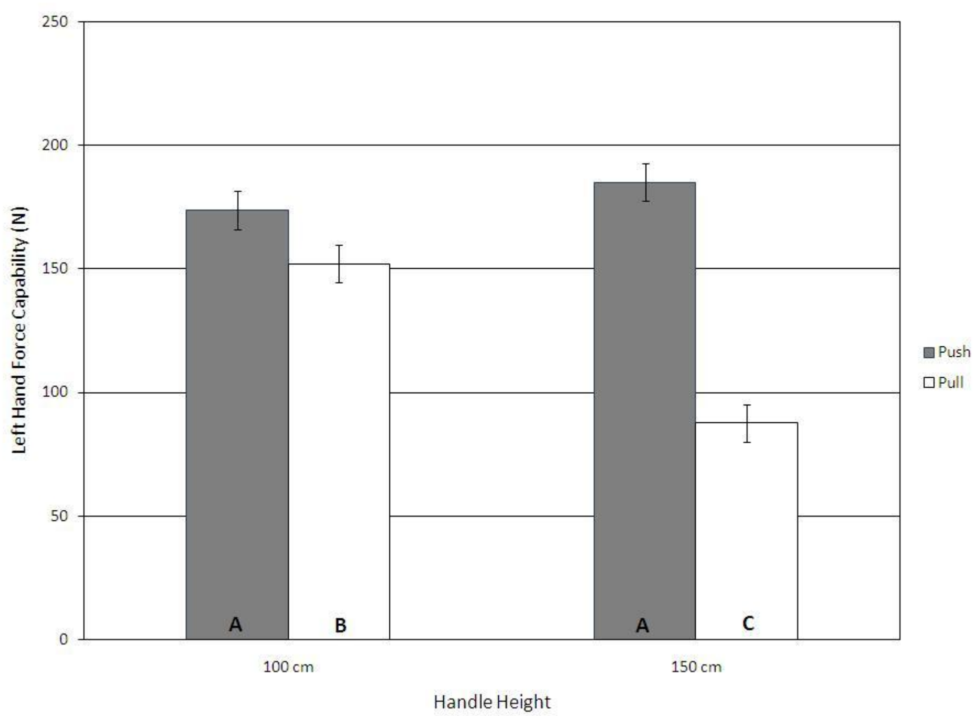

Figure 11: Effects of direction and handle height on LSM left hand force capability. Letters indicate significantly different direction by handle height interactions.

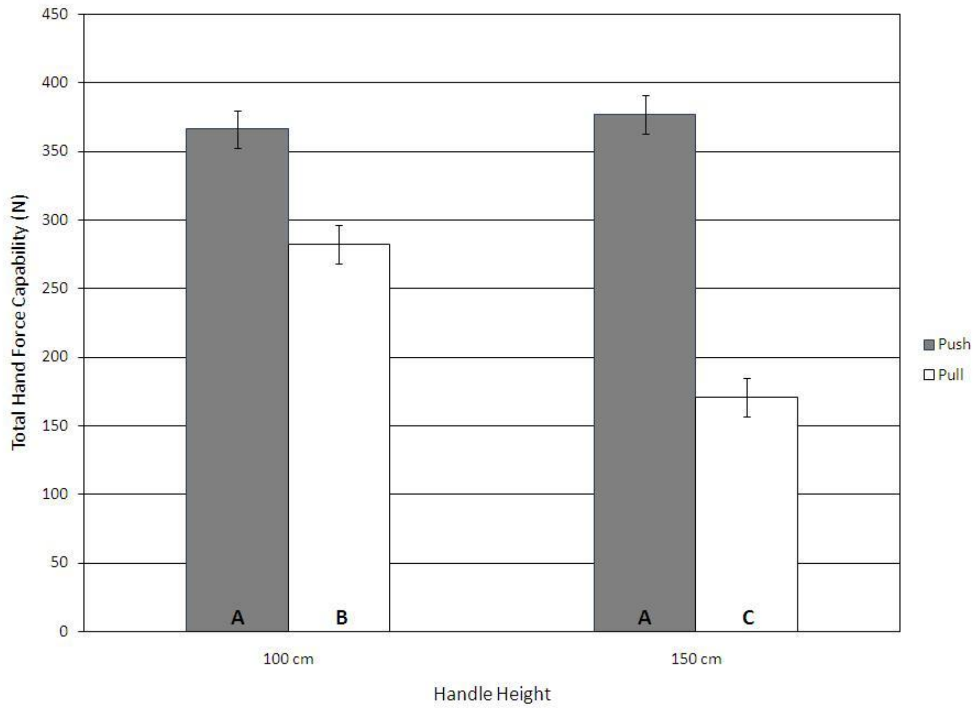

Figure 12: Effects of direction and handle height on LSM total hand force capability. Letters indicate significantly different direction by handle height interactions.

4.1.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Hand Force Magnitudes

There was a main effect of gender, mass and direction on left, right and total hand force capability, with direction having the greatest effect. Elbow angle did not have any significant effects on hand force magnitudes. Males had significantly greater left ($p = 0.0484$) and total ($p = 0.0500$) hand force capability values than females (Figure 13). Although not statistically significant, male right hand force capability values were also greater than those of females ($p = 0.0769$). A larger body mass and the push over the pull direction resulted in significantly greater left, right and total hand force capability values ($p < 0.0001$). On average, the push direction resulted in 70%, 50% and 60% greater force than the pull direction for the right, left and total forces, respectively (Figure 14).

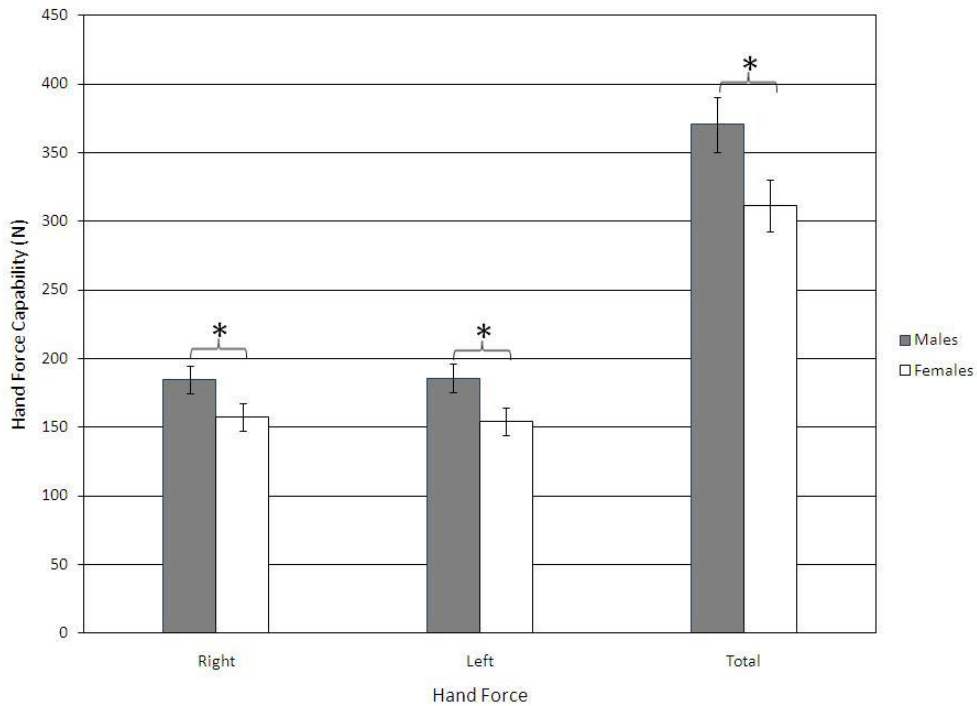

Figure 13: Effects of gender on LSM hand force capability collapsed across all conditions. * indicates significant differences.

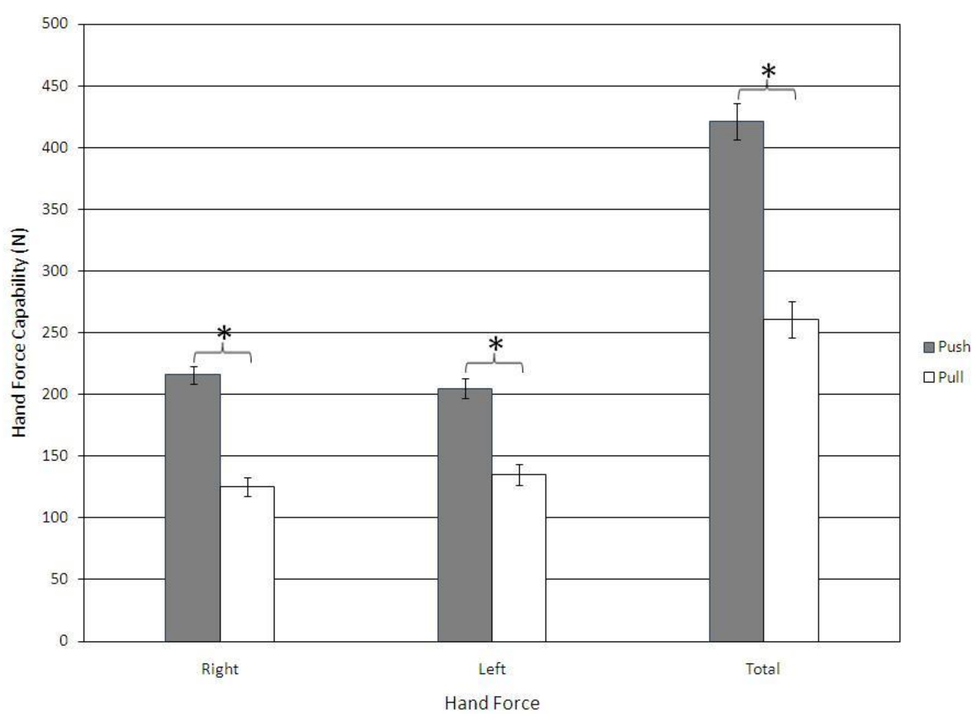

Figure 14: Effects of direction on LSM hand force capability collapsed across all conditions. * indicates significant differences.

4.1.4 Interaction Effects of Gender, Mass, Stature, Direction and Elbow Angle on Hand Force Magnitudes

An interaction effect between gender and direction on total hand force capability was evident (p = 0.0238). There was also a marked interaction effect between direction and elbow angle on left, right, and total hand force capability (p < 0.0001). An ordinal interaction occurred between direction and elbow angle, in which differences between directions with respect to force capability were larger for the flexed elbows condition than the extended elbows condition. Within each direction, greater forces were measured for males than females. Both the greatest and least hand forces resulted when the elbows were flexed as demonstrated in Figures 15-17.

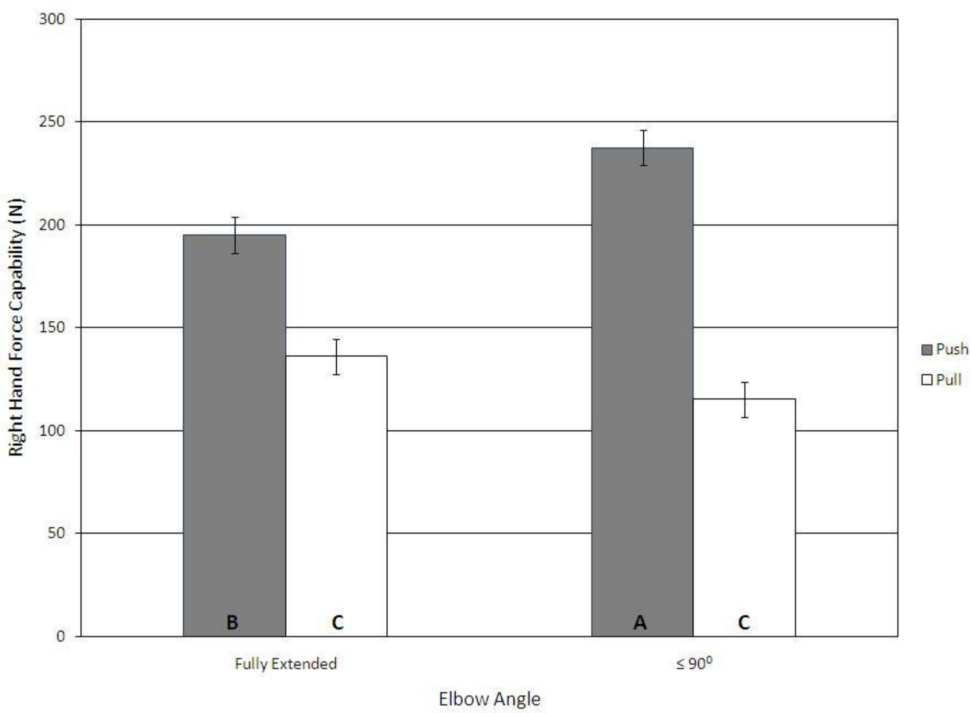

Figure 15: Effect of direction and elbow angle on LSM right hand force capability. Letters indicate significantly different direction by elbow angle interactions.

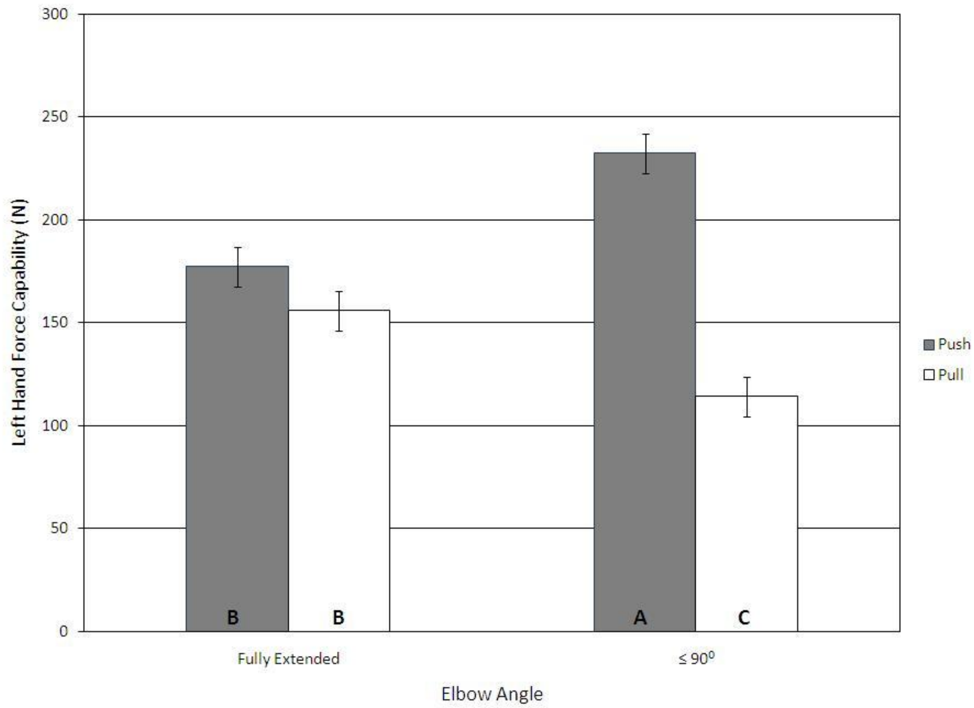

Figure 16: Effect of direction and elbow angle on LSM left hand force capability. Letters indicate significantly different direction by elbow angle interactions.

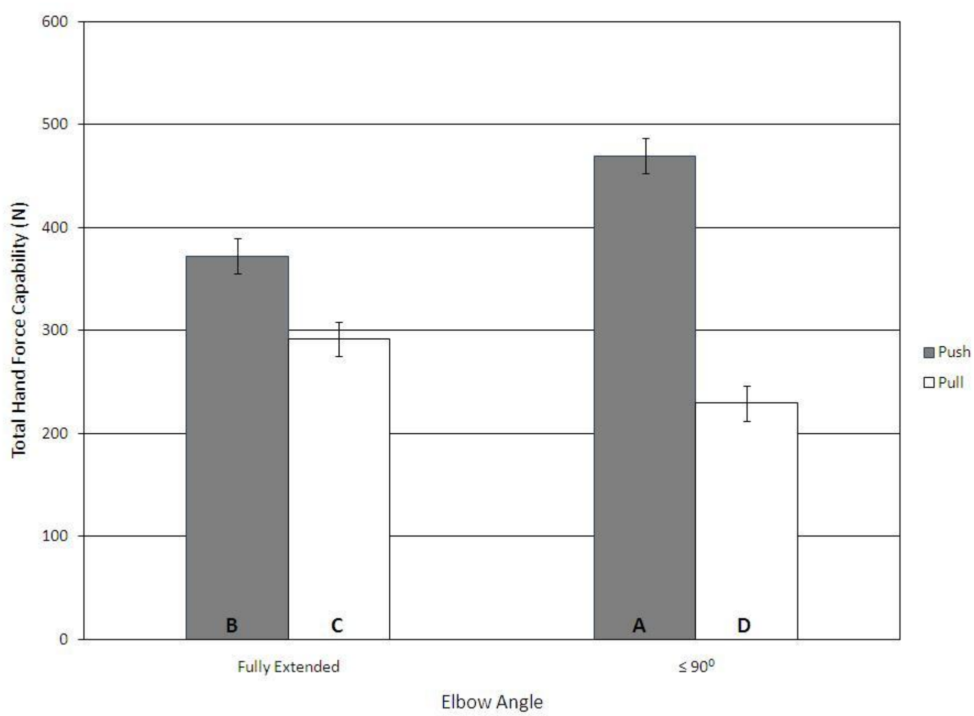

Figure 17: Effect of direction and elbow angle on LSM total hand force capability. Letters indicate significantly different direction by elbow angle interactions.

4.2 Shoulder and Low Back Moments

As external forces were predominantly generated in a plane approximating the sagittal plane, for the anterior-posterior push and pull tasks examined in the current study, the analysis focused on the flexion and extension moments at the right and left shoulder and L5/S1 joints. This was under the assumption that participants stood square to the handles for all trials. Sagittal plane moments are presented as reactive moments about the left or right shoulder or L5/S1 joint. A positive moment indicates a flexor moment at the shoulder joint or an extensor moment at the L5/S1 joint. A negative moment represents an extensor moment at the shoulder joint or a flexor moment at the L5/S1 joint. Average flexor and extensor moments at the right shoulder joint were 40.0 N.m and 28.0 N.m, respectively, across all conditions. The left shoulder joint had an average flexor moment of 37.2 N.m and an average extensor moment of 28.8 N.m, across all conditions. The L5/S1 joint had an average flexor 89.6 N.m and an average extensor moment of 143.9 N.m, across all conditions. Direction or direction by height interactions had the greatest effect on sagittal plane moments based on ω^2.

Resultant moments were calculated for the left and right shoulders and the low back at the L5/S1 joint. These helped to account for hand forces in the medial-lateral plane. Resultant right shoulder moments ranged from 5.0 N.m to 140.1 N.m with an average moment of 50.4 N.m while resultant left shoulder moments ranged from 2.2 N.m to 132.5 N.m with an average moment of 43.3 N.m, across all conditions. Resultant L5/S1 moments ranged from 12.03 N.m to 444.3 N.m with an average moment of 155.9 N.m, across all conditions. Comparisons of ω^2 showed that direction had the greatest effect on resultant moments based on the GMSDE model while the GMSDHO model was more affected by orientation effects.

4.2.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Shoulder and Low Back Moments

A main effect of direction was evident for the right shoulder joint whereby the push direction resulted in a flexor moment and the pull direction resulted in an extensor moment ($p < 0.0001$). The pull direction led to significantly greater extensor moments at the L5/S1 joint compared to the push direction ($p < 0.0001$, Figure 18). In addition, the 100 cm handle height resulted in significantly greater flexor moments at the right and left shoulder joints ($p = 0.0030$, $p = 0.0003$, Figure 19) and significantly greater extensor moments at the L5/S1 joint ($p < 0.0001$). The flexor moments at the right and left shoulder for the 100 cm handle height were, on average, 2.4 and 3.4 times greater than the flexor moments at the 150 cm handle height, respectively. The extensor moments at the L5/S1 joint for the 100 cm handle height were, on average, 2.8 times greater than the extensor moments produced at the 100 cm handle height.

A main effect of mass existed for resultant right shoulder ($p = 0.0006$), left shoulder ($p = 0.0152$) and low back ($p = 0.0117$) moments. Increased masses resulted in significantly greater moments. While the push direction led to significantly greater resultant right shoulder moments, the pull direction led to significantly greater resultant low back moments ($p < 0.0001$, Figure 20). Although not statistically significant, the push direction also led to greater resultant left shoulder moments. In addition, the 150 cm handle height ($p = 0.0107$) resulted in greater resultant low back moments (Figure 21). Finally, the horizontal handle orientation also led to greater resultant right shoulder ($p = 0.0005$), left shoulder ($p = 0.0020$), and low back ($p < 0.0001$) moments (Figure 22).

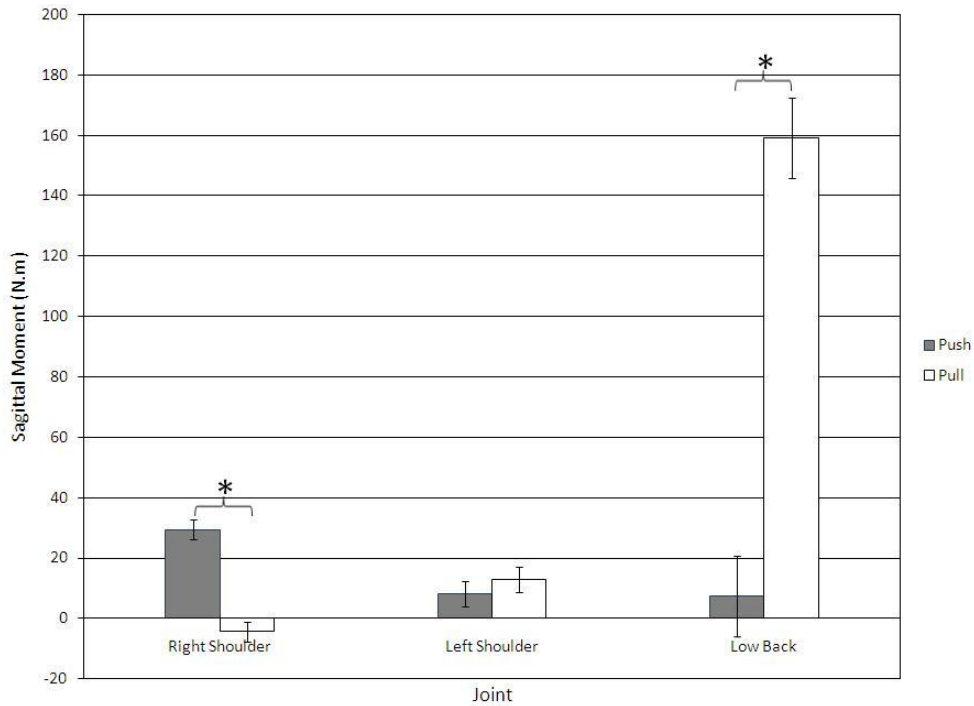

Figure 18: Effects of direction on LSM sagittal plane moments at the right and left shoulder joints and L5/S1 joint collapsed across all conditions. * indicates significant differences.

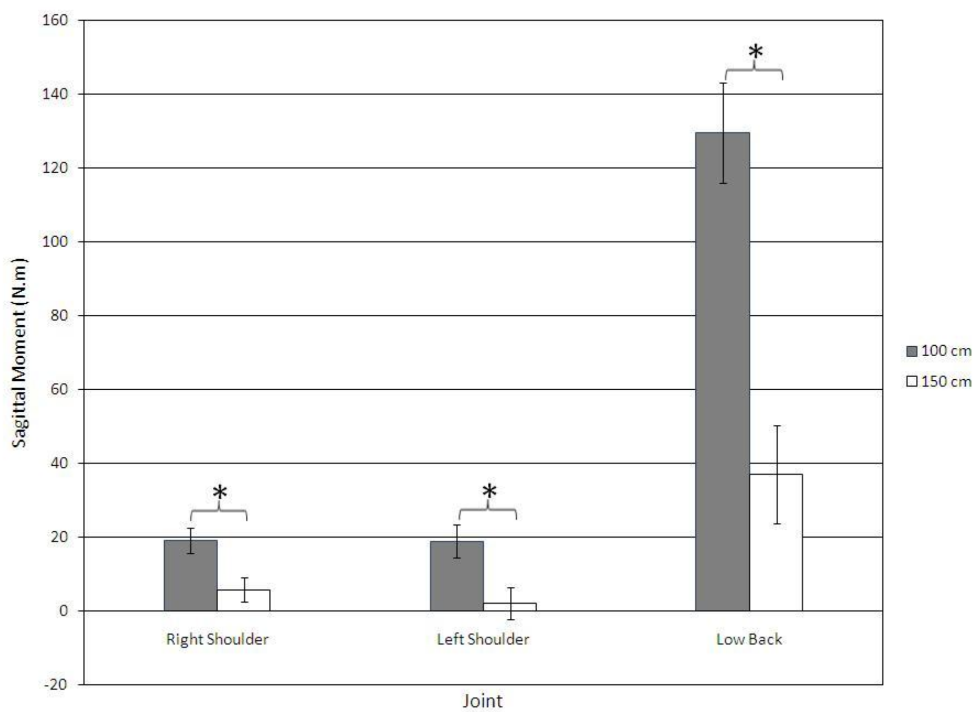

Figure 19: Effects of handle height on LSM sagittal plane moments at the right and left shoulder joints and L5/S1 joint collapsed across all conditions. * indicates significant differences.

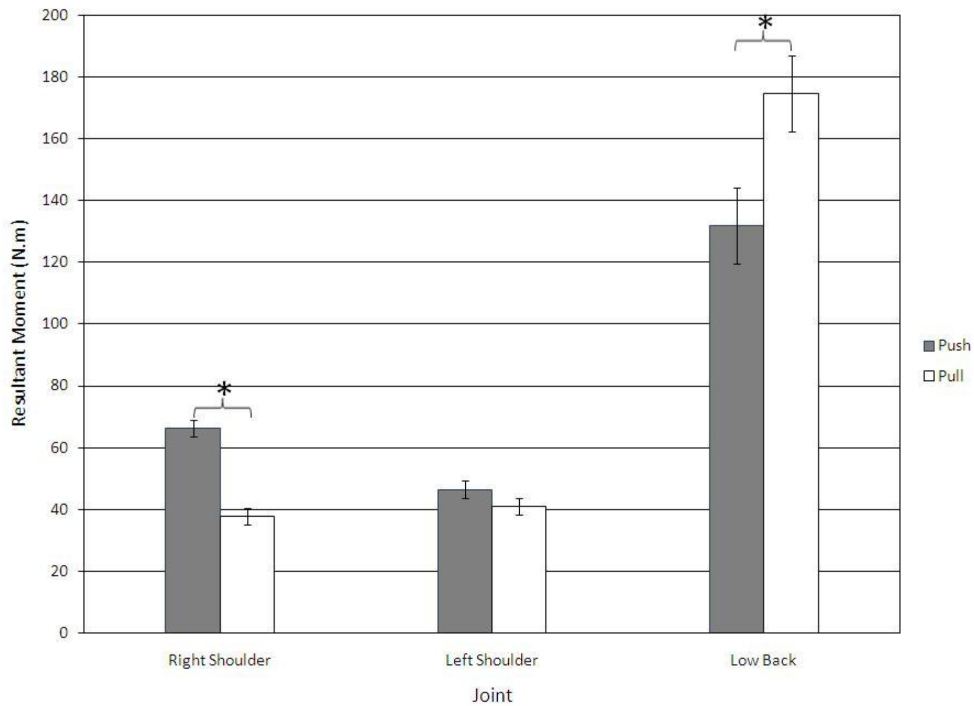

Figure 20: Effects of direction on LSM resultant moments at the right and left shoulder joints and L5/S1 joint collapsed across all conditions. * indicates significant differences.

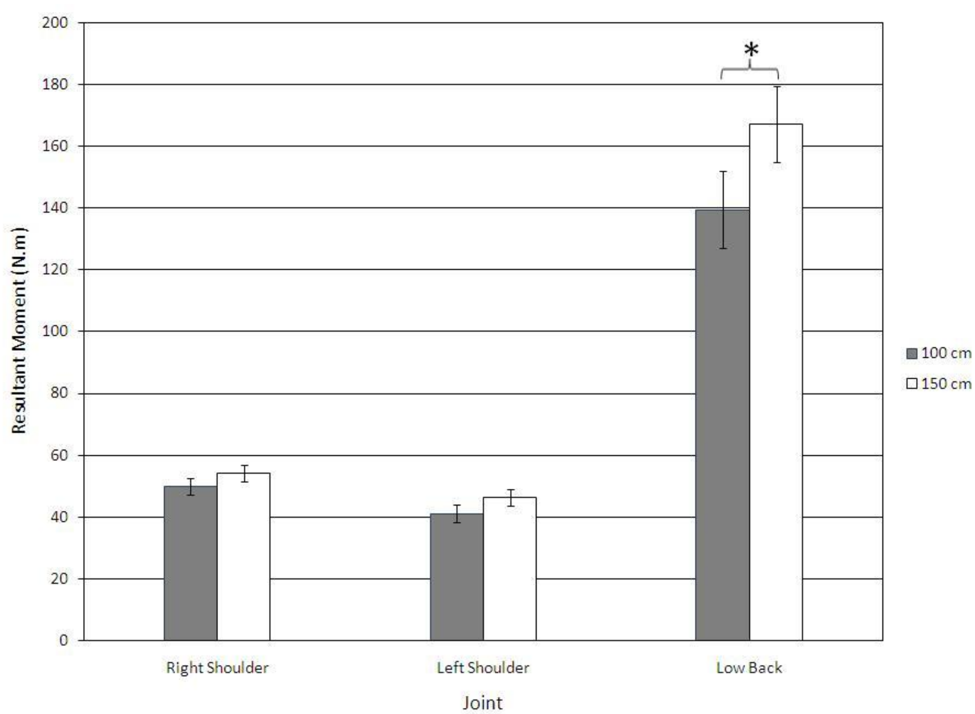

Figure 21: Effects of handle height on LSM resultant moments at the right and left shoulder joints and L5/S1 joint collapsed across all conditions. * indicates significant differences.

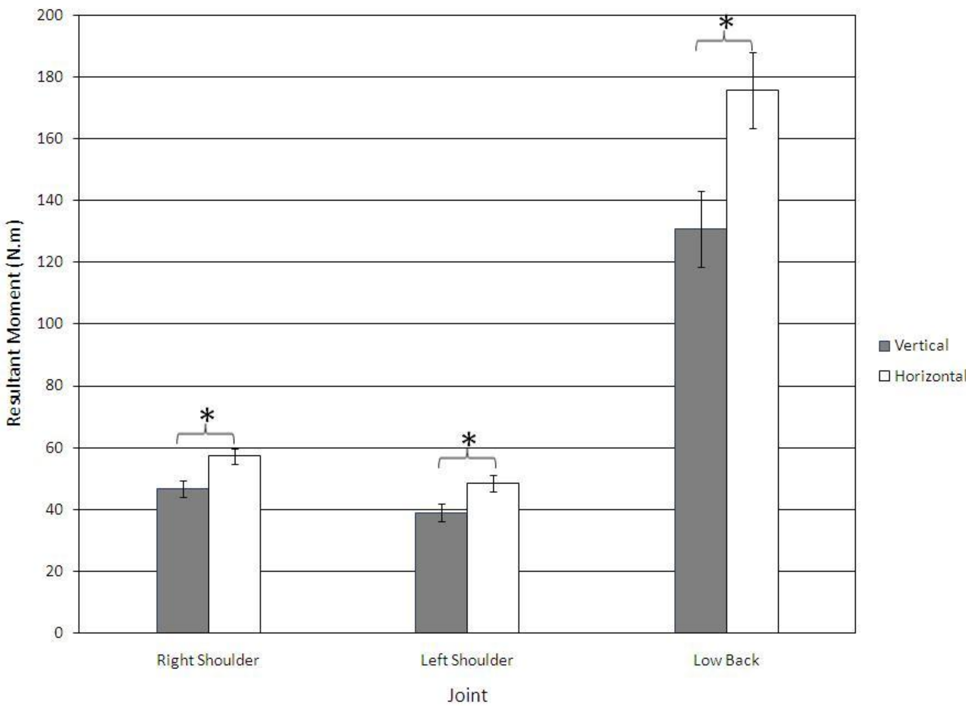

Figure 22: Effects of handle orientation on LSM resultant moments at the right and left shoulder joints and L5/S1 joint collapsed across all conditions. * indicates significant differences.

4.2.2. Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Shoulder and Low Back Moments

A significant disordinal direction by height effect was present on sagittal plane shoulder and L5/S1 joint moments such that the effects of direction are reversed as handle height changes ($p < 0.0001$). In the case of the shoulder joints, the moments reverse polarity. For the push direction, the flexor moments at the 100 cm height become extensor moments at the 150 cm height. For the pull direction, the extensor moments at the 100 cm height become flexor moments at the 150 cm height. The interaction effects of direction and height are illustrated in Figures 23-25. The interaction between direction and orientation was also significant for all sagittal plane shoulder and L5/S1 joint moments and are shown in Figures 26-28 (p <

0.0001). It appears that the push direction with the horizontal handle orientation is the combination that reduces the moments at the left shoulder and L5/S1 joints, while the pull direction with the horizontal orientation led to the greatest moments. These trends are notable as the horizontal handle orientation was found to cause the greatest resultant moments.

A significant height by orientation effect was evident on resultant right shoulder (p = 0.0059, Figure 29), left shoulder (p = 0.0004, Figure 30) and low back (p = 0.0017, Figure 31) moments. The 150 cm handle height combined with the horizontal handle orientation consistently resulted in significantly greater resultant moments compared to all other height by orientation interactions. The interaction between gender and height and was significant on resultant left shoulder moments where males and the 150 cm height generated the greatest moments while males and the 100 cm height generated the lowest moments (p = 0.0082). A significant direction by height effect was also present on resultant left shoulder moments such that the greatest moments occurred for the push direction and the 100 cm height while the lowest moments resulted for the pull direction and the 100 cm height (p = 0.0030). The interaction between gender and direction was also significant for resultant low back moments wherein females and the pull direction led to the greatest moments while females and the push direction led to the lowest moments (p = 0.0270).

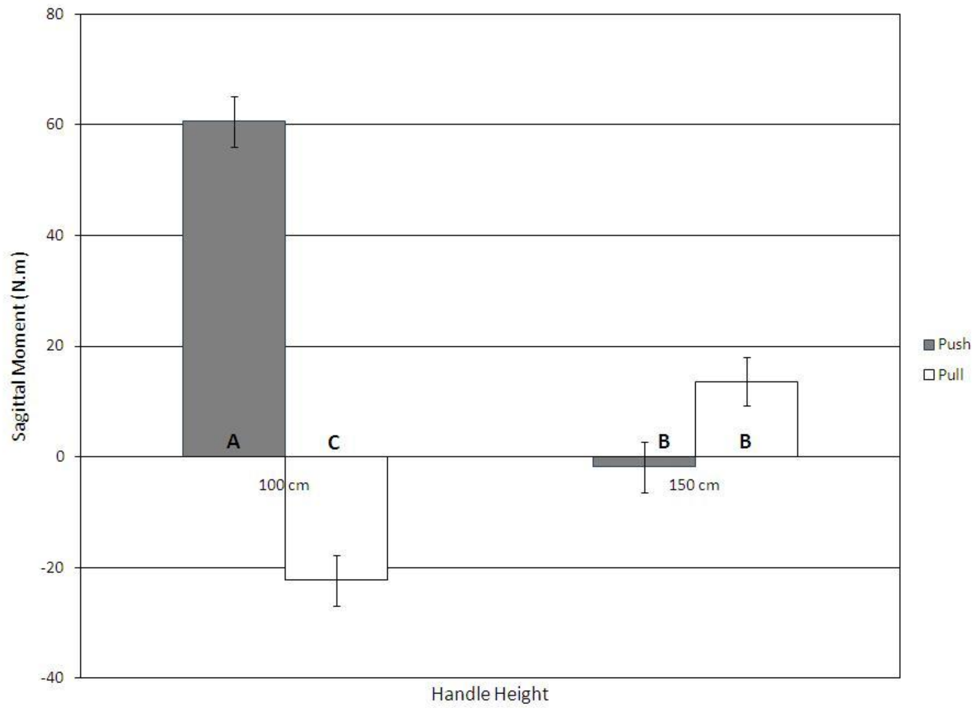

Figure 23: Effects of direction and handle height on LSM sagittal plane moments at the right shoulder joint. Letters indicate significantly different direction by handle height interactions.

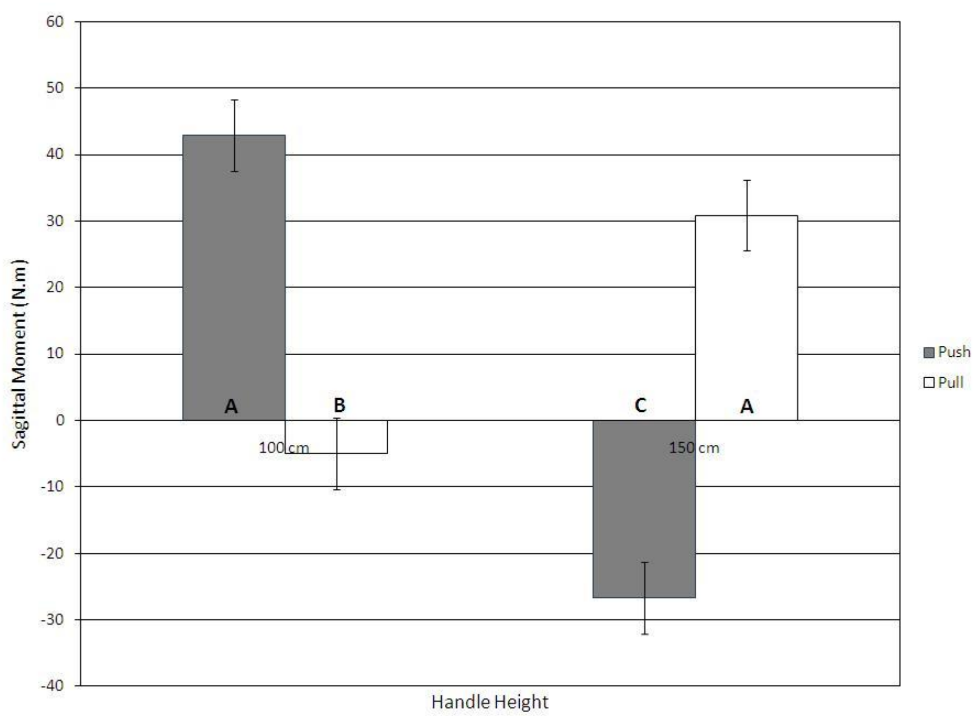

Figure 24: Effects of direction and handle height on LSM sagittal plane moments at the left shoulder joint. Letters indicate significantly different direction by handle height interactions.

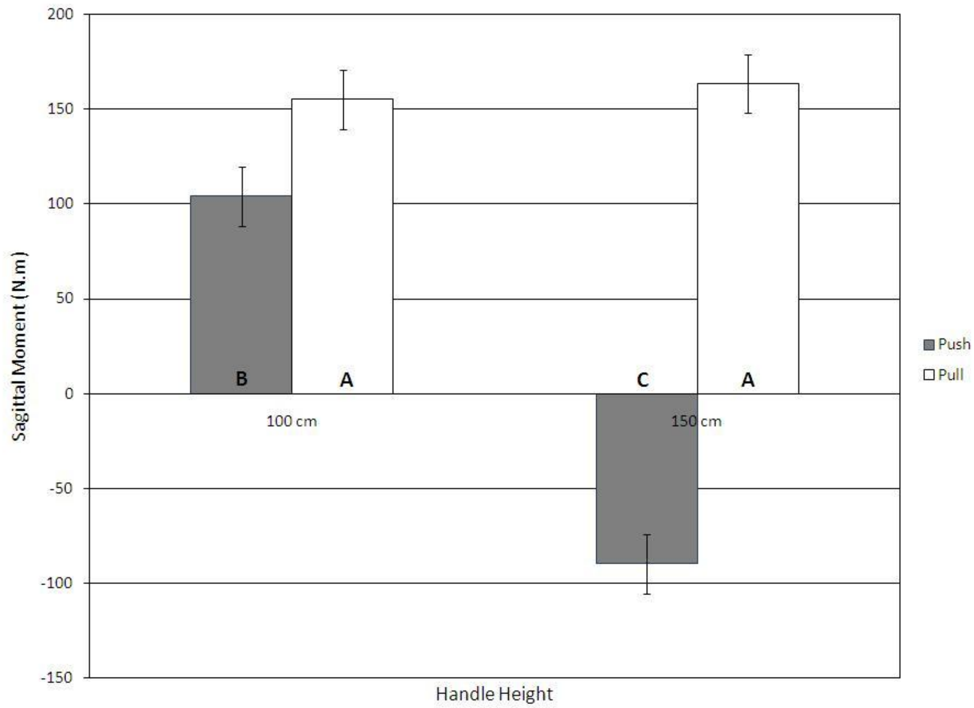

Figure 25: Effects of direction and handle height on LSM sagittal plane moments at the L5/S1 joint. Letters indicate significantly different direction by handle height interactions.

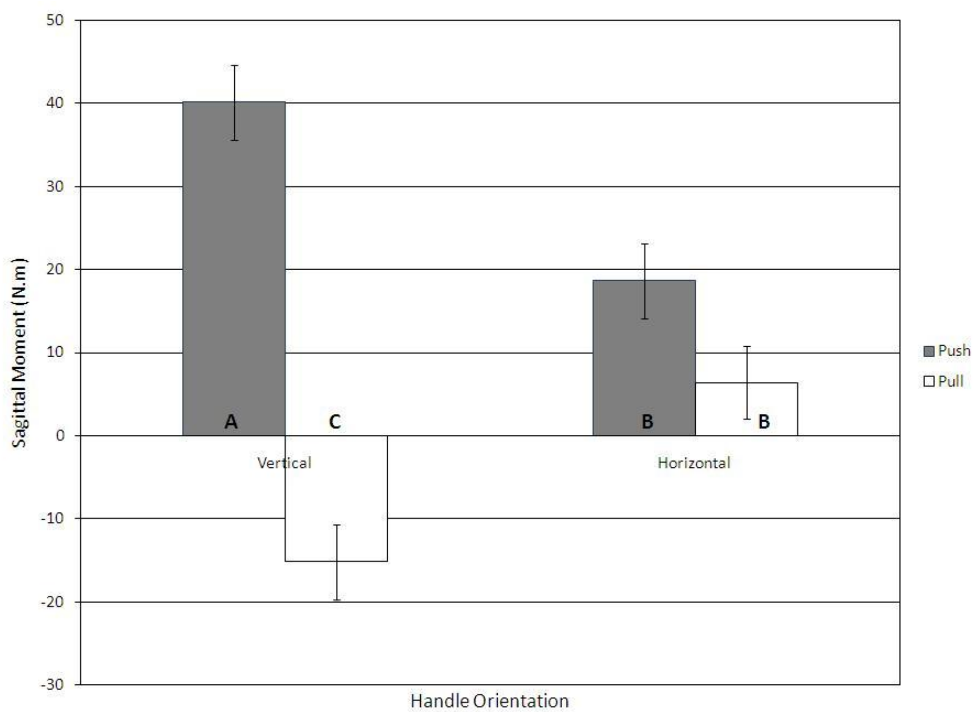

Figure 26: Effects of direction and handle orientation on LSM sagittal plane moments at the right shoulder joint. Letters indicate significantly different direction by handle orientation interactions.

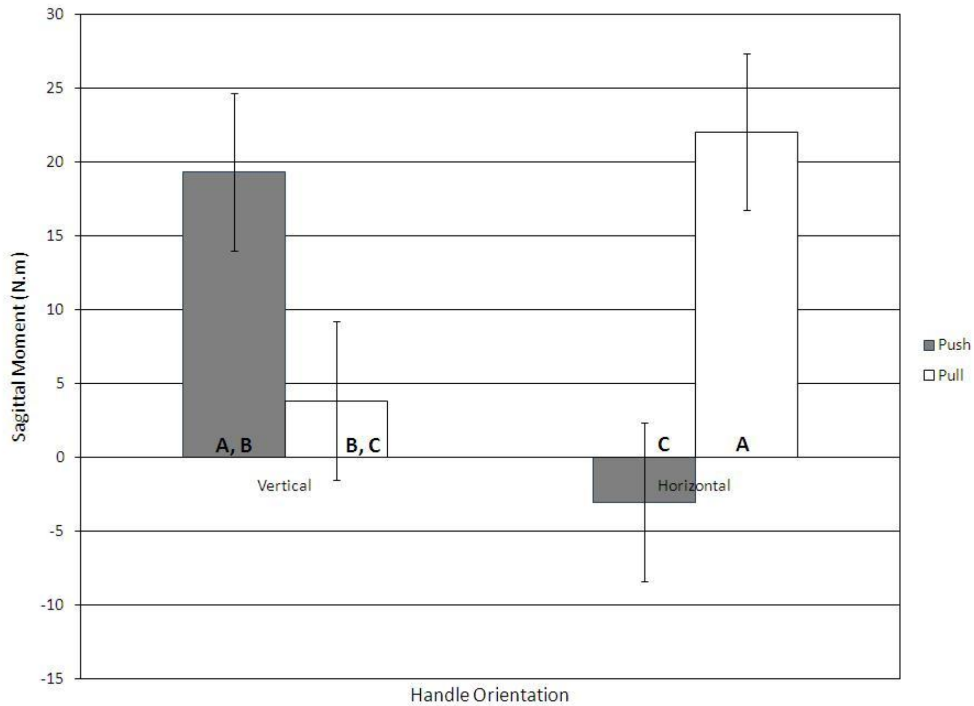

Figure 27: Effects of direction and handle orientation on LSM sagittal plane moments at the left shoulder joint. Letters indicate significantly different direction by handle orientation interactions.

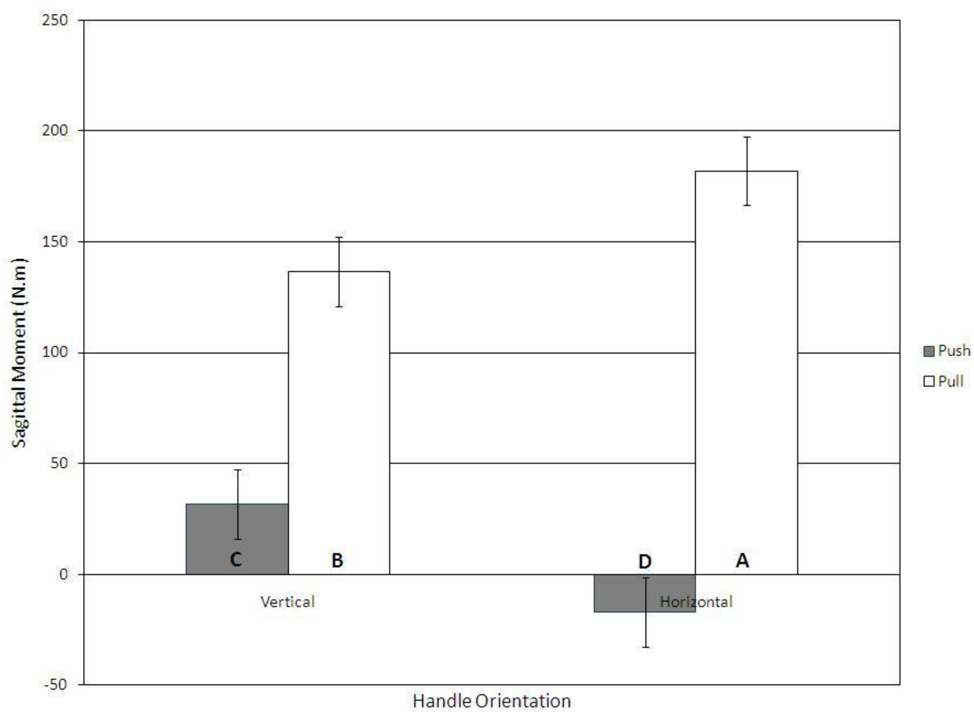

Figure 28: Effects of direction and handle orientation on LSM sagittal plane moments at the L5/S1 joint. Letters indicate significantly different direction by handle orientation interactions.

