EFFECTS OF A SIMULATED MOTION ENVIRONMENT UPON THE PHYSICAL DEMANDS OF HEAVY MATERIALS HANDLING OPERATORS

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Effects of a simulated motion environment upon the physical demands of heavy materials handling operators

By

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ABSTRACT

Maritime shipping, commercial fishing, passenger and cargo shipping and offshore oil and gas industries are all major contributors to the economies of Atlantic Canada. These industries require workers to perform heavy materials handling under harsh environmental conditions, particularly extreme deck motions. The purpose of this study was to better understand the demands of a moving environment on the ability of a person to perform specific lifting tasks.

Nineteen healthy male subjects volunteered for this study. Each subject was required to lift a 15 kg load under four lifting conditions. While performing these lifting tasks, a ship's motion simulator was used to create deck motions under foot. Three deck motions were considered: pitch, quartering and roll. A stable laboratory condition was also collected for all lifting conditions. Electromyography (EMG) histories of four muscles (erector spinae, latissimus dorsi, external oblique and trapezius) were collected bilaterally and thoracolumbar kinematics was measured throughout the experimental protocol.

A repeated measures ANOVA was employed to assess trunk motions and muscle activities across the lifting and motion conditions. There were no significant differences found due to the motion effect for any of the muscles monitored in this study. However, the lifting task did produce differences in the EMG activities for some muscles. The maximal sagittal velocities were significantly smaller for all motion states in comparison to the stable lab condition ($p \leq 0.01$) while maximum twisting and lateral bending velocities increased in the motion conditions compared to the stable lab condition

($p \leq 0.05$). Results suggest that working in a moving environment will likely increase the operator's risk for overexertion injuries, particularly to the spine.

Key Words: MMH, EMG, LMM, offshore industry, unstable environments, motion environments, simulated platform motion

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LIST OF ABBREVIATIONS

MMH	Manual Materials Handling
LBP	Low Back Pain
NIOSH	National Institute for Occupational Safety and Health
MII	Motion Induced Interruption
MIF	Motion Induced Fatigue
MIS	Motion Induced Sickness
EMG	Electromyography
LMM	Lumbar Motion Monitor
MVC	Maximum Voluntary Contraction
L	Left
R	Right
ES	Erector Spinae
LAT	Latissimus Dorsi
TRAP	Trapezius
OBLIQ	External Oblique
CH	Close High
FH	Far High
CF	Close Floor
FF	Far Floor
CoM	Center of Mass

CO-AUTHORSHIP STATEMENT

There have been a number of people who have had considerable input into this manuscript.

i) This concept of studying manual materials handling in moving environments is attributed to Dr. Scott MacKinnon.

ii) Steven Mills, Julie Matthews, Scott MacKinnon and I recruited all participants and performed all research testing on each participant.

iii) Raw data were collected by Steven Mills, Julie Matthews, Scott MacKinnon and I. With guidance from Dr. MacKinnon and Dr. Albert, Julie Matthews, Steven Mills and I reduced the data.

iv) With the guidance of Dr. MacKinnon, I prepared the following thesis.

Chapter 1
Introduction

1.1 INTRODUCTION

The human operator constantly adapts to the workplace, whether through conscious or unconscious means, trying to control the exposure of work related stresses on the body. While it seems intuitive that humans would adapt and adopt strategies to protect oneself from harm and injury, the current literature would suggest that occupational related injuries are still widespread. In many circumstances workplace production levels have exceeded the capacity of a typical operator, resulting in greater physical and mental demands and increased risk for accident and injury. Cook and Neumann (1987) suggested that one third of all industrial jobs in the United States include some form of Manual Materials Handling (MMH) activities. Chung et al. (1999) state that industrial MMH tasks are a primary source of musculoskeletal injury and that one out of every three to four overexertion injuries are attributed to MMH tasks.

Lifting, lowering, pushing, pulling, or carrying tasks are common MMH activities, which often require the operator to work in awkward bodily postures. Undesirable body postures coupled with other plausible mechanisms of injury creates significant challenges for Ergonomists developing successful intervention strategies. Static work postures as well as frequent bending and twisting, work intensity and repetition all increase the risk of injury to the worker (Dolan et al., 2001; Marras et al., 1995; McGill et al., 1987). Dolan et al. (2001) reasoned that often many typical occupational demands require the spine to function in awkward and twisted postures, rather than a safe sagittally symmetric one. Furthermore, McGill et al. (1987) suggested that high repetition lifting may lead to trauma, causing lumbar injury.

The approach of selecting an employee with compatible physical and mental characteristics necessary to perform a task in a safe and efficient manner has been criticized because it restricts fair hiring practices. The most desirable ergonomic approach has been to alter the work environment to better suit the persons capabilities. However, altering the work environment can translate into additional costs to the employer and is not always practical or effective. In some cases, opportunities to change the workplace characteristics are difficult, for example working in a marine environment.

Kumar (2001) reported that more than 20% of the world's population works under physically hazardous conditions and under high work loads. One occupation that regularly incorporates many of the MMH activities described above, into a rather hazardous work setting are offshore marine industries. Maritime shipping, commercial fishing, passenger and cargo shipping, and offshore oil and gas industries are all major contributors to the economies of Atlantic Canada. Offshore industries employ a large number of Atlantic Canadians and reflect much of the economic activity of the region. Generally offshore industries are unable to alter workstations and/or change the environmental setting to reduce the risk of injury to the operator. This in turn, proves to further increase the complexity of effectively providing safe work practice recommendations to these occupations.

Studies dating back to McLeod et al. (1980) have stated that ship motions can directly affect the performance of crew members. Wedge and Langlois (2003) stated that humans have difficulty moving in a controlled and coherent manner when ship motions are involved. Further study has revealed three major factors that will potentially affect an individual's performance: motion induced sickness, motion induced fatigue, and motion induced interruptions (Wertheim, 1998). Furthermore, these factors can be exacerbated when workers operate in cold and icy environments, on slippery ship decks, within confined spaces and over long work periods.

Arguably the most studied of all MMH activities are lifting. Ciriello et al. (1999) states that lifting comprises approximately 40% of all MMH activities, and it has become widely accepted that lifting can be directly related to low back pain (Ferguson and Marrass, 1997; Hsiang, 1997; Pope et al., 1984; Snook et al., 1978; Troup, 1965). From a biomechanical perspective, lifting becomes even more difficult when attempted in moving environments, often resulting in a loss of balance and increased risk for injury (Kingma et al., 2003). Loss of balance has been referred to as motion induced interruptions (MII), which Crossland and Rich (2000) defined as occurring when motions cause a person to lose balance and have to make postural adjustments in order to successfully perform the task at hand. Li (2002) stated that 46.2% and 19.8% of the accidents and deaths of seafarers worldwide were caused by slips/falls (i.e. MII's) during MMH activities. Torner et al. (1988) reported that the prevalence of musculoskeletal injuries was 74% in professional Swedish fishermen; with 70% of the injuries related to low back over-exertions. These data certainly suggest that engaging in MMH tasks in a moving environment can significantly increase a person's chance of injury, in particular to the trunk region.

While researchers have attempted to derive guidelines to promote safe lifting environments (Waters et al., 1993) or to assess risk of overexertion injury while lifting

(Marras et al., 1995) these typically do not consider work in motion environments. Wertheim (1998) suggests that many lifting tasks are dangerous when performed in a non moving environment, and the same lifting tasks under a moving environment may place operators at an even greater risk of injury. Ultimately, to develop acceptable limits for lifting tasks performed in offshore environments, more work considering platform motions and workstation and vessel design is required. This will provide a better understanding of the biomechanical demands placed on offshore workers engaged in MMH activities.

1.2 OPERATIONAL DEFINITATIONS

Motion-induced Interruption (MII): When platform motions are sufficiently large to cause a person to slide or lose balance unless they temporarily abandon their allotted task in order to maintain postural stability (Crossland and Rich, 2000)

Unstable Environment: A moving platform or surface that likely decreases a person's stability and equilibrium.

Stable Environment: No motion, static platform.

Six degrees of freedom: A ship can experience motion in 3–dimensions or six degrees of freedom; These six degrees of freedom are described as; the x direction (surge) or rotation around the x-axis (roll); the y direction (sway) or rotation around the y-axis (pitch); and finally the z direction (heave) or rotation around the z-axis (yaw) (Torner et al., 1994).

Motion Simulator: A full bridge ship motion simulator capable of producing computer simulated hydrodynamic motions in 6 degrees of freedom. This device is located at the Centre for Marine Simulation, Marine Institute, Memorial University of Newfoundland.

1.3 PURPOSE AND HYPOTHESIS

The purpose of this study was to better understand the demands of a moving environment on the ability of a person to perform specific lifting tasks. Such information may provide insight into the reasons people who work in a moving environments demonstrate higher incidence of overexertion injuries, particularly to the trunk region. Four experimental hypotheses are proposed:

H1: The direction of simulated platform motions will cause a significant increase in muscle activation during a lifting task.

H2: The direction of simulated platform motions will cause a significant increase in thoracolumbar kinematics during a lifting task.

H3: The characteristics of the lifting tasks will have significant effects on muscle activation during lifting conditions.

H4: The characteristics of the lifting tasks will have significant effects on thoracolumbar kinematics during lifting conditions.

1.4 ASSUMPTIONS

The following assumptions were made in this study:

1. All subjects will be prepared in the same manner with respect to electrode preparation and placement as well as Lumbar Motion Monitor placement.
2. All lifting conditions are considered regular lifting tasks that could be performed by those working in a moving environment, and all lifts fall within NIOSH guidelines (Waters et al., 1993) for safe lifting in a stable environment.

1.5 LIMITATIONS

The following limitations are recognized in this study:

1. Subjects performed repeated lifts during each motion condition; however, subjects only experienced each motion state once throughout the duration of this study.
2. Surface Electromyography and Lumbar Motion Monitor data are both considered indirect means for measures of the corresponding forces that act on the spine.
3. All subjects involved in this study were volunteer university aged participants. Further evaluation of experienced personnel may express means for ways of coping with the demands of a moving environment.

Chapter 2
Review of Literature

2.1 INTRODUCTION

Manual materials handling tasks, such as lifting, lowering, carrying, pushing and pulling features prominently in many aspects of manual labour activities, however, by far the most studied task is lifting. Many studies (Ferguson and Marras, 1997; Hsiang, 1997; Pope et al., 1984; Snook et al., 1978; Troup, 1965) have suggested an association between lifting and the occurrence of low back pain (LBP). Graves et al. (1990) estimated that between 70 and 80% of the all adults will experience LBP at some point during their lives. Despite the obvious personal tribulations that the injured person will experience with LBP, there are also extremely high costs to businesses and organizations related to the occurrence of job-related LBP. Marras and Granata (1997) state that low back disorders are amongst the leading causes for lost work days and are the most costly occupational safety and health problem common in industry.

There are large amounts of research to support the notion of LBP being related to lifting. Countless studies have been performed, with many people trying to advocate the use of proper lifting techniques. This provides a great starting point towards lowering the incidences of LBP caused by lifting, however further discussion may suggest greater concerns. Hsiang et al. (1997) defined lifting as the movement of an object from a starting position to an ending position while increasing the objects vertical position. Toussaint et al. (1998) suggested a lifting task be split into three phases; the reaching phase, which is the involvement of forward bending of the trunk while reaching for the load; the grasping phase, which is handling of the load; and finally, the lifting phase, where the load is lifted to the desired end position. Albeit the demands to perform lifting maneuvers have declined with technology and manufacturing advances, many jobs still involve the manipulation of a load from one destination to the next. Technological advances continue to provide changes to the workplace; however, humans still perform many of the required tasks by hand.

In the past, research examining low back pain associated with lifting activities, have focused predominantly on lifting techniques, the load being lifted, and workstation characteristics. Lacking from past research is an understanding of the effects the external environment has on a worker performing the lifting tasks. Many people perform lifting tasks in a moving environment, often unaware they may be placing themselves at an even greater risk for injury when compared to peers working in comparable stable environments. Kingma et al. (2003) stated that the influence of accelerations caused by the moving surface may dramatically affect low back loading. Kumar (1990) stated that "back pain can result from a single cause or from multiple pathologic causes". Moving environments place repetitive stresses on the back, while workers in this environment often perform repetitive tasks, with limited rest intervals. This could suggest that cumulative loading due to exposure of multiple, repetitive causes, should significantly increase the likelihood of chronic, overexertion injuries. Moving environments, such as ship platforms, produce very unpredictable motions and can actually produce accelerations large enough to launch a person from the floor. Large accelerations of this nature, which cause the operator to lose balance, will increase the probability of back pain from a single, acute event. By gaining a better understanding of the environmental and work circumstances that pose the greatest risk to the operator, recommendations can then be made to improve the safety of those employed in moving or unstable environments.

The following literature review will provide an overview of lifting as a manual materials handling activity. The review will focus on lifting techniques, followed by an assessment of how load characteristics, lifting speed and fatigue can affect the performance of the operator. A review of electromyography and lumbar motion during common lifting techniques will also be presented. Following these sections a discussion on lifting during a moving environment will focus on factors that promote injury and inhibit productivity. An understanding of unexpected loading on the spine, both in a moving environment and with unexpected operator loads will conclude the review.

2.2 LIFTING TECHNIQUES

Much interest has been devoted to the manner operators perform lifting tasks in order to gain a better understanding of how work-related injuries might happen. As a result, lifting tasks have become assessed on the postural techniques a person adopts during the execution of the lift.

Humans have adapted to the problems that arise during a common workday and have developed many different lifting techniques and styles to help cope with the physical demands of the job. A lifting technique is defined as the posture a person assumes immediately prior to lifting the desired load, with particular attention given to the knee joint position (Trafimow et al., 1993). Heiss et al. (1997; 2002) and Van Diëen et al. (2003) have reported upon four of the most frequent lifting techniques and describe these as the squat, semi-squat, stoop, and freestyle techniques.

The squat or frequently termed 'leg lift' is the most commonly advised lifting technique (Van Diëen et al., 1999). While there have been numerous definitions of the squat lift (Heiss et al., 2002; Hsiang et al., 1997), Straker (2003) defines the technique as being characterized with a start position of deep knee flexion with the trunk close to erect. When lifting a load from the floor, this lifting strategy is commonly described as knee flexion around 45 degrees and trunk flexion less then 30 degrees for most people (Straker, 2003).

The stoop or frequently termed 'back lift' is probably the least promoted lifting technique. The stoop method involves bending of the upper torso down and forward (Hsiang et al., 1997). During this lifting technique, the knees typically remain extended, and trunk flexion can reach upwards to 90 degrees or more (Straker, 2003).

The semi-squat lifting technique utilizes a combination of the squat lift and the stoop lift (Heiss et al., 2002). This approach incorporates moderate knee flexion and trunk inclination. Knee flexion is typically around 90 degrees and trunk flexion around 45 degrees for most workers (Straker, 2003). Without formal lifting education or training, humans generally adopt this lifting technique.

Finally, the freestyle lifting technique is generally referred to as some combination of the previous three techniques, or the technique a person uses when not given instruction as to how they should perform the task (Kumar 1984).

Studies dating back to Brackett (1924) have recommended avoiding a flexed back during lifting. As a result, the 'lifting with the legs', or squat technique has become the most widespread and universal approach to safer lifting. Contrary to popular belief, the squat lifting technique has failed to demonstrate much success of efficacy in the literature.

Garg and Saxena (1979) performed a lifting study with four different lifting frequencies and three different lifting techniques. Results showed for the same amount of physical work, metabolic cost is smallest during the free-style lifting technique and greatest during the squat technique. From a physiological perspective, a lower metabolic cost could reflect reduced levels of fatigue, suggesting the free-style lifting technique should be favored.

A study by Hagen et al. (1993) demonstrated that oxygen consumption and ventilation demands during squat lifting were greater than for stoop lifting. This is likely due to the greater muscle activity of lower body musculature during squat lifting (Straker, 2003). Kumar (1984) demonstrated that the stoop technique required less oxygen and had a lower per-minute inspiratory ventilation volume than the squat technique. It could be suggested that maximum oxygen consumption and maximum ventilation capacities, both provide equal benefits and no physiological advantage when performing a squat lifting or stoop lifting technique.

From a mechanical perspective, several studies calculated squat and stoop lifting moments about the lumbar spine to be within 5% of each other (De Looze et al., 1994; Hagen et al., 1994; Kumar, 1994). Contradicting these studies, Potvin et al. (1991) reported peak lumbar moments for stoop lifting to be 5% greater then squat lifting. Bush-Josep et al. (1988) and Dolan et at. (1994) both reported greater moments of 10 and 13% respectively, for freestyle lifting than squat lifting. Potvin et al. (1991) estimated shear forces to be 180% less for squat lifting compared to stoop lifting. Dolan et al. (1994) reported that stoop lifting resulted in a peak lumbar flexion of approximately 100% of a person's maximum, while squat lifting resulted in peak lumbar flexion of 80% of a person's maximum. As a result, Dolan et al. (1994) demonstrated stoop lifting to result in 75% more stress on passive tissues when compared to squat lifting. Van Diëen et al. (1999) reviewed 27 studies comparing mechanical loading during stoop and squat lifting techniques. When lumbar moments and compression forces were studied, conclusions suggest the squat lifting technique had the highest values. During sheer forces, passive tissue and ligament stresses, no positive correlations were found for using the squat technique over any other lifting approach. From a biomechanical perspective, the above studies would suggest that the squat lifting technique seems to impose the least amount of stress on the person performing a lift.

Schipplein et al. (1990) suggested that people who were asked to execute freestyle lifts generally performed a semi-squat technique. The semi-squat technique will place the operator in a posture that reduces the amount of work and energy expenditure when compared to the squat technique. Heiss (2002) states that the height of the body center of mass is higher and the knees are more extended, therefore less work should be required to exert the lift. Although the squat technique is often recommended, when performing repetitive lifting tasks, the semi-squat technique may be the most preferred technique, due to its limited amount of stress, and reduction in energy expenditure.

While many researchers define what they feel are the best lifting techniques to use, ultimately there is probably no irrefutable evidence to suggest one lifting technique is better then the next. The literature is quite inconclusive from both physiological and biomechanical perspectives. Before recommendations as to which lifting technique should be made, the desired lifting task needs to be fully understood and accurately evaluated.

