Neuromuscular and postural responses to sudden loading and high frequency lifting: effects of posture and fatigue

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Ehara taku toa, he taki tahi, he toa taki tini. My success should not be bestowed onto me alone, as it was not individual success but success of a collective.

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Abstract

Sudden unexpected loading and high frequency repetitive lifting have been identified as risk factors for low back injury. Despite the risk that such factors pose, there is a dearth of literature concerning the influence of posture and fatigue on muscular, postural, subjective and physiological responses in these situations. The primary aim of this thesis was to examine the effects of posture and lifting-induced fatigue on neuromuscular, postural, psychophysical and physiological responses to sudden loading and repetitive lifting.

The first two studies in this thesis focused on muscular and postural responses to sudden loading of a hand-held box in an upright and stoop standing posture. Trials were performed with and without visual warning and following lifting-induced fatigue. Sudden loading in an upright posture resulted in co-activation of oblique abdominal and erector spinae musculature, with relatively simultaneous joint initiation of the lower limb and lumbar spine. In contrast, sudden loading in the stoop posture produced minimal abdominal muscle contribution. Maximal angular joint excursion was significantly less in the stoop than that in the upright posture. Prior exposure and warning primarily affected postural responses in the upright posture with earlier activation of the trunk and lower limb musculature resulting in a reduction in joint excursion. Such a mechanism may enhance postural stability. However, the influence of warning on reducing joint motion was compromised when fatigued.

Studies three and four of the thesis focused on the effects of self-selected lifting posture (squat, mixed and stoop) and fatigue during repetitive lifting on psychophysical, physiological, neuromuscular and lumbar spine kinematic responses.

Study three compared lifting postures using ratings of perceived exertion and heart rate response patterns. Similar subjective and physiological responses were found for different lifting postures, with a progressive rise in ratings of perceived exertion and heart rate measures throughout the task. All postures exhibited evidence of erector spinae muscle fatigue at the end of the task. Study four examined the effect of self-selected lifting posture on lumbosacral kinematics and associated trunk muscle activation patterns during lifting and lowering, pre and post fatigued. When compared to squat lifting, individuals who self-selected a stoop lifting technique displayed different lower erector spinae activation patterns. The stoop lifters also flexed their lumbar spines at faster velocities and adopted greater lumbosacral flexion than squat lifters, which could potentially place greater strain on the passive structures of the spine. Fatigue influenced lumbosacral motion patterns of stoop and mixed lifters to a greater extent than squat lifters, resulting in a decreased range and velocity of lumbar extension.

Findings from this thesis provide insight into the underlying neuromuscular control of the trunk during high risk manual handling activities. It provides important information concerning the possible mechanism of postural and fatigue related musculoskeletal injury. This will assist the development of training and rehabilitation programmes targeted at neuromuscular, physiological, biomechanical and psychophysical requirements of manual handling tasks.

Chapter One

Introduction

THE PROBLEM

It has been well documented that manual handling vocations have a high incidence of low back injury (LBI) (Garg and Moore, 1992; Hoogendoorn et al., 2000; Marras et al., 1993). In 2008 LBI claims were estimated to cost New Zealand in excess of 120 million dollars per year in lost earnings and treatment related costs (Accident Compensation Corporation [ACC], 2008). Within manual handling occupations, sudden loading (perturbation) and high frequency repetitive lifting (inclusive of lifting and lowering) have long been identified as major risk factors for LBI (Kelsey et al., 1984; Magnusson et al., 1990; Manning, Mitchell, and Blanchfield, 1984; Marras et al., 1993; McCoy, Hadjipavlou, Overman, Necessary, and Wolf, 1997). Whilst sudden loading and repetitive lifting represent quite different injury scenarios both require efficient neuromuscular and postural control to minimise load on the lumbar spine.

With respect to sudden loading, postural responses have to be rapid and it has been suggested that factors such as practice and prior warning may optimise postural responses to the sudden load lessening the likelihood of LBI (Cresswell, Oddsson, and Thorstensson, 1994; Lavender, Marras, and Miller, 1993; Leinonen, Kankaanpaa, Hanninen, Airaksinen, and Taimela, 2002). However, there is uncertainty whether these postural responses to the sudden load are dependent on the posture adopted (upright vs stoop) during manual handling tasks and may differ if the handler is physical fatigued.

For repetitive lifting the focus of research has been on identifying the lifting posture (squat, mixed or stoop) that produces conditions where neuromuscular control of lumbar spine motion is optimal, preventing excessive loading on the lumbar spine and cardiorespiratory system, and is perceived by the individual as requiring an acceptable amount of effort. A number of studies have investigated neuromuscular, physiological, psychophysical and biomechanical responses during the performance of instructed lifting techniques (Potvin, McGill, and Norman, 1991; Rabinowitz, Bridger, and Lambert, 1998). However, few studies have investigated neuromuscular control of lumbar posture and associated cardiorespiratory and subjective responses to high frequency lifting when adopting self-selected lifting postures (squat, mixed or stoop), or when in a state of physical fatigue.

Sudden loading

Sudden unexpected anterior loading (such as when the contents of a box held in the hands are suddenly displaced) represents a situation where extremely rapid postural responses (less than 130 ms) are required to counter the perturbation and stabilise the body (Cresswell, 1993; Cresswell et al., 1994; Granata, Slota, and Wilson, 2004). Inadequate postural responses to unexpected loading may potentially decrease spinal stability (Cholewicki and McGill, 1996) and increase the forces imposed on the spine (Mannion, Adams, and Dolan, 2000). It has been proposed that practice or prior exposure to sudden loading may enhance postural responses and reduce spinal loading (Cresswell et al., 1994; Horak and Nashner, 1986; Lavender et al., 1993;

Leinonen et al., 2002; Pedersen, Essendrop, Skotte, Jorgensen, and Fallentin, 2004). Whilst studies have found practice to have little effect on postural muscle onset times in response to sudden loading most of these studies have provided subjects with some form of prior information (practice trials) ahead of the sudden loading event (Lavender et al., 1993; Pedersen et al., 2004; Wilder et al., 1996). Therefore, it has not been possible to establish whether postural responses would be the same as might occur in a situation where the sudden load was truly unexpected. A comparison of postural responses between truly unexpected sudden loading and subsequent perturbations is necessary in order to appreciate possible injury mechanisms and how prior exposure to sudden loading can influence postural responses to future sudden loading events.

Although there is little evidence that practice influences muscular responses to sudden loading, prior warning may enhance postural responses to a sudden load and possibly prevent injury. It has been suggested that prior warning may allow preparatory co-activation of trunk musculature in advance of the load, stiffening the spine and reducing trunk displacement (Granata, Orishimo, and Sanford, 2001; Leinonen et al., 2002; Thomas, Lavender, Corcos, and Andersson, 1998). Preparatory co-activation has been shown to occur when the perturbation is self-induced or an individual is allowed to partially track the perturbing object (Cresswell et al., 1994; Lavender et al., 1993; Thomas et al., 1998). However, similar findings have not been shown for situations involving verbal warning (Granata and Orishimo, 2001). Others have found that rather than preparatory co-activation, warning may result in earlier onset of postural muscle activation (Leinonen et al., 2002; McChesney, Sveistrup, and Woollacott, 1996). What is less clear is how warning may affect the muscle phasing patterns and the subsequent motion of the lumbar spine and lower limbs following the sudden load.

Postural responses to sudden loading may also be influenced by fatigue. To date, results from studies investigating effects of fatigue have varied, with some studies reporting delayed muscle response times to a sudden load (Magnusson et al., 1996; Wilder et al., 1996). However, others (Granata et al., 2004; Herrmann, Madigan, Davidson, and Granata, 2006) have found no change in muscle onset latencies or trunk motion following fatiguing protocols, but rather increases in preparatory co-activation of trunk musculature (Granata et al., 2001; Granata et al., 2004). Such contrasting findings may well stem from the different methods studies have employed to induce symptoms of fatigue (for example, vibration and prolonged static contraction). The effects of fatigue induced by manual handling tasks on postural responses to sudden loading have not been clearly established

Postural responses to sudden loading have primarily been investigated with individuals assessed in an upright standing position. However, during manual handling tasks individuals may adopt a range of different postures, some of which include stoop postures where the spine is considerably flexed. Sudden unexpected loading in a stoop posture (legs relatively straight and the hips and lumbar spine flexed) may potentially injure the lumbar spine if the lumbar spine is placed near its end range of motion. Towards the end range of lumbar flexion, the ability of the lower erector spinae (LES) to resist the bending moment and anterior shear forces may be compromised and the bending moment resisted by the passive tissues of the lumbar spine rapidly increases (Macintosh, Bogduk, and Pearcy, 1993; McGill, Hughson, and Parks, 2000). In contrast, an upright standing posture places the spine in a relatively neutral position which is potentially unstable (Cholewicki and McGill, 1996). In such cases, sudden loading has the potential to injure the spine if appropriate muscle responses do not occur (Cholewicki and McGill, 1996; Granata and Wilson, 2001). As both upright and stoop postures are commonly used during manual handling activities, a comparison of postural responses elicited during sudden perturbation in both upright and stoop postures would seem appropriate.

Repetitive lifting

Repetitive lifting is considered a primary risk factor for LBI in manual handling occupations, with the risk of injury increasing as frequency of lifting increases (Chaffin and Park, 1973; Frymoyer et al., 1983; Garg and Moore, 1992; Kelsey et al., 1984; Magnusson et al., 1990; Marras et al., 1993). Unlike sudden loading it is thought that injury during repetitive manual handling may occur as a result of cumulated stress and fatigue, which may lower a system's threshold for tissue damage and injury (McGill, 1997a). In order to prevent such injuries, back schools and researchers have attempted to establish safe lifting techniques, with a focus on the three main lifting postures that are adopted during manual materials handling. These include the squat or leg lift (where the knees are bent and the back is straight or erect), the stoop lift (where the knees are straight and the back was bent) and the mixed or semi squat lift which is a posture that lies in the continuum between a squat and a stoop lift (Burgess-Limerick and Abernethy, 1997; Burgess-Limerick, Shemmell, Barry, Carson, and Abernethy, 2001; Hagen, Harms-Ringdahl, and Hallen, 1994; Revuelta, Dauphin, Kowslowski, Dubois, and Thevenon, 2000; Welbergen, Kemper, Knibbe, Toussaint, and Clysen, 1991).

It has been suggested that workers may prefer to use a stoop or mixed lifting technique in the workplace as self-reported ratings of perceived exertion (RPE) and physiological measures such as heart rate (HR) are on average lower than when squat lifting, particularly at low lifting frequencies (less than six lifts per minute) (Rabinowitz et al., 1998; Straker and Cain, 1999). During lifting at higher frequencies (greater than 16 lifts per minute) RPE (Hagen, Hallen, and Harms-Ringdahl, 1993; Hagen et al., 1994) and HR (Welbergen et al., 1991) have not been found to differ between lifting postures, whereas others have reported higher HRs for the squat when compared to the stoop lift (Hagen et al., 1993; Hagen et al., 1994; Revuelta et al., 2000). One important issue when studying RPE and HR responses during repetitive lifting is how to quantify each of these measures. Typically these have involved mean and peak values which may not accurately reflect whether an individual has reached a subjective or physiological steady state. Temporal changes in RPE and HR are considered important when investigating the physiological demands of repetitive tasks (Herman, Nagelkirk, Pivarnik, and Womack, 2003; Koga et al., 1999) as a failure to maintain physiological and subjective steady state at sub-maximal workloads will lead to a gradual rise in HR and RPE, leading to eventual exhaustion and fatigue (Herman et al., 2003). To date, there appear to have been no studies which have investigated the effects of posture and high frequency repetitive lifting on temporal measures of RPE and HR.

Another approach to assessing the risk of injury from different lifting techniques is by means of biomechanical analysis. To date, a majority of the research has focused on compressive forces acting on the lumbar spine and findings indicate that there is no clear distinction between the stoop and squat lift (van Dieen, Hoozemans, and Toussaint, 1999). There seems to have been less focus on the neuromuscular control of lumbar spine kinematics when adopting different lifting techniques. Kinematic variables such as peak lumbar flexion and the rate of change in lumbar motion (velocity) are considered important risk factors for LBI in the workplace (Hoogendoorn et al., 2000; Marras et al., 1995; Punnett, Fine, Keyserling, Herrin, and Chaffin, 1991). Biomechanical studies indicate that high degrees of lumbar flexion and lumbar velocity increase the loading on the passive structures of the spine (Adams and Dolan, 1996; McGill, 1997a; Potvin et al., 1991; Wang, Parnianpour, Shirazi-Adl, and Engin, 2000). It has been shown that individuals who are instructed to perform a stoop lift produce a greater degree of lumbar flexion and less LES muscle activation than those individuals who use a squat lift (de Looze, Toussaint, van Dieen, and Kemper, 1993; Dolan, Earley, and Adams, 1994; Dolan, Mannion, and Adams, 1994; Hart, Stobbe, and Jaraiedi, 1987; Potvin et al., 1991). However, few studies have attempted to quantify lumbar kinematics and associated trunk muscle activation during lifting when adopting a self-selected lifting technique.

Fatigue may also influence lumbar spine motion and trunk muscle activation when lifting. Studies that have not differentiated lifting technique have indicated that when fatigued, individuals generally revert to a more stoop technique, increasing the extent of lumbar spine flexion (Dolan and Adams, 1998; Sparto, Parnianpour, Reinsel, and Simon, 1997a) and reducing lumbosacral velocity (van Dieen, van der Burg, Raaijmakers, and Toussaint, 1998). In contrast, others (Marras and Granata, 1997; Sparto, Parnianpour et al., 1997a) have found little change in lumbosacral velocity and decreases in peak lumbar spine flexion over time. A possible reason for the variability in findings may relate to the fact that when instructed to use a self-selected lifting technique, individuals will adopt a variety of postures (Bonato et al., 2003). Therefore, it is difficult to determine whether a specific self-selected lifting posture (squat, mixed or stoop) is more susceptible to change in lumbosacral kinematics and trunk muscle activity during repetitive fatiguing tasks.

STATEMENT OF THE PROBLEM

There appears to be a deficiency in studies that have examined the effects of prior experience and warning on postural responses to sudden loading in different standing postures. There is also limited information concerning how lifting-induced fatigue influences postural responses to sudden loading.

Furthermore, during high frequency repetitive lifting there is little information on how the initial lifting posture (squat, mixed or stoop) selected by an individual influences perceived effort, HR, lumbar spine motion and trunk muscle activation patterns during the task.

Hence, the purpose of this study was twofold. Firstly, to examine the effects of prior exposure, warning, posture and fatigue on postural responses to sudden loading. Secondly, to examine the effects of self-selected lifting posture on subjective (RPE), physiological (HR), neuromuscular (trunk muscle activation) and biomechanical (lumbar spine kinematics) during high frequency lifting to fatigue. Accordingly, there were two main aims:

Aim one: To investigate postural responses to sudden loading in different standing postures prior to and following physical fatigue. Trunk and lower limb muscle and postural responses were measured during sudden loading in upright and stoop standing, with and without warning, and prior to and following lifting-induced fatigue.

Aim two: To investigate how self-selected lifting posture (squat, mixed and stoop) influences psychophysical, physiological, neuromuscular and lumbar kinematic responses to highly repetitive lifting to fatigue. Comparisons in RPE, HR, trunk muscle activation and lumbar spine kinematics were made between individuals who performed high frequency repetitive lifting to fatigue using a self-selected lifting posture.

In respect to the above the following studies were undertaken:

AIM ONE

Study One An investigation of the effects of prior exposure, warning, and initial standing posture on muscular and kinematic responses to sudden anterior loading of a hand-held box (Chapter Three).

Study Two. An investigation of the effects of prior warning and lifting-induced fatigue on trunk muscle and postural responses to sudden loading of a hand held box (Chapter Four).

AIM TWO

Study Three. An investigation of the effects of self-selected lifting posture on RPE, HR, and ES muscle fatigue during high frequency repetitive lifting to exhaustion (Chapter Five).

Study Four. An investigation of lumbar spine kinematics and trunk muscle activation patterns of individuals adopting different lifting postures during highly repetitive lifting to fatigue (Chapter Six).

SIGNIFICANCE OF THE STUDY

Findings from this thesis have significance for those professionals involved in the prevention and management of musculoskeletal injury in industry. It provides insights into the neuromuscular and postural strategies used by individuals in response to sudden loading and repetitive manual handling. Information from this study has important implications for understanding the mechanisms of injury associated with sudden loading and repetitive handling in the workplace. Such information will further benefit the development of injury prevention and rehabilitation programmes for individuals involved in manual handling vocations.

Chapter Two

Literature Review

INTRODUCTION

In accordance with the aims and objectives of this thesis, the literature review is set out as follows. Firstly, epidemiological evidence of a relationship between sudden unexpected loading and highly repetitive lifting and LBI will be reviewed. Secondly, consideration will be given to those studies investigating the effects of sudden anterior loading of the trunk when undertaking manual handling activities will be reviewed. The emphasis will be upon the influences of prior experience, warning, lifting-induced fatigue, and the initial standing posture (upright versus stoop at the time of loading) on muscular and postural responses to sudden anterior loading of the trunk. The second key area demanding attention is the literature investigating the influence lifting posture (squat, mixed or stoop) on measures of RPE, HR, trunk muscle activity and spinal kinematics in individuals during repetitive lifting and lowering activities. Specific attention will be directed towards the influence of fatigue on these measures in individuals who adopt different lifting postures. The functional anatomy of the musculature of the trunk will then be discussed in relation to sudden loading and lifting activities. The review will conclude by discussing the relevant methodological issues associated with the primary analytical techniques (electromyography (EMG) and motion analysis) adopted by those studies reported in this thesis.

LITERATURE SEARCH STRATEGY

The literature search encompassed eight electronic databases (PubMed, MEDLINE, CINAHL, AMED, PsycINFO, ProQuest 5000, SportDiscus, and SCOPUS). A keyword list was developed for each of the following areas: epidemiology of sudden loading and lifting; sudden loading (practice, warning, posture, fatigue); repetitive lifting - posture and fatigue (subjective (psychophysical), physiological, lumbar spine kinematic and muscular responses); functional anatomy and biomechanics of trunk musculature; and methodological issues related to EMG and kinematic data collection and analysis (Table 2.1).

Table 2.1. The key themes and search terms used for the literature review.

Key themes				
Epidemiology of sudden loading and lifting	**Sudden loading**	**Repetitive lifting – posture and fatigue**	**Trunk musculature anatomy and biomechanics**	**Methodological issues: EMG and Kinematics**
		Search terms used		
epidemiology, aetiology, workplace, prevalence, risk factors, incidence, low back pain/injury/disorder, work environment, accidental, sudden loading, lifting, manual handling, frequency, fatigue, physiological, occupational, biomechanical, trunk motion, kinematics	sudden load, perturbation, translation, muscle latency, reaction time, stability, posture, practice, warning, trunk flexion, trunk loading, lordosis, bending, kinematics, electromyography (EMG), activation, motor control, neuromuscular, fatigue	lifting, lowering, technique, posture, squat, stoop, freestyle, manual handling, subjective, fatigue, perceived exertion, psychophysical, physiological, heart rate, oxygen uptake, cardiorespiratory, energy expenditure, lumbar spine, electromyography (EMG), amplitude, muscle activation /contraction, kinematics, bending, moment, forces, repetition, loading, weight lifting, frequency	muscle, stability, thoracic, lumbar spine, torso, moment, torque, biomechanics, anatomy, morphology, paraspinal, erector spinae, longissimus, iliocostalis, multifidus, abdominal, internal oblique, external oblique, rectus abdominis, co-activation, fibre, intra abdominal, thoracolumbar, lumbar fascia, electromyography (EMG), cyclic loading, muscle length, flexion, extension	skin markers, movement artefact, x-ray, radiographic, lumbar spine, reliability, validity, erector spinae, paraspinal muscle posture, length, velocity, fatigue, electromyography (EMG), Fourier, spectral, frequency analysis, electrode, noise, error

The search was limited to articles appearing in English journals prior to 2009. References were screened on the bases of title and abstract and selected for further review. Key references that were identified were directly inputted into the SCOPUS database which provided links to related articles where the reference had been cited. A manual search was also undertaken of all reference lists in order to supplement the findings of the electronic search.

EPIDEMIOLOGICAL ASPECTS OF SUDDEN LOADING AND REPETITIVE LIFTING

Low back pain is one of the most common and costly work-related problems facing western societies. The prevalence of LBI is particular evident in vocations involving manual materials handling (Garg and Moore, 1992; Hoogendoorn et al., 2000; Marras et al., 1993; Waters, Dick, Davis-Barkley, and Krieg, 2007) and approximately half of the workers that develop low back pain at work do so within the first year of employment (Van Nieuwenhuyse et al., 2004).

Accidental LBIs have been associated with longer absence from work and a higher rate of reoccurrence than non-accidental LBIs (Troup, Martin, and Lloyd, 1981). Up to 50% of incidents attributed to LBI at work involve unforeseen events or accidents (Manning et al., 1984). The majority of these accidental injuries involve sudden unexpected disturbance of postural equilibrium due to sudden unexpected loading or lifting, or arise from slips and falls (Andersson, 1981; Magora, 1973; Manning et al., 1984; Manning and Shannon, 1981; McCoy et al., 1997; Troup et al., 1981). Troup et al. (1981) reported that approximately a third of accidental LBI injuries occurred whilst handling loads and that up to 20% of accidental injuries were associated with sudden jerks or twists. Manning et al. (1984) investigated car factory workers with a history of LBI and found that a large percentage of these accidental injuries involved a sudden disturbance of equilibrium whilst carrying loads. Furthermore, up to 12% of total accidental LBI's involved a sudden unexpected load to the trunk. Examples of such events included suddenly having to catch a dropping piece of equipment or a partner dropping the end of a load (Manning et al., 1984; McCoy et al., 1997).

In addition to sudden unexpected loading, one of the most common work-related risk factors that has been associated with LBI is lifting (Frymoyer et al., 1983; Garg and Moore, 1992; Kelsey et al., 1984; Magnusson et al., 1990; Marras et al., 1993). Unlike sudden loading the majority of lifting related LBIs are often non accidental and occur as a consequence of prolonged and repeated exposure to manual handling tasks during normal working activities (McCoy et al., 1997). Those manual vocations that are at greatest risk of injury involve highly repetitive tasks performed at high repetitions. For example, Chaffin and Park (1973) reported that those jobs which involved greater than 150 lifts per day were associated with a greater incidence of LBI than jobs which involved less frequent lifting rates. Marras et al. (1995) found that those individuals undertaking manual handling jobs with a high incidence of LBI performed on average twice as many lifts per hour than those jobs considered to be at low risk of injury.

The postures (squat vs stoop vs mixed) adopted during high frequency lifting have been suggested to influence the physiological strain, spinal mechanics and associated risk of fatigue and LBI (Hagen et al., 1994; Revuelta et al., 2000). Whilst there appears to be no epidemiological studies that have investigated the association between lifting posture and LBI, a limited number of studies have reported an association between trunk posture characteristics and LBI. Most of these studies have determined trunk motion parameters from video recordings of workers performing manual handling tasks. For example, Punnett et al. (1991) found that back disorders were associated with manual handling activities that required mild (21-45 degrees) and severe (greater than 45 degrees) forward flexion. In a three year prospective study of workers from 34 companies in the Netherlands, Hoogendoorn et al. (2000) found a significant relationship between LBI and working in extreme trunk flexion (a minimum of 60 degrees flexion).

Marras et al. (1995) used a lumbar motion monitor to measure lumbar posture during different manual handling activities. Jobs were classified in terms of LBI risk (low, medium and high). It was found that in jobs that had medium and high risk of LBI, subjects flexed their spine to a greater extent during work tasks than workers in low risk occupations. Trunk velocity was also shown to be a strong predictor of LBI. In a later study, and using data from the lumbar motion monitor, it was shown that lumbar spine velocity measures were similar for the low, medium and high risk groups when the spine was flexed less than 15 degrees. However, for greater degrees of lumbar flexion the combined lumbar velocities for medium and high risk groups where twice that calculated for the low risk group (Fathallah, Marras, and Parnianpour, 1998). These findings would appear to support anatomical and biomechanical studies that have indicated that postures in extreme lumbar flexion combined with high rates of change in lumbar curvature increase loading forces on the passive tissues of the spine (Adams and Dolan, 1996; McGill, 1997a; Potvin et al., 1991; Wang et al., 2000).

Closely associated with high repetition and prolonged exposure to manual handling tasks are the effects of physical fatigue. Increased rates of LBI have been found in jobs that are highly repetitive and require high energy demands compared to jobs with low energy demands (Andersson, 1981). Increased occurrence of LBI has been associated with greater degrees of subject fatigue at the end of a working day (Sveistrup and Woollacott, 1997) and it would also seem that those individuals with poor back muscle endurance capacity are more likely to have LBI than individuals who have good muscular endurance (Biering-Sorensen, 1984). Although no clear relationship has been establish between lifting posture and fatigue, a study by Magnusson et al. (1990) found that individuals performing a squat lift had a lower lifting capacity in the afternoon compared to the morning, whereas lifting capacity did not significantly change throughout the shift when using a stoop lift.

POSTURAL RESPONSES TO SUDDEN LOADING

As epidemiological evidence has shown a clear association between sudden unexpected loading and LBI, researchers have attempted to understand possible injury mechanisms by examining postural responses to sudden unexpected loading in the laboratory setting. These studies have found that an unexpected loading event represents an injury scenario where even a relative small external load can potentially cause injury (Cholewicki and McGill, 1996; Mannion et al., 2000). In general terms, when a load is suddenly and unexpectedly applied to the trunk, the motor system most often responds within one second to complete the appropriate postural adjustments and return the body to equilibrium (Carlson, Nilsson, Thorstensson, and Zomleferm, 1981). Failure to elicit appropriate postural responses has the potential to place excessive stress on osteoligamentous structures of the spine and increase spinal loading (Mannion et al., 2000). Postural responses to sudden loading have been termed automatic responses (Diener, Horak, and Nashner, 1988; Horak and Nashner, 1986; Nashner and Cordo, 1981), and were originally thought to be responses to the stretch reflex (Liddell and Sherrington, 1924; Magnus, 1925). However, postural responses to perturbation are now considered to have greater central control with feedback from the sensory system utilised (Cordo and Nashner, 1982; Forssberg and Hirschfeld, 1994; Horak and Nashner, 1986). This has led to an interest in how factors such as practice (Lavender et al., 1993; Pedersen et al., 2004; Skotte et al., 2004), prior warning (Cresswell et al., 1994; Granata et al., 2001; Lavender et al., 1993; Leinonen et al., 2002; Thomas et al., 1998), and fatigue (Granata et al., 2001; Granata et al., 2004; Herrmann et al., 2006) may influence postural responses to sudden loading, particularly when conducted during manual handling tasks.

Sudden loading: reflex and centrally driven theories

Early schools of thought proposed that postural responses were a product of the stretch reflex and generated by peripheral feedback mechanisms (Liddell and Sherrington, 1924; Magnus, 1925). The stretch reflex explanation for postural control in response to perturbation now seems unlikely for a number of reasons. Firstly, postural musculature responses have onset latencies that range between 50 ms (medium latency) and 150 ms (long latency) (Diener et al., 1988; Henry, Fung, and Horak, 1998; Leinonen et al., 2002) which are longer than stretch reflexes elicited by the same muscle during tendon taps (Corden, Lippold, Buchanan, and Norrington, 2000; Zedka, Prochazka, Knight, Gillard, and Gauthier, 1999), but considerably shorter than voluntary responses of the same muscle groups (Cordo and

Nashner, 1982). Medium and long latency responses are considered long enough to allow signals to traverse the cerebral cortex and to be influenced by central and polysynaptic pathways and processes (Magnusson et al., 1996). Secondly, postural muscle activation sequencing often involves early activation of muscles that would not be stretched by the perturbation (Cresswell et al., 1994; Henry et al., 1998; Horak, Nashner, and Diener, 1990). Thirdly, postural response patterns are often unchanged by alterations to the sensory input in the form of amplitude and speed of the perturbing load (Diener et al., 1988; Matjacic, Voigt, Popovic, and Sinkjaer, 2001). Lastly, those individuals with sensory deprivation in the form of vestibular deficits and somatosensory loss caused by ischaemia of the lower limb appear to display similar postural responses to those with no sensory deficits (Horak et al., 1990).

Cordo and Nashner (1982), and Horak and Nashner (1986) were among the first groups of researchers to formally develop the theory that postural responses to sudden loading were of supraspinal origin. Cordo and Nashner (1982) originally found that activation of postural lower limb muscles preceded that of prime mover muscles of the upper limb when subjects were required to suddenly pull on a handle when standing upright. It was proposed that the early postural muscle activation synergies were predetermined in preparation for the perturbation. These postural muscle synergies became known as the postural set (Cordo and Nashner, 1982; Horak and Nashner, 1986). Forssberg and Hirschfeld (1994) further developed the concept of the postural set and hypothesised that the postural set was a centrally derived set of muscle patterns, which provided spatial and temporal information to muscle groups that could be modified through the interaction with peripheral multisensory information. Forssberg and Hirschfeld suggested that the spatial component of the postural set provided information regarding which specific muscles would be recruited. For example, during anterior perturbation of the trunk, the anterior and posterior musculature of the trunk and thigh are recruited (Horak and Nashner, 1986). The temporal component of the postural set was related to muscle sequencing information. This is illustrated during sudden forward trunk perturbation where abdominal musculature is often recruited in advance of one of the posterior trunk muscles and the muscles of the lower limb (Cresswell et al., 1994; Henry et al., 1998).

Sudden loading: influences of prior experience and practice

Given that the postural responses may be centrally driven it has been proposed that prior exposure to a sudden load or practice may modify postural responses to sudden loading. By providing prior exposure to a sudden loading event, the feedback gained from that exposure may allow the central nervous system to modify subsequent postural responses to a similar loading event (Horak and Nashner, 1986) potentially reducing subsequent loading on the lumbar spine (Lavender et al., 1993). Lavender et al. (1993) hypothesised that task experience would lead to a strategy of increased preparatory co-contraction activity of erector spinae (ES) and abdominal muscles, which would in turn remove the “slack” from the system, enabling a faster and stiffer response to the loading event. Lavender et al. found that repeated sudden loading over five 30 minute sessions had no consistent effect on preparatory muscle torques of the abdominal and ES muscles. When compared to initial responses some subjects showed evidence of increased preparatory torque following repeated exposure, whereas others showed a reduction in preparatory levels of muscle torque production. Nonetheless, irrespective of differing preparatory muscle strategies within the group, all subjects (4) reduced the degree of trunk flexion with repeated exposure.

The influence of prior experience on muscle onset latencies to sudden anterior loading is also unclear. Wilder et al. (1996) found no practice effect on onset latencies of the ES muscles in healthy and low back pain patients who were subjected to six trials involving a weighted tennis ball being dropped into a pan held by the subjects. This finding was supported by Pedersen et al. (2004) who applied a sudden anterior load to the trunk via a wire attached to a chest harness. Following a single practice trail, the authors found no difference in onset latencies of the ES muscle group between trials 1 and 2, and 3 to 10, although the time taken to stop forward motion of the trunk decreased with practice.

Whilst Pedersen et al. (2004) indicated that practice may have minimal effect on muscle onset latency during sudden unexpected loading, subjects were permitted to perform familiarisation trials before data collection. By allowing familiarisation trials it is not possible to compare muscle onset latency differences between a “truly” unexpected sudden loading event (where the force, magnitude, and timing of the event are unknown) and subsequent similar loading events where the individual has been previous exposed to a number of loading parameters (Chow, Cheng, Holmes, and Evans, 2003). Chow et al. (2003) commented that this experience of a sudden perturbation prior to investigation is a major limitation for a number of studies as subsequent exposures are “not entirely unexpected”. Skotte et al. (2004) were one of the few groups that recorded muscle onset responses in subjects who had not been previously exposed to sudden loading. Skotte et al. found no significant difference in the onset latency of the ES muscle group between first (no prior exposure) and subsequent exposures to direct anterior loading of the trunk via a wired harness. However, Skotte et al. did report that the average time to stop forward flexion of the trunk between trials 3 to 10 was approximately 20% less than that exhibited in the first trial.

Sudden loading: effects of warning

Another factor that may allow the central nervous system to modify postural responses to sudden loading is warning. It has been proposed that the provision of a warning signal prior to sudden loading may heighten the awareness of the central nervous system and may lead to enhancement or modification of postural responses (Lavender et al., 1993; Thomas et al., 1998). Two key theories have evolved to explain the effect of prior warning on postural responses to sudden loading. The first theory is that sufficient warning of a sudden loading event allows co-contraction of abdominal and ES musculature prior to the loading event (Granata et al., 2001; Thomas et al., 1998). This pre-activation of abdominal and ES muscles acts to stiffen the trunk in advance of the load, consequently providing greater resistance to the external load (Thomas et al., 1998). It has been shown that if individuals are permitted to visually track the trajectory of the perturbing object there are increased levels of activity of ES prior to the sudden loading event compared to loading situations where no visual warning is provided (Cresswell et al., 1994; Lavender et al., 1993; Leinonen et al., 2002). In some cases there is also associated co-contraction of the abdominal musculature prior to the load being applied (Cresswell et al., 1994; Lavender et al., 1993; Thomas et al., 1998). It has also been reported that during sudden anterior loading of the trunk, trunk displacement is reduced with the addition of visual warning (Cresswell et al., 1994; Thomas et al., 1998).