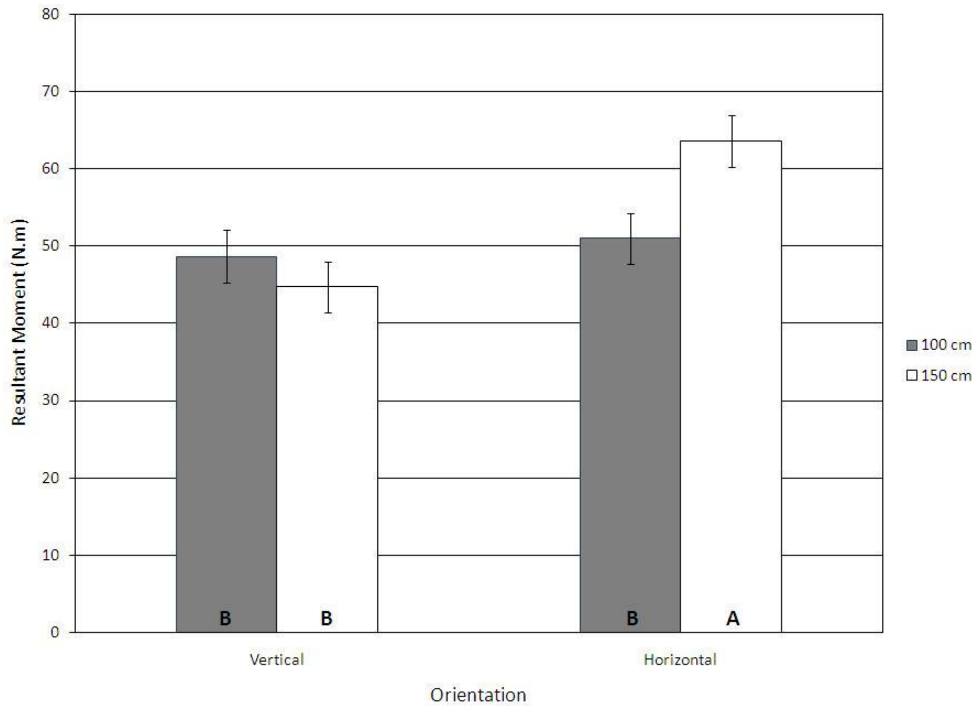

Figure 29: Effects of handle height and handle orientation on LSM resultant moments at the right shoulder joint. Letters indicate significantly different height by orientation interactions.

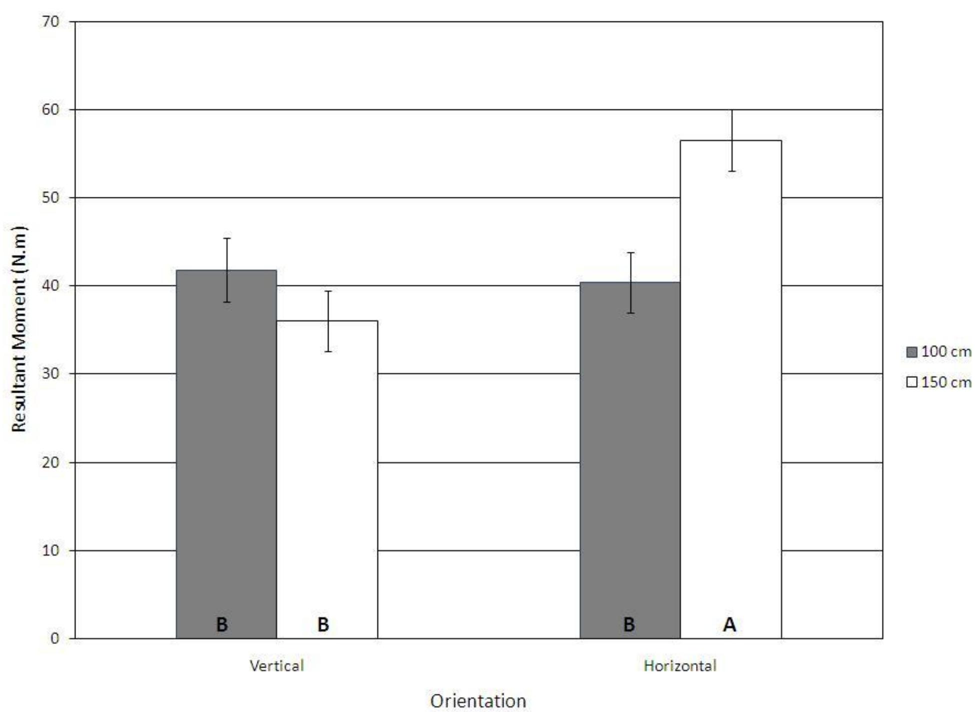

Figure 30: Effects of handle height and handle orientation on LSM resultant moments at the left shoulder joint. Letters indicate significantly different height by orientation interactions.

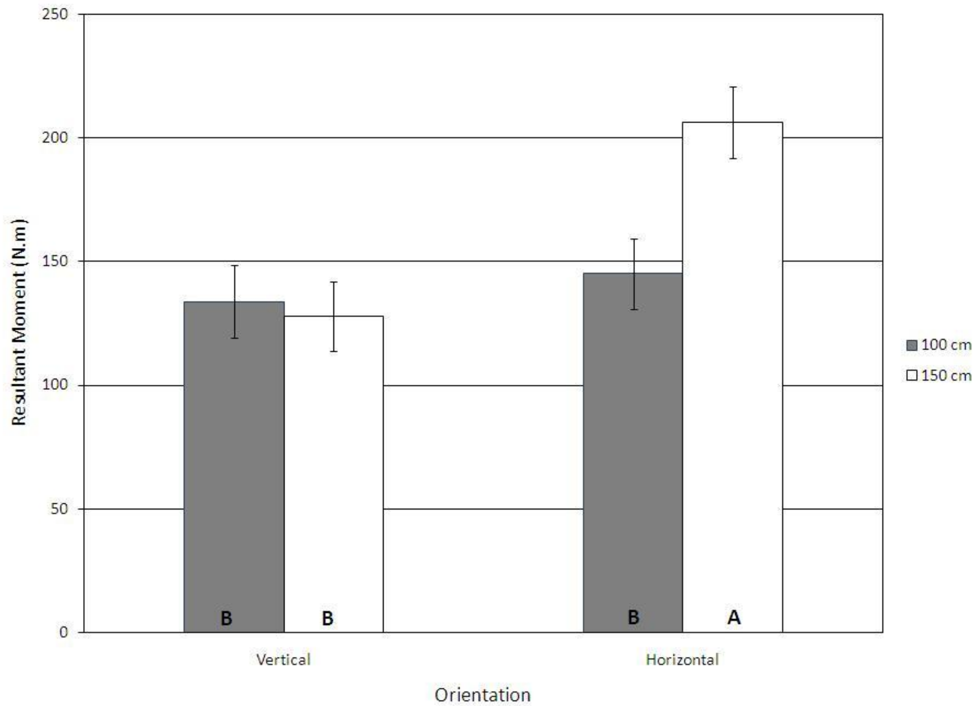

Figure 31: Effects of handle height and handle orientation on LSM resultant moments at the L5/S1 joint. Letters indicate significantly different height by orientation interactions.

4.2.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Shoulder and Low Back Moments

The push direction led to flexor moments at the shoulder joints while the pull direction resulted in extensor moments ($p < 0.0001$, Figures 32-33). The pull direction resulted in significantly greater extensor moments at the L5/S1 joint than the push direction ($p = 0.0006$, Figure 34). Mass had a main effect on resultant right shoulder ($p = 0.0064$) and low back moments ($p = 0.0001$), such that larger body masses were associated with significantly greater moments. The push direction led to significantly greater resultant right and left shoulder moments ($p < 0.0001$, p = 0.0001, Figures 35-36) whereas the pull direction resulted in significantly greater resultant low back moments ($p = 0.0004$, Figure 37). Furthermore, a main effect of elbow angle was evident on sagittal plane and resultant right shoulder moments where the flexor moments produced by extended elbows were about 3 times greater than the flexor moments caused by flexed elbows (p = 0.0022, p = 0.0049). There were no significant interaction effects.

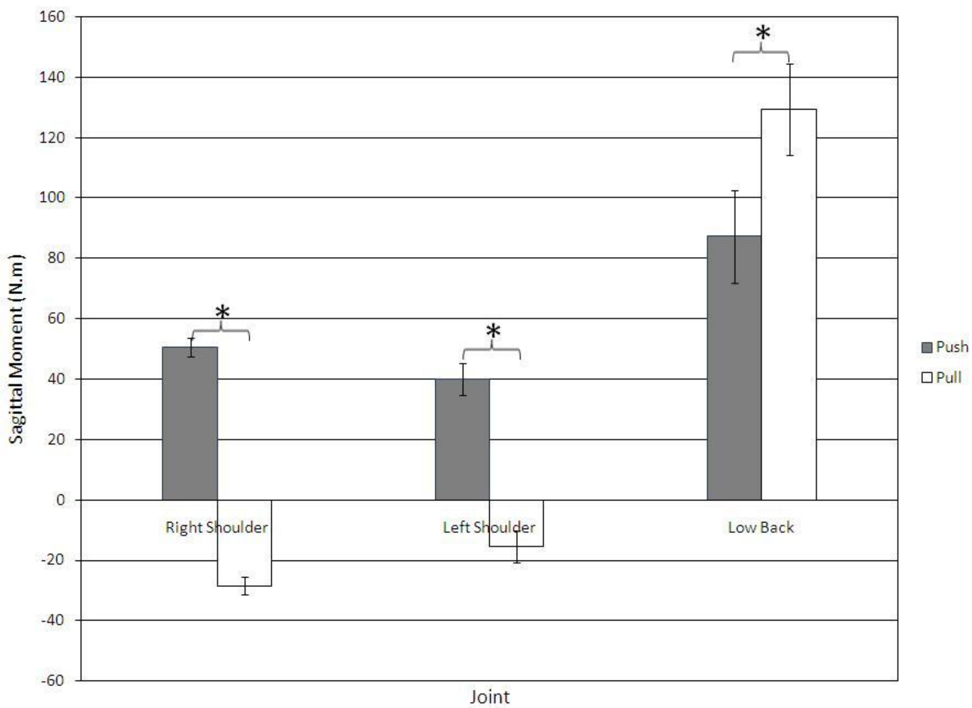

Figure 32: Effects of direction on LSM sagittal plane moments at the right and left shoulder joints and the L5/S1 joint collapsed across all conditions. * indicates significant differences.

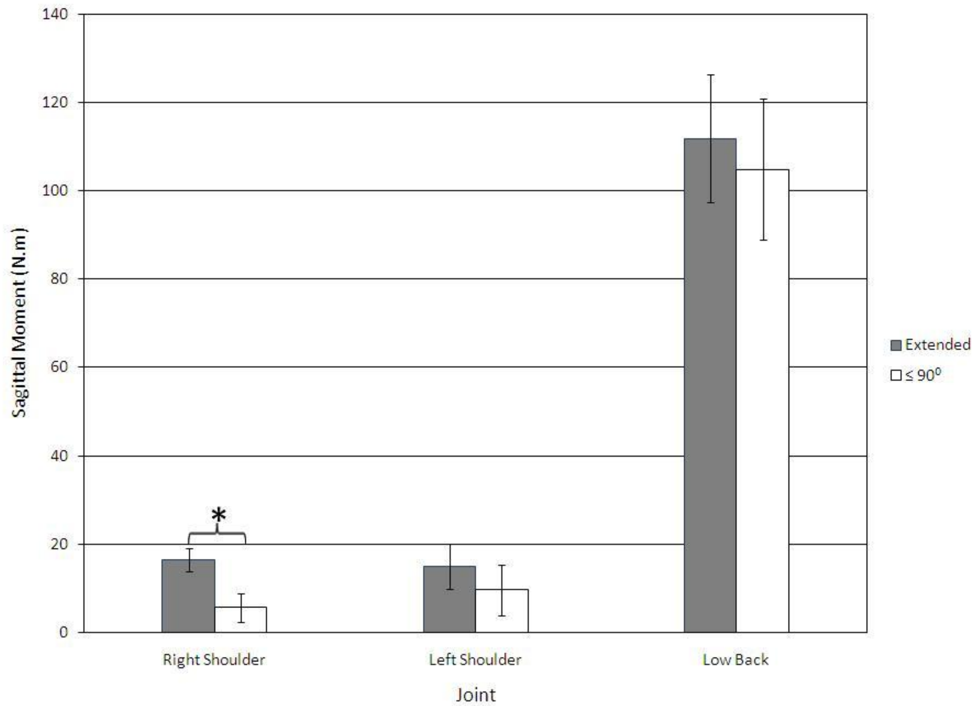

Figure 33: Effects of elbow angle on LSM sagittal moments at the right and left shoulder joints and the L5/S1 joint collapsed across all conditions. * indicates significant differences.

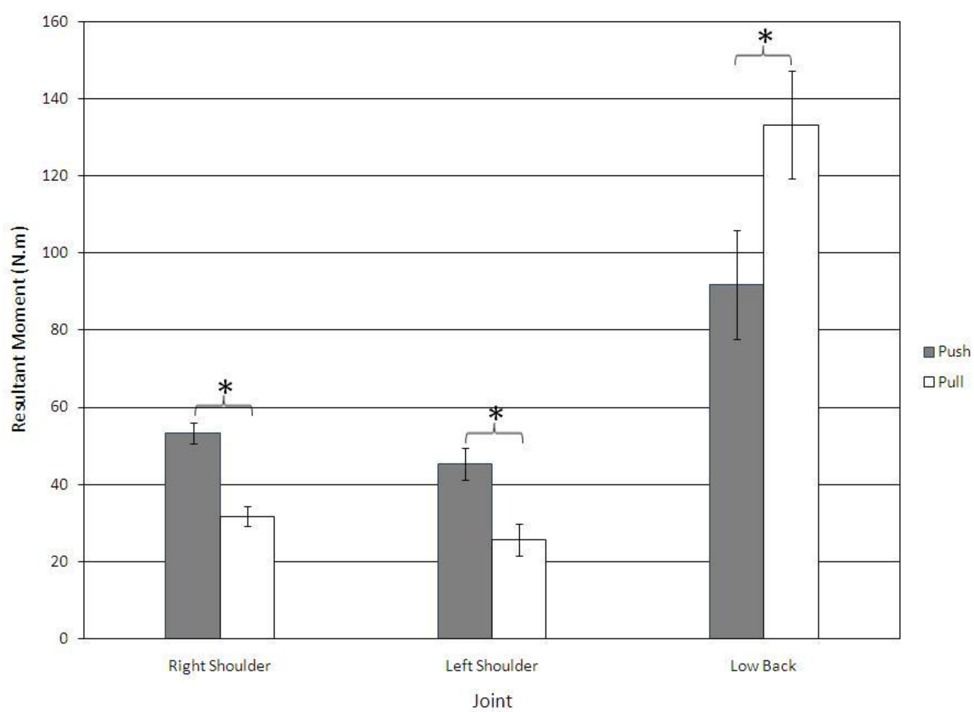

Figure 34: Effects of direction on LSM resultant moments at the right and left shoulder joints and the L5/S1 joint collapsed across all conditions. * indicates significant differences.

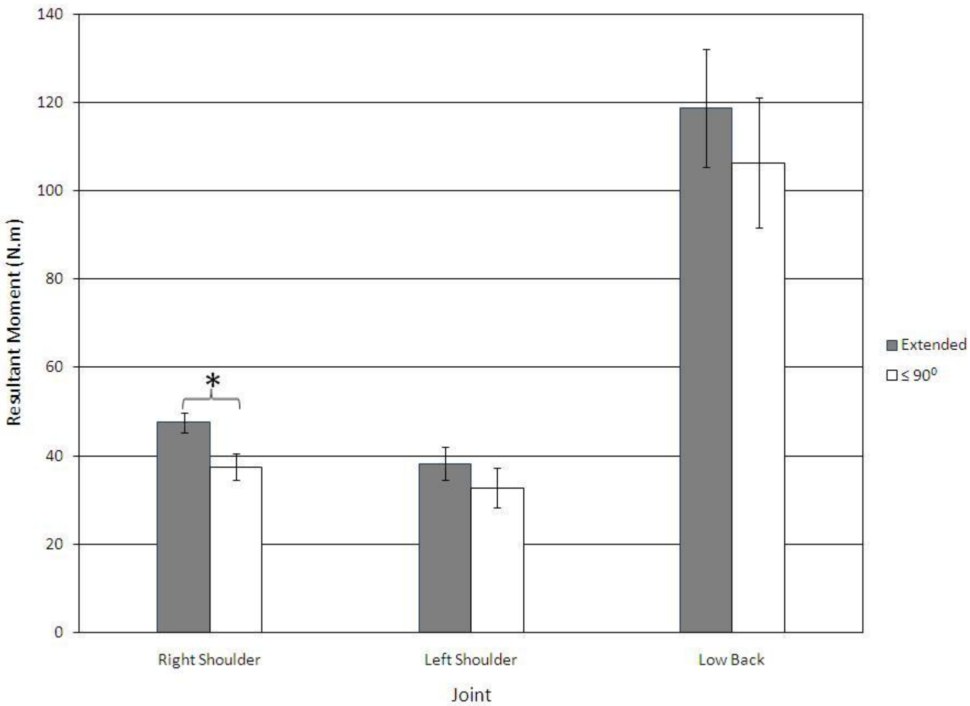

Figure 35: Effects of elbow angle on LSM resultant moments at the right and left shoulder joints and the L5/S1 joint collapsed across all conditions. * indicates significant differences.

4.3 Normalized Individual EMG

Mean percent maximal voluntary electrical activation (%MVE) of all participants by experimental conditions ranged from 6.2% to 62.0% for the fourteen muscles measured. In general, for the examined musculature, the shoulder muscles worked at a higher level of mean %MVE than the trunk muscles. Of the combined left and right muscle pairs monitored, the middle deltoid appeared to have the greatest mean %MVE while the erector spinae seemed to have the lowest mean %MVE, across conditions. Results of the GMSDHO and GMSDE models are summarized in Table 12 and Table 13, respectively. In general, direction or interaction effects between direction and height or elbow angle had the greatest influence on normalized individual EMG as determined from comparisons of ω^2.

4.3.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Normalized Individual EMG

There was a main effect of direction on mean %MVE for all muscles. Direction appeared to have the greatest effect on mean %MVE for all muscles except for the right rectus abdominis, where height was slightly more important. In general, the push direction resulted in significantly greater activation levels than the pull direction for the right and left biceps brachii, right and left triceps brachii, right and left pectoralis major, right and left middle deltoid, right and left middle trapezius, and right and left rectus abdominis muscles (Figures 36-37). Mean %MVE for the push direction was, on average, 1.76 times greater than the pull direction, pooling across muscles. Only the right and left erector spinae muscles elicited greater mean muscle activation levels for the pull direction than the push direction. The combined right and left middle deltoid muscles had the greatest mean %MVE for the push direction, but had the lowest mean %MVE for the pull direction. Height had a main effect on mean %MVE for eight of the fourteen muscles examined (Figures 38-39). Mean %MVE was greater at the 100 cm height for the right and left pectoralis major, right and left erector spinae and right biceps brachii muscles. However, mean %MVE was greater at the 150 cm height for the left triceps brachii, right middle trapezius and left rectus abdominis muscles. There was also a main effect of gender on mean %MVE for the right biceps brachii and triceps brachii muscles, in which females had greater mean %MVE values than males. A significant effect of orientation was evident for the left biceps brachii and right pectoralis major muscles where the vertical orientation resulted in greater mean %MVE estimates than the horizontal orientation. Mass only had a main effect on mean %MVE for the left deltoid muscle such that a greater mass was associated with a lower mean %MVE.

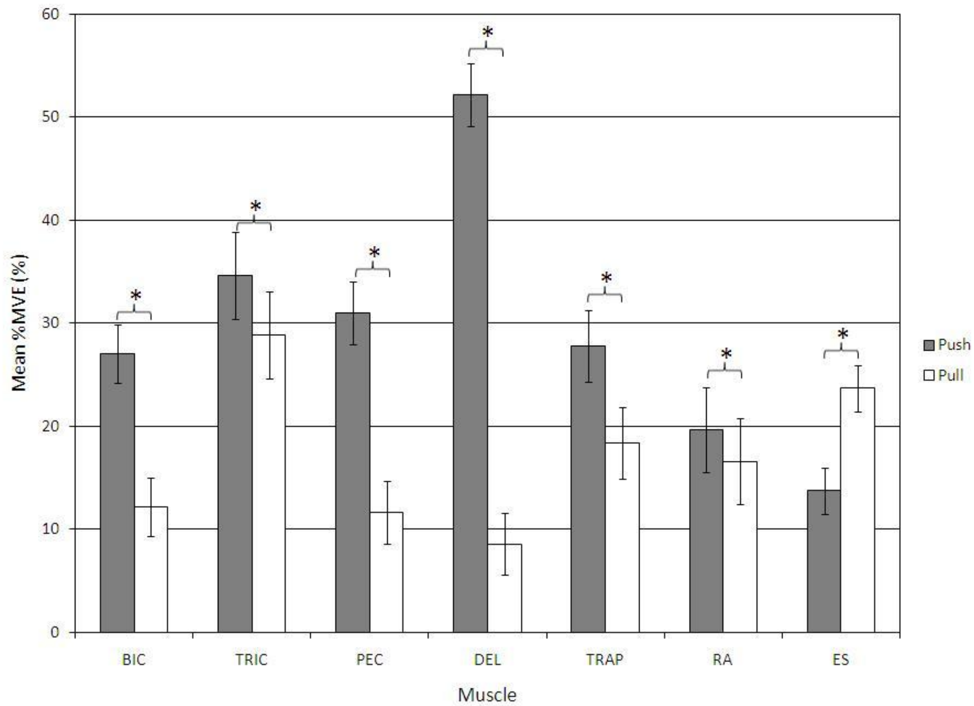

Figure 36: Effects of direction on LSM %MVE for the muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

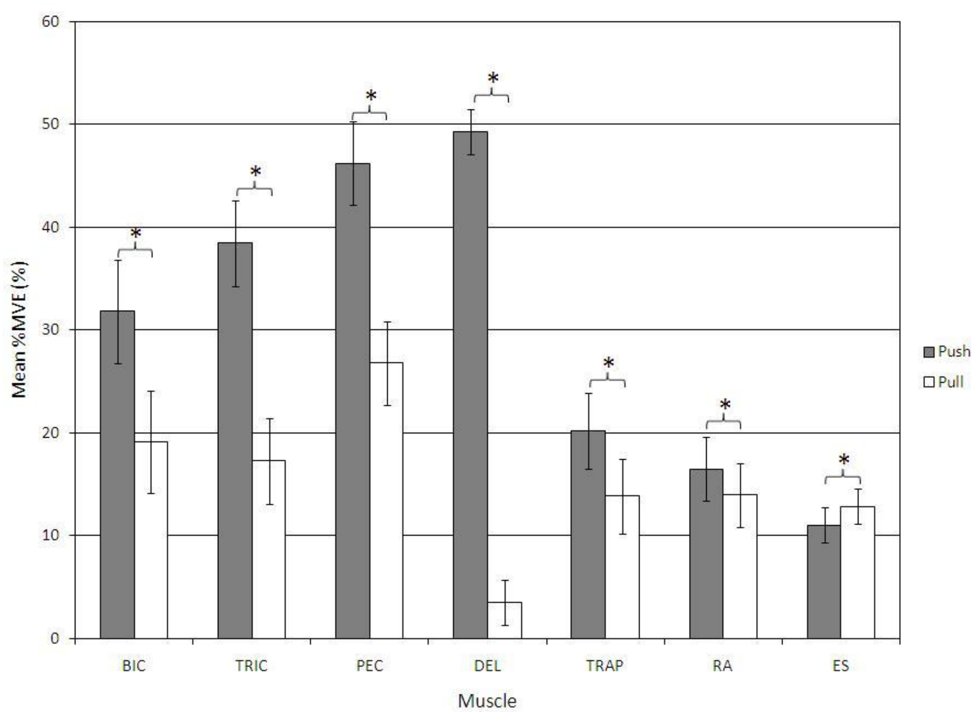

Figure 37: Effects of direction on LSM %MVE for the muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

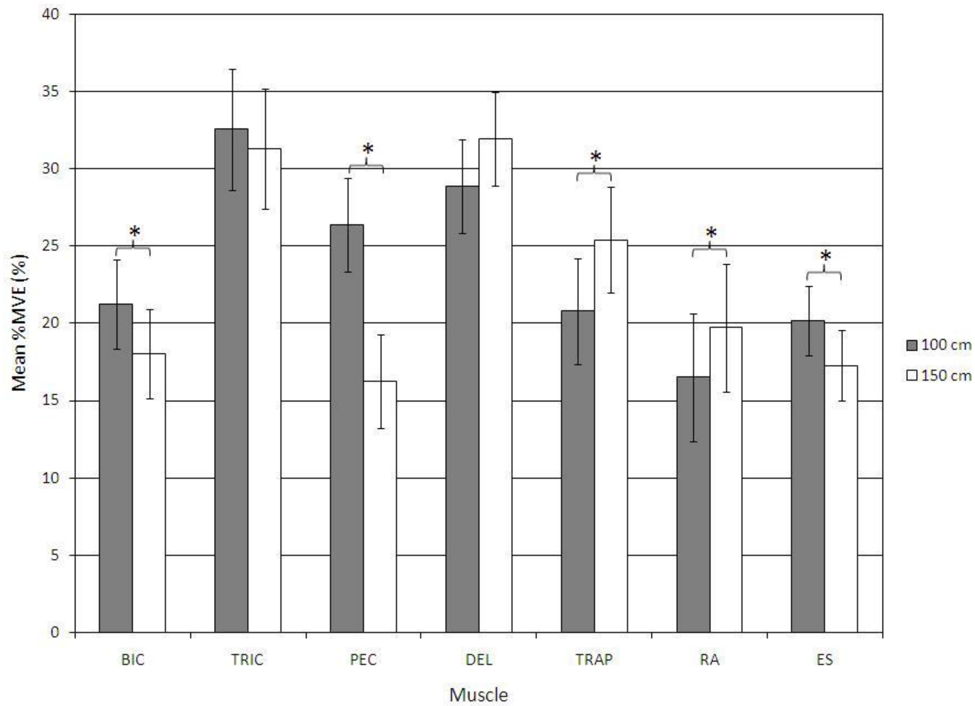

Figure 38: Effects of handle height on LSM %MVE for the muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

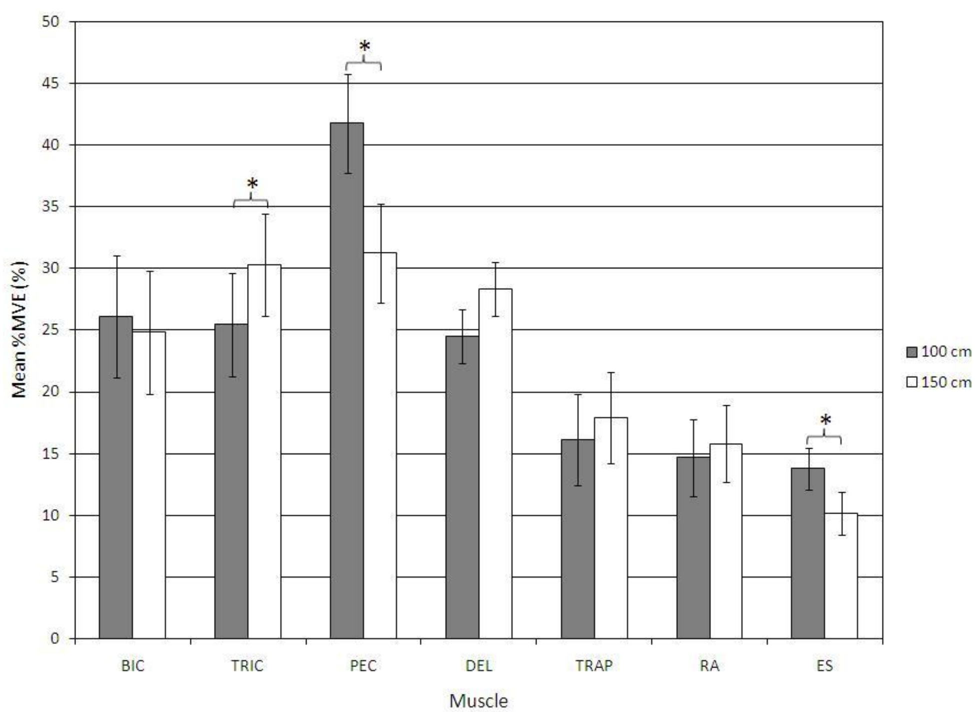

Figure 39: Effects of handle height on LSM %MVE for the muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

4.3.2 Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Normalized Individual EMG

Interaction effects existed between direction and handle height on mean %MVE for all muscles except for the left deltoid, right deltoid and left rectus abdominis muscles. Refer to Appendix A for figures of significant interaction effects between direction and handle height. An ordinal interaction occurred between direction and handle height, in which differences in mean %MVE between directions were larger for the 100 cm height than the 150 cm height for the right and left biceps brachii (Figures A1-A2) and pectoralis major muscles (Figures A5-A6). Conversely, the 150 cm height resulted in the greatest differences between direction for the left triceps brachii (Figure A4), right and left middle trapezius (Figures A7-A8), rectus abdominis (Figure A9) and right and left erector spinae muscles (Figures A10-A11). A gender by mass interaction effect was only present for the left middle deltoid muscle. Interaction effects between gender and direction were also apparent for the left biceps brachii, left middle trapezius and right pectoralis major muscles. A marked interaction effect between gender and height was evident for the left middle trapezius and right pectoralis major muscles. Furthermore, a mass by direction effect was apparent for the right and left pectoralis major muscles.

Table 12: Results of ANOVA analysis for the effects of gender, mass, direction, handle height and handle orientation on normalized individual EMG.

a) Right Muscles

Source of Variance	BIC (o = 192)	TRIC (o = 192)	PEC (o = 187)	DEL (o = 192)	TRAP (o = 192)	RA (o = 192)	ES (o = 192)
Gender (G)	0.0335*	0.0398*	0.7760	0.4873	0.9215	0.4077	0.6810
Mass (M)	0.9749	0.7553	0.1847	0.8393	0.3860	0.2725	0.3730
Stature (S)	0.2261	0.6169	0.5865	0.3280	0.8465	0.2194	0.7285
Direction (D)	<.0001*	0.0064*	<.0001*	<.0001*	<.0001*	0.0061*	<.0001*
Height (H)	0.0216*	0.5466	<.0001*	0.0742	0.0010*	0.0037*	0.0009*
Orientation (O)	0.1056	0.4263	0.0478*	0.0617	0.3532	0.8855	0.8968
G x M	0.6546	0.7448	0.8132	0.9496	0.1516	0.4367	0.4378
G x S	0.5337	0.2426	0.1269	0.6489	0.3962	0.6136	0.9346
G x D	0.0346*	0.0492*	0.0213*	0.2829	0.0513	0.4529	0.1741
G x H	0.1633	0.9117	0.0082*	0.1103	0.3084	0.0569	0.5642
G x O	0.5203	0.2949	0.6367	0.3644	0.9930	0.8399	0.6351
M x S	0.5952	0.2271	0.1550	0.1724	0.0160*	0.1647	0.7939
M x D	0.5956	0.4952	0.0069*	0.6430	0.1034	0.7078	0.8625
M x H	0.2936	0.6638	0.3414	0.7140	0.9948	0.9422	0.7455
M x O	0.3427	0.5422	0.4878	0.9472	0.8495	0.6896	0.9215
S x D	0.2599	0.0695	0.2099	0.2158	0.3774	0.0142*	0.2940
S x H	0.6538	0.2265	0.3771	0.0078*	0.8285	0.2552	0.4517
S x O	0.4203	0.5254	0.3973	0.1777	0.7468	0.7981	0.5904
D x H	0.0003*	<.0001*	<.0001*	0.0614	0.0005*	0.0394*	0.0023*
D x O	0.2277	0.4263	0.1216	0.9403	0.8193	0.9233	0.9508
H x O	0.7669	0.5196	0.4004	0.3795	0.4790	0.5887	0.8435

continued

Table 12 continued

b) Left Muscles

Source of Variance	BIC (o = 184)	TRIC (o = 184)	PEC (o = 192)	DEL (o = 190)	TRAP (o = 192)	RA (o = 192)	ES (o = 192)
Gender (G)	0.2491	0.2464	0.9372	0.1894	0.7687	0.1439	0.5505
Mass (M)	0.6915	0.5617	0.5414	0.0314*	0.6723	0.3174	0.5140
Stature (S)	0.9921	0.8755	0.4447	0.1102	0.4514	0.3757	0.3585
Direction (D)	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*	0.0020*	0.0047*
Height (H)	0.5167	0.0166*	<.0001*	0.0574	0.2021	0.1583	<.0001*
Orientation (O)	0.0157*	0.5870	0.3631	0.6808	0.9029	0.9137	0.4490
G x M	0.9844	0.4251	0.8911	0.1482	0.7779	0.8758	0.2840
G x S	0.7230	0.3303	0.6687	0.3930	0.5446	0.7991	0.4925
G x D	0.0143*	0.6904	0.3559	0.1442	<.0001*	0.3092	0.2573
G x H	0.7119	0.7008	0.4462	0.1706	0.0793	0.9243	0.3441
G x O	0.5945	0.5005	0.9217	0.6916	0.6686	0.2809	0.4623
M x S	0.6297	0.3166	0.4421	0.2279	0.3832	0.7007	0.6232
M x D	0.0838	0.0702	<.0001*	0.0036*	0.7576	0.9998	0.3825
M x H	0.1345	0.6022	0.5949	0.1641	0.7500	0.3947	0.8477
M x O	0.4230	0.9697	0.8211	0.9016	0.8013	0.7096	0.9163
S x D	0.0238*	0.2195	0.0002*	0.0083*	0.4527	0.1700	0.4022
S x H	0.1712	0.2987	0.2045	0.7955	0.3629	0.9958	0.5165
S x O	0.2477	0.5835	0.4771	0.6550	0.7321	0.6804	0.3489
D x H	0.0009*	0.0001*	<.0001*	0.3081	0.0031*	0.4787	0.0101*
D x O	0.5023	0.2668	0.0023*	0.6141	0.3325	0.9253	0.9845
H x O	0.7928	0.8217	0.7448	0.6636	0.4619	0.7482	0.4975

* indicates statistically significant differences.

4.3.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Normalized Individual EMG

There was a main effect of elbow angle on mean %MVE where greater mean %MVE resulted when the elbows were fully extended for the right triceps brachii, left pectoralis major and for the right and left middle deltoid muscles. A main effect of elbow angle was also evident for the right and left biceps brachii, right and left middle trapezius and left erector spinae muscles, in which greater mean %MVE resulted when the elbows were $\leq 90^{o}$ (Figures 40-41). There was also a main effect of direction on mean %MVE such that greater mean %MVE values arose for the push direction than the pull direction for the right and left biceps brachii, right and left pectoralis major, and right and left middle deltoid muscles. The biceps brachii, pectoralis major and middle deltoid muscles were, on average, 2.6, 3.4 and 3.1 times more active for push exertions than pull exertions, pooling across sides. The opposite trend was evident for the right triceps brachii and right erector spinae muscles (Figures 43-43). A main effect of gender and mass were only apparent for the right triceps and right deltoid muscles, respectively.

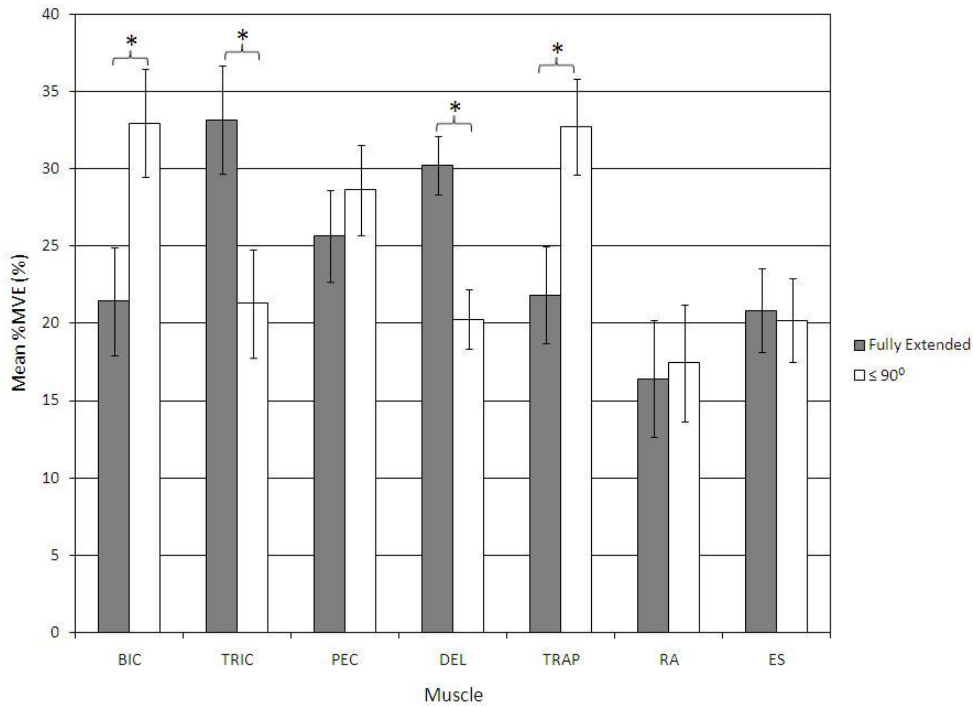

Figure 40: Effects of elbow angle on LSM %MVE for the muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

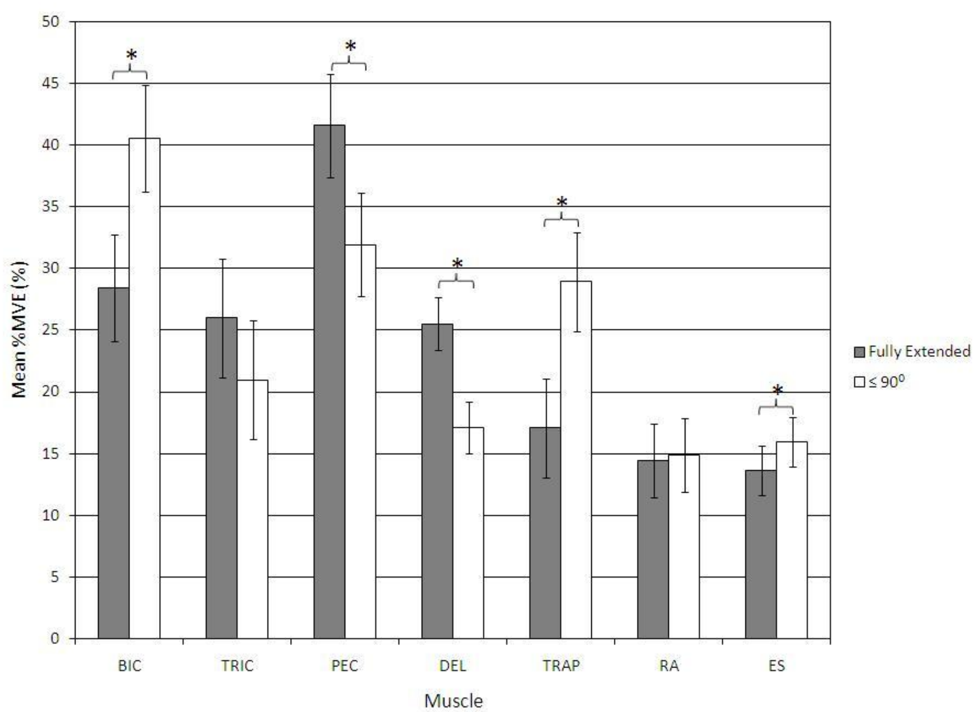

Figure 41: Effects of elbow angle on LSM %MVE for the muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

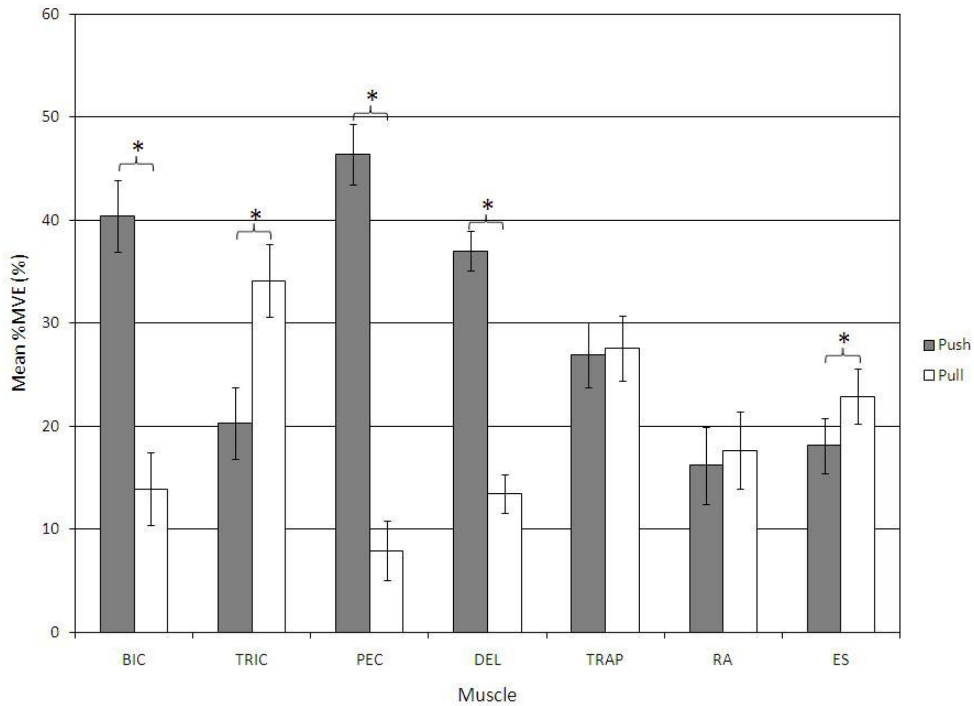

Figure 42: Effects of direction on LSM %MVE for the muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

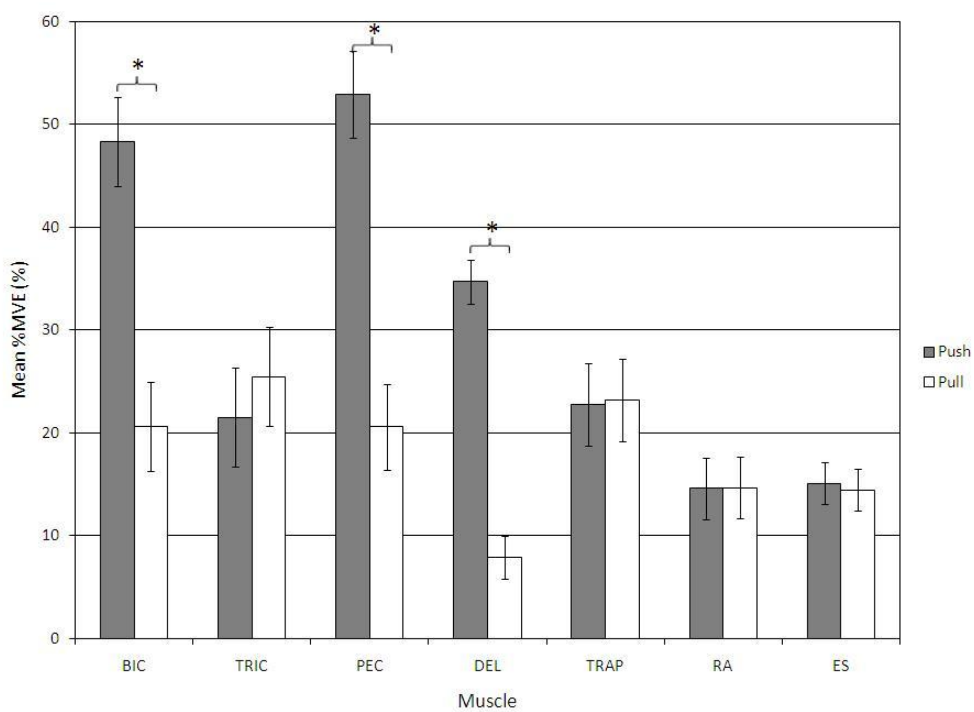

Figure 43: Effects of direction on LSM %MVE for the muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

4.3.4 Interaction Effects of Gender, Mass, Stature, Direction and Elbow Angle on Normalized Individual EMG

Interaction effects were present between direction and elbow angle on mean %MVE for half of the muscles examined (Appendix B). These included the right and left triceps brachii (Figures B1-B2), right pectoralis major (Figure B3) and right middle trapezius (Figure B6) muscles. Ordinal interactions between direction and elbow angle were also present for the right and left middle deltoid (Figures B4-B5) and right erector spinae muscles (Figure B7) such that greater mean %MVE differences between directions existed for the fully extended elbows condition. Interaction effects were also evident between gender and direction for the right and left deltoid. Further, a mass by direction effect was found for the right middle trapezius muscle and a mass by elbow angle effect was found for the left erector spinae and right middle trapeizus muscles.

Table 13: Results of ANOVA analysis for the effects of gender, mass, direction and elbow angle on normalized individual EMG.

a) Right Muscles

Source of Variance	BIC (o = 96)	TRIC (o = 96)	PEC (o = 93)	DEL (o = 96)	TRAP (o = 96)	RA (o = 96)	ES (o = 96)
Gender (G)	0.3006	0.0204*	0.7665	0.2439	0.6577	0.2506	0.6113
Mass (M)	0.8047	0.7166	0.2195	0.1162	0.9821	0.2341	0.6692
Stature (S)	0.5190	0.9332	0.5684	0.5877	0.9983	0.3151	0.6209
Direction (D)	<.0001*	<.0001*	<.0001*	<.0001*	0.7779	0.2309	0.0006*
Elbow Angle (E)	0.0006*	0.0002*	0.4162	<.0001*	<.0001*	0.4018	0.6260
G x M	0.4677	0.5840	0.7566	0.2661	0.2740	0.3384	0.4805
G x S	0.3879	0.1390	0.2225	0.5858	0.1179	0.5634	0.5832
G x D	0.6688	0.4651	0.0798	0.0035*	0.1909	0.4323	0.1337
G x E	0.8830	0.1830	0.3529	0.3297	0.2405	0.3879	0.5849
M x S	0.8520	0.2389	0.0152*	0.7600	0.0268*	0.0931	0.8968
M x D	0.3656	0.7172	0.3662	0.5275	0.1539	0.6056	0.5618
M x E	0.3934	0.7482	0.6756	0.5640	0.0050*	0.9382	0.9615
S x D	0.5778	0.5219	0.1112	0.2824	0.1062	0.1793	0.2340
S x E	0.6806	0.6210	0.6298	0.9003	0.2190	0.3589	0.6502
D x E	0.0812	0.0144*	0.0483*	<.0001*	0.0169*	0.0334*	0.0475*

continued

Table 13 continued

b) Left Muscles

Source of Variance	BIC (o = 92)	TRIC (o = 92)	PEC (o = 96)	DEL (o = 95)	TRAP (o = 96)	RA (o = 96)	ES (o = 96)
Gender (G)	0.1091	0.1548	0.9325	0.0139*	0.5083	0.1261	0.6595
Mass (M)	0.6738	0.6300	0.6581	0.7410	0.9855	0.2802	0.9951
Stature (S)	0.2144	0.5733	0.8881	0.1554	0.5941	0.3953	0.5919
Direction (D)	<.0001*	0.2634	<.0001*	<.0001*	0.8491	0.9620	0.4703
Elbow Angle (E)	0.0034*	0.1559	0.0070*	0.0003*	<.0001*	0.6723	0.0127*
G x M	0.8819	0.5020	0.9329	0.8591	0.9124	0.8575	0.2656
G x S	0.3532	0.3075	0.5862	0.2828	0.8451	0.9879	0.2446
G x D	0.6397	0.5226	0.2709	0.0061*	0.0190*	0.5717	0.4718
G x E	0.6925	0.6954	0.8465	0.6134	0.6450	0.8656	0.6556
M x S	0.7083	0.4299	0.7355	0.0327*	0.3311	0.4745	0.5702
M x D	0.8499	0.7346	0.1219	0.3499	0.9592	0.3724	0.8474
M x E	0.3572	0.6796	0.8082	0.4618	0.1037	0.4497	0.0879
S x D	0.6389	0.7873	0.1276	0.0186*	0.8492	0.3636	0.0431*
S x E	0.3515	0.8619	0.4291	0.8487	0.3941	0.5556	0.9360
D x E	0.0721	<.0001*	0.1211	<.0001*	0.1559	0.0222*	0.5133

* indicates statistically significant differences.

4.4 Total and Weighted Average EMG

Total EMG represents the collective mean %MVE for all muscles measured. For a given mean %MVE, a muscle with a larger PCSA may have the ability to contribute more to a force than a muscle with a smaller PCSA, notwithstanding the different size of the moment arms, muscle lengths and task factors. Thus, total EMG was weighted to individual muscle PCSAs (Equations 7-9). Weighted average EMG values generally decreased from their respective total EMG values, indicating that apparently the smaller muscles were proportionately more active than the larger muscles. On average, total left EMG values ranged from 15.8% to 33.2%, total right EMG values ranged from 17.4% to 32.1%, while total EMG values ranged from 16.6% to 32.4%. The lowest mean %MVE values resulted for the pull direction at 150 cm with the handles oriented horizontally and the elbows fully extended. Similar values were recorded for weighted average EMG values. Results of the GMSDHO and GMSDE models are summarized in Table 14 and Table 15, respectively. Direction had the greatest effects on total and weighted average EMG as judged by ω^2.