2.2.1 Knowledge of Load Characteristics

Generally an operator will have some prior knowledge of the characteristics of a load being lifted. However, this load mass can shift requiring the worker to make compensatory adjustments to the planned movement strategy. Commissaris and Toussaint (1997) explain that workers generally use previous experience to estimate the characteristics of an unknown load. When operators perform lifting tasks and are unaware of the mass of the load being lifted, balance issues will result in unexpected loading of the spine. Unexpected loading of the spine has also been thought to occur when operators working in a moving environment suddenly slip or fall, since added twisting and bending accelerations are expected on the spine. Despite Manning et al. (1981) suggesting that balance loss can be highly associated with the onset of LBP, little work of this nature can be found in the literature.

Heiss et al. (2002) performed a study to gain a better understanding of low back loading during balance loss. This balance loss was induced by heavier or lighter then expected loads and not as a result of a moving platform. Participants were divided into two groups, categorized as either those who lost balance or those who maintained balance. Heiss' results suggest that when the mass of the load is underestimated, too little effort and momentum occurs, making the subject lose balance in the forward direction. However, the results indicate that lifting technique also needs to be considered. Subjects who lost balance typically situated themselves with a deeper knee bend and a more vertically oriented trunk (squat lifting technique), while those who maintained balance used the semi-squat technique. This agrees with other research (Commissaris & Toussaint, 1997; Hsiang et al., 1997; Van Diëen et al., 2003) that suggests the same loss of balance due to the squat lifting technique. Van der Burg et al. (2000) also performed a study to evaluate the effects of lifting a heavier then expected load on low back loading and balance problems. This study had subjects lift a 5 or 10 kg load as fast as possible. Finding were somewhat different then Heiss and colleagues. Van der Burg suggests that the heavier then expected loads did not appear to increase balance loss or produce an increase in low back loading. While subjects were instructed to lift as fast as possible, Van der Burg suggests that as the heavier loads were introduced, subjects lifting speed slowed. This, compared to the lighter weights studied by Heiss, could attribute to discrepancies in the amount of muscle activity during heavier then expected lifting. A study by Butler et al. (1993) supports Van der Burg's conclusions of unknown heavy loads increasing mechanical loading on the spine.

Butler et al. (1993) performed a study consisting of ten subjects, each required to lift a box containing either no weight or containing one of three different loads. Subjects were unaware of which load was being attempted. Kinetics and kinematics of the lifts were considered. Results suggested that subjects approach each lift assuming a certain weight. If this assumption is incorrect, outcomes suggested by Heiss et al. (2002) tend to occur. These authors concluded that the result of unknown load masses place greater twisting and jerking motions on the subject, which in turn will create greater loading on the lower back.

De Looze et al. (2000) performed a similar study involving nine male subjects who performed a lifting task with a known mass and a lifting task where they were unaware of the load mass. However, for safety reasons, subjects knew the load would be in a certain range (6.5-16.5 kg). De Looze et al. (2000) concluded that forces and back muscle activation were higher during the unknown load conditions. These results suggest that increased forces and back muscle activation, coupled with loss of balance during lifting all pose significant hazards to the operator when load mass is unknown. Mannion et al. (2000) stated that under conditions of sudden loading, the back muscles may be unable to generate forces required to prevent excessive bending or twisting.

Marras et al. (1987) demonstrated that when a falling load was placed into the hands, trunk musculature EMG activity increased by 35%, when the load was expected and by 50% when the load was unexpected. They also proposed that compressions values increased from 200% to 350%. However, this study was performed using static postures and was not carried out under typical lifting conditions.

2.2.2 Lifting Speed

The speed at which the operator performs the lifting activity has also been reported in the literature as influencing low back loading. Bernard et al. (1999) performed a study where subjects were required to lift a light (6.82 kg) and heavy (27.3 kg) load across five different lifting speeds. Subjects could decide the lifting speed that was most comfortable and were free to perform the task using any desired lifting technique. The five speeds studied were labeled very slow, slow, normal, fast and very fast. Moments were calculated about the ankle, knee, hip, elbow and shoulder and moments were summed across all joints as an indication of cumulative stress on the entire body. Values calculated for the slowest speed were 4% smaller then those at the normal lifting speed, while moments at the fastest speed were 3% larger then normal. Not surprisingly, the speed of the lift significantly affected the average inertial moment.

Lavender et al. (2003) examined the effects of lifting speed on moments calculated about the L5/S1 segment of the spine. Subjects performed a lifting task from three different load origins, with a normal and fast lifting speed. Subjects self-determined the normal and fast lifting speeds, and were asked to choose the lifting technique that was most comfortable. Results indicate that lifting speed had a negligible effect on the moment when lifting larger loads at the knee and knuckle levels. However, there was a notable increase in moments with the faster lifting speed as lighter loads were lifted from below knuckle level (i.e. lifts originating closest to the ground). This suggests that some lifts are able to better use muscle groups other than those acting on the trunk to affect the lift, thus reducing the stresses on the lower back.

2.2.3 Lifting and Fatigue

Operators often perform repetitive lifting tasks throughout the work day. Waersted and Westgaard (1991) stated that the frequency of musculo-skeletal injuries and back pain was increased with longer workdays. Furthermore, repetitive lifting has been shown to be a risk factor for the development of back pain (Frymoyer et al. 1983). While proper lifting techniques should be common practice, if not regularly promoted, in the workplace, over long periods of time operators tend to change lifting techniques as fatigue occurs (Fogleman and Smith, 1995). Asmussen (1979) described fatigue as "a transient decrease of working capacity", while Edwards (1981) described fatigue as failure to sustain the required or expected force.

Marras et al. (1993) reported that repeated bending and lifting activities greatly increase the risk of developing low back disorders. Van Diëen et al. (1998) suggested that this increase in the risk of low back pain was the result of increased loads on passive consumptions associated with predominant leg lifting (Hagen et al., 1993). From an industrial perspective, the stoop lifting technique allows for quicker lift cycles, thus improving productivity.

2.3 ELECTRYMYOGRAPHY PROFILES OF LIFTING

Electromyography (EMG) is a technique used to measure the electrical activity of a muscle. EMG provides insight into muscle activity and recruitment patterns as well as muscle fatigue. EMG measures are generally made using either surface electrodes, commonly referred to as surface EMG, or the use of needles, referred to as fine wire (indwelling) EMG. The advancement of modern technology has developed small EMG systems, allowing researchers to successfully employ surface EMG quite easily in industry. As a result, there have been tremendous amounts of research done to evaluate the physical demands of many MMH activities.

It has been made clear throughout this chapter that a strong correlation exists between lifting and the occurrence of back pain. As a result, many authors have spent considerable time evaluating trunk musculature, in hopes of better understanding the possible muscular mechanisms that lead to back pain. To understand better these possible mechanisms, muscle activity needs to be related in terms of force output and internal forces. As a result, researchers have developed models to determine individual muscle force, as well as torsion, shear and compression forces acting on the spine.

Callaghan et al. (2001) suggest that mechanical loading (in particular peak compression) of the spine has been used to identify low back injury for years. Adams and Dolan (2005) state that the best measure of spinal compression may be the implementation of inserting a needle into the nucleus pulposus of a lumbar disk, as performed by Nachemson (1981). However, for obvious reasons, this is not a practical method for use outside laboratory settings (Kingma, 2001).

Muscle force and joint moments have also been predicted employing models driven by EMG inputs. Many researchers (Marras and Sommerich, 1991; McGill, 1992; McGill and Norman, 1985) have developed models of this nature to use EMG signals as a predictor for forces that act on the spine. These models ultimately uses EMG data to evaluate individual moments generated from both trunk extensor and flexor musculature. McGill and Norman (1985) were one of the first to develop such models and theirs consisted of 6 muscles (which ultimately were used to predict 20 different muscle forces) and 8 ligaments. This model considered only sagittally symmetric lifting tasks. While this is a rather complex model, like most, it remains difficult to validate. A model by Marras and Sommerich (1991) followed very similar methods, however successfully evaluated asymmetric lifting tasks through the use of EMG measures of 10 muscles, anthropometry, and trunk kinetics. Another method (Thelan et al., 1994) provides information into spinal loads in three dimensions, while also combining EMG activity with trunk velocity and accelerations to develop a relationship to the net torque.

Kingma et al. (2001) however would suggest that no technique is considered totally predicative, as none have been successfully validated. Furthermore, Davis and Marras (2000) suggest that few studies have successfully evaluated muscle activity during flexion, lateral bending and twisting activities.

EMG based models are often very complex and take considerable time to develop. It should also be understood that there can be many problems related to EMG measures. Prior to data collection, inherent noise in the electrical components during recording, ambient noise, motion artifacts, input impedance, and electrode stability are a few of the problems that should be considered. Winters (2005) also suggest that other variables (velocity of shortening or lengthening of the muscle, fatigue and reflex activity) will ultimately affect an EMG signal. Despite problems that can arise, EMG techniques are still widely used in research today. EMG signals can provide Ergonomists and other professionals with insight into successful industry practices. EMG data could suggest how to properly train a worker to limit the effects of fatigue by using alternative techniques. Finally, EMG data could also be used to investigate muscle impairment and dysfunction.

2.4 THORACOLUMBAR MOTIONS

Marras (2000) reports that there has been considerable debate in recent years regarding the benefits of implementing ergonomic principles and aids into the workplace. It can be argued that when monitoring workers, they often perform desired tasks at more optimal and safe levels, then when no outside observation is taking place. Despite these claims, research continues in industry to gain understanding into the causes of back pain.

There have been many risk factors demonstrated as potential mechanisms for injury. Heavy work loads, frequent bending and twisting of the trunk and whole-body vibration have all been well documented (Dolan et al., 2001; Marras et al., 1995; McGill, 1987). Marras et al. (1993, 1995) suggested dynamic trunk motions were related to spinal loading, and ultimately can be considered a biomechanical risk factor for LBP or overexertion injuries. As a result, the evaluation of trunk motion was thought to be important when making ergonomic improvements to lifting tasks. To successfully evaluate these dynamic motions in the workplace one must be able to quantify these motions. The Lumbar Motion Monitor (LMM) is an exoskeleton device. It is an electrogoniometer that measures displacement in the sagittal, lateral and twisting planes. Using differentiation techniques, the velocity and acceleration-time histories can be determined.

Marras et al. (1993, 1995) studied over 400 industrial lifting jobs in 48 varied industries, while collecting trunk motion characteristics using the Lumbar Motion Monitor. They quantified characteristics that were associated with an increase risk of occupationally related low back injuries. From these data they developed a model for LBP risk. They categorized each task as low, medium and high risk for a person to develop over-exertion back injuries. Gill et al. (1996) performed a study to find the reproducibility of the LMM for measures of range of motion, velocity and acceleration. The results indicate that overall the LMM was a valid means of producing reproducible measures. These findings provide further understanding as to why the LMM is still widely used in both industry and laboratories today.

2.5 WORK IN UNSTABLE ENVIRONMENTS

Studies dating back to McLeod et al. (1980) found that ship motions are directly related to poor performance of crew members. Whether it be a moving ship, offshore oil platform, or compliant floor structures, many people are employed to perform MMH tasks in unstable environments. An understanding of the environment in which these lifting tasks take place must be considered before deriving administrative controls to ensure the health and safety of people working in these environments. Wertheim (1998) suggested that many lifting tasks are dangerous when performed in a non moving environment, and the same lifting tasks under a moving environment may place operators at an even greater risk of injury. Kingma et al. (2003) reported that operators will demonstrate a reduction in performance when placed in a moving environment.

When working on a moving platform, motion induced performance decrements are often the result. The general effects of ship motion on the worker can be divided into motivational, energetic and biomechanical variables (Heiss et al., 2002).

2.5.1 Motivational and Energetic Variables

Wertheim (1998) described motion sickness as one of the most common and referred to phenomena in relation to a moving environment. He continues by commenting on how motion induced sickness (MIS) is a major motivational issue. He states that "motion sickness causes a massive lowering of motivation, usually resulting in a considerable slowing down of work rate, subsequently leading to a disruption in continuous work and often its complete abandonment" (pg. 1846). Heiss et al. (2002) also suggested that motivation among ship workers is often decreased due to motion sickness. To a certain extent, changes in lifting strategy can be employed to help minimize the effects of low back loading. A change in work strategies to help overcome performance problems due to motion induced sickness is limited.

Wertheim (1998) states that people doing physical work on ships (i.e. moving platforms) are more easily fatigued in comparison to the same work performed ashore. Heiss et al. (2002) states that the body constantly uses energy for the muscles to maintain or overcome loss of balance, which in turn causes motion induced fatigue (MIF). Working on a moving platform can be approximately twice as fatiguing as working in a static environment (Heiss et al., 2002). Lewis and Griffin (1995) suggested that a common fishing trawler may increase both energy expenditure and lumbar compression forces by factors of two or more during lifting. Obviously, the improvement of muscular endurance and physical fitness levels may help prolong the ill effects of fatigue, while performing MMH activities in a moving environment.

2.5.2 Biomechanical Variables

Biomechanical properties are often considered in the assessment of work in moving environments. Grinde (1985) administered a questionnaire to 878 Norwegian fishermen whom 77% of the respondents reported suffering from musculoskeletal problems. Interestingly, 51% of these problems were related to pain and discomfort specific to the lower back.

Torner et al. (1988) reported on symptoms related to musculo-skeletal injury related to working in a moving environment. They administered a questionnaire to 1243 people who were considered professional fisherman on the west coast of Sweden. The questionnaire included aspects of frequency of pain and discomfort, years in the profession, type of fishing, type of working tasks on board, physical workload and hours working. Results indicated that 74% of the fisherman responding to the questionnaire reported some kind of symptoms of musculoskeletal injury within the past 12 months. Fifty-Two percent of these were reported as symptoms of back pain. Torner et al. (1988) also asked the fisherman what factors they felt were the most stressful aspects of their occupation. Not surprisingly, ship motions were reported as the most stressful factor with 73% of those surveyed agreeing. Cold weather and the risk of tripping or slipping placed second, each having 71% agreement.

Torner et al. (1994) were interested in finding the effects of ship motions upon the biomechanical moments and forces exerted by and upon the body during MMH activities.

The results revealed that work on a moving platform causes over-stabilization of musculature deemed important for proper balance and stability. This over-stabilization could represent the excessive use of stabilizer muscles, as the operator is required to constantly contract these muscles to counteract unpredictable motions. The authors suggested that small platform motions were measured in the study, yet comparably larger moments and lumbar compression forces were calculated.

Kingma et al. (2003) also suggested that ship motions impose greater physical stresses compared to those performing similar work in stable environments. They examined the effect of ship accelerations on three-dimensional loading during lifting and pulling activities. This study acquired the accelerations experienced by a 120m frigate sailing at two different angles to the waves. Following the collection of lifting and pulling exertions under laboratory conditions, the ship acceleration profiles were superimposed upon the MMH activities' kinematics and kinetics in a pseudo inverse dynamics approach. Two sailing directions were used to produce the ship accelerations. Sailing took place at 90 degrees (waves came from the left) and 150 degrees (waves coming at an angle of 30 degrees to the left of the forward axis), while floor accelerations were recorded in two locations on the ship. They concluded that low back loading was only moderately affected by platform motions during symmetrical lifting. These findings were counter-intuitive and are likely explained by the fact that an empirical, rather than a modeling design is required under such circumstances. The authors acknowledged the following limitations: 1) angular accelerations should have been accounted for in the analysis, 2) larger ships, due to the inherent stability design generally have small acceleration profiles and perhaps motions from a smaller vessel should have been considered

Kingma et al. (2003) hypothesized that experienced seakeepers may be able to reduce loading on the spine if persons could consciously "time" the ship accelerations as they worked. Ideally, if a person could perform the upward phase of a lifting task as ship accelerations produced a downward phase, loading on the spine should be reduced. However, the authors conclude that it is almost impossible to time lifting and pulling movements in such a way that a substantial reduction of the total low back moment is obtained.

2.5.3 Description and Prediction of Motion Induced Interruptions

Wedge and Langlois (2003) suggested that humans often have difficulty moving in a controlled and coherent manner when ship motion is involved. Kingma et al. (2003) and Torner (1988 and 1994) consider musculo-skeletal injuries that could result from unfavorable motion environments. Other researchers have focused more on the biomechanical issues of loss of balance created by ship accelerations. Graham (1990) suggested that the loss of balance people often experience on a moving platform is the direct result of lateral and vertical accelerations in combination with the inclination of the deck. Balance issues often cause the worker to have problems completing desired tasks and are referred to as motion induced interruptions (MII). Crossland and Rich (2000) defined a MII as "when platform motions are sufficiently large to cause a person to slide or lose balance unless they temporarily abandon their allotted task in order to maintain postural stability".

Graham et al. (1992) developed a model to predict the number of MII a person would experience. Graham's model was a quasi-static, ridged body model, which tends to correlate well with experimental data. However, it seems to overestimate the actual number of observed MII. Consequently, models which assess deck motions to predict the incidence of MII tend to under predict the occurrence on an operator. Other MII models suggest that a person should be modeled as an inverted, articulated pendulum balanced upon two points of contact with the ground (Wedge and Langlois, 2003). This modeling approach suggests that once a person's center of mass falls outside the base of support, a stumble or MII will occur. However, these models cannot account for any conscious or reflexive corrective action an operator will take in order to maintain balance.

From a kinesiological prospective, humans rarely behave like an inverted pendulum, as anticipatory postural adjustments can easily be made to help counteract the balance issue (MacKinnon and Holmes, 2004). Of major concern is the fact that these models do not include MMH handling effects and operator experience. As a result, these models are generally restricted to the development of basic seakeeping criteria rather than being used to predict risk of injury. People who continually work in moving environments tend to adapt to their surrounds, which could impose even greater difficulties for a successful MII prediction methods.

2.5.4 Interaction between Lifting Technique and Platform Motion

When discussing the biomechanical issues surrounding performance loss, simple postural adjustments can be made to biomechanical properties that may help overcome the loss of balance often experienced in moving environments. Commissaris and Toussaint (1997), Heiss et al. (2002) , Hsiang et al. (1997) and Van Diëen et al. (2003), have all reported the squat lifting technique to be the most problematic lifting strategy with respect to balance. Quite often, humans perform the squat technique, by lifting the heels of the feel off the lifting platform, resulting in the possibility of injury due to unexpected perturbations (NIOSH, 1981). As a result, this type of lifting technique may prove harmful and should be avoided during lifting activities that occur in an offshore or moving environment where balance becomes a major concern.