Although preparatory agonist-antagonist co-contraction is evident in situations where subjects are provided with prior visual information of the perturbation, verbal warning does not seem to alter activation levels of trunk musclulature (Granata and Orishimo, 2001). Whilst preparatory co-activation of anterior and posterior trunk muscles may increase spinal stability, increases in muscle activation levels require greater energy demands and have the potential to increase spinal loading (Gardner-Morse and Stokes, 2001). This has led to the theory that warning may induce a state of readiness of the central nervous system that changes motor neuron pool excitability and elevates the sensitivity of the motor neuron pool to supraspinal commands (Hasbroucq, Kaneko, Akamatsu, and Possamai, 1999). This would potentially lead to earlier initiation of the postural set allowing muscle synergies to respond to the sudden loading stimulus more quickly, without needing to increase levels of muscle co-contraction prior to the loading event. Prior warning of a sudden loading event has been shown to reduce onset latencies of specific leg muscles during anterior trunk (Leinonen et al., 2002) and underfoot perturbation (McChesney et al., 1996). What is less evident is whether the provision of warning alters the phasing of muscle synergies (the postural set) and kinematic initiation patterns elicited in response to the sudden load, although findings from Cresswell et al. (1994) would indicate that temporal phasing patterns of trunk muscles remain unchanged when provided with prior warning.

Sudden loading: effects of fatigue

Besides the potential influence of prior experience and warning, fatigue may also affect postural responses to sudden loading. Workplace activities such as repetitive lifting have been shown to lead to trunk muscle fatigue (Bonato et al., 2003; Dolan and Adams, 1998; Sparto and Parnianpour, 1998) which may influence muscular and postural responses to sudden loading (Granata et al., 2001; Granata et al., 2004). It has been proposed that lifting-induced fatigue of the ES muscles can potentially reduce their ability to generate force in order to resist anterior loading forces and maintain spinal stiffness (Granata et al., 2004). The inability of the ES muscle group to produce sufficient spinal stiffness when in a state of fatigue may in turn reduce spinal stability during sudden anterior loading (Granata et al., 2004). One mechanism that may compensate for ES muscle fatigue is increased baseline levels of agonist and antagonist trunk muscles prior to sudden loading (Granata et al., 2004). The preparatory co-activation of abdominal and ES muscles would then increase spinal stiffness and stability in preparation for the sudden loading (Granata et al., 2004). In support of this hypothesis, increases in activation levels of ES and abdominal musculature prior to the sudden load and subsequent reductions in trunk motion have been shown following fatiguing protocols involving static trunk flexion and short duration (two minutes) high frequency repetitive lifting (Granata et al., 2001; Granata et al., 2004).

Mixed findings have also been found for the influence of fatigue on muscle onset latencies in response to sudden loading. Wilder et al. (1996) showed that following fatigue induced by a prolonged period of exposure to low frequency vibration that the onset latency of the ES muscle group was longer in response to the sudden loading of a hand-held box after vibration-induced fatigue. Chow, Man, Holmes, and Evans (2004) also found that following fatigue, there was an increase in the ES muscle relaxation times in response to sudden unloading. In this instance, sudden unloading was performed in a stoop posture. Although no physiological explanations were given for increases in ES muscle latencies following fatigue, local muscle fatigue has been associated with decreased central motor neuron excitability and a metabolically inhibitory effect induced from group III and IV free nerve endings (Avela and Komi, 1998; Balestra, Duchateau, and Hainaut, 1992; Bigland-Ritchie, Dawson, Johansson, and Lippold, 1986). This inhibitory effect has been suggested to be responsible for the decrement in muscle reflex responses during fatiguing contractions (Avela and Komi, 1998; Balestra et al., 1992; Bigland-Ritchie et al., 1986). Of note, when interpreting the findings from Wilder et al.'s (1996) study it should be considered that vibration-induced fatigue may influence postural responses in a different manner from lifting-induced fatigue. For example, it has been shown that vibration of the muscle tendons independent of local muscle fatigue can increase the length of short and medium latency muscle reflex responses (Bove, Nardone, and Schieppati, 2003).

The implications of using vibration as a fatiguing protocol may partly explain why the findings from Wilder et al.'s (1996) study are not supported by other investigators who have used dynamic protocols to fatigue the ES muscle group. These studies (Granata et al., 2004; Herrmann et al., 2006) have reported that despite electrophysiological evidence of ES muscle fatigue there was no change in the onset latencies of the ES and abdominal muscles, or subsequent spinal motion following fatiguing interventions. Some studies have even reported that following fatigue there were decreases in onset latencies of postural muscles in response to voluntary trunk perturbation (hand raising) (Allison and Henry, 2002; Strang and Berg, 2007). Allison and Henry (2002) suggested that the early activation of postural muscles in response to self-induced perturbation when fatigued may be an adaptation which takes place to allow more time for the muscles to reach the force requirements of the task. However, voluntary induced postural perturbation differs from that of an unexpected load, as the focal movement inducing the perturbation is initiated internally and onset latencies of postural muscles in these situations occur well in advance of the perturbing movement (Allison and Henry, 2002; Strang and Berg, 2007).

Sudden loading: effects of initial standing posture

The majority of studies investigating sudden anterior loading of the trunk have often induced perturbation whilst the subject has been in an upright standing posture (Cresswell et al., 1994; Lavender et al., 1993; Pedersen et al., 2004; Wilder et al., 1996). However, individuals involved in manual handling activities who are exposed to sudden loading often adopt postures that use a large range of trunk flexion (Marras et al., 1995). Of concern, postures that have a greater degree of trunk flexion in the sagittal plane have been shown to be at increased risk of LBI when compared to postures where the lumbar spine is less flexed (Marras et al., 1995).

Sudden loading of tissue in a stoop posture (where the legs are relatively straight and the hips and lumbar spine are flexed) may present a greater risk of injury to the spine when compared to standing upright. At the end range of lumbar spine flexion shear forces (McGill et al., 2000; Potvin et al., 1991) and intradiscal pressure (Takahashi, Kikuchi, Sato, and Sato, 2006) are greater than when the spine is in a neutral upright position. When approaching the end range of lumbar flexion the LES muscles change geometry and their ability to exert an extension moment and support anterior shear forces may be compromised (Macintosh et al., 1993; McGill et al., 2000). At this range of flexion, the LES may also become “electrically silent” (Floyd and

Silver, 1955), even when holding a moderate load in the hands (Gupta, 2001; Takahashi et al., 2006). In this situation, if the shear forces are close to the threshold for soft tissue damage, then even a small additional load could impart enough shear force to potentially damage the spine.

Conversely, the increased recruitment of the posterior ligamentous system in combination with the passive stretch of the ES muscles that takes place with increased trunk flexion may increase stability (Arjmand and Shirazi-Adl, 2005) and provide greater resistance to sudden loading. In this respect, Mannion et al. (2000) found that during sudden loading in a flexed posture, although peak total compressive forces of the spine were greater than in erect standing, the addition of compressive forces arising from the load were similar between postures. Furthermore, there was evidence that in a slightly flexed posture the addition of compression forces generated from the sudden load were less than those generated in erect standing.

It also seems likely that sudden anterior loading in upright posture may also pose a risk of injury, but the underlying mechanism may be different from that of a flexed spine. In a neutral lumbar spine there is minimal ligamentous contribution to resisting a bending moment and passive structures of the lumbar spine contribute to total spinal stiffness considerably less than when the spine is flexed (Ebenbichler, Oddsson, Kollmitzer, and Erim, 2001; McGill, Seguin, and Bennett, 1994). Modelling techniques taking into consideration the passive and active elements of the spine indicate that lumbar spine stability in upright standing is considerable less than in lifting positions where the trunk is inclined forward (Cholewicki and McGill, 1996; Granata and Wilson, 2001). Therefore, it has been hypothesised that injury risk may be as similar in situations where lumbar spine instability is high (upright standing) and those where there is high task demand (such as lifting a load where the trunk is inclined forward) (Cholewicki and McGill, 1996; McGill, 1997a).

Cholewicki and McGill (1996) also suggested that in an upright position with minimal muscle activation, the large contraction forces (as reported during unexpected loading) generated by the more global spanning muscles of the trunk and without co-ordinated activation of more local intersegmental muscles (for example, multifidus) can potentially reduce stability and increase the risk of injury to the spine (Cholewicki and McGill, 1996; McGill, 1997a). Evidence in support of reduced stability when standing upright may be reflected by greater rate of change in lumbar curvature observed during sudden loading in an upright posture when compared to a similar loading event in a posture where that spine is moderately flexed (Mannion et al., 2000).

In summary, whilst it is evident that sudden unexpected loading is associated with a number of LBIs, the mechanisms of injury are not clearly understood. Although it is generally agreed that postural responses are centrally driven there is still controversy as to how ergonomic interventions such as prior exposure (practice), warning, and work-related factors such as the posture adopted by the worker during the task and fatigue may influence postural responses to sudden loading. Whilst a number of studies have investigated postural responses to sudden loading in upright standing, few investigations have compared these responses to those elicited when standing in a stoop posture, and when physically fatigued by work-related manual handling activities.

REPETITIVE LIFTING

Vocations involving repetitive lifting have been shown to be at an increased risk of LBI (Frymoyer et al., 1983; Garg and Moore, 1992; Kelsey et al., 1984; Magnusson et al., 1990; Marras et al., 1993). Evidence also suggests that at higher lifting frequencies the risk of LBI also increases (Chaffin and Park, 1973; Marras et al., 1995). Some manual handling occupations have been found to involve lifting rates in excess of 1500-2500 lifts per hour (Marras et al., 1995; Marras et al., 1993), with air cargo workers able to sustain lifting and carrying rates of between 16-22 times per minute over a 2 hour period (Mital, Foononi-Fard, and Brown, 1994). Continued and repeated exposure to manual handling at these high rates, even at sub maximal effort, can increase the physical and physiological demands on the body beyond that which can be realistically sustained over time (Mital et al., 1994). In such cases prolonged repetitive lifting may lead to premature termination of the lifting task due to fatigue (Fernandez, Ayoub, and Smith, 1991).

Physical fatigue has been defined as a state of disturbed homeostasis attributed to work and the working environment (E. H. Christensen, 1960). This disturbance in homeostasis may be reflected by a failure to maintain steady state in central cardio-respiratory measures, such as HR and oxygen consumption (VO_2) (Gaesser and Poole, 1996; Herman et al., 2003), and can result in changes in muscle activation characteristics and lumbar spine motion patterns (Bonato et al., 2003; Dolan and Adams, 1998). Often associated with these physiological changes is an increased perception of subjective effort and a decrease in efficiency or performance (Basmajian and De Luca, 1985; Fernandez et al., 1991; Grandjean, 1979). The assessment of physical fatigue during high frequency repetitive lifting provides a challenge as the original posture adopted (stoop, squat and mixed) may influence physiological parameters and subjective measures (Garg and Saxena, 1979; Hagen et al., 1993; Hagen and Harms-Ringdahl, 1994; Rabinowitz et al., 1998; Straker and Cain, 1999). Therefore, this section of the literature review will focus on how the original lifting posture adopted influences subjective (RPE), physiological (HR) and postural responses during repetitive lifting, and the implications of such responses in respect to physical fatigue.

Repetitive lifting: posture and ratings of perceived exertion

A common measure of fatigue during manual handling activities is a rating of perceived exertion (RPE) (Asfour, Ayoub, Mital, and Bethea, 1983; Hagen et al., 1993; Hagen and Harms-Ringdahl, 1994; Hagen et al., 1994; Hattori et al., 1996; Rabinowitz et al., 1998; Straker and Cain, 1999). Developed by Borg (1970) the RPE scale is a 15 point categorical scale with end points ranging from 6 and 20. Adjacent to specific points along the scale are verbal descriptions of perceived effort. For example a score of 7 indicates a perceived effort of “very, very light” and a score of 19 indicates a perceived effort of “very, very hard” (Borg, 1970, 1982). It has been suggested that RPE is influenced by two key physiological factors. Firstly, it is believed that central physiological factors associated with the stimulation of the organs of circulation and respiration influence an individual’s RPE (Borg, Ljunggren, and Ceci, 1985). A positive correlation between RPE and central physiological measures, such as HR and VO_2, has been shown during repetitive lifting at different frequency-weight combinations (Asfour et al., 1983; Hagen et al., 1994). The fact that correlations between RPE and central physiological responses are considerably weaker during stoop than squat lifting (Hagen et al., 1994) suggests that changes in RPE cannot be solely attributed to the increasing demands of the cardiorespiratory system. Other factors that may influence RPE are local physiological responses, such as muscle fatigue (Dedering, Nemeth, and Harms-Ringdahl, 1999) that contribute to perception of strain and effort from of the working muscles and joints (Borg et al., 1985).

Given that RPE may reflect cardio-respiratory stress and/or local muscle sensations associated with fatigue, it has been argued that the lifting posture (squat, stoop or mixed) that elicits the lowest RPE for a given work load would be preferable and possibly present a lower risk of injury (Straker and Cain, 1999). At sub-maximal effort and when lifting at low frequencies (less than six lifts per minute) RPE has been shown to be higher for the squat lift when compared to other lifting techniques (stoop and semi-squat) (Rabinowitz et al., 1998; Straker and Cain, 1999). The RPE for either posture in these studies was between "fairly light" and "somewhat hard", and within the range that workers involved in highly repetitive manual handling would consider acceptable when continually handling materials over a two hour period (Mital et al., 1994). During sub-maximal lifting effort at lifting frequencies between 3 and 12 lifts per minute, the mass of the maximal acceptable weight of lift (MAWL) has also been found to be greater when subjects are instructed to execute stoop when compared to a squat lift (Garg and Saxena, 1979). Furthermore, Kumar (1984) found that when lifting a 10 kg basket at a lifting rate of 10 lifts per minute over a six minute period subjects rated the squat lift as the most tiring in over 74% of all lifts, whereas, only 17% of stoop lifts were rated as most tiring.

As lifting frequency increases, RPE increases, but the difference in RPE across lifting postures is less evident than at lower lifting rates (Hagen et al., 1993; Hagen et al., 1994; Mital et al., 1994). Hagen and Harms-Ringdahl (1994) required subjects to lift at five different frequency/weight combinations with lifting frequencies up to 20 lifts per minute. Although RPE increased with concurrent increases in the lifting frequency and weight lifted, there were no differences in RPE between lifting postures (Hagen et al., 1994). When the same group of subjects executed lifting to maximal effort no differences in RPE were observed between stoop and squat lifting postures despite significant increases in cardiorespiratory responses for the squat lift compared to stoop lift (Hagen et al., 1993).

Whilst there is less evidence of clear differences in central RPE between lifting postures when performing manual materials handling at high lifting rates, perception of local muscular effort may differ between postures. Hagen and Harms-Ringdahl (1994) asked subjects to rate local thigh and low back RPE after sub-maximal high frequency lifting using squat and stoop techniques. It was found that at a sub-maximal level, RPE for the thigh was higher for the squat lift than that recorded when stoop lifting, whereas low back RPE were higher for the stoop lift when compared to the squat lift (Hagen and Harms-Ringdahl, 1994).

Ratings of perceived exertion: instructed versus self-selected lifting posture

An issue when assessing lifting and its effects on RPE is that posture may be confounded by the novelty of learning a new technique. In the majority of studies, subjects were instructed on how to lift, rather than categorised into lifting technique using a self-selected method (Hagen and Harms-Ringdahl, 1994; Hagen et al., 1994; Rabinowitz et al., 1998; Straker and Cain, 1999). The difficulties that arise with instructing individuals to perform a novel lifting technique are that the instructed stoop or squat lifts are perceived as requiring more effort than a self-selected lifting technique (Garg and Saxena, 1979; Kumar, 1984). Furthermore, when instructed on a specific lifting technique a large percentage of individuals perform the technique at a low skill level (Hart et al., 1987). For example, when individuals are asked to perform instructed squat and stoop lifts there appears to be no pattern in subjective preference of a preferred lifting technique (Rabinowitz et al., 1998). In contrast, when subjects are asked to compare self-selected with instructed squat and stoop lifting techniques, subjects rate the self-selected lifting technique as the least tiring lift in the majority of lifts (Kumar, 1984). In addition, Garg and Saxena (1979) found that subjects perceived they could lift a heavier weight when they were permitted to use a self-selected technique, than when instructed to use a stoop or squat technique. Given that self-selected lifting can be categorised into squat, stoop and intermediate lifting techniques (Bonato et al., 2003; Burgess-Limerick et al., 2001; van Dieen et al., 1998) a better method of assessing the effects of posture on RPE during repetitive lifting would be to allow individuals to use their preferred technique that would typically be used in the workplace.

Repetitive lifting: posture and heart rate

Heart rate (HR) is one of the most commonly used methods to assess physiological stress, particularly in the workplace. HR is a relatively easy and simple measure to use and has a close relationship with more laboratory based measures of energy expenditure, such as VO_2 (Swain, Leutholtz, King, Haas, and Branch, 1998). The National Institute for Occupational Safety and Health (NOISH) guidelines have no specific recommendations for HR limits during manual handling (Waters, Putz-

Anderson, Garg, and Fine, 1993), however, studies reporting recommended maximal HR for sustainable continuous work over an eight hour shift propose a range between 90 and 130 beats per minute (Legg and Myles, 1985; Mital, 1984). Interestingly, Mital et al. (1994) found individuals involved in highly repetitive lifting tasks sustained HRs between 150-162 beats per minute over a two hour period.

On the premise that increases in HR for a given workload may place increased physiological stress on the worker, it seems reasonable to assume that the lifting posture (squat, stoop or mixed) that elicits the lowest HR would be preferable. It has been proposed that during repetitive lifting and lowering at a given workload the squat lift places greater demand on the cardiovascular system than the stoop lift, as it requires the additional activation of the leg musculature (Duplessis et al., 1998; Hagen et al., 1993; Revuelta et al., 2000). The effects of lifting posture at sub-maximal effort during low frequency (less than six lifts per minute) repetitive lifting indicates that average HR appears lower for mixed (semi-squat) and stoop lifting when compared to squat lifting (Rabinowitz et al., 1998; Straker and Cain, 1999). For example, Rabinowitz et al. (1998) reported that when lifting a beer crate weighing 20% of the subjects body mass from the floor onto a 75 cm high surface, average HR for a squat lift was 123 beats per minute, whereas employing a stoop lift elicited an average HR of less than 100 beats per minute. Straker and Cain (1999) allowed subjects to adjust the weight of box in order to determine the MAWL over a 20 minute period whilst repetitively lifting from floor to knuckle height at a frequency of 4.3 lifts per minute. HRs elicited during this period were on average 119 and 109 beats per minute for the squat and stoop lifts, respectively. It should be noted that whilst HRs were found to be higher for squat lifters during sub-maximal effort (Rabinowitz et al., 1998; Straker and Cain, 1999) these HRs were still within the limits considered safe during the performance of manual handling activities over an eight hour period (Legg and Myles, 1985; Mital, 1984).

Studies have shown that with increases in lifting frequency there is a corresponding increase in HR (Hagen et al., 1994; Welbergen et al., 1991). However, the effects of lifting posture on HR during repetitive lifting at high lifting rates (greater than 16 lifts per minute) is less apparent than at lower lifting rates. Welbergen et al. (1991) investigated students with previous weight lifting experience who repetitively lifted and lowered a 19 kg barbell from floor to waist height at a frequency of 10 and 20 lifts per minute. This task was performed in a stoop and squat posture for a period of five minutes. Whilst HRs were similar for stoop and squat lifting postures Welbergen et al. reported a significant increase in HR for the higher lifting rate. In contrast, Hagen et al. (1994) found that experienced forestry workers lifting at rates of up to 20 lifts per minute for a 4-6 minute period produced higher HRs for the squat than stoop lift. Hagen et al. (1993) also had the same cohort of subjects perform an incremental lifting task to ascertain peak VO_2. The task required subjects to lift and lower at a rate of 20 lifts per minute with the addition of a 2.5 kilogram weight every two minutes until exhaustion. Maximal HRs recorded during the incremental maximal lifting test were on average 12 beats higher when using the squat lift than when using a stoop lift, despite the weight of the box at maximal effort being similar for both postures. The studies untaken by Hagen and colleagues indicated that whist stoop lifting at sub-maximal effort elicited lower HRs for a given load than when squat lifting, squat lifters were able to sustain higher HRs at maximal effort.

The inconsistency in peak and average HR measures during repetitive lifting highlight the difficulty of interpreting the effect of lifting posture on cardiovascular strain and subsequent fatigue (Asfour, Genaidy, and Mital, 1988). A different approach to predicting excessive strain on the cardiorespiratory system during repetitive exercise, such as lifting, is to observe HR over time. When performing any exercise at a constant load at low or moderate intensity there is usually a mono-exponential increase in HR and VO_2 until a steady state is reached within approximately 2-3 minutes (Bouckaert, Jones, and Koppo, 2004; Deley, Millet, Borrani, Lattier, and Brondel, 2006). If the body is unable to meet the work requirements aerobically, instead of attaining a steady state, there will be a slow rise in HR and RPE over time (Herman et al., 2003), which in turn may lead to depletion in energy stores and eventual fatigue (Gaesser and Poole, 1996). This has been termed the "slow component" rise and differs from the cardiovascular drift observed during prolonged exercise in that the slow rise in physiological variables can occur within two minutes of initiating the exercise (Barstow and Mole, 1991; Bouckaert et al., 2004). The slow component rise usually occurs during constant load exercise which exceeds lactate threshold but is below maximal oxygen uptake (VO_2 max) (Barstow and Mole, 1991; Herman et al., 2003). It has been suggested that the slow component rise may be due to factors associated with fatigue, such as increases in blood lactate, catecholamine accumulation, and elevation of muscular temperature (Gaesser and Poole, 1996). It has been proposed that during constant heavy load exercise the recruitment of fast twitch fibres, which are less aerobically efficient than slow twitch fibres may contribute to a slow component rise in HR and VO_2 (Barstow, Jones, Nguyen, and Casaburi, 1996; Krustrup, Soderlund, Mohr, and Bangsbo, 2004).

To date, few repetitive lifting studies have investigated HR and RPE patterns over time in individuals using different lifting postures. There is some evidence that workers may tend to overestimate the maximal weight that can be repeatedly lifted over prolonged periods, leading to increases in subjective effort to a point where they have to prematurely terminate the lifting task (Fernandez et al., 1991). Hagen et al. (1994) did not specifically investigate HR or VO_2 kinetics, but did state that physiological measures of varying workloads were taken at 4-6 minutes until a steady state VO_2 was observed. These observations indicated that when lifting at frequencies as high as 20 lifts per minutes, forestry workers were able to generate and maintain consistent levels of VO_2 for both stoop and squat postures. Rabinowitz et al. (1998) made comment that when subjects used a stoop lifting technique, steady state HR was attained relatively quickly, whereas for the squat lift a continual increase in HR was observed. Rabinowitz et al. suggested that the increase in HR was an attempt to meet increasing physiological demands of the squat lift, and reflected more rapid fatigue.

In summary, there is a dearth of information investigating the effects of self-selected lifting posture on temporal patterns of RPE and HR. An understanding of temporal changes in RPE and HR patterns during the course of repetitive lifting may provide insight into the effects of lifting posture on task demands during manual handling activities. Slow rises in physiological and subjective measures are likely to indicate that an individual is unable to sustain task demands leading to fatigue and eventual exhaustion (Herman et al., 2003).

Repetitive lifting: posture and lumbar spine kinematics

Although RPE and HR provide an indication of the subjective effort and physiological strain imposed by repetitive lifting, lumbar spine kinematics can provide insight into the biomechanical stress imposed on the lumbar spine when adopting different lifting postures. The degree of lumbar flexion and rate of change in lumbar spine curvature (velocity) are considered two important kinematic variables for assessing repetitive lifting activities (Adams and Dolan, 1991, 1996; de Looze et al., 1993; Dolan, Mannion et al., 1994; Marras et al., 1995; Potvin et al., 1991). This point has been demonstrated through *in vitro* and epidemiological studies which have shown a strong association between lumbar spine kinematics (flexion and velocity) and LBI (Adams and Dolan, 1991, 1996; Dolan, Mannion et al., 1994; Marras et al., 1995; Wang et al., 2000). Whilst LBI has been associated with lumbar spine kinematics, less is known about how self-selected posture (squat, stoop or mixed) influences lumbar spine kinematic parameters during repetitive lifting, particularly when in a state of physical fatigue.

Lumbar flexion

Epidemiological evidence indicates that individuals who perform manual handling activities in high degrees of lumbar flexion are at greater risk of injuring their lumbar spine than those who flex their spine less during manual handling activities (Hoogendoorn et al., 2000; Marras et al., 1995; Punnett et al., 1991). The degree of lumbar flexion at the initiation of a lift and at the end of the lowering phase is important as its extent determines the bending moment acting on the passive tissues of the spine (Adams and Dolan, 1991; Dolan, Mannion et al., 1994; Potvin et al., 1991). Cadaver studies and *in vivo* experiments have found that the bending moment resisted by the ligaments and discs of the spine increases exponentially when the spine is flexed beyond 80% of maximal *in vivo* flexion (Adams and Dolan, 1991; Dolan, Mannion et al., 1994). It has been argued that high lumbar flexion involves the recruitment of passive tissues of the spine which may serve to decrease the activation requirements of the ES muscles (Gracovetsky, Farfan, and Lamy, 1977). However, others predict the recruitment of the interspinous and supraspinous ligaments during lumbar flexion leads to an increase in the anterior shear forces acting on the spine (McGill, 1997a; Potvin et al., 1991).

The potential for injury when the spine is placed in near end range of lumbar flexion without adequate muscle involvement increases as motion is repeated over time. For example, Adams and Dolan (1996) found that repeated flexion of cadaver segments to end range of motion (as determined from *in vivo* flexion) at a rate of 20 cycles per minute for 5 minutes resulted in a 15% decrease in passive tissue resistance to bending moment. Based on lumbar segment tolerance studies (Adams and Hutton, 1985), it was suggested that repetitive loading had the potential to injure soft tissue structures of the spine in situations where there was insufficient muscle activation (Adams and Dolan, 1996). Gallagher et al. (2005) used loading parameters on lumbar spine cadaver segments that simulated repeatedly lifting a 9 kg load at 0.33 Hz (20 lowers per minute) at 0, 22, and 45 degrees of lumbar flexion. These researchers found that loading conditions associated with full flexion (45 degrees) resulted in spinal tissue damage in an average of 263 cycles compared to 3257 and 8253 cycles required to induce spinal injury when the spine was flexed to 22 and 0 degrees respectively. These findings suggested that when the osteoligamentous spine was exposed to a simulated manual handling rate of 20 cycles per minute in near full range of lumbar flexion the time to failure would have been as little as 13 minutes.

The increased injury risk associated with performing manual handling activities at end range of lumbar flexion has led to an interest in the degree of peak lumbar flexion for different lifting postures. It has generally been found that when subjects are instructed to perform a stoop lift they flex their spine to a greater extent than when instructed to use a squat lift (de Looze et al., 1993; Dolan, Earley et al., 1994; Dolan, Mannion et al., 1994; Potvin et al., 1991). However, only a few studies have quantified the extent of lumbar flexion as a percentage of the individual's full range of flexion (Dolan, Earley et al., 1994; Dolan, Mannion et al., 1994; Potvin et al., 1991). Potvin et al. (1991) had subjects lift a box (handle height of 15 cm) from the floor using an instructed squat and stoop lift. Subjects who were instructed to lift a moderate mass (14 kg) using a squat lift flexed their lumbar spine to approximately 67% of its maximal range (based on full range flexion measures). The model developed by Potvin et al. predicted that the ligaments and discs of the spine had a minor role in resisting bending moments during squat lifting and that most of the peak extensor moment was produced by the ES muscles. When instructed to perform a stoop lift, subjects flexed the lumbar spine to 86% of their maximal *in vivo* limit, which resulted in a greater percentage of bending moment being resisted by the ligaments (Potvin et al., 1991).

In a series of studies conducted by Dolan and colleges (Dolan, Earley et al., 1994; Dolan, Mannion et al., 1994) it was found that when males were instructed to lift a weight lifters disc weighing 10 kg from the floor using a squat lift, peak lumbar flexion was only slightly less (81% of the *in vivo* maximal limit) than that exhibited by the stoop lifters in the Potvin et al. (1991) study. However, when asked to perform a stoop lift the lumbar spine was flexed to between 95% and 97% of maximal range of motion (Dolan, Earley et al., 1994; Dolan, Mannion et al., 1994). Using equations developed by Adams and Dolan (1991), it was predicted that the maximal peak flexion was 17%-20% and 31% of the elastic failure limit of the ligaments and discs during the squat and stoop lift, respectively (Dolan, Earley et al., 1994; Dolan, Mannion et al., 1994). Based on these predictive equations, Dolan, Earley et al. (1994) concluded that even when in maximal *in vivo* lumbosacral flexion there is a margin of safety against injuring the ligaments and discs of the lumbar spine. This safety margin has been attributed to structures such as the thoracolumbar fascia, skin, viscera, and the passive elastic properties of the ES muscles resisting a substantial percentage of the bending moment when in 100% *in vivo* flexion (Adams and Hutton, 1986; McGill et al., 1994). It has been predicted that the additional resistance of these passive structures at 100% maximal *in vivo* flexion would only allow the ligaments and discs of the lumbar spine to be stretched to less than 40% of their elastic limit (Adams and Dolan, 1991; Dolan, Earley et al., 1994).

Although postures that exhibit near end range maximal flexion may not approach the predicted tissue failure limits of the discs and ligaments, modelling studies have shown that as the lumbar spine approaches end range of flexion the recruitment of supraspinous and interspinous ligaments can substantially increase anterior shear forces acting on the lumbar spine (McGill, 1997a; Potvin et al., 1991). Furthermore, repetitive high frequency loading of the spine in near range end range flexion can potentially inflict damage to the vertebral endplate in a relatively short number of cycles when compared to loading the spine in a less flexed position (Gallagher et al., 2005).

Whether the difference in peak lumbar flexion observed in those individuals who are instructed to perform a stoop or squat lift is evident in subjects who naturally adopt different postures (squat, stoop or mixed) remains speculative. In this respect, McGill (1997a) argued that the extent of lumbar flexion may not be solely related to the lifting technique (stoop or squat). This notion is supported by evidence that subjects are able to exhibit a lifting posture that has substantial knee and hip flexion (as illustrated when using the squat technique) whilst flexing the spine to near end range lumbar flexion (Delitto and Rose, 1992; Dolan, Earley et al., 1994; McGill, 1997a). It has also been shown that in a bent knee posture with maximal lumbar flexion during the initiation of the lift LES may be electrically silent and all of the bending moment is resisted by passive tissues of the spine (McGill, 1997a; Potvin et al., 1991).

Some researchers have visually identified lifting posture (squat, mixed or stoop) from groups of subjects required to use a self-selected lifting technique during repetitive handling (Bonato et al., 2003; Burgess-Limerick et al., 2001; van Dieen et al., 1998). However, the majority of these studies have not attempted quantify lumbar motion for each identified posture. For example, Dolan, et al. (1994) found that when subjects were asked to perform a "freestyle" (self-selected) lifting technique they flexed their lumbar spine to a similar level to that produced during instructed squat lifts. van Dieen et al. (1998) used single subject data to illustrate lumbar angular motion differences between a self-selected "leg" (squat) and "back"(stoop) lifting technique. From the graphed lumbar motion data, an individual using a leg technique flexed their spine up to 20 degrees more than the subject using a back (stoop) technique. It should be noted that in squat lifting other factors such as relative arm length, the load lifted, and handle distance from the floor may influence lumbar curvature.

Lumbosacral velocity

Though substantial focus has been on peak lumbar flexion, less is known about the rate of change in lumbar curvature (velocity) for different self-selected postures during lifting and lowering. The speed at which lumbar curvature changes is important as postures that elicit higher lumbosacral velocities during manual handling activities are at more risk of LBI when compared to those that exhibit slower velocity (Marras et al., 1995). These epidemiological findings are supported by studies utilising cadaver and modelling techniques that indicate postures that induce large lumbosacral velocities during combined lumbar spine compression and flexion (as exhibited during lowering activities) increase loading forces on the passive structures of the spine when compared to flexing the lumbar spine at slower velocities (Adams and Dolan, 1996; Wang et al., 2000). For example, Adams and Dolan (1996) found more rapid increases in the bending moment of cadaver lumbar segments that were flexed over a one second period compared to a 10 second period. Wang et al. (2000) used a finite element model of the L2-3 lumbar motion segment to simulate combined flexion and compression of the lumbar spine during a lowering task at different velocities. These authors predicted that a higher rate of flexion loading increased the resisted bending moment near the end of motion and induced higher stresses on the disc when compared to slower rates of loading.