4.4.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Total and Weighted Average EMG

There was a main effect of direction on all measures of total and weighted average EMG, in which the push direction resulted in an average of 71% greater total and weighted average EMG than the pull direction (Figure 44). There was also a main effect of gender on total left and total EMG as well as on weighted average left and total EMG such that females activated their muscles to a greater extent than males (Figure 45). Female left and total EMG values were 1.2 times greater than male values while female weighted average left and total EMG values were 1.3 times greater than male values.

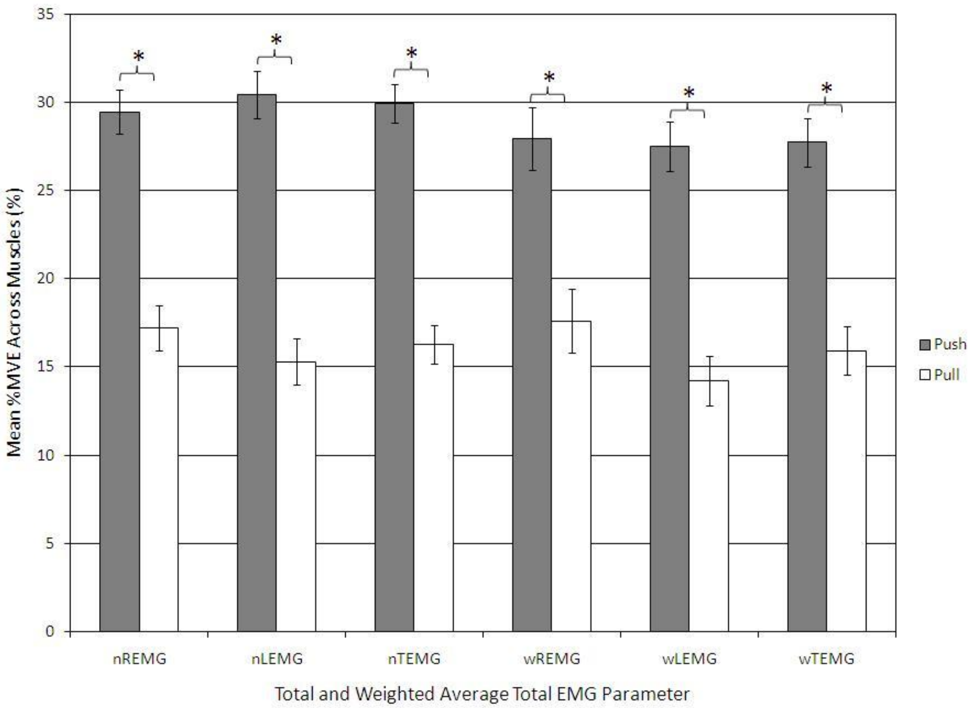

Figure 44: Effects of direction on LSM total and weighted average total EMG collapsed across all conditions. * indicates significant differences.

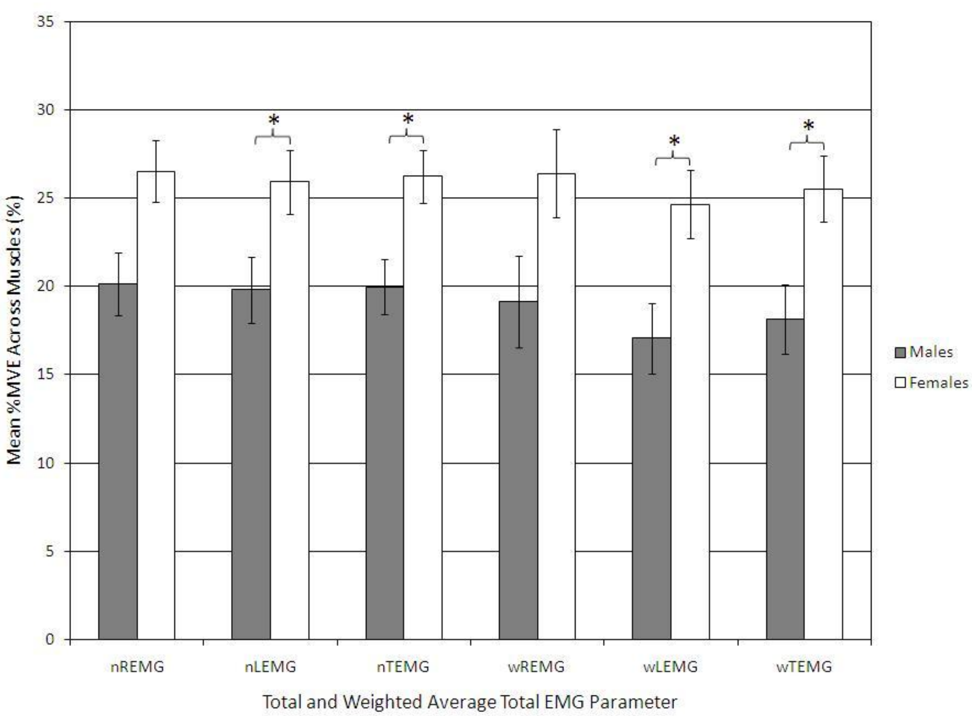

Figure 45: Effects of gender on LSM total and weighted average total EMG collapsed across all conditions. * indicates significant differences.

4.4.2 Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Total and Weighted Average EMG

There was an interaction effect between gender and mass on left total EMG and weighted average left EMG. There was also an ordinal interaction between direction and height on weighted average right EMG such that differences in mean %MVE between directions were larger at the 150 cm handle height than at the 100 cm handle height (Figure 46). In addition, both the greatest and least total mean %MVE occurred at the 150 cm height.

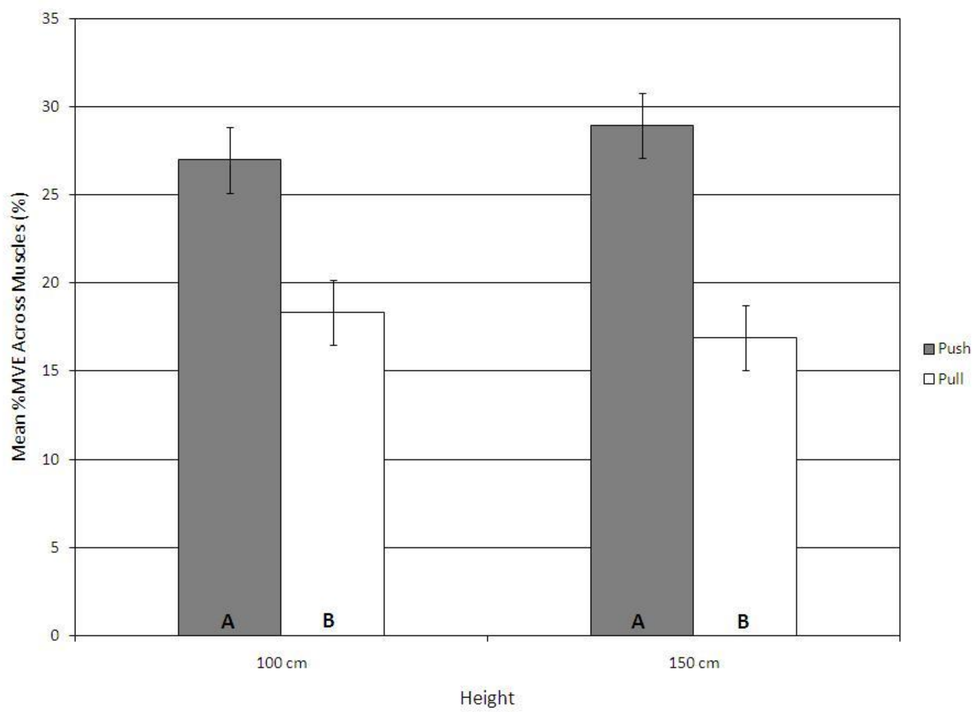

Figure 46: Effects of direction and handle height on LSM weighted average right EMG. Letters indicate significantly different direction by handle height interactions.

Table 14: Results of ANOVA analysis for the effects of gender, mass, direction and handle height on total and weighted average EMG.

	Total EMG			Weighted Average EMG		
Source of Variance	nREMG	nLEMG	nTEMG	wREMG	wLEMG	wTEMG
Gender (G)	0.0209*	0.0339*	0.0107*	0.0638	0.0155*	0.0160*
Mass (M)	0.2552	0.5265	0.3010	0.3052	0.3889	0.2692
Stature (S)	0.3662	0.7832	0.7226	0.2206	0.8565	0.3664
Direction (D)	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*
Height (H)	0.1294	0.3854	0.1821	0.6907	0.5040	0.5397
Orientation (O)	0.7998	0.4188	0.6965	0.5030	0.6799	0.9066
G x M	0.4170	0.3950	0.9628	0.5156	0.3828	0.9808
G x S	0.0502	0.9665	0.2353	0.1942	0.8549	0.3428
G x D	0.0577	0.8814	0.3914	0.1218	0.2911	0.8435
G x H	0.9978	0.4828	0.6659	0.2726	0.4552	0.3009
G x O	0.6483	0.9931	0.8108	0.5492	0.8794	0.8115
M x S	0.0030*	0.4787	0.0266*	0.0241*	0.3519	0.0500
M x D	0.0195*	0.0449*	0.0137*	0.1052	0.2686	0.1253
M x H	0.1791	0.5991	0.3062	0.3294	0.5877	0.3936
M x O	0.5735	0.6922	0.5842	0.6995	0.7076	0.6628
S x D	0.0282*	0.0991	0.0321*	0.0058*	0.2307	0.0279*
S x H	0.1558	0.9433	0.4426	0.0464*	0.7316	0.1973
S x O	0.1697	0.4268	0.2203	0.2697	0.5565	0.3401
D x H	0.4383	0.2848	0.7812	0.0066*	0.7183	0.0880
D x O	0.1807	0.0572	0.0578	0.3485	0.1420	0.1683
H x O	0.4013	0.8530	0.5781	0.3080	0.7684	0.4670

* indicates statistically significant differences.

4.4.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Total and Weighted Average EMG

There was a main effect of direction on all total and weighted average EMG measures, in which the push direction yielded, on average, between 1.3 and 1.6 times greater total and weighted average EMG than the pull direction (Figure 47). There was also a main effect of gender on total left EMG and weighted average left and total EMG measures such that females had greater total and weighted average EMG levels than males (Figure 48). There was no effect of elbow angle on total or weighted average EMG.

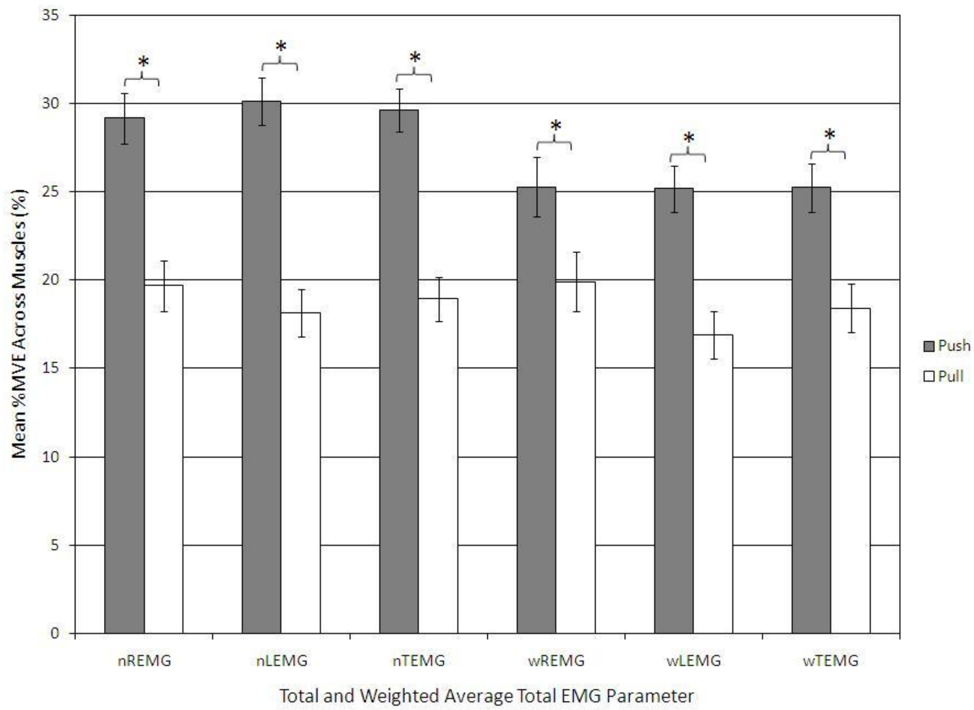

Figure 47: Effects of direction on LSM total and weighted average total EMG collapsed across all conditions. * indicates significant differences.

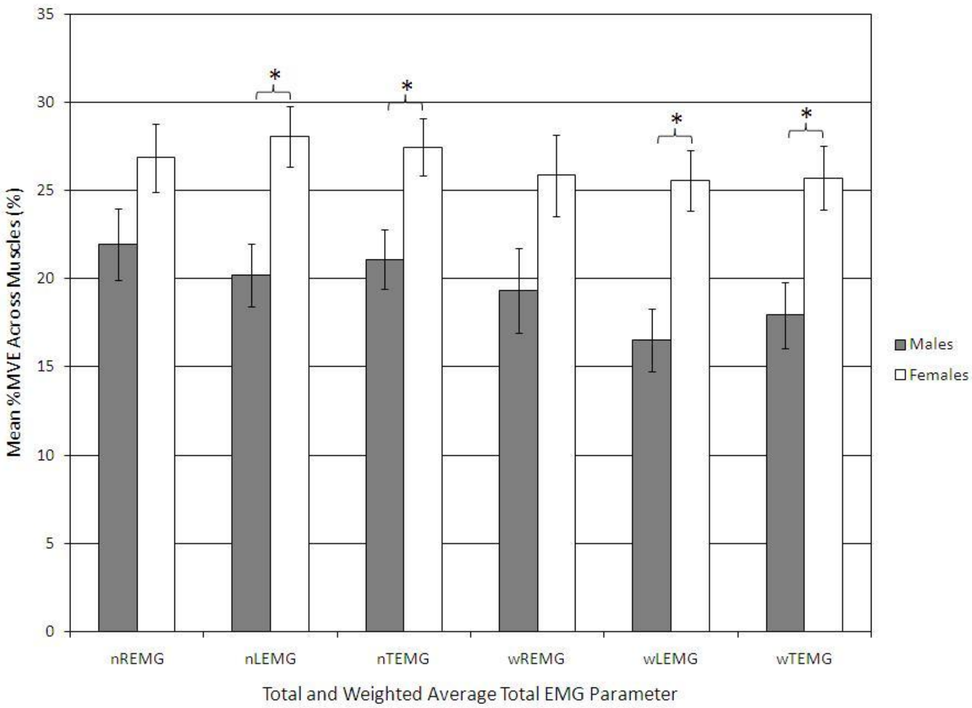

Figure 48: Effects of gender on LSM total and weighted average total EMG collapsed across all conditions. * indicates significant differences.

4.4.4 Interaction Effects of Gender, Mass, Stature, Direction and Elbow Angle on Total and Weighted Average EMG

An ordinal interaction effect between direction and elbow angle was evident for all measures of total and weighted average EMG, in which differences in mean %MVE between directions were larger for the fully extended elbows condition than the flexed elbows condition. The direction by elbow angle effects for total EMG are shown in Figures 49-51, while those for weighted average EMG are displayed in Figures 52-54. A gender by mass interaction effect was also present for the measure of weighted average left EMG.

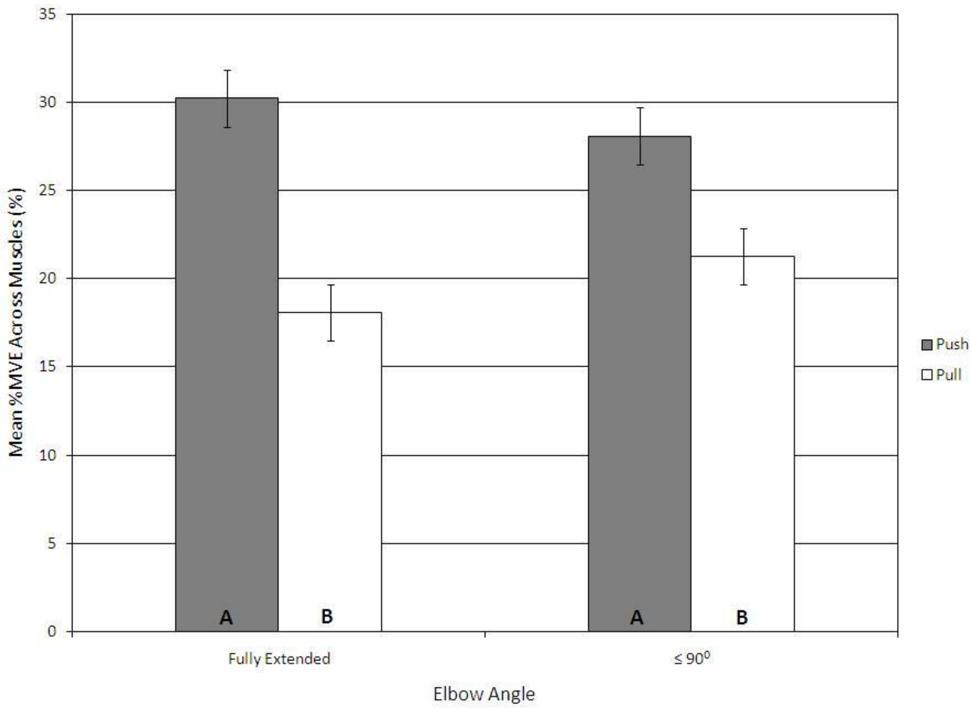

Figure 49: Effects of direction and elbow angle on LSM total right EMG. Letters indicate significantly different direction by elbow angle interactions.

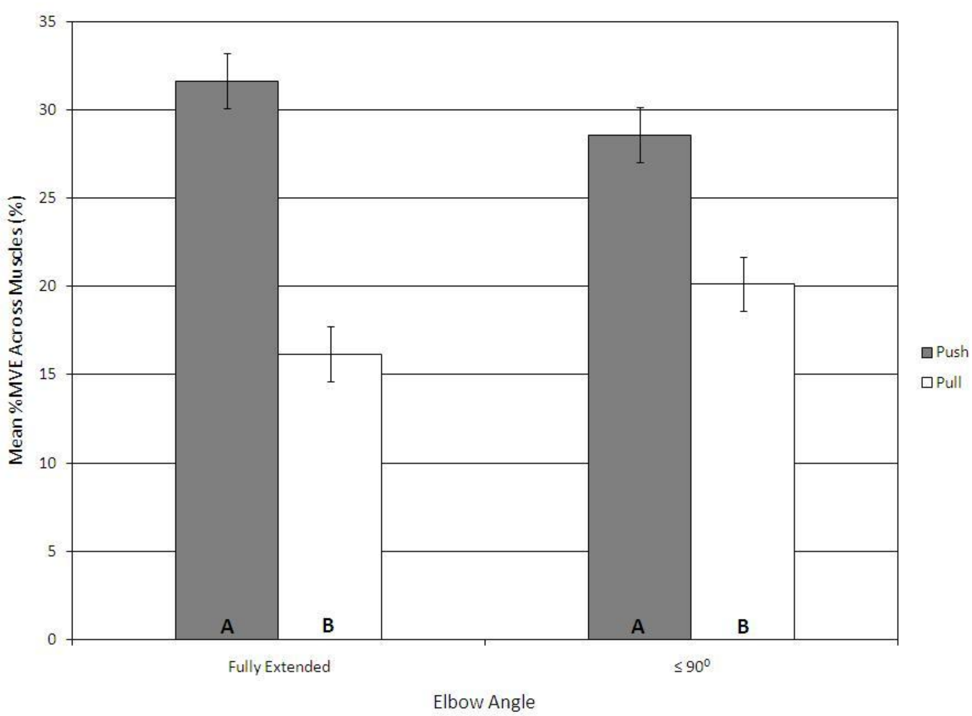

Figure 50: Effects of direction and elbow angle on LSM total left EMG. Letters indicate significantly different direction by elbow angle interactions.

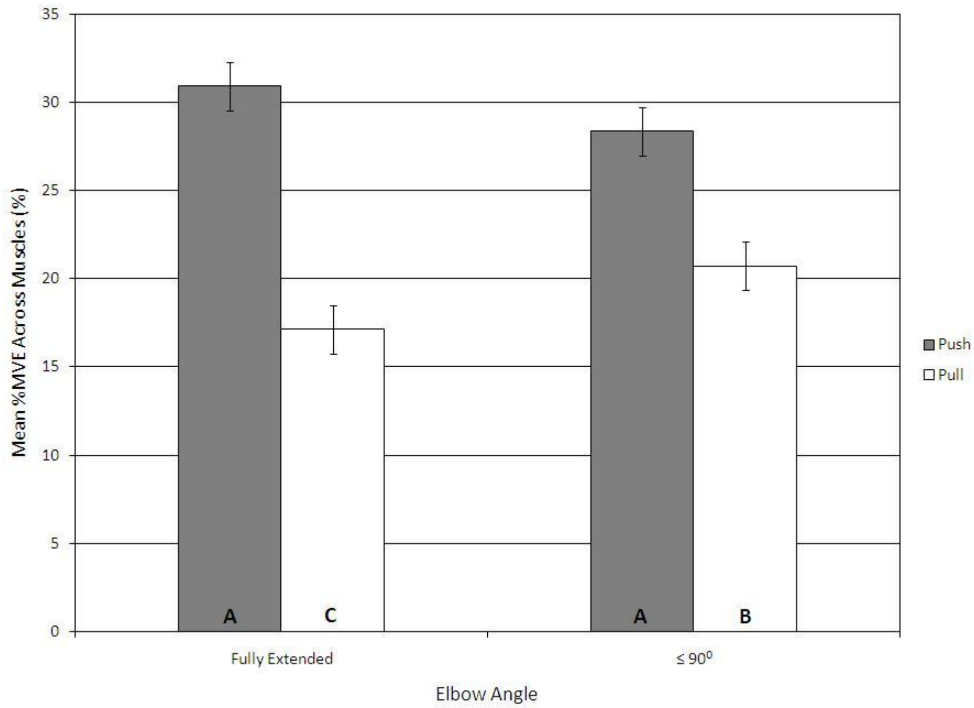

Figure 51: Effects of direction and elbow angle on LSM total EMG. Letters indicate significantly different direction by elbow angle interactions.

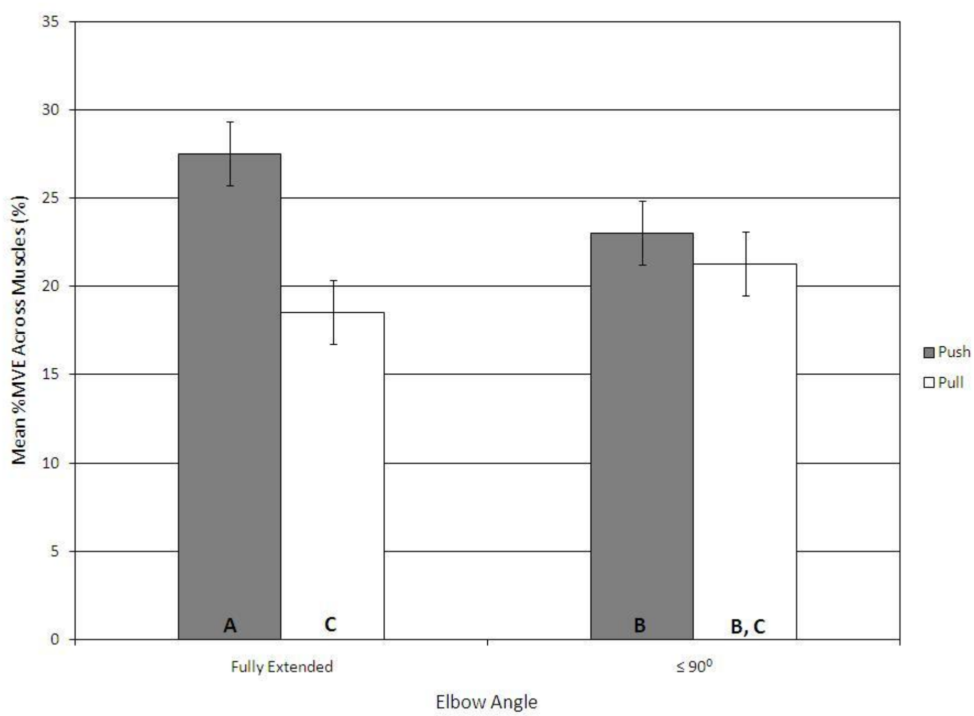

Figure 52: Effects of direction and elbow angle on LSM weighted average right EMG. Letters indicate significantly different direction by elbow angle interactions.

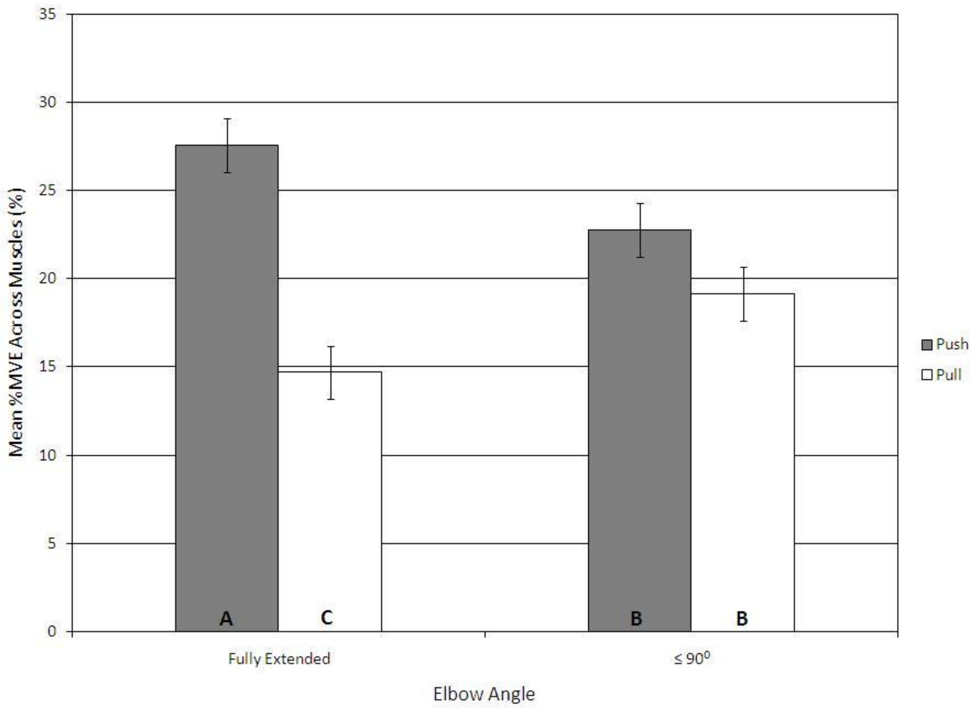

Figure 53: Effects of direction and elbow angle on LSM weighted average left EMG. Letters indicate significantly different direction by elbow angle interactions.

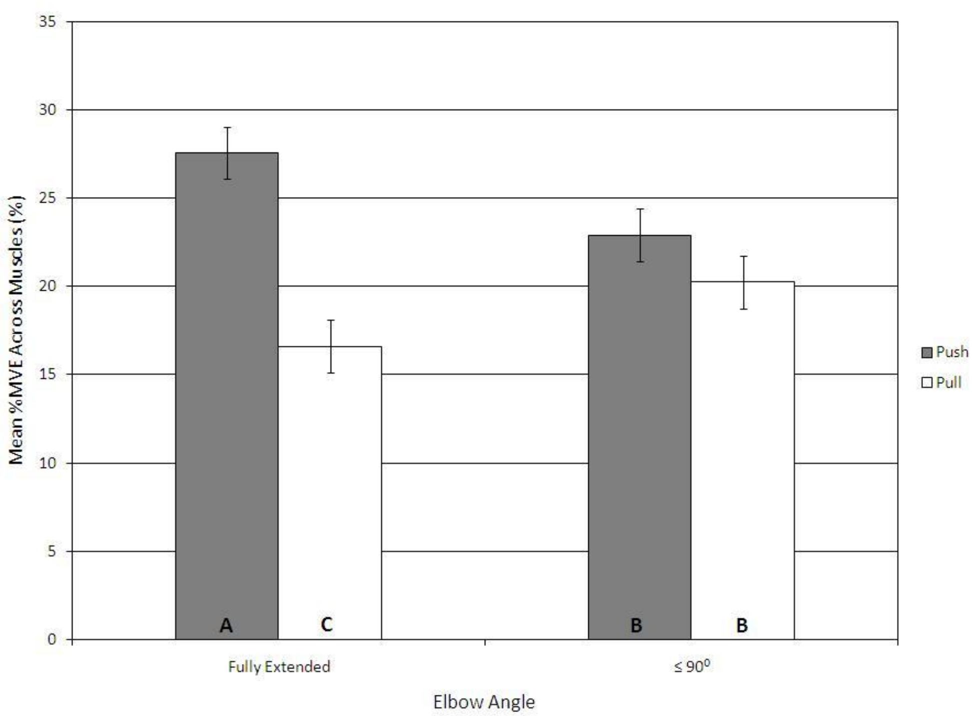

Figure 54: Effects of direction and elbow angle on LSM weighted average total EMG. Letters indicate significantly different direction by elbow angle interactions.

Table 15: Results of ANOVA analysis for the effects of gender, mass, direction and elbow angle on total and weighted average EMG.

	Total EMG			Weighted Average EMG		
Source of Variance	nREMG	nLEMG	nTEMG	wREMG	wLEMG	wTEMG
Gender (G)	0.1017	0.0067*	0.0161*	0.0693	0.0024*	0.0092*
Mass (M)	0.7885	0.5416	0.6385	0.4690	0.2599	0.3220
Stature (S)	0.7954	0.2989	0.6732	0.4622	0.7040	0.7794
Direction (D)	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*
Elbow Angle (E)	0.5813	0.6719	0.5635	0.3446	0.8667	0.5537
G x M	0.8165	0.5357	0.8486	0.6837	0.4866	0.9383
G x S	0.0697	0.1739	0.0758	0.1422	0.2016	0.1271
G x D	0.7418	0.5192	0.8459	0.4549	0.8964	0.6226
G x E	0.2509	0.6402	0.7219	0.1276	0.7905	0.5135
M x S	0.0183*	0.0931	0.0233*	0.0306*	0.0666	0.0266*
M x D	0.2797	0.9129	0.5181	0.5256	0.6390	0.9629
M x E	0.4800	0.9322	0.6431	0.6555	0.6935	0.9964
S x D	0.7312	0.9155	0.7776	0.9278	0.8769	0.9503
S x E	0.6631	0.2681	0.3456	0.5057	0.3820	0.3774
D x E	0.0074*	0.0020*	0.0008*	0.0002*	<.0001*	<.0001*

* indicates statistically significant differences.

4.5 Normalized Individual EMG/force Ratios

EMG/force ratios were calculated to provide a measure of the mean %MVE associated with each unit of force (Equations 10-11). Normalized individual EMG/force ratios provide an estimate of individual muscle contributions to total hand force capability. Smaller ratios indicate that the exertion required less activation of the specific muscle per unit of force while larger ratios indicate that the exertion required more activation of the specific muscle per unit of force. Ratios ranged from 0.046 to 0.39. On average, the left erector spinae EMG/force ratios were the lowest and the right triceps brachii EMG/force ratios were the highest. Results of the GMSDHO and GMSDE models are summarized in Table 16 and Table 17, respectively. Comparisons of ω^2 from the ANOVA output showed that direction or direction interactions with height or elbow angle explained the greatest proportion of the variance among normalized individual EMG/force ratios.

4.5.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Normalized Individual EMG/force Ratios

A main effect of direction was evident where higher ratios resulted for the pull direction than the push direction for the right triceps brachii, right and left middle trapezius, right and left rectus abdominis, and right and left erector spinae muscles. The reverse effect was apparent for the right and left deltoid (Figures 55-56). Height also had a main effect on EMG/force ratios such that the 150 cm height resulted in higher ratios than the 100 cm height. This applied to the left biceps brachii, left triceps brachii, right and left deltoid, right and left middle trapezius, right and left rectus abdominis, and right erector spinae (Figures 57-58). Gender had a main effect on the right biceps brachii, right and left triceps brachii and left deltoid EMG/force ratios with females having in higher EMG/force ratios than males (Figures 59-60). Finally, mass had a main effect on the right triceps brachii, right and left deltoid and right erector spinae EMG/force ratios where higher ratios were seen with smaller body masses.

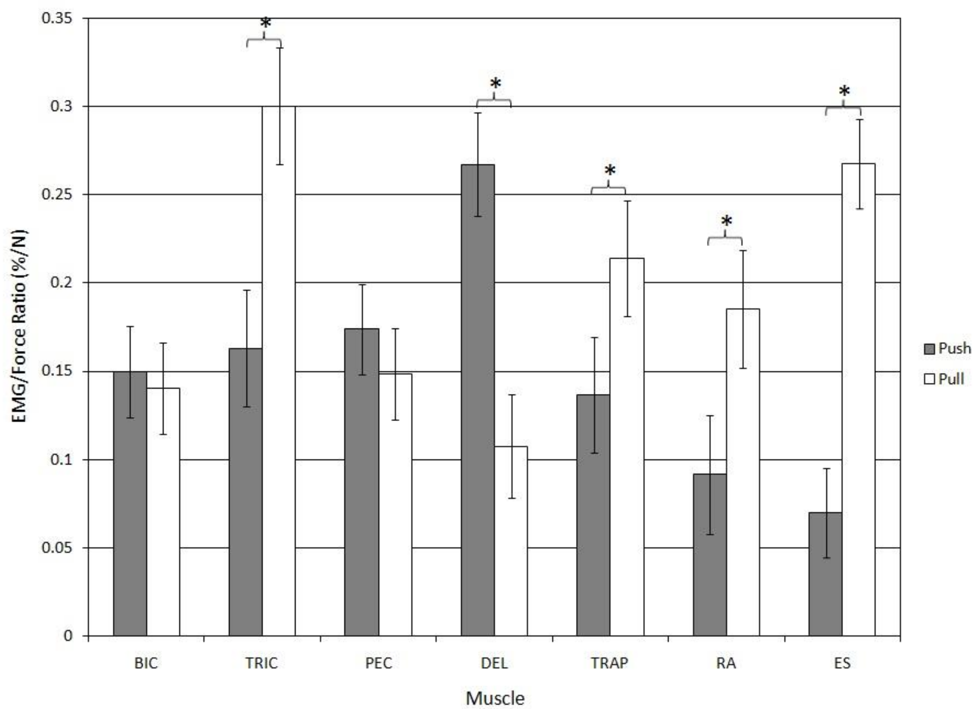

Figure 55: Effects of direction on individual muscle LSM EMG/force ratios (right side of the body) collapsed across all conditions. * indicates significant differences.

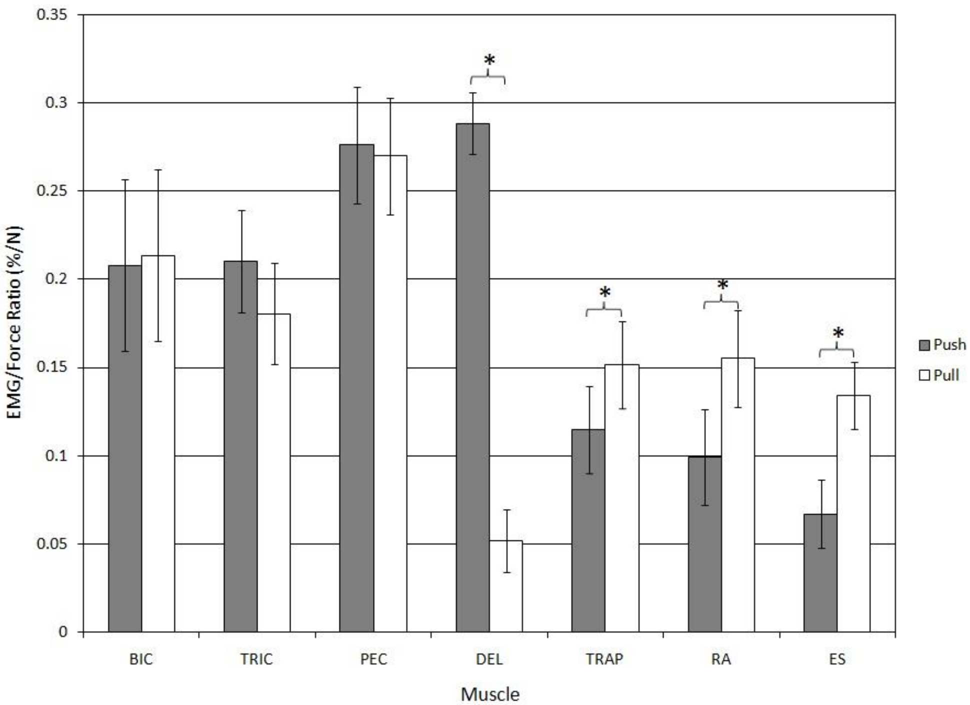

Figure 56: Effects of direction on individual muscle LSM EMG/force ratios (left side of the body) collapsed across all conditions. * indicates significant differences.

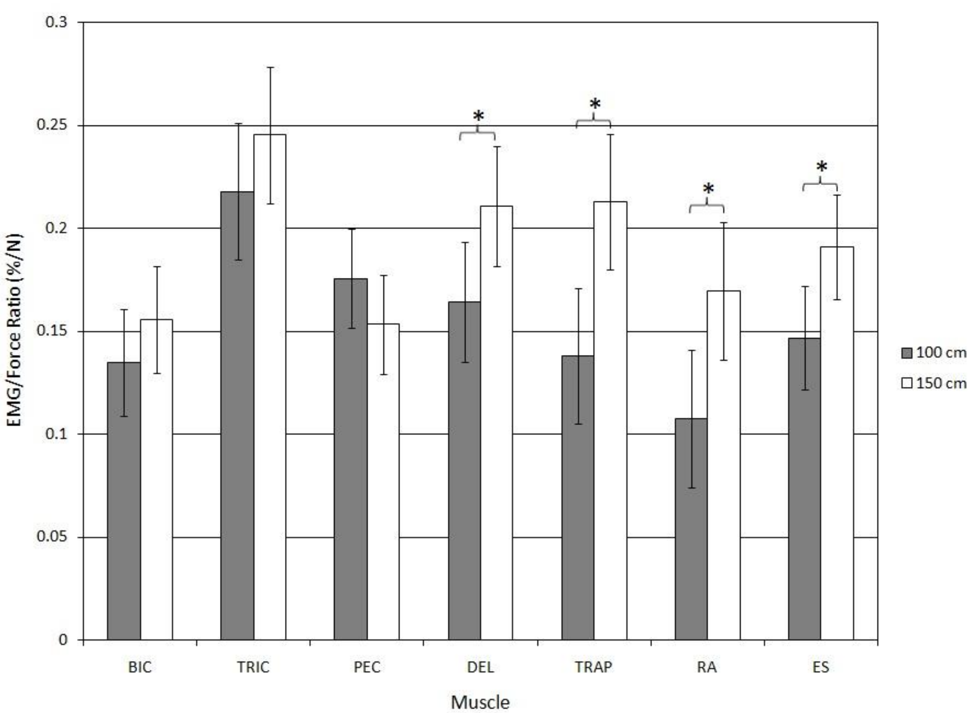

Figure 57: Effects of handle height on individual muscle LSM EMG/force ratios (right side of the body) collapsed across all conditions. * indicates significant differences.

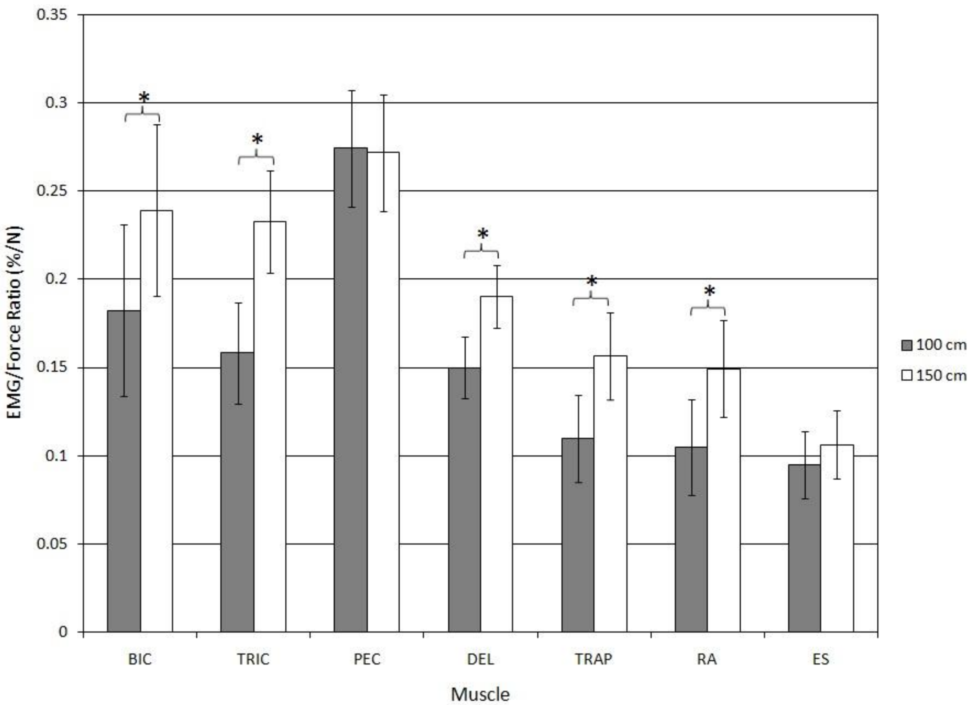

Figure 58: Effects of handle height on individual muscle LSM EMG/force ratios (left side of the body) collapsed across all conditions. * indicates significant differences.

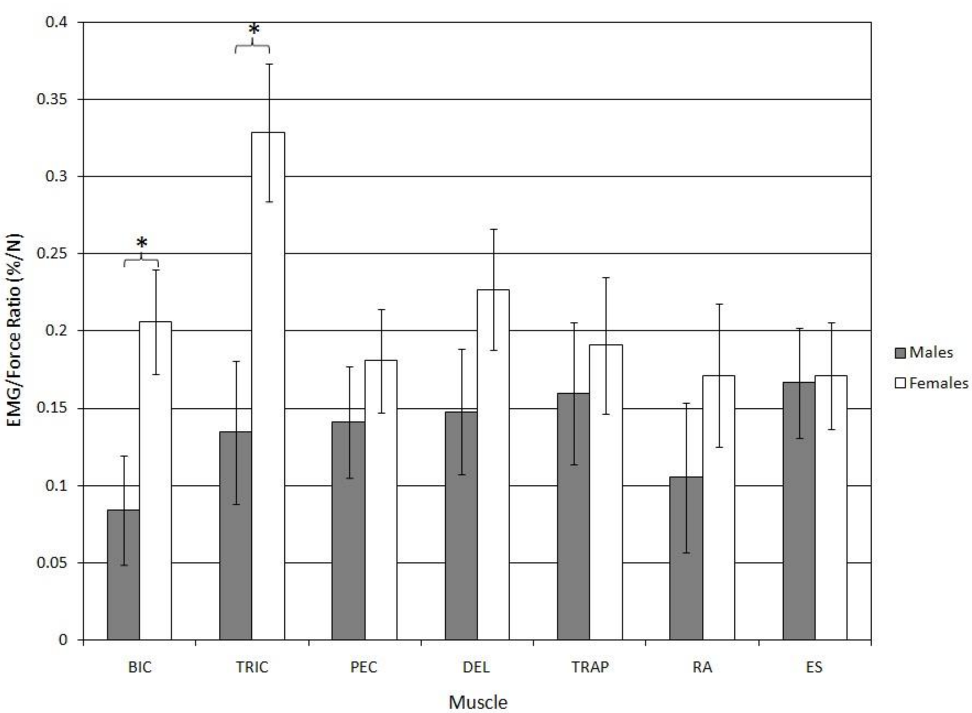

Figure 59: Effects of gender on individual muscle LSM EMG/force ratios (right side of the body) collapsed across all conditions. * indicates significant differences.

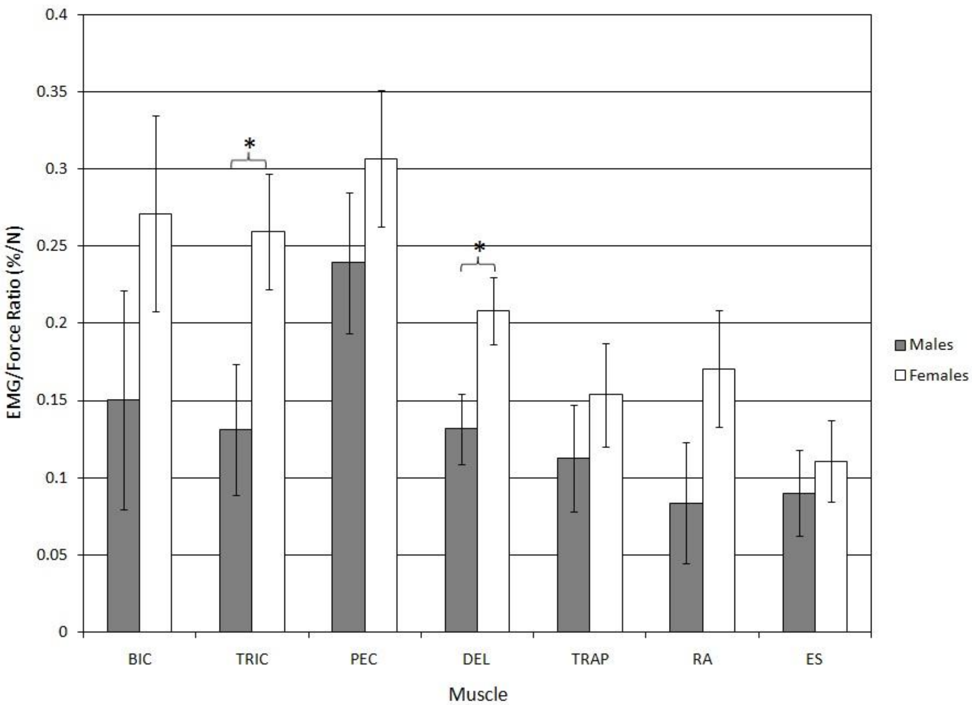

Figure 60: Effects of gender on individual muscle LSM EMG/force ratios (left side of the body) collapsed across all conditions. * indicates significant differences.

4.5.2 Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Normalized Individual EMG/force Ratios

A significant interaction effect existed between direction and height on most of the muscles examined (Appendix C). These included the pairs of right and left biceps brachii (Figures C1-C2), right and left pectoralis major (Figures C3-C4), right and left rectus abdominis (Figures C6-C7), right and left erector spinae (Figures C8-C9), and left middle trapezius (Figure C5) muscles. Further, a gender by direction effect was significant for the right and left deltoid, and left middle trapezius muscles. A mass by direction effect was also evident for the right triceps brachii, right and left pectoralis major, left deltoid, right middle trapezius, right rectus abdominis, and right and left erector spinae muscles. A gender by mass effect was only present for the left deltoid muscle.