Van Diëen et al. (1998) stated that an increase in spinal flexion, often caused by fatigue, may reduce ones ability to adjust in a timely manor to external perturbations, hence, leading to excessive loading of the spine. An increase in spinal flexion may also be the result of a stoop lifting technique, which could also influence the onset of fatigue. The result of a stoop lifting technique (increase spinal flexion), coupled with an increase of fatigue as an indirect result of ship motions could further support Van Diëen and colleagues suggestion of fatigue leading to excessive loading on the spine.

2.6 CONCLUSION

There has been a large amount of research into MMH activities and their subsequent involvement in injury mechanisms. Ferguson and Marras (1997) conclude that lifting, which is one of the most common MMH activities is a well documented risk factor for LBP. Hsiang et al. (1997) suggest that low back pain is an established problem that generates tremendous cost and suffering to both workers and employers. Klein et al. (1984) suggest that LBP is one of the most expensive medical problems in industry. As a result, the abundance of research aimed towards proper lifting techniques, workstation design and the resultant spinal loading has been well justified.

It can be concluded that both the characteristics of the lifting activity and the environment in which it is being performed should be considered when examining the risk of operator injury and accident. As concluded earlier, no single lifting technique should be suggested without proper evaluation of the required lifting activity. There has been limited research done on the effects of a moving platform on the incidence of injury. Even less research has been done, suggesting the effect a moving environment has on the execution of lifting tasks as well as the potential increased risks of back pain.

Chapter 3
METHODOLOGY

3.1 SUBJECTS

Nineteen healthy male subjects from a university population volunteered for this study. Subject characteristics are reported in Table 3.1. As part of the selection criteria, all participants confirmed they had no previous history of low back pain. The experiment was explained verbally to all participants and they were encouraged to ask questions. The subjects were reminded that they could discontinue the experiment at any time. Once subjects agreed to participate, they read and signed a consent form. The participant's personal information was gathered after the consent form was signed and a Physical Activity Readiness Questionnaire (PAR-Q) was completed. The study was approved by the Memorial University of Newfoundland Human Investigation Committee.

Table 3.1 – Subject demographics

	Mean	**SD**
Age (Years):	22.78	1.72
Stature (cm):	180.93	5.67
Mass (kg):	82.42	12.08

3.2 PROTOCOL

Subjects were asked to perform lifting tasks under three motion conditions and one no motion (i.e. stable) condition. While exposed to the four motion environments, subjects completed four repetitive lifting tasks.

3.2.1 Description of Load

Subjects performed all lifting tasks with a 15 kg load that was equipped with a uniaxial load cell that served as both a connector between the handle and the load and a means of measuring the load acting through the hands. The load apparatus was also equipped with a three-dimensional accelerometer mounted on the load cell near the point of contact with the hands (Figure 3.1). Both load acceleration and load cell data were sampled at 60 Hz using an IOtech data acquisition system (A-D board) (ACA Tmetrix, Mississauga, Ont.). The load could be easily gripped symmetrically by two handles. The dimensions of the load apparatus required the participant's hands to be placed on the designated handles, 330 mm apart from each other. The distance between the load and the handles was 375 mm.

3.2.2 Description of the Lifting Tasks

The 15 kg load was lifted under four different conditions. One condition, termed 'close floor' had the load lifted from the floor through a displacement of 750 mm. This lifting situation positioned subjects with a starting horizontal distance of 160 mm (from the participant's hands to the middle of their ankles) and a final horizontal distance of 600 mm, from hands to the middle of ankles (measured while the load was at its 750 mm height) (Figure 3.2a). Another condition, termed 'far floor' had the same 750 mm displacement height from the floor. However, the initial horizontal distance was 260 mm with a final horizontal distance of 700 mm (Figure 3.2b). The final two conditions consisted of the load being placed on a 250 mm platform (riser) and then displaced through the same 750 mm height. This lifting condition was performed with the same initial and final horizontal measurements as described for the close floor and far floor conditions, and were termed, 'close high' and 'far high' respectively (Figure 3.2c, Figure 3.2d). During all lifting conditions subjects' feet were positioned behind a starting line and were instructed to assume a shoulder width stance. All lifting tasks conformed to safe manual materials handling guidelines established by the National Institute for Occupational Health and Safety (NIOSH, 1981).

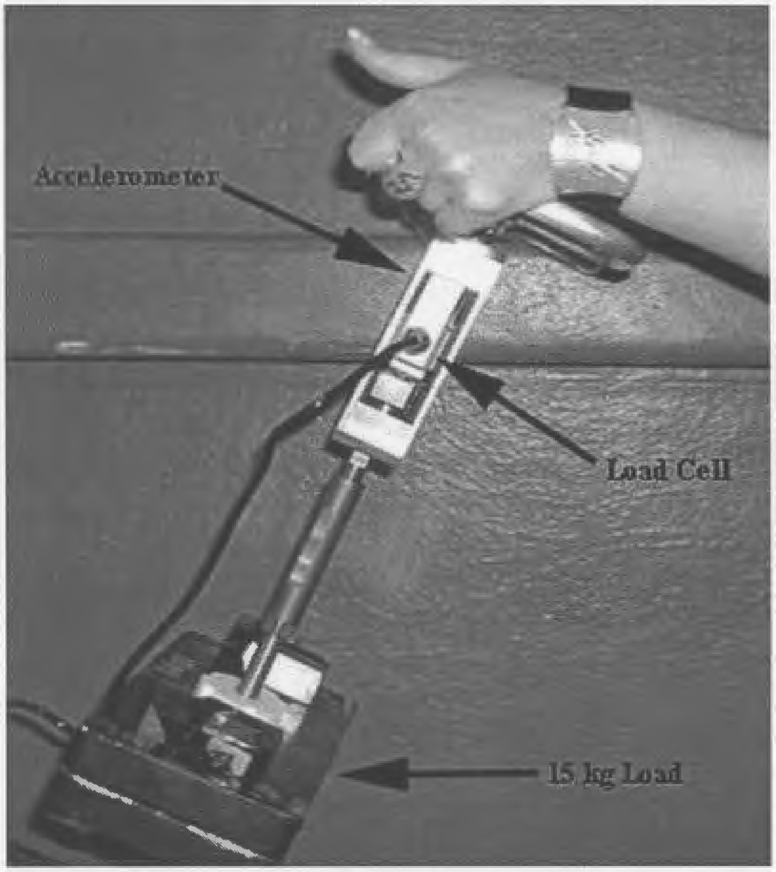


Figure 3.1: The load apparatus (the location of load cell, accelerometer and load mass are indicated)

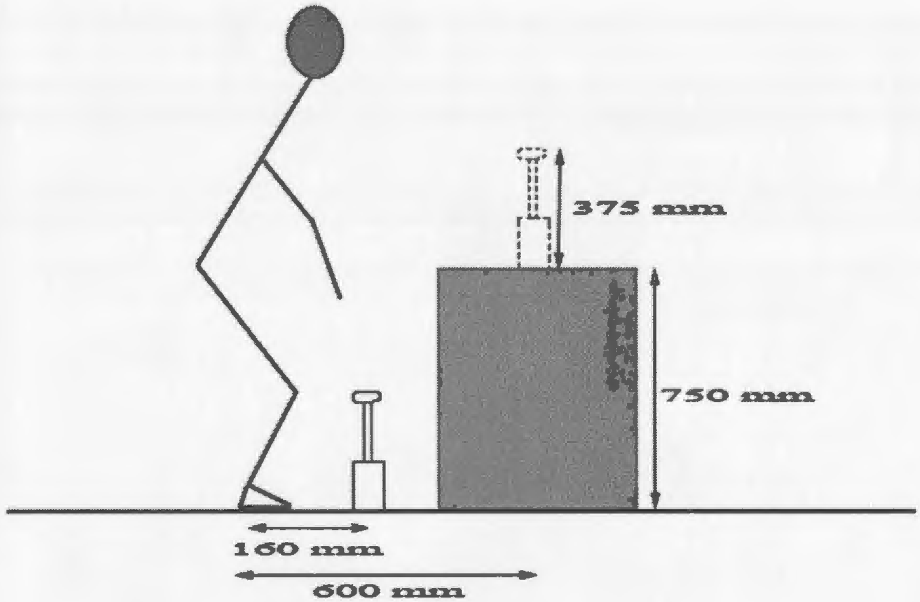


Figure 3.2 a: Close Floor lifting condition

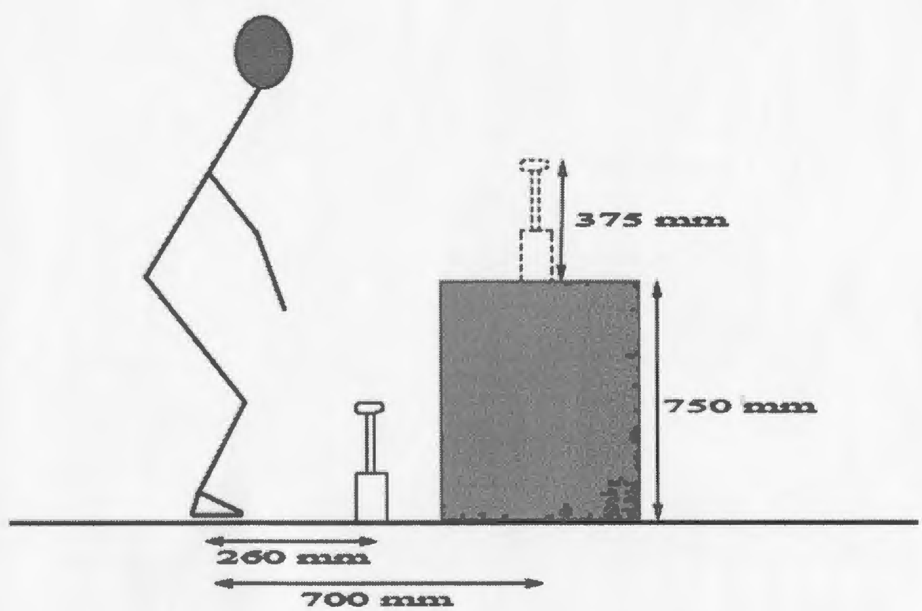


Figure 3.2 b: Far Floor lifting condition

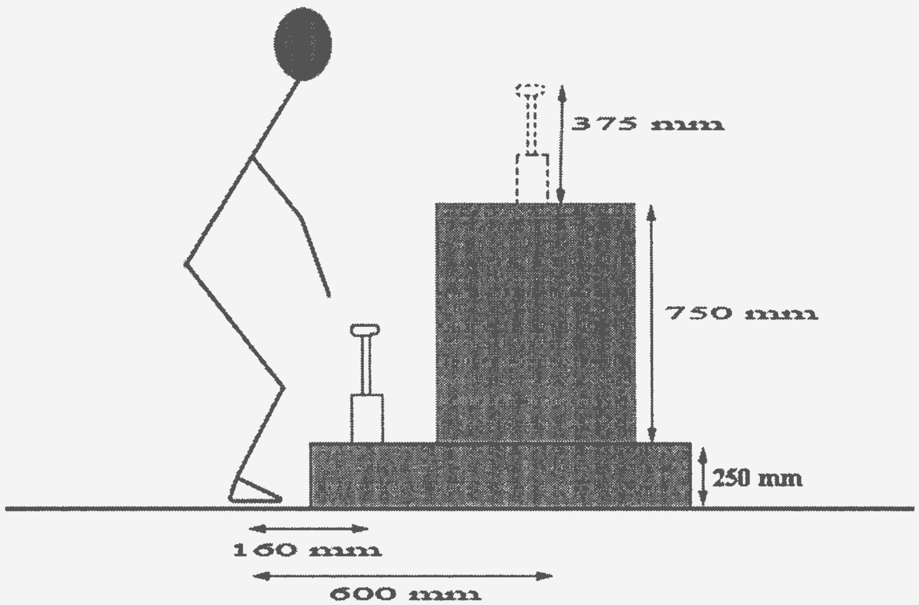


Figure 3.2 c: Close High lifting condition

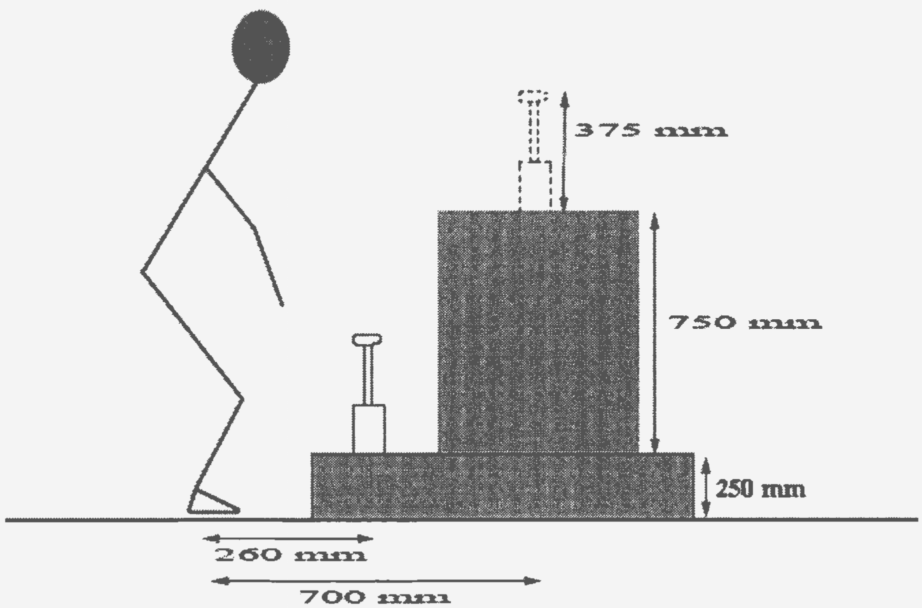


Figure 3.2 d: Far High lifting condition

3.2.3 Floor Motions

This study consisted of a no motion (i.e. stable floor) condition and three dynamic moving floor conditions. A 6 degree of freedom ship's motion simulator was used to create a moving environment for subjects to perform the lifting conditions (Figure 3.3). The motion based simulator is located at the Marine Institute's Centre for Marine Simulation, Memorial University of Newfoundland. These six degrees of freedom are described as; the x direction (surge) or rotation around the x-axis (roll); the y direction (sway) or rotation around the y-axis (pitch); and finally the z direction (heave) or rotation around the z-axis (yaw) (Figure 3.4). During all motion conditions, the simulator executed a motion profile based on a numerical model derived from an existing 45 foot coast guard supply vessel experiencing approximately seven metre wave (maximum) conditions with 5-10 second wave periods.

Table 3.2 – Simulator motions

		Roll-Rate	Pitch-Rate	Yaw-Rate
deg/s	+	9.47	3.86	0.95
	-	-9.87	-3.54	-1.12

For purposes of this experiment, the simulator produced a large roll motion (x-rate) as described in table 3.2. For further clarification of motion direction refer to figure 3.4. Subjects orientated themselves in 3 different positions on the simulator, which would consequently place them in three different motion states. These states included a roll motion (rotation about the x axis), a pitch motion (rotation about the y axis), where subjects positioned themselves 90 degrees from the roll position and finally a combination of pitch and roll, referred to as the quartering condition (45 degrees to the roll motion).

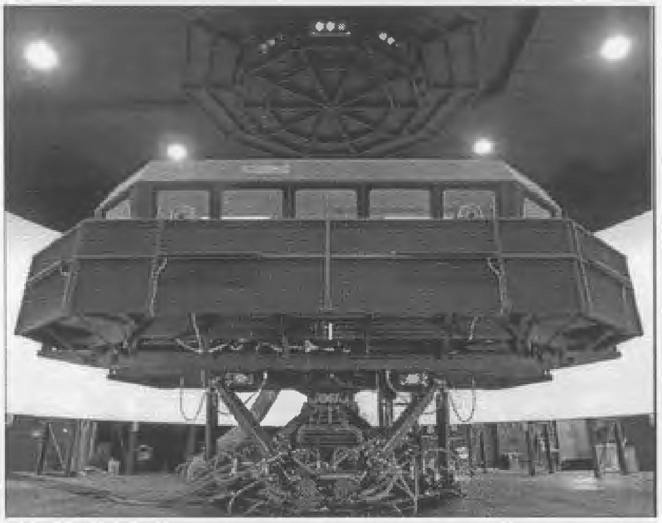


Figure 3.3: 6 degree of freedom motion based simulator

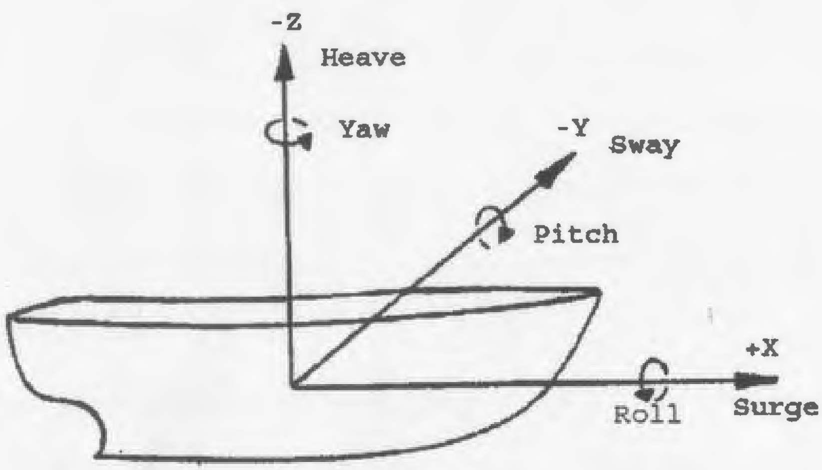


Figure 3.4: Description of motions produced about each principle axis

3.2.4 Experimental Sessions

Each subject attended three experimental sessions over a period of two weeks. The first two sessions took place at the Centre for Marine Simulation. During the first session, two randomly selected motion states were selected and subjects were oriented accordingly within the simulator. The same lifting tasks were performed with the remaining (third) motion state during the second session. All lifting conditions were completed in each of the motion states. Following completion of all motion sessions, subjects were required to attend a final session in the Biomechanics/Physiology laboratories at the School of Human Kinetics and Recreation, Memorial University of Newfoundland. During this session the stable, or no motion, condition was collected. Each motion condition took approximately 1.5 hours to complete and the laboratory session required approximately 45 minutes.

3.2.5 Measurement Equipment Preparation

Upon arrival for each session, the subject was prepared for electromyography (EMG) electrode placement. EMG activity was recorded on eight superficial muscles that were considered important for these lifting activities. Skin preparation for all electrodes included: removal of hair by shaving and removal of dead epithelial cells with an abrasive paper over the designated areas followed by cleansing with an isopropyl alcohol swab.