In *in vivo* studies, Dolan and Adams (1993b) reported that lifts inducing greater lumbosacral velocity significantly increased peak extensor moment when compared to lifts with slower rates of change in lumbar curvature. It was also predicted that the higher extensor moment would contribute to an increase in peak spinal compression at faster rates of lumbar motion. Granata and Marras (1995) used lumbar kinematic and trunk EMG data from dynamic and isokinetic lifts at different velocities to estimate the effect of increases in lumbosacral velocity on spinal loading. Granata and Marras found that when subjects extended from a flexed position (45 degrees of lumbar flexion) increases in lumbosacral velocity resulted in increased activation levels and associated force production of the ES and abdominal musculature of the trunk. It was predicted that increased lumbar extension velocity not only increased compressive forces but also substantially increased anterior shear forces acting on the lumbar spine.

Although the relationship between increased rate of change of lumbar curvature and spinal loading seems apparent, there have been few studies that have investigated lumbosacral velocity measures during squat, mixed and stoop lifting. Using graphs of the L5/S1 angle in de Looze et al. (1993) it can be observed that subjects who were instructed to use a back (stoop) technique had a steeper slope and greater total range of motion during lifting and lowering than those using a leg (squat) lifting technique. Both of these observations would indicate that rate of change of L5/S1 motion was greater in a stoop lift when compared to a squat lift. These findings were supported by van Dieen et al. (1998) who plotted lumbosacral displacement of individual subjects using a stoop and squat lifting technique. This showed that during lifting the average rate of change in the lumbosacral angle over one second for the stoop lifter was twice that of a squat lifter.

Lifting postures may also display different lumbar spine velocity patterns throughout a lifting cycle. Individual subject data would indicate that when performing a squat lift there is minimal change in lumbar motion for greater than half of the lifting period, whereas change in lumbosacral angle occurs early during the stoop lift (van Dieen et al., 1998). In contrast, data from graphs in de Looze et al. (1993) showed that both the squat and stoop postures had very little change in lumbar spine motion during the first 25% of the lift and last 25% of the lowering cycle.

Repetitive lifting: trunk muscle activation

Lumbar spine extensor muscles

An understanding of the patterns of trunk muscle activation during repetitive lifting is important as it can provide insight into the active contribution of musculature to the control of lumbar spine motion. One of the most commonly investigated trunk muscles during lifting and lowering activities is the LES muscle group. The level of LES activation during lifting and lowering may be dependent on the lifting posture adopted, as both the degree of lumbar flexion and velocity at which the lumbar spine moves may influence the magnitude of LES muscle activity (Dolan and Adams, 1993b; Raschke and Chaffin, 1996; A. L. Roy, Keller, and Colloca, 2003). During the initial stage of a lift, the magnitude of LES muscle activation seems to be influenced by differences in spinal posture. When lifting an object of moderate mass (11-16 kg) activation levels of the LES at the L3-4 level have been reported to be lower for a flexed lumbar posture compared to lifts where the lumbar spine is considerably more lordotic (Delitto and Rose, 1992; Hart et al., 1987; Potvin et al., 1991; Vakos, Nitz, Threlkeld, Shapiro, and Horn, 1994).

The lower levels of LES muscle activation in a less lordotic lumbar posture have been attributed to the flexed lumbar spine altering lumbar LES muscle length (Macintosh et al., 1993), the change in fibre orientation decreasing the mechanical advantage of LES (Macintosh et al., 1993; McGill et al., 2000), as well as the recruitment of passive tissues of the lumbar spine (Delitto and Rose, 1992; Holmes, Damaser, and Lehman, 1992). For example, when compared to a lordotic posture, a fully flexed lumbar posture results in the LES muscle fascicles being elongated by an average of 39% (Macintosh et al., 1993). Due to the viscoelastic properties of the stretched muscle this elongation generates passive tension which can compensate for the loss of active tension (Macintosh et al., 1993). The recruitment of passive posterior structures of the spine in a flexed posture has the advantage of lessening the required activation levels of the ES without significantly altering or potentially reducing compressive forces (Potvin et al., 1991). However, in a flexed spine, the altered orientation of LES to the lumbar spine (Macintosh et al., 1993; McGill et al., 2000) dramatically reduces the ability of the LES muscles to resist anterior shear forces (McGill et al., 2000).

Although LES activity at the L3-4 level differs between lifting postures at the start of the lift, few studies have investigated LES EMG activity in relation to lumbar motion throughout the remainder of the lifting and lowering cycle. When executing a squat lift in a lordotic posture the initial peak of LES EMG is followed by a progressive decrease in activation until the lifting phase is completed (Delitto and Rose, 1992; Holmes et al., 1992). LES muscle activation patterns through the lifting cycle tend to differ when the lumbar spine is more flexed. Delitto and Rose (1992) found that when subjects were asked to posteriorly tilt their pelvis LES muscle activity did not differ between the first and second half of the lift. Abdoli, Agnew, and Stevenson,

(2006) and Vakos et al. (1994) showed similar findings in graphs illustrating a steeper decline in LES activation levels throughout the lifting phase in the lordotic lift compared to a kyphotic (stoop) lift. Others have reported that when subjects use a kyphotic posture or a "back lift" (stoop lift) there is minimal LES activity at the start of the lifting phase, with activity levels peaking in the middle lifting phase and diminishing again towards the end of the lifting phase (de Looze et al., 1993; Holmes et al., 1992; Toussaint et al., 1995). The differential findings from these studies may be related to the instruction given to subjects, as some instructions have focused on the degree of pelvic tilt (Delitto and Rose, 1992), whereas others have focused on the degree of knee flexion (Dolan, Earley et al., 1994).

Few studies have examined LES EMG during the lowering phase. It has been generally found that irrespective of technique (stoop vs squat) there is less activation of the LES during lowering compared to lifting (de Looze et al., 1993; Kumar and Davis, 1983). This decrease in levels of LES muscle activation has been attributed to the different types of muscle action required when lifting and lowering and the greater inertial force that needs to be overcome during lifting (de Looze et al., 1993; Tesch, Dudley, Duvoisin, Hather, and Harris, 1990). It has been commonly found that for a given tension lower EMG activity is recorded during eccentric (as would be expected during lowering) compared to concentric (as would be expected during lifting) muscle actions (Tesch et al., 1990). Studies examining LES muscle activation patterns throughout the lowering phase when using different lifting postures have differing results. Delitto and Rose (1992) found no difference in LES activation levels during lowering when subjects were asked to use squat lifting techniques with a posteriorly or anteriorly tilted pelvis. de Looze et al. (1993) illustrated that when lowering using a "leg" (squat) technique, levels of LES activity recorded during the initiation were half the magnitude of that produced at the end of lowering. When using a “back” (stoop) lift LES activity peaked in the middle of the lowering phase, subsequently decreasing in the final quarter of the lowering phase.

The role of upper erector spinae (UES) during lifting and lowering is less well defined than that of LES. Knowledge of UES muscle activation patterns during repetitive lifting using different postures is important, as the anatomy of UES is considerably different from that of the LES (Macintosh and Bogduk, 1987, 1991). Despite such differences, few studies have compared the activation levels of UES and LES when using different lifting postures. Potvin et al. (1991) found that at the instant of peak extensor moment during a lifting task UES activity was greater in the stoop lift when compared to the squat lift, whereas LES muscle activity in the stoop lift was significantly less than that produced when performing an instructed squat lift. Data from graphs in Abdoli et al. (2006) indicated that when subjects were instructed to use a stoop and squat lifting technique UES displayed similar activation patterns for both postures, peaking near the start of the lift and decreasing in a relatively linear fashion throughout the remainder of the lifting cycle. Toussaint et al. (1995) analysed paced lowering and lifting using an instructed stoop lifting technique. It was found that whilst LES displayed near electrical silence in the period from the end of lowering to the initiation of the lift, UES did not display such a pattern, and tended to increase activation when levels of LES activation were low. Toussaint et al. suggested that the lever arm length and orientation of UES is less affected by lumbar flexion than LES and that even when the spine is flexed the UES are still at a mechanical advantage to resist bending moments.

Abdominal muscles

Although the ES muscle contribution to repetitive manual handling has been extensively investigated, less is known about the role of abdominal muscles during lifting and lowering activities. The magnitude of abdominal muscle activation during different phases of lifting and lowering is likely to be important as increased abdominal muscle co-activation has the potential to increase lumbar spine stability (Cholewicki, Panjabi, and Khachatryan, 1997; Gardner-Morse and Stokes, 1998; Granata and Marras, 2000). However, this is often at the cost of increased loading on the lumbar spine (Granata and Marras, 2000; Granata et al., 2004).

Irrespective of posture, the amount of abdominal activity during lifting of a moderate mass is relatively low compared to ES muscle activity (Delitto and Rose, 1992; Kingma et al., 2006; Potvin et al., 1991). Potvin et al. (1991) reported levels of internal oblique (IO) and external oblique (EO) activity of between 14%-15% of maximal voluntary contraction (MVC) when lifting a moderate mass with a stoop lift which was reduced by approximately 50% for the squat lift. Delitto and Rose (1992) reported that during lifting the abdominal muscle activity ranged between 10% and 15% of MVC, but found no difference in abdominal activity across lifts involving either anterior and posterior tilting of the pelvis.

Delitto and Rose (1992) also explored the patterns of abdominal muscle activity during lifting and found that oblique abdominal muscle activity was higher during the first half of a lift as opposed to the second half of the lift. The opposite was apparent during lowering, where greater abdominal activity was recorded in the second half of the lowering phase. Posture (anterior and posterior pelvic tilt) was not found to influence these activation patterns. Vakos et al. (1994) reported similar findings to Delitto and Rose (1992) and identified no difference in rectus abdominis (RAB) and oblique abdominal activity between kyphotic and lordotic lifting styles, but reported greater activity of the abdominal muscles during the beginning of the lift compared to the later stages of the lift. However, it should be noted that peak abdominal muscle activation did not exceed 3% of MVC throughout the lifting cycle (Vakos et al., 1994). Others (Cappozzo, Felici, Figura, and Gazzani, 1985; de Looze, Groen, Horemans, Kingma, and van Dieen, 1999; McGill and Norman, 1986) who have not differentiated between lifting posture have reported higher levels of abdominal muscle activity near the end of the lifting phase (approaching an upright position) when compared to the beginning of the lift. It has been argued that abdominal activation is more important in the upright position as lumbar spine stability is considerably lower compared to more forward flexed positions where additional abdominal muscle activation may further increase the higher compressive loads (Cholewicki et al., 1997; Granata and Marras, 2000; Granata, Rogers, and Moorhouse, 2005).

Repetitive lifting: effects of fatigue on lumbar spine kinematics

Another important aspect during the performance of repetitive lifting tasks is fatigue. A number of studies have reported that subjects performing repetitive lifting tend to change to a more stooped posture and increase peak lumbar flexion over time (Dolan and Adams, 1998; Sparto, Parnianpour et al., 1997a). However, these studies fail to differentiate between lifting postures. Sparto, Parnianpour et al. (1997a) had subjects lift 25% of their maximal lifting capacity (mean=16.7 kilograms) at a maximal lifting rate (39 lifts per minute) from mid-shank to waist level until they could no longer continue to lift or until their HR was greater than 180 beats per minute. Average endurance time was 135 seconds and 9 of the 16 subjects were required to terminate the task because HR exceeded 180 beats per minute. Sparto, Parnianpour et al. found that although total lumbar range of motion did not change over time there was reduction in lumbar extension and an increase in peak lumbar flexion from 35 to 38 degrees. Sparto, Parnianpour et al. suggested that the increase in lumbar flexion at the end of the lifting task was representative of a transformation to the less physiologically demanding stoop lift. Dolan and Adams (1998) found that when subjects were required to perform self-paced lifting and lowering of a weight lifters disc over 100 continuous cycles using a freestyle technique, peak lumbar flexion increased from approximately 81% of maximal flexion at the beginning of the session to 89% of maximal flexion at the end of the session. It was predicted that the increase in lumbar flexion over time increased the bending moment on the lumbar spine, although its extent was still within the elastic limits of the osteoligamentous spine (Dolan and Adams, 1998).

In contrast, other researchers have found little change in lifting posture at the end of a repetitive lifting task (Bonato et al., 2003) or decreases in lumbar flexion (Marras and Granata, 1997). For example, Marras and Granata (1997) recruited 10 experienced material handlers who transferred 23 kg boxes from pallets at a lifting rate of two lifts per minute. Marras and Granata found a 15% decrease in peak sagittal trunk angle over the five hour period of lifting and lowering. The authors proposed that this adaptation was to reduce the trunk mass (and therefore trunk centre of gravity) that was flexed forward of the lower lumbar spine. Having a more upright trunk and the centre gravity closer to the lumbar spine would in turn reduce the bending moment on the lower lumbar spine. It was also suggested that the reduction in trunk angle when fatigued may have served to maximise the ES length-tension relationship so the optimal ES muscle force potential was reached.

Studies that have investigated the effects of fatigue on lumbosacral velocity have not differentiated between lifting postures. The findings from these studies appear equivocal. van Dieen et al. (1998) found that when using a self-selected lifting technique lumbosacral velocity near the initiation of the lift decreased when fatigued, whereas lumbosacral velocity near the end of the lift increased when compared to a non-fatigued state. Marras and Granata (1997) found that peak trunk extension velocities decreased during a series of standardised lifting tests throughout a five hour manual handling task. Others (Sparto, Parnianpour et al., 1997a) have found little change in lumbosacral velocity measures during short duration high frequency lifting and lowering to fatigue. It has also been shown that in activities that require large ranges of lumbar flexion and extension (as would be seen when stoop lifting) there is a slowing of lumbosacral velocity and decreases in range of motion of the lumbar spine in the sagittal plane over time (Parnianpour, Nordin, Kahanovitz, and Frankel, 1988). This may indicate that the lumbosacral velocity of those postures that initially have large range and rate of change in lumbosacral motion may be affected by fatigue to a greater extent than lifting postures that have a reduced lumbosacral excursion and velocity.

The contrary nature of the findings concerning the influence of lifting-related fatigue on lumbosacral kinematics may reflect the variety of postures adopted by individuals using freestyle (self-selected) lifting techniques. When examining the effects of fatigue on lumbosacral kinematics during repetitive freestyle lifting it is important to establish the type of posture initially adopted by the handler, and in turn the degree of lumbar flexion and speed of lumbosacral motion occurring for a given posture. This is important as lifting-induced fatigue may affect lifting postures that have differing lumbosacral kinematics in different ways.

Repetitive lifting: Muscle EMG changes with fatigue

A possible explanation for the observed changes in lumbosacral kinematic changes during repetitive lifting is trunk muscle fatigue (van Dieen et al., 1998). Subjective sensations of localised low back muscle fatigue have been considered a primary factor contributing to the termination of incremental lifting tasks (Kell and Bhambhani, 2003). It would also seem that during repetitive manual handling stoop lifting induces higher local back RPE and pain ratings than squat lifting (Hagen and Harms-Ringdahl, 1994; Rabinowitz et al., 1998).

Subjective perception of local back muscle fatigue seems to be supported by local EMG measures of ES muscle fatigue. Dolan and Adams (1998) assessed ES muscle fatigue by analysing changes in median frequency (MDF) of the EMG power spectrum during sub-maximal pulls from a fixed handle bar before and immediately after 100 lifts using a self-selected technique. Dolan and Adams found a decrease in MDF of the LES indicating muscle fatigue, but no changes were noted in MDF of UES following the lifting intervention. Using a similar static testing apparatus Bonato et al. (2003) reported that the right ES muscle at L2 level showed a significant decrease in MDF following 4.5 minutes of lifting of a 13 kg box at a lifting frequency of 12 lifts per minute. However, at the end of the lifting task there were no significant decreases in MDF at UES and LES (at the L5 level) during maximal static lifts (Bonato et al., 2003). Potvin and Norman (1993) had subjects perform a 20 minute repetitive manual handling task, lifting and lowering 20 kg at a lifting rate of eight lifts per minute. In this study MDF measures were recorded during static contractions before and throughout the lifting task. Similar to the findings of Bonato et al. (2003), Potvin and Norman (1993) found that the LES showed evidence of a decline in MDF over the 20 minute session of lifting whereas there was no significant change in the MDF of UES.

Unfortunately these studies did not differentiate between lifting postures and it would seem reasonable to assume that the lumbar posture adopted may influence ES fatigue. Interestingly Potvin and Norman (1993) commented that two subjects who used substantial flexion of the spine during lifting exhibited greater evidence of ES fatigue than two subjects who used a technique that placed greater demand on the legs and involved limited lumbar spine flexion.

Trunk muscle amplitude measures have also been used to study fatigue during lifting and lowering. Again, these studies have failed to differentiate between lifting postures. Petrofsky and Lind (1978b) reported increases in levels of ES muscle activation throughout an hour of repetitive lifting, and this increase in ES muscle activation was more pronounced when lifting at a higher percentage of maximal aerobic capacity. Marras and Granata (1997) found that during pallet loading and unloading subjects performing a series of five standardised lifts from knee height to the upright position increased the level of ES EMG per unit of trunk moment over time. Though no frequency measures of EMG were taken, they suggested that the increase in EMG to moment ratios reflected ES fatigue and less efficiency of activation in relation to moment. Potvin and Norman (1993) found that despite reductions in maximal extensor moment, amplitude of the LES and UES activity increased in static contractions performed in the later stages of repetitive lifting.

The effects of fatigue on abdominal muscles have been less extensively investigated than that of ES. During fatiguing isometric trunk extension manoeuvres an increase in the activity of the IO (Reeves, Cholewicki, Milner, and Lee, 2008; Sparto, Parnianpour, Marras et al., 1997) and EO (Reeves et al., 2008) has been observed. Based on previous findings that showed increased abdominal muscle co-contraction improved spinal stability (Cholewicki et al., 1997; Gardner-Morse and Stokes, 1998; Granata and Marras, 2000) it has been suggested that the increased abdominal co-contraction observed during sustained static extension is a mechanism to stiffen the spine and aid spinal stability when the ES muscles are fatigued (Reeves et al., 2008).

Abdominal co-activation may be an important adaptation during more dynamic ES muscle fatigue. Sparto and Parnianpour (1998) examined the effects of fatigue using a repetitive isokinetic trunk extension task where the pelvis was restrained and subjects repeatedly extended their trunk at 35% and 70% of their dynamic trunk MVC. Subjects continued the repeated trunk extensions until there was a 35% reduction in dynamic maximal torque, or subjects felt that they could no longer continue the task, or 30 minutes had elapsed. Using an EMG assisted model it was predicted that when performing repeated lumbar extensions at a rate of up to 10 times per minute over a range of 35 degrees from the neutral, there was increased force (activation) generated by EO and little change in predicted force production of IO or RAB (Sparto and Parnianpour, 1998). Marras and Granata (1997) observed EMG activity of abdominal muscles whilst manual workers performed repetitive moving of pallets. Marras and Granata found that IO activation tended to increase during standard static lift exertions measured throughout the task. It was suggested that the increased activation of the IO was a compensatory mechanism whereby some of the lifting moment was transferred from the fatigued ES muscle to the IO.

In summary, there is lack of evidence on how the self-selected freestyle lifting posture (squat, mixed or stoop) adopted during repetitive lifting influence lumbosacral kinematics and associated muscle activity. Furthermore, information is lacking about the effects of lifting-induced fatigue on lumbosacral kinematics and associated trunk muscle activity in individuals who adopt a variety of self-selected lifting postures.

FUNCTIONAL ANATOMY OF THE TRUNK MUSCLES: IMPLICATIONS FOR POSTURAL CONTROL DURING SUDDEN LOADING AND REPETITIVE LIFTING

The combining of kinematic and EMG data provides a medium for assessing muscle activity in relation to motion during sudden loading, and repetitive lifting (Dolan and Adams, 1993b; Granata et al., 2004; Pedersen et al., 2004; Potvin et al., 1991). However, knowledge of the functional anatomy of musculature of the lumbar spine is necessary in order to interpret the relationships between EMG and trunk motion findings. During anterior sudden loading and lifting in the sagittal plane the musculature of the trunk must activate to counter the forces and moments imposed on the spine. Without the trunk musculature, compression forces as low as 90 N can buckle the spine (Crisco, Panjabi, Yamamoto, and Oxland, 1992). Furthermore, during lifting compression forces may be within acceptable limits, yet, without adequate posterior trunk muscle activation in flexed postures, anterior shear may exceed forces considered to be safe (McGill, 1997a).

The erector spinae muscles

The major trunk muscle group responsible for resisting and controlling the bending moment and anterior shear forces imparted during sudden anterior loading and lifting is the ES muscle group (Macintosh and Bogduk, 1986; McGill, Patt, and Norman, 1988). Though there is some debate as to which posterior trunk muscles constitute the ES muscle group (Macintosh and Bogduk, 1986), the muscles within this group include the longissimus thoracis, iliocostalis lumborum (Macintosh and Bogduk, 1986), sacrospinalis, and multifidus muscles (McGill and Norman, 1987a). The ES muscle group have a large percentage of Type I fibres particularly at the thoracic level (Mannion et al., 1997; Sirca and Kostevc, 1985; Thorstensson and Carlson, 1987) making them quite suited to repetitive lumbar activities that require high levels of muscular endurance. However, because the ES type I fibres have a larger diameter than the type II fibres (Mannion et al., 1997; Sirca and Kostevc, 1985) they have a greater potential per fibre for force production than would be typically found in type I fibres of the peripheral limb muscles.

The role of the ES muscles in resisting forces during loading events may differ depending on their attachments to the spine and pelvis. For example, through detailed examination Macintosh and Bogduk (1986) have suggested that the upper and lower portions of longissimus thoracis and iliocostalis lumborum are quite anatomically separate, and therefore have different functional roles in relation to the control of lumbar spine motion. Bergmark (1989) furthered this concept and divided the ES muscles into those that connect the thoracic cage to the pelvis (global) and those that act at a segmental level (local). The global ES or upper erector spinae (UES) include the thoracic fibres of iliocostalis lumborum and longissimus thoracis. Thoracic fibres of longissimus and Iliocostalis lumborum arise from the ribs and transverse processes of T2 to T12 (Macintosh and Bogduk, 1987). All of the thoracic fibres of iliocostalis lumborum and the lower fibres of longissimus thoracis span the entire lumbar spine forming the ES aponeurosis which moves freely over the lumbar ES muscles (Macintosh and Bogduk, 1994), connecting to the sacrum and posterior superior iliac spine (Macintosh and Bogduk, 1987). This allows the UES to have a 'bowstring' effect on the lumbar spine to maintain or accentuate the lumbar lordosis. The thoracic portions of longissimus and iliocostalis lumborum have the greatest moment arm of all the lumbar extensors (Daggfeldt and Thorstensson, 2003) which allows them to generate a large extensor moment that resists bending forces produced by forward flexion of the trunk (Macintosh and Bogduk, 1987) with minimal compressive loading on the spine (Callaghan and McGill, 1995). It has been predicted that the thoracic fibres contribute to between 70% to 90% of the total extensor moment to the upper lumbar spine and up to 50% of the total extensor moment exerted on L4-L5 (Bogduk, Macintosh, and Pearcy, 1992; Daggfeldt and Thorstensson, 2003). However, the global ES have little capacity to resist anterior shear forces (Callaghan and McGill, 1995).

The local subgroup of ES muscles are those muscles whose fascicles originate and insert on the vertebrae of the lumbar spine (Bergmark, 1989). This group primarily includes the poly segmental muscles – the lumbar components of longissimus and iliocostalis, and multifidus (Bogduk and Twomey, 1987), and are often termed the lower erector spinae (LES). The lumbar fibres of iliocostalis lumborum (iliocostalis lumborum pars lumborum) and longissimus thoracis (longissimus thoracis pars lumborum) have the potential to act directly on the lumbar vertebrae (Macintosh and Bogduk, 1987). These fibres arise from the lumbar accessory processes and the L1-L4 transverse processes and insert independently of the ES aponeurosis into the ilium (Macintosh and Bogduk, 1987). The lumbar fibres of iliocostalis lumborum and longissimus thoracis are more angulated relative to the vertebral column than multifidus, with a substantial increase in obliquity in the L4-L5 region (Macintosh and Bogduk, 1991). Therefore, when contracted bilaterally during a symmetrical activity, such as lifting, the lumbar fibres of iliocostalis lumborum and longissimus thoracis have the potential to produce large posterior translation and shear forces at the lower lumbar spine (Macintosh and Bogduk, 1991). The lumbar fibres of iliocostalis lumborum and longissimus have a closer proximity to the spine and therefore have less ability to resist bending moment than UES (Callaghan and McGill, 1995), and because of their fascicle obliquity are less able to resist anterior sagittal rotation than multifidus (Macintosh and Bogduk, 1991).

Another key muscle included in the local subgroup of ES is multifidus. Multifidus consists of multiple overlapping layers of fibres (Bojadsen, Silva, Rodrigues, and Amadio, 2000) and is the largest and most medial muscle that spans the lumbosacral area (Macintosh, Valencia, Bogduk, and Munro, 1986). Each fascicle arises from a common tendon attached to the spinous process of individual lumbar vertebrae with fascicles spanning to attach to the mamillary process of the inferior vertebrae, the iliac crest and the sacrum (Macintosh and Bogduk, 1986). At each lumbar vertebral level the fascicles are innervated by the medial branch of the dorsal ramus of the inferior vertebra (Bogduk, Wilson, and Tynan, 1982; Jonsson, 1969; Macintosh et al., 1986). This segmental fascicle arrangement and innervation gives multifidus the potential to control motion of individual vertebra of the lumbar spine (Bogduk et al., 1982). Fascicles of multifidus arise from a common tendon and form a vertical vector that acts at approximately 90 degrees to the spinous process. This vector lies behind the axis of sagittal rotation giving multifidus a mechanical advantage to posteriorly sagittally rotate the vertebrae from which the fascicle originates (Macintosh and Bogduk, 1986). The activation of multifidus therefore has the ability to produce an anti-flexion (extension) moment needed to balance the anterior sagittal rotation generated from the contraction of the internal and external abdominal oblique muscles (Macintosh and Bogduk, 1986). Fascicles of multifidus are orientated at right angles to the vertebra of origin and therefore the primary action of multifidus is to exert a compressive force on the vertebra of origin (Macintosh and Bogduk, 1991). However, because there is minimal obliquity of multifidus fibre orientation in the sagittal plane (small horizontal vector), multifidus has limited ability to produce substantial amounts of posterior translation or shear (Macintosh and Bogduk, 1986).

The abdominal muscles

Although the ES muscles are the most active muscle group during both sudden anterior loading and lifting, there is also notable activation of the abdominal musculature (Carlson et al., 1981; Cresswell, 1993; Delitto and Rose, 1992; Henry et al., 1998; Potvin et al., 1991). The abdominal muscles consist of RAB, IO, EO and transversus abdominis. The combined action of the abdominal muscles is to apply a flexion moment on the lumbar spine (Dumas, Poulin, Roy, Gagnon, and Jovanovic, 1988; Gatton, Pearcy, and Pettet, 2001; McGill, 1991; Reid and Costigan, 1985). Therefore, these muscles have the potential to accentuate flexion and potentially de-stabilise the spine during lifting and sudden forward loading of the trunk. However, a number of researchers have suggested that activity of anterio-ventral muscles of the trunk serve to stabilise the spine (Bartelink, 1957; Cholewicki, Juluru, and McGill, 1999; Gracovetsky, Farfan, and Helleur, 1985).

Two main theories have been developed to explain the stabilising role of abdominal muscles during sudden loading and lifting and lowering. The first key theory to explain the activation of the abdominal musculature during sudden anterior loading of the lumbar spine and lifting activities is the intra-abdominal pressure (IAP) theory (Bartelink, 1957; Cresswell, 1993; Cresswell et al., 1994; Gracovetsky et al., 1985). Bartelink (1957) first introduced the IAP or "balloon" theory, proposing that the spine could be stiffened to resist flexion through intra-abdominal pressure acting upwards on the diaphragm. The mechanism by which IAP was increased was through hoop tension generated by activation of the abdominal muscles in combination with contraction of the diaphragm, reducing the volume of the abdominal cavity (Bartelink, 1957; Cresswell, 1993; Gracovetsky et al., 1985). Gracovetsky et al. (1985) proposed that the subsequent increase in IAP provided a decompressive effect on the lumbar spine.

McGill and Sharrott (1990) were one of the first groups of investigators to apply the intra-abdominal balloon theory to a simulated perturbation situation. They had subjects contract the abdominal muscles as fast as possible in standing whilst simultaneously measuring IAP. Correlations between IAP and each of the three abdominal muscles were found to be greater than 0.86. Based on these findings, McGill and Sharrott suggested that the early increase in IAP in response to abdominal muscle activation was a possible mechanism for stiffening the spine during sudden perturbations. Subsequently, Cresswell et al. (1994) developed an experimental paradigm to investigate the relationship between IAP and abdominal muscle activity during sudden unexpected loading. Subjects had their pelvis and lower limbs fixated, and an unexpected forward perturbation was elicited through a wire attached to a vest worn by the subject. Cresswell et al. observed that early activation of abdominal muscles coincided with an increase in IAP, and peak IAP occurred between 7-9 ms after that of abdominal EMG. From this observation Cresswell et al. suggested that there was a strong relationship between activation of the anterior-ventral muscles of the trunk and increased IAP, and that this increase in IAP increased spinal stiffness.

Whilst relationships between IAP and abdominal muscle activity have been established during sudden perturbation in the upright posture, the relationship between IAP and abdominal muscle activation when lifting in different postures is less clear. Simulated static lifting postures tend to yield relative high correlations between IAP and abdominal muscle activity. During static lumbar exercises (including trunk extension) high correlations (0.87-0.89) between IAP and EMG activity of the oblique abdominal muscles have been shown (Cholewicki, Ivancic, and Radebold, 2002). It was concluded that in a number of static exercise scenarios it was not possible to generate IAP without concurrent co-contraction of abdominal and ES muscles. In an earlier investigation Cholewicki et al. (1999) suggested that the interaction between abdominal muscle activity and IAP could potentially stiffen the spine without further increase in activation level of the extensor muscles of the trunk. Cholewicki et al. (1999) developed a model that predicted that if the abdominal muscles were contracted against the hydrostatic pressure exerted in the abdominal cavity then the net moments would approach zero. Based on this finding, it was predicted that spinal stability would increase without the need for any additional trunk extensor activity. Cholewicki et al. (2002) suggested that the IAP mechanism was advantageous during activities such as lifting as it would allow greater capacity for the trunk extensors to exert extensor forces while still maintaining spinal stability.

It would also seem that the activation of abdominal musculature and the associated unloading effect of IAP may be dependent on lumbar posture. A biomechanical model developed by Daggfeldt and Thorstensson (1997) predicted that contraction of muscles of the abdominal wall with transverse fibre orientation (transversus abdominis and IO) led to IAP pressure having an unloading effect on the spine, that was more optimal if the spine was slightly flexed. In a subsequent study, Daggfeldt and Thorstensson (2003) reported that during maximal isometric extension effort, IAP was significantly greater in effort performed in a flexed posture when compared to when the lumbar spine was more extended. Through modelling it was found that the addition of IAP into the model decreased compression forces at all spinal levels. However, the total decrease in compressive force was greater in the flexed postures. These findings were supported by Arjmand and Shirazi-Adl (2006) who found that when lifting a 180 N mass in a forward flexed posture moderate levels of abdominal muscle activity were produced without negating the effect of IAP. However, in the upright posture minimal levels of abdominal muscle activation were required to negate the beneficial effects of IAP (Arjmand and Shirazi-Adl, 2006).

Other researchers (Cholewicki et al., 2002; Gracovetsky et al., 1985; Kingma et al., 2006; McGill and Norman, 1987b) have questioned whether elevated IAP reduces extensor muscle requirements during lifting, as often the flexion moment generated through co-contraction of the abdominal muscles to raise IAP during lifting is large enough to offset or exceed the IAP extension moment. In addition, the relationship between abdominal muscle activation levels and IAP does not seem to transfer to more dynamic lifting tasks. For example, McGill and Sharrott (1990) investigated time histories of IAP and abdominal activity throughout a dynamic lifting cycle and found that only between 48% and 55% of IAP variance during dynamic lifting could be explained by abdominal muscle activity.

A second explanation that has been advanced for abdominal muscle activation during sudden anterior loading and lifting activities is that abdominal muscle activation can indirectly apply an "anti-flexion" moment to the lumbar spine via abdominal muscle attachments to the thoracolumbar fascia (Gracovetsky et al., 1985). The posterior layer of the thoracolumbar fascia consists of a series of overlapping triangles (Gracovetsky et al., 1985; Macintosh and Bogduk, 1987) and attaches to the spinous processes of the vertebral bodies of the mid to lower lumbar spine. Some of the abdominal muscles (in particular IO and transversus abdominis) attach to the lateral raphe of the thoracolumbar fascia and are able to apply a lateral pull or hoop tension to the thoracolumbar fascia. Due to the triangular shape of the posterior layer of the thoracolumbar fascia, a lateral pull created by bilateral contraction of the abdominal muscles can be converted to longitudinal tension in the midline. This tension can approximate the spinous processes of the lumbar spine, producing an anti-flexion or an extension moment acting on the lumbar spine (Gracovetsky et al., 1985; Tesh, Dunn, and Evans, 1987). However, the triangular arrangement of the thoracolumbar fascia is limited to the mid and lower lumbar spine and only allows the thoracolumbar fascia to apply an anti-flexion moment on L2 through to L5 spinous processes (Bogduk and Macintosh, 1984).