Table 16: Results of ANOVA analysis for the effects of gender, mass, direction, handle height and handle orientation on normalized individual EMG/force ratios.

a) Right Muscles

Source of Variance	BIC F	TRIC F	PEC F	DEL F	TRAP F	RA F	ES F
Gender (G)	0.0253*	0.0398*	0.4348	0.1874	0.6360	0.3462	0.9297
Mass (M)	0.4431	0.7553	0.9859	0.4493	0.9779	0.6983	0.1036
Stature (S)	0.5094	0.6169	0.5025	0.1676	0.5709	0.1204	0.3692
Direction (D)	0.6162	0.0064*	0.1343	<.0001*	0.0002*	<.0001*	<.0001*
Height (H)	0.2750	0.5466	0.1927	0.0066*	0.0003*	<.0001*	0.0006*
Orientation (O)	0.3013	0.4263	0.0579	0.0309*	0.4175	0.4263	0.3455
G x M	0.8511	0.7448	0.4451	0.6204	0.0763	0.4317	0.4584
G x S	0.4150	0.2426	0.2222	0.4702	0.1932	0.6694	0.9025
G x D	0.5865	0.0492*	0.6172	0.0049*	0.3415	0.0682	0.6813
G x H	0.1801	0.9117	0.0730	0.2903	0.7215	0.2772	0.9132
G x O	0.4733	0.2949	0.3198	0.6409	0.4785	0.6468	0.8212
M x S	0.4376	0.2271	0.1601	0.0190*	0.0026*	0.0903	0.9785
M x D	0.1896	0.4952	0.1065	0.8157	0.0674	0.0837	0.0062*
M x H	0.7287	0.6638	0.6287	0.9154	0.3391	0.5766	0.3734
M x O	0.4720	0.5422	0.2915	0.9132	0.7700	0.8515	0.8606
S x D	0.9796	0.0695	0.8935	0.0699	0.0917	0.0042*	0.0163*
S x H	0.1969	0.2265	0.3185	0.0272*	0.8017	0.2083	0.7802
S x O	0.8330	0.5254	0.6726	0.1455	0.6988	0.9734	0.6610
D x H	0.0003*	<.0001*	<.0001*	0.2936	0.1963	0.0002*	<.0001*
D x O	0.3134	0.4263	0.0515	0.5420	0.6119	0.6213	0.4057
H x O	0.4840	0.5196	0.5956	0.9414	0.3425	0.8733	0.5409

continued

Table 16 continued

b) Left Muscles

Source of Variance	BIC F	TRIC F	PEC F	DEL F	TRAP F	RA F	ES F
Gender (G)	0.2283	0.0411*	0.3088	0.0288*	0.4122	0.1357	0.6012
Mass (M)	0.8137	0.6672	0.5581	0.0043*	0.8662	0.7760	0.2133
Stature (S)	0.9608	0.9746	0.4360	0.4715	0.9135	0.2237	0.8985
Direction (D)	0.8326	0.0536	0.7771	<.0001*	0.0071*	<.0001*	<.0001*
Height (H)	0.0351*	<.0001*	0.9128	0.0148*	0.0007*	<.0001*	0.1466
Orientation (O)	0.0912	0.8799	0.1727	0.8082	0.9208	0.5006	0.6252
G x M	0.8860	0.7529	0.9765	0.1178	0.8083	0.8940	0.3667
G x S	0.6954	0.4800	0.6397	0.3495	0.4018	0.4806	0.7323
G x D	0.9450	0.7728	0.5471	0.0050*	0.0106*	0.7265	0.3903
G x H	0.1717	0.3471	0.3044	0.4358	0.6573	0.3346	0.5371
G x O	0.9255	0.2395	0.8740	0.7585	0.7823	0.3620	0.7551
M x S	0.4476	0.2690	0.4786	0.2073	0.6384	0.8884	0.9450
M x D	0.1588	0.8580	0.0043*	0.0002*	0.3171	0.1455	0.1634
M x H	0.8678	0.9151	0.6919	0.1630	0.8595	0.5610	0.4652
M x O	0.1934	0.3994	0.2575	0.6638	0.1402	0.2302	0.5231
S x D	0.0844	0.0903	0.0704	0.0012*	0.0120*	0.0149*	0.0731
S x H	0.8864	0.5163	0.1193	0.8505	0.9866	0.3774	0.6628
S x O	0.2218	0.1686	0.2658	0.5649	0.3581	0.4124	0.8214
D x H	<.0001*	0.2633	<.0001*	0.1131	0.0415*	<.0001*	<.0001*
D x O	0.3627	0.2784	0.0099*	0.7767	0.5681	0.3014	0.8050
H x O	0.7904	0.9317	0.6731	0.8019	0.8172	0.4806	0.9817

* indicates statistically significant differences.

4.5.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Normalized Individual EMG/force Ratios

A main effect of direction was evident for EMG/force ratios for all individual muscles. The push direction resulted ratios that were 1.5, 2.1 and 1.8 times greater than the pull direction for the right and left biceps brachii, right and left pectoralis major, and right and left middle deltoid EMG/force ratios, respectively. The opposite was apparent for the right and left triceps brachii, right and left middle trapezius, right and left rectus abdominis, and right and left erector spinae EMG/force ratios (Figures 61-62). Elbow angle also had a main effect on EMG/force ratios, in which higher ratios were calculated for fully extended elbows compared to flexed elbows for the right triceps brachii, left pectoralis major and left middle deltoid muscles. Higher ratios for the flexed elbow condition arose for the right and left biceps brachii, and right and left middle trapezius muscles (Figures 63-64). It was further evident that females had higher ratios than males for the right biceps brachii, right triceps brachii and left deltoid muscles. Larger body masses were also associated with lower ratios, particularly for the right biceps brachii, triceps brachii, pectoralis major, middle deltoid, middle trapezius and erector spinae muscles, as well as the left middle deltoid.

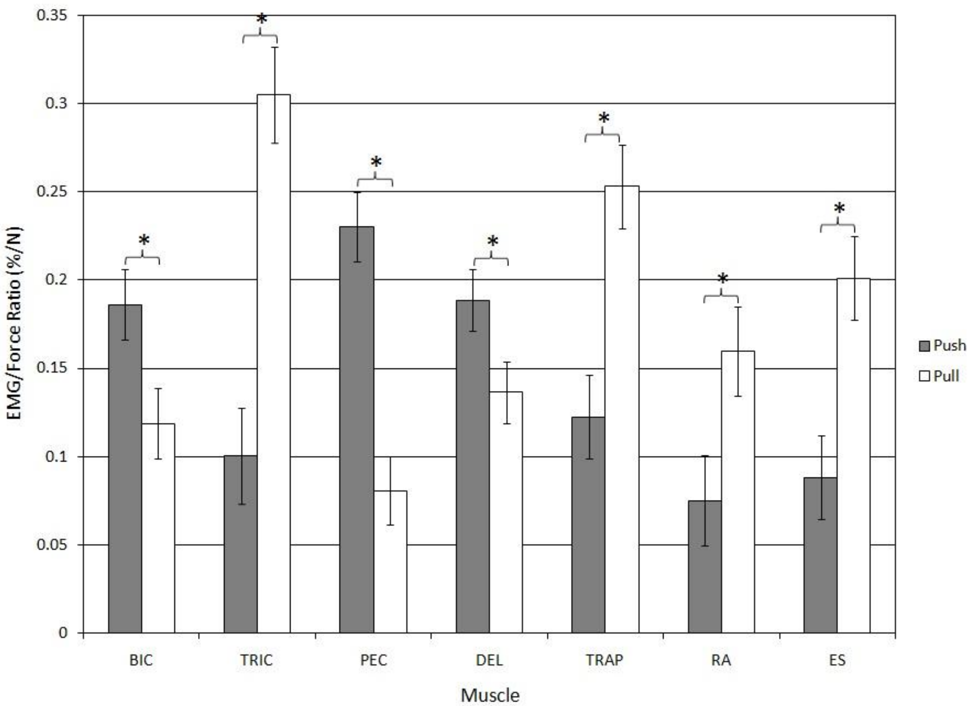

Figure 61: Effects of direction on individual muscle LSM EMG/force ratios for muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

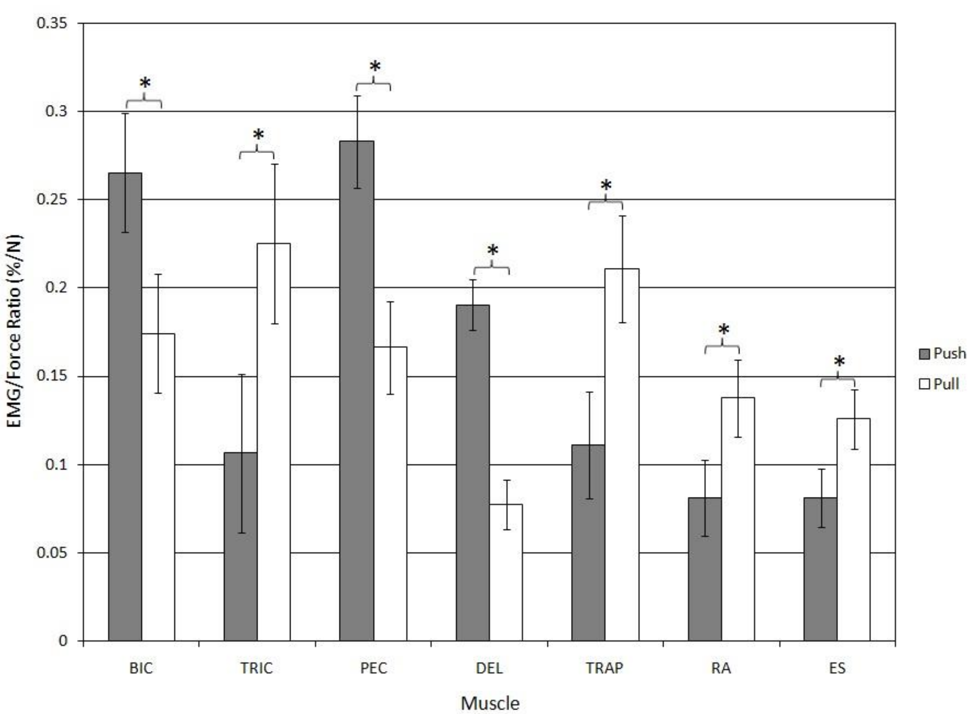

Figure 62: Effects of direction on individual muscle LSM EMG/force ratios for muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

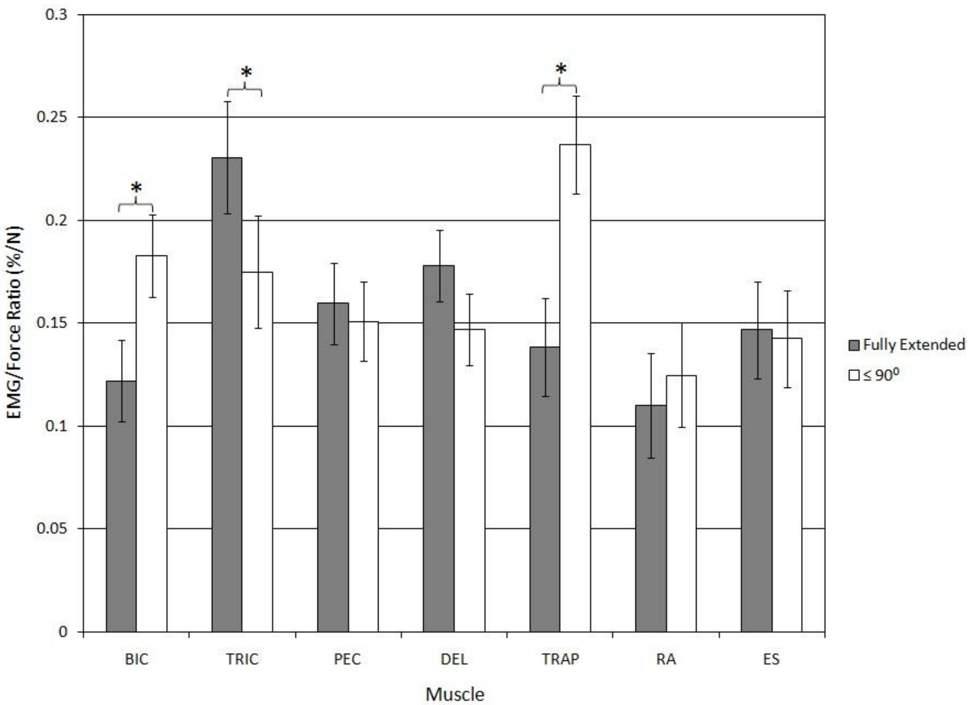

Figure 63: Effects of elbow angle on individual muscle LSM EMG/force ratios for muscles on the right side of the body collapsed across all conditions. * indicates significant differences.

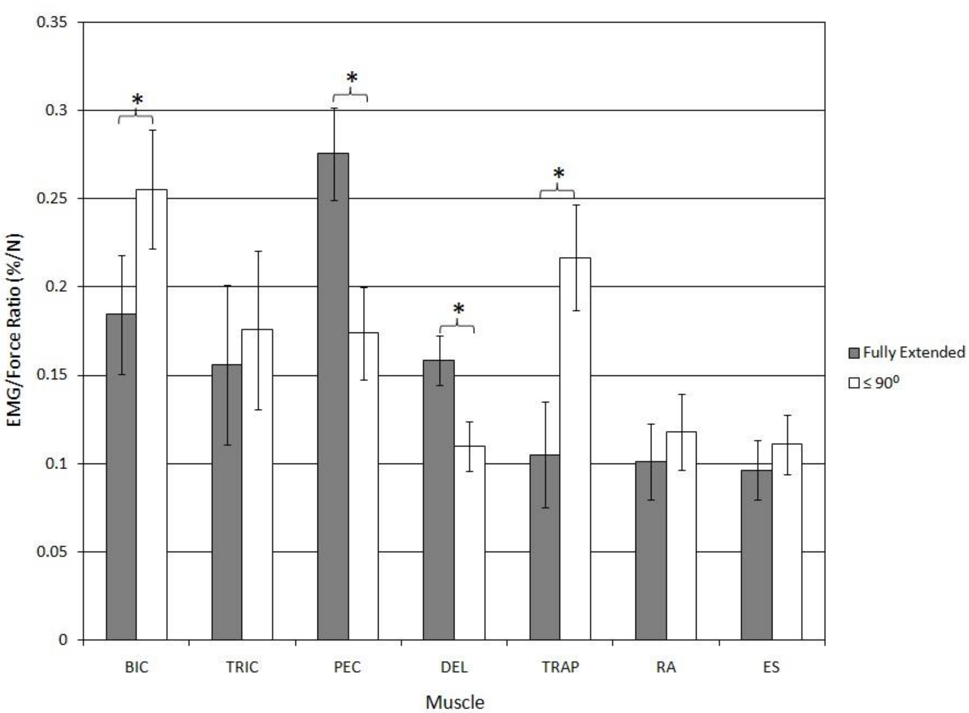

Figure 64: Effects of elbow angle on individual muscle LSM EMG/force ratios for muscles on the left side of the body collapsed across all conditions. * indicates significant differences.

4.5.4 Interaction Effects of Gender, Mass, Stature, Direction and Elbow Angle on Normalized Individual EMG/force Ratios

A significant direction by elbow angle effect was demonstrated by the many of the individual muscles (Appendix D). These included the pairs of right and left triceps brachii (Figures D1-D2), middle deltoid (Figures D3-D4), middle trapezius (Figures D5-D6) and rectus abdominis (Figures D7-D8) muscles. The left erector spinae muscle also exhibited a direction by elbow angle effect (Figure D9). In the case of the triceps brachii, middle trapezius, rectus abdmonis and erector spinae muscles, the highest and lowest ratios occurred when the elbows were flexed. For the middle deltoid muscles, the highest and lowest ratios occurred when the elbows were fully extended. There was also an interaction effect between gender and direction for the right and left deltoid, and left middle trapezius muscles.

Table 17: Results of ANOVA analysis for the effects of gender, mass, direction and elbow angle on normalized individual EMG/force ratios.

a) Right Muscles

Source of Variance	BIC F	TRIC F	PEC F	DEL F	TRAP F	RA F	ES F
Gender (G)	0.0467*	0.0236*	0.2120	0.3676	0.5857	0.2519	0.9886
Mass (M)	0.0437*	0.4883	0.5879	0.0200*	0.2933	0.5878	0.3126
Stature (S)	0.8233	0.3647	0.3506	0.0657	0.1495	0.1088	0.2770
Direction (D)	0.0060*	<.0001*	<.0001*	0.0152*	<.0001*	<.0001*	<.0001*
Elbow Angle (E)	0.0126*	0.0381*	0.7197	0.1383	<.0001*	0.2006	0.7270
G x M	0.2438	0.5830	0.5568	0.4507	0.2354	0.3492	0.5773
G x S	0.2661	0.2864	0.3633	0.8997	0.0501	0.7041	0.8655
G x D	0.6429	0.4798	0.2473	0.0081*	0.0256*	0.4396	0.3048
G x E	0.6492	0.3375	0.1754	0.1673	0.2450	0.5736	0.7692
M x S	0.3792	0.2743	0.0177*	0.6152	0.0044*	0.0816	0.8664
M x D	0.9432	0.5026	0.6676	0.7882	0.0498*	0.7880	0.6986
M x E	0.5398	0.7261	0.6946	0.6613	0.0105*	0.9305	0.6927
S x D	0.3688	0.2981	0.3158	0.0687	0.0071*	0.0508	0.0177*
S x E	0.7868	0.6472	0.8028	0.5517	0.8959	0.6669	0.8258
D x E	0.5322	0.0250*	0.3612	<.0001*	<.0001*	0.0021*	0.6408

continued

Table17 continued

b) Left Muscles

Source of Variance	BIC F	TRIC F	PEC F	DEL F	TRAP F	RA F	ES F
Gender (G)	0.1461	0.1609	0.2905	0.0016*	0.3712	0.1104	0.4585
Mass (M)	0.1227	0.9083	0.6411	0.1523	0.4686	0.6759	0.3924
Stature (S)	0.5138	0.8671	0.3930	0.8224	0.7786	0.1349	0.9358
Direction (D)	0.0069*	0.0049*	<.0001*	<.0001*	<.0001*	0.0003*	<.0001*
Elbow Angle (E)	0.0334*	0.6254	<.0001*	0.0070*	<.0001*	0.2577	0.0905
G x M	0.5058	0.6928	0.8081	0.8820	0.8440	0.9452	0.4033
G x S	0.6019	0.6468	0.5778	0.2527	0.9468	0.4356	0.4765
G x D	0.8194	0.6011	0.7746	0.0067*	0.0698	0.8571	0.6064
G x E	0.7931	0.9886	0.9981	0.2146	0.7597	0.8655	0.7614
M x S	0.8541	0.5262	0.6983	0.0174*	0.4448	0.8435	0.7631
M x D	0.4238	0.3424	0.5488	0.0753	0.5008	0.0704	0.5263
M x E	0.1724	0.3263	0.6035	0.2538	0.2258	0.3491	0.2949
S x D	0.5765	0.6320	0.4162	0.0192*	0.0788	0.1447	0.0188*
S x E	0.6005	0.8333	0.4670	0.3667	0.8529	0.5703	0.5015
D x E	0.5411	0.0006*	0.7454	<.0001*	<.0001*	0.0008*	0.0024*

* indicates statistically significant differences.

4.6 Total and Weighted Average EMG/force Ratios

Total and weighted average EMG/force ratios (Equations 15-17) represent an estimate of the mean %MVE associated with each unit of force, taking into account all muscles recorded. On average, total left EMG/force ratios ranged from 0.12 to 0.24, total right EMG/force ratios ranged from 0.14 to 0.27, while total EMG/force ratios values ranged from 0.07 to 0.27. Weighted average EMG/force ratios were comparable. Smaller ratios indicate that the exertion required less combined activation of the shoulder and trunk muscles per unit of force while larger ratios indicate that the exertion required more combined activation of the shoulder and trunk muscles per unit of force. Results of the GMSDHO and GMSDE models are summarized in Table 18 and Table 19, respectively. Total and weighted average EMG/force ratios were most affected by direction and height or direction and elbow angle interactions, based on ω^2 comparisons.

4.6.1 Main Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Total and Weighted Average EMG/force Ratios

There was a main effect of height on all total and weighted average EMG/force ratios (Figure 65). The 150 cm height resulted in higher EMG/force ratios than the 100 cm height for the left, right and total EMG/force ratios as well as for the weighted average left, right and total EMG/force ratios. The ratios for the 150 cm height were, on average, 24% higher than the ratios for the 100 cm height. A main effect of gender was evident for all total and weighted average EMG/force ratios such that EMG/force ratios were higher for females than males (Figure 66). On average, female left and total EMG/force ratios were 1.9 times higher than those of males. Likewise, female weighted average left and total EMG/force ratios were 2.0

times higher than those of male ratios. Mass had a main effect on left, right and total EMG/force ratios and weighted average right and total EMG/force ratios. Larger masses tended to have lower EMG/force ratios. Furthermore, there was a main effect of direction on the right and weighted average right and total EMG/force ratios where the pull direction was associated with higher EMG/force ratios (Figure 67). No significant effects of handle orientation on EMG/force ratios were found.

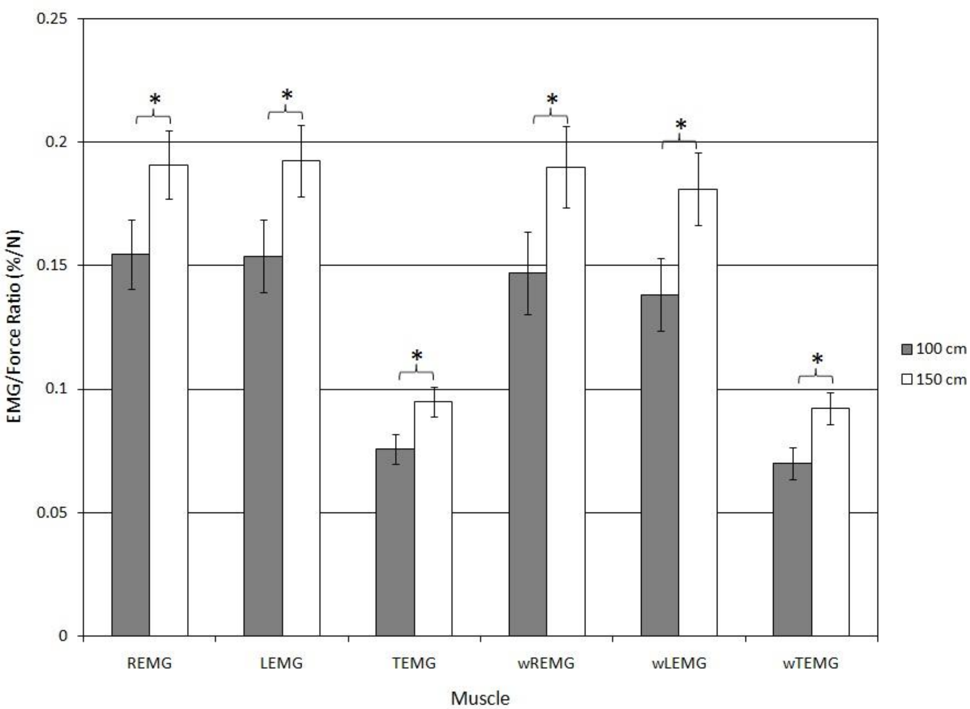

Figure 65: Effects of handle height on LSM total and weighted average total EMG/force ratios collapsed across all conditions. * indicates significant differences.

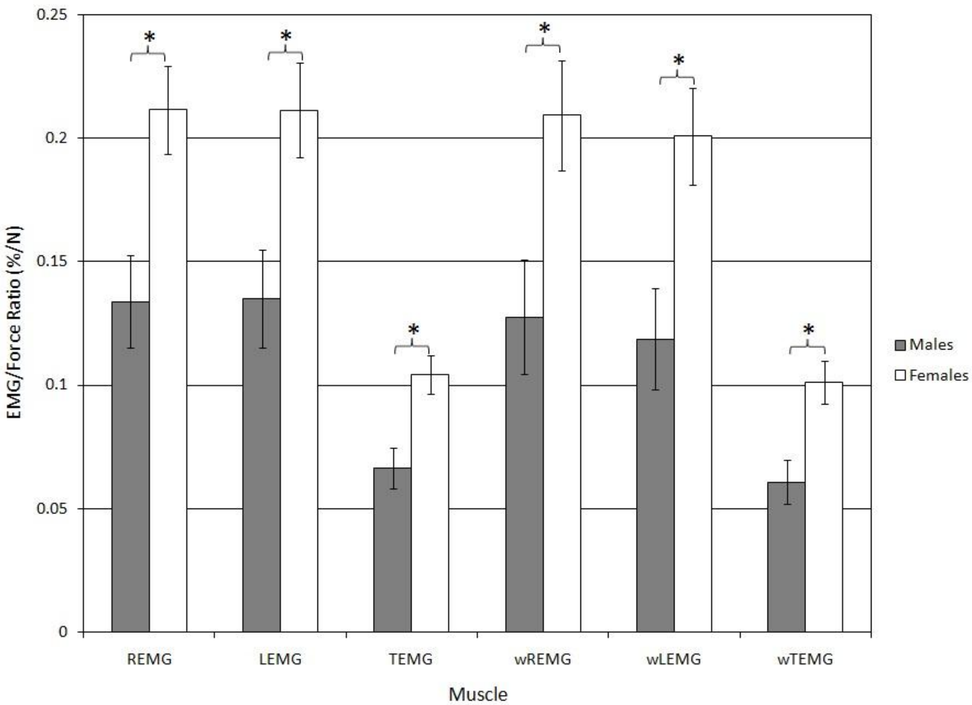

Figure 66: Effects of gender on LSM total and weighted average total EMG/force ratios collapsed across all conditions. * indicates significant differences.

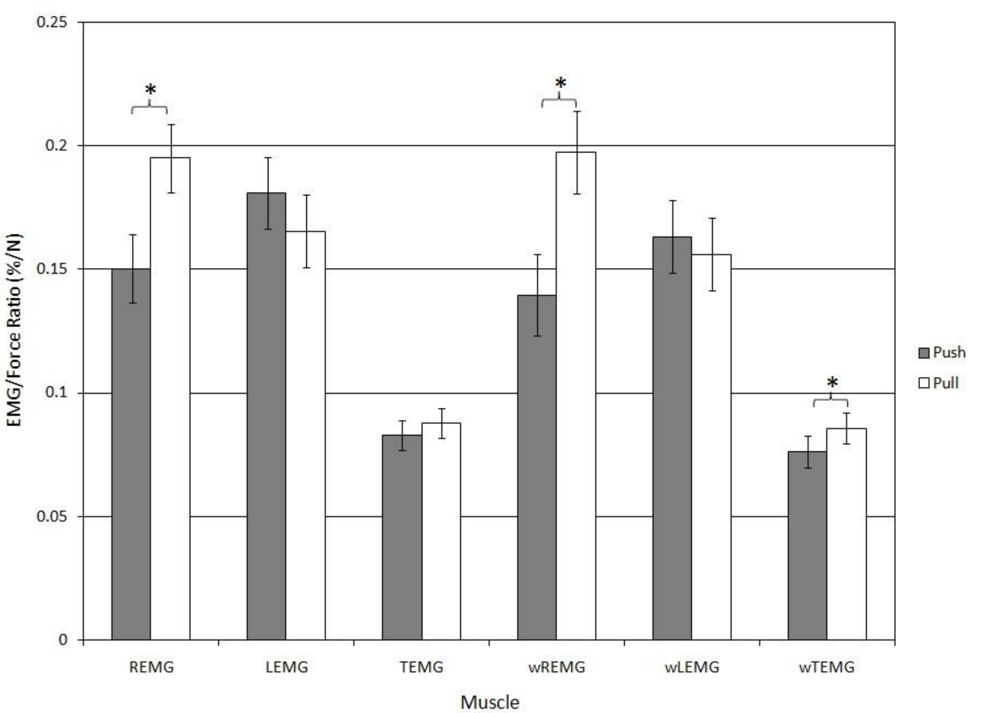

Figure 67: Effects of direction on LSM total and weighted average total EMG/force ratios collapsed across all conditions. * indicates significant differences.

4.6.2 Interaction Effects of Gender, Mass, Stature, Direction, Handle Height and Handle Orientation on Total and Weighted Average EMG/force Ratios

There was a significant interaction effect between direction and height on right, left and total (Figures 68-70) as well as weighted average right, left and total (Figures 71-73) EMG/force ratios. The pull direction combined with the 150 cm height always yielded the highest EMG/force ratios. A mass by direction effect was also evident for the right and total, and weighted average right and total EMG/force ratios.

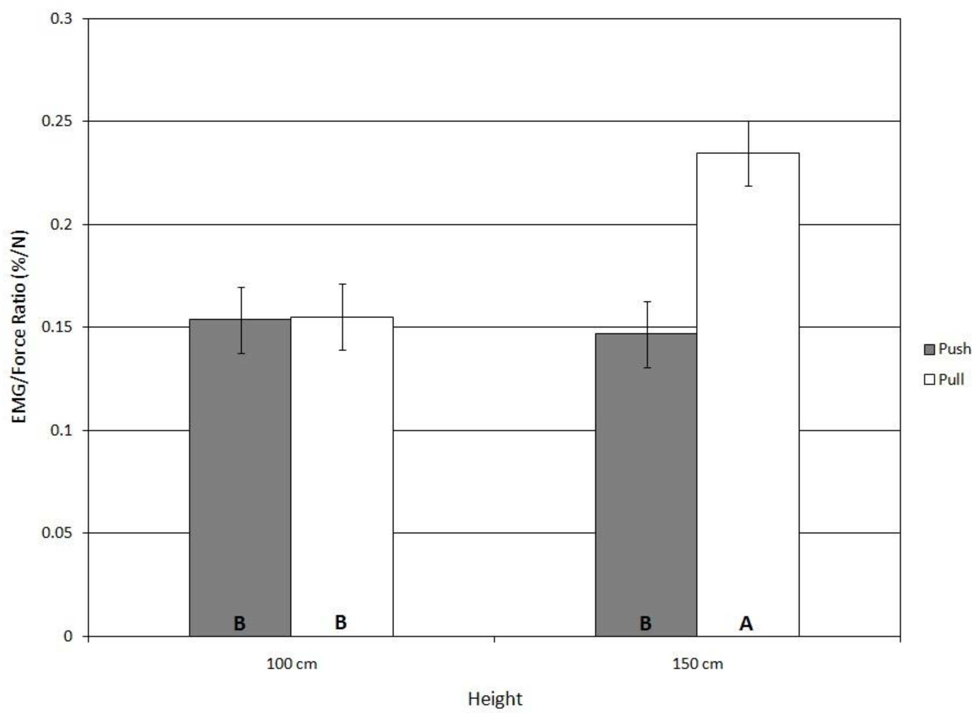

Figure 68: Effects of direction and handle height on LSM total right EMG/force ratios. Letters indicate significantly different direction by height interactions.

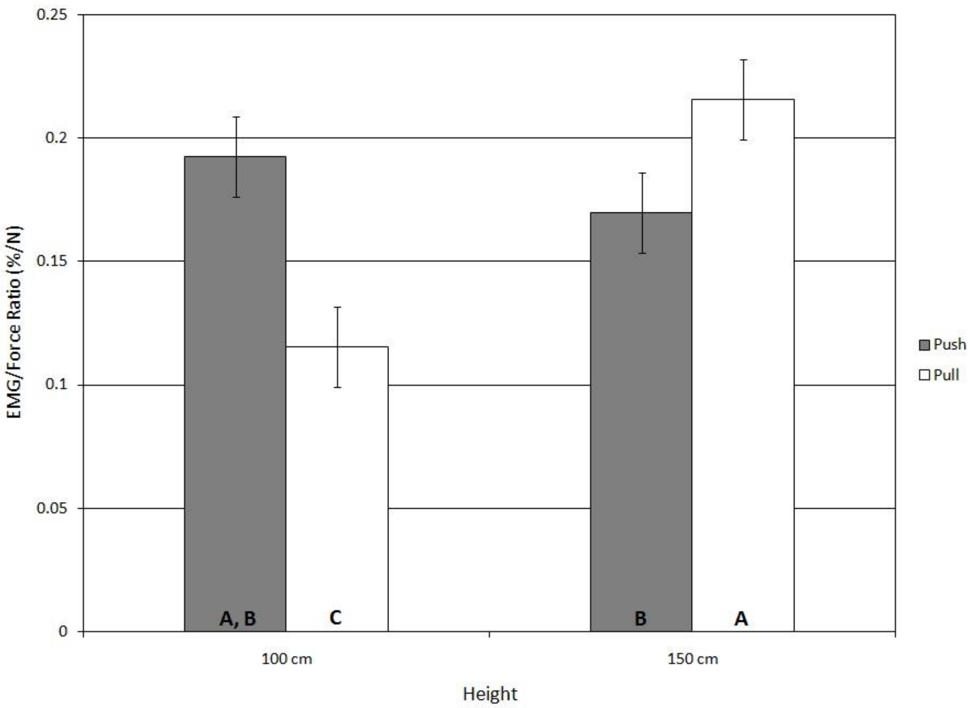

Figure 69: Effects of direction and handle height on LSM total left EMG/force ratios collapsed across all conditions. Letters indicate significantly different direction by height interactions.

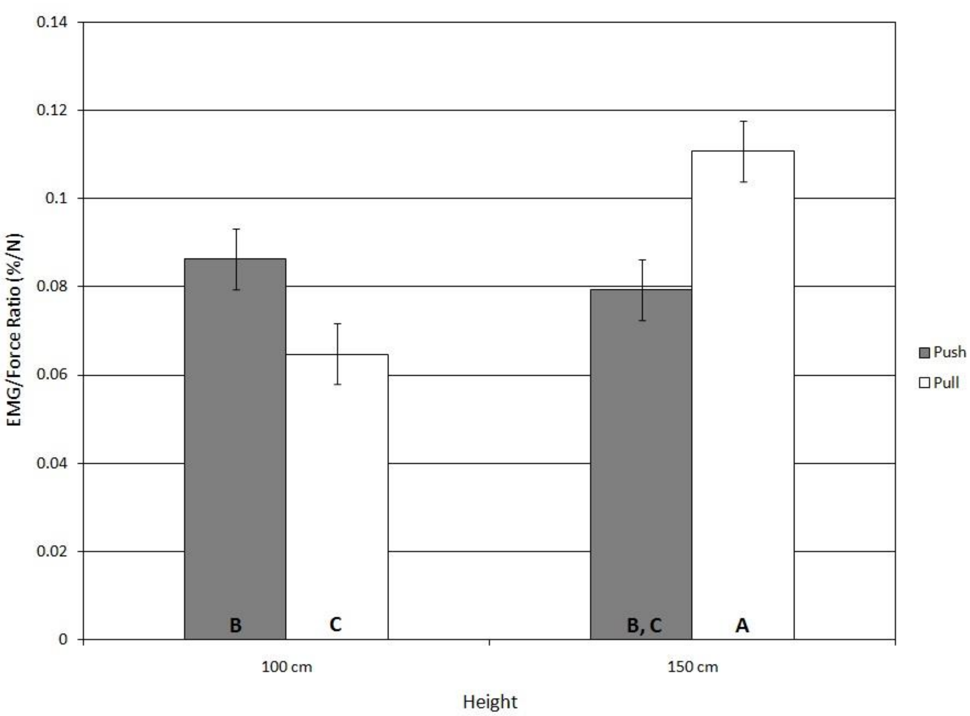

Figure 70: Effects of direction and handle height on LSM total EMG/force ratios collapsed across all conditions. Letters indicate significantly different direction by height interactions.

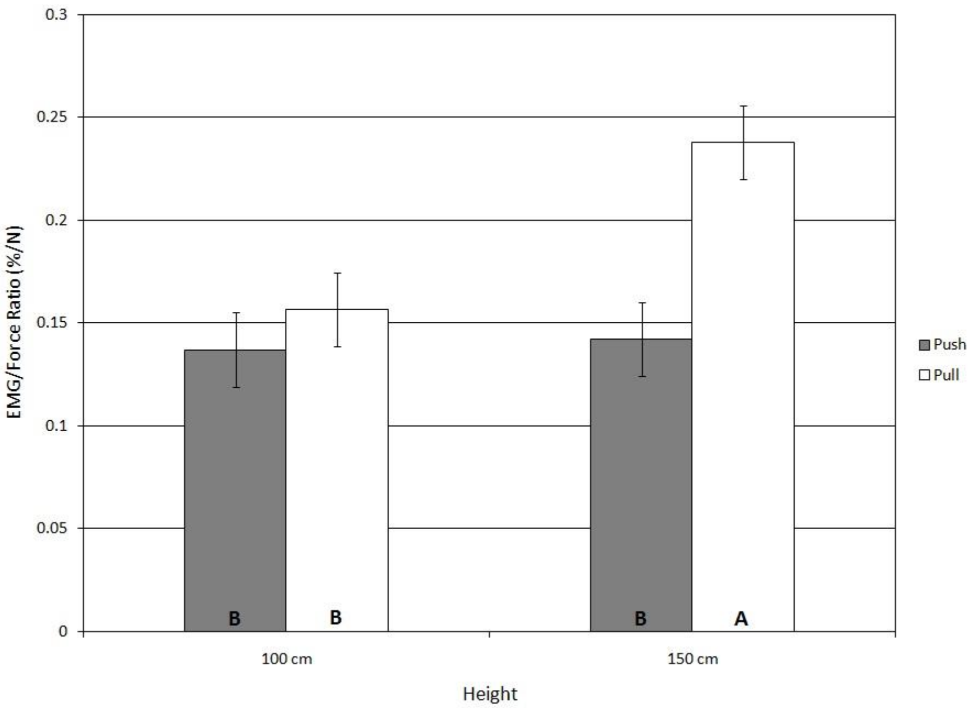

Figure 71: Effects of direction and handle height on LSM weighted average right EMG/force ratios. Letters indicate significantly different direction by height interactions.

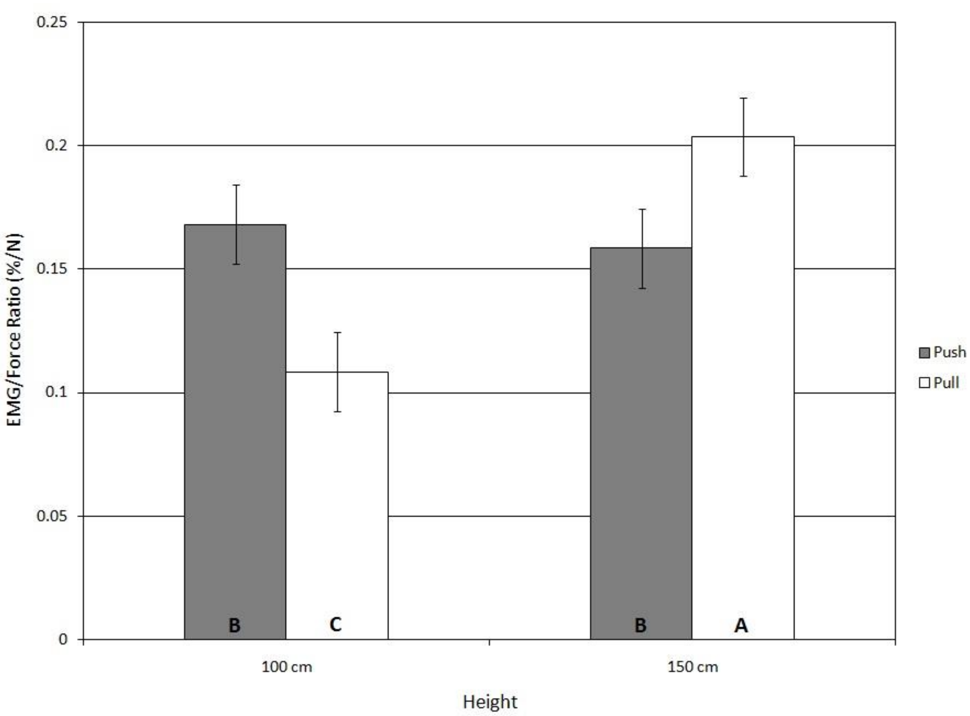

Figure 72: Effects of direction and handle height on LSM weighted average left EMG/force ratios. Letters indicate significantly different direction by height interactions.

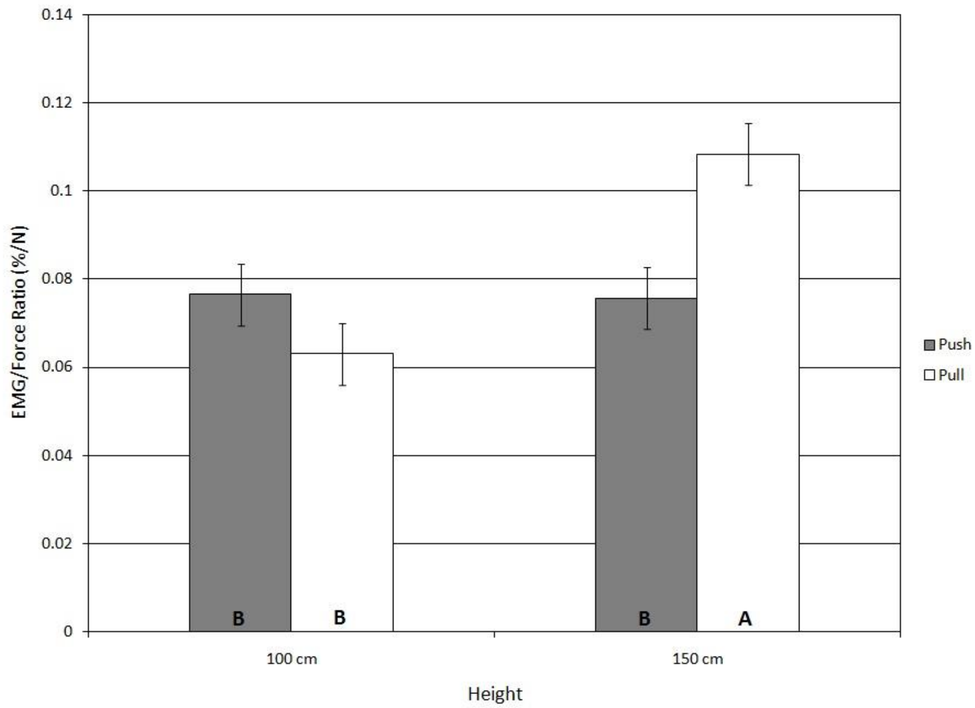

Figure 73: Effects of direction and handle height on LSM weighted average total EMG/force ratios. Letters indicate significantly different direction by height interactions.

Table 18: Results of ANOVA analysis for the effects of gender, mass, direction and handle height on total and weighted average EMG/force ratios.

	Total EMG			Weighted Average EMG		
Source of Variance	nREMG F	nLEMG F	nTEMG F	wREMG F	wLEMG F	wTEMG F
Gender (G)	0.0089*	0.0144*	0.0043*	0.0232*	0.0111*	0.0054*
Mass (M)	0.3073	0.2591	0.2191	0.5448	0.4270	0.4266
Stature (S)	0.0896	0.5534	0.2110	0.0629	0.3568	0.1051
Direction (D)	<.0001*	0.1389	0.2911	<.0001*	0.4200	0.0144*
Height (H)	0.0014*	0.0003*	<.0001*	<.0001*	<.0001*	<.0001*
Orientation (O)	0.4754	0.4417	0.8156	0.2439	0.7676	0.8076
G x M	0.2507	0.5601	0.8079	0.3697	0.5493	0.8511
G x S	0.0597	0.7856	0.4169	0.1717	0.7423	0.5581
G x D	0.3591	0.9664	0.5671	0.1849	0.6733	0.3061
G x H	0.5985	0.3992	0.4935	0.8949	0.3520	0.5840
G x O	0.6204	0.8853	0.8470	0.7337	0.9841	0.8264
M x S	0.0005*	0.9800	0.0443*	0.0029*	0.8852	0.0560
M x D	0.0197*	0.8445	0.1973	0.0157*	0.4831	0.3751
M x H	0.6954	0.6784	0.6341	0.6483	0.7957	0.6547
M x O	0.6784	0.1210	0.2556	0.8001	0.1152	0.2696
S x D	0.0418*	0.3378	0.1102	0.0080*	0.0523	0.0122*
S x H	0.9473	0.8628	0.8180	0.5561	0.7607	0.3677
S x O	0.5925	0.1476	0.3894	0.5359	0.1437	0.3861
D x H	0.0002*	<.0001*	<.0001*	0.0002*	<.0001*	<.0001*
D x O	0.9484	0.2636	0.3685	0.7911	0.5808	0.6408
H x O	0.4455	0.9254	0.7657	0.5441	0.8412	0.9809

* indicates statistically significant differences.

4.6.3 Main Effects of Gender, Mass, Stature, Direction and Elbow Angle on Total and Weighted Average EMG/force Ratios

There was a main effect of gender on left and total as well as weighted average left and total EMG/force ratios with higher ratios arising for females than males (Figure 74). On average, female left and total EMG/force ratios were 1.8 times higher than those of males. Likewise, female weighted average left and total EMG/force ratios were 1.9 times higher than those of male ratios. A main effect of mass was also apparent for right, left and total, and weighted average right and total EMG/force ratios. Larger masses were associated with lower EMG/force ratios. Finally, direction had a main effect on right and weighted average right and total EMG/force ratios such that the pull direction was an average of 1.2, 1.4 and 1.2 times higher EMG/force ratios than the push direction, respectively (Figure 75). There was no main effect of elbow angle on total or weighted average EMG/force ratios.

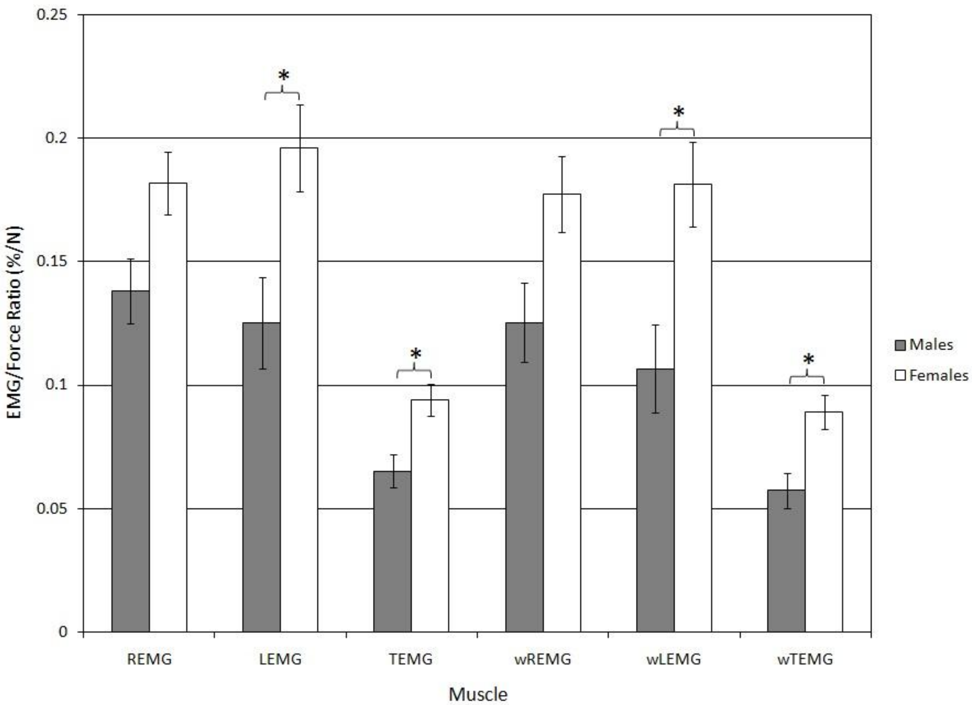

Figure 74: Effects of gender on LSM total and weighted average total EMG/force ratios collapsed across all conditions. * indicates significant differences.

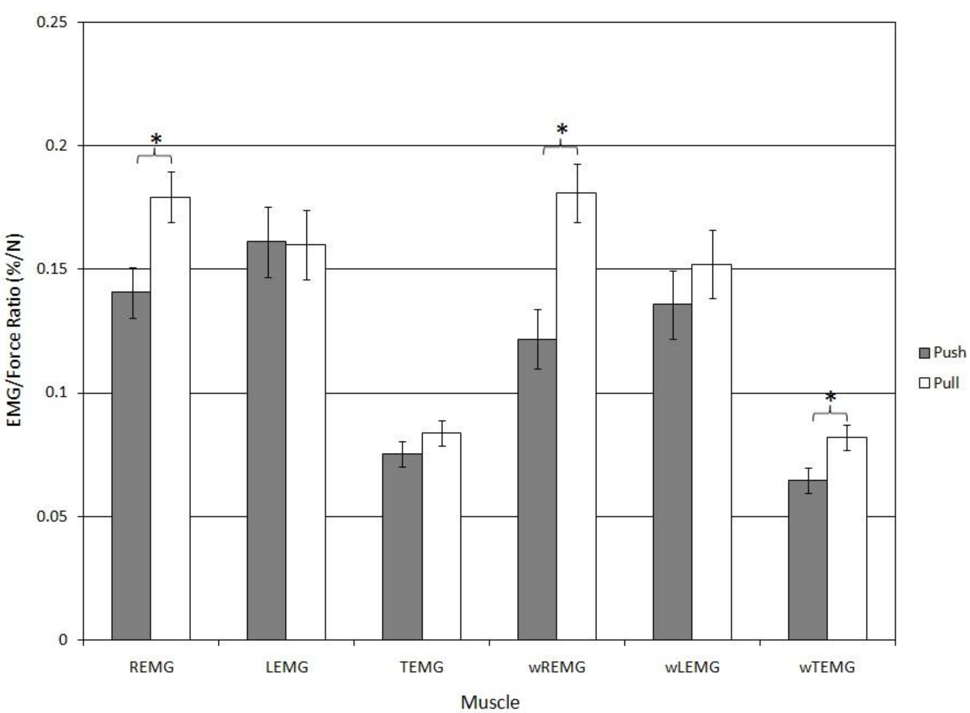

Figure 75: Effects of direction on LSM total and weighted average total EMG/force ratios collapsed across all conditions. * indicates significant differences.

4.6.4 Interaction Effects of Gender, Mass, Stature, Direction and Elbow Angle on Total and Weighted Average EMG/force Ratios

A marked interaction effect existed between direction and elbow angle on all total and weighted average EMG/force ratios. Together, the pull direction and flexed elbows consistently resulted in the highest EMG/force ratios while the push direction and flexed elbows generally resulted in the lowest EMG/force ratios. Figures 76-78 demonstrate the disordinal interaction effects between direction and elbow angle on total EMG/force ratios such that in changing postures from fully extended to flexed elbows, the EMG/force ratios increased for the pull direction while EMG/force ratios decreased for the push direction. Figures 79-81 show the interaction effects on weighted average EMG/force ratios. The interaction effect between mass and direction was also significant for the right and weighted average right EMG/force ratios.