Surface electrodes (Kendall ® Medi-trace 133 series, Ag/AgCl, Chikopee, MA) were placed bilaterally on the subject's erector spinae (located at the fourth and fifth lumbar vertebrae), latissimus dorsi (located 2 cm below the inferior angle and 3 cm distally), external oblique (located at the midpoint between the ASIS and naval), and trapezius muscles (located 2 cm lateral of the midpoint between C2 and the acromion processes) (Figure 3.5 a-b). A ME3000P (Mega Electronics Ltd, Kuopio, Finland) unit and the MegaWin Version 1.21 software (Mega Electronics Ltd.) was used to collect the electromyography profiles. The EMG unit was connected to the communications port of a personal computer, via an optic cable, for online data collection. Each channel was sampled at 1000 Hz, band-pass filtered between 20 Hz and 500 Hz, amplified (differential amplifier, common mode rejection ratio $\geq$ 130 dB, gain x 1000, noise $\leq$ 1 μV) and analogue-to-digitally converted (12-bit), and stored on personal computer for further analysis. The amplification of the biological signal was done at the grounding electrode site, which effectively minimizes signal artifacts caused by movements and external noise.

(a)

(b)

Figure 3.5: (a) EMG electrode placement of Trapezius, Latissimus Dorsi and Erector Spaine muscles. (b) EMG electrode placement of external oblique muscles

Once the electrodes were mounted on the participant, the subject was required to perform a maximal voluntary contraction (MVC) for each muscle and was instructed to hold the MVC for a period of three seconds. Each MVC was conducted isometrically and performed twice for each muscle, with ample rest given between trials. The erector spinae MVC was obtained using a Modified Sorensen Back Extension Test (Biering-Sorensen, 1984). The latissimus dorsi MVC was obtained using a modified lat pull-down posture. The external oblique MVC was obtained using an oblique crunch. The trapezius MVC was obtained using a modified shoulder shrug. Postures employed to obtain the MVC were standardized across subjects and are depicted in Figure 3.6 a-d.

Following MVC's the subject was fitted with an AcuPath Industrial Lumbar Motion Monitor (LMM) (BIOMEC Inc. Cleveland, OH) (Figure 3.7). The LMM is an exo-skeleton device, employed to measure the displacement - time-series data for side bending, flexion/extension and rotation of the thoraco-lumbar spine region (Marras et al., 1992). LMM data were collected at a rate of 60 Hz. Standing heights of all subjects were carefully observed and the appropriate size adjustments were made on the LMM. Following size selection, the LMM was calibrated following manufacturer's instructions. This calibration or 'zeroing' was necessary to position the LMM in a neutral position. Calibration was performed before each motion condition.

Once equipment preparation was complete, the subject was reminded of the requirements for the lifting tasks. Each subject was given a randomly selected motion profile, as well as a random order for each of the four lifting conditions required.

(a)

(b)

(c)

(d)

Figure 3.6: MVC postures for the (a) Erector Spinae, (b) Latisimus Dorsi, (c) External Oblique and (d) Trapezius musculature

Figure 3.7: AcuPath Industrial Lumbar Motion Monitor (LMM)

3.2.6 Lifting Procedure

The subject was instructed to perform each lift using any desired lifting strategy, while keeping his feet fixed with a normal shoulder width stance. The subject was informed that once a lifting technique was adopted, to continue using the technique throughout the entire study. An audible signal was employed to direct the participant to execute a lift every 10 seconds. Once a lift was complete an experimenter lowered the load back to the desired starting position and the subject began preparation for the next lift. Each trial lasted approximately two minutes, for a total of approximately 12 lifts.

Sufficient rest was given between each of the four lifting situations, minimizing the potential effects of fatigue.

3.3 DATA ANALYSIS

3.3.1 Determining a Lift

An event marker, connected to a channel on the A/D board, was used to indicate when a person began execution of a lift and was held until the load had reached the final position. To determine the beginning and end of a successful lift, the event marker, strain gauge and accelerometer data were all considered in identifying the start and end of the lift cycle. A lift cycle was defined as when the load first leaves the ground until final contact at the lift destination. When the load cell recorded a reading above 15 kg (mass of the load being lifted) it was believed at this instance the load was lifted off the ground (or platform). A sudden increase in the vertical acceleration was observed as the load made contact at its destination. The data point prior to this large increase was used as the endpoint for completion of the lift.

A second event marker, connected to one channel of the A/D board, was used to indicate when a person experienced a motion induced interruption (MII). A MII was recorded when a participant was unable to complete the desired lift without moving his feet to help maintain balance. Only successful lifts, those without a MII, were subsequently analyzed.

3.3.2 Synchronization of the Data Streams

During data collection the A/D collection was first started, followed simultaneously by the LMM and EMG data streams. Thus these time histories required synchronization to determine the start of the trial. Figure 3.8 depicts how data channels were synchronized at the start of the trial. It should be noted that the A/D and LMM sampled data at 60 Hz while the EMG was sampled at 1000 Hz and thus data streams had to be normalized in time prior to further analysis.

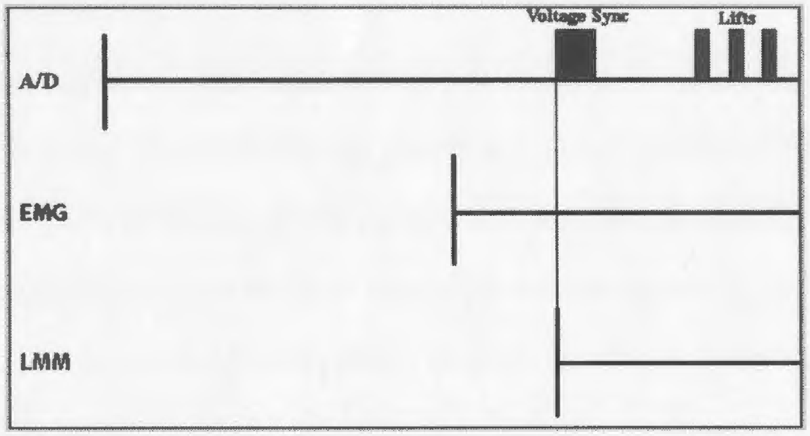


Figure 3.8: Depiction of how data streams where synchronized to the start of each lifting trial

3.4 STATISTICAL ANALYSIS

Data were analysed by a repeated measures ANOVA (4x4) (SPSS 11.5 for windows, SPSS Inc., U.S) to determine whether there were significant main effects or interactions for motion states and lifting conditions.

Chapter 4
Results

4.1 INTRODUCTION

Subjects performed four lifting tasks while exposed to 3 platform motions (roll, quarter and pitch) and a stable laboratory condition (i.e. no motion). Lifting tasks were named according to load height and feet placement characteristics. The lifting tasks close high, far high, close floor and far floor have been abbreviated as CH, FH, CF, and FF respectively and are used in figure and table descriptions. Similarly for figure and table legends, muscle names are abbreviated: erector spinae (ES), latissimus dorsi (LAT), trapezius (TRAP) and external oblique (OBLIQ) and the left and right side are referred to as L and R. These data represent successful lifting attempts only; excluded are attempts during these trials which the participant stumbled or had to make abrupt postural adjustments to maintain balance.

Table 4.1 includes the mean (standard deviation) time required to complete a lift for all subjects across conditions. A repeated measures ANOVA revealed that there were no significant differences in the time to complete a lift across the four motion conditions. However, when assessing lifting tasks the close high condition took a significantly longer amount of time to complete a lift than all other lifting conditions ($p \leq 0.01$). The post h c analysis revealed no other significant pairwise differences in the times of the lifting tasks.

Table 4.1: Mean time (standard deviation) in seconds to complete a lift

Motion State	**Lab**	**Roll**	**Quarter**	**Pitch**
Mean	1.25	1.24	1.22	1.25
SD	0.14	0.15	0.16	0.17
Lifting Condition	**CH**	**FH**	**CF**	**FF**
Mean	1.35	1.21	1.18	1.21
SD	0.46	0.34	0.34	0.40

4.2 ELECTROMYOGRAPHY (EMG) RESULTS

The raw EMG signal was full-wave rectified and low-pass filtered at 4 Hz (2^{nd} order butterworth). The mean and maximum values are expressed as a percentage of the maximal voluntary contraction of each muscle tested. The largest mean MVC of the two isometric contractions were used to normalize the EMG values. The electromyography data were reduced further so that the mean and maximum values, as well as relative time (i.e. % of lift cycle) when the maximum EMG occurred in the time history were determined for each lift.

Table 4.2 is a summary of the repeated measures ANOVA analysis that considered overall condition and motion effects for maximum EMG values, mean EMG values and when the maximum EMG signal occurred in the lift cycle, for all muscles. There were no significant differences found during motion states for any of the muscles monitored in this study. There were differences found between the four lifting conditions, which are further described below.

Table 4.2: Overall condition and motion effects

MAXIMUM EMG

	LES	RES	LLAT	RLAT	LTRAP	RTRAP	LOBLIQ	ROBLIQ
Lifting Condition	*	**	*	**	**	**	*	*
Motion	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*

MEAN EMG

	LES	RES	LLAT	RLAT	LTRAP	RTRAP	LOBLIQ	ROBLIQ
Lifting Condition	*	**	**	**	**	**	**	**
Motion	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*

PERCENTAGE OF LIFT CYCLE WHEN THE MAXIMUM EMG VALUE OCCURRED

	LES	RES	LLAT	RLAT	LTRAP	RTRAP	LOBLIQ	ROBLIQ
Lifting Condition	*NS*	*NS*	*NS*	*NS*	**	**	*NS*	*NS*
Motion	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*	*NS*

(* = $p \leq 0.05$, ** = $p \leq 0.01$, *NS* = No significance)

4.2.1 Bilateral Muscle Recruitment Differences

Muscle activity was monitored bilaterally on all subjects. A series of paired-sample t-tests were used to determine if the maximum and mean EMG activities, as well as when the maximum occurred were significantly different between the left and right musculature. Left and right musculature differences were considered in this study because significant differences could reflect muscular imbalances. If imbalances are large enough, asymmetrical loading on the relevant soft tissues may occur, leading to increases in injury. These results are presented in tables 4.3, 4.4 and 4.5, respectively.

4.2.1.1 Maximum EMG Differences

Table 4.3 reports the summary of the statistical analyses for the data described in this section. The left and right external oblique muscles demonstrated no significant differences during all motion states, and during all lifting conditions. The left and right erector spinae muscles demonstrated a significant difference for the close high lifting task during the pitch motion state ($p \leq 0.01$). The left and right trapezius muscles demonstrated a significant difference during the far floor lifting task during the roll motion state ($p \leq 0.05$). The latissimus dorsi muscle demonstrated consistent differences between the left and right side activities throughout the experimental conditions. During the close floor lifting condition the latissimus dorsi showed a significant difference during the roll motion ($p \leq 0.01$), and $p \leq 0.05$ for both quarter and pitch. The far floor lifting condition produced differences between the left and right latissimus dorsi during both the roll and quarter motion states ($p \leq 0.01$ and $p \leq 0.05$, respectively). The close high lifting condition also had the left and right latissimus dorsi muscles to be significantly different during both the roll and quarter motion states ($p \leq 0.05$). The far high lifting condition had the left and right latissimus dorsi significantly different during the pitch and quarter motion states ($p \leq 0.05$).

Table 4.3: Statistical summary of maximum EMG differences between the left (L) and right (R) side musculature for each lifting condition and motion state

Close High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*	*NS*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	**	*NS*	*NS*	*NS*
Far High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	*NS*	*	*NS*	*NS*
Close Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	**	*NS*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	*NS*	*	*NS*	*NS*
Far Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	**	*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	*NS*	*NS*	*NS*	*NS*

(* = $p \leq 0.05$, ** = $p \leq 0.01$, *NS* = No significance)

4.2.1.2 Mean EMG Differences

Table 4.4 reports the summary of the statistical analyses for the data described in this section. The external oblique muscles demonstrated no significance during all motion states, and lifting conditions. The left and right trapezius demonstrated a significant difference during the far floor lift under the roll condition ($p \leq 0.05$).

During the close high lifting condition, the left and right erector spinae had significant differences during the pitch motion state ($p \leq 0.05$). The far high condition demonstrated significant differences in both the quarter and pitch motion states ($p \leq 0.05$). Finally, the close floor and far floor conditions had no significant differences between the left and right erector spinae during any motion states.

During the close high and far high lifting condition, the left and right latissimus dorsi had a significant difference for the quarter motion state ($p \leq 0.05$ and $p \leq 0.01$, respectively). The close floor condition demonstrated significant differences during the roll and quarter motion states ($p \leq 0.01$ and $p \leq 0.05$, respectively). The far floor condition had significant differences of during both the roll and quarter motion states ($p \leq 0.01$). Finally the far floor condition had a significant difference in the pitch motion ($p \leq 0.05$).

Table 4.4: Statistical summary of mean EMG differences between the left (L) and right (R) side musculature for each lifting condition and motion state

Close High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	*	*NS*	*NS*	*NS*
Far High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*	**	*NS*	*NS*
Pitch	*	*NS*	*NS*	*NS*
Close Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	**	*NS*	*NS*
Quarter	*NS*	*	*NS*	*NS*
Pitch	*NS*	*NS*	*NS*	*NS*
Far Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	**	*	*NS*
Quarter	*NS*	**	*NS*	*NS*
Pitch	*NS*	*	*NS*	*NS*

(* = p $\leq$0.05, ** = p $\leq$0.01, *NS* = No significance)

4.2.1.3 Left and right side differences in the time (expressed as a percent of lift cycle) that the maximum EMG value occurred

Table 4.5 reports the summary of the statistical analyses for the data described in this section. Both the left and right erector spinae and latissimus dorsi muscles demonstrated no significant differences during any motion state or lifting condition. The left and right trapezius muscle had a significant difference during the far high lifting condition and quarter motion state only ($p \leq 0.01$). Finally, the external oblique muscles produced a significant difference during the far high and quarter setup, while also expressing significance during the close floor, lab trials ($p \leq 0.01$ and $p \leq 0.05$, respectively).

Table 4.5: Statistical summary of relative time the maximum EMG differences occurred (as percent of lift cycle) between the left (L) and right (R) side musculature for each lifting condition and motion state

Close High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*NS*	*NS*	*NS*
Pitch	*NS*	*NS*	*NS*	*NS*
Far High				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*NS*	**	**
Pitch	*NS*	*NS*	*NS*	*NS*
Close Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*NS*	*NS*	*NS*
Pitch	*NS*	*NS*	*NS*	*NS*
Far Floor				
	LES/RES	LLAT/RLAT	LTRAP/RTRAP	LOBLIQ/ROBLIQ
Lab	*NS*	*NS*	*NS*	*NS*
Roll	*NS*	*NS*	*NS*	*NS*
Quarter	*NS*	*NS*	*NS*	*NS*
Pitch	*NS*	*NS*	*NS*	*NS*

(* = $p \leq 0.05$, ** = $p \leq 0.01$, *NS* = No significance

4.2.2 Maximum EMG Values

Figures 4.1, 4.2, 4.3 and 4.4 describe the maximum EMG values (and standard deviations) of the left and right erector spinae, latissimus dorsi, trapezius and external oblique muscles, respectively. These values are expressed as a percentage of MVC for each lifting condition, during each of the motion states. Due to the number of pair-wise comparisons performed in the post hoc analyses (and for clarity purposes), these differences are described in the text rather than on the figures themselves throughout the remainder of this chapter.

4.2.2.1 Erector Spinae Musculature

Figure 4.1 contains the data described in this section. In both the left and right muscles, the far high and close floor lifting condition produced maximum EMG activities that were significantly less than the far floor condition ($p \leq 0.01$). During the close high condition the left erector spinae demonstrated significantly less muscle activity when compared to both the far high and far floor conditions ($p \leq 0.05$ and $p \leq 0.01$, respectively). The right erector spinae demonstrated the same results, while also expressing the close high condition to be significantly less than the close floor condition ($p \leq 0.01$).

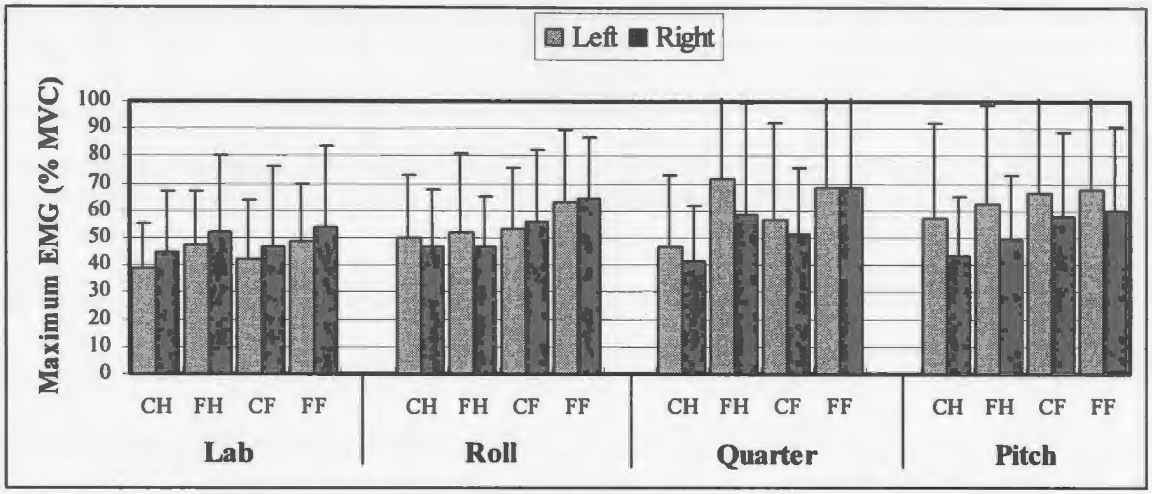


Figure 4.1: Maximum EMG values for the Erector Spinae muscle

4.2.2.2 Latissimus Dorsi Musculature

Figure 4.2 contains the data described in this section. Significant differences were found between the far high and close floor condition as well as the close floor and far floor conditions ($p \leq 0.01$). The left latissimus, during the close high condition, produced less muscle activity than both the far high and close floor conditions ($p \leq 0.05$ and $p \leq 0.01$, respectively). For the close high condition, the right latissumus produced less muscle activity than both the far high and far floor conditions ($p \leq 0.01$ and $p \leq 0.01$, respectively).