To assess the influence of the abdominal muscles on the thoracolumbar fascia, Tesh et al. (1987) simulated the hoop tension of the abdominal muscles in the absence of increased IAP by inflating a centrally located balloon in the abdominal cavity of a cadaver. The findings showed that inflation of the balloon resulted in "small" extension of the lumbar spine in the sagittal plane. Similar findings were reported by Macintosh and Bogduk (1987), who through anatomical observation of abdominal muscle attachments to the thoracolumbar fascia, calculated the extension moment acting on the lumbar spine to be less than six Nm. The role of IO in the anti-flexion mechanism has also been questioned as fibres of IO vary in their attachment to the thoracolumbar fascia (Bogduk and Macintosh, 1984), and in a majority of cadavers it is only the posterior fibres of IO that attach to the lateral raphe of the thoracolumbar fascia (Bogduk and Macintosh, 1984; Macintosh et al., 1987). In addition, Vleeming, Pool-Goudzwaard, Stoeckart, van Wingerden, and Snijders (1995) reported that there was no motion in the deep lamina of the thoracolumbar fascia in response to traction of IO. Vleeming et al. have, however, reported that traction of EO muscle had variable effect on the super layer of the thoracolumbar fascia in different cadaver preparations. The EO muscle group had not been thought to have role in the thoracolumbar fascia anti-flexion mechanism previously, as anatomical studies had not reported attachment of EO to the thoracolumbar fascia (Bogduk and Macintosh, 1984; Macintosh et al., 1987).

METHODOLOGICAL ISSUES

Electromyography

The most extensively used tool to measure muscle recruitment and activation levels during sudden loading and lifting is electromyography (EMG) (Cresswell, 1993; de Looze et al., 1993; Delitto and Rose, 1992; McGill, 1997a; Pedersen et al., 2004). Since the first development of metal surface electrodes by Hans Piper (1912) EMG measurement techniques have advanced considerably. However, the electrophysiological principles for recording EMG remain the same. A combination of action potentials from the numerous muscle fibres innervated by a single motor neuron summate to produce electrical activity that represents the motor unit action potential. It is the change in electrical activity of the motor unit action potentials of the muscle fibres underlying a single or number of electrodes that can be measured, through EMG recording techniques (Basmajian and De Luca, 1985; Hagg, 1992; Kamen and Caldwell, 1996; Soderberg and Knutson, 2000).

The two primary methods used for the collection of muscle EMG during sudden loading and lifting included fine wire and surface EMG (sEMG). The most direct method of measuring physioelectrical activity of a muscle is the fine wire technique. The use of fine wire or needle EMG requires the insertion of a needle, which guides a wire into the muscle of interest. Wire electrodes have the advantage of being relatively easily manipulated to attain a clear signal of single motor unit activity and are sensitive to muscle activity during the production of low force muscle contraction (Basmajian and De Luca, 1985). In sudden loading and lifting studies fine wire electrodes have been primarily used for the measurement of the activity of the deep musculature of the trunk, such as transversus abdominis and quadratus lumborum (Cresswell, 1993; Cresswell et al., 1994; Hodges and Richardson, 1997; McGill, Juker, and Kropf, 1996). Despite these advantages fine wire EMG has the disadvantages of being invasive and requires ultrasonic guidance to ensure the wire is inserted into the correct muscle (Hodges and Richardson, 1997; McGill et al., 1996). There is also an associated risk of infection and because wire electrodes only measure the activity of the closest muscle fibres, they may not provide a true representation of the electrophysiological activity of the whole muscle (Disselhorst-Klug, Bahm, Ramaekers, Trachterna, and Rau, 2000; Disselhorst-Klug, Rau, Schmeer, and Silny, 1999).

The most common non-invasive technique used for recording electrical activity of superficial trunk muscles during sudden loading and lifting is sEMG. sEMG recording techniques are often chosen in preference to wire EMG because they are relatively painless, easy to administer and have good repeatability when measuring over long time periods (Kamen and Caldwell, 1996). sEMG recordings can vary from the simplest form of the monopolar electrode to linear electrode arrays capable of detecting individual motor unit activity (Enoka, 1994; Farina, Gazzoni, and Merletti, 2003; Merletti, Rainoldi, and Farina, 2001). Traditionally sEMG recording techniques have been used to provide global time domain information such as muscle onset latency, phasing of muscle groups (Disselhorst-Klug et al., 2000) and sEMG amplitude measures that provide an estimation of net motor unit activity (Farina, Merletti, and Enoka, 2004). More recently the development of sEMG frequency domain analysis has allowed an insight into localised trunk muscle fatigue during and after repetitive lifting (Bonato et al., 2003; Dolan and Adams, 1998).

Electrode and collection system influences on the sEMG signal

Before the sEMG signal can be interpreted there are a number of methodological concerns associated with the acquisition and analysis of the sEMG signal that must be understood. The electrical signal produced by a contracting muscle is of extremely low voltage and has a large range of frequencies (Kamen and Caldwell, 1996). These two characteristics make it difficult to distinguish signal and the background electrical activity (noise) arising from surrounding electrical appliances, including recording equipment (De Luca, 1997). Other sources of noise include movement artefacts and electrical signals generated from the heart (De Luca, 1997; Skotte et al., 2004). In order to maximise the signal to noise ratio several factors with respect to the collection of sEMG signals should be taken into consideration. These factors include the characteristics of the recording electrode and electrode placement in relation to muscle morphology.

Recording electrode characteristics

One factor to consider with respect to noise elimination is the electrode-skin interface. Passive surface electrodes do not have high electrical input resistance, and are therefore sensitive to skin resistance (Türker, 1993). Skin resistance can be reduced by shaving the skin surface followed by rubbing the surface of the skin with abrasive gel and cleaning with alcohol (Hermens, Freriks, Disselhorst-Klug, and Rau, 2000; Soderberg and Cook, 1984; Sparto, Parnianpour, Reinsel, and Simon, 1997c; Türker, 1993). An input resistance at the skin surface of no more than 10 K ohms is considered acceptable (Hermens et al., 2000; Mannion and Dolan, 1996). Skin preparation is less important with active differential electrodes as they have high input resistance (Türker, 1993). Active differential electrodes also have circuitry that is able to differentiate the sEMG signal by subtracting the signal of one recording site from an adjacent site. The differential technique eliminates any noise that is common to both recording sites and the resultant difference (differential signal) is then pre-amplified (De Luca, 1997). A differential amplifier which accurately rejects noise common to both electrodes has a high Common Mode Rejection Ratio (CMRR). Therefore a pre-amplified differential electrode with a CMRR of 32,000 or greater than 80 dB has been considered sufficient to eliminate unwanted electrical noise from the sEMG signal (De Luca, 1997; Ng, Richardson, Kippers, Parnianpour, and Bui, 1996).

Electrode configuration also has an effect on the quality of the sEMG signal. There has been some debate as to the ideal distance between recording electrodes. Larger inter-electrode distances (20 mm vs 12 mm) are associated with greater inter-test day reliability (Koh and Grabiner, 1993) and less variability than smaller inter-electrode distances when assessing ES muscle amplitude and fatigue parameters (Farina et al., 2003). However, with increasing inter-electrode distance there is more likelihood of measuring activity from the muscles adjacent and deep to that being measured (De Luca, 1997). Bipolar electrodes with smaller inter-electrode distances (5 mm) are not recommended for the recording of ES sEMG (Farina et al., 2003) as electrodes with a small inter-electrode distance are susceptible to changes in electrode location (Farina et al., 2003). There is also the increased likelihood of electrical activity recorded in one interface being shunted to the other, increasing noise and eradicating high frequencies (De Luca, 1997).

Inter-electrode distance may also alter amplitude and the spectral parameters of the sEMG signal. For example, Elfving, Liljequist, Mattsson, and Nemeth (2002) reported that doubling inter-electrode distance from 2 to 4 cm decreased MDF measure of ES muscle group by 8%. Others have also reported that with increases in inter-electrode distance over the ES muscle group there is corresponding increase in the sEMG amplitude measures (Farina et al., 2003; Mannion and Dolan, 1996) and a decrease in MDF and mean power frequency (Rosenburg and Seidel, 1989; Zedka, Kumar, and Narayan, 1997). However, Mannion and Dolan (1996) reported that MDF attained from sEMG signals recorded from the ES muscles in the thoracic and mid-lumbar regions were similar for electrodes with inter-electrode distances of 25, 30 and 35 mm. Although there has been much debate as to the ideal inter-electrode distance it has been suggested that for optimal measurement of the sEMG signal, the inter-electrode distance should be 10-25mm when using bipolar measurement techniques (De Luca, 1993, 1997; Farina et al., 2004; Hermens et al., 2000; S. H. Roy, De Luca, and Casavant, 1989).

Filtering and sample rate

Even with differential pre-amplification and appropriate selection of electrode characteristics, noise can still be attained via external sources including the amplifier. Band-pass filtering and appropriate selection of sampling frequency are two procedures used to further improve the signal to noise to ratio. Typically the frequency content of the sEMG signal ranges from as low 5 Hz (De Luca, 1997; Dolan, Mannion, and Adams, 1995) and seldom exceeds 500 Hz (Clancy, Morin, and Merletti, 2002; Kamen and Caldwell, 1996). As most of the frequency spectrum of sEMG data is between this range, it has been suggested that an active electrode with a built in bandwidth filter set between 10-20 Hz and 500 Hz can be used to eliminate any low frequency movement artefacts and noise (De Luca, 1997; Farina et al., 2003;

Granata et al., 2001; S. H. Roy et al., 1989; Türker, 1993). More recently there has been debate as to the ideal low frequency cut off when collecting sEMG data. De Luca (1997) argued that a low frequency cut off of 20 Hz is necessary in order to improve sEMG signal stability and stationarity, and to eliminate motion artefacts. However, others have found that during isometric fatiguing contractions of ES muscles power spectral changes are observed in frequency bands between 5-30 Hz, and have used low frequency cut offs as low as 5 Hz when assessing muscle fatigue (Dolan et al., 1995).

The sampling rate at which the sEMG signal is collected is also important for maximising signal to noise ratio. The sampling theorem states that the sample rate should be at least twice the highest frequency component of the signal (Clancy et al., 2002). This is termed the Nyquist limit and prevents high frequency aliasing (Dolan et al., 1995). Given that the sEMG signal does not contain signals that exceed 500 Hz the sampling rate at which sEMG data is collected during sudden loading and lifting experiments is most often 1000 Hz or greater (Granata et al., 2001; Vakos et al., 1994).

<u>Electrode placement in relation to muscle morphology</u>

Though sophisticated recording equipment has been developed to eliminate noise from the sEMG signal, the most important factors that affect the quality of the sEMG signal include the orientation and location of the electrode relative to the underlying muscle. Given that an action potential conducts along the length of a muscle fibre, it has been suggested that electrode orientation should be placed parallel to the fibres (Kamen and Caldwell, 1996). De Luca (1997) reported that up to 50% of the amplitude of the signal may be lost if electrodes are placed perpendicular to the muscle fibres. Proximity of electrodes to the midline will also influence sEMG spectral parameters. Sparto, Parnianpour et al. (1997c) investigated the placement of electrodes on the skin surfaces covering the medial and lateral aspects of the ES muscles at the levels of L1 and L3. Medially located electrodes produced higher initial MDF than those electrodes placed more laterally. Location of the electrode relative to spinal level will also influence spectral parameters of the sEMG signal collected from ES musculature. Electrodes placed over ES muscle at the lower lumbar spine produce a higher initial MDF than electrodes located more cephalid during static contractions (S. H. Roy et al., 1989; Sparto, Parnianpour et al., 1997c).

Another factor to take into consideration when orientating recording electrodes is the location of the electrode relative to the muscle's innervation zone. Propagation of action potentials from the innervation zone of the muscle is bi-directional (Kamen and Caldwell, 1996). Electrodes placed over the innervation zone tend to produce smaller signals with greater noise than electrodes placed to the side of the innervation zone (Merletti et al., 2001). Electrodes located near the innervation zone and musculotendinous junction also tend to produce higher frequencies than electrodes located over the muscle away from these areas (S. H. Roy, De Luca, and Schneider, 1986). To minimise this variability it has been suggested that electrode placement should be between the muscle's nearest innervation zone and the musculotendinous junction (De Luca, 1997; Farina et al., 2003). Whilst the innervation zones of large lower limb muscles have been established (Beck et al., 2008; Rainoldi, Melchiorri, and Caruso, 2004), the location of innervation zones of the ES muscles are difficult to detect via sEMG (Farina et al., 2003; Shiraishi, Masuda, Sadoyama, and Okada, 1995). Furthermore, the innervation zones of the LES muscles may be spread throughout different regions of the muscle (Farina et al., 2003). This may partly explain the higher MDF that tends to be recorded in the LES (S. H. Roy et al., 1989; Sparto, Parnianpour et al., 1997c).

Muscle fibre type characteristics may also affect sEMG amplitude and spectral measurements (Gerdle, Karlsson, Crenshaw, and Friden, 1997; Mannion, Dumas, Stevenson, and Cooper, 1998). ES muscle morphology is such that the slow twitch (type I) muscle fibres of the ES muscle group tend to be larger in size and occupy a relatively greater area than fast twitch (type II) fibres (Bagnall, Ford, McFadden, Greenhill, and Raso, 1984). This differs from more peripheral muscles where those muscles with larger fibre diameter are fast twitch and have faster conduction velocities and tend to produce higher frequencies in the power spectrum than the same muscle with a larger percentage of smaller diameter slow twitch fibres (Gerdle et al., 1997). Mannion et al. (1998) investigated the relationship between ES fibre morphology and MDF. The authors found that during sub-maximal static trunk holds those subjects with a larger relative area occupied by slow twitch fibres had less rapid decline in MDF and the longer contraction times than those subjects with reduced relative area of slow twitch fibres. Mannion et al. attributed the slower decline in MDF in subjects with larger percentages of slow twitch fibres to the reduced breakdown of ATP and less production of metabolites observed in those muscles consisting of predominantly type I muscle fibres, when compared to those muscles with higher percentages of fast twitch fibres (Katz, Sahlin, and Henriksson, 1986).

Irrespective of fibre morphology the spinal position and muscle fibre length can also affect sEMG amplitude and spectral parameters recorded at the lumbar spine (Mannion and Dolan, 1996; Nargol, Jones, Kelly, and Greenough, 1999). This is of particular importance when interpreting sEMG parameters during lifting and lowering where the lumbar spine may flex from a relatively neutral position in the upright standing position to near 100% of maximal flexion during the initiation of the lift (Dolan and Adams, 1998 183). Nargol et al. (1999) investigated the relationship between lumbar posture and sEMG parameters of the ES muscles. Nargol et al. found that the lordotic posture, where the ES muscle group was in a shortened position, resulted in significantly higher amplitude and MDF measures than a kyphotic posture, where the ES muscles were in a lengthened position. Rosenburg and Seidel (1989) also reported that during static trunk holds in the prone position with the pelvis supported that with greater trunk inclination the ES signal amplitude and mean power frequency were lower than with less trunk flexion. Mannion and Dolan (1996) compared MDF of the back extensor muscles when subjects held forces whilst flexed between 30% and 90% of maximal spinal flexion. They reported significantly higher MDF at 30% compared to 60% or 90% of maximum flexion. Mannion and Dolan attributed lower MDF in the elongated position to a reduction in conduction velocity of the stretched fibres. The influence of posture on sEMG measures from the ES muscles highlights the need to take into consideration lumbar posture during repetitive lifting. It also highlights the need to ensure that lumbar spine position is consistent when assessing fatigue measures such as MDF.

Even with appropriate electrode placement, nearly one fifth of the signal recorded by the surface electrode can be activity from adjacent or deeper musculature (De Luca and Merletti, 1988). This phenomenon is termed crosstalk and can be significantly reduced by using double differential techniques. The double differential technique requires the collection of EMG from three electrode sources placed in series. These three electrodes are able to produce two bipolar signals. The double differential signal eliminates voltage gradients with constant or similar slopes. In doing so the double differential recording technique is more selective (van Vugt and van Dijk, 2001) and produces sEMG with higher amplitude measures than the single differentiated techniques. It has been reported that double differentiation techniques can reduce crosstalk from adjacent muscles (Koh and Grabiner, 1993) by up to six fold when compared to single differentiation techniques (van Vugt and van Dijk, 2001).

EMG frequency domain analysis and muscle fatigue

A key area of interest when assessing repetitive lifting is the local fatigue of the ES musculature (Bonato, Boissy, Della Croce, and Roy, 2002; Bonato et al., 2003; Dolan and Adams, 1998), as it can compromise the ability of the ES muscles to resist anterior shear and bending moments, and increase the loading of the passive structures of the lumbar spine (Dolan and Adams, 1998). In the past, time domain analysis has highlighted inconsistencies in results with respect to muscle fatigue. This has drawn focus to spectral analysis of the frequency component of the sEMG signal and its relationship with other physical and physiological indicators of muscle fatigue (H. Christensen and Fuglsang-Frederiksen, 1988; Hermens, Baten, Van Bruggen, Rutten, and Zilvold, 1991; Lindstrom, Magnusson, and Petersen, 1970; Weytjens and van Steenberghe, 1984). As sEMG recorded during a sustained static contraction is considered to be stochastic with a Gaussian distribution (Hagg, 1992; Stulen and DeLuca, 1981) the mathematical process of Fast Fourier Transformation (FFT) (Cooley and Tukey, 1965), can be used to transform time domain sEMG epochs into a power spectrum (Basmajian and De Luca, 1985; Herzog, Guimaraes, and Zhang, 1994).

It has been observed by a number of researchers that during a sustained, fatiguing contraction, there is compression of the power spectrum. This compression of the power spectrum is due to a progressive loss of power in the high frequency region and an increase in power in the low frequency region of the power spectrum (Figure 2.1) (De Luca, 1984; Hagberg, 1981; Lindstrom et al., 1970; Merletti, Knaflitz, and De Luca, 1990).

A B

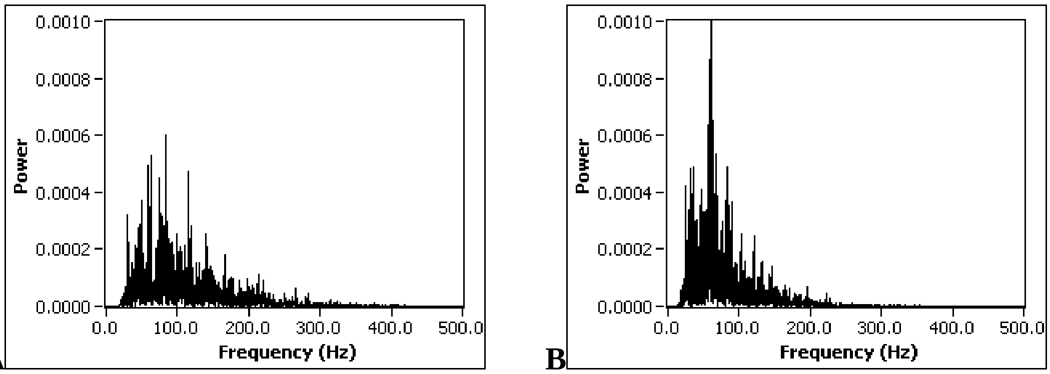


Figure 2.1. Power spectrum of the LES muscle group during static MVC: A) before and B) after repetitive lifting to exhaustion. Note the reduction in power in the high frequency component of the signal and the increase in the power of the low frequency component of the signal.

The shift of power to lower frequencies has been associated with peripheral (change in shape of the motor unit action potential) and central (motor unit recruitment) factors of fatigue (Bigland-Ritchie, Donovan, and Roussos, 1981; Broman, Bilotto, and De Luca, 1985; H. Christensen and Fuglsang-Frederiksen, 1988; Hermens et al., 1991; Stulen and DeLuca, 1981; Weytjens and van Steenberghe, 1984). The primary peripheral parameter that has been associated with the change in the power spectrum during fatigue is the shape of motor unit action potential, in particular the increase in time duration of the motor unit action potential (De Luca, 1997). During the 1970's Lindstrom et al. (1970) proposed that with fatigue there was a slowing of muscle fibre conduction velocity, which in turn increased the time duration of the motor unit action potential, shifting the power towards the lower frequency component of the power spectrum. The conduction velocity hypothesis devised by Lindstrom et al. was based on the finding that during a sustained muscle contraction there was limited blood flow into the muscle resulting in an accumulation of metabolic by-products such as lactic acid which reduced intracellular pH (Mortimer, Magnusson, and Petersen, 1970). With alteration of pH level, the depolarisation along the muscle fibre diminished, which was reflected by an increasing conduction time between measurement electrodes. To explore the conduction velocity hypothesis, Lindstrom et al. (1970) developed a mathematical model to investigate the relationship between conduction velocity and power spectrum changes during a sustained fatiguing contraction of the biceps brachii muscle. Their model predicted power spectral shifts to lower frequencies during sustained maximal contraction and that the spectral shifts to lower frequencies were proportional to decreases in conduction velocity (Lindstrom, Kadefors, and Petersen, 1977; Lindstrom et al., 1970).

Based on the findings of Lindstrom et al. (1970), Stulen and DeLuca (1981) developed a mathematical model to investigate whether a single index could accurately represent spectral power shift observed during muscle fatigue. The indices that were investigated included the mode (most commonly occurring frequency), mean (average frequency) and MDF (the frequency at which the power spectrum was divided into two regions of equal power). Stulen and DeLuca's findings indicated that both MDF and mean frequency were linearly proportional to conduction velocity. Their results showed that noise-related error was less for MDF compared to mean frequency, and concluded that MDF provided the most accurate estimate of conduction velocity.

Subsequent studies have substantiated the relationship between the reduction in conduction velocity observed during fatiguing muscle contractions and spectral compression towards lower frequencies (Arendt-Nielsen and Mills, 1985; Sadoyama and Miyano, 1981). Investigations using magnetic resonance imaging of interstitial muscle pH have reported a high correlation between muscle hydrogen levels and decreases in MDF (Vestergaard-Poulsen et al., 1992). However, the relative decrease in MDF is disproportional to the decrease in conduction velocity (Bigland-Ritchie et al., 1981; Broman et al., 1985; Dimitrova and Dimitrov, 2003). Furthermore, the relationship between conduction velocity and the power spectral changes of trunk musculature has not been clearly established as propagating potentials in the ES are difficult to detect using single differential sEMG techniques (Farina et al., 2003).

A number of investigators have questioned the conduction velocity theory and suggested that central factors related to motor unit recruitment correlate well with changes observed in the power spectrum (H. Christensen and Fuglsang-Frederiksen, 1988; Hermens et al., 1991; Weytjens and van Steenberghe, 1984). For example, Weytjens and van Steenberghe (1984) developed a model based on the synchronization of two motor unit action potential trains. The model predicted that motor unit synchronization led to an absolute increase of power at low frequencies and to a relative decrease of power at high frequencies. Subsequently, Hermens et al. (1991) simulated sEMG by modelling the summation of a series of motor unit action potentials. They reported that average firing frequency had little influence on spectral parameters, whereas spectral indices were altered considerably by motor synchronisation. De Luca, Roy, and Erim. (1993) argued that synchronisation would potentially have little effect on the amplitude and power spectrum, as they found evidence of synchronisation in only 8% of motor unit firings and the synchronisation occurred in sporadic bursts, with no more than two consecutive firings. In addition, it was also concluded that the influence of firing characteristics was limited to the lower part of the power spectrum (De Luca, 1984).

Reliability of spectral indices of back muscle fatigue

The MDF index has been used extensively as an index of ES muscle fatigue in normal (Kankaanpaa et al., 1998; Mannion et al., 1998; Ng, Richardson, and Jull, 1997; Sparto, Parnianpour et al., 1997c) and low back pain populations (Ng, Richardson, Parnianpour, and Kippers, 2002; S. H. Roy et al., 1989). To perform FFT effectively it is assumed that the sEMG signal is stochastic; thus the majority of studies examining the reliability of the MDF have used isometric contractions at varying percentages of MVC over a set time period (Dolan et al., 1995; Elfving, Nemeth, Arvidsson, and Lamontagne, 1999) or until there was an inability to maintain a constant position or force (Dedering, Roos af Hjelmsater, Elfving, Harms-Ringdahl, and Nemeth, 2000).

Elfving et al. (1999) tested the reliability of MDF during the performance of sustained back extension in upright sitting throughout a 45 second period at 80% MVC. Improved reliability was reported when measures were conducted on the same day as opposed to repeated over consecutive days. Variability, as expressed as a coefficient of variation, was less than 10%, with intraclass correlation coefficients varying between 0.4 and 0.7 (Elfving et al., 1999). Dedering et al. (2000) reported better reproducibility of MDF measures at L1 and L5 levels of the lumbar spine using the modified Sorenson test. They also reported intraclass correlation coefficients of 0.73 to 0.89 between MDF measures taken during the modified Sorenson test at the beginning and the end of a three-week period, with error measurements of between 7.2% and 10.6%. Ng and Richardson. (1996) also used a trunk holding test to evaluate within and between days reliability of MDF measures for multifidus and iliocostalis lumborum. They reported initial MDF value intraclass correlation coefficients of between 0.79 and 0.94. These researchers also examined the rate of change in MDF (M-slope) over time. They noted that there was a poor correlation in M-slope for iliocostalis lumborum, whereas intraclass correlation coefficients for the M-Slope of multifidus were between 0.78 and 0.82. Roy et al. (1989) evaluated reliability of MDF indices in standing using a device that restricted pelvic and lower limb motion to isolate back extensor contraction at 80% MVC. Although the subject group was small, numbering four in total, initial MDF and M-Slope measures had intraclass correlation coefficients that exceeded 0.94 with error values as low as 2%. Such high reliability may be attributed to the fact that the subjects were rested for a fifteen-minute period prior to re-testing and that electrodes were not removed.

Time domain measures of EMG

The measurement of sEMG signal amplitude has been used extensively to calculate baseline muscle activity and muscle onset latency during sudden loading of the trunk, and to assess phasing and magnitude of muscle activation over time during lifting and lowering activities (Cresswell et al., 1994; Delitto and Rose, 1992;

Granata et al., 2001; Lavender et al., 1989; Pedersen et al., 2004; Vakos et al., 1994). Amplitude measures include the average rectified value and root mean square (RMS) (Farina et al., 2003; Merletti et al., 1990). To determine the average rectified value the demeaned sEMG signal undergoes either half wave rectification, or more commonly full-wave rectification. Full wave rectification is preferred over half wave rectification, as half wave rectification deletes all of the negative values in the signal. Following rectification, the signal is often smoothed using a low pass filter (Kleissen, 1990). The smoothed signal can then be used for detection of muscle onset, or undergo integration to calculate the amount of muscle activity.

The second common method of quantifying the sEMG signal is to calculate the RMS of the signal. Similar to full wave rectification the RMS produces absolute values by initially squaring all data values. The final RMS value, however, equates to the square root of the average of the squared values. The RMS method has been reported to have a stronger mathematical basis than rectification with smoothing as it provides an instantaneous measure of the average power of the signal (De Luca, 1997; Soderberg and Cook, 1984).

Normalisation

Whether using rectification or RMS, in order to accurately compare sEMG amplitude between different muscle groups and across individuals the sEMG amplitude recorded during an activity should be referenced (normalised) to a standard value (Soderberg and Knutson, 2000). This is because factors such as skin preparation, electrode placement and the depth of subcutaneous tissue influence absolute amplitude measures of sEMG signal (Hemingway, Biedermann, and Inglis, 1995; Nordander et al., 2003). For example, there is an inverse relationship between local skin fold thickness and sEMG amplitude recordings from the ES (Hemingway et al., 1995; Mannion and Dolan, 1996; Nordander et al., 2003). In order to avoid such errors sEMG normalisation methods have been developed. The normalisation technique divides the amplitude value recorded during the selected activity by the value recorded during maximum effort where theoretically there is near to maximal recruitment of motor units (Soderberg and Knutson, 2000). This value is then multiplied by 100 and expressed as a percentage of MVC (Soderberg and Knutson, 2000). The process of normalisation is important during the assessment of manual handling activities such as lifting and lowering as it allows an appreciation of the level of specific muscle contribution to resisting loads on the spine (Cresswell et al., 1994; Delitto and Rose, 1992; Granata et al., 2001; Leinonen et al., 2002; Thomas et al., 1998).

Muscle EMG onset

Muscle EMG onset is another standard temporal measure that is commonly used during trunk perturbation experimentation to detect the time when a muscle activates in response to a sudden load (Cresswell et al., 1994; Henry et al., 1998; Pedersen et al., 2004; Wilder et al., 1996). There is little consensus in the literature regarding methods for selecting muscle EMG onset (Hodges and Bui, 1996). Several computer based techniques have been designed to automatically detect muscle EMG onset. These methods usually incorporate some form of low pass filtering of the rectified signal from as low as 6 Hz to as high as 1000 Hz (Hodges and Bui, 1996). Computer-based onset latency is usually detected as a change in baseline EMG activity of greater than one to three standard deviations from baseline (Di Fabio, 1987; Eriksson Crommert and Thorstensson, 2008), or a low percentage (for example, 5%) of the peak value recorded during activity (Bullock-Saxton, 1994). It has been reported that computer based onset determination methods compute different muscle onset times from those which are determined visually (Hodges and Bui, 1996). Despite the differences between computer and visual onset detection, Hodges and Bui (1996) found the error to be systematic, with a high correlation between computer and visual onset determination. Even though computer onset latency detection may be highly reliable error may still be present. In this regard, Eriksson, Crommert and Thorstensson (2008) reported that in some perturbation experiments up to 58% of computer detected onset and shut off latencies from abdominal and ES muscles had to be corrected manually via visual determination. These findings emphasise the need to manually check computer detected onset latencies.

Although computer onset detection protocols have been developed, a number of researchers assessing muscle responses to sudden loading prefer to use visual analysis for the determination of muscle onset during trunk perturbations (Pedersen et al., 2004; Skotte et al., 2004; Urquhart, Hodges, and Story, 2005). Visual determination of EMG onset with minimal filtering (10-1000 Hz) where the assessor is blind to movement onset has been shown to have high between day reliability (Hodges and Bui, 1996). The other advantage of visually detecting muscle EMG onset is that an experienced examiner can appreciate the amount and length of a burst of muscle activity relative to baseline activity and establish whether an artefact may be present (Soderberg and Knutson, 2000). Irrespective of computer or visual detection, high or variable levels of baseline EMG activity prior to the perturbation have been shown to reduce the ability to detect onset latencies. Skotte et al. (2004) had two experienced investigators visually determine EMG onset from sEMG signals collected from the ES muscles during a sudden trunk loading experiment. For each trial the investigator rated visual determination of onset as "easy" or "difficult". Large differences between observers were found in onset latency determination in trials classified as "difficult" with large background EMG.

Summary

In summary there are a number of factors that can influence the recording and quality of the sEMG signal. There are also a number of issues that are specific to the ES muscle group when interpreting amplitude and spectral measures from different portions of the ES muscle group during manual handling in different postures. These should be taken into consideration when interpreting time and frequency domain measures of sEMG during sudden loading and repetitive lifting.

Motion analysis

A number of methods have been used to assess trunk and lower limb motion during manual handling activities, including video fluoroscopy, electromagnetic tracking devices, and goniometry (Cholewicki and McGill, 1992; Dolan and Adams, 1998; Marras et al., 2006; Sparto, Parnianpour et al., 1997a). However, motion analysis systems used to track markers attached to a persons' body are one of the most common methods used to assess and record human movement during lifting and sudden loading experimentation (Arjmand and Shirazi-Adl, 2005; Bonato et al., 2003; Burgess-Limerick et al., 2001; Gill, Bennett, Savelsbergh, and van Dieen, 2007; van Dieen et al., 1998). Most two dimensional marker tracking systems are designed to track spherical markers that are brighter than the corresponding background, to enable the location of a marker in space and the attainment of its "X" and "Y" co-ordinates from a video signal (Chiari, Della Croce, Leardini, and Cappozzo, 2005; Pedotti and Ferrigno, 1995). These marker systems are designed to provide an accurate and reliable measure of skeletal motion. However, in the process of estimating bony movement there are two main sources of error that occur - those associated with skin movement artefacts, and instrumental error related to the collection and conversion of video data into kinematic variables (Cappozzo, Catani, Leardini, Benedetti, and Croce, 1996). An appreciation of the limitations of passive marker kinematic analysis has implications for the interpretation of movement patterns observed during sudden loading and lifting tasks.

Errors related to skin movement artefact

One of the key sources of error during the collection of kinematic data using passive marker systems is skin movement artefacts. Skin movement artefacts are due to the relative movement of the soft tissue underlying the marker and the anatomical point of interest. Because skin movement artefacts are unable to be filtered effectively the relative and absolute error associated with them must be taken into consideration when interpreting motion characteristics of individuals performing an activity. To assess error arising from skin movement artefacts, co-ordinates from passive surface markers have been compared to that of bone embedded markers. Bone embedded markers have included intra-cortical pins and external fixators (Cappozzo et al., 1996; Reinschmidt, van den Bogert, Nigg, Lundberg, and Murphy, 1997).