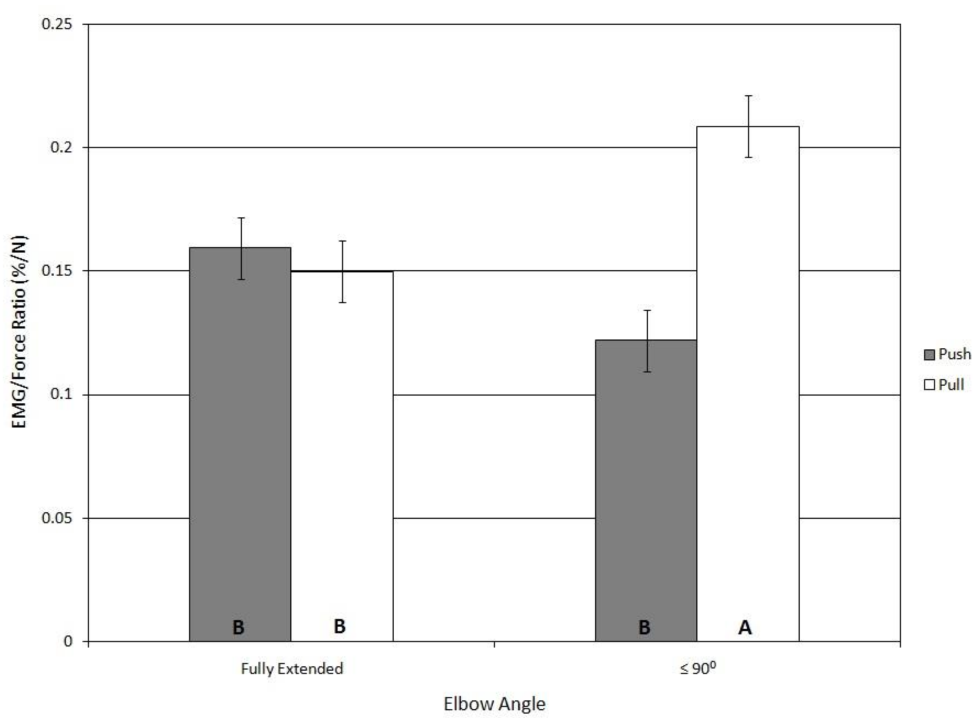

Figure 76: Effects of direction and elbow angle on LSM total right EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

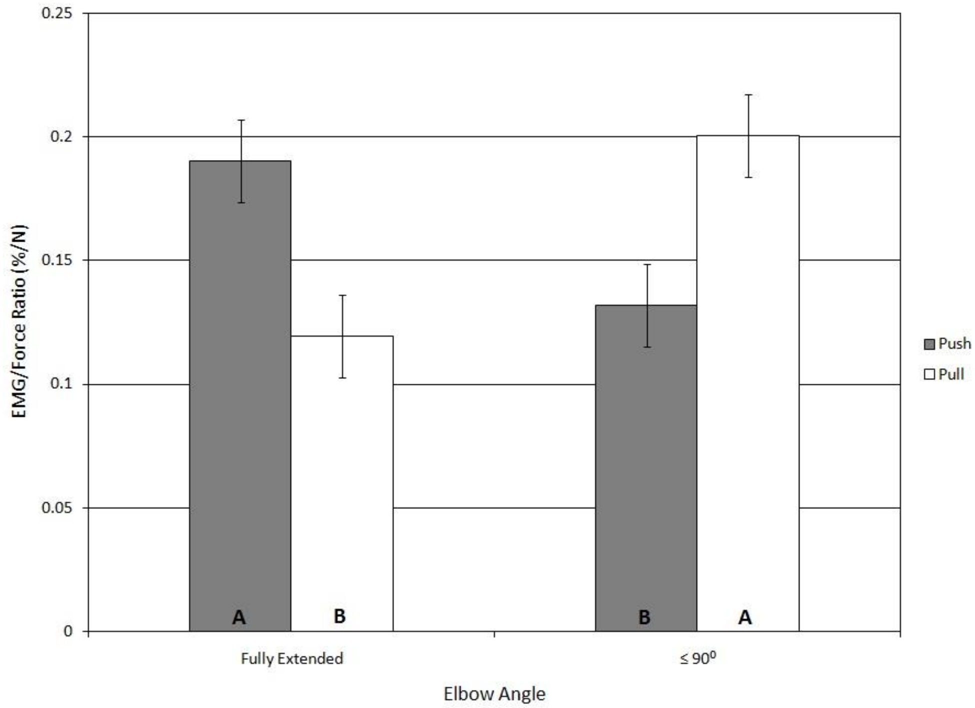

Figure 77: Effects of direction and elbow angle on LSM total left EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

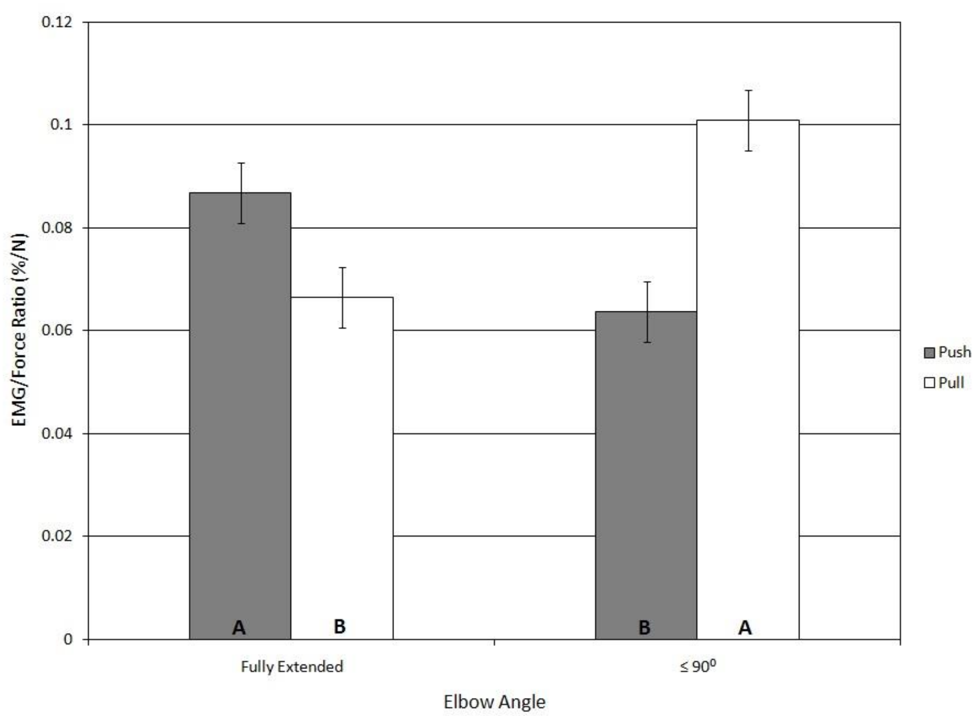

Figure 78: Effects of direction and elbow angle on LSM total EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

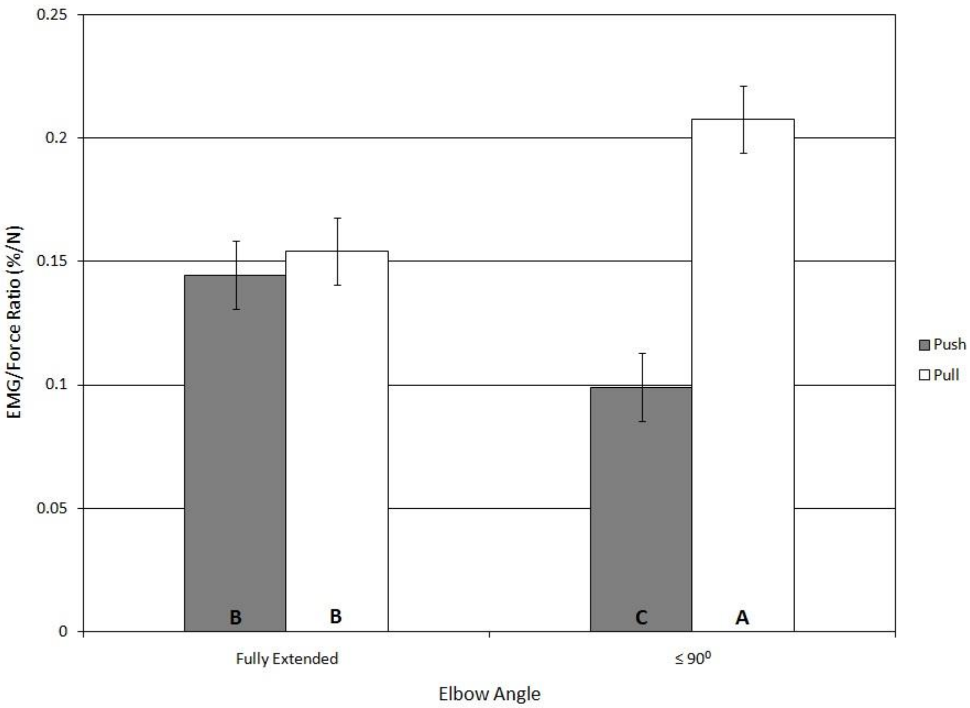

Figure 79: Effects of direction and elbow angle on LSM weighted average right EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

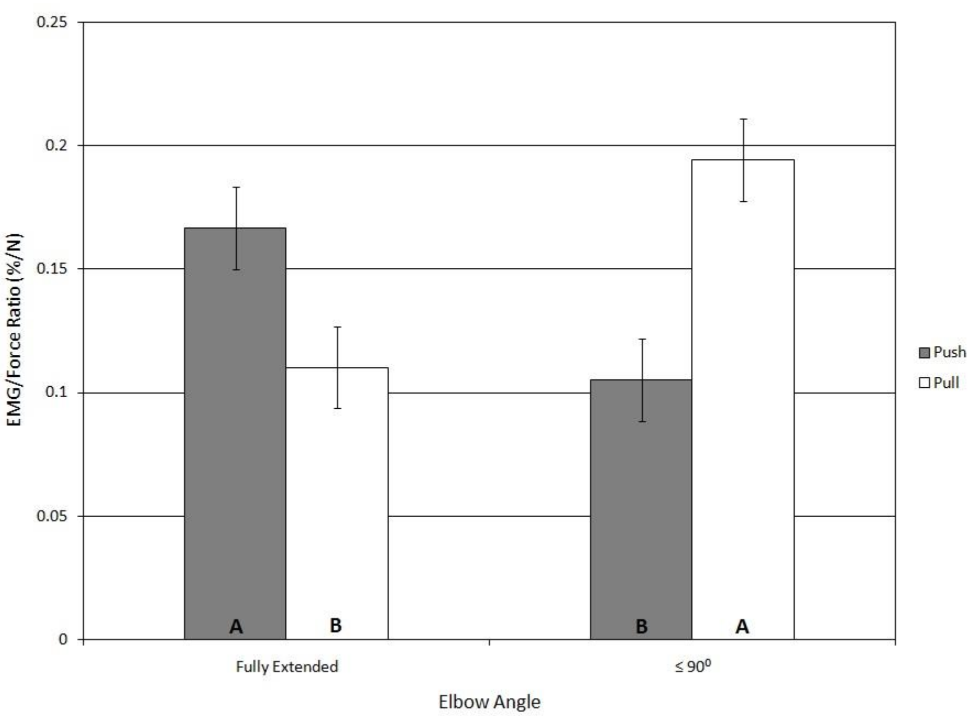

Figure 80: Effects of direction and elbow angle on LSM weighted average left EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

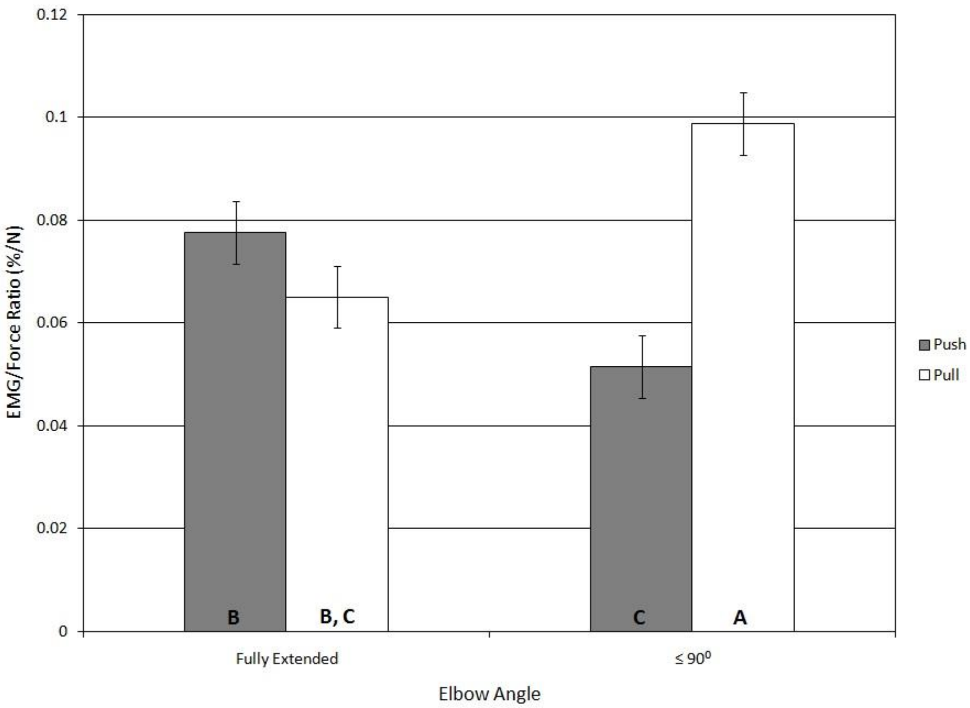

Figure 81: Effects of direction and elbow angle on LSM weighted average total EMG/force ratios. Letters indicate significantly different direction by elbow angle interactions.

Table 19: Results of ANOVA analysis for the effects of gender, mass, direction and elbow angle on total and weighted average EMG/force ratios.

	Total EMG			Weighted Average EMG		
Source of Variance	nREMG F	nLEMG F	nTEMG F	wREMG F	wLEMG F	wTEMG F
Gender (G)	0.0297*	0.0144*	0.0073*	0.0359*	0.0089*	0.0057*
Mass (M)	0.0452*	0.3032	0.0835	0.2781	0.5882	0.3131
Stature (S)	0.0259*	0.6366	0.1642	0.0259*	0.3704	0.0746
Direction (D)	0.0003*	0.9296	0.0577	<.0001*	0.2149	<.0001*
Elbow Angle (E)	0.2912	0.3780	0.1974	0.6803	0.3965	0.3633
G x M	0.8752	0.5462	0.7516	0.7389	0.5794	0.8867
G x S	0.0826	0.6505	0.2473	0.2312	0.8686	0.4231
G x D	0.3277	0.6491	0.8300	0.3213	0.8525	0.7027
G x E	0.5037	0.9464	0.6620	0.3365	0.8644	0.4975
M x S	0.0048*	0.4211	0.0568	0.0168*	0.4022	0.0684
M x D	0.4789	0.1632	0.6130	0.6088	0.0856	0.3552
M x E	0.4235	0.7359	0.7066	0.6878	0.4515	0.8774
S x D	0.0116*	0.1517	0.0151*	0.0068*	0.1135	0.0090*
S x E	0.9412	0.9070	0.9410	0.8459	0.7208	0.8624
D x E	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*	<.0001*

* indicates statistically significant differences.

4.7 Foot Stance

As participants were given the freedom to position their left foot in any position in which they felt would help enable them to exert their maximum push and pull forces, left foot positions could vary between and within participants across experimental trials. Foot stance ranged between 14.8 cm and 96.0 cm.

When considering gender, mass, stature, direction, height and handle orientation, height was found to have the greatest effect on foot stance in the anterior-posterior plane. Height had a main effect on foot stance with the 100 cm height leading to, on average, 7.7 cm wider foot stances than the 150 cm height ($p < 0.0001$). Taller statures were associated with wider foot stances ($p = 0.0020$). Direction also had a main effect on foot stance, in which participants selected, on average, 4.2 cm wider foot stances for the push direction compared to the pull direction ($p = 0.0021$). Participants further tended to adopt a rearward left foot position for push trials and a forward left foot position for pull trials, with respect to the fixed right foot position. When accounting for gender, mass, stature, direction and elbow angle, direction was shown to have the greatest effect on foot stance. Both the push direction ($p < 0.0001$) and taller statures ($p = 0.0031$) lead to wider foot stances. Participants also elected a wider foot stance for the extended elbow conditions ($p = 0.0042$). Furthermore, a pronounced direction by elbow angle effect was evident, in which the push direction interactions resulted in wider foot stances than the pull direction ($p = 0.0032$). Both the widest and narrowest foot stances chosen were those for the flexed elbow conditions.

4.8 Left and Right Asymmetry

As left and right hand forces were measured independently using two force transducers, left and right hand forces could be compared and contributions of each hand force to the total two-handed exertion could be determined. In addition, the muscles measured on the right side of the body were also recorded for the left side of the body. This allowed comparisons between left and right muscles and contributions toward the total left and total right mean %MVE estimates.

4.8.1 Asymmetry between Hand Force Magnitudes

According to the GMSDHOA models, differences between right and left hand force magnitudes across conditions were determined to be insignificant ($p = 0.1992$). However, interaction effects between direction and hand were statistically significant ($p < 0.0001$). Post hoc analyses revealed that hand forces generated by the push direction were consistently and significantly greater than hand forces generated by the pull direction. Within the push direction, right hand forces tended to be greater than left hand forces, but these differences were not significant. Within the pull direction, left hand forces were significantly greater than right hand forces. Figure 82 illustrates the interaction effect. In accounting for the GMSDEA models, differences between right and left hand force magnitudes were also found to be insignificant across conditions ($p = 0.9684$). Again, a direction by hand interaction effect was present ($p = 0.0254$), where differences between directions with respect to force capability were larger for the right hand than the left hand (Figures 83).

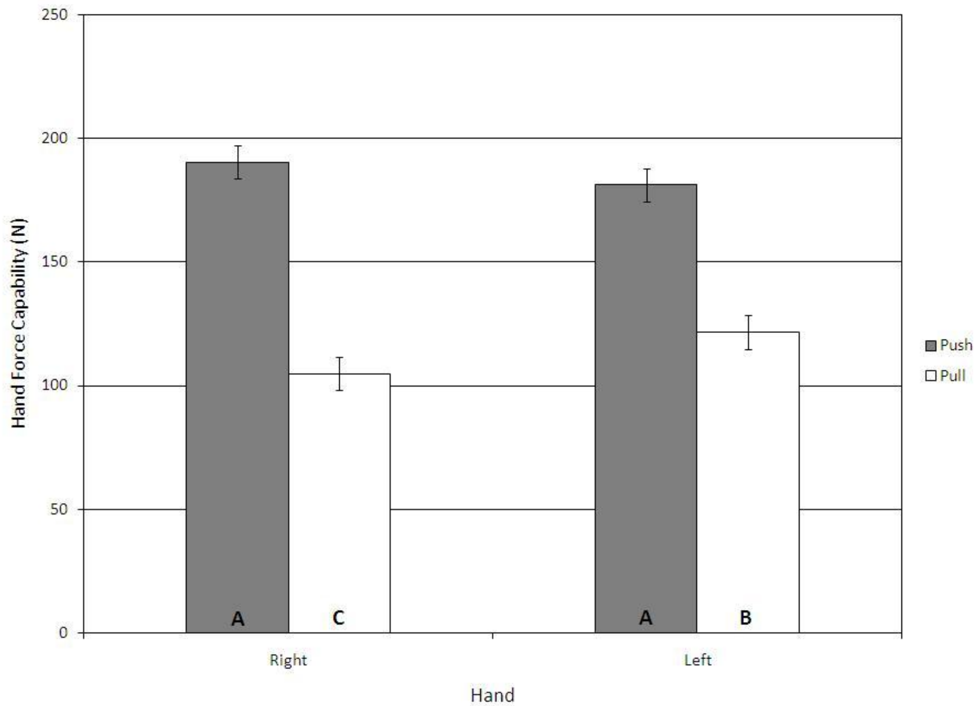

Figure 82: Effects of direction and hand on LSM hand force capability based on the GMSDHOA model. Letters indicate significantly different direction by hand interactions.

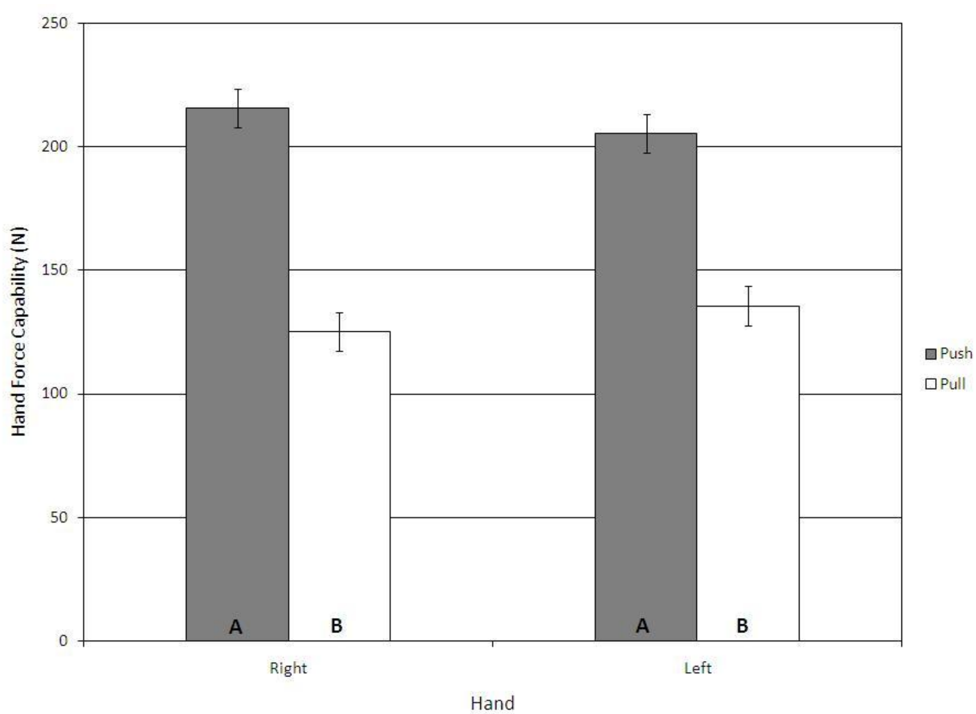

Figure 83: Effects of direction and hand on LSM hand force capability based on the GMSDEA model. Letters indicate significantly different direction by hand interactions.

4.8.2 Asymmetry between Individual EMG, Total EMG and Weighted Average Total EMG

The seven muscles measured from each side of the body were compared to determine any significant differences in mean %MVE. Total and weighted average total EMG from each side of the body were also compared. Based on the GMSDHOA models, hand had a main effect on mean %MVE for all individual muscles except for the triceps brachii (Figure 84). The mean %MVE of the right sided muscles were significantly greater than mean %MVE of the left sided muscles for the middle deltoid ($p = 0.0065$), middle trapezius ($p < 0.0001$), rectus abdominis ($p = 0.0038$) and erector spinae muscles ($p < 0.0001$). The reverse trend occurred for the biceps brachii and pectoralis major muscles ($p < 0.0001$). Weighted average total mean %MVE was also significant between sides and was greater for the right sided muscles than the left ($p = 0.0070$). Upon determining the effects using the GMSDEA models, significant differences were found for the biceps brachii, pectoralis major and erector spinae (Figure 85). Again, the mean %MVE of the right erector spinae was greater than the mean %MVE of the left erector spinae ($p < 0.0001$). The opposite trend was apparent for the biceps brachii ($p = 0.0093$) and pectoralis major ($p = 0.0158$) muscles.

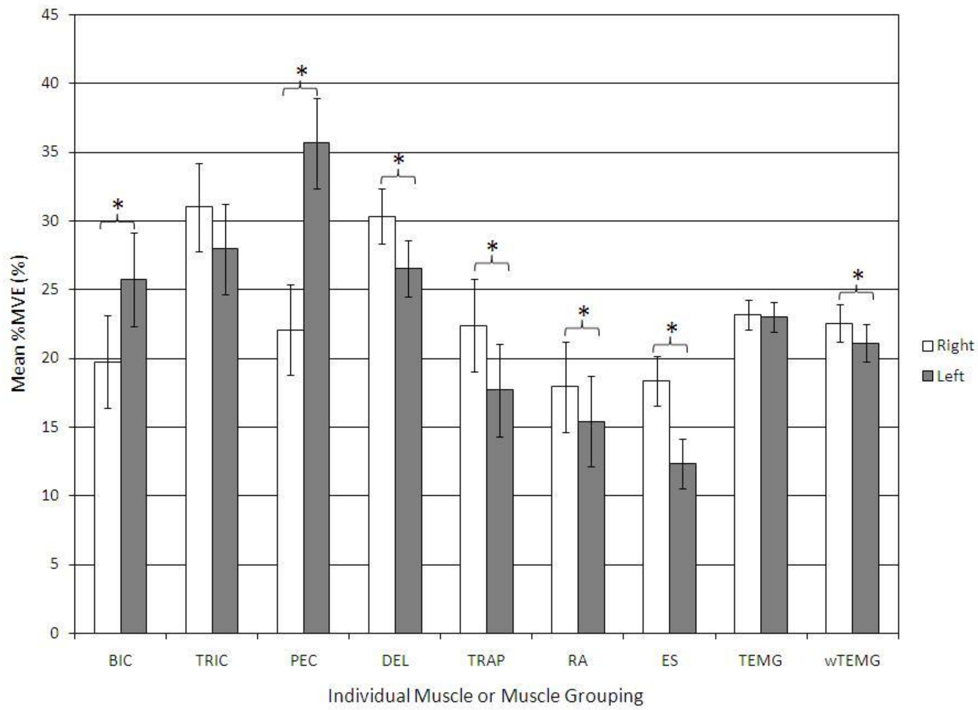

Figure 84: Effects of hand on individual muscle LSM %MVE, total EMG and weighted average total EMG based on the GMSDHOA model. * indicates significant differences.

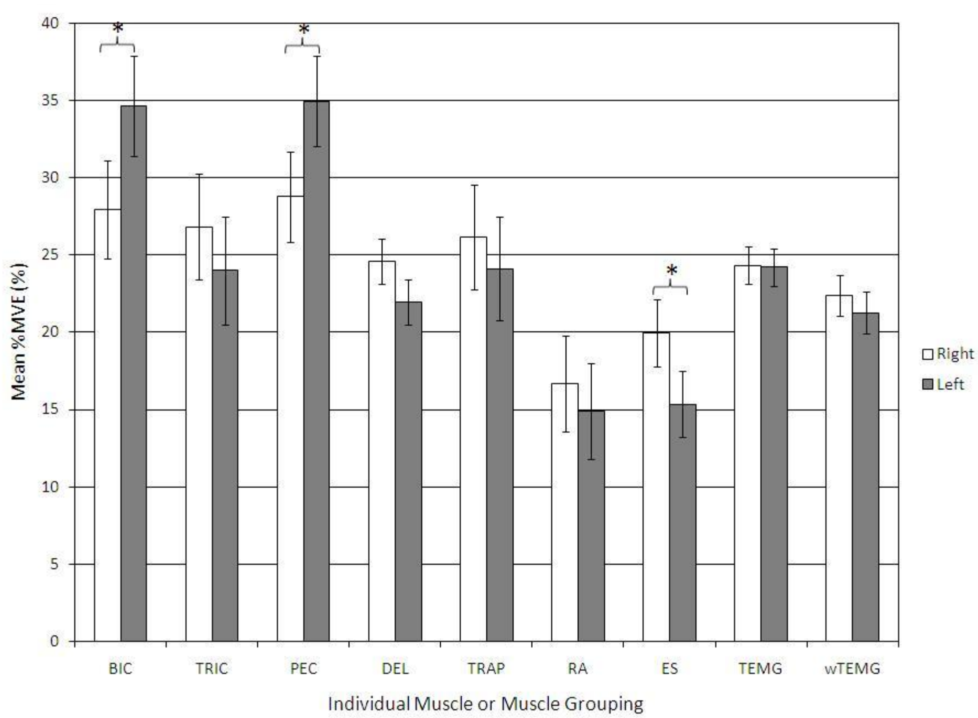

Figure 85: Effects of hand on individual muscle LSM %MVE, total EMG and weighted average total EMG based on the GMSDEA model. * indicates significant differences.

4.8.3 Asymmetry between EMG/force Ratios

The computed EMG/force ratios of the seven muscles measured from each side of the body as well as total and weighted average total EMG/force ratios for each side were compared to determine any significant differences in mean %MVE. In determining the effects with the GMSDHOA models, differences between right and left sides of the body were significant for all EMG/force ratios, except for the total EMG/force ratio (Figure 86). Trends were similar as those reported for mean %MVE such that higher ratios were determined for the left biceps brachii ($p = 0.0066$) and pectoralis major ($p < 0.0001$) muscles, while higher ratios were found for the right side of the triceps brachii ($p < 0.0001$), middle deltoid ($p = 0.0065$), middle trapezius ($p < 0.0001$), rectus abdominis ($p = 0.0043$), and erector spinae ($p < 0.0001$) muscles as well as for weighted average total mean %MVE ($p = 0.0015$). The GMSDEA models disclosed significant differences in right and left EMG/force ratios for the biceps brachii, pectoralis major and erector spinae muscles (Figure 87). Higher ratios were found for the left biceps brachii ($p = 0.0154$) and pectoralis major ($p = 0.0085$) muscles compared to the contralateral side while the opposite was evident for the erector spinae muscle ($p < 0.0001$).

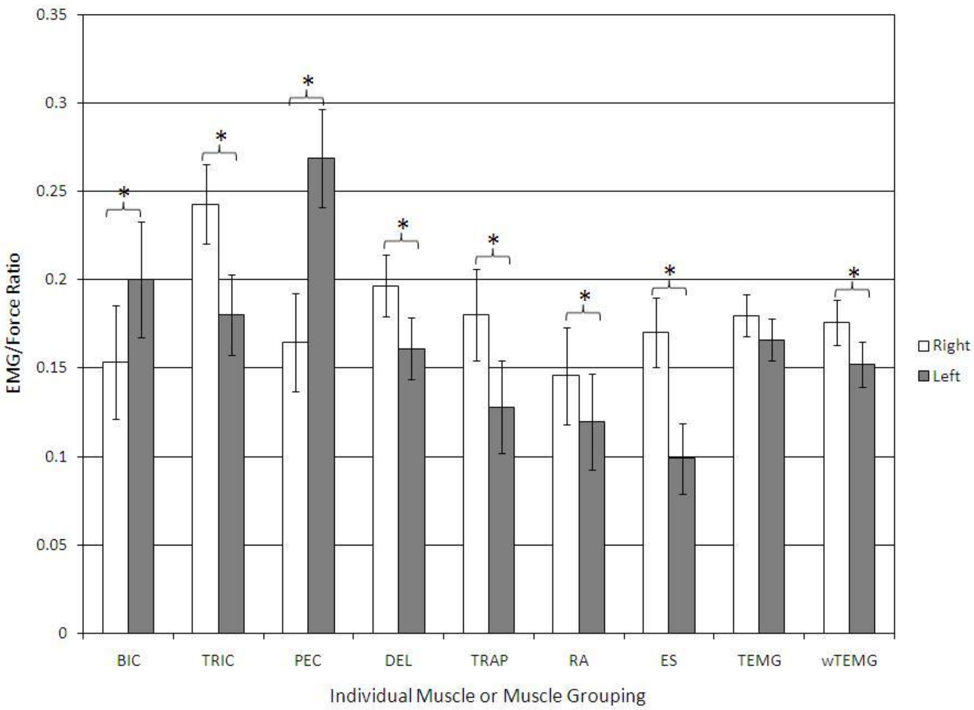

Figure 86: Effects of hand on individual muscle LSM EMG/force ratios, total EMG/force ratios and weighted average total EMG/force ratios based on the GMSDHOA model. * indicates significant differences.

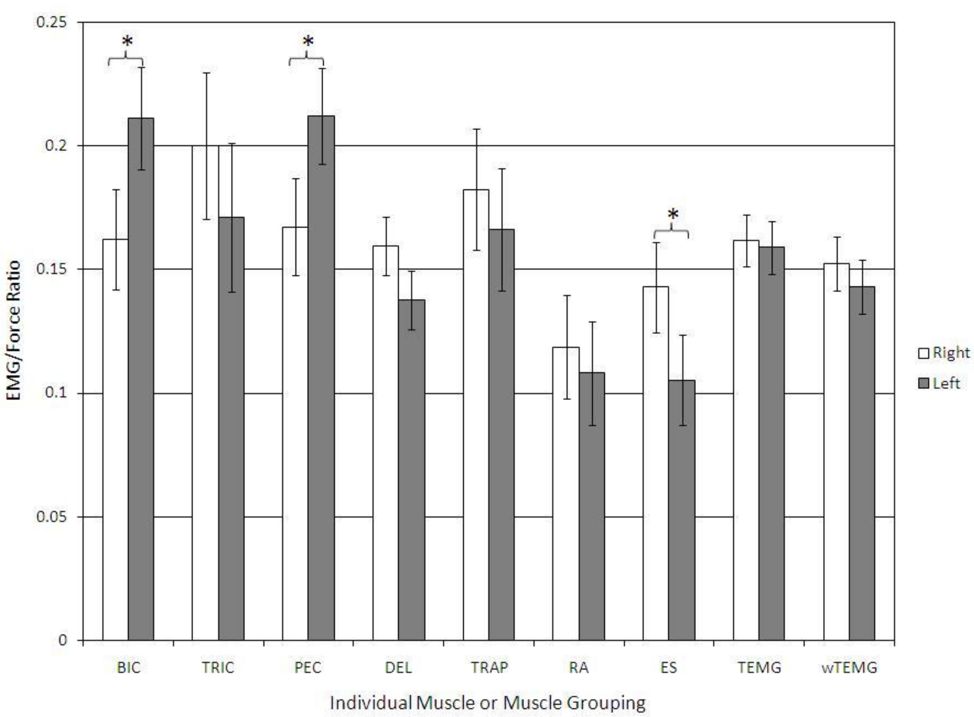

Figure 87: Effects of hand on individual muscle LSM EMG/force ratios, total EMG/force ratios and weighted average total EMG/force ratios based on the GMSDEA model. * indicates significant differences.

5.0 Discussion

The purpose of this study was to quantitatively evaluate the effects of gender, mass, stature, direction, handle height, handle orientation and elbow angle on force-, moment- and EMG-based exposure estimates during two-handed maximal voluntary isometric contractions. Investigating maximal hand forces is important because they are often referenced when setting force limits in the workplace. Consideration should be given when interpreting the reported values as although these are maximal hand forces under the experimental conditions, this does not imply that the participating muscles are maximally activated. It should also be noted that larger muscles may contribute more to these forces than smaller muscles. Furthermore, although only 14 muscles were monitored, it is likely that the force outputs are the result of a synergy of muscles from the entire body, among other tissues.

5.1 Addressing the Hypotheses

Hypothesis 1

It was hypothesized that there would be statistically significant differences in hand force magnitudes, shoulder and low back moments, normalized EMG values and EMG/force ratios between males and females. Males had significantly greater hand force magnitudes than females. Gender only had a main effect on two or three normalized individual EMG and individual EMG/force ratios from each set of ANOVAs. However, gender differences were generally significant for mean %MVE and EMG/force ratios based on total left and total EMG as well as weighted average left and total EMG parameters. In all cases, females had greater mean %MVE magnitudes and EMG/force ratios than males. (Partially supported)

Hypothesis 2

It was hypothesized that there would be statistically significant differences in hand force magnitudes, shoulder and low back moments, normalized EMG values and EMG/force ratios obtained between push and pull directions. The push direction resulted in greater forces than the pull direction. The push direction also caused greater resultant shoulder moments while the pull direction caused greater extensor and resultant moments about the L5/S1 joint. In addition, the push direction resulted in greater mean %MVE magnitudes for most muscles, except for the erector spinae muscles, in which the opposite trend was observed. Mean %MVE magnitudes for total and weighted average variables were consistently greater for the push direction than the pull direction. Direction had a main effect on most of the individual EMG/force ratios and primarily the total right and weighted average right and total EMG/force ratios. Together with height or elbow angle, several interaction effects also existed for the EMG/force ratios. (Supported)

Hypothesis 3

It was hypothesized that there would be statistically significant differences in hand force magnitudes, shoulder and low back moments, normalized EMG values and EMG/force ratios as the handle height increases from 100 cm to 150 cm. All hand force magnitudes at the 100 cm height were greater than those at the 150 cm height, but were also accompanied by greater sagittal plane moments. Height had a main effect on mean %MVE for eight of the fourteen muscles examined, but none of the mean %MVE values for total or weighted average EMG parameters were significantly different between heights. The 150 cm height yielded higher EMG/force ratios for nine of the fourteen muscles measured and for all total and weighted average EMG/force ratios, compared to the 100 cm height. (Supported)

Hypothesis 4

It was hypothesized that there would be statistically significant differences in hand force magnitudes, shoulder and low back moments, normalized EMG values and EMG/force ratios between vertical and horizontal handle orientations. A majority of the differences in all dependent variables between vertical and horizontal handle orientations were not significant. Resultant shoulder and low back moments were found to be significantly greater for horizontal handle orientations. Significant interactions with direction and handle height were also evident on sagittal plane moments. Handle orientation otherwise did not contribute greatly in explaining any of the differences in dependent variables. (Partially supported, predominantly for moments)

Hypothesis 5

It was hypothesized that there would be statistically significant differences in hand force magnitudes, shoulder and low back moments, normalized EMG values and EMG/force ratios between fully extended elbows and elbow angles of less than or equal to 90 degrees. Differences in hand force magnitudes between fully extended elbows and elbow angles of less than or equal to 90 degrees were not significant. However, interaction effects between direction and elbow angle were significant. Greater resultant and flexor moments were calculated for the right shoulder for the extended elbow conditions. Elbow angle did have a main effect on mean %MVE magnitudes for nine of the fourteen muscles measured, but like height, none of the mean %MVE values for total or weighted average EMG parameters were significantly different between elbow angle conditions. With respect to EMG/force ratios, elbow angle was found to be significant for seven of the fourteen muscles examined, but not for any of the total or weighted average EMG/force ratios. (Partially supported)

5.2 Anthropometrics

Gender and mass significantly influenced hand force capability. Males and larger body masses were associated with greater left, right and total hand force capability. In the present study, females had a mean total hand force capability (THF) of 254.9 N while males had a mean THF of 381.8 N. Several previous studies have also found female hand forces to be lower than those of males (Al-Eisawi, 1999; Ayoub & McDaniel, 1974; Chaffin & Andres, 1983; Fothergill et al., 1991; Kumar, 1995; Kumar et al., 1995; Snook, 1978; Snook & Ciriello, 1974, 1991; van der Beek et al., 2000). Kumar et al. (1995) reported female forces to be between 1% and 29% lower than male forces. Larger differences of 67% have also been found (Chengalur et al., 2004). In particular, mean female strengths for the lower extremity are more comparable to males than for the upper extremity and trunk (from Laubach, in Webb Associates, 1978). A gender by direction interaction effect was evident on all hand force capability estimates, in which the push direction generated greater forces than the pull direction, but within each direction, males generated greater forces than females (Figures 7-9). van der Beek et al. (2000) similarly found that for both genders, maximum strength was significantly greater for pushing than pulling. Furthermore, the lower female hand force capability estimates support the use of females as the reference population often used for design thresholds, especially for pull exertions or upper limb tasks, such as seated work.

Anthropometric characteristics have often been used to explain differences between genders. Kumar et al. (1995) determined that mass was the best predictor in explaining peak isometric pull forces, in which mass accounted for 20.3% of the variance. One supposed mechanism is that a greater body mass decreases the required coefficient of friction to prevent slipping during pushing and pulling (Lee et al., 1992). Ignoring hand forces during push exertions, the forward turning moment about the centre of pressure (COP) at the feet is the product of body weight and the horizontal distance between the centre of mass (COM) of the body and the COP. Thus, an increase in either body mass or the moment arm of the COM will increase the forward turning moment. To preserve balance, it was reasoned that the increase in the forward turning moment of the body must be counteracted by an increase in the backward turning moment generated by hand forces (Hoozemans, 1998). This may explain why larger body masses also resulted in greater resultant moments at all joints examined. Thus, it is plausible that the exerted push force may increase with increases in body mass, for a given posture.

In the current study, when mass and stature were accounted for, gender differences persisted, as has also been suggested in previous work. Thus in addition to mass and stature, males and females must have further characteristics to explain the differences in hand forces. Laubach and McConville (1969) and Nordgren (1972) also found that stature was not well correlated with strength for males or females. It has been proposed that muscular exertions about the shoulder are more difficult for females due to the smaller muscle moment arms associated with the smaller average female shoulder and thoracic skeletal frame (Stobbe, 1982). Kumar (1991) further showed that static strengths generally produce larger differences than dynamic strength, thus during sustained pushing or pulling of carts, for example, gender differences may be reduced. The physiological cross sectional areas of muscles are presumably greater for males than females, but also vary widely within each gender. Bishop, Cureton and Collins (1987) showed that muscle size accounted for the greatest proportion of the total variance in population strength data, based on either an individual's lean body mass (body mass corrected for fat) or the cross-sectional area of body segments. Laubach and McConville (1969) found that both total body weight and lean body mass were correlated with muscle static strengths. Pandya, Hasson, Aldridge, Maida, and Woolford (1992) further confirmed the importance of lean body mass relative to total body weight and stature in estimating muscle strengths. Thus, lean body mass and size may be better predictors of hand force magnitudes than gross mass and stature.

In addition to males being more physically adept at generating force, for a given force level, it is likely that females would need to activate their muscles to a greater extent than males. The current study demonstrated that gender played a significant role on mean %MVE for individual and total EMG parameters. On average, female mean %MVE ranged from 14.6% to 40.5% while male mean %MVE ranged from 13.4% to 34.3%. Females tended to have greater mean %MVE values than males for the right biceps brachii and triceps brachii muscles. There was also a main effect of gender on total left and total EMG as well as on weighted average left and total EMG such that females activated their muscles to a greater extent than males (Figures 45 and 48). As males generated greater push and pull forces than females with reduced muscle activity levels, it is possible that males employ a different technique to exert force. For example, it is possible that males used their lower body to generate force compared to their upper body (Figure 88). Although not statistically significant, males did adopt a wider foot stance than females, which may indicate that males took greater advantage of their body weight to contribute to the force exertions. Thus, exertion technique may be a method to reduce mean %MVE and should be further investigated.

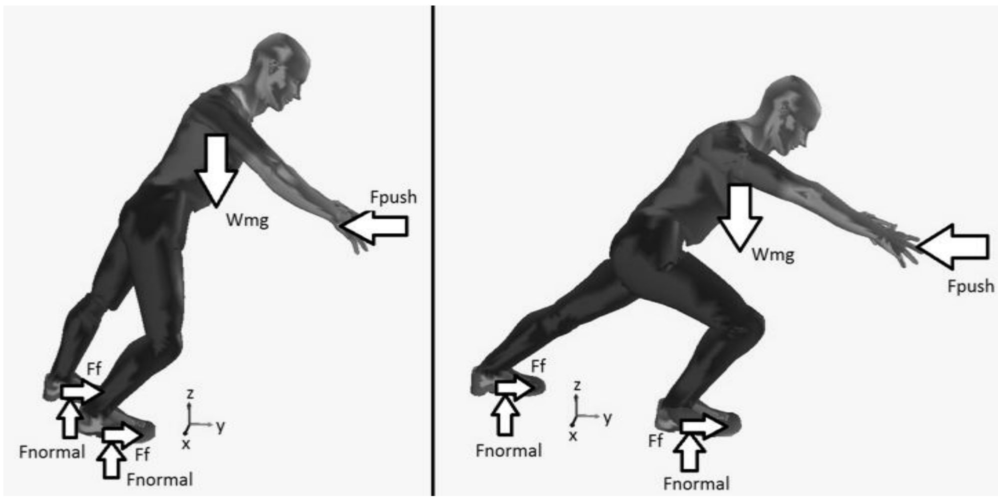

Figure 88: Free body diagrams showing the use of a wider foot stance as a possible mechanism to increase hip flexion to increase the forward turning moment about the COP at the feet for push exertions at 100 cm. Free body diagrams were simulated using 3DSSPP 6.0.4 (University of Michigan, Ann Arbor, MI).

The effects of mass on mean %MVE were limited. In general, a larger mass was associated with a lower mean %MVE. It was shown that larger gross body masses are positively correlated with larger lean body masses and that together these parameters helped to predict muscle strengths (Pandya et al., 1992). This effect may have been more prominent if lean body masses were measured. Reduced mean %MVE estimates may have also resulted if participants with a larger gross body mass elected to lean forward more at the hips to further increase the forward turning moment, which would need to be resisted by an increased push force on the handles. Likewise, participants may have leaned backward at the hips or transferred their weight to their rearward leg by sitting backward thereby increasing the supportive pull force at the hands. The use of the body weight to increase the forward and backward turning moments transfers the force generating role from the smaller upper limb muscles to the larger back and lower limb muscles. It may have been interesting to examine fixed body postures to determine the extent that gross body mass contributes to increasing the forward or backing turning moment about the COP towards increasing the push and pull force magnitudes, respectively.

In addition to females having lower hand force capability, their corresponding EMG/force ratios were also larger. Likewise, smaller body masses were generally accompanied by higher EMG/force ratios. Females had higher right biceps brachii, right and left triceps brachii, left and total, and weighted average left and total EMG/force ratios, based on the GMSDHO model. Based on the GMSDE model, females had higher ratios for the right biceps brachii, right triceps brachii and left deltoid muscles. Across models, mass also had a main effect on left, right and total EMG/force ratios and weighted average right and total EMG/force ratios. Larger masses tended to have lower EMG/force ratios. As predicted by Schaefer (2007), the EMG/force ratios indicate that females must activate their muscles to a greater extent than males, for a given force level. As a result, it is possible that females may be at greater risk for fatigue while performing repetitive or sustained exertions at the same absolute force level as males. Sustained contractions of greater than 10% MVE were found to induce localized muscle fatigue (Sjøgaard, Savard, & Juel, 1988). While push and pull tasks requiring maximum forces may not often be sustained, it is likely that the %MVE ranges observed in the current study may lead to muscular fatigue under repetitive conditions. It thus seems reasonable that females were found to have a significantly higher prevalence rate ratio when compared to males for both low back and shoulder complaints (Hoozemans, 2002).

5.3 Direction

Direction had a significant effect on hand force capability and joint moments. Hand force capability for the push direction was consistently greater than hand force capability for the pull direction (Figures 4 and 14). The push direction caused greater resultant shoulder moments while the pull direction caused greater extensor and resultant moments about the L5/S1 joint (Figures 18 and 20). Hoozemans et al. (2004) also found net moments at the low back to be higher during pulling compared to pushing. Hand force interaction effects between gender and direction also revealed that differences between genders with respect to force capability were larger for the push direction than the pull direction (Figure 7). In the current study, two-handed force capability for combined genders across conditions was found to be 400.4 N for the push direction and 236.2 N for the pull direction. Several studies have also found maximum push forces to be greater than maximum pull forces. Chaffin and Andres (1983) found two-handed maximum forces across genders to be 302 N for pushes and 235 N for pulls, with heights set at 68 cm, 109 cm and 152 cm. These are comparable to the results of the current study. Warwick reported maximum push and pull forces at shoulder height to be 292 N and 170 N, respectively, where feet were side by side. These forces are lower than those reported in the current study, but may be explained by foot stance. Chaffin and Andres (1983) found that when the feet were allowed to be staggered in the anterior-posterior plane, significantly greater push forces than pull forces were recorded. Likewise, MacKinnon (1998) found that maximum pull forces produced under set foot positions were about 36% lower than forces generated with freely chosen foot positions. In a psychophysical study, Ciriello et al. (1993) found maximum acceptable initial and sustained two-handed push and pull forces at about 95 cm to be 300.2 N and 200.1 N and 260.0 N and 159.9 N, respectively.

These forces are generally lower as they represent maximum acceptable forces rather than maximum hand force capability. Higher two-handed push forces have also been reported by Ayoub and McDaniel (1974), Snook (1978) and van der Beek et al. (2000). Other studies found maximum pull forces to be greater than maximum push forces (Das & Wang, 2004; Kumar, 1995, Kumar et al., 1995; Seo, Armstrong & Young, 2010). Many of these studies imposed postures by seating participants or stabilizing the lower extremities by securing them to a metal base. The results of the current study reiterate the importance of direction in generating maximal push or pull forces.