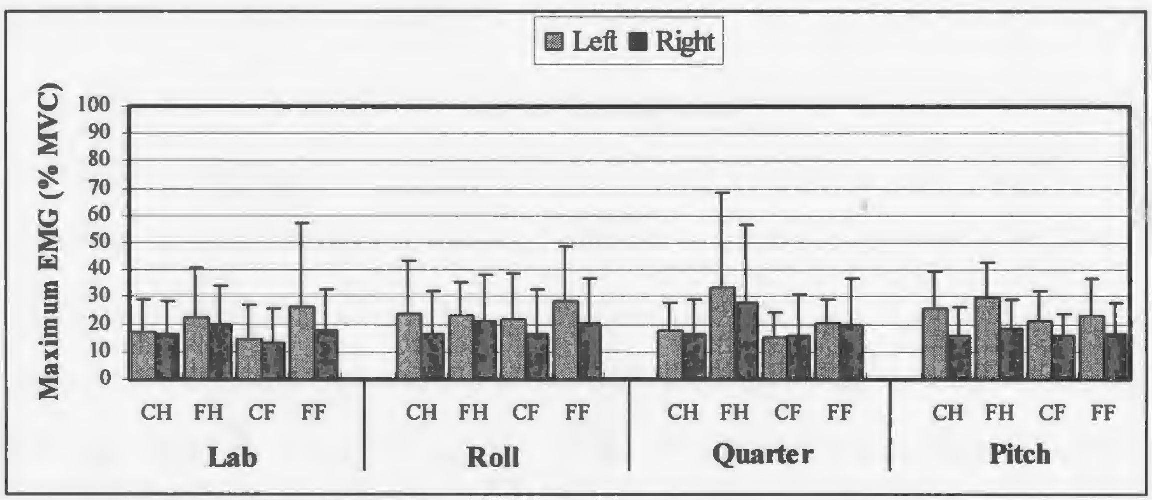


Figure 4.2: Maximum EMG values for the Latissimus Dorsi muscle

4.2.2.3 Trapezius Musculature

Figure 4.3 contains the data described in this section. The left and right trapezius had greater maximum muscular activity during the close high condition than the close floor condition ($p \leq 0.01$). Both the left and right also demonstrated the far high condition to have more activation than the close floor condition ($p \leq 0.01$). There was also greater trapezius activity during the far floor condition than the close floor condition ($p \leq 0.05$ for the right and $p \leq 0.01$ for the left).

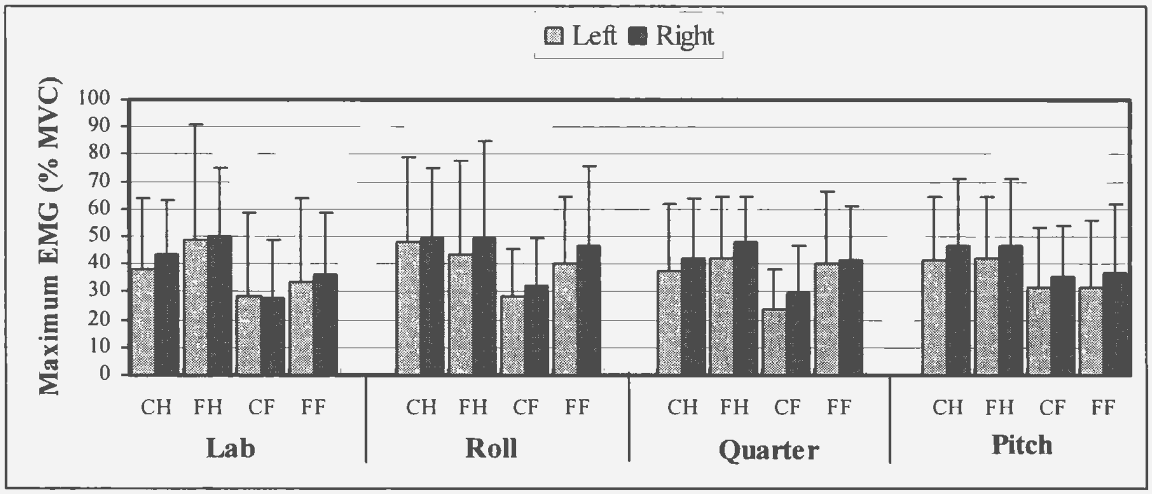


Figure 4.3: Maximum EMG values for the Trapezius muscle

4.2.2.4 External Oblique Musculature

Figure 4.4 contains the data described in this section. The left external oblique muscle had a significantly greater maximum activity during the far high condition when compared to the close high condition ($p \leq 0.05$). The far floor condition also had significantly greater activity than the close floor condition ($p \leq 0.05$). The right external oblique had significantly greater activity during the far high condition than when compared to the close high condition ($p \leq 0.01$). The far floor condition also had significantly greater activity when compared to the close high condition ($p \leq 0.05$). The far high condition had significantly greater activity than the close floor condition and the far floor had greater activity than close floor condition ($p \leq 0.05$ and $p \leq 0.01$, respectively).

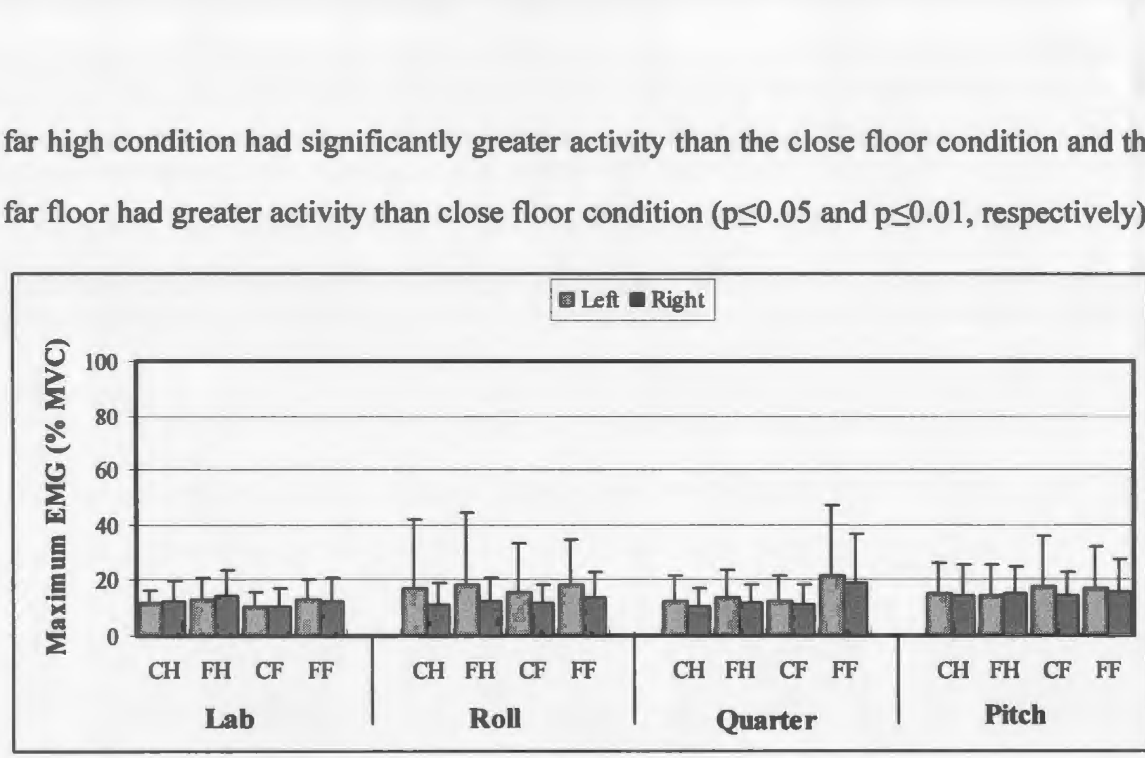


Figure 4.4: Maximum EMG values for the External Oblique muscle

4.2.3 Mean EMG Values

Figures 4.5, 4.6, 4.7 and 4.8 below represent mean EMG values (and standard deviations) of the left and right erector spinae, latissimus dorsi, trapezius and external oblique musclues, respectively. These values are expressed as a percentage of MVC for each lifting condition, during each of the motion states.

4.2.3.1 Erector Spinae Musculature

Figure 4.5 contains the data described in this section. The left erector spinae muscle demonstrated significantly less activation during the close high condition when compared to the far high, close floor or far floor conditions ($p \leq 0.05$, $p \leq 0.01$ and $p \leq 0.001$, respectively). The far floor condition was significantly greater than both the far high and close floor conditions ($p \leq 0.05$ and $p \leq 0.01$). The right erector spinae muscle also demonstrated similar results to the left erector spinae. The close high condition was significantly less than the far high, close floor and far floor conditions ($p \leq 0.01$). The far floor condition was similar to the left erector and again significantly greater than the far high and close floor conditions ($p \leq 0.01$). The far high condition had significantly less activation than the close floor conditions ($p \leq 0.05$).

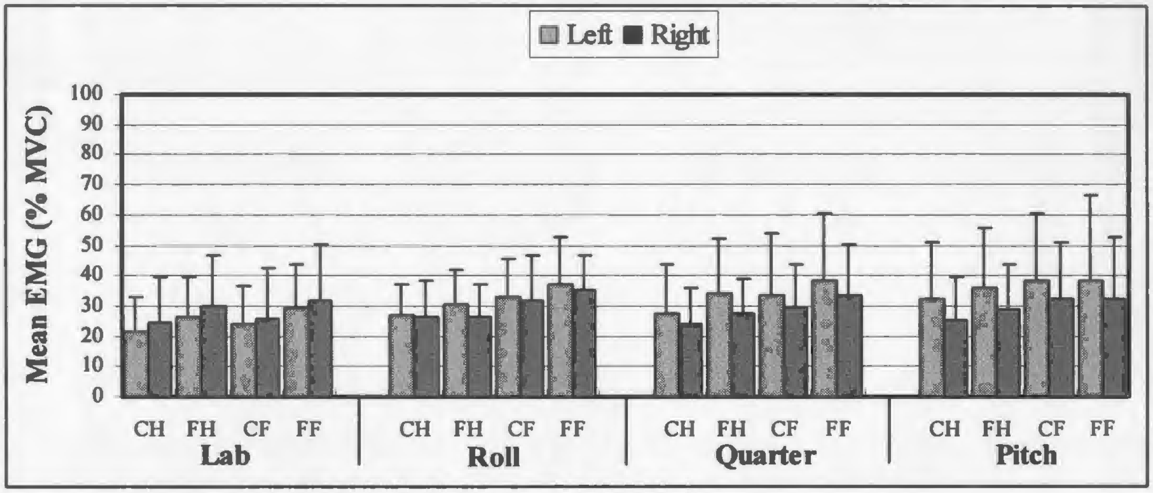


Figure 4.5: Mean EMG values for the Erector Spinae muscle

4.2.3.2 Latissimus Dorsi Musculature

Figure 4.6 contains the data described in this section. The left and right latissimus dorsi, demonstrated the close high condition to be significantly less than the far high condition ($p \leq 0.05$ and $p \leq 0.01$ respectively). The close floor condition was significantly less than both the far high and far floor conditions ($p \leq 0.01$). Finally, the right latissimus dorsi had the close high condition significantly less than the far floor conditions ($p \leq 0.01$).

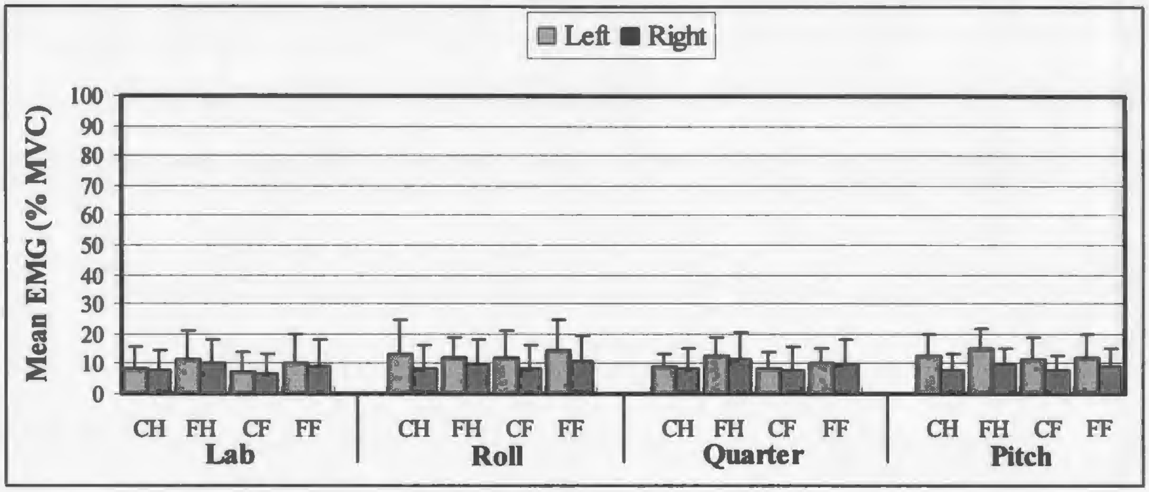


Figure 4.6: Mean EMG values for the Latissimus Dorsi muscle

4.2.3.3 Trapezius Musculature

Figure 4.7 contains the data described in this section. The trapezius demonstrated greater mean EMG activity during the far high condition when compared to the close floor and far floor conditions ($p \leq 0.01$). The left and right trapezius had the close high condition being significantly greater than the far floor condition ($p \leq 0.01$ and $p \leq 0.05$, respectively). Both the left and right trapezius also demonstrated a greater significance when comparing the close high to close floor lifting condition ($p \leq 0.01$).

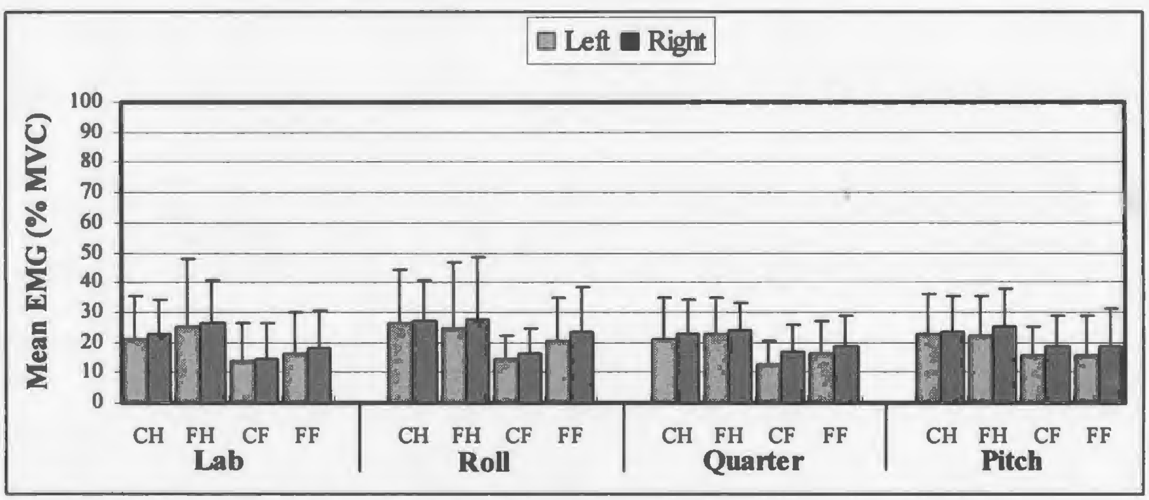


Figure 4.7: Mean EMG values for the Trapezius muscle

4.2.3.4 External Oblique Musculature

Figure 4.8 contains the data described in this section. The right external oblique muscle demonstrated mean EMG activity to be less during close floor conditions, when compared to far high and far floor conditions ($p \leq 0.01$). The close high condition was significantly less than the far floor conditions ($p \leq 0.01$). The left external oblique showed the close high condition to be significantly less than the far high and far floor conditions ($p \leq 0.05$ and $p \leq 0.01$, respectively). The close floor condition was also significantly less than the far high and far floor conditions ($p \leq 0.01$ and $p \leq 0.05$, respectively).

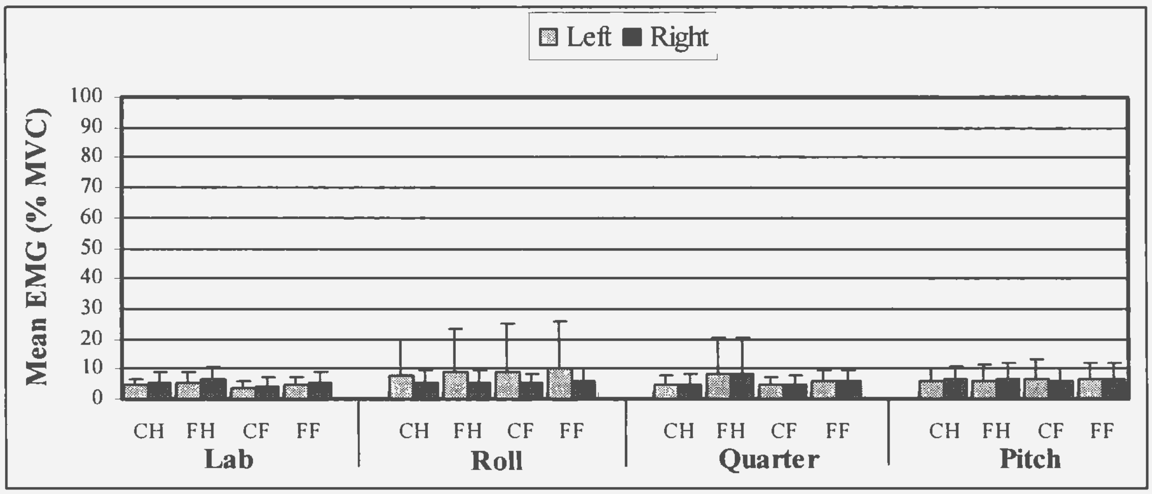


Figure 4.8: Mean EMG values for the External Oblique muscle

4.2.4 The time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred

Figures 4.9, 4.10, 4.11 and 4.12 below represent the time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred for the left and right erector spinae, latissimus dorsi, trapezius and external oblique muscles, respectively.

The left and right erector spinae, latissimus dorsi and external oblique muscles demonstrated no significant differences during lifting conditions and motion states. The left and right trapezius both had lifting condition close high to be significantly different than all other lifting conditions ($p \leq 0.01$). The left trapezius maximum EMG occurred significantly earlier in the close floor lifting condition compared to the far floor condition ($p \leq 0.05$).