Due to the invasive nature of bone embedded markers, the few studies which have investigated the errors arising from skin movement artefact have often involved low subject numbers. Cappozzo et al. (1996) attached three markers to external fixation devices on the lower limb of seven injured patients. Subjects were required to perform a number of functional tasks, which included walking and cycling. Co-ordinates from markers attached to the external fixation device were then compared to those recorded from markers attached to the skin surface above bony prominences of the femur, tibia and fibula. Results showed that skin movement artefacts were variable between subjects with position errors of up to 40 mm for the greater trochanter. Errors of between 15-25 mm were recorded at the femoral condyle and head of the fibula. These errors increased with greater range of hip and knee motion (Cappozzo et al., 1996). Cappozzo et al. also reported that markers located over large muscle bellies produced much larger errors than those placed over bony prominences. It is important to note that findings from Cappozzo et al. may not be directly transferable to samples without pathology, as the subjects examined were recruited from an injured population, and the effect of the physical presence of an external fixation device on lower limb motion is unknown (Alexander and Andriacchi, 2001).

Some studies have inserted bone pins into subjects. Reinschmidt et al. (1997) used bone pins to assess the validity of skin surface markers on five healthy subjects during the performance of functional tasks. Each subject had intercortical Hoftman bone pins with triads of markers embedded in the posterio-lateral calcaneous and lateral tibial condyle. Movement artefacts were determined by the root mean square (RMS) difference between skin and bone markers. The authors reported average errors of 21% relative to the full knee range of motion in the sagittal plane. When observing tibio-calcaneal displacement, a shift in curves was noted between bone and skin surface marker estimates. When this shift was adjusted, skin and pin tibial rotation estimates determined by skin and in bone markers were similar. It was concluded that skin markers could reliably determine flexion/extension at the tibiofemoral joint. Benoit et al. (2006) used a similar approach to Reinschmidt et al. (1997) and compared intra-cortical pins (triads of reflective markers) with skin surface markers during walking and more rapid cutting movements. These authors reported absolute sagittal plane errors between skin and pin markers of less than 3 degrees during walking and up to 4.2 degrees during cutting movements. Within subject data was highly repeatable when using skin surface markers for both walking and cutting movements. In agreement with Reinschmidt et al. (1997), Benoit et al. (2006) also observed similar angular profiles of surface and cortical markers in the sagittal plane (knee flexion-extension) yet the direction of error was more consistent during cutting movements when compared to walking.

Instrumental error

Another source of error associated with data collected from passive marker systems is instrumental error. Instrumental error can occur during the acquisition of video footage and processing of co-ordinate data. Sampling frequency and shutter speed are two variables that can contribute to error during data collection (Winter, 1990). If the sampling frequency is too low then frame to frame stepping can create false frequencies, termed aliasing errors. To avoid these aliasing errors, the sample rate should be at least double the highest frequency present in the signal (Winter, 1990). For most kinematic analysis (for example, gait and lifting) the standard recording rate of 25 frames per second is considered sufficient to minimise analysing errors (Winter, 1982). During sudden loading activities more rapid movement can occur and sampling rates up to 156 Hz have been used (Carlson et al., 1981). Some digital cameras such as the DVL 9800 have the ability to capture video images at high capture rates. This is achieved by vertical cropping which divides each field into two vertical images. Although vertical cropping reduces the field image by half, it does enable data to be collected at 120 images per second.

Shutter speed is also important when recording high-speed movement. The shutter speed is the length of time that shutter allows light onto the charged-couple device in the camera. When recording faster movements, like those observed during sprinting, lower shutter speeds of 1/60 may result in image blurring, reducing the accuracy of digitising (Greaves, 1995). Winter (1990) suggests that to avoid blurring when recording fast movements a shutter speed of 1/250 is preferable. However, the disadvantage of a high shutter speed is that more external lighting is required to observe the image of interest (Winter, 1990).

During processing of data from video recording, errors arise from electronic devices, the spatial precision of the digitising system and human digitising error. Fortunately most human movement data is of low frequency and the noise arising from the processing of the video data is often random and of high frequency and therefore, can be filtered or smoothed (Chiari et al., 2005; Winter, 1990). Smoothing involves the selection of an appropriate cut-off frequency that will eliminate unwanted noise whilst retaining as much of the original signal as possible. The cut-off frequency is the key frequency component of the raw signal frequency, above which unwanted signals will be excluded. The cut-off frequency can be estimated through harmonic or spectral analysis of the raw data. Harmonic analysis provides a plot of the power of each frequency in the raw signal. From this plot it is possible to estimate the frequency range in which the majority of the power of the signal is contained

(Winter, 1982). Frequencies above this point can then be attenuated without eliminating much of the task specific data. As lifting and lowering is a relatively slow and rhythmical motion, most investigators use a low cut off frequency equal to or less than 6 Hz (van Dieen et al., 1998). For more rapid movement tasks such as those observed during sudden perturbation where there is rapid change in joint angle a higher frequency cut-off is required. Fourier analysis indicates that a cut-off frequency of 12 Hz is appropriate to retain 95% of the original signal during these rapid movements (Benoit et al., 2006).

Measurement of lumbar posture

Unlike peripheral joints, lumbar spine movement is often described as a change in curvature or lordosis, and represents the motion at multiple vertebral levels. Therefore, lumbar displacement is measured differently from peripheral joints. Lumbar curvature is often quantified as the lumbosacral angle, which is the angle between two lines perpendicular to the tangent at the estimated levels of T12-L1 and L5–S1 (Adams, Dolan, Marx, and Hutton, 1986 ; Dolan and Adams, 1993a; Dolan, Adams, and Hutton, 1988). Because of the invasive nature of embedded markers, the lumbosacral angle calculated from sagittal plane radiographs of the lumbar spine has been considered the best estimate of *in vivo* lumbar spine curvature.

The reliability of x-ray measures of lumbar curvature has been well established. For example, Adams et al. (1986) compared inter-observer reliability of x-ray measures of the lumbar curvature and reported average absolute and relative errors for lumbar curvature of 1.5° and 0.5° respectively. Polly et al. (1996) assessed the reliability of measuring lumbar lordosis from the lateral radiographs. Intra-observer and inter-observer reliability coefficients ranged from 0.82-0.92. From these measures it was concluded that skilled persons could reliably reproduce lumbar lordosis angles from radiographs (Polly et al., 1996).

On the assumption that x-ray measures are a reliable measure of spinal curvature, a number of researchers have used radiographic images to validate external measurement devices, including inclinometers, goniometers (Adams et al., 1986; Newton and Waddell, 1991; Portek, Pearcy, Reader, and Mowat, 1983) electromagnetic tracking devices and skin markers used with video analysis (Chen and Lee, 1997; Gracovetsky et al., 1990). External estimation of lumbar curvature was originally based on the work of Adams et al. (1986) who quantified lumbar curvature using electronic goniometers attached to the skin surface above S1 and L1. The lumbosacral angle was calculated as the difference in the inclinometer tangents measured at S1 and L1. Adams et al. reported no significant difference between x-ray and inclinometer measures of lumbar flexion angles, with correlation coefficients of 0.91.

Radiographic images of the lumbar spine have also been used to validate skin surface markers used for video analysis. Researchers have addressed the validity of palpating bony landmarks for marker placements (Chen and Lee, 1997; Harlick, Milosavljevic, and Milburn, 2007) and assessed the accuracy of skin marker estimation of lumbar curvature (Chen and Lee, 1997; Gracovetsky et al., 1990). Chen and Lee (1997) investigated the ability of an experienced orthopaedic surgeon to place radio opaque markers on the skin surface adjacent to the spinous processes of L1, L2, L5 and S1. Marker locations were compared to bony landmarks of the lumbar spine on x-ray film. They reported that in 10% of the trials the external markers were located 0.5 cm above or below the spinous process, as exhibited on the x-ray. In a similar experiment Harlick, Milosavljevic, and Milburn (2007) investigated the palpation error associated with attaching skin markers by highly experienced physiotherapists. The mean inaccuracy of the therapist in locating skin surface markers was 19.3 mm when compared to radiographic images.

Few researchers have investigated the accuracy of skin surface markers to measure lumbar curvature in various degrees of forward flexion. Gracovetsky et al. (1990) taped steel balls on the skin surface above the spinous processes between the vertebral levels of T12 and L5 in 27 subjects with spinal disorders. Lateral radiographic images of the lumbar spine that were taken in standing postures between 0 and 60 degrees of lumbar flexion were compared to the lumbar angle calculated from the skin surface markers. Preliminary data from two subjects showed that intra-subject correlation coefficients were greater than 0.90. However, the slopes of the linear relationship between radiographic and external marker angle differed between subjects. Gracovetsky et al. also reported that the lumbar angle determined by external markers tended to underestimate radiographic derived angles. It was concluded that external markers could be useful in showing relative changes in lumbar angle, as opposed to being an absolute measure of *in vivo* lumbar curvature during a flexion/extension manoeuvre.

Subsequently Chen and Lee (1997) compared radiographic and videotaped vertebral angles between L1 and S1 in healthy subjects in four torso positions between upright standing and 90 degrees of torso flexion. Lateral radiographs and video recordings of each position were taken in sequence and lumbosacral angle was calculated using the vertebral angle at L1 and S1. These authors reported no significant differences between radiographic and videographic vertebral angle at L1. However, large vertebral angle discrepancies of up to 30 degrees were reported for the vertebral angle at S1. When linear regression models were developed adjusting for the discrepancies at S1, predictive R^2 values for lumbosacral angle were 0.846. These findings supported those of Gracovetsky et al. (1990), showing that skin surface markers can provide an accurate measure of relative change in lumbar curvature, with most measurement error occurring at the S1 level.

Whilst video analysis of lumbosacral angle can only accurately provide a relative measure of lumbar spine motion, the reliably of videographic measurement techniques is high. Levine and Whittle (1996) tested the reliability of a three dimensional video analysis system for measuring lumbar lordosis of subjects in normal standing, and when performing maximum posterior and anterior tilting of the pelvis. It was reported that the lumbosacral angle range of motion between maximum anterior and posterior pelvic tilt was on average 65 degrees and intraclass correlation coefficients for all three positions were greater than 0.90. Although the results of this study indicated that the video analysis was reliable at reproducing lumbar angle predicted by skin surface markers, Levine and Whittle were unable to establish the systems validity as no comparisons were made with a "gold standard" measure such as x-rays.

In summary bone marker and x-ray studies indicate that there are limitations when using passive surface marker systems to recorded absolute peripheral skeletal and lumbar spine motion. However, passive marker systems are reasonably accurate and reliable at measuring relative change of motion in the sagittal plane provided that appropriate data sampling frequency and filtering processes are used. This is why a number of lifting and sudden loading studies choose to measure joint angle motion relative to that collected in an upright standing posture (Adams et al., 1986; Dolan and Adams, 1993b, 1998).

Chapter Three

The effects of prior exposure, warning, and initial standing posture on muscular and kinematic responses to sudden loading of a hand-held box.

PRELUDE

The first aim of the thesis was to examine the effects of prior exposure and warning on muscular and kinematic responses to sudden anterior loading when adopting upright and stoop postures. Accordingly, trunk and lower limb muscle onset latencies and postural responses were recorded during sudden anterior loading of a box held in upright and stoop postures with and without forewarning. In order to assess a situation where the load was truly unexpected the subjects were given no indication of the type, magnitude or timing of the sudden loading event in the first trial. A paper from this chapter has been published in the journal "Clinical Biomechanics" (Mawston, McNair, and Boocock, 2007a)

INTRODUCTION

Up to 12% of low back related injuries have been attributed to sudden loading events (Manning et al., 1984), such as when an additional load is suddenly applied to an object held in the hands, or the contents of a container move unexpectedly. During such a perturbation, the neuromotor system must react quickly by activating appropriate muscles and generating kinematic responses of the trunk and lower limbs in order to maintain balance while preventing excessive loading (Carlson et al., 1981; Cresswell et al., 1994; Diener et al., 1988; Henry et al., 1998). The inability to execute postural responses appropriately in response to these sudden loads has the potential to increase loading on the lumbar spine and surrounding tissue (Mannion et al., 2000).

Although postural responses elicited during sudden loading are rapid (Diener et al., 1988; Henry et al., 1998; Leinonen et al., 2002), they have the potential to be affected by the central nervous system. It has been proposed that by providing prior exposure to a sudden loading event, the information gained from that exposure allows the central nervous system to influence subsequent postural responses to a similar loading event (Horak and Nashner, 1986). The effect of prior exposure on postural responses after sudden loading is not clear, as most sudden loading experiments have provided subjects with familiarisation trials prior to data acquisition (Cresswell et al., 1994; Henry et al., 1998). For example, Pedersen et al. (2004) reported that following a practice trial repeated exposure to sudden loading did not alter ES muscle onset latency. Hence, it has not been possible to establish to what extent postural responses differ in response to a completely novel sudden loading event, which may be considered more reflective of that occurring in the work environment, compared to an event where the individual has been previously exposed to loading parameters.

In addition to prior exposure to a perturbation, providing a warning signal may also influence postural responses. With warning, a better estimation of the timing onset of a sudden loading event may allow the neuromuscular system to respond more swiftly to the perturbation (Leinonen et al., 2002; McChesney et al., 1996).

Forewarning of a sudden loading event has been shown to reduce muscle onset latencies of specific trunk and leg muscles during anterior trunk (Leinonen et al., 2002) and underfoot perturbation (McChesney et al., 1996). However, it is unclear whether the provision of a warning signal alters the muscle and kinematic phasing patterns elicited in response to a sudden loading event. Some authors have shown that, with warning, preparatory co-contraction of anterior and posterior trunk musculature is often observed (Thomas et al., 1998), whilst others have noted no change in activation patterns (Granata et al., 2001).

Postural responses to sudden loading with varying levels of warning have been primarily assessed in an upright standing position. Evidence exists, however, that loading of tissue in a stoop posture (in which the legs are relatively straight and the hips and lumbar spine are flexed) may present a greater risk of injury to the spine when compared to standing upright. For example, as the lumbar spine approaches end range flexion, paraspinal and hamstring muscle group geometry and level of activation change, potentially reducing their ability to resist the bending moments imparted on the trunk (Macintosh et al., 1993; McGill et al., 2000). As both upright and stoop postures are commonly used during manual handling activities, a comparison of responses elicited to sudden perturbation in both upright and stoop posture would seem appropriate.

The aim of the current study was to investigate the effects of prior exposure, warning and standing posture (upright versus stoop) on muscular and kinematic responses to sudden anterior loading. Subjects had no prior experience of the perturbation event, thereby allowing comparisons to be made with subsequent perturbations with and without prior warning. These findings may provide insight into mechanisms of injury and preventative strategies for reducing the incidence of musculoskeletal injury.

METHODOLOGY

Subjects

Thirty healthy male volunteers with a mean age of 25 years (SD = ± 4.8 years), mean mass of 76.3 kg (SD = ± 10.6 kg) and mean height of 1.78 m (SD = ± 0.07 m) participated in the study. None of the subjects had experienced an episode of low back pain within the twelve months prior to the commencement of this study. Written and verbal explanations of the experimental procedures were provided in accordance with the requirements of the Auckland University of Technology Ethics Committee, and subjects provided written consent prior to their participation (see Appendices 1 and 2 for Participant Information Sheet and Subject Consent Forms).

Experimental protocol

A mechanical device (Figure 3.1) was used to produce a sudden downward motion of an aluminium box held by subjects. The first trial undertaken by all subjects (no prior exposure) was a completely novel event where the subjects were not advised of the direction, magnitude, and timing of the sudden unexpected load. This was followed by three subsequent perturbations in an upright (with and without warning) and stoop posture (with and without warning). Trial order for warning and posture (standing versus stoop) was randomised for each subject. In the upright standing posture, subjects held a box (20 cm in height x 23 cm wide x 25 cm deep) so that its handles were at mid-thigh level, adjacent to, but not touching the thighs. In the stoop posture, they stood on a raised platform (20 cm high) and were instructed to bend forward so that the handles of the box were at the height of the patella. In this position the knees were fully extended and lumbosacral flexion was, on average, 48 degrees from the erect standing position. The raised platform allowed the box to be closer to the floor while still allowing sufficient space between the floor and the box for the loading device to be attached. In both the upright and stoop postures the subject's feet were at the same horizontal distance (10 cm) from the loading apparatus, taking care to ensure that the cable attached to the box exerted a force in a vertical downward direction.

Sudden downward motion of the box was initiated with or without warning whilst subjects stood in an upright or stoop posture. An adjustable chain, connected to a steel cable that was attached to a constant pre-load of 9.8 N, was bolted to the bottom of the box (Figure 3.1). An electromagnetic switch triggered the release of a 30 N weight attached to a pendulum arm, which suddenly increased the tension in the pre-loaded steel wire. This produced a sudden downward loading force on the box of approximately 100 N. An accelerometer attached to the base of the box was used to determine the moment at which downward movement occurred, as well as provide a means by which to synchronise kinematic data collected from a marker located on the right hand side of the box.

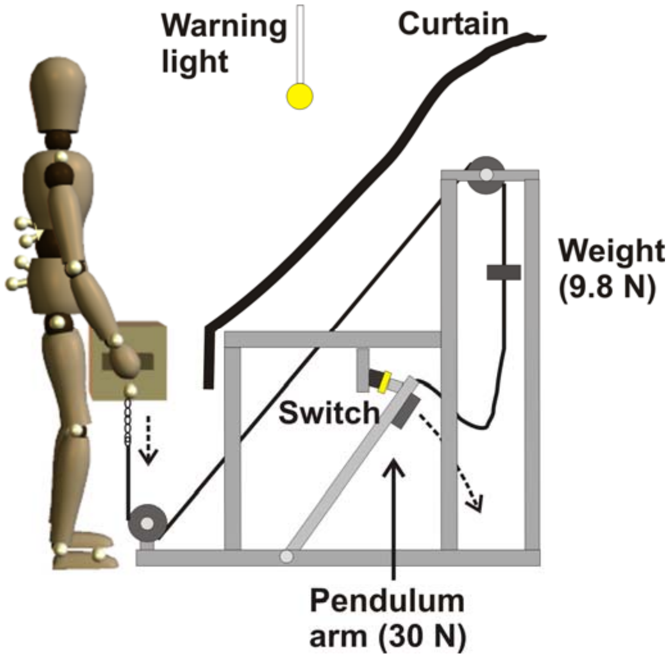


Figure 3.1. Sagittal view of the experimental setup with the subject in an upright standing posture.

In both the upright and stoop postures, subjects were advised that a warning light located at eye level would be illuminated 1-3 seconds prior to the application of a sudden load to the box. During warning trials, a light located directly in front of the eyes was illuminated while simultaneously transmitting a 3 volt signal to the computer acquisition system. During the first trial and no-warning trials, the load was applied without the light being illuminated. To avoid hearing of keyboard strokes and switch initiation, a beating metronome and relatively loud background music were played.

Electromyography (EMG)

The muscle activity of seven major muscle groups located on the right side of the body were recorded using surface EMG. The skin was shaved and cleaned with alcohol swabs prior to application of the electrodes. Electrodes were placed approximately parallel to the muscle fibre direction of the biceps femoris (BF), and vastus lateralis (VL) muscles in accordance with the protocol of Merletti et al. (2005). In addition, electrodes were placed on the skin surface superficial to the LES (approximately 2 cm lateral to the L4-5 interspinous space), UES (5 cm lateral to the T9 spinous process), RAB (3 cm from the midline just above the umbilicus), IO (midway between the anterior superior iliac spine and symphysis pubis, superior to the inguinal ligament) and EO (15 cm lateral to the umbilicus) (Cholewicki and McGill, 1996).

Active electrodes (Delsys DE02.3, Delsys Inc., Boston, USA) with an inter-electrode distance of 10 mm were used. EMG signals were collected at a sampling rate of 1000 Hz and band-pass filtered (20-450 Hz). EMG data from each muscle were demeaned and rectified. To determine the onset latency time for each muscle, a blinded, visual observation method with no reference points was used (Hodges and Bui, 1996).

Kinematic analysis

Two-dimensional kinematic data of subjects' postural movements in the sagittal plane were collected using a digital video camera (JVC GR-DVL 9800) located 4 m from the subject. Retro-reflective markers were attached to the skin surface overlying anatomical locations on the body in order to measure joint angles. The angle at the ankle joint was measured using markers placed on the base of the fifth metatarsal, the lateral malleolus of the ankle and the lateral femoral condyle of the knee joint. Markers placed on the lateral malleolus, the lateral femoral condyle of the knee and the greater trochanter provided a measure of knee joint angle. Hip joint angle was represented by markers located on the lateral femoral condyle of the knee joint, the greater trochanter, and the sacrum. Measures of lumbosacral angle were determined from the angle formed by the tangents of two pairs of markers placed on the skin surface overlying the first lumbar vertebrae and sacral level one aspect of the spine (Caldwell, McNair, and Williams, 2003) (Figure 3.1). To allow an appreciation of the degree of lumbosacral flexion adopted in the stoop posture, total range of lumbosacral flexion was determined as the difference between lumbosacral angle during upright standing and that in maximal lumbar flexion (Adams et al., 1986).

The video camera recorded digital images of the illuminated retro-reflective markers at a sampling rate of 120 Hz. These were synchronised with EMG signals via a switch which illuminated a light located posterior to, but outside the field-of-view of the subject. The switch also passed a simultaneous voltage (3 volts) to the computer acquisition system. After data collection, digital images were transferred to a computer and appropriate frames were identified for digitising. The positional co-ordinates of individual markers were digitised into respective 'x' and 'y' co-ordinates using a video analysis software package (ARIEL Performance Analysis System, San Diego, USA). Co-ordinate data were low pass filtered using a digital filter with a cut off frequency of 12 Hz. The cut off frequency of 12 Hz was determined based upon a spectral analysis of the frequency component of individual markers. Filtered data were converted to angular data using LabVIEW version 6i development software (National Instruments, Austin, TX, USA). The kinematic variables of joint onset latency (time) and maximal angular displacement (degrees) were determined using a visual cursor routine, where the assessor was blinded to trial order, warning condition and the posture adopted.

Statistical analysis

Descriptive statistics were calculated and the data were checked for assumptions related to normality. The dependent variables measured were muscle onset latency, joint onset latency, and maximal joint displacement. The independent variables of interest were prior exposure, warning, and posture. Firstly, a 2 factor ANOVA using the factors: warning (1st trial, no warning, warning) and muscles (VL, HM, UE, LE, IO, EO), or joints (ankle, knee, and lumbosacral angle) was undertaken for the upright standing posture. Thereafter, a 3 factor ANOVA involving posture (upright-standing and stoop), warning (warning and no warning) and muscle (VL, HM, UE, LE, IO, EO,) or joint (ankle, knee, and lumbosacral angle) was performed. Where main effects/interactions were identified, pairwise comparisons of means were performed using the Bonferroni test (Ottenbacher, 1991). An alpha level of 0.05 was considered significant for all statistical tests. All statistical analysis was conducted using SPSS 13.0 for Windows (SPSS Version 13, SPSS Inc., 2003).

RESULTS

Prior exposure and warning

Muscle onset latencies showed main effects for muscle ($P<0.05$), warning ($P<0.05$), and a significant interaction between muscle and warning ($P<0.05$) (Figure 3.2A). In the upright standing posture all muscles of the trunk had significantly shorter onset latencies when the warning condition was compared to the first trial ($P<0.05$). Comparing no warning versus the first trial, some muscles (IO, EO, and LES) had shorter onset latencies in the no warning condition ($P<0.05$). Comparing onset times for the no warning versus warning, most muscles (UES, LES, IO, VL, and BF) had shorter onset times with warning ($P<0.05$). Muscle activity of the RAB was excluded from all data analysis as only 4 of the 30 subjects displayed observable changes in baseline activity during perturbation. Joint onset latencies and maximal angular displacement also showed main effects for warning and joint in the upright standing posture ($P<0.05$), with a significant interaction between warning and joint ($P<0.05$). In the warning condition all joint onset latencies were significantly shorter than the first trial and no warning conditions ($P<0.05$). Comparing the no warning condition with the first trial, the ankle and knee joints had significantly shorter latencies in the no warning condition ($P<0.05$) (Figure 3.2B).

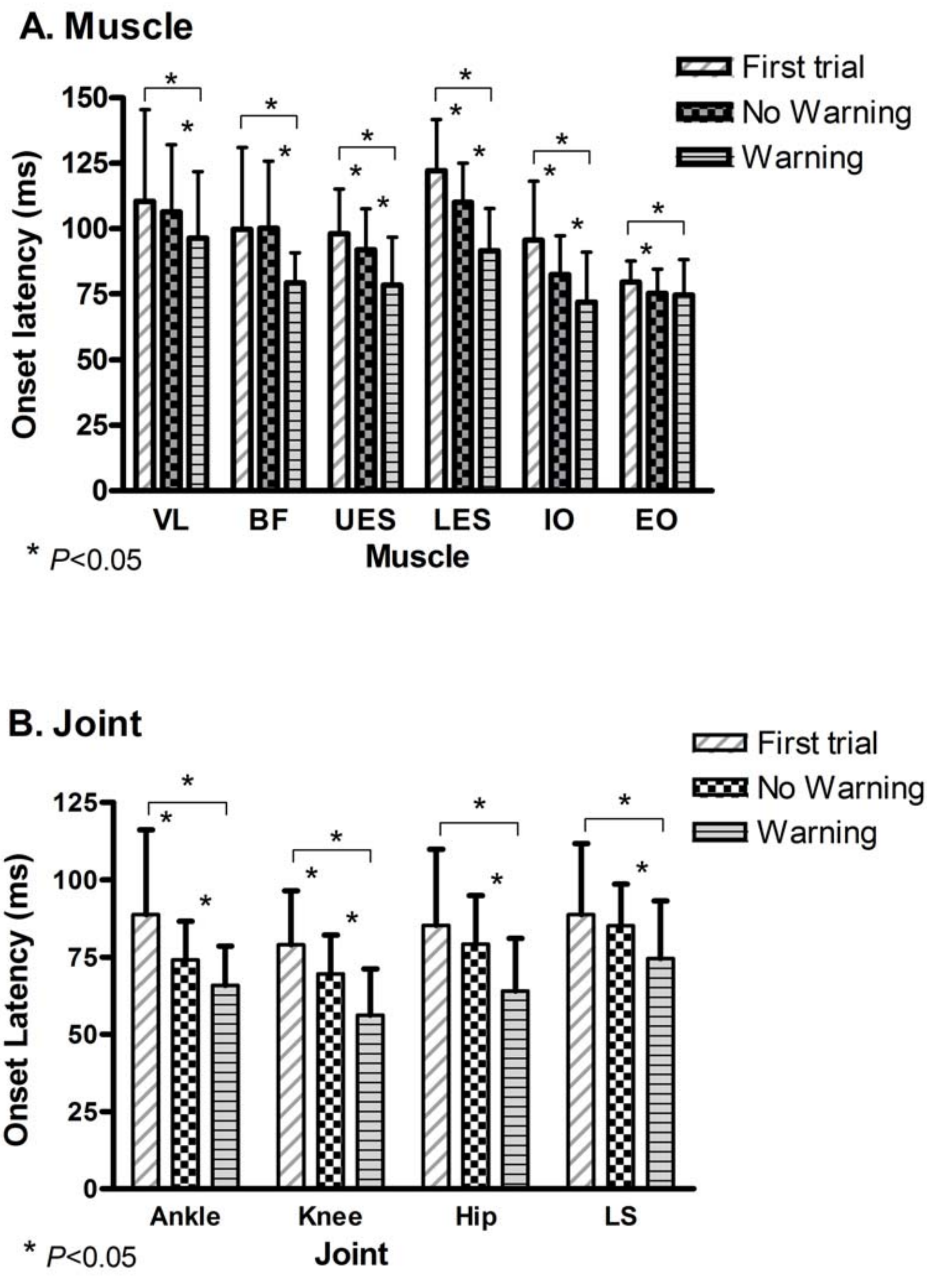


Figure 3.2. Onset latencies of the A) muscles and B) joints of the trunk (lumbosacral angle (LS)) and lower limb in response to sudden loading for the first trial and different levels of warning in an upright standing posture. * with brackets indicates a significant difference across first trial and warning conditions while a lone * indicates significance across adjacent bars.

Maximal angular displacement of the knee, hip and lumbar spine was significantly greater in the first trial than in the warning and no warning conditions (*P*<0.05). When comparing warning with no warning, the knee and hip joints displayed significantly less angular displacement in the warning condition (*P*<0.05) (Figure 3.3). It should be noted that in the first trial, data from 11 of the 30 subjects who either dropped the box or released one of its handles were excluded from comparisons between first, no warning and warning trials in the upright posture.

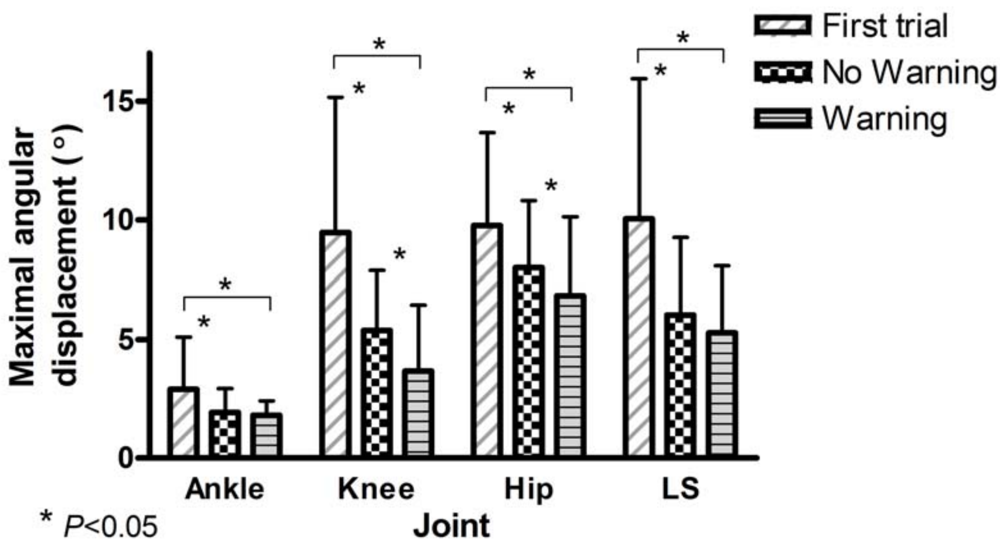
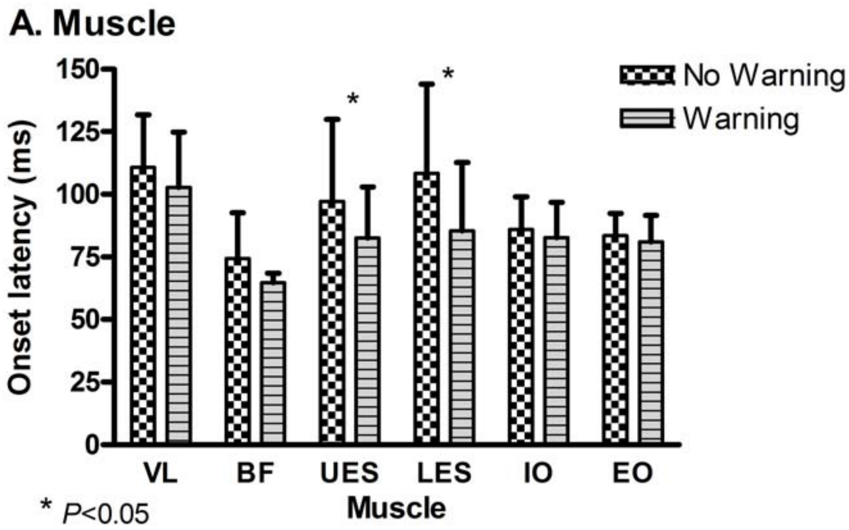


Figure 3.3. Maximal angular displacement of the trunk (lumbosacral angle (LS)) and lower limb for the first trial and different levels of warning in an upright standing posture. * with brackets indicates a significant difference across first trial and warning conditions while a lone * indicates significance across adjacent bars.

Posture

For muscle onset latencies there was no main effect for posture, but there was a significant posture by muscle interaction (*P*<0.05). Subsequent individual muscle analysis showed that EO and IO onset latencies were delayed in the stoop posture compared to upright standing (P<0.05). In contrast BF onset latencies were shortened (P<0.05). It should be noted that in the stoop posture eight subjects had no detectable EMG activity from EO and IO.