Foot position may explain the greater hand force magnitudes in pushing compared to pulling. In the current study, it was observed that participants often selected a rearward left foot position for pushing and a forward left foot position for pulling, with respect to the fixed right foot position. For pulling, Ayoub & McDaniel (1974) suggested placing the forward foot close to or under the handle to enable the individual to lean back further, pivoting about the forward foot to increase force exertion capability and possibly prevent falling. Likewise in pushing, a wider foot stance as found in the current study is a feasible mechanism towards increasing the push force capability by increasing the ability to lean forward (Figure 89). Chaffin and Andres (1983) and Hoozemans et al. (1998) similarly contended that staggered foot positions may increase the forward and backward turning moments. A rearward foot position in pushing enables participants to lean forward more, rotating about their rearward foot while using the forward foot as additional weight to increase the forward turning moment of the body around the COP. Likewise, a forward foot position in pulling allows participants to lean backward more or sit backward, rotating about their forward foot to increase the rearward turning moment. In fact, Ayoub and McDaniel (1974) had their participants lift their forward foot off the ground during pushes and their rearward foot off the ground during pulls. It was reasoned that any downward force on the non-force bearing leg (forward foot in pushes, rearward foot in pulls) would cause an upward vertical reactive force opposing body weight and thus reduce the forward or backward turning moments. Hoozemans et al. (1998) further argued that to preserve postural balance in pushing, for example, the increase in the forward turning moment of the body caused by a rearward foot position needs to be resisted by an equal increase in the backward turning moment. This may be achieved by increasing the push force at the hands and/or by activating the extensor musculature (assuming a neutral lordotic posture is adopted (McGill, 2007)). Foot stance appears to play a crucial role in the development of force and should be considered when examining hand force magnitudes in the laboratory or when allocating space for work tasks.

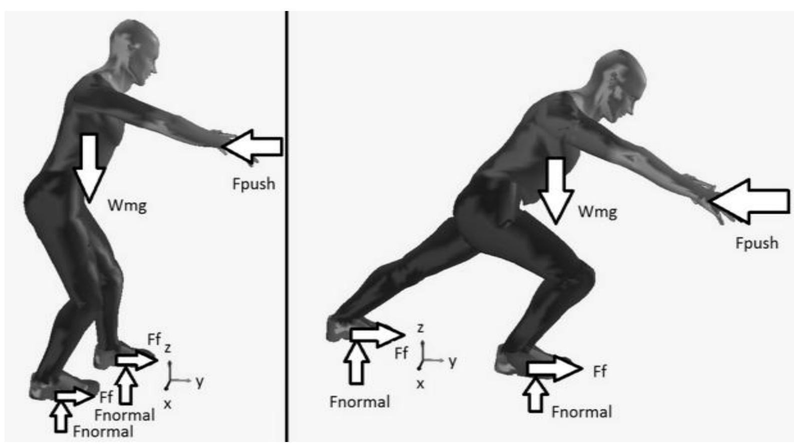

Figure 89: Free body diagrams to show how a wider foot stance and use of a rearward left foot position with respect to the fixed right foot position during push exertions may enable participants to increase hip flexion and make use of body weight and/or the lower limbs to increase the forward turning moment about the COP and increase push capability. Free body diagrams were simulated using 3DSSPP 6.0.4 (University of Michigan, Ann Arbor, MI).

The ability to generate moments based on the postures adopted may explain the interactions between direction and height and direction and elbow angle on force capability and reactive moments. The current study found that the moments developed by hand forces at the higher height were generally greater than the moments caused by body weight. At higher heights, the vertical moment arm of the hand force increases and the anterior moment arm of body weight decreases as participants adopted a narrower foot stance, thus restricting hip flexion. Thus, this may explain why pushing at the 150 cm height led to flexor moments at the L5/S1 joint while pushing at the 100 cm height led to extensor moments (Figure 25). Ayoub and McDaniel (1974) similarly found that increasing height for push exertions led to reduced extensor moments or even flexor moments. Ayoub and McDaniel (1974) also found that push forces and extensor moments at the hip, particularly for females, increased with increasing foot distance, which effectively decreases the handle height. Moreover, when pushing, the ability to flex at the hips and thus increase the forward turning moment increases when the elbows are flexed, whereas in pulling, the ability to extend at the hips and further augment the backward turning moment increases with extended elbows. This notion was realized by the interaction effects between direction and elbow angle which showed that the push direction yielded greater forces than the pull direction, but both the greatest and least forces were generated when the elbow were flexed (Figures 15-17). These interaction effects emphasize the importance of posture as affected by individual preference and working conditions when evaluating joint moments.

Cocontraction of all muscles was evident during all pushing and pulling tasks. This seems reasonable as the trunk muscles, for instance, are purposely employed to ensure sufficient stability of the spine in order to withstand loading and sustain postures and movement (McGill, 2007). Likewise, the deltoid muscles help stabilize the shoulder joint and hold the humeral head within the glenoid cavity during arm movements (Moore & Dalley, 1999). Thus, cocontraction is an integral part of all tasks and both agonist and antagonist muscles should be monitored when evaluating the effect of potential risk factors on muscle activation levels.

The shoulder musculature and abdominal muscles may be more prone to musculoskeletal complaints for push exertions while the low back extensors may be more at risk for pull exertions. The shoulder muscles and the rectus abdominis muscles elicited greater mean %MVE levels during push exertions, which were also greater than the low back extensors, with the middle deltoid having the greatest mean %MVE of all muscles monitored. In addition, the erector spinae muscles exhibited greater mean %MVE during pull exertions (Figures 36-37). MacKinnon and Vaughan (2005), whom investigated one-handed submaximal pulling exertions, found that the shoulder complex musculature tended to work at a greater %MVC level than the trunk musculature. This corroborates the finding that pushing and pulling were associated with greater shoulder complaints than low back complaints as suggested by higher odds ratios (Hoozemans et al., 2002). MacKinnon and Vaughan (2005) further noted that the deltoid elicited the greatest relative activity of all shoulder muscles measured and would likely be at the greatest risk for onset of fatigue. This differs from the results of the current study. When accounting for direction, the middle deltoid appeared to elicit the greatest mean %MVE of all shoulder muscles for the push direction, but the lowest mean %MVE for the pull direction. Thus, there appears to be a dependence on force direction when considering the risk of developing MSDs.

The effect of direction on mean %MVE estimates was different for shoulder and trunk muscle groups. Mean %MVE estimates were greater for the push direction than the pull direction for all muscle pairs except for the erector spinae muscles, based on the GMSDHO models (Figure 36-37). Similar findings have been found where, on average, normalized muscle activities in right and left erector spinae during pull exertions were greater than those during push exertions while the reverse was found for the middle deltoid and trapezius muscles (Jongkol, 2006). Given that the force levels were also significantly lower for the pull direction than the push direction, it is possible that participants primarily relied on active hip extension, as indicated by the increase in erector spinae activity, to generate pull forces. As it was observed that participants adopted a narrower foot stance in the anterior-posterior plane for the pull conditions compared to the push conditions, it is also possible that the angle of hip extension during pull exertions was less than that of hip flexion during push exertions. Consequently, the horizontal distance between the COM of the body and the COP, and thus the resulting flexor moment that hip extension may have caused, would likely be smaller. It may thus be reasonable to postulate that to preserve postural balance for pull exertions, the counteracting forward turning moment accomplished by an increase in the pull force at the hands may not need to be as large in magnitude. In reference to pushing, greater shoulder muscle activity was observed. This may indicate that the reactive extensor moment may have been primarily generated by larger hand forces due to the active contraction of the shoulder musculature than through hip extension. With respect to reach distance, MacKinnon and Vaughan (2005) found that a shoulder strategy was employed for near pulls as the shoulder muscles exhibited high muscle activation levels while limited erector spinae muscle activity was observed. For greater reach distances, a trunk motion strategy seemed to emerge as the relative activity of the shoulder muscles was reduced while increasing erector spinae activity was evident. Within the constraints of the current study, it may be that participants adopted a shoulder strategy for push exertions and a low back strategy for pull exertions.

By examining hand force capability and mean %MVE estimates as separate entities, it would appear that the push direction enables greater force production albeit at the expense of greater muscle activation, compared to the pull direction. However, higher EMG/force ratios resulted for the pull direction (Figure 67). This indicates that for a given force level, the pull direction requires individuals to activate their shoulder and trunk muscles to a greater extent than the push direction, based on the muscles monitored. The pull direction also led to greater extensor moments at the L5/S1 joint than the push direction (Figure 20). Direction must be considered with height when examining moments at the shoulder joints. While the push direction with the 150 cm height resulted in relatively lower moments, the push direction with the 100 cm height caused the greatest shoulder moments (Figures 23 and 24). Ayoub and McDaniel (1974) also recommended larger foot distances as a means to delay the onset of fatigue as measured by the time until a 10% decrease in force was observed. As participants in the current study adopted a wider stance with push exertions, the rearward force bearing leg was a greater distance from the handles. Thus, push exertions may also delay the onset of fatigue, as corroborated by the EMG/force ratios. Across models, higher EMG/force ratios resulted for the pull direction than the push direction for the triceps brachii, middle trapezius, rectus abdominis and erector spinae muscles. The reverse effect was apparent for the biceps brachii, pectoralis major and middle deltoid muscles. It was further found that the right and weighted average right and total EMG/force ratios for the pull direction were associated with higher EMG/force ratios. In general, the push direction may be more favourable at producing the greatest force while reducing the level of muscle activity from most muscles examined. Furthermore, the push direction reduces the L5/S1 joint extensor moments and more specifically, with the 150 cm, may reduce the shoulder moments. It is evident that the push direction is more favourable for force production and also proves effective in alleviating the muscular demands and resulting joint moments.

5.4 Handle Height

Handle height influenced hand force capability and joint moments. Hand force magnitudes and positive sagittal plane moments were significantly greater for the 100 cm height compared to the 150 cm height (Figure 6 and Figure 19). This may suggest that the 100 cm height may result in greater hand forces at the expense of greater moments. It was previously shown that increased net moments at the low back were associated with increased cart weights, which have been shown to be related to exerted forces (Hoozemans et al., 2004). However, forces and moments also depend on directional effects. An interesting direction by height effect was found whereby the push direction always yielded greater hand forces than the pull direction (Figures 10-12). Within the pull direction, hand forces were significantly greater at the 100 cm height. Within the push direction, the two handle heights were not significantly different from each other in terms of magnitude, but the 150 cm height generally resulted in greater left and total hand forces while the 100 cm height tended to produce greater right hand forces. Snook and Ciriello (1991) similarly found that participants were able to generate greater push forces at higher handle heights whereas greater pull forces were generated at lower handle heights. Likewise, MacKinnon (1998) found that pull forces increased with lower heights. Kumar et al. (1995) found that both push and pull forces were greatest at the 100 cm height compared to 50 cm and 150 cm heights. Pinder et al. (1995) found that participants generated significantly lower forces at the 150 cm handle height than at the 100 cm and 50 cm handle heights, across six directions (Pinder, 1995). Furthermore, Ayoub and McDaniel (1974) found that push forces increased with increasing height from 60% to 100% of shoulder height while the opposite was shown for pull forces. With respect to moments, while the push direction and the 100 cm height resulted in the largest moments at the shoulder joints the same interaction resulted in significantly lower extensor moments at the L5/S1 joint (Figures 23-25). Thus, decreasing moments at one joint may effectively increase moments at other joints. To maximize forces, the push direction with the 150 cm height or the pull direction with the 100 cm height appear to be the most effective force generating combinations while also helping to minimize the magnitudes of all joint moments, under the conditions examined.

The lower handle height may be associated with increased stability of the human-machine interface. At the lower handle height, the height of the pull or pull exertion approaches the height of the whole-body COM thus increasing the stability, allowing for greater magnitudes in horizontal force production (MacKinnon, 1998). This may explain why participants in the current study adopted a wider foot stance for the 100 cm height than the 150 cm height. Daams (1993) also found that the shoulder height tended to result in a narrower foot stance compared to the elbow height for push and pull exertions, which similarly resulted in lower hand forces. In comparing male and female push forces, Ayoub and McDaniel (1974) noted very small differences when the foot distances from the handles were small, suggesting that it may be more difficult to take advantage of a larger body mass (often associated with males) in generating force. With respect to pulling, increased trunk flexion which occurs more often with lower handle heights due to physical constraints, may allow the participant to take advantage of the high inertial properties of the upper-body to generate momentum in the extension direction to pull against the handle (MacKinnon & Vaughan, 2005). In terms of pushing, an increase in stature (by means of a lower fixed handle height) may increase the forward turning moment of the body around the COP, which in turn may be resisted by an equal increase in the backward turning moment (in the form of a push force) to maintain postural balance (Hoozemans et al., 1998). Although the lower handle height may allow for further hip flexion when pushing, or hip extension when pulling to effectively use the weight of the body to contribute to hand forces, there is also a greater risk of falling forward or backward, respectively, if the foot should slip. However, if the individual is able place one foot in front of the other given space constraints, then the forward foot in pushing or the backward foot in pulling may enable the person to catch themselves and prevent a fall, which is also more feasible at lower heights. This potential mechanism is illustrated in Figure 90. These points suggest that the greater hand forces achieved with the lower handle height may be the result of the increased ability to improve stability.

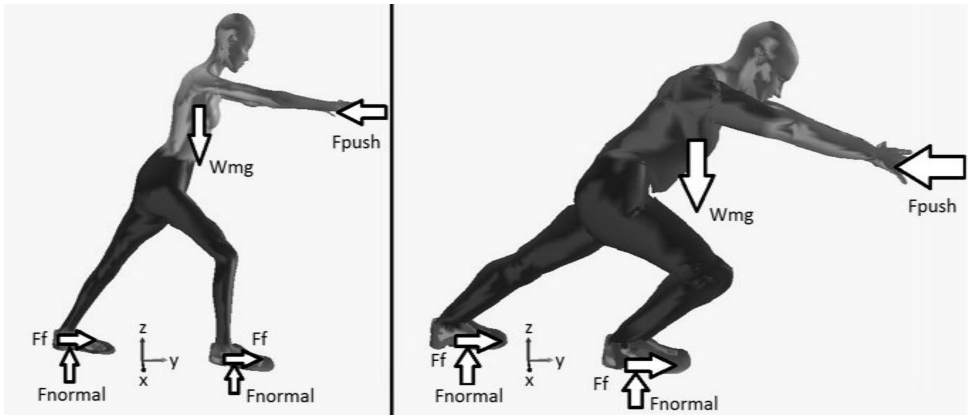

Figure 90: Free body diagrams to show the possible increase in hip flexion achieved with an increase in stature. The mannequin on the left demonstrates a posture illustrative of a 5th percentile North American female while the mannequin on the right depicts a 5th percentile North American male pushing at 100 cm. Free body diagrams were simulated using 3DSSPP 6.0.4 (University of Michigan, Ann Arbor, MI).

The line of action of the force exertion may explain the significant differences observed between push and pull forces at the 150 cm handle height. Pheasant et al. (1982) found that peak force vectors occurred in the lift-push direction, primarily at the 100 cm and 175 cm heights, in which force was exerted in a line approximating a straight line between the hands and feet. Chow and Dickerson (unpublished data) similarly found that anterior push exertions at or above shoulder height were generally accompanied by upward forces. Fothergill et al. (1991) proposed that these postures primarily employed lower extremity muscle groups. In the current study, participants tended to adopt forward left foot positions for pull exertions and rearward left foot positions for push exertions. Thus, under the push conditions, participants may have been better able to make use of their lower extremity muscles than during the pull conditions, attributing to the greater hand forces measured. However, lower extremity muscles were not measured in the current study. The current study further showed that the sagittal plane moments for the shoulder and L5/S1 joints were found to be lower at the 150 cm height (Figure 19). In comparing push and pull directions for the 150 cm, the magnitudes of the sagitttal plane moments were also smaller for pushing (Figures 23-25). Hoozemans et al. (2004) and Abel and Frank (1991) similarly found net moments to be lower at higher handle heights. Thus, it may be thought that the postural configurations associated with the 150 cm handle height better enable participants to keep the shoulder and low back joints close to the line of action of the exerted force and may have helped to reduce the moments at those joints (Figure 91). Thus, reducing the demands on the smaller muscles to resist counter moments by employing the use of the larger lower limb muscles also helps to effectively increase hand forces.

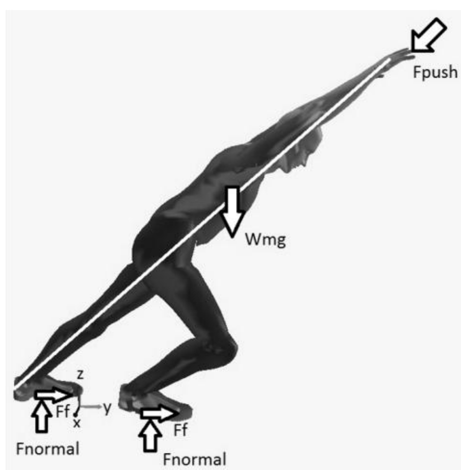

Figure 91: Free body diagram of the general posture adopted during push exertions at the 150 cm handle height. The shoulder and low back joints are approximately in line with the direction of force exertion as indicated by the solid white line. Free body diagram was simulated using 3DSSPP 6.0.4 (University of Michigan, Ann Arbor, MI).

The objective to minimize external moments, whether intentional or fortuitous, appears to be coupled with the goal to minimize muscle activation levels. The reduction in sagittal plane moments with increasing handle heights (Figure 19) corresponded to the decrease in mean %MVE for the right biceps brachii, right and left pectoralis major, and right and left erector spinae muscles (Figures 38 and 39). By exerting force in a direction that approximates a line connecting the hands to the feet, counter moments may be minimized, which would otherwise be resisted by the muscles (Pheasant et al., 1982). Due to the observed position of the hands being at or above shoulder level when the handles were set at the 150 cm heights, this explanation appears tenable. Particularly for pushing, in which participants generally adopted a rearward left foot position, it is likely that the joints were brought even closer in line with the direction of force exertion. This may explain the reduction in mean %MVE of the biceps brachii, pectoralis major and erector spinae muscles in moving from the lower to higher handles height for the push direction (Figures 40, 41, 44, 45, 49 and 50). Jongkol (2006) found similar trends for pushes in the normalized EMG of the right and left erector spinae muscles. The results demonstrate that minimizing counter moments is an effective approach to minimizing mean %MVE, which may ultimately delay the onset of fatigue.

Muscle function plays an important role in explaining muscle activation levels. In the current study, as the height increased to 150 cm during pulls, the mean %MVE decreased for the right and left triceps brachii and middle trapezius muscles but, as the height increased during pushes, the mean %MVE increased for these muscles (Figures 42, 43, 46 and 47). Jongkol (2006) found similar trends for the middle trapezius muscles. The triceps brachii muscles are primarily responsible for extension of the elbow but may also aid is extending the shoulder. Since the long head of the triceps brachii crosses the shoulder joint, it has the ability to help stabilize the glenohumeral joint by resisting inferior displacement of the head of the humerus (Moore & Dalley, 1999). Extension of the elbow often occurs with push exertions while shoulder extension and inferior displacements are more likely to result from exertions causing positive moments at the shoulder. Positive moments may be created with pushes at 150 cm or pulls at 100 cm. Based on the current findings, it appears that the triceps brachii muscles were employed for both shoulder extension and stabilization as well as for elbow extension. The middle trapezius muscles are involved in scapular stability, bracing the shoulder by pulling the scapulae posteriorly and superiorly (Moore & Dalley, 1999). Thus, it is likely that the increased mean %MVE of the middle trapezius for push exertions at the 150 cm acted to stabilize the protracted scapulae in this posture. The current study also found decreases in mean %MVE of the erector spinae muscles for the 150 cm during pull exertions. This may have been due to the inability for participants exerting force at the higher handle height to take advantage of the inertial properties of the trunk in pulling. Jongkol (2006) found the reverse trend for pulls where normalized EMG increased with an increase in height. This contradiction may be attributed to the difference in foot positions used between the studies, where participants were able to adopt asymmetric foot positions in the current study. It may be that the decision to give participants some control over foot positions may have undoubtedly given participants greater control over the ability to use the trunk to help generate forward or backward turning moments, particularly for the lower handle height. Thus, the inherent properties of the triceps brachii and erector spinae muscles seem to be plausible explanations for the observed interaction effects between direction and handle height.

Height also had a main effect on EMG/force ratios. In general, the 150 cm height resulted in higher ratios than the 100 cm height (Figures 57 and 58). The ratio may have been more affected by the force component than mean %MVE as hand force magnitudes were consistently and significantly greater for the 100 cm height than the 150 cm height. A significant interaction effect existed between direction and height between the pairs of right and left biceps brachii (Figures C1 and C2), pectoralis major (Figures C3 and C4), rectus abdominis (Figures C6 and C7), erector spinae (Figures C8 and C9), and the left middle trapezius (Figure C5) muscles. Based on the ratios, it would appear that the interaction between the pull direction and the 150 cm height tended to result in the highest ratios. This may suggest that pull exertions at high heights should be avoided as greater muscle activity is required for a given force level relative to other direction and height combinations. There was a main effect of height on all total and weighted average EMG/force ratios (Figure 65). The 150 cm height also led to higher EMG/force ratios than the 100 cm height for all total and weighted average EMG/force ratios. Furthermore, the push direction with the 150 cm height and the pull direction with the 100 cm height generally led to lower EMG/force ratios. These results support the force and moment data that suggest that these combinations may provide the greatest force output while minimizing the muscular demands and joint moments.

5.5 Handle Orientation

The vertical and horizontal handle orientations examined in this study did not show any significant effects on hand force capability estimates as defined in this study. Comparable results were found in a study examining horizontal and vertical handle orientations for two-handed isometric push exertions at elbow height (Olanrewaju & Haslegrave, 2008). Olanrewaju and Haslegrave (2008) found that vertical handle orientations, on average, produced greater left, right and total hand forces in the anterior direction, however, differences in force between the two orientations were not statistically significant. It has further been recommended that the hand should be allowed to exert force to a handle through compression rather than shear (Pheasant & Haslegrave, 2006). A method in which force may be exerted through compression exists when the long axis of a cylindrical handle is oriented perpendicular to the direction of force exertion. In contrast, force is exerted to a handle through shear often when the long axis of a cylindrical handle is oriented parallel to the direction of force exertion. When the long axis of the handles are oriented perpendicular to the force exertion direction, the push or pull forces are the result of the normal force and friction forces acting in the direction of the resultant push or pull, however, force generated by parallel handles rely on friction alone for the coupling between the hand and handle (Seo et al., 2010). In addition, a perpendicular handle provides mechanical interference during pushing to prevent the hands from slipping, thus upper-body push forces may be applied directly against the handle in the anterior direction (Seo et al., 2010). Consequently, it was found that maximum anterior push forces for perpendicular handles were, on average, 52% greater than maximum anterior push forces for parallel handles (Okunribido & Haslegrave, 2008). Seo et al. (2010) also reported that mean maximum push and pull forces increased by 11% when perpendicular handles were used instead of parallel handles. The results of the current study suggest that as long as forces are exerted through compression, either vertically or horizontally oriented handles may be employed, without impairing force performance.

Handle orientation had a significant effect on resultant right shoulder, left shoulder, and low back moments, such that the horizontal orientation caused greater resultant moments. The 150 cm handle height with the horizontal handle orientation further resulted in greater resultant moment at all joints (Figure 22). However, when examining sagittal plane moments, the significant interaction effect between direction and orientation revealed that the push direction with the horizontal orientation resulted in the smallest and negative left shoulder and L5/S1 joint moments while the pull direction with the horizontal orientation led to the greatest and most positive moments. Thus, these results demonstrate how resultant moments mask the directionality of moments. Ultimately cocontraction will occur to stiffen the spine and brace it from buckling (McGill, 2007) but, the moment direction will determine whether the back or abdominal muscles must further activate to support the moment and maintain equilibrium. For example, the large positive moment at the L5/S1 joint must be supported by the back extensors. These are large muscles and have the greatest possible moment arm, which allows them to generate a large extensor moment with minimal compression on the spine (McGill, 2007). Thus, although the pull direction may generate greater extensor moments at the L5/S1 joint, the body may be more adept at supporting the moment, within compressive limits. The results also show the effect that height has on shoulder and L5/S1 joint moments when considering different handle orientations. In examining both interactions, the vertical handle orientation and the 100 cm height for pushes or pulls may be more conservative with respect to moments, based on the conditions examined.

Handle orientation only had a significant main effect on mean %MVE estimates for two of the fourteen muscles examined, which is partially supported by earlier studies. In a study investigating meat cutting, it was predicted that modifying the knife handle to allow the use of a stab grip (handle is perpendicular to direction of force exertion) rather than a slice grip (handle is parallel to direction of force exertion) could double force exertion capacity and reduce muscle activation by as much as 80% (Grant & Habes, 1997). This prediction was based on the premise that the stab grip promoted a neutral wrist posture while the slice grip was often associated with ulnar deviation, particularly with increased work heights. In the present study, it was hypothesized that the vertical and horizontal handle orientations would result in significant differences in mean %MVE for the muscles examined. At the 100 cm height, it was thought that vertical handles may promote neutral forearm postures while horizontal handles required the forearms to be pronated. At the 150 cm height, it was thought that vertical handled would result in ulnar deviation at the wrists while horizontal handles permitted neutral wrist postures. As the only forearm muscles examined in the current study were the biceps brachii and triceps brachii muscles, perhaps differences in mean %MVE may have resulted if wrist and additional forearm muscles were monitored, such as wrist deviators or forearm pronators and supinators. Further investigation with the integration of more wrist and forearm muscles needs to be undertaken before the effects of handle orientation on muscular demands may be concluded.

5.6 Elbow Angle

Elbow angles have a large influence on the magnitude of the moment arms about the joints and COP. Elbow angle had a significant effect on sagittal plane and resultant right shoulder moments where extended elbows resulted in significantly greater flexor moments than flexed elbows (Figures 33 and 35). Extending the elbow effectively increases the moment arm in the anterior direction, which may explain the larger moments experienced by the right shoulder joint. Only the interaction between direction and elbow angle had an effect on left, right, and total hand force capability, in which the push direction generated greater forces than the pull direction (Figures 15-17). Both the greatest and least hand forces resulted when the elbows were flexed. This corresponds with the pronounced direction by elbow angle effect shown for selected foot stance, in which the push direction resulted in wider foot stances than the pull direction and both the widest and narrowest foot stances chosen were those for the flexed elbow conditions. When pushing, the ability to flex at the hips and thus increase the forward turning moment about the COP increases when the elbows are flexed, whereas in pulling, the ability to extend at the hips and further augment the backward turning moment increases with extended elbows (Figure 92). Likewise, a wider foot stance when pushing or pulling will help to increase the respective forward and backward turning moments about the COP. This notion was realized by the interaction effects between direction and elbow angle which showed that for pushing, greater forces resulted with flexed elbows while for pulling, greater forces were generated with extended elbows. The elbow angle adopted clearly depends on the goal of the task and the spatial constraints of the work environment.

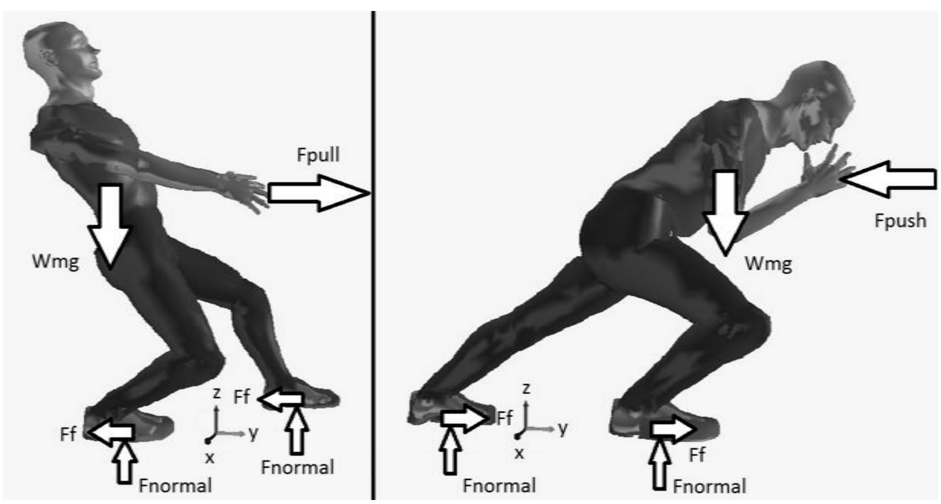

Figure 92: Free body diagrams to demonstrate the increased ability to extend at the hips in pulling with extended elbows (left) or flex at the hips in pushing with flexed elbows (right). Free body diagrams were simulated using 3DSSPP 6.0.4 (University of Michigan, Ann Arbor, MI).

Many of the muscle activation trends can be explained by muscle function. Greater mean %MVE resulted when the elbows were fully extended for the right triceps brachii, left pectoralis major and for the right and left middle deltoid muscles, while greater mean %MVE resulted when the elbows were $\leq 90^{o}$ for the right and left middle trapezius muscles (Figures 40-41). Shoulder flexion is primarily dependent upon the activation of the anterior and middle thirds of the deltoid, such that the arm cannot be held against gravity without the deltoid's contribution (Rockwood, Matsen, Wirth, & Lippitt, 2009). Thus, it appears reasonable that the right and left middle deltoid muscles exhibited greater mean %MVE when the elbows were extended, a condition requiring increased shoulder flexion to accommodate for the increased horizontal reach, compared to the flexed elbows condition. Furthermore, the deltoid is a large broad muscle accounting for about 20% of the shoulder muscles (Rockwood et al., 2009). Thus, the stabilizing function of the deltoid is thought to be significant, in which the shoulder may be less stable under elevated conditions (Rockwood et al., 2009). Furthermore, as the extended elbow conditions resulted in greater flexor moments at the right shoulder joint, it may be expected that the right middle deltoid would responsible for supporting the moment, whereas the right triceps brachii may play more of a stabilizing role. Although the clavicular head of the pectoralis major plays a smaller role in shoulder flexion, the increased shoulder flexion resulting from the extended elbows conditions accounts for the greater mean %MVE observed. Furthermore, in the flexed elbows conditions, participants were observed to abduct the shoulder slightly. This would require upward rotation of the scapula, which involves the activity of the middle trapezius muscle for stabilization. The shoulder muscles measured demonstrate the important role of muscles for moment production and joint stabilization.

Extending the elbows may be a useful strategy to reduce the demand on the smaller forearm musculature. In the current study, right and left biceps brachii mean %MVE were reduced when the elbows were extended (Figures 40-41). In examining maximal push and pull forces, Ayoub and McDaniel (1974) restricted participants from flexing the elbows as it was thought that when the elbows were slightly flexed, maximum forces would be limited by shoulder and elbow strength. It has also been suggested that the extended elbows posture stabilizes the elbow which may enable the transmission of higher forces to the hand by the stronger shoulder or trunk muscles (Haslegrave, 1990). This idea was not completely supported in the current study as although the biceps brachii muscle activity decreased with extended elbows, the left erector spinae muscle activity also decreased. However, greater forces were recorded with extended elbows for the pull direction.

Trends in muscle tension that occur with changes to muscle length may similarly apply to the measured EMG. Linear enveloped EMG has often been used as a rough estimator of muscle tension under relatively static contractions (Winter, 2005). At rest length, there are a maximum number of cross bridges between the myosin and actin filaments, a condition conducive for maximum tension (Winter, 2005). As the muscle lengthens, the myosin and actin filaments are pulled apart, decreasing the number of cross bridges and the resulting tension developed (Winter, 2005). As the muscle shortens beyond resting length, the cross bridges overlap and insufficient cross bridges are formed, also reducing the amount of tension that can be developed (Winter, 2005). For the biceps brachii muscle, resting length occurs when the elbows are flexed to 90 degrees with the forearms supinated. Thus, it may be expected that under the extended elbows condition, fewer cross bridges were formed which may have resulted in reduced contribution of the biceps brachii muscle to force generation as observed with the reduced mean %MVE levels. When the elbows were flexed, more cross bridges may have been able to form thus generating greater tension as measured by the increased mean %MVE. It was not expected that mean %MVE levels reported in this study for the biceps brachii would reach 100% MVE as participants were free to adopt elbow postures ≤ 90 degrees, the forearm was either neutral or pronated and the force direction for the push-pull task is different from the muscle's line of action. Thus, the importance of elbow angle is demonstrated by its ability to dictate active tension.

The push direction and flexed elbows provide the greatest force output and are relatively less demanding on the shoulder and trunk. Elbow angle also had a main effect on EMG/force ratios, in which higher ratios were calculated for fully extended elbows compared to flexed elbows for the right triceps brachii, left pectoralis major and left middle deltoid muscles. Higher ratios for the flexed elbow condition arose for the right and left biceps brachii, and right and left middle trapezius muscles. These ratios corroborate the above findings in mean %MVE and demonstrate the greater impact that muscle activation appears to have on EMG/force ratios than force. In addition, Vredenbregt and Rau (1973) found nonlinear relationships between wrist force and EMG for the elbow flexors over a wide range of joint angles. For example, with an elbow angle of 56^{o}, it appeared that less force was developed for a given level of muscle activation, compared to elbow angles of 138^{o} and 162^{o}. Thus, with decreasing joint angles, it may be expected that for a given force level, greater muscle activation may be required. This was evident in the current study for the biceps brachii, in which higher EMG/force ratios were recorded for the flexed elbow conditions. Significant direction by elbow angle effects were demonstrated by most of the muscles examined. A marked interaction effect also existed between direction and elbow angle for all total and weighted average EMG/force ratios where the lowest ratios occurred for the push direction with flexed elbows (Figures 76-81). Thus, from the force data alone, it might be supported that the push direction with flexed elbows generates the greatest force. As the EMG/force ratios support the EMG data, it may be recommended that the push direction with flexed elbows provide a good compromise between force output and muscular demands.

5.7 Left and Right Asymmetry

Even though mean %MVE varied between left and right muscles, left and right hand forces remained equal indicating that some muscles may be compensating. The current study found no significant differences between left and right hand force magnitudes. This agrees with previous research examining two-handed push exertions, in which mean anterior push forces during sustained exertions were not significantly different for left and right hands (Olanrewaju & Haslegrave, 2008). Al-Eisawi et al. (1999) also found no significant difference between left and right hand push and pull forces. However, the results revealed an interaction effect between direction and hand such that within the push direction, right hand forces were generally greater than left hand forces and within the pull direction, left hand forces were significantly greater than right hand forces. The force-bearing leg is often the rearward leg in pushing and the forward leg in pulling, about which the entire body pivots (Ayoub & McDaniel, 1974). Participants often adopted a rearward left foot position in pushing and a forward left foot position in pulling. During a push exertion, if the left foot acted alone, the weight of right lower limb in a forward right foot position may cause the right hip to rotate anteriorly about the spine (vertical axis). As a result, the right hand may resist this moment by increasing the push force. Likewise, when pulling, the weight of the right lower limb in a backward right foot position perhaps causes the left hip to rotate anteriorly about the spine, resulting in the need for an increased left hand force.

In general, the mean %MVE values were greater for the right sided muscles than the left sided muscles. This might be expected, specifically for the weighted average EMG estimates. The sum of the PCSAs for the left muscles was greater than the sum of the PCSAs for the right muscles (Table 9). In addition, the differences between total left and total right EMG values were found to be insignificant. Thus, in dividing the total left EMG estimate by a larger total PCSA, a smaller magnitude would result. All participants in the current study were right-hand dominant. Thus, it may be expected that the right muscles may be larger than the contralateral muscles due to greater use, particularly for the shoulder muscles. As a result, a larger muscle may require less muscle activation than a smaller muscle to achieve the same force output. Upon examining the differences in mean %MVE between the right and left muscles, it was interesting that the sum of the differences in mean %MVE for the biceps brachii and pectoralis major shoulder muscles, which were greater for the left side than the right side, were almost equal to the sum of the differences in mean %MVE between the remaining five muscles recorded (-19.6 %MVE vs. 20.1 %MVE for GMSDHOA model and -12.9 %MVE vs. 13.9 %MVE for the GMSDEA) (Figures 84 and 85). Thus, although hand force magnitudes were similar between right and left hands, this was achieved at the expense of some muscles more than others. Based on the mean %MVE estimates together with the EMG/ratios, it may be suggested that where single hand forces are required or when hand forces are primarily generated by the upper body, such as seated tasks, it may be advised that guidelines be developed using strength capabilities of the non-dominant hand.

5.8 Study Limitations

The effects of gender, mass, stature, direction, handle height, handle orientation and elbow angle on maximal hand force magnitudes, shoulder and low back moments, mean %MVE, EMG/force ratios and foot stance are specific to the population and conditions used in the current study. Participation was voluntary and the participant pool was limited to university-aged students with little to no manual materials handling experience. A larger, more diverse working population may provide greater insight into the influence of examined factors. However, participants were recruited to cover the male and female North American 5th to 95th percentiles with respect to stature, upon which many design thresholds are based.

The use of static exertions was used to facilitate the use of EMG, but may not be representative of dynamic task situations. For initial forces, static exertions have been used to estimate dynamic tasks. Hoozemans et al. (1998) cited several studies that found shoulder complaints in relation to isometric loading of the shoulder muscles during pushing and pulling tasks. Previous literature have also shown that the results obtained using a stationary bar and moveable cart were comparable in terms of force exertion and body posture (de Looze et al., 2000). In addition, it was found that dynamic tasks may be simulated with static forces when the applied tasks involve slow movements (Grant & Habes, 1997, Resnick & Chaffin, 1995). Thus, it appears that the static exertions used in the current study may be applicable to maximal push and pull tasks performed outside of the laboratory.

The influence of the examined factors on the dependent measures investigated may only be applied to tasks where maximal forces are required. Maximal exertions were examined as it is often that the initial force required to accelerate an object is used to determine the limits of maximal acceptable forces and load weights, as it is recognized as the largest force requirement in pushing and pulling (Jung et al., 2005). In general, the risk of developing musculoskeletal disorders, particularly as a result of overexertion injuries, increases when the exerted push or pull forces approximate the maximum force generating capability or maximum acceptable forces. Thus, determining maximal forces and the associated muscle activity under maximal force conditions is important when trying to establish strength profiles design purposes.

Upon examining maximal exertions, fatigue may have been a limiting factor due to the number of trials performed. However, several precautions were taken to reduce the effects of fatigue on the results. Elbow angles were only examined under a subset of conditions to reduce the total number of maximal contractions performed by the participants. A minimum rest period of 2 minutes was given between all MVE and MVIC trials to avoid muscular fatigue as suggested by Chaffin (1975) and Mathiassen et al. (1995). In addition, fifteen minutes of rest was provided between the set of MVE and MVIC trials to further mitigate the effects of any localized muscle fatigue. Moreover, a previous study that made use of 29 experimental trials found that fatigue was not a confounding variable, in which a similar protocol to the current study was used (Kelly, Kadrmas & Speer, 1996).

The collection of EMG data presents several limitations. As EMG is sensitive to posture, the lack of control over flexed elbow angles may have overestimated or underestimated mean %MVE values reported. Despite having right foot locations marked on the floor, left foot positions varied between participants and within participant repetitions. As a result, shoulder and L5/S1 joint angles also varied, even after controlling for arm posture under the extended elbow conditions. In testing for the effect of elbow angle, elbow angles were self-selected by participants for the "$\leq 90^{o}$" conditions. Based on qualitative observations, pulling had greater elbow angles than pushing when elbows angles were $\leq 90^{o}$. In calculating joint moments, wrist, elbow, shoulder and L5/S1 joint angles were accounted for in the model, however, the remaining force- and EMG-based estimates were averaged over repetitions. Postural variations among the participants due to individual preference as well as constraints of the experimental conditions indicate that the results cannot be interpreted based on specific body postures, but rather on the experimental conditions examined in this current study. Averaging would generally have the effect of underestimating magnitudes. With respect to averaging forces, it might be thought of as a conservative approach to setting strength limits. The potential underestimation of mean %MVE estimates may imply that the muscles are active to a lesser extent than in reality. Thus guidelines based solely on mean %MVE are not recommended and may lead to and increased onset of muscle fatigue.

In normalizing EMG obtained from the trials, it was assumed that the MVE postures chosen would elicit 100% muscle activity and that participants were sufficiently motivated to elicit their maximal muscle activation. Three repetitions of the MVE trials were performed and peak EMG amplitudes were used for normalization. A lack of motivation during the MVE trials may have led to overestimation of mean %MVE during the experimental conditions. From a safety perspective, overestimations may be viewed as being more conservative, but could also identify tasks as being more demanding and more at risk for leading to MSDs than in reality. All participants were verbally encouraged throughout all trials. It was further assumed that participants were equally motivated during the MVE trials and the experimental conditions such that differences in %MVE due to lack of motivation would be nullified upon normalization. Only a small number of the trials from the test conditions resulted in some muscles exceeding 100 %MVE indicating that participants were highly motivated, particularly for the MVE trials.

The populations from which the PCSAs were acquired were different from the population used in the current study. PCSAs used for the calculation of weighted average EMG were obtained from studies that made use of living male participants for the trunk muscles and cadaver male participants for the shoulder muscles. The use of cadavers to estimate PCSAs may have the effect of underestimating the areas of the male participants in the current study and underestimating or overestimating the areas of the female participants. Although, muscles sizes between males and females may be proportional, the use of male PCSA may have masked gender differences. However, gender differences persisted in EMG-based estimates. Thus, there is some confidence that differences in the weighted average estimates are relative, but may underestimate female EMG magnitudes.

The ability to draw conclusions regarding all shoulder and trunk muscle groups was limited by the inability to measure all muscles. A total of seven superficial bilateral shoulder and trunk muscles were examined. The number of muscles monitored was partially restricted due to the number of leads that could be connected to the transmitter. In addition, the use of surface electrodes have the increased potential for cross-talk and are not able to access the deep muscles, but they are noninvasive to the participants and were relatively inexpensive considering the number of muscles being measured and number of participants being tested. Although only seven bilateral muscles were monitored, these muscles were thought to be representative of the muscles involved in the push and pull exertions examined in this study.

The assumption that participants maintained the same postures between repetitions may have been a limiting factor. Other studies investigating push and pull forces that have made use of nonstandardized postures have found acceptable repeatability with respect to forces and postures. Chaffin & Andres (1983) investigated force exertion while standing with one foot in front of the other and found that mean angles varied by an average of about 2 degrees between repetitions while foot placements varied by about 2 cm. It was further noted that the mean forces were not significantly different (at $p < 0.05$). Daams (1993) also found that the use of a functional posture (average position assumed by participants in the free posture conditions from a preliminary study) yielded an average reproducibility (correlation coefficient r between the forces exerted between repetitions) of 0.90. Daams also established that identical handle heights are a necessary condition for repeatability of forces. As the current study utilized fixed heights, it is possible that forces and postures may be less free to vary between repetitions. Participants were further restricted by the use of two hands and fixed right foot positions, which further constrains joint range of motion.

Additional equipment artifact may have limited the ability for participants to assume natural postures or limited their ability to exert their greatest for due to the need to support the extra weight of the equipment. Precautions were taken to secure the EMG leads together on one side of the body while providing sufficient slack in the wires so as not to restrict normal range of motion. The battery pack and transmitters were wireless and thus the participants were not tethered to the recording and storage devices. Markers were light and so as not to hamper movement due to excessive load and they were reflective such that movements could be captured using cameras outside the task volume. The selected equipment and setup were thought to promote more natural working conditions given the requirements for testing.

5.9 Suggestions for Future Investigations

The current study was able to demonstrate significant effects of the gender, mass, stature, direction, handle height, handle orientation and elbow angle on most of the dependent variables examined. Further studies should include other age groups as well as participants with greater manual materials handling experience to determine how well trends from the current study can be applied to the working population. Future research could also be examined to include other shoulder muscles, forearm, wrist and lower limb muscles as these muscles may provide greater insight into some the observed findings. It may then be able to make more definitive conclusions on the particular muscle groups examined. As restrictions on lifting the heels off the floor were imposed in the current study, it may be interesting to examine the effects of lifting the heels to aid in pushing and pulling through increased contributions of the lower limbs. With the use of static contractions, the application of indwelling electrodes may be possible and may enable greater understanding of the role of the deeper muscles. Furthermore, as only two handle heights were examined, more handle heights could be examined to determine if the measured variables could be predicted from interpolations, which could provide a useful tool in ergonomics applications. This study was able to demonstrate the influence of several factors on force-, moment- and EMG-based measures, but identifies the need for continued research into push-pull tasks.

The results of this study can also be examined with the use of biomechanical models. As only net moments and sagittal plane moments were examined, the use of a biomechanical model could further aid in resolving internal forces to provide insight into the mechanical stresses at the joints (i.e., compressive and shear forces at the intervertebral discs). The current investigation, which incorporated hand force magnitudes, muscle activation and whole-body kinematics for two-handed push and pull forces provide a large set of data which could be used to validate models used to predict muscle activation based on anthropometric, postural and hand force data. It is important for models to be validated for both males and females due to their integrated use in industry for job assessment and design purposes.

5.10 Relevance to Ergonomics and Work Design

The results of this study demonstrate which factors or combinations of factors had the greatest influence on force-, moment- and EMG-based estimates. The quantification of the effects of the potential risk factors examined for pushing and pulling tasks is pertinent to implementing appropriate ergonomics interventions. Gender, mass, stature, direction, handle height, handle orientation and elbow angle are all important factors that should be considered in task design or modification. For example, the current study found that the larger total pull forces of 282.4 N (28.8 kg) at the 100 cm handle height resulted in similar extensor moments at the L5/S1 joint as total pull forces of 170.6 N (17.4 kg) at the 150 cm handle height. EMG/force ratios were also 69 percent larger for the latter condition. This suggests that for a given mechanical load, changing the working conditions, say from a lower to higher work height, might require a reduction in the maximum force exertion requirement, or when more favourable working conditions are implemented, greater maximum acceptable forces and reduced muscular demands may result. Furthermore, the significant effects of gender emphasize the importance of providing strength profiles of both the male and female populations, to provide guidelines in tasks where strength may be a limiting factor.

The control of particular aspects of the experimental conditions for empirical testing often makes it difficult to apply the results to workplace settings. The decision to examine fixed handle heights, rather than heights with reference to body landmarks, is justified in that the measured parameters are more applicable to tasks than postures. In addition, the use of participants spanning the heights of the male and female North American population facilitates greater generalizations of the results. Thus, the results may assist in making more accurate estimations of hand force magnitudes based on workplace conditions, to the extent of the factors examined, and thus make better predictions of muscular demands and internal shoulder and trunk loads. The end goal is to provide engineers and ergonomists the means to better assess and quantify push and pull tasks to enable evidence-based recommendations and preventative measures to reduce the incidence of shoulder and low back injuries, thereby decreasing the associated worker absences and health care costs.

6.0 Conclusions

The purpose of this study was to quantitatively evaluate the influence of gender, direction, handle height, handle orientation and elbow angle on force-, moment- and EMG-based exposure estimates. The following conclusions may be made when considering two-handed maximal voluntary push and pull exertions:

- Direction had the greatest effect on most outcome measures.
- Push exertions should be recommended over the pull exertions as evidenced by greater hand force magnitudes, reduced L5/S1 extensor moments, reduced shoulder moments (for the 150 cm handle height) and reduced shoulder and trunk muscular demands.
- The 100 cm handle height resulted in greater hand force magnitudes and reduced shoulder and trunk muscular demands for a given force level compared to the 150 cm handle height, however the 100 cm handle height was accompanied by greater sagittal plane moments.
- Reiterating previous literature, females, on average, had 67% of the maximal hand forces of males. Females must further activate their muscles to a greater extent than males for the same absolute force level, as demonstrated by EMG/force ratios.
- Larger body masses were associated with greater hand force capability and greater resultant moments.
- The flexed elbows condition with force exerted in the push direction produced the greatest hand force magnitudes with reduce expense to shoulder and trunk muscular demands. Only moment magnitudes at the right shoulder were significantly

influenced by elbow angle.

- Either vertically or horizontally oriented handles may be implemented without compromising force performance; however, the horizontal handle orientation caused greater resultant moments at all joints. Minimal effects of handle orientation were found for mean %MVE estimates.

Historically, past research on push and pull tasks have focused on the mechanical loading of the low back, resulting in inadequate knowledge of the upper extremities. Previous literature has extensively made use of male participants, though design thresholds often support the use of female populations as the reference population. Furthermore, the associated muscle activation patterns for push and pull tasks have been quite limited, especially for the female population. In collecting male and female whole-body kinematic data, bilateral hand forces and the corresponding EMG of seven bilateral shoulder and trunk muscles, this study was able to successfully evaluate the specific tissue loads on the shoulder and trunk musculature associated with the measured hand forces and calculated shoulder and low back moments for different task conditions. The results further contribute to the database of female and male isometric hand force capabilities for two-handed anterior-posterior push and pull exertions. With the use of EMG/force ratios, the study was further able to determine which conditions are able to maximize hand force capability while minimizing the muscular demands to the shoulder and trunk musculature examined. The results of this study have important ergonomics implications for evaluating, designing or modifying workstations, tasks or equipment towards improved task performance and the prevention of musculoskeletal complaints.