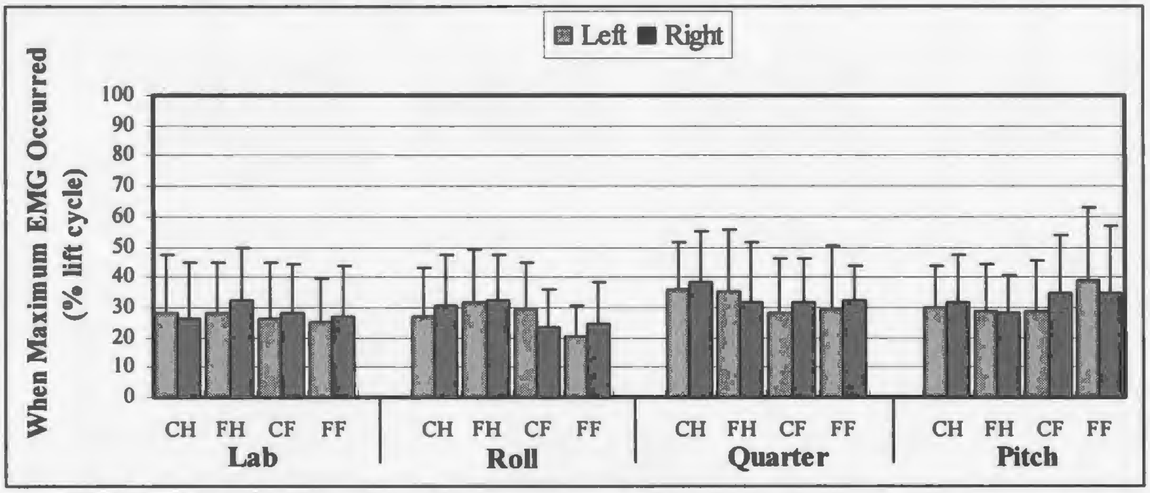


Figure 4.9: The time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred for the Erector Spinae muscle

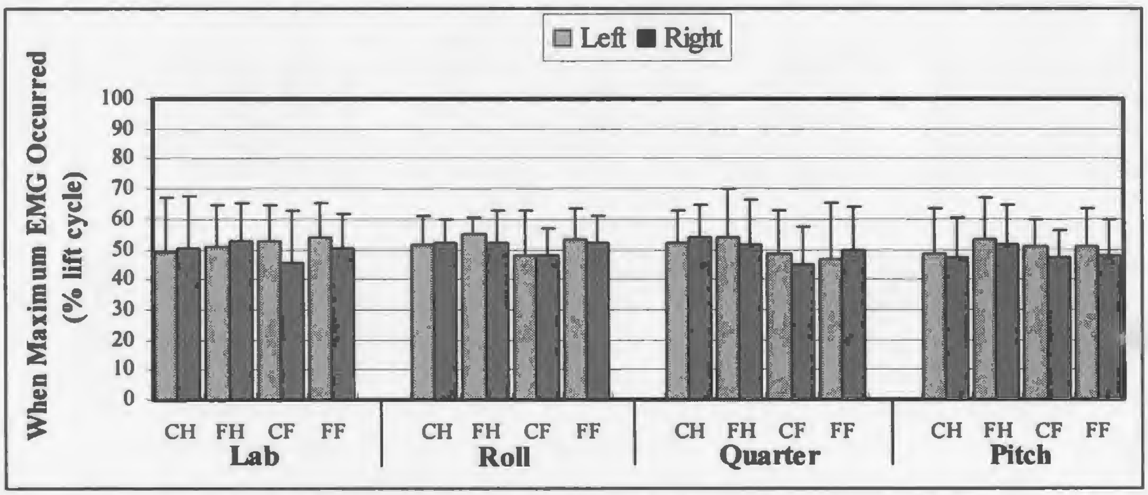


Figure 4.10: The time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred for the Latissimus Dorsi muscle

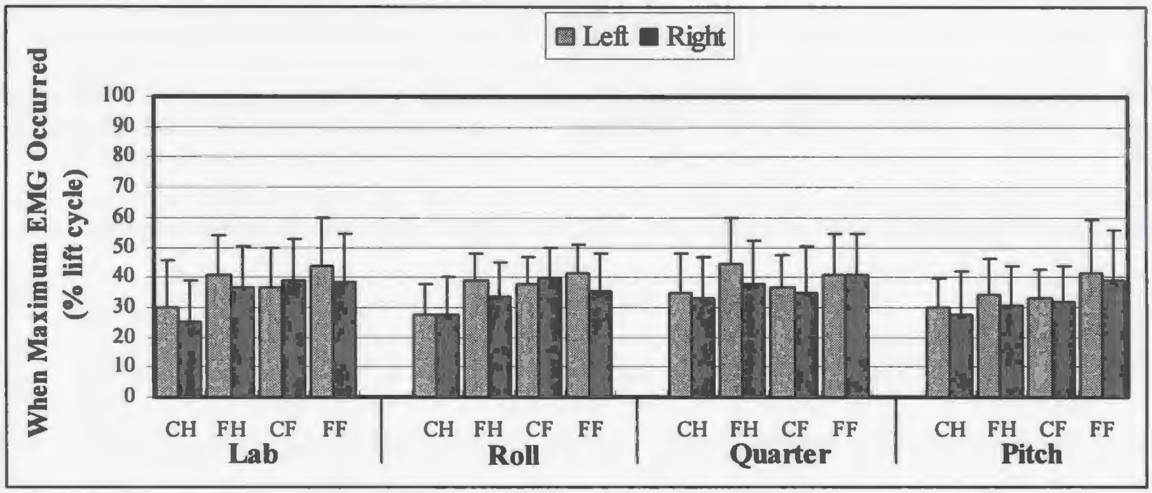


Figure 4.11: The time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred for the Trapezius muscle

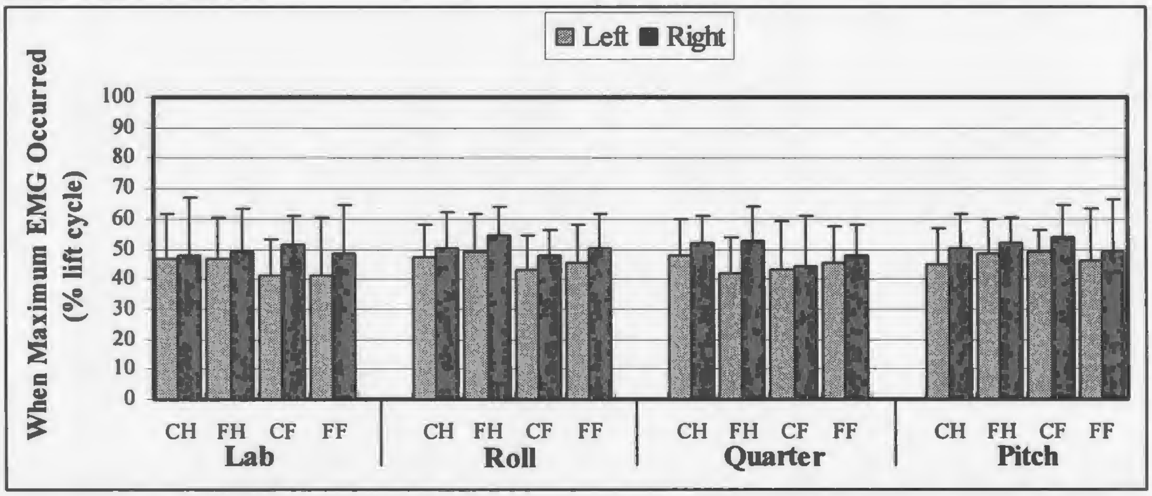


Figure 4.12: The time (expressed as a percent of lift cycle duration) that the maximum EMG value occurred for the External Oblique muscle

4.3 LUMBAR MOTION MONITOR (LMM) RESULTS

The LMM collected thoracolumbar displacement data in its three planes of motion (sagittal, lateral and twisting planes). The velocity-time profiles were derived using a first order differentiation technique. The velocity data were reduced so that for each trial the mean and maximum values, as well as relative time (i.e. % of lift cycle) when the maximum velocity occurred in the time history were determined.

Tables 4.6 is a summary of the repeated measures ANOVA analysis that considered the overall condition and motion effects for maximum LMM values, mean LMM values and % of lift cycle for all planes of motion. Simply for comparison purposes, a page is devoted (displaying all three planes of motion) to each of maximum velocities, mean velocities and when the maximum velocity occurred.

Table 4.6: Overall condition and motion effects

MAXIMUM LMM VELOCITY

	Sagittal	Twist	Lateral
Lifting Condition	*	*	*
Motion	*	**	*

MEAN LMM VELOCITY

	Sagittal	Twist	Lateral
Lifting Condition	**	*	*NS*
Motion	*	**	*

PERCENTAGE OF LIFT CYCLE TIME WHEN THE MAXIMUM LMM VELOCITY OCCURRED

	Sagittal	Twist	Lateral
Lifting Condition	*NS*	*NS*	*
Motion	*	*NS*	*NS*

(* = $p \leq 0.05$, ** = $p \leq 0.01$, *NS* = No significance)

4.3.1 Sagittal Velocity

4.3.1.1 Maximum Velocity

Figure 4.13a represents maximum sagittal velocities (and standard deviations) for each lifting condition and motion state. With respect to the maximum sagittal velocities, both the close high and far high conditions were significantly smaller than the close floor and far floor conditions ($p \leq 0.05$). During the lab condition, the maximal sagittal velocity was significantly greater than during all motion states ($p \leq 0.01$). The sagittal velocity also demonstrated significantly greater values for the roll motion when compared to the pitch motion ($p \leq 0.05$).

4.3.1.2 Mean Velocity

Figure 4.14a represents mean sagittal velocities (and standard deviations) for each lifting condition, during each of the motion states. Results for the mean sagittal velocities showed the far high lifting condition to be significantly smaller than both the close floor and far floor conditions ($p \leq 0.01$). The close high lifting condition was also significantly smaller than both the close floor and far color conditions ($p \leq 0.01$ and $p \leq 0.05$, respectively). Both the roll motion and lab condition was significantly greater than the pitch motion ($p \leq 0.05$ and $p \leq 0.01$, respectively).

4.3.1.3 Time (expressed as percent of lift cycle) that the maximum LMM velocity occurred

Figure 4.15a represents the time (expressed as a percent of lift cycle time) that the maximum sagittal velocities (and standard deviations) occurred, for each lifting condition and motion state. There were no significant differences for the relative time when the maximum velocity occurred, for all lifting conditions. During the sagittal velocity, the peak velocity occurred significantly earlier in the lab trial compared to the pitch and quarter motions ($p \leq 0.01$ and $p \leq 0.05$, respectively).

4.3.2 Twisting Velocity

4.3.2.1 Maximum Velocity

Figure 4.13b represents maximum twisting velocities (and standard deviations) for each lifting condition, during each of the motion states. The maximum twisting velocities found the far high condition to be significantly lower than the close floor and far floor conditions ($p \leq 0.05$). Results revealed that maximum twisting velocities for the lab trials were significantly smaller than all motion trials ($p \leq 0.05$ and $p \leq 0.01$ for the roll motion). There were no significant differences produced among motion states.

4.3.2.2 Mean Velocity

Figure 4.14b represents mean twisting velocities (and standard deviations) for each lifting condition, during each of the motion states. The only difference found during mean twisting velocities was the far high condition being significantly lower than the far floor conditions ($p \leq 0.05$). Results for the maximum twisting velocity revealed the lab trials to be significantly smaller than all motion trials ($p \leq 0.05$).

4.3.2.3 Time (expressed as percent of lift cycle) that the maximum LMM velocity occurred

Figure 4.15b represents the time (expressed as a percent of lift cycle time) that the maximum twisting velocities (and standard deviations) occurred for each lifting condition, during each of the motion states. There were no significant differences in the relative time when the maximum velocity occurred during the lifts for the twisting velocities. However, during motion conditions, the peak twist velocities occurred significantly earlier during the lab trials when compared to the roll motion trials ($p \leq 0.05$).

4.3.3 Lateral Velocity

4.3.3.1 Maximum Velocity

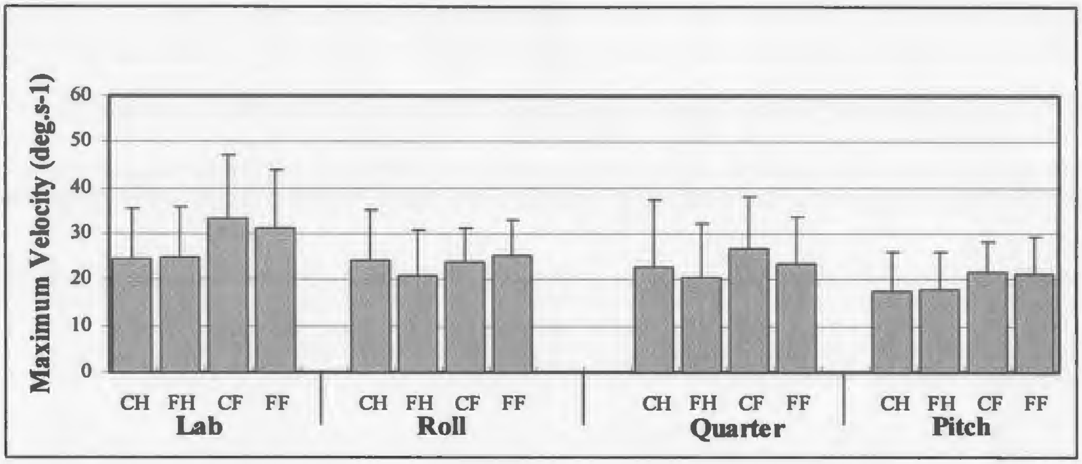
Figure 4.13c represents maximum lateral velocities (and standard deviations) for each lifting condition, during each of the motion states. The far high lifting condition was significantly smaller than the close floor and far floor conditions ($p \leq 0.05$). The lateral velocity demonstrated the smallest values during the lab condition than during any other motion state ($p \leq 0.05$).

4.3.3.2 Mean Velocity

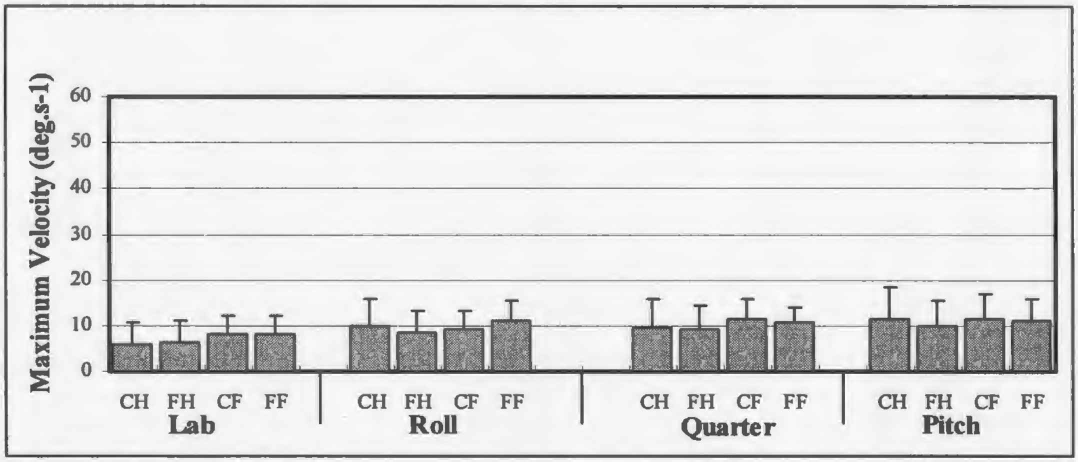
Figure 4.14c represents mean lateral velocities (and standard deviations) for each lifting condition, during each of the motion states. There were no significant differences for the mean lateral velocities for any of the lifting conditions. The lab trials were significantly smaller than all other motion states ($p \leq 0.05$).

4.3.3.3 Time (expressed as percent of lift cycle) that the maximum LMM velocity occurred

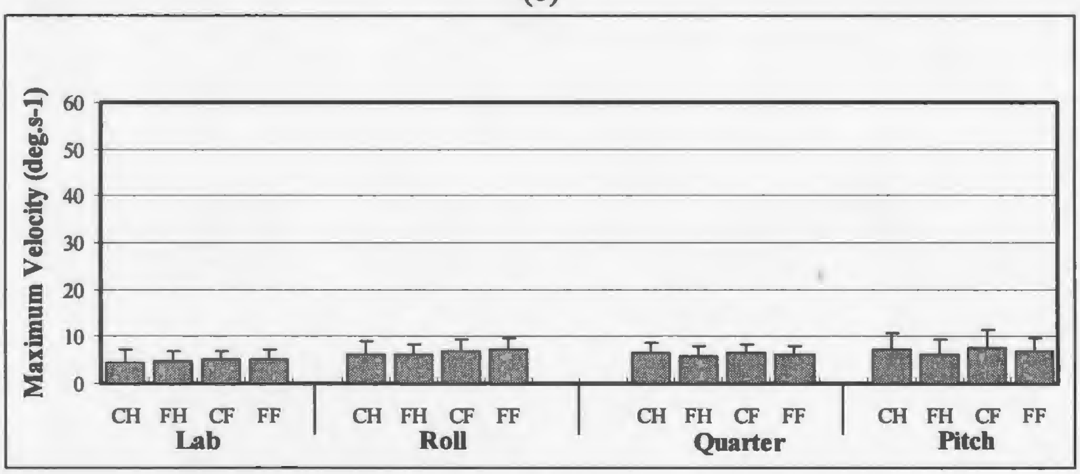
Figure 4.15c represents the time (expressed as a percent of lift cycle time) that the maximum lateral velocities (and standard deviations) occurred, for each lifting condition, during each of the motion states. The lateral velocities demonstrated a significantly greater amount of time during the far high condition when compared to the far floor condition ($p \leq 0.05$). The maximum velocity occurred significantly earlier in the lab trial than the roll motion ($p \leq 0.05$).