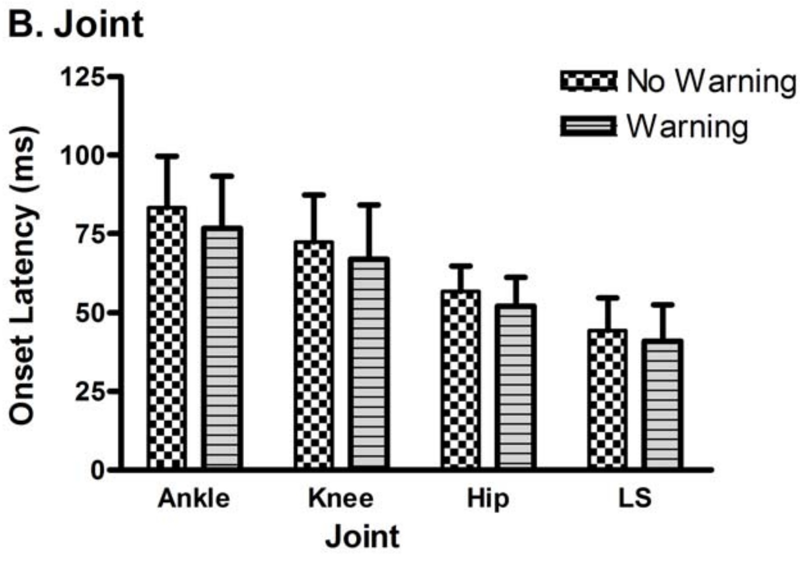


Figure 3.4. Onset latencies of the A) muscles and B) joints of the trunk (lumbosacral angle (LS)) and lower limb in response to sudden loading for the no warning and warning condition in a stoop standing posture.

In the stoop standing posture there was a main effect for warning ($P<0.05$) and muscle ($P<0.05$) with an interaction between warning and muscle ($P<0.05$). In the stoop posture the UES and LES muscles had shorter onset latencies in the warning compared to the no warning condition ($P<0.05$) (Figure 3.4A).

Joint onset latencies and maximal angular displacement showed main effects for posture ($P<0.05$), with a posture by joint interaction ($P<0.05$). The onset of joint motion in a stoop posture differed from that in upright standing. There was a more sequential pattern in the former posture, with progressively significant decreases in onset latencies for those joints more proximal to the perturbation ($P<0.05$) (Figure 3.4B). Across joints (Figure 3.4.B), there were no significant differences in joint onset latencies between the warning and no warning conditions in the stoop posture. Overall, the maximal angular displacement in the stoop posture (Figure 3.5) was significantly less than that for upright standing (Figure 3.3) ($P<0.05$). With respect to the stoop posture there was a significant effect of warning ($P<0.05$), with a significant interaction between warning and joint ($P<0.05$). With the addition of warning there was a reduction in angular displacement of the knee and hip in the stoop posture ($P<0.05$) (Figure 3.5).

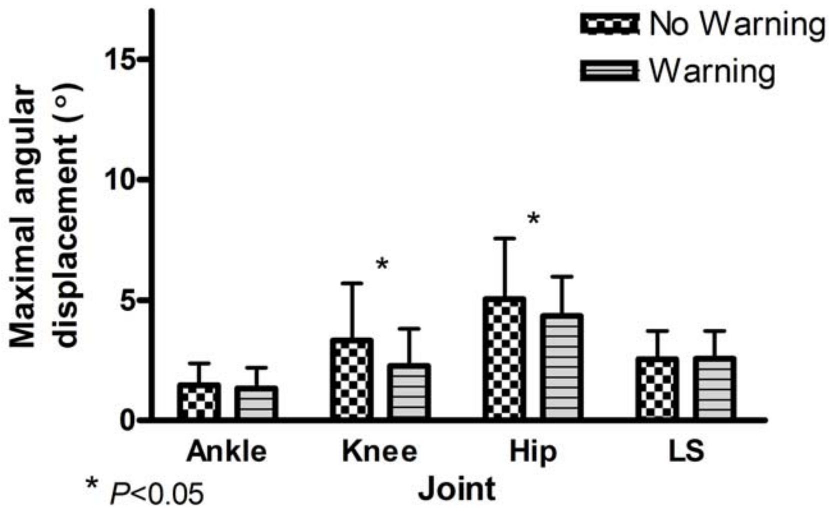


Figure 3.5. Maximal angular displacement of the trunk (lumbosacral angle (LS)) and lower limb for the no warning and warning condition in a stoop standing posture.

DISCUSSION

This study found that postural responses to loading of a box held in the hands were influenced by prior exposure to, and visual warning of the sudden loading event, and that these responses were affected by standing posture. After the exposure of a single trial in the upright standing position, most trunk muscles, with the exception of UES, activated earlier, and knee and ankle joint response times were decreased. This finding was in contrast to that of Pedersen et al. (2004) who found no difference in muscle onset latency across repeated trials. However, it should be noted that Pedersen et al. provided subjects with a sudden loading experience in advance of actual data collection. Our findings suggest that a single exposure to a sudden loading event provides sufficient feedback to improve the response times of trunk musculature for subsequent exposures, but that further repeated exposure has little effect on muscle response latencies. The earlier activation of oblique abdominal and

LES muscles following a single exposure may aid stability of the lumbar spine, particularly as these muscles have been suggested to act as key stabilisers of the lumbar spine (Bergmark, 1989; Gracovetsky et al., 1985; Macintosh and Bogduk, 1986). For example, the LES includes the muscle fascicles of multifidus which have the potential to stabilise the lumbar spine at a segmental level (Bergmark, 1989; Macintosh and Bogduk, 1986). Though controversial, the IO muscle, through its attachments to the thoracolumbar fascia may resist bending moments on the lumbar spine (Gracovetsky et al., 1985). It has also been suggested that the IO and EO play an important role in increasing intra-abdominal pressure which may also serve to improve lumbar spine stability (Bartelink, 1957; Cresswell et al., 1994).

This study found that subjects not only responded more slowly during the first trial, but also exhibited increased joint angular displacements. For the lower limbs and trunk, there was between 18% and 43% greater angular displacement during the first trial compared to the no-warning condition. Although no previous perturbation research has compared novel loading events with subsequent trials, it has been shown that repeated exposure to sudden trunk loading reduces the total range of lumbar flexion when compared to previous loading events (Pedersen et al., 2004). Increased flexion of the hip and lumbar spine, in particular, may move the centre of gravity further forward of the base of support, increasing the likelihood of the individual losing balance and experiencing a fall.

The provision of a warning signal when standing upright, further reduced muscle onset latencies of the LES, IO, and UES when compared to the no warning condition. These findings support those of Cresswell et al. (1994) and Leinonen et al. (2002) who found faster activation of anterior and posterior trunk muscles during self-loading and forewarning conditions. The reduction in muscle onset latencies in the presence of advanced warning of a sudden loading event may be best explained by an increased state of readiness. When an individual is provided with a warning signal prior to a reaction time event, changes in motor neuron pool excitability occur, increasing motor neuron pool sensitivity to supraspinal commands (Hasbroucq et al., 1999). This increased sensitivity during warning conditions may have led to a more rapid muscle response to the sudden loading event and earlier muscle activation in order to decrease joint range of motion during trials in which there was warning.

Although muscular response times decreased with prior exposure and warning, the findings showed that the selection and relative phasing of muscles and joint movements remained unchanged irrespective of condition. The UES and abdominal muscles showed evidence of co-activation, followed 14-40 ms later by increased activation of the LES. Activation of the anterioventral muscles prior to the posterior muscles of the trunk during sudden anterior loading has been well documented (Cresswell et al., 1994; Henry et al., 1998) and it has been proposed that abdominal muscle activation and the associated increased intra-abdominal pressure serve to stabilise the spine (Cresswell et al., 1994). However, our finding of delayed phasing of the LES relative to the UES contrasts with that of Leinonen et al. (2002) who found simultaneous responses of the UES and LES during sudden box loading. This difference may be related to experimental design, as Leinonen et al. fixated the lower limb. Our findings may reflect the difference in functional roles of the UES when compared to LES. When perturbed, the UES may have activated earlier as it has a greater mechanical advantage than the LES when resisting bending moments on the spine (Macintosh and Bogduk, 1987). The initiation of joint movement patterns was consistent for all conditions in the upright standing posture and displayed a general pattern that was similar to muscle onset latencies, whereby there was concurrent movement of the trunk and lower limb. This simultaneous execution of knee extension, ankle plantar flexion, hip flexion and lumbosacral flexion produced by most subjects in the current study has been reported as a technique to minimise anterior displacement of the centre of gravity during voluntary self-initiated rapid trunk flexion manoeuvres (Alexandrov, Frolov, and Massion, 1998).

The contribution of the abdominal musculature was less evident in the stoop posture, in that irrespective of warning, the subject's abdominal muscles were either not activated or they were delayed. Minimal activation of abdominal muscles has also been illustrated during load lifting in the stoop posture where the spine is near end range flexion (McGill, 1997b). The implication might be that the posterior muscles and passive ligamentous structures of the trunk are more likely to contribute to resisting the bending moments associated with anterior loading in a stoop posture (Dolan, Mannion et al., 1994; McGill, 1997b).

Large differences between stoop and upright standing were seen in the pattern of joint latency onsets. In contrast to upright standing, perturbation in the stoop posture resulted in decreased movement onset latencies of the hip and lumbar spine, and an increase in ankle joint onset time. The changes observed in the stoop posture led to a movement initiation sequence where latencies increased sequentially from those joints closest to the perturbation to those most distal from the perturbation. The shortening of hip and lumbar spine movement onset latencies was most probably due to the additional tensioning of the system. When subjects adopted the stoop posture, on average they flexed their lumbar spine to 90% of the total range of lumbar motion. Towards the end range of lumbar flexion, the ligamentous tissues of the lumbar spine have been shown to increase tension and begin resisting bending moments on the spine (Dolan, Mannion et al., 1994; McGill, 1997b). These passive tissues of the spine attach to the pelvis and through their connections with the sacrotuberous ligament (Vleeming, Stoeckart, and Snijders, 1989) have links to the biceps femoris muscle (Snijders, Vleeming, and Stoeckart, 1993). With increased tension of the passive system, energy can be transferred quickly in a sequential manner from the site of perturbation to the trunk and pelvis, and then to the lower limb. As Cordo and Nashner (1982) have suggested, the staggered sequencing of joints may permit each successive joint to resist smaller inertial loads.

CONCLUSION

The findings of the current study have ramifications for the risk of musculoskeletal injury following sudden anterior loading of the body. In upright standing, prior exposure to, and warning of a sudden loading event enabled postural responses to be elicited more quickly reducing the extent of subsequent joint motion. This emphasises the importance of knowledge about load parameters and utilisation of visual warning cues during manual handling activities. In contrast, the sequential movement pattern initiated during perturbation in the stoop posture resulted in less joint excursion than the upright posture, potentially generating less disruption to the body's equilibrium.

Chapter Four

The effects of prior warning and lifting-induced fatigue on trunk muscle and postural responses to sudden loading during manual handling.

PRELUDE

In the previous chapter prior experience and warning were shown to influence muscular onset latencies and postural responses to sudden anterior loading. Furthermore, the effect of warning was more influential on postural response to sudden loading in the upright posture. However, the influence of warning and fatigue on muscle activation levels prior to the sudden load were not investigated. Moreover, the effects of lifting-related physical fatigue and its effects on trunk muscle activation patterns and joint kinetics during sudden loading, with and without prior warning, warranted further investigation. Accordingly, trunk muscle activation and postural responses to sudden anterior loading in an upright posture, with and without forewarning, were recorded before and after repetitive lifting to physical fatigue. A paper from this chapter has been published in the journal, “Ergonomics” (Mawston, McNair, and Boocock, 2007b), and hence there is some overlap with the previous chapter.

INTRODUCTION

Sudden unexpected loading during manual handling activities has been highlighted as a high risk factor for LBI in the workplace (McCoy et al., 1997). A proposed mechanism for these injuries is the inability of the neuromuscular system to elicit appropriate postural responses to counter the forces generated by the sudden load (Mannion et al., 2000; Thomas et al., 1998). Factors such as insufficient warning and fatigue may contribute to these inappropriate postural responses, which may potentially impair the control of lumbar spinal segment motion (Diener et al., 1988) and increase the forces acting on the lumbar spine (Mannion et al., 2000; Thomas et al., 1998).

Given the potential risk for injury during sudden unexpected loading, it has been suggested that the provision of a warning signal prior to sudden loading may improve the ability of the neuromuscular system to respond to the load and return the body to a state of equilibrium, reducing the likelihood of injury to the spine (Magnusson et al., 1996). Two schools of thought have evolved regarding the influence of warning on postural responses to sudden loading. Firstly, it has been suggested that advanced warning promotes preparatory co-activation of the agonist and antagonist muscles of the trunk in advance of the sudden load (Granata et al., 2001; Thomas et al., 1998). This co-activation of agonist and antagonist muscle groups may serve to increase joint stiffness, in turn increasing the stability and the resistance of the trunk to the external load (Thomas et al., 1998). However, evidence of agonist and antagonist muscle co-activation is equivocal. For example, it has been reported by Cresswell et al. (1994) and Thomas et al. (1998) that visual warning was associated with preparatory co-contraction of the anterio-ventral and posterior trunk musculature, and decreased trunk motion when compared to situations where visual tracking was absent. In contrast, Granata et al. (2001) found no evidence of preparatory co-activation of abdominal and posterior trunk muscles in subjects who were given a single verbal warning prior to sudden loading.

A second school of thought argues that rather than preparatory co-contraction occurring, the provision of warning allows the earlier initiation of muscle and joint synergies following the sudden loading event. This concept is based on the premise that muscle responses to sudden loading are not simple reflexes, but a set of specific muscle activation patterns influenced by the central nervous system, which is often referred to as the postural set (Cordo and Nashner, 1982; Horak and Nashner, 1986). Providing a warning signal prior to sudden loading could potentially increase the central nervous system state of alertness, reducing muscle onset latencies and facilitating earlier joint motion. Evidence exists that during sudden underfoot and anterior trunk loading warning significantly reduces the onset latencies of individual postural muscles (Leinonen et al., 2002; McChesney et al., 1996). However, there is little information whether muscle and joint initiation patterns remain the same with or without warning. Cresswell et al. (1994) found that although muscle onset latencies in response to sudden loading decreased with preparatory warning, the relative phasing patterns of the abdominal and posterior trunk muscles remained the same irrespective of a warning event.

Whilst warning may modify muscle activity and joint motion, uncertainty surrounds the role that fatigue may play in altering muscle and kinematic responses to sudden loading. Fatigue has been identified as an important risk factor for LBI, and repetitive lifting has been associated with low back pain and fatigue-related injuries

(Masset and Malchaire, 1994; Svensson and Andersson, 1989). Granata et al. (2001) found that the level of trunk muscle activity prior to sudden loading was increased following repetitive lifting when compared to that observed in a non-fatigued state. Wilder et al. (1996) also showed that following fatigue (prolonged period of exposure to low frequency vibration) muscle onset latency of the ES was longer in response to the sudden loading of a hand-held box. In contrast, no changes in trunk muscle onset latencies were evident when subjects were exposed to sudden loading following a fatigue-inducing lifting and back extension tasks (Granata et al., 2004; Herrmann et al., 2006).

The effect of fatigue on kinematic responses to sudden loading has not been examined extensively, though fatigue may change co-ordination patterns in voluntary movement tasks (Forestier and Nougier, 1998). In addition there is no clear evidence of how lifting-related fatigue may influence the effects of warning on postural responses to sudden loading. Hence, the purpose of the current study was three fold. Firstly, to investigate the effects of advanced warning on preparatory muscle activity and postural responses to sudden loading. Secondly, to investigate the influence of lifting-induced fatigue on modifying muscle and postural responses to sudden anterior loading, and thirdly, whether the observed muscle and postural responses to prior warning were influenced by fatigue.

METHODOLOGY

Subjects

Thirty-one healthy male volunteers were randomly assigned to two groups: 1) experimental group and 2) control group. No significant differences were found between groups for age (exp = 25.4 yr, standard deviation (SD) = ±4.4 yr; control = 24.5 yr, SD = ±5.6 yr), mass (exp = 78.5 kg, SD = ±9.0 kg; control = 73.2 kg, SD = ±12.2 kg), stature (exp = 1.79 m, SD = ±0.08 m; control = 1.76 m, SD = ±0.07 m), and resting HR (exp = 64.6 beats per minute, SD = ±11.9 beats per minute ; control = 63.8 beats per minute, SD = ±4.7). All subjects were right hand dominant and had not experienced an episode of low back pain within a period of twelve months prior to the commencement of the study. Written and verbal explanations of experimental procedures were provided in accordance with the requirements of the Auckland University of Technology Ethics Committee, and subjects provided written consent prior to their participation (see Appendices 1 and 2 for Participant Information Sheet and Subject Consent Forms).

Experimental design

A repeated measures factorial design was used. Following a practice trial, all subjects received six randomly allocated sudden loading trials, three with and three without warning whilst holding a box. Subjects then performed an exercise task which either involved lifting-induced fatigue (experimental group) or light calisthenic exercises (control group). Immediately after exercise subjects repeated the sudden loading trials randomised with and without warning.

Experimental protocol

To induce sudden anterior loading of the body a purpose-built loading apparatus was constructed (Figure 4.1). This comprised a 30 N weighted pendulum arm that on release created tension in a steel cable attached to the underside of aluminium box held by subjects. Standing in an upright posture, subjects stood with their feet 10 cm from the loading apparatus holding the 3 kg box (20 cm in height by 23cm wide by 25 cm deep) in both hands at mid-thigh level. The steel cable from the pendulum arm was connected to the box via a chain which enabled the height of the box to be easily adjusted to suit individual subjects' stature. The steel cable was pre-tensioned with a load of 9.8 N. Sudden loading of the box involved the triggering of an electromagnetic switch that released the 30 N pendulum arm. This swung through an arc of 45 degrees and applied a downward vertical force of approximately 100 N to the box. Initial downward motion of the box was detected by an accelerometer attached to the base of the box which enabled movement of the box to be synchronised with recordings of postural motion and muscle activity.

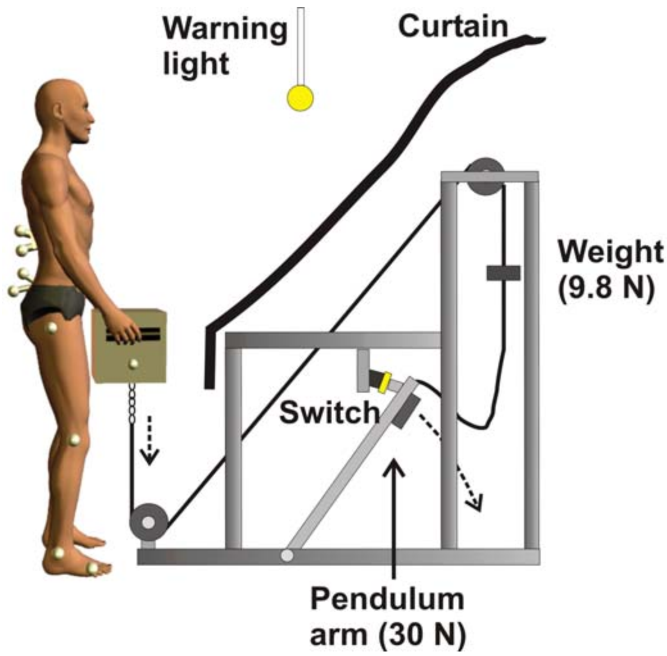


Figure 4.1. Sagittal view of the experimental setup

Prior warning of sudden loading

To provide subjects with prior warning of the sudden loading event, all subjects were advised that a warning light located at eye level would be illuminated 1-3 s prior to the release of the 30 N load (Figure 4.1). During warning trials, the light was illuminated at randomly selected periods of between 1 to 3 s prior to the loading event. To reduce the likelihood of subjects anticipating movement of the box during the no warning conditions, subjects were not informed when the loading event would take place. To exclude potential auditory cues of the sudden loading event, a combination of metronome tapping and loud music was played.

Lifting-induced fatigue and light exercise tasks

The subjects were randomly allocated to either: 1) a lifting-induced fatigue task (experimental group) (n=18); or 2) a light exercise group (control group) (n=13).

Lifting-induced fatigue consisted of a dynamic lifting task based on the Progressive Isoinertial Lifting Evaluation (PILE) protocol described by Mayer et al. (1988). This involved subjects lifting a 10 kg box from a raised platform (15 cm in height) that was located directly in front of the subject's feet to waist height. The subjects were required to lift to the beat of a metronome that equated to a rate of 20 lifts per minute. Subjects were instructed to continue lifting until they either felt excessive discomfort or subjective fatigue, their HR exceeded 85% of an age predicted maximum HR (220-age), or they were unable to keep pace with the metronome (Mayer et al., 1988). To determine perceived effort, the Borg rating scale of perceived exertion (RPE) was used (Borg, 1982). In addition to RPE, subjects were asked to identify the body location where discomfort was most evident: i.e. whole body, shoulders, wrists, thighs, buttocks, upper back or lower back. The subjects in the control group undertook callisthenic exercises that involved a combination of self-paced stepping and arm circular motion for a period of five minutes.

Electromyography (EMG)

Surface EMG signals were recorded from five trunk muscles. An initial pilot study found no significant difference between onset latencies for the right and left paraspinal muscles during sudden loading and therefore, EMG recording was restricted to the right side of the body. Active electrodes (Delsys DE02.3, Delsys Inc., Boston) with an inter-electrode distance of 10 mm were placed on the skin surface superficial to the muscle fibres of LES, UES, RAB, IO, and EO in accordance with the protocols of Cholewicki and McGill (1996). Prior to the application of all electrodes, the skin was shaved and cleaned with alcohol swabs.

EMG signals from each of the five muscle groups was sampled at a rate of 1 KHz and band-pass filtered (20-450 Hz). To normalise baseline EMG data prior to sudden loading (preloading), a 2 second segment of muscle activity from three static MVCs was recorded by calculating the root mean square (RMS) of the EMG signal. The highest RMS value from the three maximal efforts was then used to express the preloading muscle activity as a percentage of MVC.

Figure 4.2 illustrates the primary measures used to analyse the EMG signal. For all trials a 200 ms period of muscle activity (preloading RMS) was analysed immediately prior to sudden loading and expressed as a percentage of MVC. This period was chosen as it occurred at least 800 ms after the warning signal, allowing an appreciation of any changes in preloading muscle activity following the warning signal. In order to determine muscle onset latencies, EMG data from each muscle were demeaned and rectified. A blinded visual observation method with no reference points was used to establish the onset latencies for each muscle (Hodges and Bui, 1996).

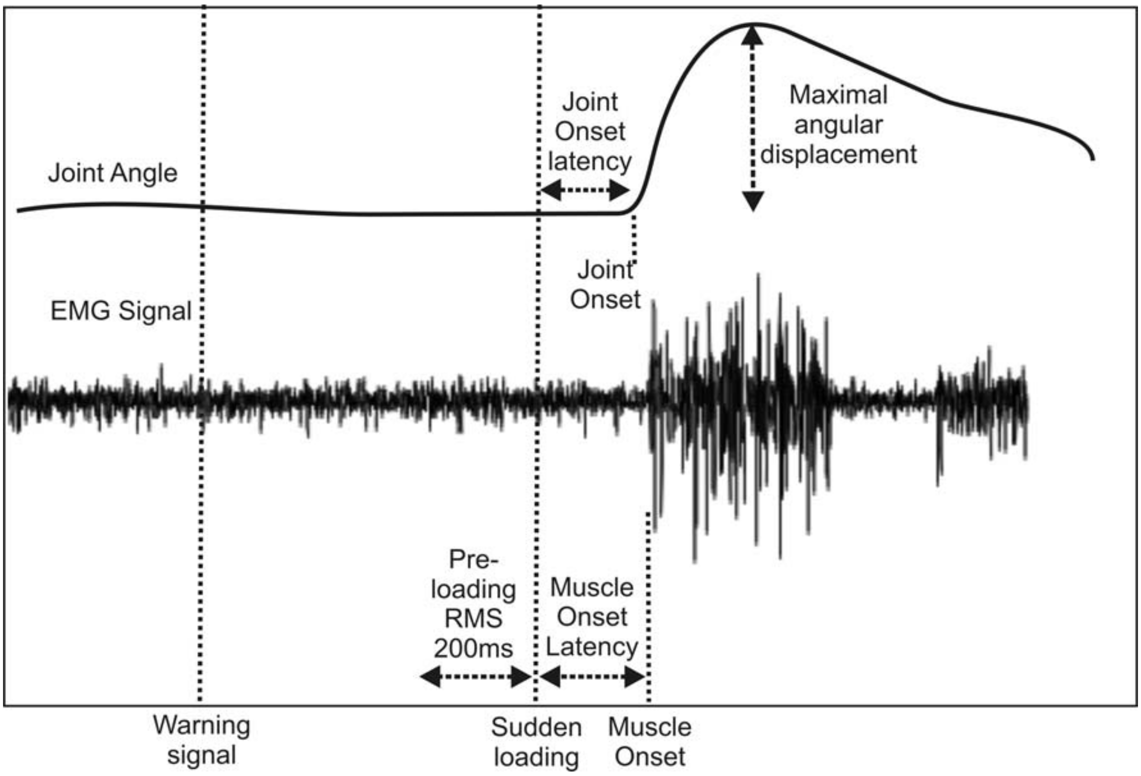


Figure 4.2. Muscle activity and kinematic variables recorded from the EMG signals and measures of joint angles.

Assessment of local muscle fatigue

To assess local muscle fatigue, EMG signals from the UES and LES were recorded during a static MVC of experimental subjects prior to sudden loading and following the exercise-induced fatigue. This test was performed in a modified Biering-Sørensen position, with the subjects lying on a plinth in a prone position with the trunk unsupported and the lower limbs strapped to the plinth (Biering-Sorensen, 1984). Two seconds of consistent EMG data were selected from the five second isometric contraction. Pre-test and post-test MVCs were filtered with a Hanning Window and a Fast Fourier transformation was used to derive a power spectrum of the EMG signal, from which the MDF was calculated. A decrease in MDF of the ES was considered to provide evidence of local muscle fatigue (S. H. Roy et al., 1989).

Kinematic analysis

A digital video camera (JVC GR-DVL 9800) located four metres from the subject recorded sagittal plane, two dimensional kinematic data of subjects' postural movements. In order to measure joint angles, retro-reflective markers were attached to the skin surface overlying anatomical locations at the ankle, knee and hip joints. Lumbosacral angle was also determined and derived from two pairs of markers placed at the first lumbar (L1) and sacral levels (S1). Using each pair of markers, tangents were drawn and the difference between the lumbar and sacral tangents provided a measure of lumbosacral angle (Caldwell et al., 2003). Digital images of the retro-reflective markers were captured at a sampling rate of 120 Hz. These images were synchronised with EMG signals via a switch which illuminated a light positioned posterior to and outside the field of view of the subjects. On initiation of the switch, a simultaneous voltage (3 volts) was sent to the computer data acquisition system. Digital images were transferred to a computer and individual markers were digitised into respective x and y co-ordinates, using commercially available motion analysis software (ARIEL Performance Analysis System, San Diego, USA). Co-ordinate data was filtered using a 12 Hz low pass filter, the optimal frequency being determined based on a spectral analysis of signal profiles. Joint angles were derived from the filtered data using LabVIEW version 6i development software (National Instruments, Austin, TX, USA). A visual cursor routine, where the assessor was blinded with respect to joint, warning condition and trial number, was used to determine joint onset latency and peak joint angles.

Statistical analysis

To investigate the effects of warning and lifting-induced fatigue on muscle activity, a 2 (light exercise vs. lifting-induced fatigue) x 2 (pre and post exercise) x 2 (no warning or warning) x 5 (UES, LES, RAB, IO, EO) mixed model MANOVA with repeated measures was implemented. A second similar repeated measures MANOVA ((2 (light exercise vs. lifting-induced fatigue) x 2 (pre and post exercise) x 2 (no warning or warning) x 4 (ankle, knee, hip, lumbosacral angle)) was also used to investigate the effects of warning and lifting-induced fatigue on joint onset latencies and maximal joint displacements. The within group factors were exercise (pre vs post exercise) and warning (no warning vs. warning), and the between group factor was exercise type (light exercise vs. lifting-induced fatigue). The muscle and joint dependent variables included: preloading RMS and onset latencies for UES, LES, RAB, IO, and EO muscles; and onset latencies and maximal displacement of the lower limb joints (ankle, knee, hip) and the lumbosacral angle. Significant main effects were followed up with post hoc tests (Bonferroni test) of individual muscles and joints. All statistical analyses were conducted using SPSS for windows version 13 software package (SPSS Version 13, SPSS Inc., 2003). The alpha level was set at 0.05.

RESULTS

Warning

Muscle activity

For muscle activity prior to sudden loading (preloading RMS EMG) there was a significant main effect for warning ($P<0.05$). Subsequent analysis showed that LES was the only muscle to significantly increase levels of muscle activity prior to sudden loading when the warning condition was compared to the no-warning condition ($P=0.05$) (Figure 4.3A). For muscle onset latencies there was also a significant main effect for warning ($P<0.05$), with all muscles exhibiting a decreased onset latency for the warning condition when compared to the no-warning condition ($P<0.05$) (Figure 4.3B).

It should be noted that RAB was excluded from an analysis of muscle onset latency, as only 4 of the 30 subjects showed any observable change in baseline RAB muscle activity in response to sudden loading, whereas only 10% of other muscle onset latencies were not detected because of no observable change in baseline muscle EMG activity, inability to distinguish muscle onset because of elevated baseline muscle activity, or movement artefact.

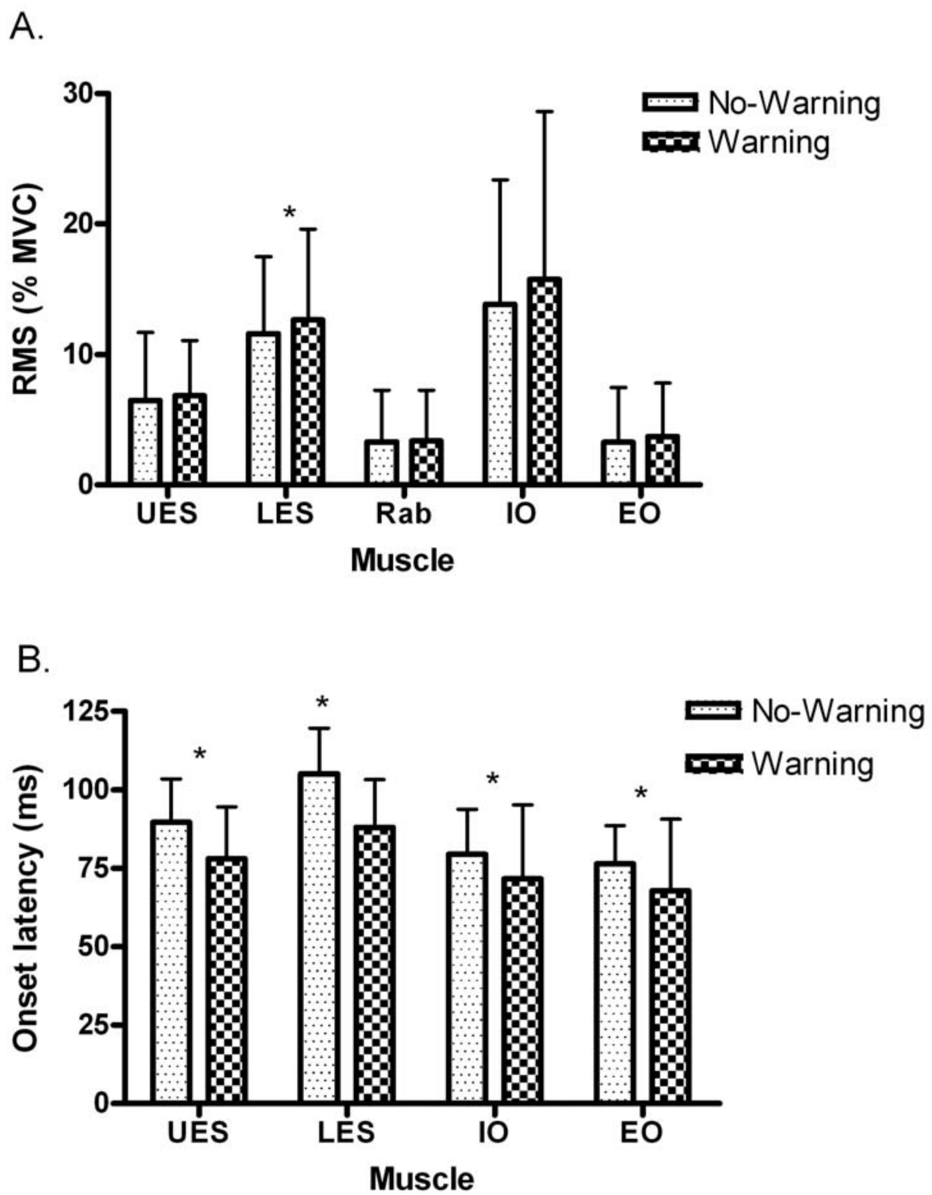


Figure 4.3. A) Mean (SD) preloading RMS expressed as a percentage of MVC for 5 muscle groups for the no warning and warning conditions. B) Mean (SD) onset latencies for each of the 4 individual muscle groups for the no warning and warning conditions. * $P<0.05$.

Joint Kinematics

Joint onset latency showed a significant main effect for warning ($P<0.05$), with all joint onset latencies being significantly shorter for the warning condition when compared to the no warning condition (*P*< 0.05) (Figure 4.4). For maximal joint displacement there was also a significant main effect for warning (*P*= 0.05) and joint (*P*< 0.05). In general, subjects decreased maximal joint displacement of the knee (*P*< 0.05) and hip (*P*= 0.05) when a warning signal was present.