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APPENDICES

Appendix A: Effects of Direction and Handle Height on Individual Muscle LSM %MVE

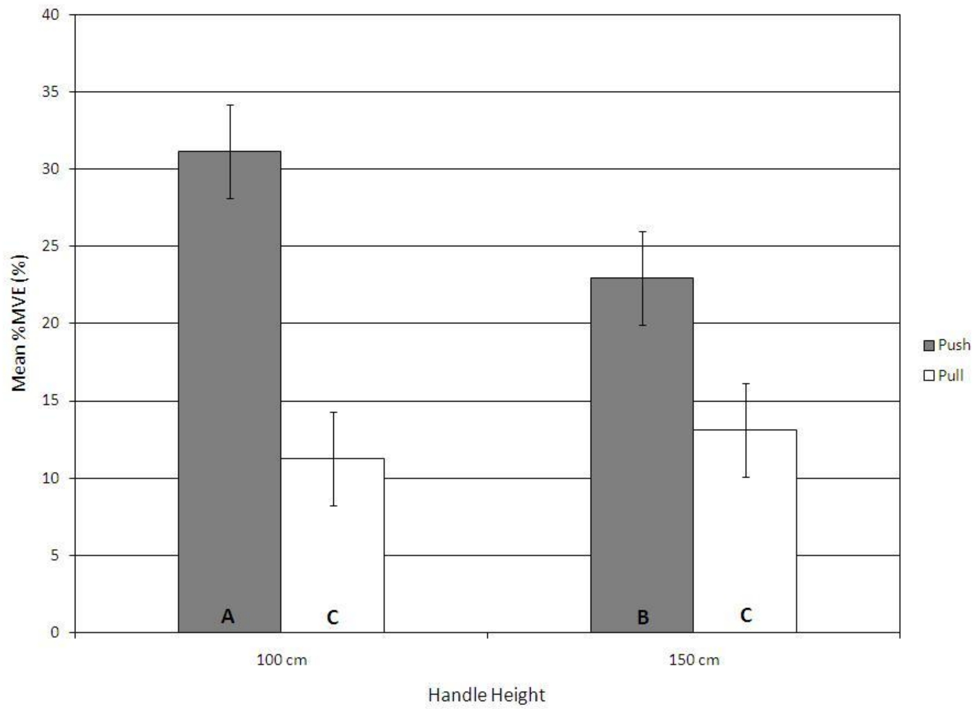


Figure A1: Effect of direction and handle height on LSM %MVE for the right biceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

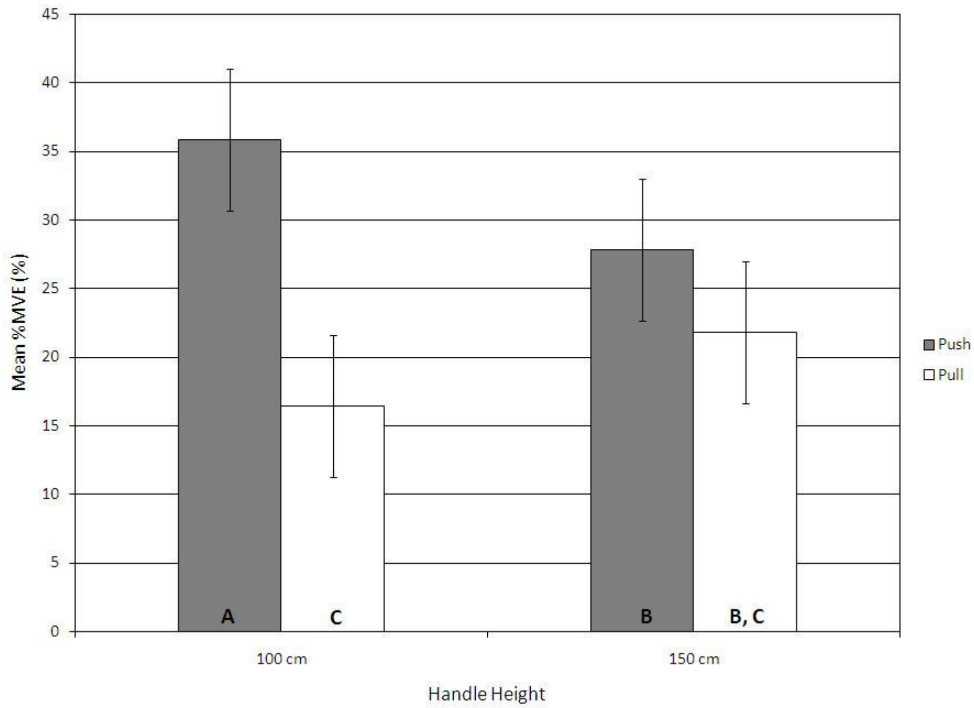


Figure A2: Effect of direction and handle height on LSM %MVE for the left biceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

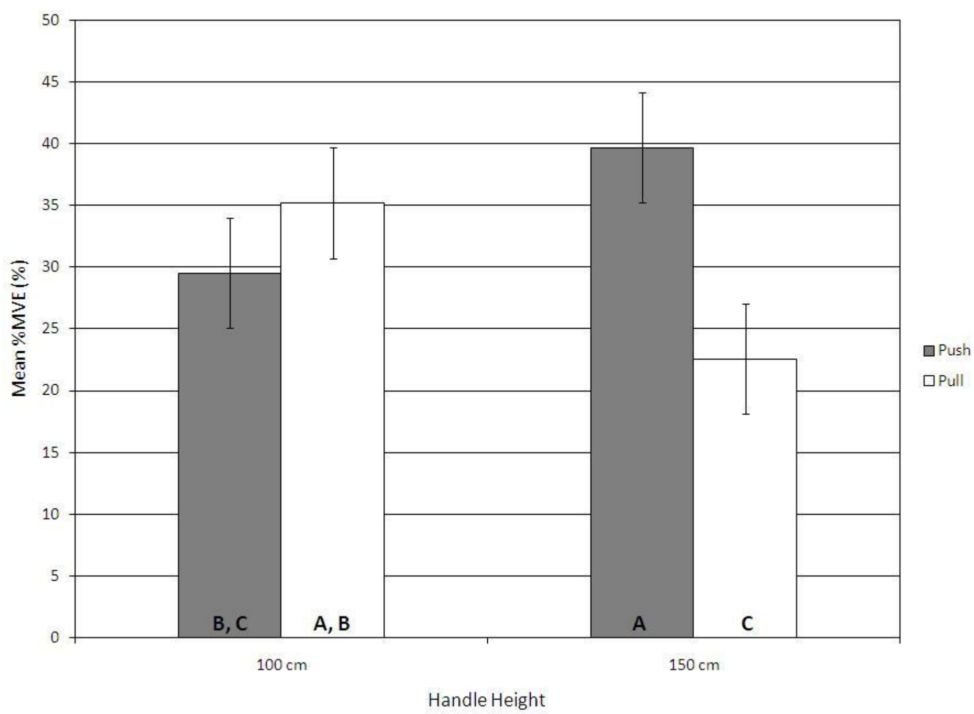


Figure A3: Effect of direction and handle height on LSM %MVE for the right triceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

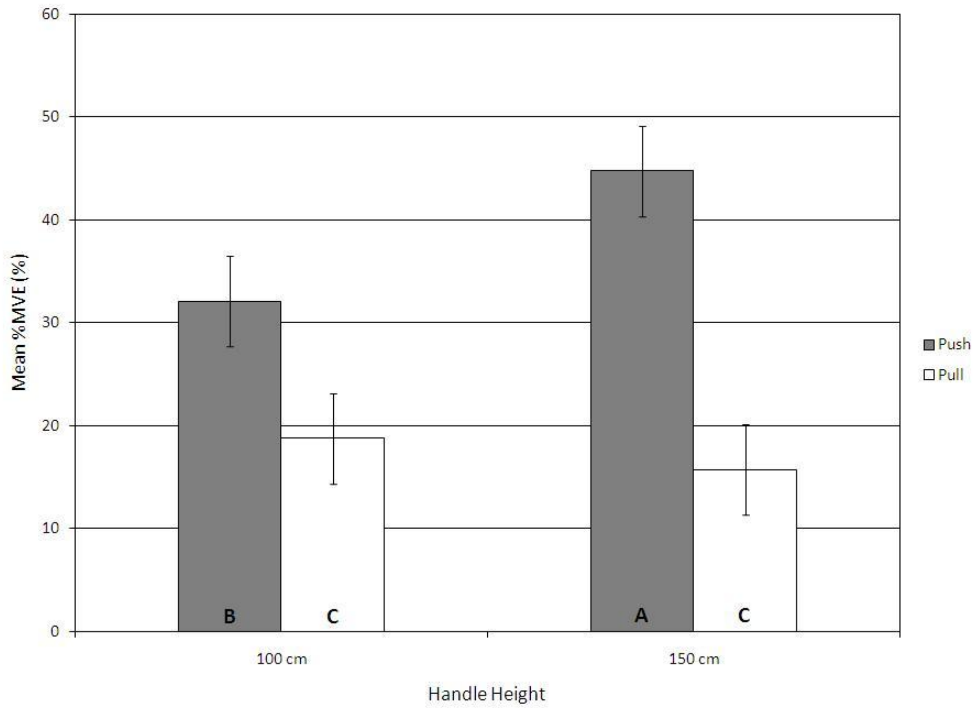


Figure A4: Effect of direction and handle height on LSM %MVE for the left triceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

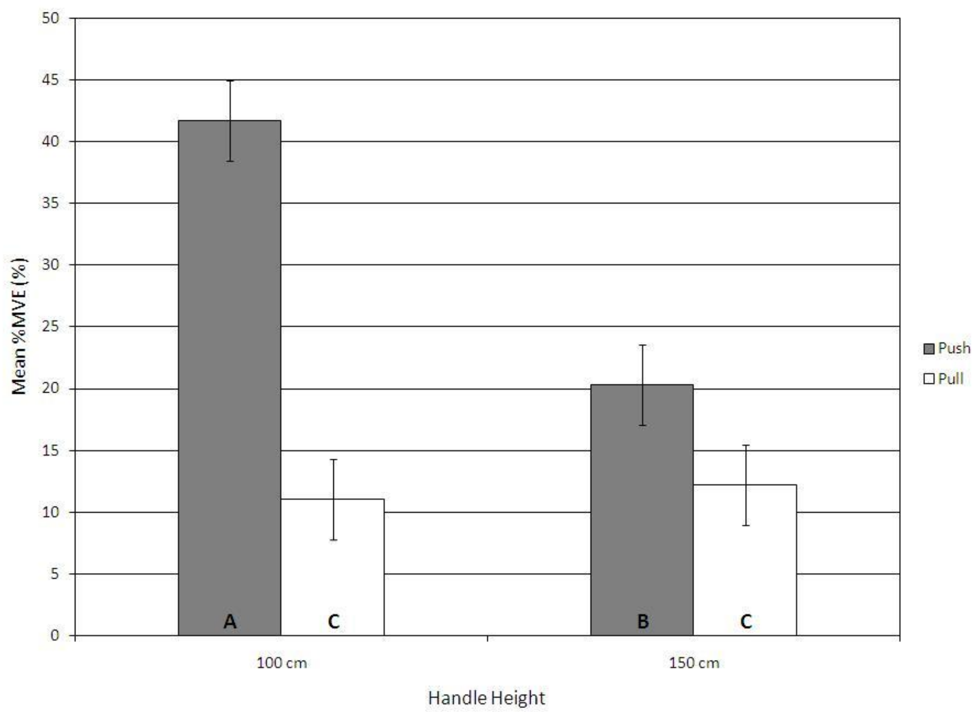


Figure A5: Effect of direction and handle height on LSM %MVE for the right pectoralis major muscle. Letters indicate significantly different direction by handle height interactions.

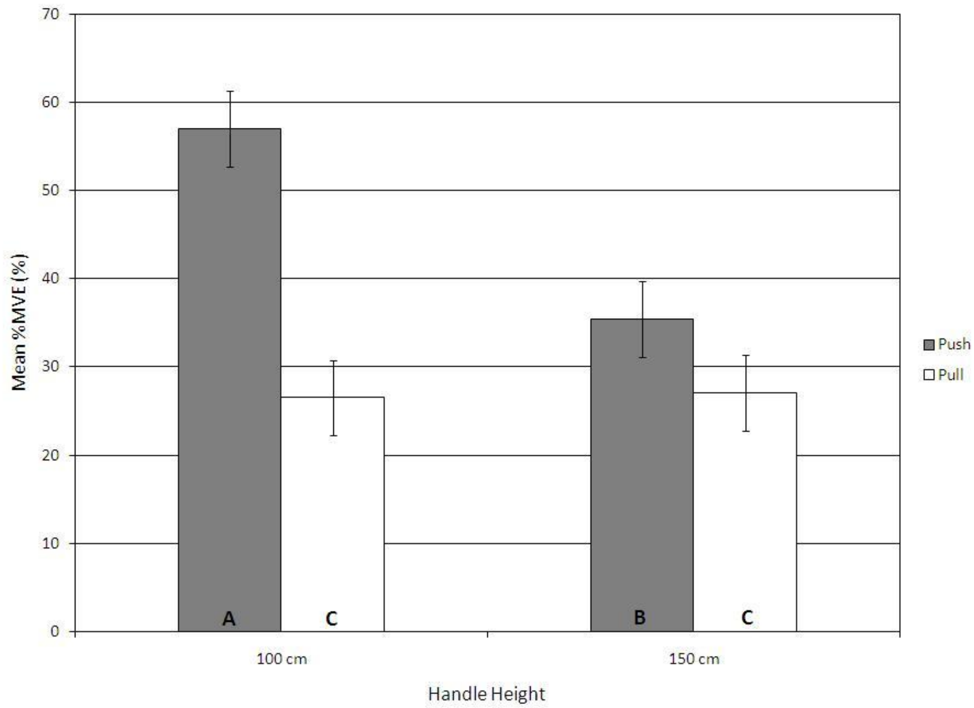


Figure A6: Effect of direction and handle height on LSM %MVE for the left pectoralis major muscle. Letters indicate significantly different direction by handle height interactions.

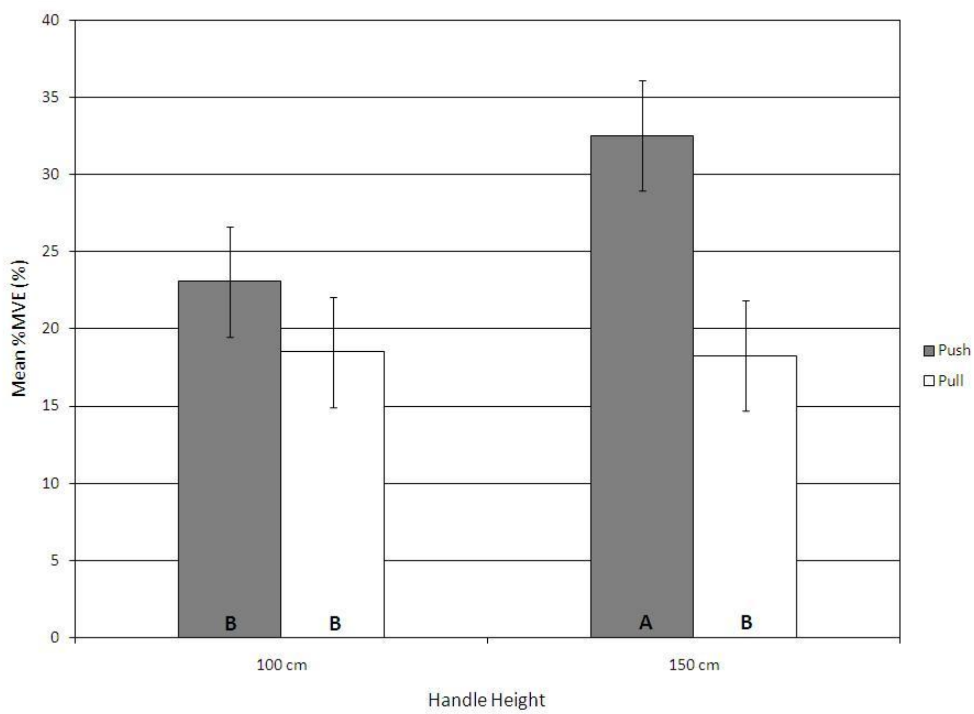


Figure A7: Effect of direction and handle height on LSM %MVE for the right middle trapezius muscle. Letters indicate significantly different direction by handle height interactions.

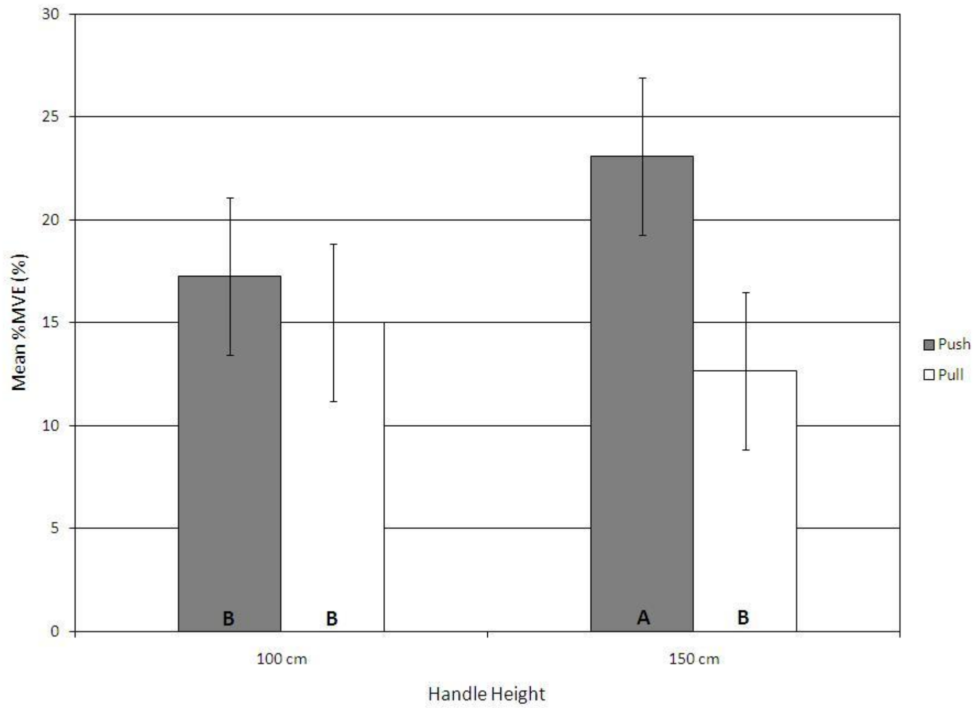


Figure A8: Effect of direction and handle height on LSM %MVE for the left middle trapezius muscle. Letters indicate significantly different direction by handle height interactions.

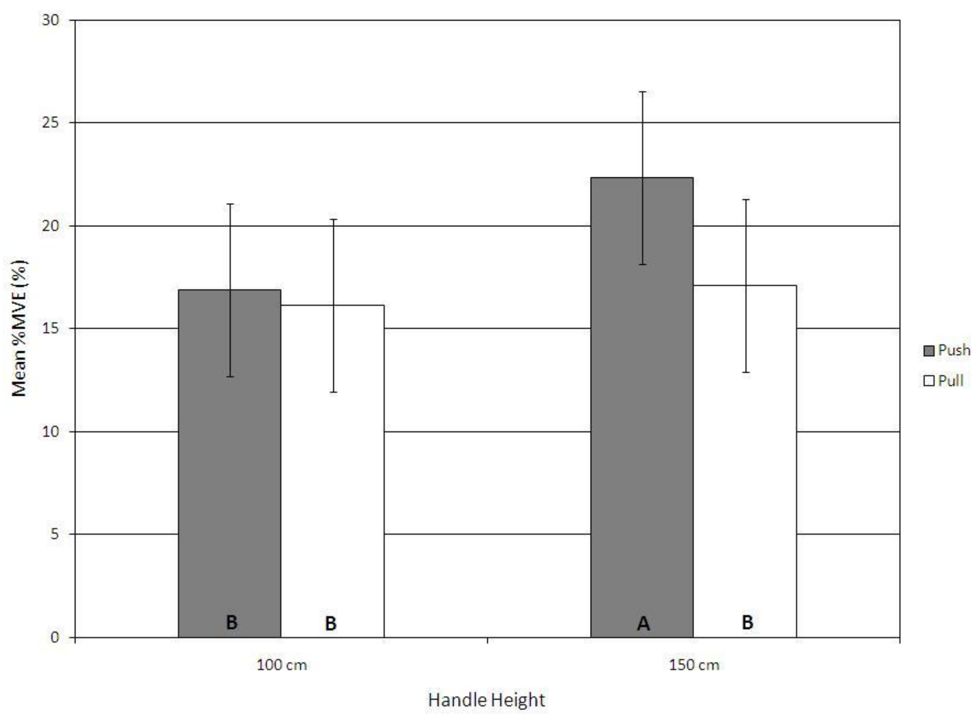


Figure A9: Effect of direction and handle height on LSM %MVE for the right rectus abdominis muscle. Letters indicate significantly different direction by handle height interactions.

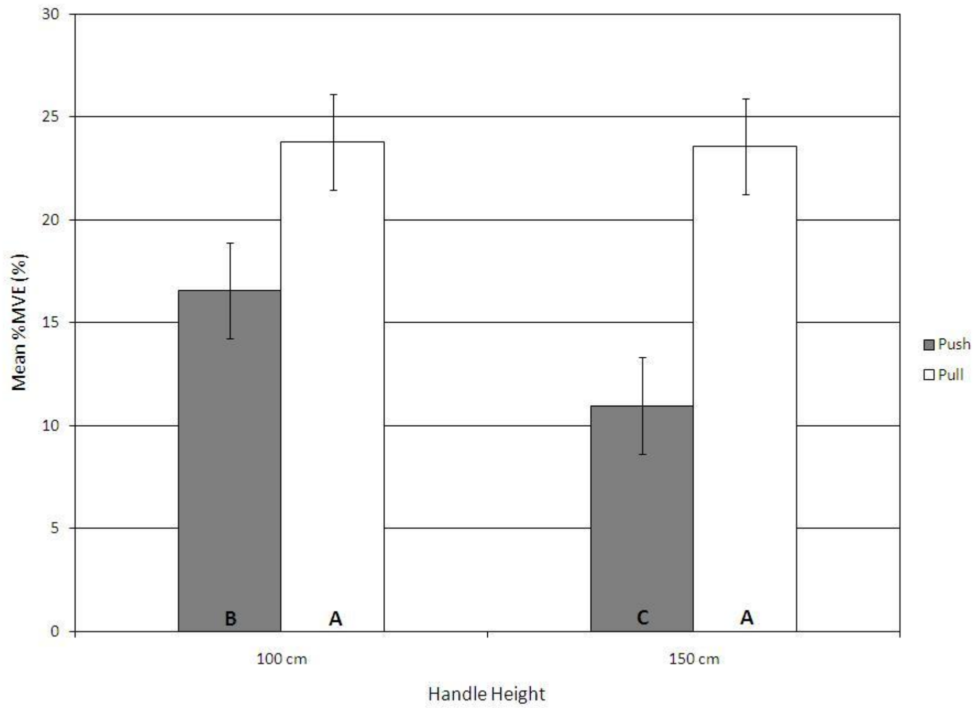


Figure A10: Effect of direction and handle height on LSM %MVE for the right erector spinae muscle. Letters indicate significantly different direction by handle height interactions.

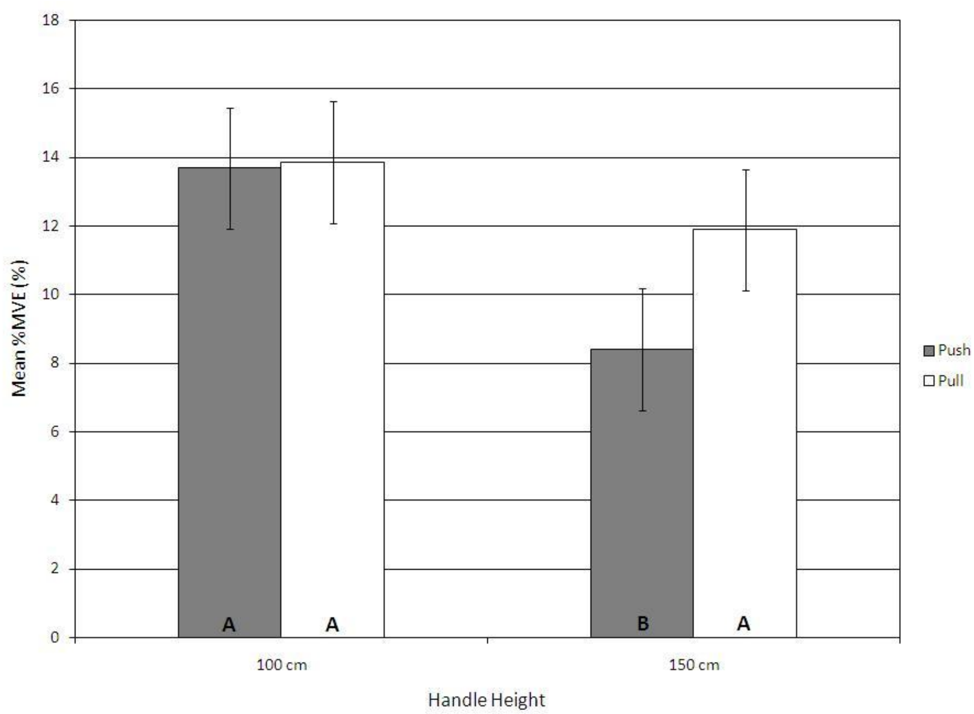


Figure A11: Effect of direction and handle height on LSM %MVE for the left erector spinae muscle. Letters indicate significantly different direction by handle height interactions.

Appendix B: Effects of Direction and Elbow Angle on Individual Muscle LSM %MVE

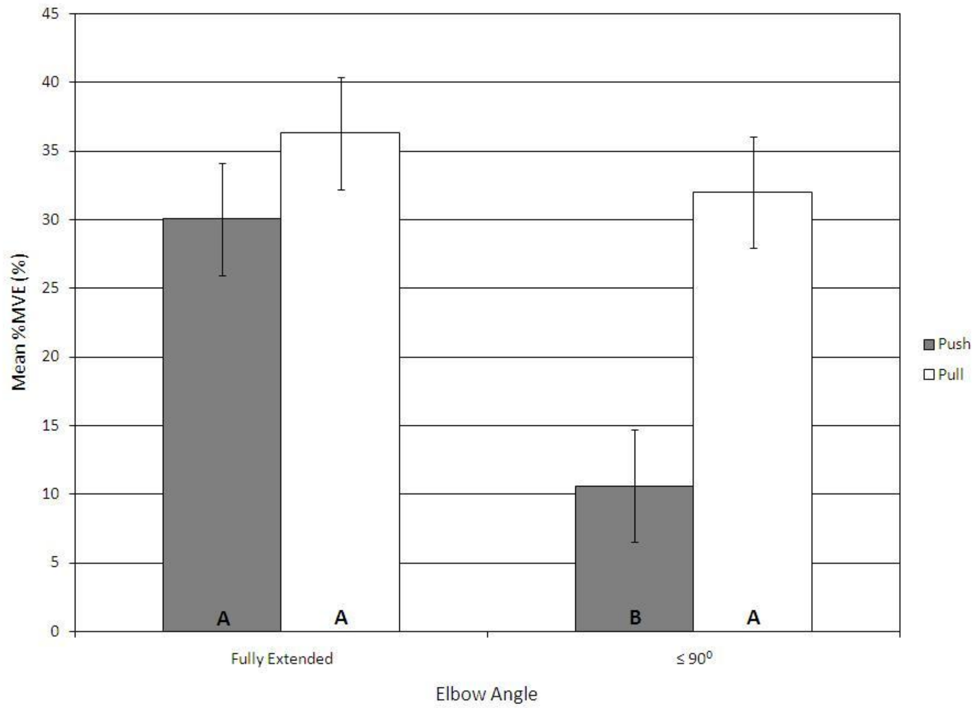


Figure B1: Effect of direction and elbow angle on LSM %MVE for the right triceps brachii muscle. Letters indicate significantly different direction by elbow angle interactions.

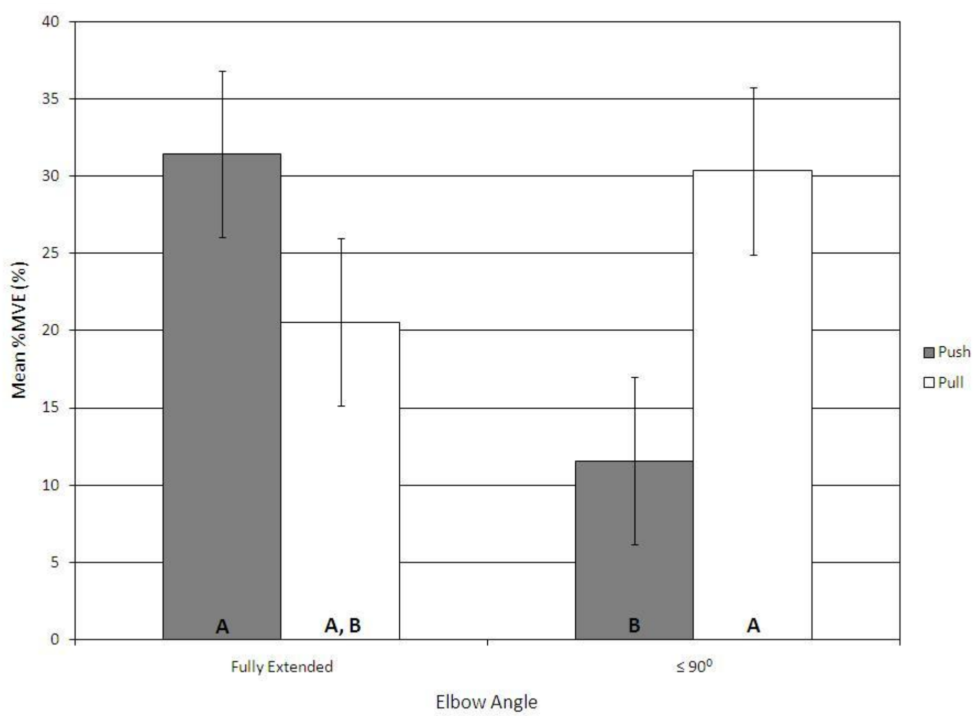


Figure B2: Effect of direction and elbow angle on LSM %MVE for the left triceps brachii muscle. Letters indicate significantly different direction by elbow angle interactions.

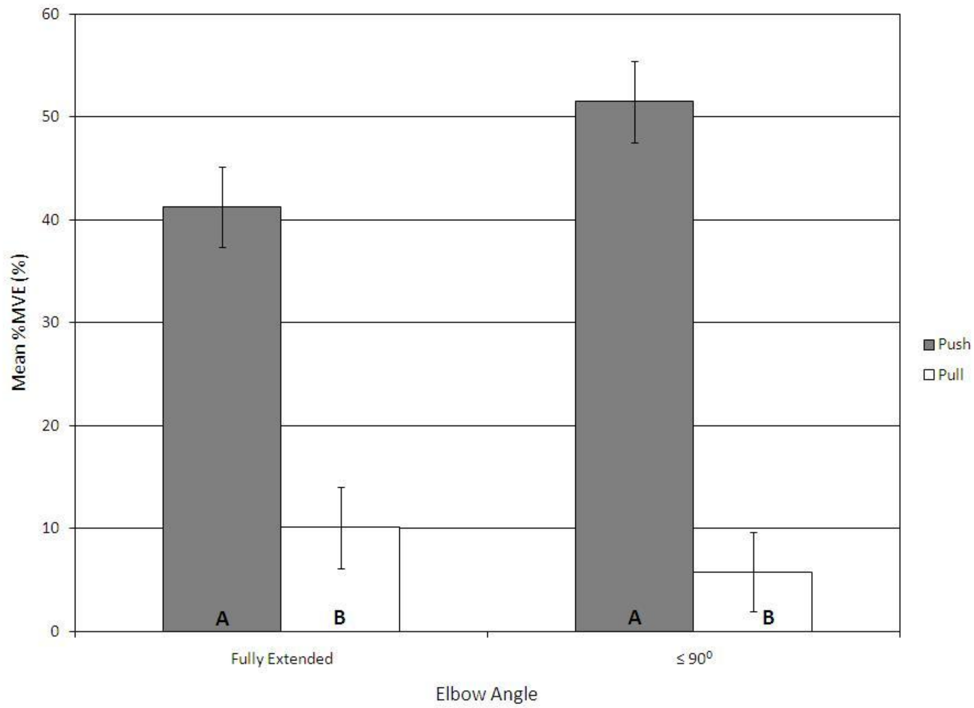


Figure B3: Effect of direction and elbow angle on LSM %MVE for the right pectoralis major muscle. Letters indicate significantly different direction by elbow angle interactions.

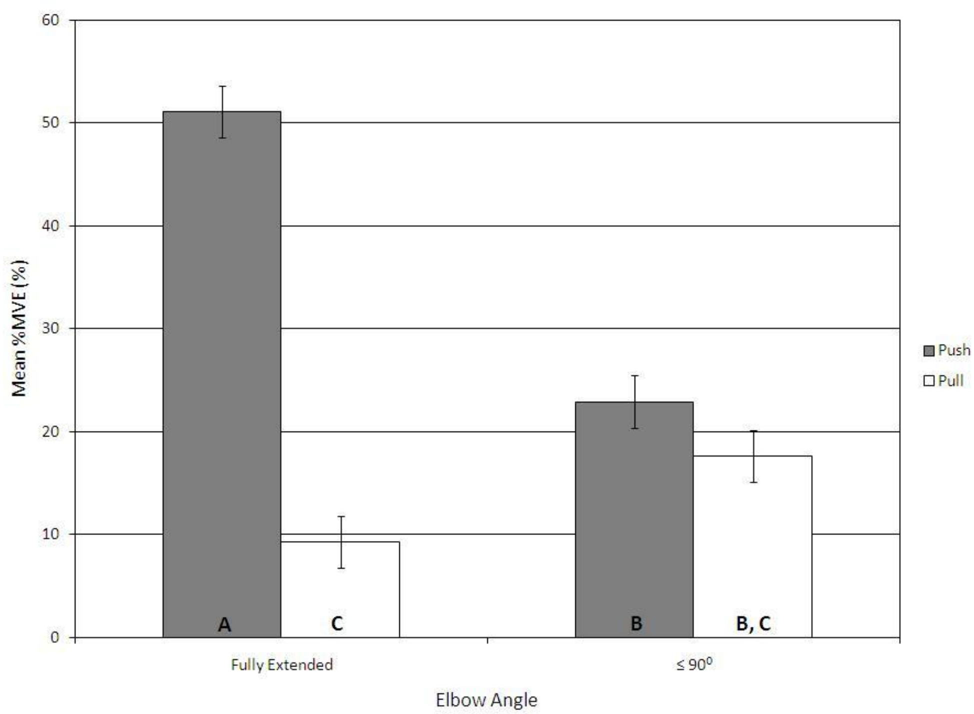


Figure B4: Effect of direction and elbow angle on LSM %MVE for the right middle deltoid muscle. Letters indicate significantly different direction by elbow angle interactions.

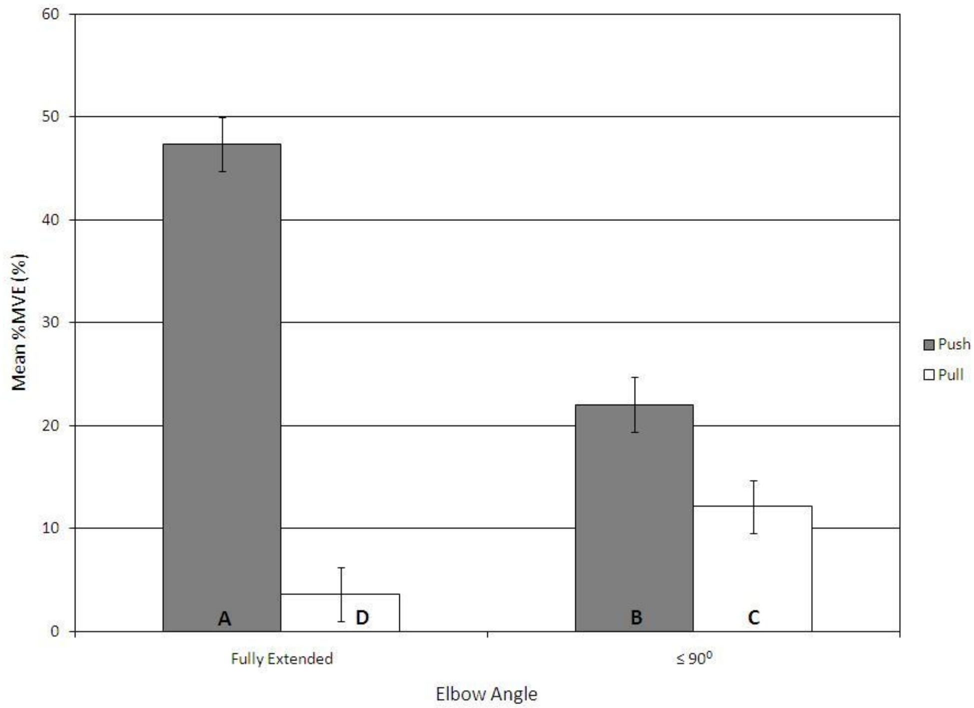


Figure B5: Effect of direction and elbow angle on LSM %MVE for the left middle deltoid muscle. Letters indicate significantly different direction by elbow angle interactions.

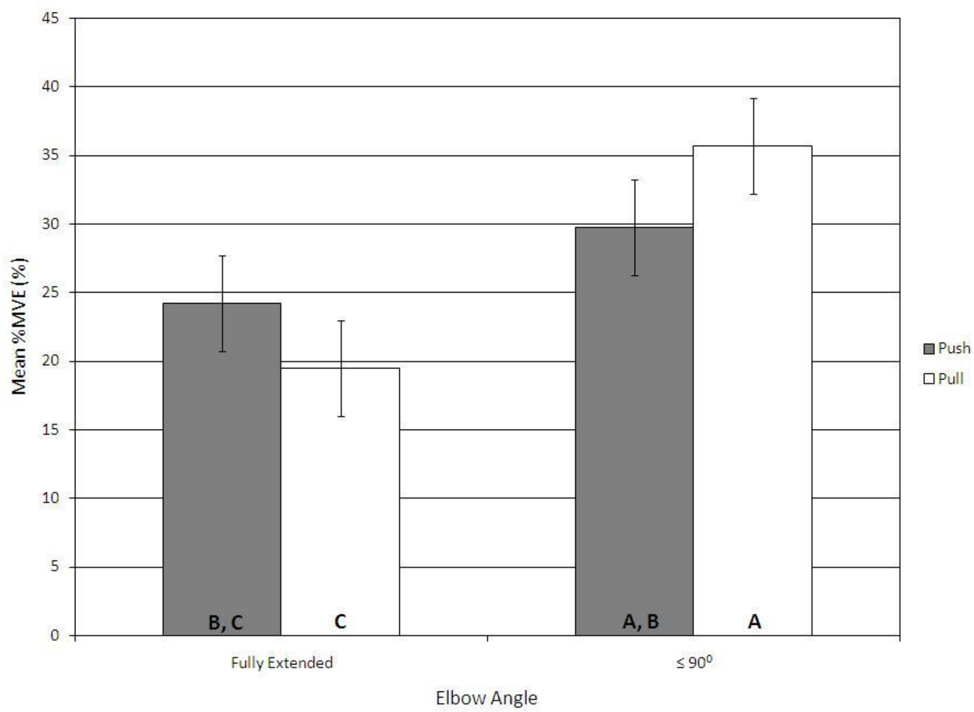


Figure B6: Effect of direction and elbow angle on LSM %MVE for the right middle trapezius muscle. Letters indicate significantly different direction by elbow angle interactions.

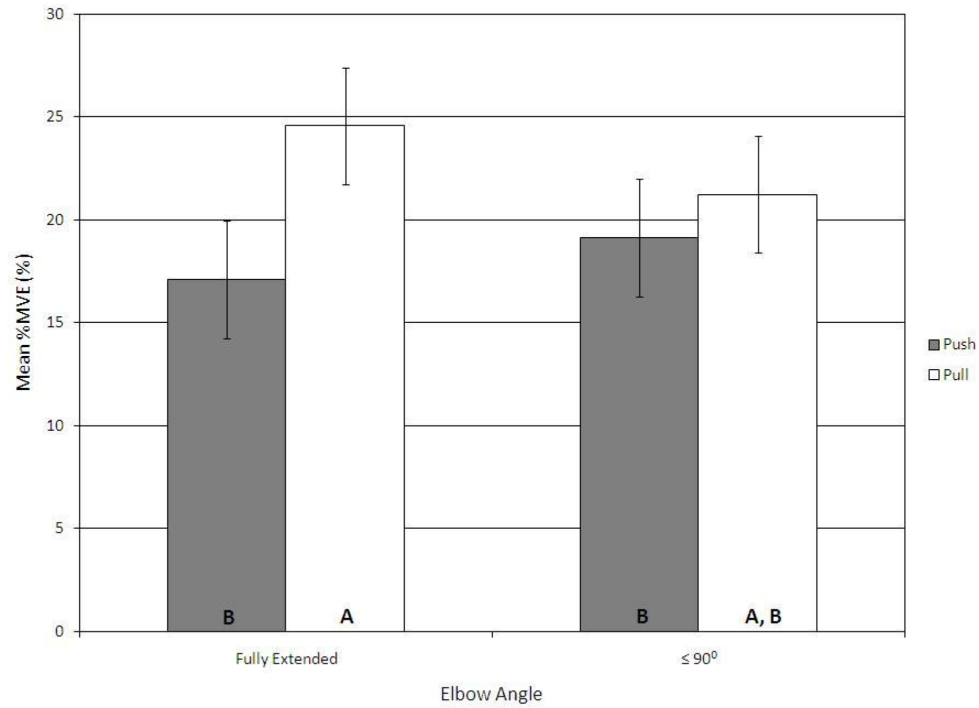


Figure B7: Effect of direction and elbow angle on LSM %MVE for the right erector spinae muscle. Letters indicate significantly different direction by elbow angle interactions.

Appendix C: Effects of Direction and Handle Height on Individual Muscle LSM EMG/force Ratios

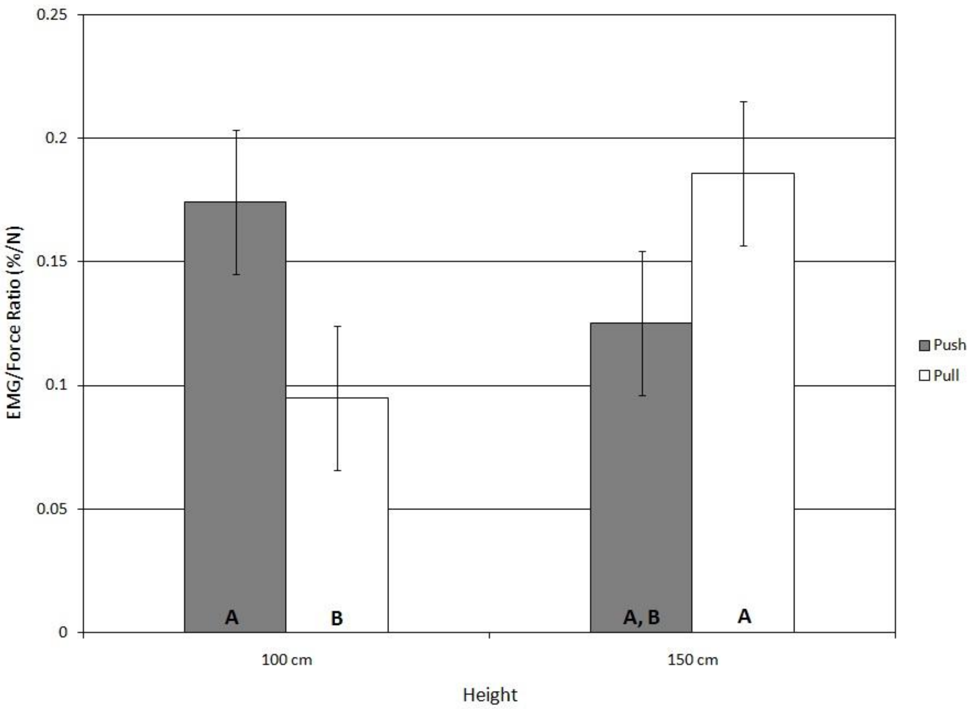


Figure C1: Effect of direction and handle height on LSM EMG/force ratios for the right biceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

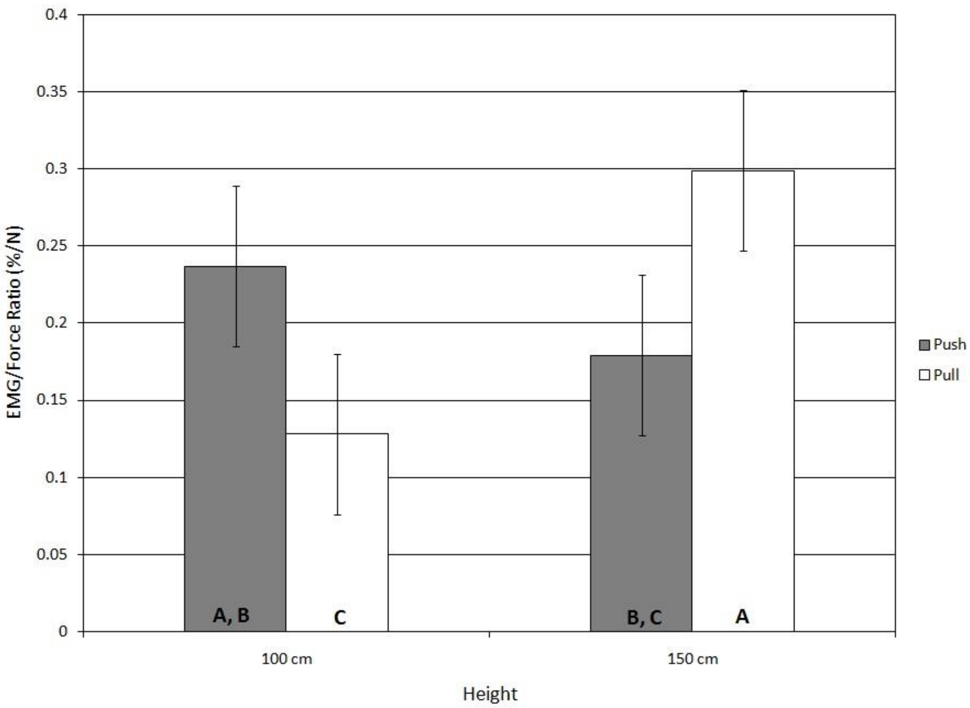


Figure C2: Effect of direction and handle height on LSM EMG/force ratios for the left biceps brachii muscle. Letters indicate significantly different direction by handle height interactions.

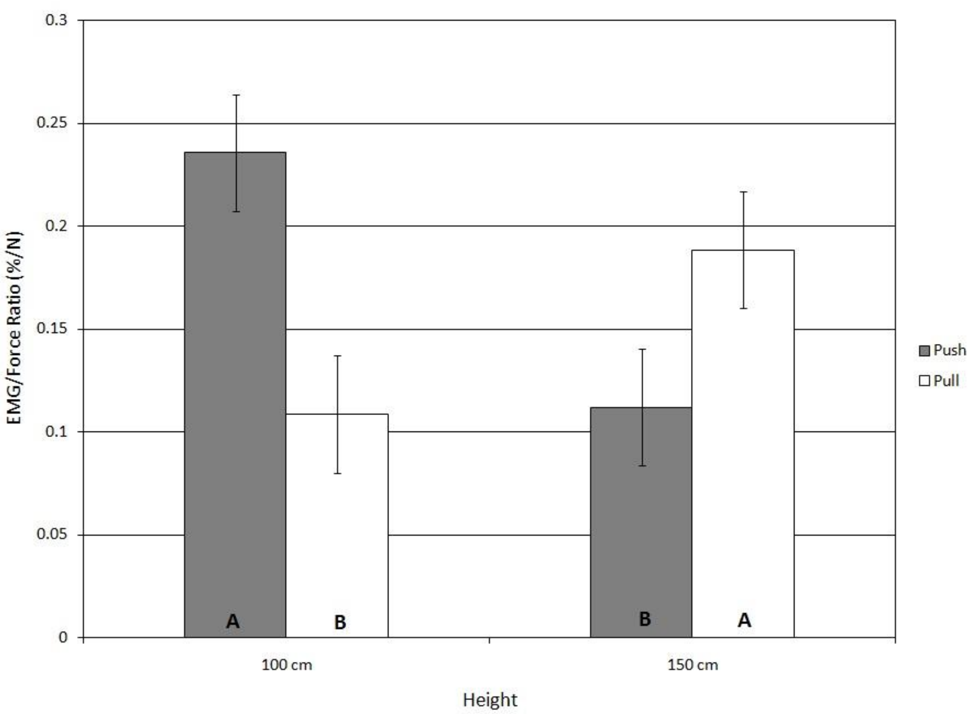


Figure C3: Effect of direction and handle height on LSM EMG/force ratios for the right pectoralis major muscle. Letters indicate significantly different direction by handle height interactions.