(a)

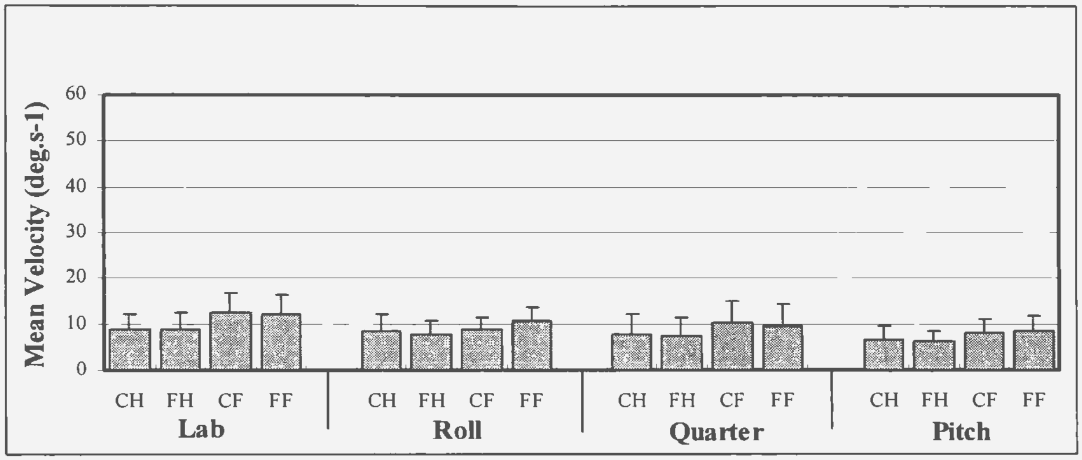


(b)

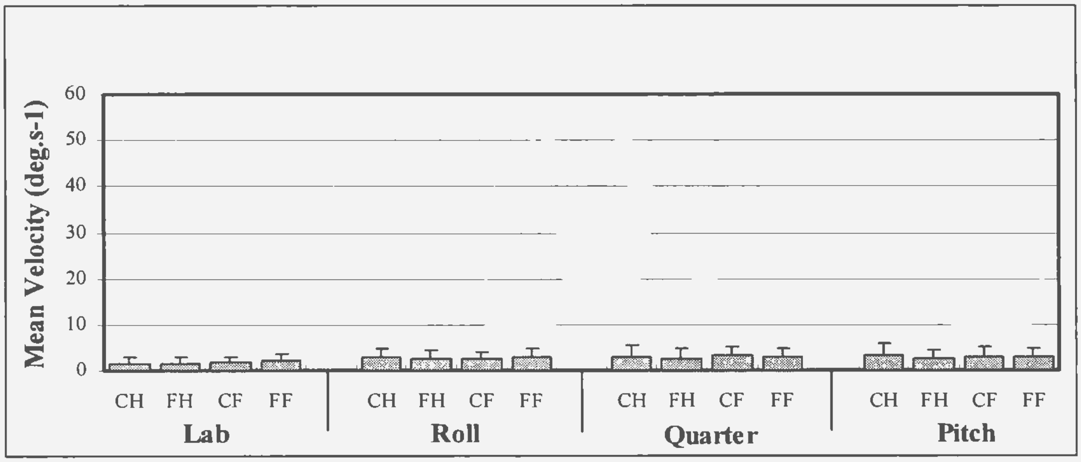


(c)

Figure 4.13: Maximum sagittal (a), twisting (b) and lateral (c) LMM velocities



(a)



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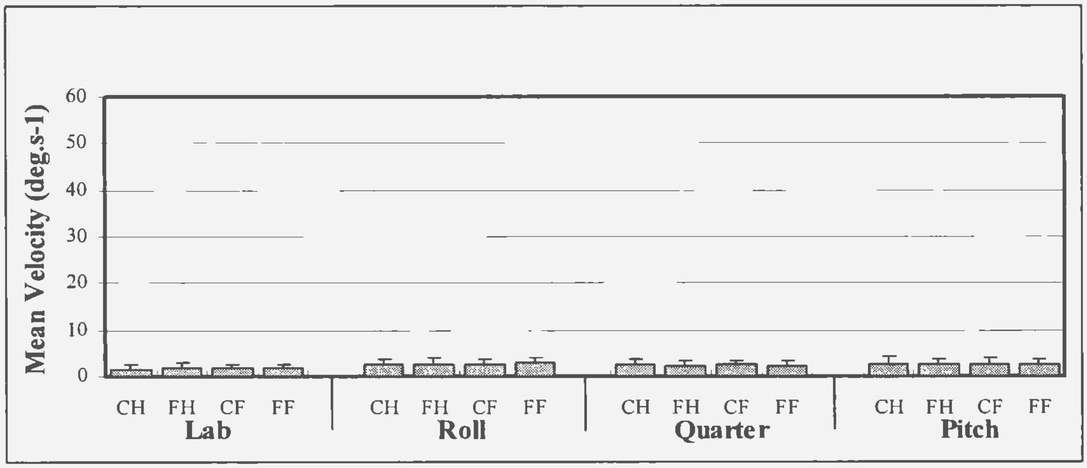
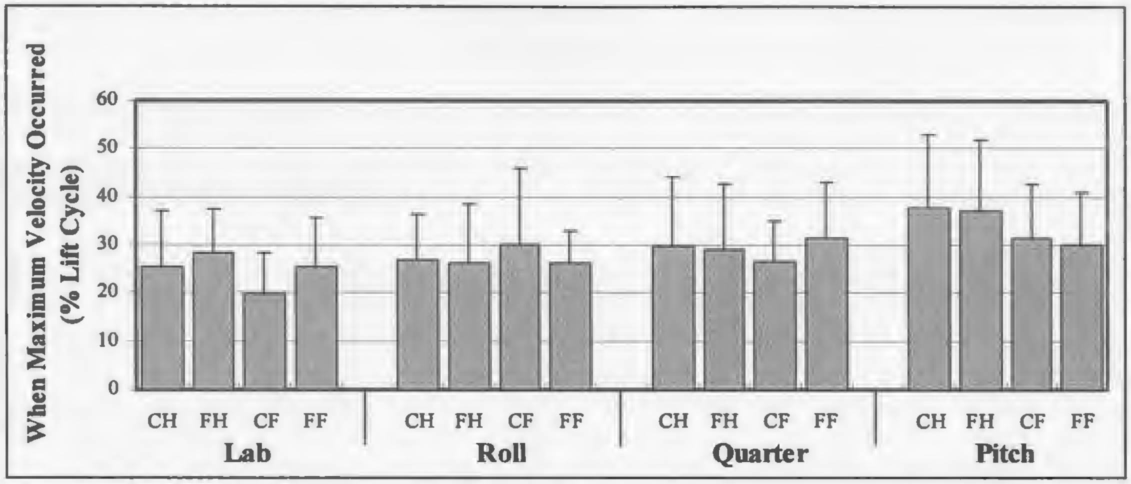
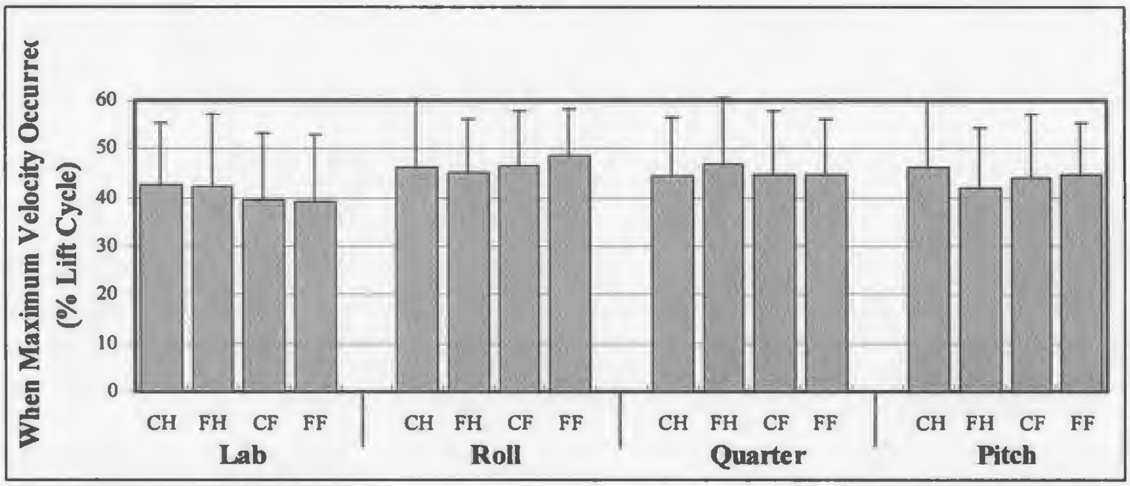


(c)

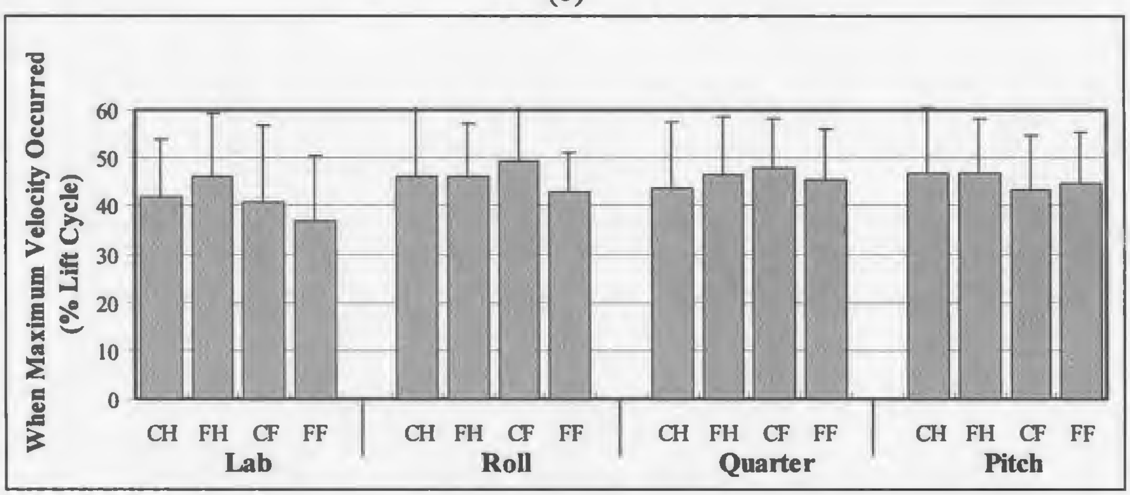
Figure 4.14: Mean sagittal (a), twisting (b) and lateral (c) LMM velocities



(a)



(b)



(c)

Figure 4.15:Time maximum sagittal (a), twisting (b) and lateral (c) LMM velocities occurred

Chapter 5
Discussion

5.1 INTRODUCTION

There has been limited research on operators performing lifting tasks in moving environments. While the effects of ship accelerations on low back loading have been explored (Kingma et al., 2003; Torner et al., 1994) the research tends to focus on kinetic modeling approaches. While providing some insight into potential mechanisms of injury, methodological constraints really do limit the utility of these findings. There has been limited research into how the back functions when placed in different motion-rich environments. Evaluating the effects of different lifting tasks, while monitoring muscle activity and lumbar kinematics during different deck motion profiles has not been successfully attempted. This empirical approach should lead to further insight into how ship motions may contribute to the high incidence of low back overexertion injuries common to persons working in maritime environments (Grinde, 1985; Torner et al., 1988 and 1994).

The advantages of using a ship motion simulator to reproduce sea-like conditions are numerous. For example, motion induced sickness and motion induced fatigue can be experimentally controlled. While in any experiment it is important to maintain ecological validity, it is also important to eliminate or minimize the effects of known confounding factors. In this case, subjects were able to perform the lifting tasks free from common occupational challenges likely to influence the manner in which manual materials handling tasks are performed.

The most obvious advantage to employing a motion simulator was the capacity to create and systematically reproduce a deck motion throughout the duration of the study. Previous studies have used questionnaires to estimate injury rates under real sea conditions (Grinde, 1985, Torner et al., 1988). However, unpredictable sea conditions varied task demands. As a result, being able to compare musculoskeletal loading across subjects can be problematic as each individual is likely to experience different external perturbations over the course of a day or even lifetime. During this study, subjects performed each of the lifting tasks over two minute trials and despite randomization of the lifting and motion conditions, all experienced similar motion profiles over the duration of the trials.

The three motions simulated in this study were pitch, quarter and roll. Experienced mariners would likely agree that a pitch motion orientation would be most hazardous to the worker. Research exists to support this notion (Kingma et al., 2003). Kingma and colleagues (2003) suggested that balance considerations are challenged more when lifting in a pitch motion, as compared to a roll orientation. MacKinnon and Holmes (2005) state that maintaining balance requires the vector projection from a person's centre of mass (CoM) to remain within the boundaries of the base of support. In the para-transverse plane located at the foot-floor interface the shortest distance this projection has to travel to leave the boundaries of the base of support is the antero-posterior direction, thus a pitch motion is likely to be the hardest position to maintain balance. During quarter motion profiles, the subject was positioned at a 45 degree angle to the pitch motion. Often times, subjects were able to alter the position of the load, effectively helping them maintain CoM equilibrium during times when the motion could project their

CoM outside the base of support. For the roll motion condition, the subject was oriented 90 degrees from the original pitch condition. It can be reasoned that the roll motion was the easiest to maintain balance. The subject was positioned perpendicular to the motion and could lean their center of mass over their base of support to the opposite side in which the motion was extended. (i.e. lean to one side). In relative terms, the subject had the largest base of support in the roll condition.

While only successful lifts, free from motion induced interruptions, were considered in this research, it was observed that stumbles and loss of balance resulting in the subject not being able to complete the lift did occur, particularly in the pitch motion condition. The motion profiles selected for this study are based on mathematical representations of data collected in situ on seagoing vessels. Natural motions are not "sinusoidal" and repetitive in nature and generally can be assumed to be unpredictable with wave profiles ranging in height and period. It is likely that the "successful" lifts occurred when the motion profiles were smaller in magnitudes and the stumbles occurred when the values were higher.

To successfully evaluate the demands placed on the human operator while performing lifting tasks in a moving environment, an assessment of lift times, muscle activity and thoracolumbar kinematics were of primary focus. This chapter will attempt to identify the biomechanical changes that may have developed as a direct result of the motion profiles and furthermore, identify those conditions that might place an operator at greater risk for overexertion injuries during lifting activities.

5.2 LIFT TIMES

It was expected that both the close high (CH) and close floor (CF) lifting conditions, which involved transfer of loads that were initially closer to the body, to take a shorter amount of time to complete (see figure 3.2). However, results indicate a significantly longer amount of time was required to move the close high load, when compared to all others. Because subjects were positioned close to the structure upon which the destination target was located they may have been more concerned about a load collision and thus took a longer time to complete the lift. Furthermore, higher lifts mean that the centre of mass of the subject and load system are also relatively higher, reducing the stability of the system. Time might have been longer as more attention was needed to keep the system in a state of equilibrium. Although the CH condition took significantly longer to complete, the time difference was generally less than 0.15 seconds. Mean time across all load conditions to complete the full lift was just over 1.2 seconds.

Results from this study indicate that none of the motion conditions produced significant differences for the time it took subjects to complete a lift. Relatively benign motion conditions were selected for this study, mostly as an ethical consideration in managing the subject's exposure to risk. The implementation of more substantial sea like conditions may provide significant variations in time to complete a lift. It should again be reminded, that only successful lifts were analyzed in this study. Future analysis of lifts where subjects showed balance issues may reveal further lift time differences.

5.3 ELECTROMYOGRAPHY (EMG)

5.3.1 Left and Right Erector Spinae Activity

Contrary to what was hypothesized, there were limited significant differences found during motion states in the EMG activities. Maximum left erector spinae activity demonstrated a significant difference between the lab trial and the quarter motion trials (see figure 4.1). There was also a trend towards the maximum left erector spinae activity pitch motion producing greater activations than the lab trials (see figure 4.1). Generally the pitch motion produced the greatest change, however no significance was found, possibly as a result of large standard deviation values. Figure 5.1 indicates the percent change from lab trials to pitch motion trials, across lifting conditions for the maximum EMG activations. The left erector spinae showed 30, 40 and 50 % increases in activation during the pitch motion compared to the lab condition. The right erector spinae failed to demonstrate similar directional changes and in half of the conditions an opposite trend was observed. In these situations, an asymmetrical loading of the spine is likely occurring, thus producing a higher opportunity for overexertion injury.

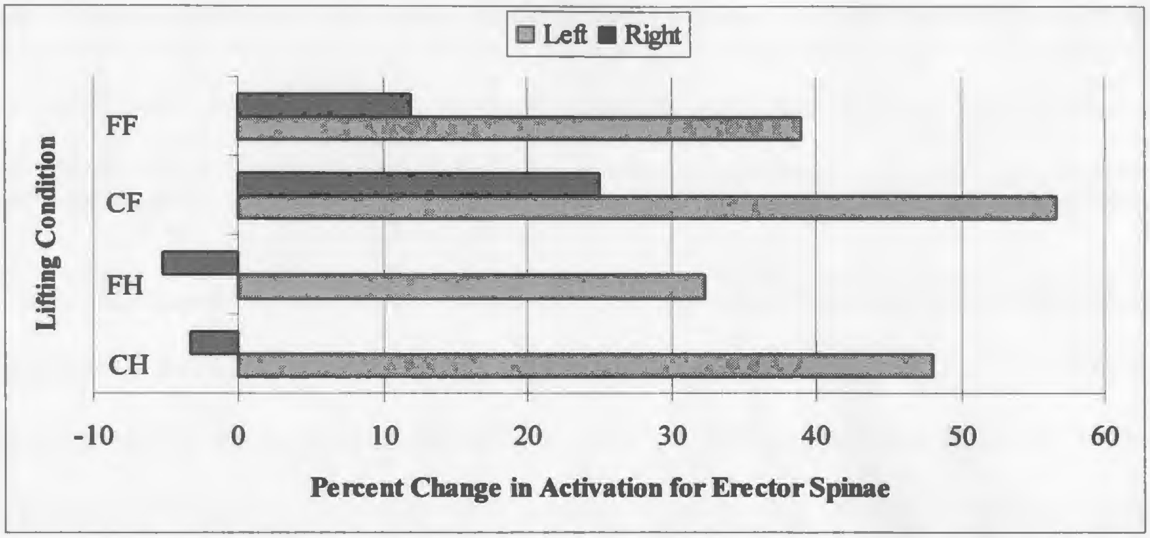


Figure 5.1: Percent change from lab condition for each lifting task in the pitch condition.