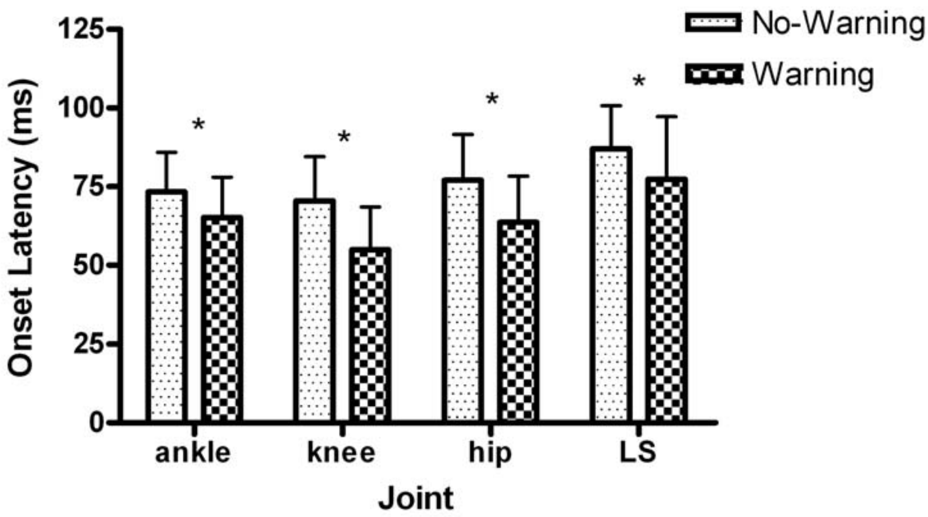


Figure 4.4. Mean (SD) joint onset latencies of individual joints for the no-warning and warning conditions. * *P*<0.05.

Fatigue

Lifting-induced fatigue

All subjects who undertook repetitive lifting stopped lifting because of self-reported fatigue, with a RPE of 19 (very very hard). The mean level of RPE for the control group was 7 (very very light). The most common reason why subjects in the experimental group stopped lifting was due to perceived discomfort in the lower back. EMG frequency analysis of experimental subjects following lifting-induced fatigue revealed a 14.5% and 11% decrease in the MDF of the UES and LES muscle groups, respectively (*P*= 0.05).

Muscle activity

Muscle activity prior to sudden loading (preloading RMS EMG) showed no main effect for exercise or exercise type, indicating that the activation levels of all muscles during box holding were not changed by fatigue. Muscle onset latencies, however, showed significant main effects for exercise (*P*<0.05) and muscle (*P*<0.05), but not for exercise type. There was also a significant exercise by muscle interaction (*P*=0.05). Subsequent analysis showed that LES and EO onset latencies were significantly reduced for both the control and experimental groups following exercise (*P*<0.05) (Figure 4.5).

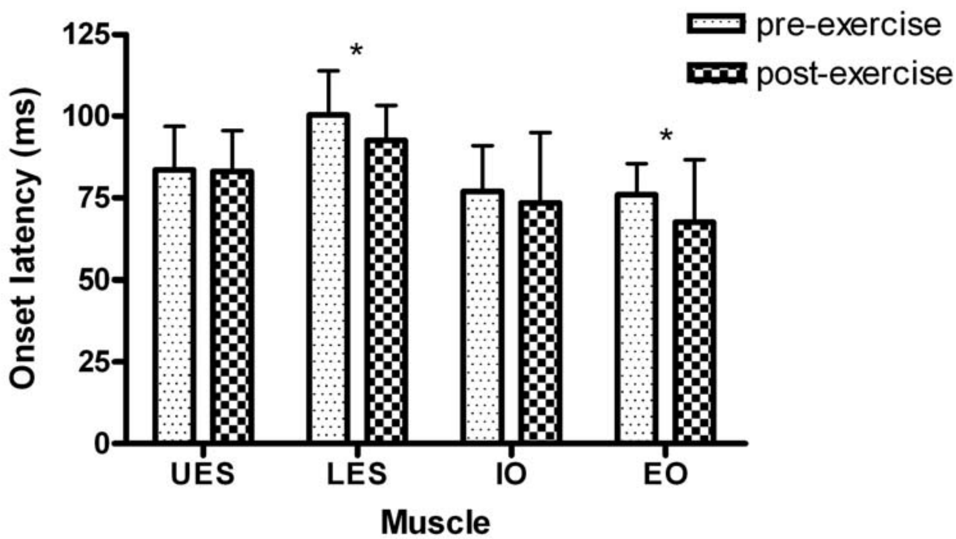


Figure 4.5. Mean (SD) muscle onset latencies for each of the 4 muscle groups pre and post-exercise intervention. * *P*<0.05.

Joint kinematics

For joint onset latencies there was no main effect of exercise, indicating that onset latencies for all joints post exercise remained similar to those pre-exercise for both the experiment and control group. However, for maximal joint displacement, significant main effects were found for exercise ($P< 0.05$) and joint ($P< 0.05$). There was also a significant exercise by warning by exercise type interaction ($P= 0.05$). Subsequent analysis showed that post-exercise both the experimental and control groups exhibited a decrease in the maximal displacement of the knee ($P< 0.05$) and hip ($P=0.05$) joints compared to pre-exercise conditions. With fatigue, the experimental group did not alter maximal joint displacement when provided with a warning, whereas after light exercise the control group significantly decreased knee, hip and lumbar displacement when comparing the warning condition to the no-warning condition ($P< 0.05$) (Figures 4.6A and B).

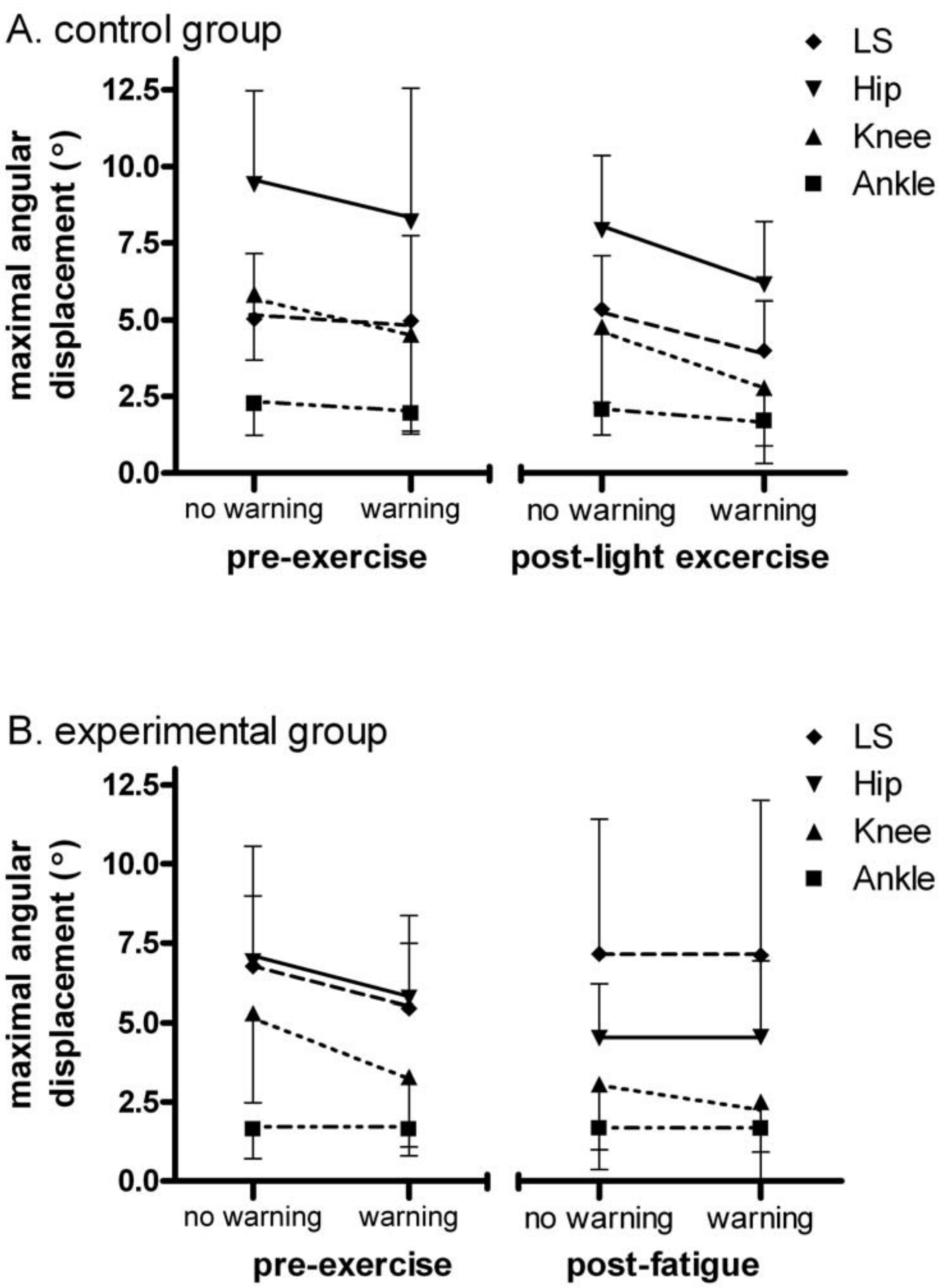


Figure 4.6. Mean maximal angular displacement of the lumbosacral (LS) region and lower limb joints pre and post-exercise for the A) control (light exercise) and B) experimental group (lifting-induced fatigue).

DISCUSSION

Warning

Providing subjects with a visual warning prior to sudden loading of a hand-held box did not change the level of baseline EMG activity for most trunk muscles, but did produce shorter onset latencies across all trunk muscles investigated in this study. These findings support those of Granata et al. (2001) who also reported no evidence of preparatory co-activation of agonist and antagonist trunk muscles when subjects were verbally warned of a sudden loading event. This is in contrast to a number of other studies which have shown an increase in the level of co-activation of anterior and posterior trunk muscles prior to sudden loading (Cresswell et al., 1994; Leinonen et al., 2002; Thomas et al., 1998). The reasons for these contrasting findings may stem from the nature of the warning information provided to subjects and/or the methods of sudden loading used by the different studies. Firstly, in the latter three studies (Cresswell et al., 1994; Leinonen et al., 2002; Thomas et al., 1998) subjects were permitted to track the trajectory of the perturbing object in the warning condition. This may have provided more information about the timing of the sudden loading event in comparison to a single warning light, as used in the present study. Secondly, in these studies (Cresswell et al., 1994; Thomas et al., 1998) sudden loading was initiated via a wire attached to a chest harness, with the lower limbs fixated or supported (Leinonen et al., 2002; Thomas et al., 1998). Cordo and Nashner (1982) reported that externally fixating part of the body during upper limb sudden loading can alter muscle responses when compared to unsupported standing.

The LES was the only muscle to consistently increase levels of baseline activity prior to sudden loading when subjects were presented with a warning signal. Electrode placement for this muscle was such that the electrodes were superficial to the region where the multifidus portion of the LES is most superficial, at a level where there is an absence of other ES fibres (Bojadsen et al., 2000). Bergmark (1989), and Macintosh and Bogduk (1986) showed that the muscle fascicles of multifidus in the lower lumbar spine are arranged in such a way that they can influence inter-segmental lumbar spine stability and even small increases in muscle activity could enhance lumbar spine stability significantly (Cholewicki and McGill, 1996). Thus, with warning, subjects may have increased activation levels of LES prior to sudden loading in order to aid local lumbar spine segmental stability in advance of the sudden load.

While there was no evidence of preparatory activation with warning for most of the postural muscles, there was a decrease in all muscle onset latencies following the provision of a warning signal. A possible explanation for this decrease in the muscle onset latencies was that warning facilitated a state of alertness influencing motor neuron pool excitability (Hasbroucq et al., 1999), which may have allowed postural muscle responses to be executed more quickly when the perturbation occurred. Gardner-Morse and Stokes (2001) suggest that earlier execution of muscular responses following sudden loading has several advantages over preparatory co-activation, as there is less metabolic energy consumption and compressive forces acting on the spine are likely to be lower.

Despite decreases in onset latencies muscle phasing patterns remained the same irrespective of warning condition. One pattern that did emerge, irrespective of warning condition, was the activation of one or more abdominal muscles prior to both posterior trunk muscles. This is consistent with studies involving sudden loading in a neutral standing posture (Cresswell et al., 1994; Henry et al., 1998). In an upright posture there is little or no contribution of posterior ligamentous structures to resist anterior loads (McGill, 1988). Hence, the early activation of abdominal muscles may serve to increase intra-abdominal pressure, potentially stiffening and stabilising the lumbar spine (Bartelink, 1957; Cresswell et al., 1994). A second consistent phasing pattern observed during this study was the shorter onset latency of UES when compared to that of LES. A possible advantage of activating UES earlier is that UES spans multiple segments of the lumbar spine and, because of its length and increased distance from the spinal vertebrae when compared to LES, has greater potential to resist bending moments imposed on the spine (Macintosh and Bogduk, 1987).

With regard to joint displacements, the provision of a warning signal decreased onset latencies for all joints and altered joint phasing patterns. During sudden unexpected loading there was relative simultaneous initiation of all joints, and only the knee joint onset latency was significantly faster than that of the lumbar spine. For the warning condition, subjects switched to a strategy whereby joint motion was primarily initiated at the knee (extension) and all lower limb motion occurred prior to that of the lumbar spine. The early initiation of knee extension, combined with hip flexion and ankle plantar flexion during forward trunk flexion will act to minimise anterior displacement of the body's centre of gravity (Alexandrov et al., 1998). Following initial knee extension, subjects then switched to knee flexion whilst continuing to flex the hip and lumbar spine. The addition of a warning signal also served to decrease the maximal amplitude of hip and knee flexion. Perhaps the earlier initiation of muscular onset latencies and joint movement in the warning condition enabled the system to stiffen more quickly, reducing joint excursion. This reduction in joint motion could potentially decrease motion of the centre of gravity, enabling equilibrium to be maintained more efficiently.

Fatigue

Exposure to lifting-induced fatigue resulted in no significant change in baseline muscle activity prior to sudden loading. These findings are in contrast to those of Granata et al. (2001) who reported small, but significant increases in levels of preparatory contra-lateral muscle activity of the oblique and ES muscles following a static crate holding task to fatigue. However, unlike the current study, Granata et al.'s subjects adopted forward flexion (45 degrees) of the trunk when holding the box, which may have led to a significant increase in trunk muscle activation to resist the additional bending moment generated by the weight of the flexed trunk, when compared to an upright trunk posture (Peach, Sutarno, and McGill, 1998).

In the current study fatigue was found to have a minimal effect on muscle onset latencies and no effect on joint onset latencies in response to sudden loading. These findings differed from those of Wilder et al. (1996) who reported an increase in ES muscle onset latency times of 20 ms after prolonged exposure to seated vibration. It is important to note that using vibration as a fatiguing task can induce other neurological effects. For example, vibration of specific tendons can lead to increased peripheral drive to motor neurons, in turn influencing medium latency responses to perturbation (Bove et al., 2003). As Wilder et al. (1996) provided no indication of local muscle fatigue it is difficult to know whether the increase in ES onset latency was due to neurological stimulation associated with vibration or that arising directly from the muscle fatigue. A more recent study by Granata et al. (2004) who used a lifting protocol to induce fatigue reported a decrease in the MDF of the ES, providing evidence of local muscle fatigue. Following sudden loading, these authors found that there was no alteration to the response latencies of anterior and posterior trunk muscles. These findings are also supported by a subsequent study by Herrmann et al. (2006) who used a combination of dynamic and static back extension exercises to fatigue subjects and reported no changed in LES (L4) muscle response times to sudden loading following the fatiguing protocol.

The results from this study showed that the LES and EO were the only muscles that significantly reduced their onset latencies after exercise, irrespective of warning or whether subjects were fatigued. This reduction in muscle onset latencies can be explained in two ways. Firstly, although there has been no report of decreased muscle onset latencies following fatigue in sudden external loading studies (Granata et al., 2004; Herrmann et al., 2006) it has been reported that during sudden voluntary unilateral arm raising tasks that the onset latencies of some postural muscles are shortened following a static and dynamic fatiguing task, when compared to performing the same task in a non-fatigued state (Allison and Henry, 2002; Strang and Berg, 2007). It was suggested that during voluntary arm raising the shorter postural muscle onset latencies exhibited after fatigue was a response of the central nervous system to compensate for decreased force generating capability of the muscle following fatigue (Allison and Henry, 2002; Strang and Berg, 2007). This explanation is unlikely in our study, as control subjects also showed a similar decrease in EO and LES muscle onset latencies following gentle exercise. A second and more probable reason may have been a warm-up or practice effect resulting from the exercises. Both exercise groups performed whole body exercises and although the experimental group displayed local muscle fatigue, the peak HR during the lifting task was on average 66% of their predicted maximal HR. Whilst the effects on muscle latencies is not fully understood, it has been reported that performing aerobic activity of between 50%-70% of a person's VO_2 max can decrease voluntary movement initiation times (Davranche, Burle, Audiffren, and Hasbroucq, 2005; McMorris, Delves, Sproule, Lauder, and Hale, 2005). It would seem that the repetitive lifting task performed by our subjects primarily produce fatigue at a local muscle level (as indicated by a decrease in UES and LES MDF), but had minimal effect on the ability of the central nervous system to initiated muscle responses to sudden loading.

Following exercise, the experimental group decreased the amplitude of joint motion at the knee and hip in response to sudden loading. This was unlikely to be due to fatigue, as the control group also significantly decreased the magnitude of lower limb motion. Therefore, it appears more likely that reduced lower limb ranges of motion were brought about by a learning effect, with subjects learning to decrease excursion of their body's centre of gravity.

The findings from this study showed that following fatigue knee, hip and lumbosacral displacements were unaltered by warning. In contrast, following gentle exercise the control group reduced their maximal knee, hip and lumbosacral displacement in response to warning when compared to no warning. Granata et al. (2004) also showed that following fatigue-induced lifting there was no change in the range of lumbosacral motion when exposed to sudden loading. However, some caution is needed when interpreting these findings as Granata et al. had no control group against which to make comparisons. Thus based upon the findings from the current study, it would appear that fatigue influences the ability to control joint motion and any potential benefits afforded by prior warning of sudden loading are negated.

CONCLUSION

The findings from this study showed that a visual warning signal prior to sudden loading resulted in earlier initiation of postural muscle responses and a switch in joint initiation strategy from a simultaneous movement pattern to one that is initiated at the knee joint. This strategy, in combination with pre-activation of inter-segmental lumbar stabilisers, may help reduce subsequent joint motion and maintain a state of stability more effectively. When fatigued, the potential benefits of warning for reducing lower limb and lumbosacral range of motion may be compromised.

Chapter Five

The effects of self-selected lifting posture on ratings of perceived exertion, heart rate, and erector spinae muscle fatigue during high frequency repetitive lifting to exhaustion.

PRELUDE

Whilst Chapter Three and Chapter Four focused on a single sudden loading event, Chapter Five focuses on the subjective and physiological demands during highly repetitive lifting and lowering over time. Of particular interest was the effect of self-selected lifting posture (squat, stoop or mixed) on RPE and HR responses, and ES muscle fatigue throughout a constant-load lifting and lowering task to fatigue. Accordingly, RPE, HR, and measures of ES muscle fatigue were recorded before, during and at the end of the task and comparisons were made between different self-selected lifting postures.

INTRODUCTION

The association between repetitive lifting and LBI has been well documented (Frymoyer et al., 1983; Garg and Moore, 1992; Kelsey et al., 1984; Magnusson et al., 1990; Marras et al., 1993). When performed at a high repetition rate (frequency), lifting significantly increases the risk of injury (Chaffin and Park, 1973) which may be due to physiological demands exceeding those that the individual can adequately sustain. In order to minimise physiological costs of repetitive lifting attention has often focused on lifting techniques (i.e. squat vs stoop vs mixed or semi-squat). Evidence indicates that when lifting a constant load at low frequency (less than six lifts per minute) RPE is higher for the squat lift when compared to stoop or semi-squat techniques (Rabinowitz et al., 1998; Straker and Cain, 1999). During high frequency lifting (greater than 16 lifts per minute) no differences in overall RPE have been found for different lifting postures (Hagen et al., 1993; Hagen et al., 1994). However, when trunk and lower limb exertion are assessed independently and compared for high frequency stoop and squat lifts, lower back RPE has been shown to be higher for stoop and lower for the squat lift (Hagen et al., 1993). The influence of lifting posture on HR appears to be more variable. It would seem that HR is higher during low frequency squat lifting when compared to stoop lifting (Rabinowitz et al., 1998; Revuelta et al., 2000). At higher lifting frequencies, Welbergen et al. (1991) found no difference in the effect of lifting posture on HR, whereas others have reported higher HRs for the stoop when compared to the squat lift (Hagen et al., 1993; Hagen et al., 1994; Revuelta et al., 2000).

Temporal changes in RPE and HR are considered important when investigating the physiological demands of repetitive tasks (Herman et al., 2003; Koga et al., 1999). Evidence from non-lifting exercise indicates that failure to maintain physiological and subjective steady state with constant sub-maximal workloads will lead to a gradual rise (slow component) in HR and RPE (Herman et al., 2003). When a steady state is not observed, the profile of HR and RPE appears to have two distinct phases, an initial rapid rise that is followed by a more gradual but sustained increase in these variables. To date, few studies have investigated the effects of different postures during repetitive lifting on temporal measures of RPE and HR. Observations of repetitive lifting (5 lifts per minute) by Rabinowitz et al. (1998) led them to conclude that subjects using a stoop lift reach a steady state HR, whereas those employing a squat technique fail to attain a steady state.

One proposed mechanism for the slow component rise in HR and RPE during repetitive activities involving a constant load is local muscle fatigue (Yano, Yunoki, and Ogata, 2001). Studies that have investigated local muscle fatigue during repetitive lifting have often focused on the ES muscle group (Bonato et al., 2003; Dolan and Adams, 1998). Whilst these studies have identified fatigue in some portions of the ES muscles they have not examined the effects of different lifting techniques on fatigue, nor have they assessed RPE and HR profiles in relation to muscle fatigue, particularly during high frequency lifting scenarios which are observed in occupations such as air cargo handling, packaging and distribution (Mital et al., 1994). Such work provides an insight into the temporal changes of commonly used physiological and perceptual measures during repetitive lifting. Given the conflicting findings relating to different postures and the lack of information concerning temporal changes in physiological and psychophysical variables during repetitive lifting, this study investigated the effects of self-selected lifting posture on HR, RPE and ES muscle fatigue during a high frequency repetitive lifting to exhaustion. Information of this kind was considered important if the physiological demands and the effects of fatigue arising from manual handling tasks are to be more clearly understood.

METHODOLOGY

Subjects

Thirty five healthy male volunteers participated in the study. Demographic and anthropometric measures (age, height, weight, body mass index) were also recorded (see Table 5.1 for demographic details). No subjects had experienced an episode of low back pain within a period of twelve months prior to commencing the study. In order to assess current levels of activity subjects were requested to complete a Habitual Activity Questionnaire (Baecke, Burema, and Frijters, 1982), from which indices of work, sport, and leisure activity were calculated (see Appendix 5). Written and verbal explanations of experimental procedures were provided in accordance with the requirements of the Auckland University of Technology (AUT) Ethics Committee, and subjects gave written consent prior to their participation in the study (see Appendices 3 and 4 for Participant Information Sheet and Subject Consent Forms).

Experimental protocol

Procedure

Subjects were required to undertake a repetitive lifting task to exhaustion. This required the subjects to lift and lower an aluminium box (20 cm in height by 23 cm wide by 25 cm deep, with handles 15 cm from the base of the box) loaded with a 10 kg weight (13 kg total mass) to and from a platform 15 cm above the floor at a lifting rate of 20 lifts per minute. A 13 kg mass was considered an acceptable weight and common to highly repetitive lifting and carrying tasks (Mital et al., 1994).

Subjects began the lifting task in an upright standing posture with the box at hip height. The box was initially lowered onto the platform before being lifted back to an upright body position. Subjects repeated the task at a rate of 20 lifts per minute, the rate of lifting being controlled by a metronome positioned close to the subjects. On one beat of the metronome each subject rested the box on the platform, and within the next beat returned the box to hip height. Subjects repeated the task until they reached a state of fatigue, adjudged to have occurred when subjects felt such excessive discomfort or subjective fatigue that stopped them from lifting, or they were unable to keep pace with the metronome (Mayer et al., 1988). Prior to the lifting task, subjects were given a period of familiarisation which involved practising lifting and lowering the empty box in time with the metronome. Once comfortable with the task, subjects then rested for approximately five minutes before undertaking repetitive lifting to fatigue.

Lifting posture

A digital video camera (JVC GR-DVL 9800) was used to record lifting postures in the sagittal plane throughout the duration of the lifting task. Two experienced physiotherapists were required to visually analyse each subject’s video recordings in order to classify lifting postures. Postures were classified into three common lifting categories - squat; stoop; and mixed - according to the definitions proposed by several authors (Burgess-Limerick and Abernethy, 1997; Burgess-Limerick et al., 2001; Hagen et al., 1994; Revuelta et al., 2000; Welbergen et al., 1991). A squat lift was defined as a lift in which the knees were bent (flexed) and the back was straight (erect). A stoop lift was defined as a lift where the knees were straight and the back was bent (flexed). It should be noted that these definitions although commonly utilised (Burgess-Limerick and Abernethy, 1997; Burgess-Limerick et al., 2001; Hagen et al., 1994; Revuelta et al., 2000; Welbergen et al., 1991), do not stipulate the degree of lumbar curvature at the initiation of the lift. Those subjects who were considered not to adopt a clearly defined squat or stoop posture were classified as having a mixed lifting technique.

Lifting posture assessment was conducted on four consecutive lifts occurring shortly after the initiation of the repetitive lifting task and four consequitive trials at the end of the lifting task. An assessment of inter-rater reliability between the two physiotherapists in assessing lifting posture revealed good agreement, with a Kappa statistic of 0.723 (P<0.001). If the two therapists identified lifting postures that were not in agreement, a third therapist was required to determine posture. In all instances at least two physiotherapists reached agreement over the lifting technique.

Ratings of perceive exertion and heart rate

RPE and HR (recorded via a polar 625x HR monitor, Polar Electro Oy, Finland) were recorded at rest, at the end of each minute during the lifting task, and on termination of the task. HRs were expressed as a percentage of predicted maximum HR (percentage HR max) ([HR lift ⁄ (220 – age)] x 100). Subjects perceived exertion was determined using Borg's RPE scale (Borg, 1982). This requires subjects to rate their perceived exertion on a scale from 6 to 20, where 6 is "very, very light" and 19 is "very, very hard". In addition to RPE subjects were asked whether they felt discomfort, and if so, the body location where most discomfort was evident; i.e. shoulders, wrists, thighs, buttocks, upper back, or lower back.

Electromyography (EMG)

In accordance with the electrode placement described by Cholewicki and McGill (1996), muscle activity of the right LES and right UES were measured using surface electrodes. Active electrodes (Delsys DE02.3, Delsys Inc., Boston) with a separation distance of 10 mm were placed directly on the skin of each subject. To facilitate good electrical conductance, the skin was shaved and cleaned with alcohol swabs. All EMG signals were pre-amplified (x 500), sampled at a rate of 1 KHz and band-pass filtered (20-450 Hz).

Prior to and immediately following the repetitive lifting task, subjects undertook three MVC's of the back extensors. This lasted for a period of 5 s during which EMG signals were recorded. All MVC's were performed with subjects in a modified version of the Biering-Sørensen position (Biering-Sorensen, 1984), such that subjects were lying prone on a plinth with their lower limbs strapped to the plinth and the trunk extending in an unsupported position beyond the edge of plinth. Each MVC involved maintaining the upper body into a horizontal position at maximal effort while a restraining force was applied to the shoulders. A two second segment of each EMG signal recorded during the MVC's was filtered using a Hanning window. A Fast Fourier Transformation was then used to convert time based data into a power spectrum. From the resulting power spectrum, the MDF of UES and LES was calculated. A decrease in MDF between pre and post lift was considered to be an indicator of muscle fatigue.

Statistical analysis

Single factor analysis of variance (ANOVA) was performed to determine the main effects of lifting posture on RPE and HR. The independent variable included in the model was posture (i.e. squat, mixed or stoop), with the dependent variables being peak HR, and RPE, and total number lifts completed when exhausted. As assumptions of normality were not valid for measures of peak RPE and time to exhaustion, differences in lifting posture for these measures were analysed using the Kruskal-Wallis test. To determine the effects of posture on localised muscle fatigue, a 3 x 2 repeated measures ANOVA was used for each muscle group (UES and LES); the between subject factor being posture (squat, mixed or stoop) and the within subject factor being time (before vs after the lifting task).

To investigate the effects of posture on RPE and HR throughout the lifting task, RPE and percentage HR max for each subject over the duration of the lifting task were plotted using spaghetti plots and Lowess curves (Dupont, 2002). A Lowess curve is a nonparametric mean function estimator. This enabled the variability and patterns of RPE and HR to be assessed over time, and to determine whether any influential or aberrant data points existed. As the sparseness and variability of the data increased with time, both untransformed and logarithmically transformed time variables were assessed using Bayesian Information Criterion (BIC) (Schwarz, 1978). The BIC is an information criterion which considers both the complexity of non-nested models and their goodness-of-fit to the data. The preferred model balances these competing demands and culminates in the lowest BIC value. Based on the patterns identified in the Lowess curves, mixed-effects models using low dimensional polynomial functions of time were compared. As all RPE values at baseline had a common value (6), a fixed intercept random slope model was employed for analyses of this variable. As no such constraint existed for percentage HR max, a random intercept and random slope model was applied to the data. Once the mixed-effects models were established, the relationship of posture alone and its interaction over time was investigated. Residual and influence diagnostic checks were undertaken using Studentised residuals, the PRESS statistic and Cook's distance measure (Dupont, 2002). All statistical analyses were performed using SPSS version 14, SAS version 9, and Stata version 8.0 software packages. A significance level of $P<0.05$ was used to determine statistical significance for all tests.

RESULTS

Posture classification and fatigue

The therapist's assessment of video footage of lifting posture identified 13 (37%) subjects employing a squat lift, 6 (17%) using a stoop lift and 16 (46%) using a mixed technique. Analysis of anthropometric measures and activity indices (work, sport and leisure) showed no significant difference between posture classification groups, suggesting that characteristics were similar across all groups (Table 5.1).

Table 5. 1. Mean (± SD) anthropometric measures and activity indexes for subjects grouped by lifting posture (squat, mixed, stoop).

Lifting posture	N		Age (years)	Stature (cm)	Weight (kg)	Body mass index	Work index	Sport index	Leisure index
Squat	N=13	Mean (± SD)	24.23 (4.75)	177.23 (8.56)	74.62 (11.38)	23.68 (2.63)	2.41 (0.49)	3.76 (0.99)	2.69 (0..46)
Mixed	N=16	Mean (± SD)	26.13 (4.80)	177.81 (6.75)	78.04 (10.83)	24.69 (3.16)	2.35 (0.37)	3.61 (0.98)	2.95 (0.44)
Stoop	N=6	Mean (± SD)	25.67 (5.42)	184.17 (3.13)	79.58 (6.92)	23.43 (1.37)	2.31 (0.28)	3.93 (0.57)	2.95 (0.48)
Pooled	N=35	Mean (± SD)	25.34 (4.82)	179.74 (7.34)	77.41 (10.61)	23.93 (2.70)	2.36 (0.39)	3.76 (0.91)	2.86 (0.46)

All subjects stopped lifting because of subjective fatigue, with a median RPE at exhaustion of 19 and a mean peak HR of 67% of predicted maximal HR (Table 5.2). Irrespective of lifting posture, for all individuals, the low back was identified as the body region for which there was most discomfort at the end of the lifting task, and was the reason for discontinuing the task. The median time to exhaustion was similar for all lifting postures, and peak RPE and HR did not differ significantly between postures (Table 5.2).

Table 5.2. Median and mean time to exhaustion, peak RPE and percentage HR max for individuals adopting the squat, mixed, and stoop lifting postures.

Lifting posture	N	Median time to fatigue (s) (range)	Median total lifts (range)	Median peak RPE (Range)	Mean Peak %HR max (SD)
Squat	N=13	300 (180-960)	100 (60-320)	19 (17-19)	67 (9.17)
Mixed	N=16	245 (150-600)	81.67 (50-200)	19 (15-19)	64 (10.40)
Stoop	N=6	213 (180-360)	71 (60-120)	19 (19-19)	69 (13.20)

Temporal patterns of RPE and HR

For the 35 subjects, lifting time to exhaustion ranged between 150 and 960 s, with a median of 250 s. As time distributions were highly skewed with a sparseness of observations at the higher time values, time was truncated to 540 s for the mixed-effects model analyses. Figure 5.1 presents the spaghetti plots of RPE and percentage HR max for the 35 subjects, according to lifting posture (squat, stoop or mixed). Visual inspection of the graphs revealed considerable variability between subjects over time. However, the intra-subject variability was relatively small after accounting for intercept and slope components.