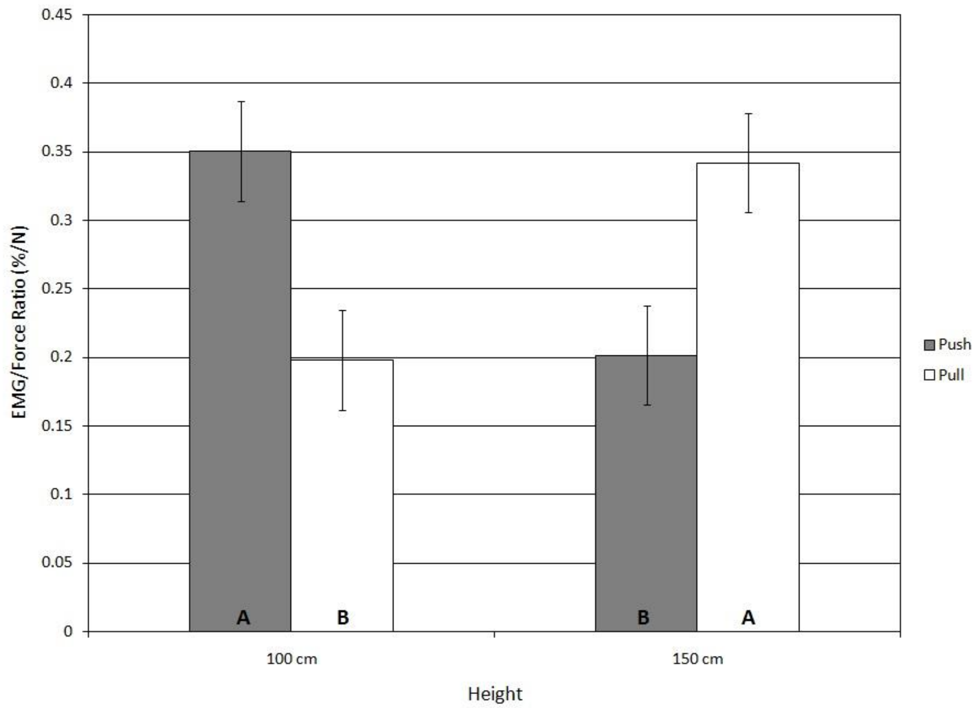


Figure C4: Effect of direction and handle height on LSM EMG/force ratios for the left pectoralis major muscle. Letters indicate significantly different direction by handle height interactions.

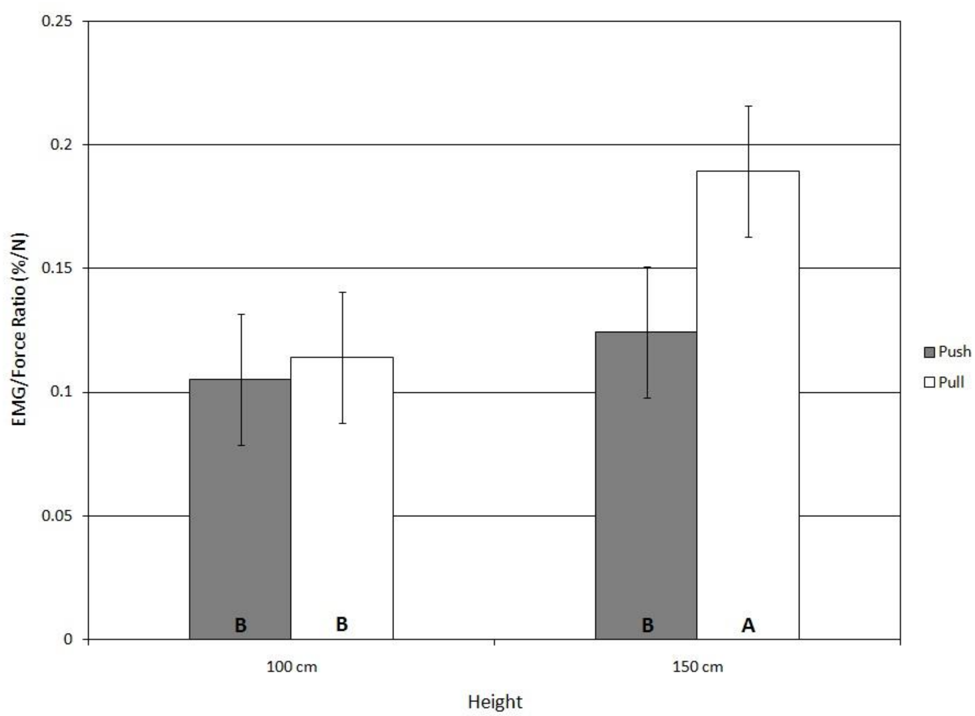


Figure C5: Effect of direction and handle height on LSM EMG/force ratios for the left middle trapezius muscle. Letters indicate significantly different direction by handle height interactions.

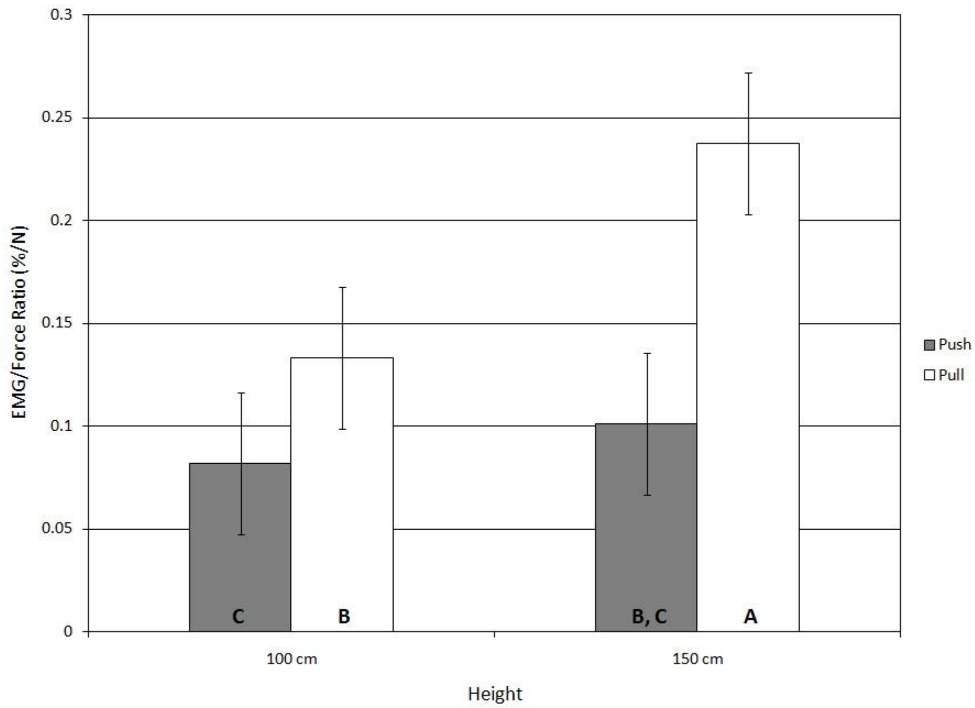


Figure C6: Effect of direction and handle height on LSM EMG/force ratios for the right rectus abdominis muscle. Letters indicate significantly different direction by handle height interactions.

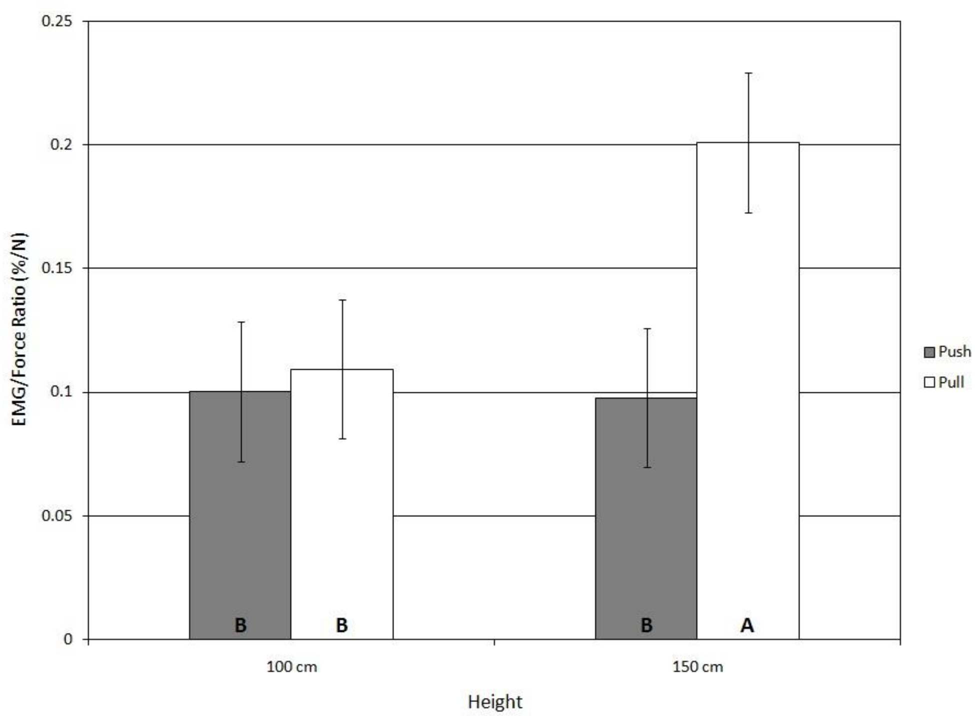


Figure C7: Effect of direction and handle height on LSM EMG/force ratios for the left rectus abdominis muscle. Letters indicate significantly different direction by handle height interactions.

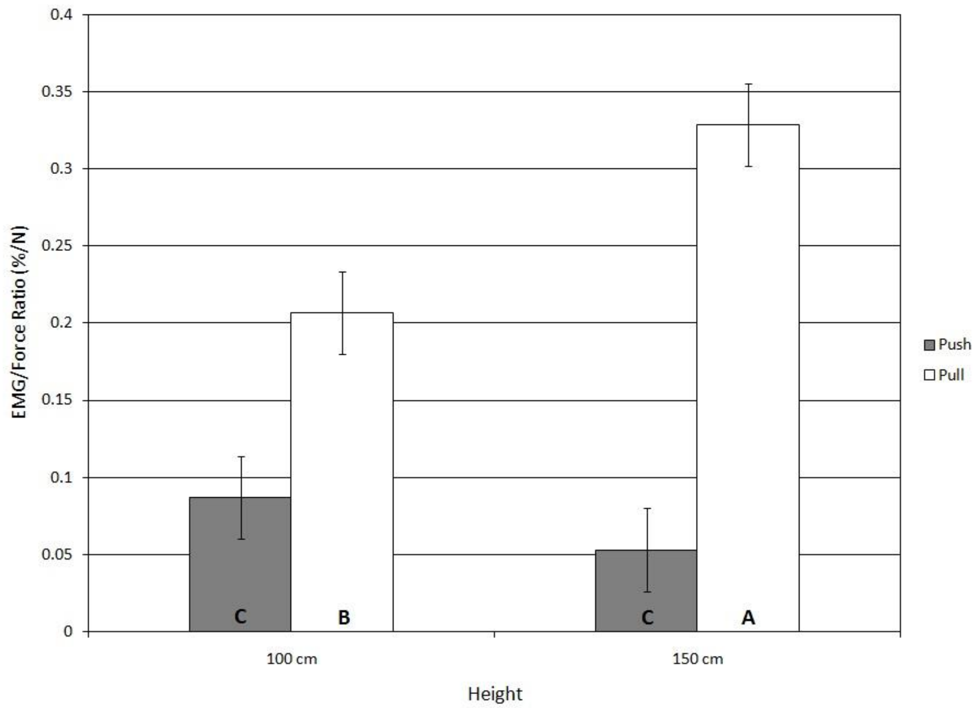


Figure C8: Effect of direction and handle height on LSM EMG/force ratios for the right erector spinae muscle. Letters indicate significantly different direction by handle height interactions.

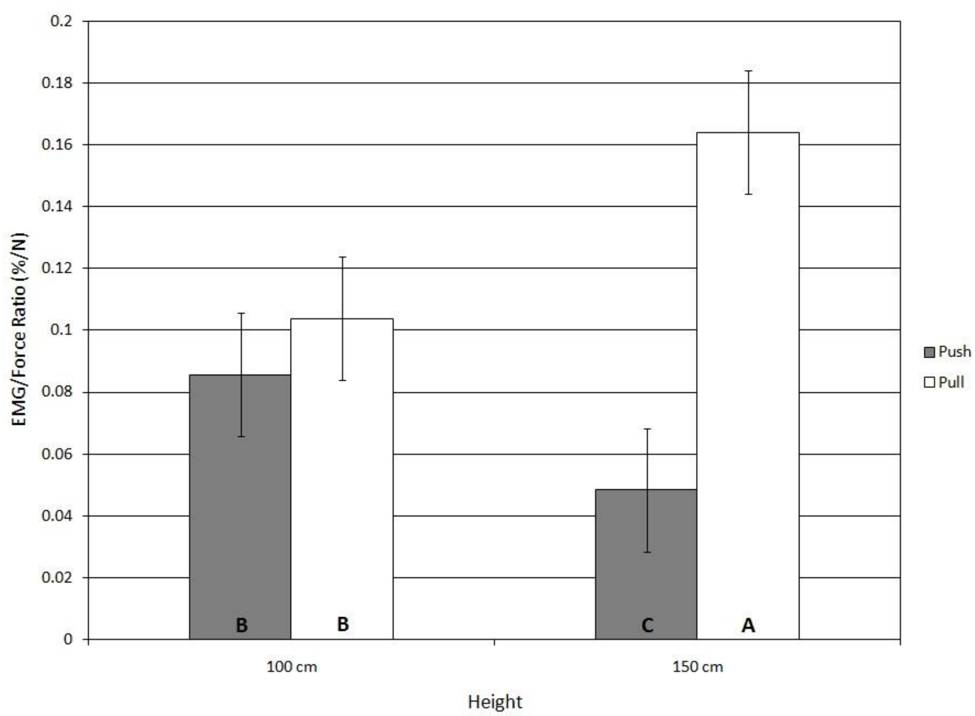


Figure C9: Effect of direction and handle height on LSM EMG/force ratios for the left erector spinae muscle. Letters indicate significantly different direction by handle height interactions.

Appendix D: Effects of Direction and Elbow Angle on Individual Muscle LSM EMG/force Ratios

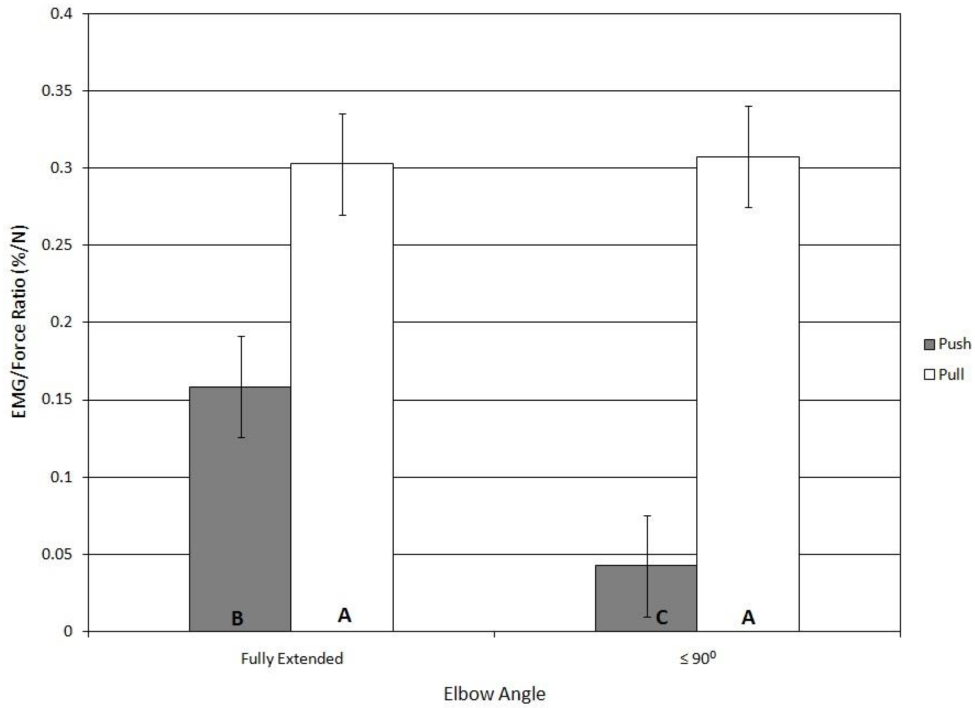


Figure D1: Effect of direction and elbow angle on LSM EMG/force ratios for the right triceps brachii muscle. Letters indicate significantly different direction by elbow angle interactions.

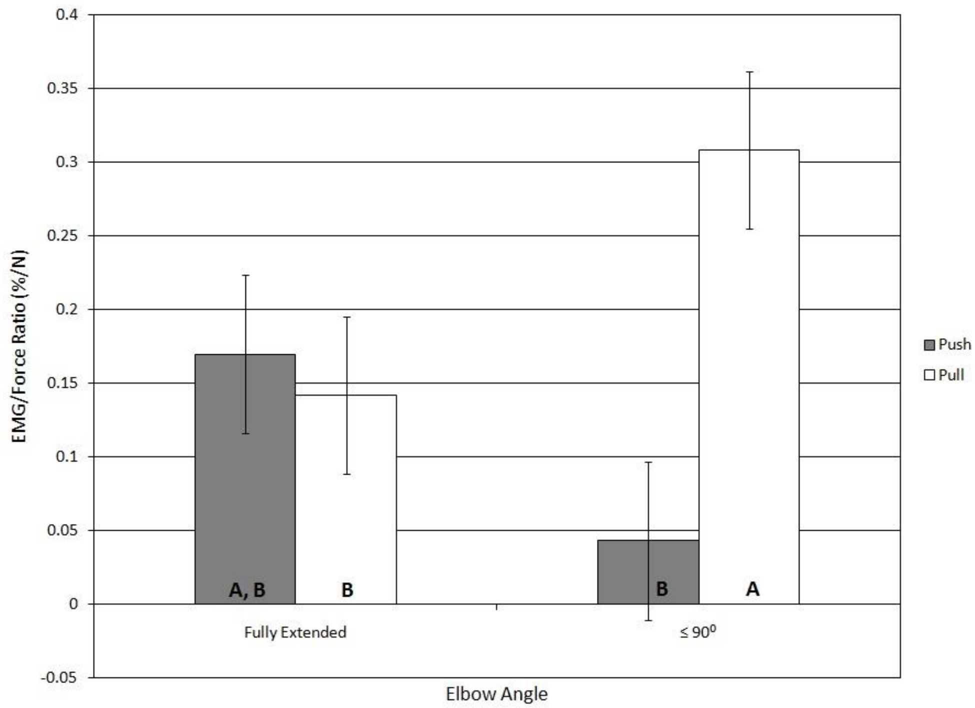


Figure D2: Effect of direction and elbow angle on LSM EMG/force ratios for the left triceps brachii muscle. Letters indicate significantly different direction by elbow angle interactions.

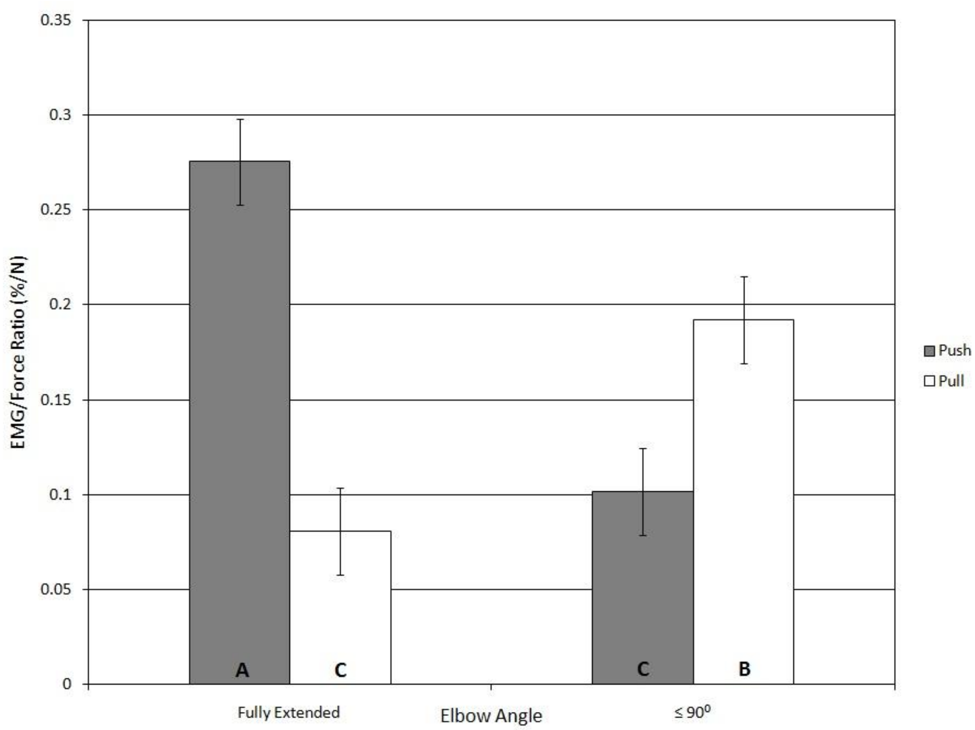


Figure D3: Effect of direction and elbow angle on LSM EMG/force ratios for the right middle deltoid muscle. Letters indicate significantly different direction by elbow angle interactions.

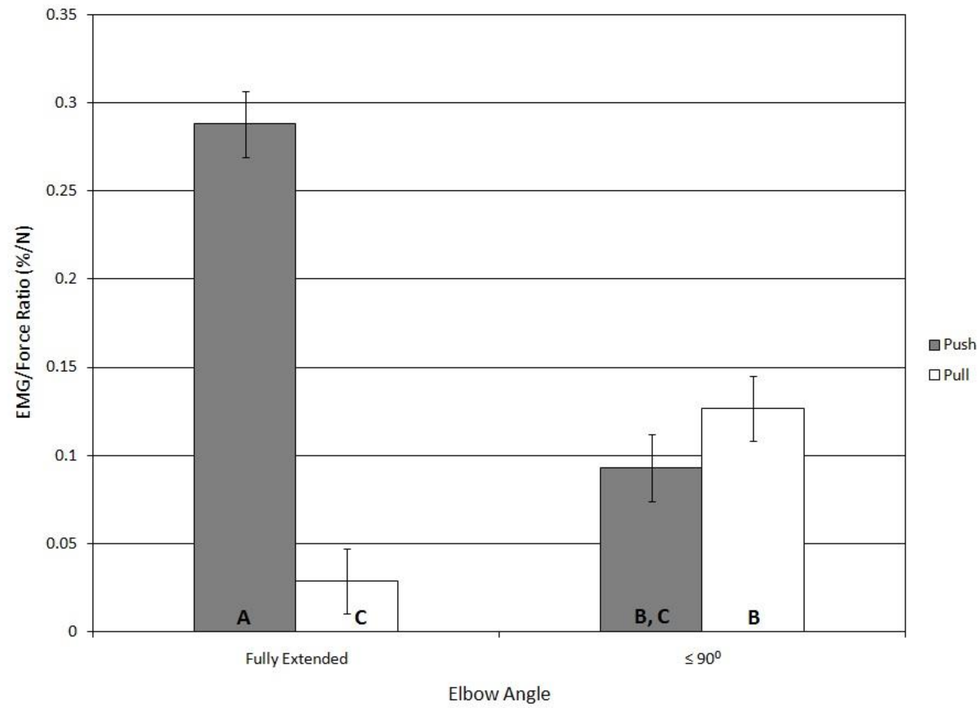


Figure D4: Effect of direction and elbow angle on LSM EMG/force ratios for the left middle deltoid muscle. Letters indicate significantly different direction by elbow angle interactions.

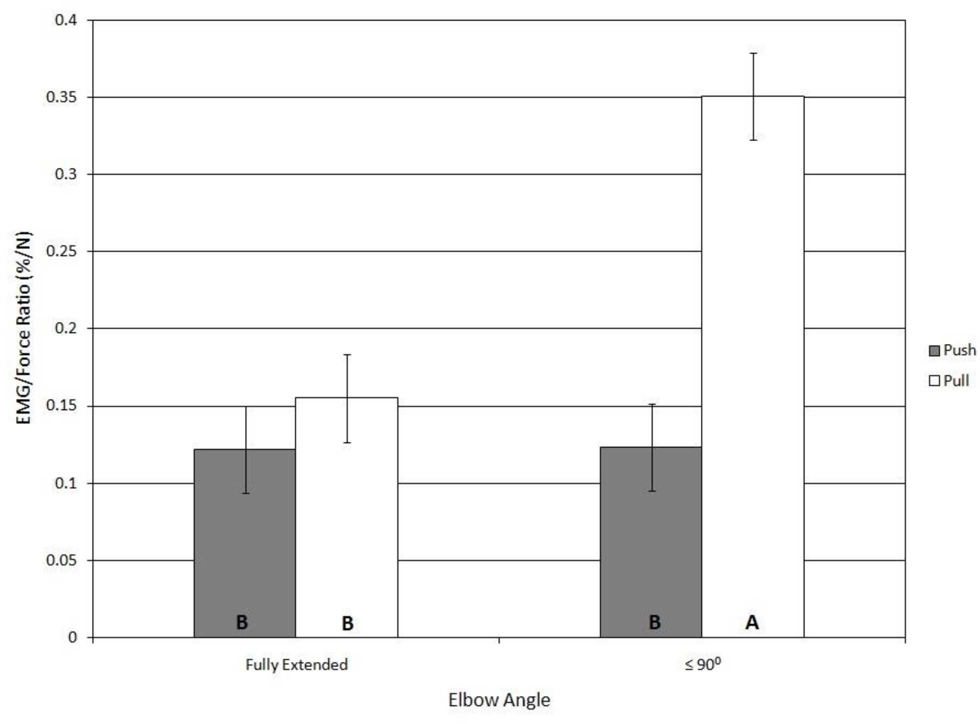


Figure D5: Effect of direction and elbow angle on LSM EMG/force ratios for the right middle trapezius muscle. Letters indicate significantly different direction by elbow angle interactions.

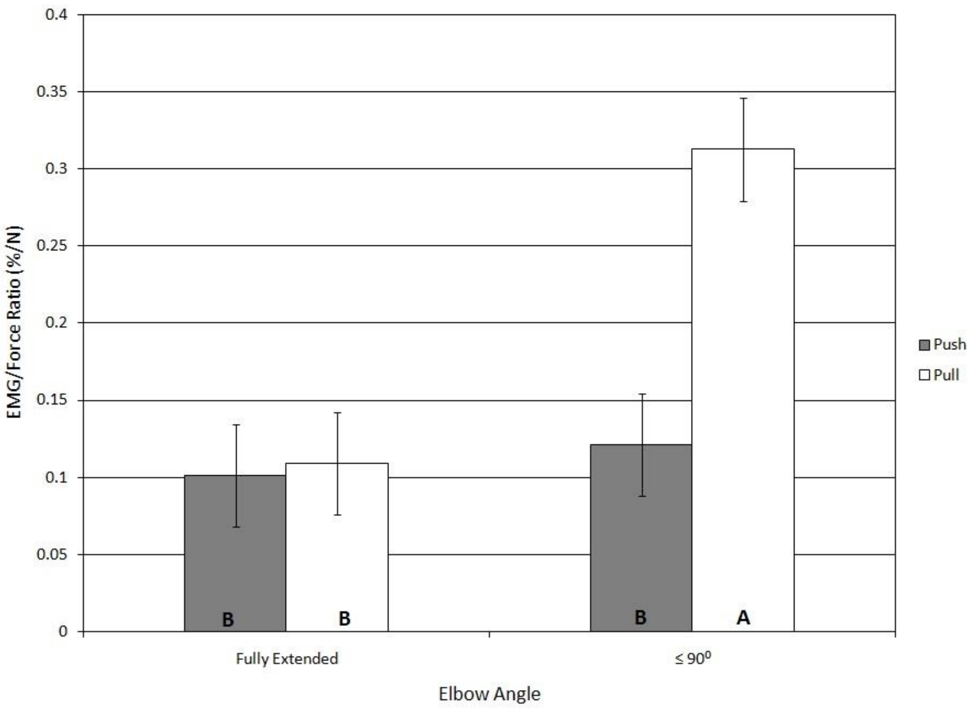


Figure D6: Effect of direction and elbow angle on LSM EMG/force ratios for the left middle trapezius muscle. Letters indicate significantly different direction by elbow angle interactions.

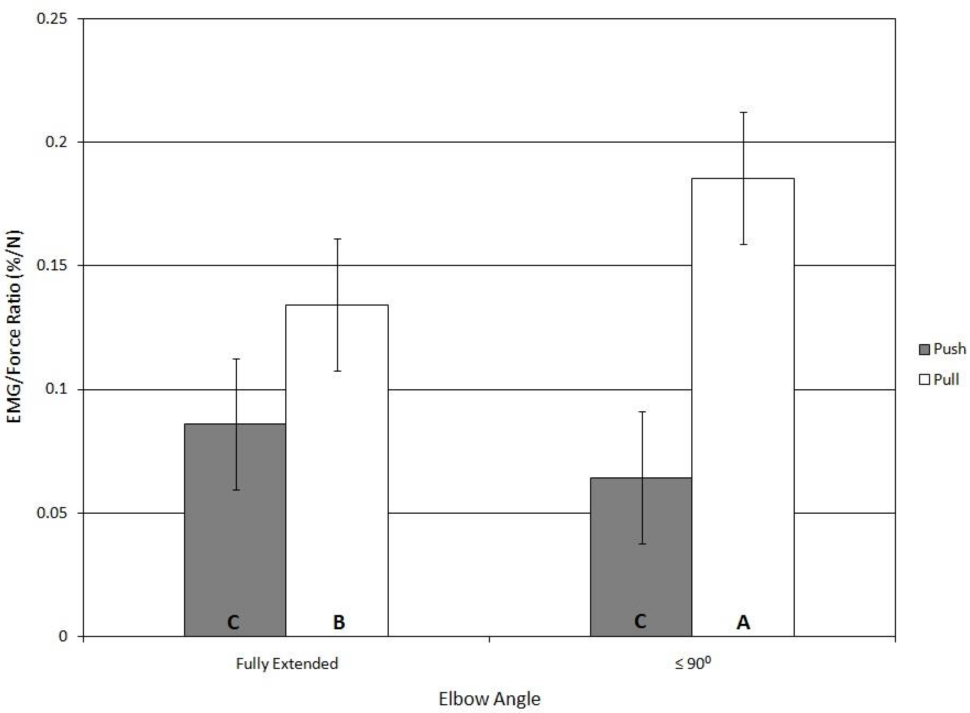


Figure D7: Effect of direction and elbow angle on LSM EMG/force ratios for the right rectus abdominis muscle. Letters indicate significantly different direction by elbow angle interactions.

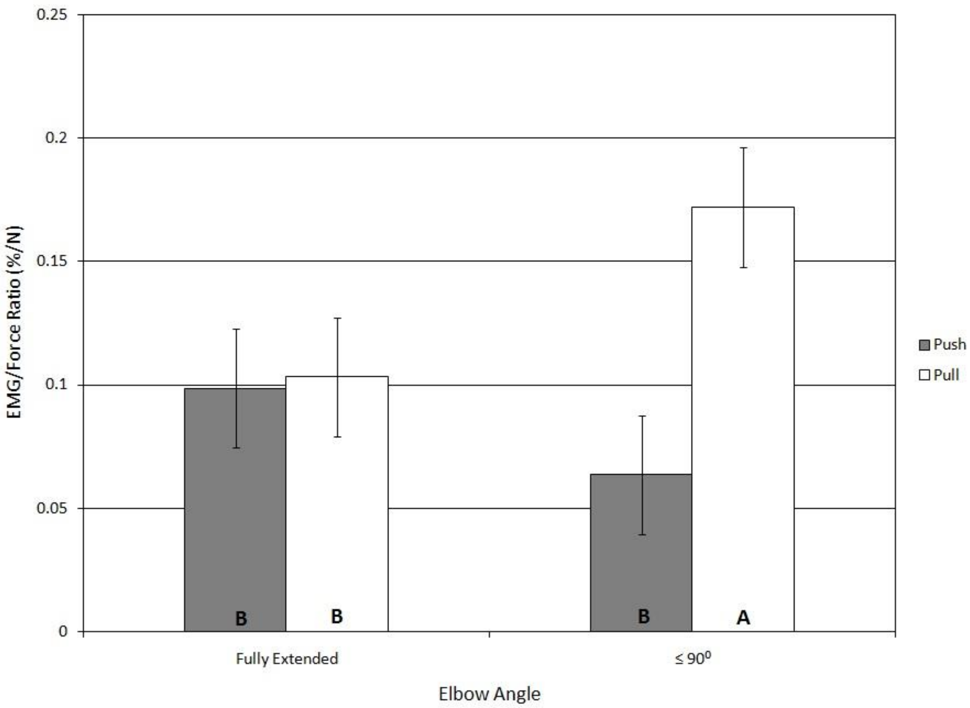


Figure D8: Effect of direction and elbow angle on LSM EMG/force ratios for the left rectus abdominis muscle. Letters indicate significantly different direction by elbow angle interactions.

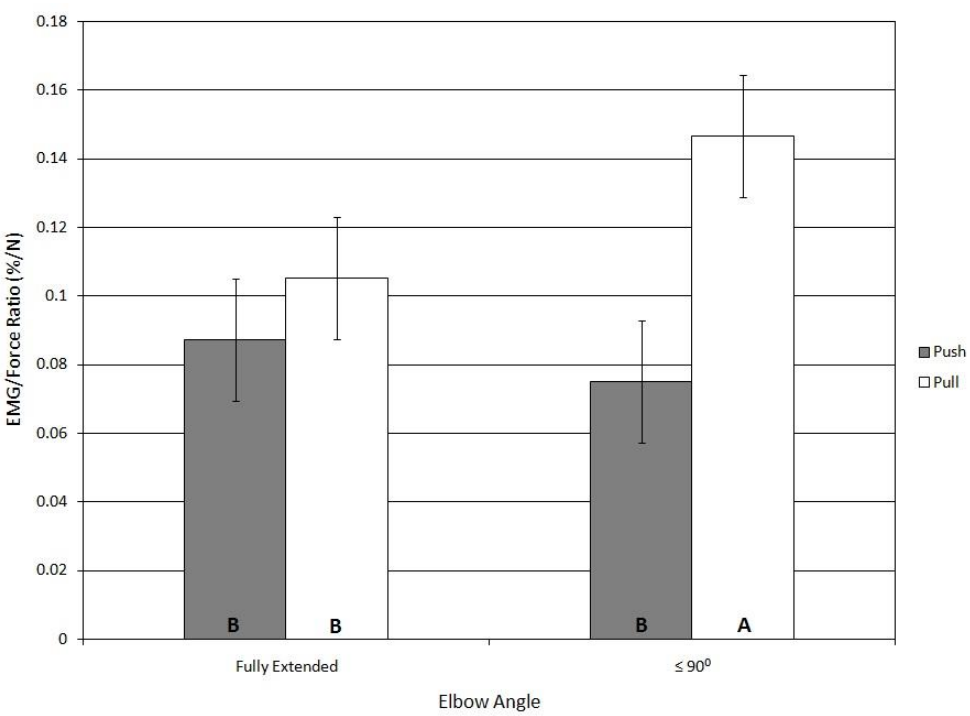


Figure D9: Effect of direction and elbow angle on LSM EMG/force ratios for the left erector spinae muscle. Letters indicate significantly different direction by elbow angle interactions.

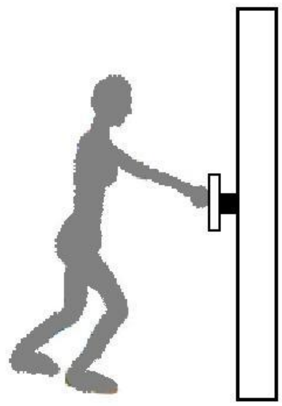
Appendix E: Recruitment Poster

Department of Kinesiology

PARTICIPANTS NEEDED

for a study investigating strength and shoulder muscle activity during push and pull tasks.

- **MALE and FEMALE participants, right-hand dominant**
- **NO past shoulder or back injury**
- **NO shoulder pain or discomfort in the past year**
- **Approximately a 3 hour time commitment**



For the testing session, the following will be performed:

- ✓ **14 surface electrodes will be placed on the right and left shoulder, arm, chest, abdominal and back muscles**
- ✓ **2 moderate intensity (females-10 kg, males-19 kg) push & pull trials**
- ✓ **27 maximum intensity muscle contraction trials**
- ✓ **10 push and 10 pull strength tasks common in industry**

Participants will be asked to wear a comfortable loose fitting t-shirt and females will also be asked to wear a sports bra

If interested, please contact Amy: aychow@uwaterloo.ca

This study has been reviewed by, and received ethics clearance through, the University of Waterloo's Office of Research Ethics

Appendix F: Information and Consent Form

INFORMATION CONSENT FORM

Study Title

Investigation of Hand Forces, Shoulder Muscle Activation Patterns and EMG/force Ratios in Push and Pull Exertions

Research Team

Student Investigator	*Faculty Supervisor*
Amy Chow, MSc. Student Dept. of Kinesiology University of Waterloo 519-888-4567 ext. 36162	Clark Dickerson, PhD Dept. of Kinesiology University of Waterloo 519-888-4567 ext. 37844

Purpose of the Study

The objective of this study is to determine the effects of force direction, handle height, handle orientation and elbow angle on hand forces and muscle activation levels of 14 muscles during two-handed pushing and pulling strength tasks. It is anticipated that the results can assist in designing or modifying workstations, tasks and equipment to allow the greatest force while minimizing muscular strain, as well as provide guidelines for male and female strength limits.

Procedures Involved in this Study and Time Commitment

As a participant in this research study, you will be asked to take part in one testing session that is approximately 3 hours in length. During the session you will be asked to perform a total of 2 submaximal push and 2 submaximal pull trials (trials requiring moderate force), 27 (9x3) maximal muscle activation trials (maximum muscle contractions), as well as 10 (5x2) maximal push and 10 (5x2) maximal pull force trials, or strength trials. The data collection procedures are as follows:

Instrumentation

- Upon your arrival, the skin overlying several upper body muscle groups will be shaved and cleansed by the student investigator or lab assistant associated with this study so that surface electromyography (EMG) electrodes can be attached (with tape). A new disposable razor will be used for each participant. Due to the locations of the electrodes, all participants will be required to wear a loose fitting t-shirt and women are asked to also wear a sports bra. During the EMG setup, participants will be asked to remove their t-shirt. After placing the electrodes, you have the option of putting your t-shirt back on for the remainder of the study. A female researcher will place electrodes on female participants and a male lab assistant will place electrodes on male participants. Upper limb muscle EMG will be collected throughout all of the procedures using 14 electrode pairs. In addition, a ground electrode will be placed on the right collar bone. Electrode pairs will be placed on the left and right sides of the body on the middle deltoid (shoulder), sternal insertion of pectoralis major (upper chest), middle trapezius (back), biceps (arm), triceps (arm), rectus abdominis (abdominal) and erector spinae (back) muscles.
- Eighty-nine skin surface markers will be taped to the skin overlying the whole body with the use of double-sided tape. These markers will be used, in conjunction with VICON cameras, to document the movement of the entire body. Marker placements include the left and right hands (2nd and 5th metacarpal heads), left and right forearms (ulnar styloid, radial styloid, cluster (4) on lateral surface), left and right upper arms (medial epicondyle, lateral epicondyle, cluster (4) on lateral surface, acromion), torso (suprasternal notch, cluster (5) on chest, inferior tip of the xiphoid process, C7/T1 joint, L5/S1 joint, left and right anterior superior iliac spine), head (left and right

ear), left and right thighs (greater trochanter, cluster (5) on anterior surface), left and right shanks (medial condyle, lateral condyle, medial malleolus, lateral malleolus, cluster (5) on anterior surface), left and right feet (1st and 5th metatarsal heads, distal bisection of the calcaneous, cluster (5) on superior surface).

Procedures

- Prior to application of the instrumentation, your height, weight and functional arm reach will be measured and recorded.
- For all trials, you will be given a chance to practice the postures prior to recording the trial. You will be asked to attempt to maintain your posture during each trial, in which each trial will last 6 seconds.
- Following the application of the instrumentation, you will be asked to complete a submaximal push and pull trial at 100 Newtons of force (about 10 kg), if you are a female or at 190 Newtons of force (about 19 kg), if you are a male. You will be asked to place your feet side by side and shoulder width apart such that your elbows are outstretched upon griping the handles.
- Each maximal muscle activation trial will be performed while sitting or lying on a clinical test bench and you will be pushing against manual resistance provided by the student investigator. Once the 9 maximal muscle activation trials are completed, they will be repeated two more times for a total of 27 maximal muscle activation trials. You will be given a minimum of 2 minutes of rest between each trial as well as a minimum of 15 minutes of rest before completing the maximal push and pull force trials.
- For the maximal push and pull force trials, you will be asked to stand and place your feet side by side, shoulder width apart, and stand centred in front of the test handles. Your arms will either be outstretched or bent at the elbows. The right foot positions will be marked on the floor with tape to ensure consistent placement. You may then position your left foot in whatever position you feel will enable you to produce maximal force. For each trial, you will be informed of the direction in which to exert force on the handle (either push or pull) and the trial will begin with you moving your hand towards the handle position, grabbing it and pushing forwards or pulling backwards. Again, a minimum of 2 minutes of rest will be provided between trials. This step will be repeated for each of the 10 conditions, which will be repeated one more time, for a total of 20 maximal push and pull force trials.
- The initial 6-second submaximal push and pull trials will be repeated after the maximal push and pull force trials.
- On occasion, participants are photographed to capture specific postures or equipment setup that would be helpful in teaching or when presenting the study results at a scientific conference or in a publication. You are not required to have your photo taken to participate in this study and photographs will only be taken with your permission and signed consent. Participants are not identified by name and facial images will be blocked out. No photographs will be taken during the setup (i.e., EMG electrode placements).

Potential Risks and Associated Safeguards

- Some participants may experience mild skin irritation/redness from the tape used to attach the instrumentation to the skin. This is similar to the irritation that may be caused by a bandage and typically fades within 1 to 3 days. The occurrence of this skin irritation is rare amongst participants.
- If you have an allergy or sensitivity to rubbing alcohol or electrode gel, please inform the investigators associated with this study. Rubbing alcohol must be used to cleanse the skin prior to electrode attachment. As this is a mandatory step in the procedure, you will not be able to participate in the study if you have an allergy or sensitivity to rubbing alcohol or electrode gel.
- The portable parts of the electrical recording systems are battery operated and isolate you from the main power lines. There is no risk of electrical shock. The instrumentation is CSA approved.
- You may experience muscle fatigue or mild discomfort from the maximal contractions that should disappear in 1 to 3 days.

__Changing Your Mind about Participation__

You may withdraw from the study at any time without penalty. To do so, indicate this to the investigators by saying, “I no longer wish to participate in this study”.

__Inclusion / Exclusion Criteria__

Individuals who have undergone shoulder or back surgery, have had an upper extremity or low back disorder within the past year, or those who have experienced pain or discomfort in their shoulder or back in the past year will be excluded from this study. Individuals with an allergy to isopropyl alcohol or electrode gel will also be excluded.

__Potential Benefits of Participation__

By participating in this study, you will have the opportunity to gain or further your knowledge and understanding of experimental procedures and theories in human movement research. Participants will also be provided with feedback on how varying postures through modifications or design of workstations, tasks or equipment affect strength and shoulder and back muscle activation. The knowledge gained from this research may assist in the reduction of upper extremity injury risk in the workplace.

__Confidentiality and Security of Data__

Each participant will be assigned a 3-letter identification code. Only the investigators associated with this study will have access to this code. All data will be stored indefinitely on computer hard drives (password protected) and/or digital storage media (locked in the investigator’s filing cabinet). A separate consent will be requested in order to use photographs for teaching, for scientific presentations, or in publications of this work. Once all the data are collected and analyzed for this study, it is our intent to share this information with the research community through seminars, conferences, presentations, and journal articles.

__Concerns about Participation__

We would like to assure you that this study has been reviewed by, and received ethics clearance through, the University of Waterloo’s Office of Research Ethics (ORE). However, the final decision about participation is yours. In the event you have any comments or concerns resulting from your participation in this study, please contact Dr. Susan Sykes (Director ORE) at (519) 888-4567 ext. 36005 or ssykes@uwaterloo.ca.

__Questions about the Study__

If you have any further questions or want any other information about this study, please feel free to contact Amy Chow or Dr. Clark Dickerson (contact information provided below).

Sincerely Yours,

Amy Chow, MSc Student
Dept. of Kinesiology
University of Waterloo
519-888-4567 ext. 36162

Clark Dickerson, PhD
Dept. of Kinesiology
University of Waterloo
519-888-4567 ext. 37844

CONSENT OF PARTICIPANT

I have read the information presented in the information letter about a study being conducted by Amy Chow (Student Investigator) and Dr. Clark Dickerson (Faculty Supervisor) of the Department of Kinesiology at the University of Waterloo. I have had the opportunity to ask any questions related to this study, to receive satisfactory answers to my questions, and any additional details I wanted. I am aware that I may withdraw from the study without penalty at any time by advising the researchers of this decision.

This project has been reviewed by, and received ethics clearance through, the University of Waterloo's Office of Research Ethics (ORE). I was informed that if I have any comments or concerns resulting from my participation in this study, I may contact Dr. Susan Sykes (Director, ORE) at (519) 888-4567 ext. 36005 or ssykes@uwaterloo.ca.

With full knowledge of all foregoing, I agree, of my own free will, to participate in this study.

Participant's Name (Please Print): ____

Participant's Signature: ____

Dated at Waterloo, ON: ____

Witnessed: ____

CONSENT TO USE PHOTOGRAPHS IN TEACHING, PRESENTATIONS, and/or PUBLICATIONS

Sometimes a certain photograph clearly demonstrates a particular feature or detail that would be helpful in teaching or when presenting the study results at a scientific conference or in a publication.
I agree to allow photographs in which I appear to be used in teaching, scientific presentations and/or publications with the understanding that I will not be identified by name and facial images will be blocked out. I am aware that I may withdraw this consent at any time without penalty, and the photograph will be confidentially shredded.
I was informed that if I have any comments or concerns resulting from my participation in this study, I may contact Dr. Susan Sykes (Director, Office of Research Ethics) at (519) 888-4567 ext. 36005 or ssykes@uwaterloo.ca.

Participant's Name (Please Print): ____

Participant's Signature: ____

Dated at Waterloo, ON: ____

Witnessed: ____

Appendix G: Feedback Letter

FEEDBACK LETTER

Study Title: Investigation of Hand Forces, Shoulder Muscle Activation Patterns and Muscular Efficiency in Push and Pull Exertions

Dear Participant:

We would like to thank you for your participation in this study. As a reminder, the purpose of this study was to examine the muscle activation patterns of the shoulder and back while completing a series of push and pull strength exertions common in industrial tasks. It is anticipated that the results can assist in designing or modifying workstations, tasks and equipment to allow maximal force while minimizing muscular strain, as well as provide guidelines for female strength limits.

Please remember that any data pertaining to you as an individual participant will be kept confidential. Each participant will be assigned a 3-letter identification code. Only the investigators will have access to this code. All data will be stored indefinitely on computer hard drives (password protected) and/or digital storage media (locked in the investigator's filing cabinet). Once all the data are collected and analyzed for this study, it is our intent to share this information with the research community through seminars, conferences, presentations, and journal articles. If you are interested in receiving more information regarding the results of this study, or if you have questions or concerns, please contact us via phone or e-mail (details listed at bottom of page). If you would like a summary of the results, please let us know by providing us with your contact information. When the study is completed, we will send it to you. The expected date for the study findings to be available is August 31, 2010.

As with all University of Waterloo projects involving human participants, this project was reviewed by, and received ethics clearance through, the University of Waterloo's Office of Research Ethics (ORE). Should you have any comments or concerns resulting from your participation in this study, please contact Dr. Susan Sykes (Director of ORE) at (519) 888-4567 ext. 36005 or ssykes@uwaterloo.ca.

Thank you again for your participation in this study.

Sincerely Yours,

Amy Chow, MSc Student	Clark Dickerson, PhD
Dept. of Kinesiology	Dept. of Kinesiology
University of Waterloo	University of Waterloo
519-888-4567 ext. 36162	519-888-4567 ext. 37844

Further Reading:

Hoozemans, M. J. M., Kuijer, P. P. F. M., Kingma, I., van Dieën, J. H., de Vries, W. H. K., van der Woude, L. H. V., et al. (2004). Mechanical loading of the low back and shoulders during pushing and pulling activities. *Ergonomics, 47*(1), 1-18.

I participated in the study: Investigation of Hand Forces, Shoulder Muscle Activation Patterns and Muscular Efficiency in Push and Pull Exertions

I would like to receive a summary of the results.

Name: ______________________________

Address: ______________________________

E-mail: ______________________________