The muscle recruitment increases also expose the added physiological costs of an activity performed in moving environments. An increase in muscle recruitment could represent an increase in energy expenditure, resulting in a faster time to fatigue. This could explain the anecdotal reports of excessive fatigue reported by seagoing personnel (Wertheim, 1998). Perhaps complimentary work-rest ratios should be developed for persons performing tasks in a moving environment.

Both the left and right muscles, during the far high and close floor lifting condition produced maximum EMG activities that were significantly less than the far floor condition. This was to be expected, as the far floor condition would require a greater extensor moment due to the forward flexed trunk position. This is known to produce greater activity in the erector muscles. Waters et al. (1993) provides insight into injury risk as a result of high erector spinae activations. They suggested that greater muscle activations are experienced when the load is on the floor, and that lifting objects from the floor causes compression of the spine that could exceed acceptable levels.

Given the added stability and physiological demands of working in a moving environment, lifts originating from the deck should be avoided and workstations should be redesigned to accommodate these needs.

5.3.2 Left and Right Latissimus Dorsi Activity

The latissimus dorsi musculature was the only muscle monitored throughout the study to consistently show left and right muscle differences (table 5.1). The quarter motion profile produced significant left and right differences throughout all lifting conditions. The roll and pitch motions only produced significant differences across certain lifting conditions. The lab trials showed no significant differences and thus it could be concluded that motion was a contributing factor to the observed bilateral differences. The largest bilateral differences occurred during quarter and roll motions. During these conditions, subjects may have needed to counter the load forces opposite to the motions. While this strategy could reduce the risk of stumbling, an asymmetrical posture created bilateral differences in muscle recruitment, likely resulting in changes to the 3-dimensional loading of the spinal anatomy. A question remains whether balance and stability are more important to injury prevention than symmetrical manual material handling postures. Furthermore, whether or not these goals are of equal importance when comparing the risk of either acute or chronic injuries, require further consideration.

Table 5.1: Maximum EMG values for the left and right latissimus dorsi muscle

		Left Latissimus Dorsi		Right Latissimus Dorsi		
Lifting Conditions	Motion	Mean	SD	Mean	SD	Significance
CH	Lab	16.84	11.86	16.17	12.07	NS
	Roll	23.92	19.71	16.45	15.70	*
	Quarter	18.05	9.31	16.71	12.36	*
	Pitch	25.48	14.25	16.00	10.41	NS
FH	Lab	22.47	18.03	19.60	14.54	NS
	Roll	23.35	11.88	20.76	17.15	NS
	Quarter	33.38	35.13	27.36	29.10	*
	Pitch	29.33	13.28	18.42	10.43	*
CF	Lab	14.55	12.29	13.26	12.14	NS
	Roll	21.70	16.81	16.53	16.13	**
	Quarter	14.99	9.28	15.63	15.46	*
	Pitch	21.09	11.13	15.48	8.32	*
FF	Lab	26.50	30.46	17.73	15.25	NS
	Roll	28.07	20.91	20.48	16.53	**
	Quarter	20.41	8.67	19.46	17.33	*
	Pitch	23.17	13.51	16.65	10.92	NS

(* = $p < 0.05$, ** = $p < 0.01$, *NS* = No significance)

5.3.3 Left and Right Trapezius Activity

For the majority of motion states and lifting conditions, the trapezius muscle reached maximum activation levels of 40% MVC. During only the far floor, roll condition was there a difference in the left and right trapezius muscle (see table 4.3 and figure 4.3). While further research is needed with respect to the upper extremities involvement during these lifting tasks under motion profiles, 40% MVC activations will suggest that fatigue within these muscles may develop over prolonged periods of continuous manual materials handling, potentially increasing injury risk. The close high, far high and far floor lifting conditions had greater activations than the close floor condition. This suggests that lower muscle activations during the close floor condition will limit the amount of muscular loading placed on the trapezius muscles. Continued over-exertion and repetitive lifting at shoulder level or higher should be eliminated or minimized as much as possible from proper workstation design.

5.3.4 Left and Right External Oblique Activity

There were no significant differences found for both mean and maximum left and right external oblique muscles. Maximum external oblique muscle activity occurred during the quarter motion orientation, during the far floor lifting condition for both the left and right muscles (see table 4.3 and figure 4.4). During this time, the left and right muscles produced activations of 21.28 ± 26.02 % MVC and 19.01 ± 17.77 % MVC, respectively. The magnitudes of these activities are large, considering the external obliques are not prime movers for these movement activities. There were no differences observed between activation levels for the lab and motion trials. Barr et al. (2005) discussed the multifidus and transverse abdominis as being deep stabilizer muscles that function to prevent excessive bending and stiffen the spine. This suggests that if increased core stabilization occurred for the motion trials compared to the lab trials it may have been attributed to deeper abdominal muscles not measured in this study. Barr et al. (2005) also stated that additional muscles, such as paraspinal (i.e. erector spinae) and iliopsoas muscles also assist in core stabilization. They state that these muscles prevent unwanted trunk movements. With the observed recruitment of the external obliques and the increases in erector spinae activities (see figures 4.1 and 4.5) and the increased twisting and bending velocities (see figures 4.13 and 4.14) observed in the motion trials (compared to the lab condition) it is clear that a greater risk for overexertion injury is likely when motion complicates the execution of a lifting task. Additional research should be given with respect to activation patterns of deep abdominal musculature.

The EMG activity for all muscles monitored in this study followed the expected trends. Erector spinae and trapezius muscle activities were large in magnitude and would suggest significant loading to the pertinent joint segments. As a result of these high EMG activations, over-exertion injuries are likely to occur. Due to the continuous stabilization required of trunk musculature in a moving environment, performing repetitive lifting tasks for long periods of time would certainly suggest localized fatigue to specific muscles active during the required tasks. It is recommended therefore, that shorter bouts of activity be performed while in these environments.

5.4 LUMBAR MOTION MONITOR (LMM)

Marras et al. (1995) studied over 400 industrial lifting jobs in 48 varied industries, while collecting trunk motion characteristics. They were able to quantify which characteristics, such as workstation, load and personnel were associated with an increase risk of reported occupationally related low back injuries. From these data they assigned categories of low, medium and high risk for people who may develop low back over-exertion injuries as a result of lifting activities. It has been reported that an increase in trunk motions during lifting activities will increase a person's chance of developing low back injuries (Marras et al. 1995, Norman et al., 1995). Marras et al. (1995) reported that maximal trunk velocities were the most significant predictors of risk for low back overexertion injuries. The results obtained in this study will be reported and compared to those obtained by Marras et al. (1995). However, it should be noted that of the 48 industries studied by the Marras et al. (1995), none were measured in moving environments. Nor were there comparable industries where workers could be characterized as working in unstable environments, such as mining, forestry and work in sand or snow. A summary of Marras et al. (1995) maximum LMM velocities for low, medium and high risk are reported in table 5.2.

Table 5.2: Maximum Lumbar Motion Monitor data reported by Marras et al. (1995)

Trunk Motion	Low Risk ($deg.s^{-1}$)	Medium Risk ($deg.s^{-1}$)	High Risk ($deg.s^{-1}$)
	Mean ± Std. Dev	**Mean ± Std. Dev**	**Mean ± Std. Dev**
Sagittal Plane	38.69 ± 26.52	53.69 ± 36.37	59.00 ± 36.19
Twisting Plane	38.04 ± 17.51	48.48 ± 6.86	49.72 ± 27.64
Lateral Plane	35.45 ± 12.88	45.14 ± 18.97	44.58 ± 17.47

There has been limited research on successfully evaluating common MMH tasks, such as lifting, in motion-rich environments. Waters et al. (1993) and Marras et al. (1995) both provide empirical data which can be used as guidelines to assess the suitability of a lifting task. However, the utility of such information may be limited if applied to work in moving environments and as such; comparisons to this literature must be done with caution.

5.4.1 Sagittal plane thoracolumbar kinematics

Maximal sagittal velocity was significantly different for all motion states in comparison to the lab condition (see figure 4.13a.). The four lifting conditions consistently produced lower maximum sagittal velocities during the lab trials than during motion states (see figure 4.13a). Participants generally demonstrated greater erector spinae muscle activity during the motion conditions. This increased activity is reflected in the decreases in thoracolumbar velocities as trunk stabilization likely increases. Unfortunately, dynamic trunk motions, with increased paraspinal activities, have been associated with greater spine loading (Marras et al., 1984; McGill, 1991a, 1991b).

Davis and Marras (2000) suggest that trunk motion significantly reduces an individual's ability to produce force. When relating this to an unstable environment, these results suggest that a decrease in trunk motion, coupled with an increase in trunk muscle activation will be necessary to maintain stability and balance.

The lifting conditions that began with the load on the floor (close floor and far floor) produced significantly greater maximum velocities when compared to the conditions which began with the load on the riser (refer to figure 4.13a). By having to extend the trunk further to pick up the load from the floor, it can be reasoned that subjects felt their back extensor muscles were lengthened to a point where they were required to use more force to displace the load. Having to bend more, may cause additional balance problems as well. Hence, a 'jerking motion' may have been required at the beginning of the exertion (MacKinnon and Li, 1998), producing maximum velocities greater than those lifts when the back was in a more upright position (load starting on the riser). This increased velocity is further supported with the increased erector spinae muscle activity found during the close floor lifting condition (see figure 4.1). Further evidence to support this notion can be seen when evaluating the relative time at which the maximum velocities occurred during the lift. While only an observed trend, the maximum velocities occurred earlier in the trials lifted from the floor, when compared to those from the riser (see figure 4.15a.). The left erector failed to support this observation, however, the maximum activity of the right erector spinae during the close floor condition occurred significantly earlier in the trial than the far high condition ($p<0.05$). There was also a trend ($p<0.073$) to support the notion that floor conditions occurred earlier in the trials, when the close high condition was compared with the close floor condition.

In comparison to data presented by Marras et al. (1995) (see table 5.2), maximum sagittal velocities from this study were consistently below those considered to put an operator in the low risk category for developing an overexertion injury. Often, poor workstation design influences the manner at which a person exerts a lift task. However in this protocol subjects were asked to perform lifting tasks which conformed to NIOSH (Waters et al. 1993) guidelines for safe lifting practices. This would certainly be a reason why Marras et al. (1995) reported much higher velocities. Further reasoning may be that subjects felt lifting in a slow and controlled manner (although producing high EMG recruitment values) which would provide greater trunk stabilization, ultimately helping maintain balance or better prepare themselves for unexpected perturbations during the lift. However, these values are interpreted against those obtained in stable environments. There is no doubt that working in moving environment creates added biophysical stresses, such as fatigue and occasion for loss of stability.

Injuries occur when balance and equilibrium are disrupted. At these times, thoracolumbar motions would be higher in comparison to successful lifts executed in a controlled manor. Muscle activities would also be higher in order to stabilize segments in hopes of regaining balance and improving personal safety. Higher LMM velocities and EMG activities during these cases would likely be precursors to overexertion injuries of the low back. Cholewicki et al. (2000) determined that immediately after a perturbation, the trunk muscles contribute to the prevention of large spinal motions. There may be evidence to suggest that this idea could be extended to an unstable environment, and successful, unperturbed lifts.

5.4.2 Twisting plane thoracolumbar kinematics

Bending and twisting have been assumed to be associated with the development of low back pain (MacKinnon, 1998; Mital 1997; Troup et al. 1970, Van Diëen, J.H., 1996). During this study, maximum twisting velocities measured during the motion conditions were significantly higher than the lab condition (see figure 4.13b). It must be assumed that it was the accelerations of the floor in the directions other than the sagittal plane of the lift that induced these increased thoracolumbar motions. If one was to consider typical seagoing working environments, operators would often be exposed to more substantial deck motions, as well as other external conditions (i.e. wind and slippery floors) that will induce even greater thoracolumbar twisting velocities. However, an additional increase in platform motions may also cause the operator to decrease the throacolumbar velocity, as a means of protection and increased stability. Chiang and Potvin (2001) suggest that during more stable conditions there will be a smaller angular displacement of the trunk after the perturbation is experienced. Van der Burg et al. (2004) suggested that the amount of trunk rotation after a perturbation will depend on spine stability just prior to the stumble. Perhaps when working in a moving environment experienced operators come to expect perturbations and would adopt lifting strategies to increase personal safety. The majority of these data suggest larger velocity kinematics of the spine during lifting in a moving environment and it can be argued that spinal stability is certainly lowered during motion conditions. If operators can increase the strength of the supporting musculature, such as incorporating core stabilization exercises, they could lower their risk of injury.

Marras et al (1995) reported maximum twisting velocities of 38.4 ± 17.51 deg.s^{-1} as a limit for the low risk category. Maximum values reported in this study (see figure 4.1b) are lower then those reported in table 5.2. The smaller values might be expected as twisting was not necessary to complete the lifting task under normal conditions and any observed motion in this plane should be due to the floor motions. Kingma et al. (2003) suggests there may actually be less twisting during balance loss experienced in the pitch motion orientation. Subjects in this study usually extend one foot in front of their body to help maintain balance during a stumble in the pitch trials. Thus depending upon which foot is moved to regain balance and the direction of the deck accelerations in the twisting plane, the operator would be at a very different risk of overexertion injury, as the pelvis-spine orientations would differ considerably.

5.4.3 Lateral plane thoracolumbar kinematics

The maximum lateral velocities demonstrated similar trends as both the sagittal and twisting planes. Significantly lower values during the lab condition than all other motion states were observed (see figure 4.13c). These values were expected as subjects were performing only a sagittal plane lift. These values were below the low risk standard set by Marras et al. (1995) (see table 5.2). However, lateral bending velocities were increased in the motion conditions compared to the lab condition. Similar to discussion of the twisting velocities, it can be assumed that risk for overexertion injury is therefore increased.

Lateral bending places additional shear forces on the spine and it is likely that these forces increase if bending occurs during work in moving environments. Participants in this study leaned to one side as a means of counteracting the out of sagittal plane accelerations.

5.5 Limitations

There were a number of limitations to this study. This study is the first of its kind to evaluate the effects of floor motion on muscular activity and thoracolumbar kinematics. Relatively benign motion conditions were selected for this study, mostly as an ethical consideration in managing the risk exposure of the subject. Furthermore, data collection under any moving condition is complex and fraught with methodological issues so the number of dependent variables considered was restricted. Often times, offshore operators perform similar lifting tasks as performed in this study while experiencing slippery floors, icy conditions, and extremely cold temperatures. Previous work (MacKinnon and Holmes, 2004) performed in a real moving environment show similar thoracolumbar kinematics to those obtained in these simulator trials. This suggests that there is some ecological validity in testing under simulated motion conditions, which would be of tremendous advantage for future research proposals. However, further insight into the causes of overexertion injuries would be obtained if more extreme sea conditions were employed in this study.

It should be noted that this study recruited healthy, male university population participants. Kingma et al. (2003) suggested that the prediction and/or anticipation of platform motions are not possible by an operator; however, further analysis of lifting in a moving environment with experienced seagoing personnel may be of interest.

EMG analysis during this study was limited to 4 muscles measured bilaterally. The muscles monitored during this study were considered important core stabilizer muscles as well as muscles actively involved during lifting. However, all muscles were upper body measures and further analysis of lower body muscle activity would be of interest. It was apparent that during this study subjects used lower body strength to help counter motion effects and maintain stability.

5.6 Future Research

All initial lifting conditions during this study were restricted to the sagittal plane. Quite often, offshore workers perform lifting tasks that involve twisting and lateral bending as a result of poor workstation design, confined work spaces or other ergonomic or engineering issues. Many authors have suggested an increase in the risk of LBP during asymmetric lifting (Anderson, 1981; Granata and Marras, 1993; Kyserling et al., 1988; Marras et al., 1995; Marras and Davis, 1998). Couple this increased risk caused by twisting and bending, with the effects of motion and increased muscular activity and injury risk should certainly increase. Further insight into the effects of motion on asymmetrical lifting tasks should be further considered.

Further studies involving lifting in a moving environment should also consider changes in centre of pressure motions under foot. Foot pressure data is related to balance and stability and these data can be a means of quantifying the level of balance and stability during a lift. This type of measurement may help understand better the changes in a person's centre of mass and help provide significant insight into improving motion induced interruption model predictions.

Chapter 6
Conclusions

This study has successfully contributed to a better understanding of the demands placed on offshore workers who often perform lifting tasks in a moving environment. It is the first study of its kind to evaluate muscle activity and trunk kinematics during motion conditions. Both thoracolumbar kinematics and muscle activity during different motion states were analyzed, providing insight into the four hypotheses proposed in this study. Hypothesis one stated that the direction of simulated platform motions will cause a significant increase in muscle activation during a lifting task. While not significant, there were trends to support the notion that muscular activity was increased with certain motion conditions. The lab trails consistently produced lower activation levels then the motion trials, while the pitch and quarter motion trials generally produced greatest muscular activity. This suggests that greater deck motions likely induces greater levels of muscular activity. It can be speculated that onset of muscular fatigue would be earlier in lifts performed in moving environments compared to stable floor conditions, all other factors being equal.

The pitch motion provided the lowest sagittal thoracolumbar kinematic velocities, while lab trials produced the greatest velocities. This trend fails to support hypothesis number two, which suggested that the motion state would significantly increase thoracolumbar kinematic values. However, data collected on the lumbar spine measured in the twisting and bending plane did support the second hypothesis, as increases in both the twisting and lateral bending velocities were experienced during motion conditions. It is likely that these twisting and lateral bending spinal movements place operators at increased risk for injury.

Hypothesis three and four stated that the characteristics of the lifting task would have significant effects on both muscular activity and thoracolumbar kinematics. Both were supported by the data from this study. Certain lifting conditions provided both greater muscular activations as well as thoracolumbar kinematic velocities. Both the left and right erector spinae muscles, during the far high and close floor lifting condition produced maximum EMG activities that were significantly less then the far floor condition. The lifting conditions close floor and far floor produced significantly greater maximum velocities when compared to the conditions which began with the load on the riser. Particular attention needs to be given towards better design of workstations in order to reduce the stresses placed upon the body. It cannot be assumed that workstations designed for stable environments are necessarily adequate for use in moving environments.

Many authors have speculated that platform motion is related to an increased risk of injury to the operator (Kingma et al., 2003; Wertheim, 1998; Torner et al., 1988 and 1994; Grinde, 1985). Given the large muscle activations and significant thoracolumbar velocities observed in this study, these data suggest that performing tasks in moving environments, especially over extended periods of time, will place an operator at greater risk of MMH-related overexertion injuries.

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