Figure 5.2 shows the Lowess curves of RPE and percentage HR max over time for the 35 subjects, according to lifting posture (squat, mixed or stoop). It is apparent from these curves that the relationship between RPE and percentage HR max was non-linear, showing an initial rapid increase in RPE and HR from rest followed by a slow rise in both variables over time (Figure 5.2). Therefore, 2nd order polynomials were fitted to these data.

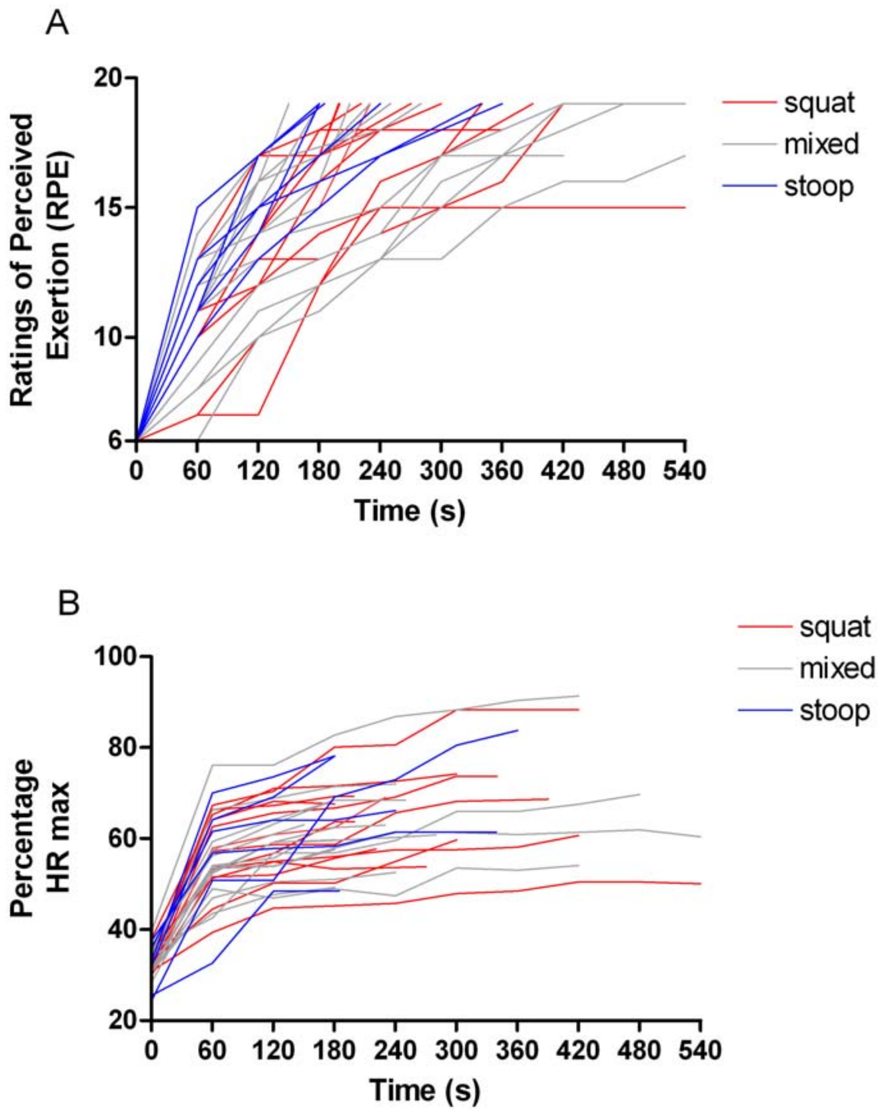


Figure 5.1. Spaghetti plots of A) RPE and B) HR responses expressed as percentage HR max for each subject and according to whether they adopted the squat, mixed or stoop posture over the duration of the lifting task to exhaustion.

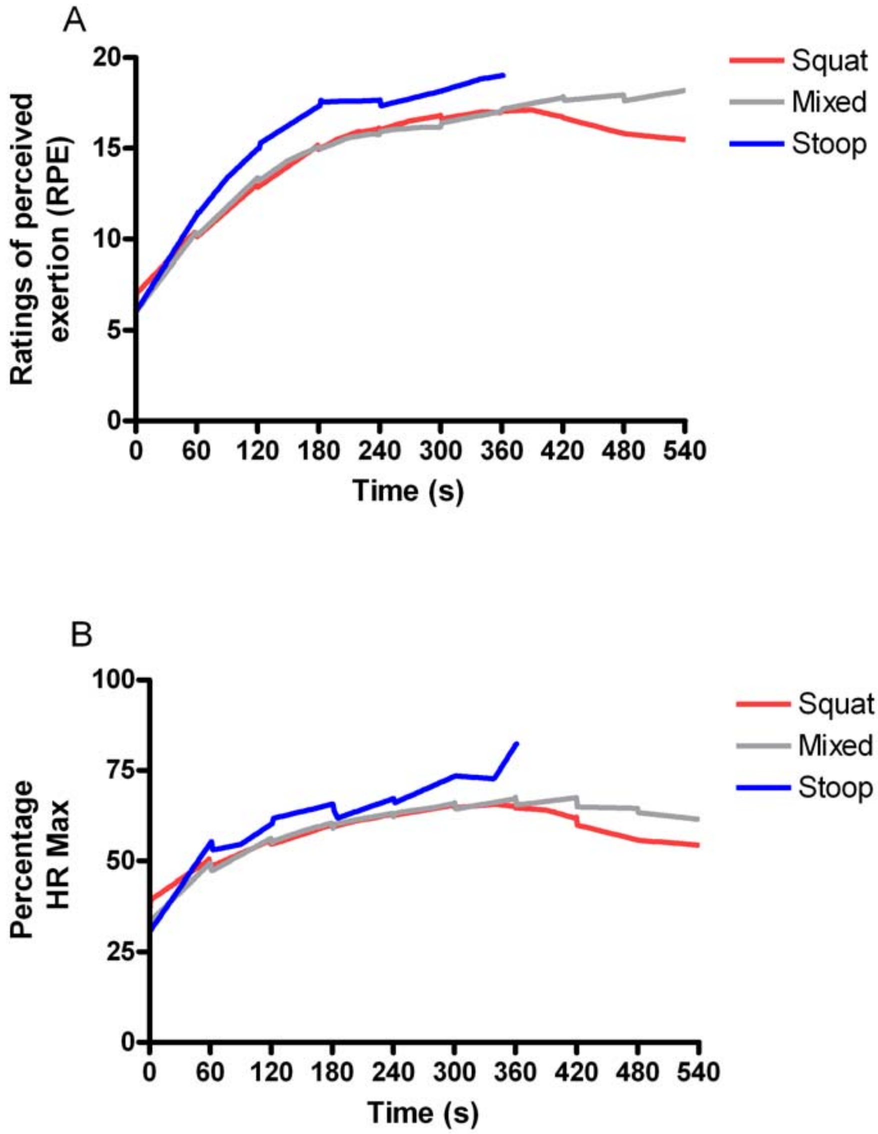


Figure 5.2. Lowess curves for A) RPE and B) HR responses expressed as percentage HR max for the squat, mixed, and stoop postures over the duration of the lifting task to exhaustion.

When fitting linear and quadratic functions over time for untransformed and logarithmically transformed time variables, both functions were statistically significant for RPE and percentage HR max. However, the BIC was superior for the logarithmically transformed time variable compared to its untransformed characterisation for both RPE (BIC: 753.7 vs 874.8) and percentage HR max (BIC:

1,114.0 vs 1,354.5) and so was used in a subsequent analysis of the effects of time and posture (Table 5.3).

Table 5.3 provides the regression coefficients and standard errors (SE) for separate mixed-effects models of posture and RPE, and posture and percentage HR max. Posture was not found to have a significant effect on the temporal patterns of RPE and HR throughout the lifting task. Moreover, there was no significant interaction between posture and time. Residual and diagnostic checks using Studentised residuals, the PRESS statistic and Cook's distance measures revealed no important violations of the mixed-effects models assumption.

Table 5.3. Regression coefficient estimates and standard errors (SE) from separate mixed-effects model analysis of posture and rate of perceived exertion (RPE), and posture and percentage HR max, after adjusting for $\log_e$(time) and $\log_e$(time)^2.

	Rate of perceived exertion (RPE)			Percentage of predicted maximal heart rate (% HR max)		
	Estimate	(SE)	*P*-value	Estimate	(SE)	*P*-value
$\log_e(time)$	-1.10	(0.19)	<0.05	4.19	(0.57)	<0.05
$\log_e(time)^2$	0.58	(0.03)	<0.05	0.30	(0.09)	<0.05
Posture						
Stoop	reference		0.85	reference		0.55
Squat	-0.16	(0.47)		1.77	(1.91)	
Mixed	-0.26	(0.46)		2.00	(1.89)	
Posture x $\log_e(time)$						
Stoop	reference		0.43	reference		0.49
Squat	-0.24	(0.20)		-0.77	(0.89)	
Mixed	-0.24	(0.19)		-1.05	(0.88)	

Figure 5.2 also shows that the Lowess curves for the stoop posture may be different from squat and mixed postures. Hence, in an effort to increase statistical power, all the analyses contained in Table 1 were repeated using this dichotomous variable for

posture rather than the original trichotomous characterisation. In the subsequent analysis of RPE, posture (P=0.62) and the interaction between posture and time (P=0.19) remained non-significant. Similarly, for HR max, posture (P=0.27) and the interaction between posture and time (P=0.26) were non-significant.

Localised muscle fatigue

Figure 5.3 shows the changes in MDF of UES and LES when performing the MVC before and after the lifting task. ANOVA showed a main effect for fatigue (P<0.05), but no significant effect for posture. Both UES and LES showed significant decreases in MDF following repetitive lifting (P<0.05).

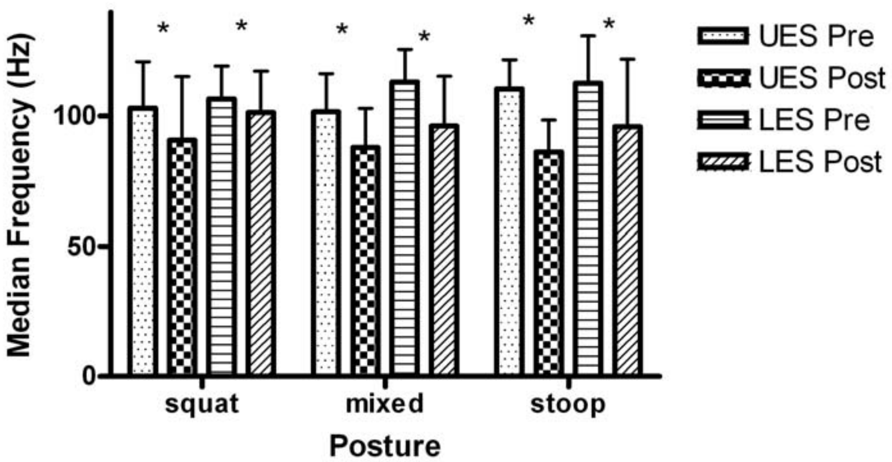


Figure 5.3. Median frequency (MDF) values for UES and LES during the performance of MVC in the modified Biering-Sørensen position before (pre) and after (post) repetitive lifting to exhaustion. * P<0.05.

DISCUSSION

This study showed that HR and RPE responses to fatigue were not affected by self-selected lifting posture throughout the lifting task to exhaustion. Furthermore, the lifting posture did not affect time to exhaustion and maximal RPE at exhaustion. Hagen et al. (1994) also found no difference in RPE for the squat and stoop lifting postures when subjects lifted a variety of weights over a range of different lifting frequencies at sub-maximal effort and to exhaustion. In contrast, Rabinowitz et al. (1998) and Straker and Cain (1999) found higher RPE for the squat lift when compared to the stoop (Rabinowitz et al., 1998) and semi-squat techniques (Straker and Cain, 1999) when repetitive lifting was performed at sub-maximal effort. The differences in these findings and those of the current study may be partly attributed to the higher lifting frequencies used in the current study (20 lifts per minute) compared to those of Rabinowitz et al. (1998) and Straker and Cain (1999). Furthermore, in contrast to the current study Rabinowitz et al. (1998) and Straker and Cain (1999) instructed subjects on the different lifting techniques to adopt. Thus, subjects may have been unaccustomed to these lifting styles which may have affected their perceptual responses. In this regard, Kumar (1984) reported that subjects performing squat, stoop and self-selected techniques during a variety of lifting activities rated the self-selected freestyle lift as the least tiring lifting technique on 75% of the occasions. Garg and Saxena (1979) also showed that when subjects adopted a self-selected lifting technique the maximal acceptable weight lifted (MAWL) was heavier than when performing an instructed squat or stoop lifts.

All subjects stated that local back discomfort was the main reason for terminating the lifting task, irrespective of lifting posture. This is in agreement with the findings of

Kell and Bhambhani (2006) and Fernandez et al. (1991) who also found that following repetitive lifting to fatigue subjects reported the highest regional specific RPE in the back (Kell and Bhambhani, 2006) and identified low back soreness as the reason for terminating a lifting task lifting prematurely (Fernandez et al., 1991).

The termination of the lifting task due to back discomfort may be related to muscle fatigue as there was found to be a significant decrease in the MDF of UES and LES muscles following the lifting task. This was irrespective of posture exhibited during the lifting task. Other lifting studies have shown fatigue of the mid lumbar portion ES muscle group following repetitive lifting (Bonato et al., 2003; Dolan and Adams, 1998). The most probable explanation for UES and LES fatigue is due to a decrease in blood flow to, and thus oxygenation of the ES muscles (Kell and Bhambhani, 2006) contributing to the accumulation of metabolites (Brody, Pollock, Roy, De Luca, and Celli, 1991) and alteration in pH (Vestergaard-Poulsen et al., 1995).

As with RPE, maximal HR did not differ according to lifting postures. These findings were similar to those of Welbergen et al. (1991) who also found HR to be similar for the squat and stoop posture when repetitively lifting at rates up to 20 lifts per minute. In contrast, however, Hagen et al. (1993) and Rabinowitz et al. (1998) found significantly higher HRs when lifting the same weight and frequency in the squat posture when compared to the stoop posture at sub-maximal and maximal efforts. These differences in findings may again be due to methodological differences in which subjects in the Hagen et al. (1993) and Rabinowitz et al. (1998) studies were instructed to use a novel lifting technique. It has been shown that novel tasks demand greater VO_2 and elicit higher HRs when compared to well-practiced tasks (Lay, Sparrow, Hughes, and O'Dwyer, 2002; Lay, Sparrow, and O'Dwyer, 2005). Consequently, instructing individuals about how to lift, particularly for unaccustomed tasks, may influence physiological demands and corresponding HR responses.

Peak HRs for all subjects did not reach their predicted maximum and were on average 67% of their predicted HR max at exhaustion. Other studies have also reported significantly lower maximum HR and VO_2 for maximal lifting exertion when compared to maximal effort during cycling, stepping and treadmill running (Asfour, Genaidy, Khalil, and Muthuswamy, 1986; Hagen et al., 1993; Nindl et al., 1998; Petrofsky and Lind, 1978a). This may be due to a number of reasons. Firstly, it has been shown that despite increased ventilation during lifting, the accessory muscles (abdominal and intercostal muscles) involved in breathing already have an active role specific to the lifting task and their ability to contribute to respiration may be compromised, ultimately decreasing cardiorespiratory capacity (McGill, Sharratt, and Seguin, 1995). Secondly, lifting may reduce maximal cardiorespiratory capacity due to increases in intra-abdominal pressure (IAP), inhibiting venous return and reducing cardiac output (Nindl et al., 1998). Studies involving animals have shown that increased IAP can reduce cardiac output, although the effects on HR are equivocal (Vivier et al., 2006). Thirdly, lifting involves both static and dynamic (rhythmical) components (Petrofsky and Lind, 1978a, 1978b). In this respect, Petrofsky and Lind (1978b) found that during lifting at a low level of aerobic demand (work rate inducing 25% of VO_2 max) muscle groups associated with static components of the lift exhibited significantly decreased levels of endurance capacity (20% to 30%). Saito and Mano (1991) also showed that when comparing dynamic and static exercise at the same workload, there was an increased sensation of local fatigue for static exercise. This may be due to restricted blood flow to key muscles during repetitive lifting (Kell and Bhambhani, 2006) and the accumulation of metabolites in those muscles working statically. Also afferent somatosensory activity during dynamic muscle contraction has been shown to have greater influence on HR when compared to static muscle contractions (Saito and Mano, 1991; Smith et al., 2003).

Temporal patterns of HR and RPE did not differ between each lifting posture. Mixed model analysis showed an initial rapid rise in RPE and HR for the first 100 - 120 s of the lifting task followed by a gradual increase in both variables until exhaustion. This pattern of initial rapid increase followed by slow component rise without attainment of steady state has been well documented for RPE, HR and VO_2 responses during sub-maximal constant heavy load exercise (above lactate threshold but below VO_2 max) (Barstow and Mole, 1991; Herman et al., 2003). It has been suggested that the slow component rise may be due to factors associated with fatigue, such as increases in blood lactate, elevation of muscular temperature (Gaesser and Poole, 1996) and increased oxygen cost associated with recruitment of fast twitch fibres which are less efficient aerobically compared to slow twitch fibres (Barstow et al., 1996; Krustrup et al., 2004). Whether fatigue can be prevented by adjusting workload to accommodate steady state RPE and HR was beyond the scope of the current study but requires further investigation. There is evidence that during cycling at high intensity sub-maximal workloads adjustments to external work load in order to maintain steady state RPE can significantly reduce and even eliminate the slow component rise in HR and VO_2 (Herman et al., 2003), leading to decreased energy expenditure and potential fatigue. Fernandez et al. (1991) has shown that during moderate lifting rates (8 lifts per minute) when individuals are allowed to continually adjust the MAWL, subjects are able to maintain a steady state HR throughout an eight hour period. In contrast, when no adjustments were made to the initial MAWL, Fernandez et al. observed increases in HR over time with most subjects (9/16) terminating the lifting task within 5 hours (Fernandez et al., 1991).

CONCLUSION

For a repetitive lifting task self-selected lifting posture was not found to affect RPE, HR, and the degree of local ES muscle fatigue. However, the number of subjects who were stoop lifters was low and hence the ability to detect a difference was compromised. Temporal patterns of HR and RPE during the lifting task were similar for all postures and did not attain a steady state. This highlights the importance of continuously monitoring of RPE and HR if fatigue during repetitive lifting activities is to be prevented. Measures of local back discomfort and associated local back muscle fatigue appear to be important determinants of the likelihood of fatigue during high repetitive lifting.

Chapter Six

Lumbosacral kinematics and trunk muscle activation patterns during highly repetitive lifting: effects of initial self-selected lifting posture and fatigue.

PRELUDE

In Chapter Five it was found that RPE and HR responses were similar between self-selected lifting postures showing a slow incline until exhaustion. In addition, maximal HR responses indicated that excessive cardiovascular demand did not seem to be the limiting factor for terminating a high frequency lifting task. However, subjective and objective EMG measures of ES muscle fatigue indicated that local neuromuscular control of lumbar spine motion may be an important factor for maintaining work output during highly repetitive lifting. This led to the final study which focused on the effects of self-selected lifting posture and fatigue on neuromuscular control of lower lumbar spine kinematics during high frequency repetitive lifting. Accordingly, lumbosacral kinematics and trunk muscle activation patterns were recorded during lifting and lowering using a self-selected posture at the beginning and end of a high frequency repetitive lifting task to exhaustion.

INTRODUCTION

Lumbosacral kinematics and their associated levels of trunk musculature activation are important issues to consider with respect to developing safe lifting techniques and reducing the likelihood of injury during manual handling activities (Davis and Marras, 2000; Marras et al., 1995; Marras et al., 1993; Norman et al., 1998). It is evident that during repetitive lifting tasks, individuals will adopt one of three main types of lifting technique. These have been termed squat, stoop, and intermediate/mixed (Bonato et al., 2003). A number of studies have highlighted the positive and negative elements of each of these lifting techniques (Potvin et al., 1991; van Dieen et al., 1999). Where studies have compared these lifting techniques, subjects have been provided with instruction on how to undertake each lift. Such a strategy is problematic, as the individual is not likely to be fully attuned to all of the techniques (Hart et al., 1987). That is, they may not be performing one or all of the tasks efficiently from a physiological or technical perspective due to a lack of sufficient practice in all techniques. Thus, it would be advantageous to compare groups of subjects who typically utilise one specific type of lifting technique as part of their daily work and leisure activities.

Those studies which have used instructed lifting techniques have often focused on the initial stages of the lift where the lumbosacral velocity is typically low and the extensor moment is at its greatest (Potvin et al., 1991). During this phase of the lift, subjects who are instructed to perform a stoop lifting technique flex the lumbar spine to a greater extent than that those who are instructed to perform a squat lift (de Looze et al., 1993; Dolan, Mannion et al., 1994; Potvin et al., 1991). Also, LES levels of muscle activation have been reported to be lower in the stoop lift when compared to the squat lift (Hart et al., 1987; Potvin et al., 1991; Vakos et al., 1994). The extent of lumbosacral flexion is important during repetitive lifting because as the lumbar spine approaches 80% of maximal flexion there is a rapid increase in resistance to the bending moment from the passive tissues of the lumbar spine (Dolan, Mannion et al., 1994; Potvin et al., 1991). Furthermore, the ability of the ES muscle group to resist anterior shear forces is compromised near end range lumbar flexion (Macintosh et al., 1993; McGill et al., 2000). Whilst research focusing on the initiation of the lift has provided useful information, examination of lumbosacral kinematics and associated trunk muscle activation throughout the remainder of the lifting and lowering cycle may provide a better understanding of the neuromuscular control of lumbar motion, and allow comparisons across self-selected lifting postures. In doing so, this might provide insight into the likelihood of injury associated with lifting postures.

Another factor that may influence lumbar motion and trunk muscle activation when lifting and lowering is fatigue. Studies that have not differentiated lifting technique have indicated that when a person becomes fatigued, they revert to a more stoop technique, increase lumbar spine flexion (Dolan and Adams, 1998; Sparto, Parnianpour et al., 1997a), and reduce lumbosacral velocity at the beginning of the lift (van Dieen et al., 1998). In contrast, others have found little change in lifting posture (Bonato et al., 2003), lumbosacral velocity (Sparto, Parnianpour et al., 1997a), and decreases in peak lumbar flexion (Marras and Granata, 1997) over time. The amount of lumbar motion used during a particular lifting technique may also influence lumbar spine kinematics over time. For example, isokinetic tasks that require large ranges of lumbar flexion and extension exhibit decreases in lumbosacral velocity and a reduction in lumbar spine range of motion (sagittal plane) over time (Parnianpour et al., 1988). Such changes in lumbar spine kinematics may be associated with the findings that ES muscle fatigue may be more pronounced in individuals who use a substantial amount of spinal flexion when lifting (Potvin and Norman, 1993).

Given the above considerations there were two aims of the current study. The first aim was to examine differences in lumbosacral kinematics and associated trunk muscle activation patterns of individuals adopting self-selected postures (squat, mixed or stoop) across the lifting and lowering cycle. The second aim was to examine the influence of lifting-induced fatigue on lumbosacral kinematics and trunk muscle activation patterns throughout the lifting and lowering cycle.

METHODOLOGY

Subjects

Thirty-one male volunteers with a mean age of 25 years (SD = ± 4.9 years), mean mass of 76 kg (SD = ± 11.2 kg) and mean height of 1.78 m (SD = ± 0.08 m) gave written consent to participate in the study. All subjects were in good health and had no previous history of low back, hip, knee or ankle injury within twelve months prior to the study. Written and verbal explanations of the experimental procedures were provided in accordance with the requirements of the Auckland University of Technology Ethics Committee (see Appendices 3 and 4 for Participant Information Sheet and Subject Consent Forms).

Experimental protocol

Lifting task

Firstly, subjects performed a five minute warm up of general callisthenic exercises and three static MVC trials were completed for each trunk muscle group. Retro-reflective markers were then attached to each subject and lumbosacral motion measures were taken in upright standing and maximal flexion in standing. Prior to the lifting task, subjects practised lifting an empty box (3 kilograms) to the beat of a metronome. The lifting task required subjects to lift and lower a hand-held box (13 kg in total) from a raised platform (15 cm in height) that was located directly in front of the subject's feet to waist height, to the beat of a metronome set at a rate of 20 lifts per minute. Subjects were provided with no instruction about a preferred lifting technique and they were required to continue lifting and lowering the box until they either felt excessive discomfort or exhaustion, or they were unable to keep pace with the metronome (Mayer et al., 1988). On termination of the task subjects were asked to rate their level of perceived exertion using the Borg rating scale of perceived exertion (RPE) (Borg, 1982).

Electromyography (EMG)

Surface EMG signals were recorded using active electrodes (Delsys DE02.3, Delsys Inc., Boston) with an inter-electrode distance of 10 mm. Prior to application of the electrodes the skin was shaved and cleaned with alcohol swabs and electrodes were applied. Electrodes were placed on the skin surface of the trunk superficial to the muscle fibres of the UES (5 cm lateral to the T9 spinous process), LES (approximately 2 cm lateral to the L4-5 interspinous space), RAB (3 cm from the midline just above the umbilicus), IO (midway between the anterior superior iliac spine and symphysis pubis, superior to the inguinal ligament) and EO (15 cm lateral to the umbilicus) (Cholewicki and McGill, 1996). EMG signals from the five muscle groups were collected at a sample rate of 1000 Hz and band-pass filtered (20-450 Hz). EMG activity during lifting and lowering was normalised to EMG recorded during a static maximal voluntary contraction (MVC). The MVC of the ES muscles was performed in the modified Biering-Sørensen position, with the subjects lying on a plinth in a prone position with the trunk unsupported and the lower limbs strapped to the plinth (Biering-Sorensen, 1984) whilst maximal manual resistance was applied at the T6 level. The MVC for RAB was recorded with the feet strapped to the plinth and the trunk flexed to approximately 30 degrees whilst maximal manual resistance was applied against trunk flexion. The IO and EO MVCs were recorded in the same degree of trunk flexion as RAB, but the trunk was rotated to the left and the right while manual resistance was applied. The highest RMS value over a two second epoch was chosen for each of the oblique abdominal muscles (White and McNair, 2002).

In order to assess local muscle fatigue, EMG signals from the UES and LES were recorded during a static MVC in the modified Biering-Sørensen position before and immediately after the termination of the lifting task. A two second segment of EMG data from the MVC trials was filtered using a Hanning window. Fast Fourier transformation was applied to the resulting signal to derive a power spectrum, and MDF. The MDF's for each MVC prior to and after the lifting task were compared to assess whether local muscle fatigue had occurred.

Kinematic analysis

Two pairs of retro-reflective markers were attached to the skin surface covering the first lumbar (L1) and sacral (S1) levels. A digital video camera (JVC GR-DVL 9800) was located perpendicular to the sagittal plane of motion, four metres from the subject and recorded marker motion at a sampling rate of 120 frames per second. Digital video data were recorded continuously throughout the lifting and lowering task. Two lifting and lowering cycles from at the start (between lifts 6-10) and end of the trial (within the last 10 lifts) were used in the subsequent analysis.

Motion analysis digitising software (ARIEL Performance Analysis System, San Diego, USA) was used to digitise movement of the lumbar and sacral (x and y co-ordinates). Individual marker displacements were low pass filtered at 6 Hz and used to calculate lumbar curvature using the method developed by Adams et al. (1986). Lumbar curvature (lumbosacral motion) was expressed as a percentage of the subject's full range of lumbar motion using the calculations developed by Dolan, Mannion et al. (1994). Briefly, the full range of lumboscral flexion was established as the difference in lumbosacral angle (LSA) recorded during upright standing subtracted from the lumbosacral angle recorded during extreme toe touching (Dolan, Mannion et al., 1994). Subsequent lumbosacral motion during the lifting and lowering trial was expressed as a percentage of the subject's maximal lumbosacral flexion using the following formulae (Dolan, Mannion et al., 1994):

Percentage of lumbosacral flexion = 100 x {[LSA (lifting and lowering) - LSA(standing)] / [LSA(full flexion)-LSA(standing)]}

Quantifying posture

Posture was quantified using a modification of the postural index developed by Burgess-Limerick and Abernethy (1997). Briefly, retro-reflective markers attached to the skin surface overlying the base of the fifth metatarsal, lateral malleolus, knee joint line, greater trochanter and sacrum, were used to calculate ankle, knee, and hip angles of the right lower limb. These angles were used to calculate the postural index during the initiation of the lift. The postural index equates to the amount of knee flexion relative to normal standing divided by the sum of ankle, hip, and lumbar flexion relative to normal standing (Burgess-Limerick and Abernethy, 1997). A postural index closer to 1 is representative of a more squat lifting posture, whereas a postural index closer to zero is more representative of a stoop lifting technique (Burgess-Limerick and Abernethy, 1997).

Phasing of kinematic and EMG data

EMG data were synchronised with kinematic data via a switch that illuminated a light located posterior to the subject and which sent a simultaneous 3 volt signal to the acquisition system. Data from an accelerometer fixed to the bottom of the box was collected simultaneously with the EMG collection system and used to detect initial box contact with the platform. Accelerometer data on initial contact of the box with the platform was synchronised with vertical velocity data from a marker located at the side of the box which registered zero on box contact with the platform. Pilot testing showed that the accuracy of this technique was within 8 ms. A complete lifting and lowering cycle was defined as the period from between consecutive box contacts with the platform. The lifting and lowering cycle was divided into four phases: 1) contact with the platform; 2) the lifting phase; 3) the holding phase, where the box was stationary whilst the subject was in the upright position, and 4) the lowering phase. Lifting and lower phases were further divided into three equal divisions (initial, middle, and final) and the box contact and hold periods were divided into two equal divisions (Figure 6.1).

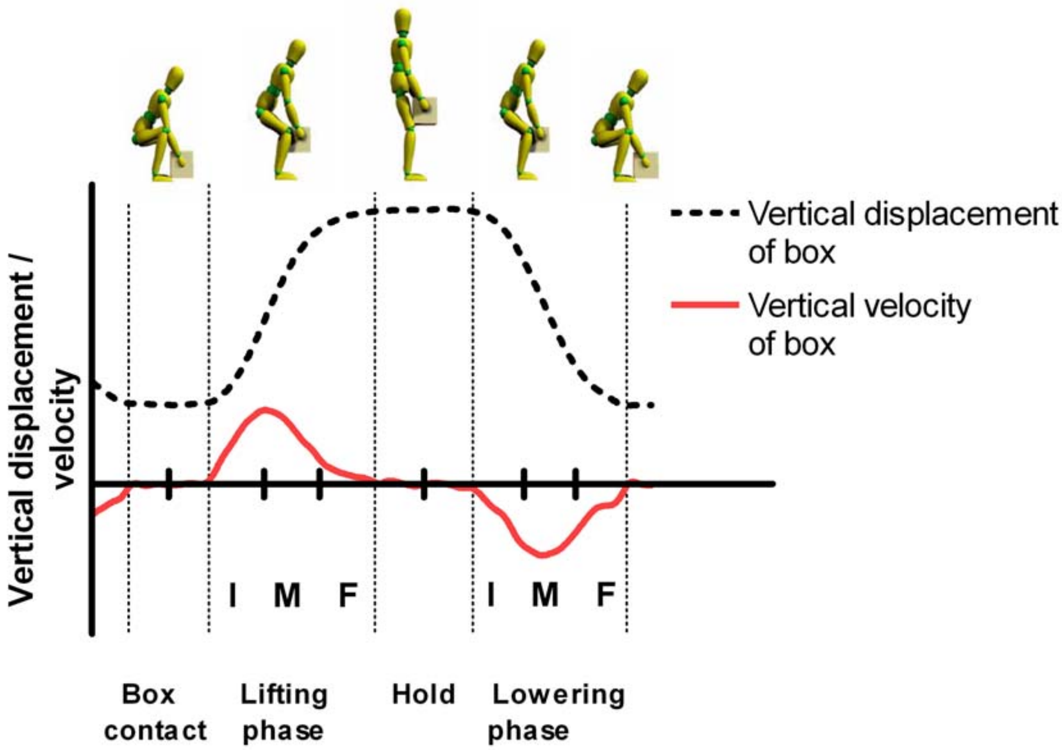


Figure 6.1. Box vertical displacement and velocity measures used to identify the four main phases of the lifting and lowering cycle (box contact, lifting, hold and lowering phase) (dotted lines). The lifting and lower phases were further divided into initial (I), middle (M), and final (F) divisions. Box contact and hold phases were divided into equal divisions for analysis.

For each of the respective divisions of each phase average lumbosacral motion and trunk muscle RMS values were calculated.

Therapist classification of lifting posture

Lifting postures were identified via visual analysis by an experienced physiotherapist of video footage of the subject lifting between lifts 6-10 and during four consecutive trials at the end of the lifting task. Postures were classified into three common lifting categories (squat; stoop; and mixed) according to the definitions proposed by a number of authors (Burgess-Limerick and Abernethy, 1997; Burgess-Limerick et al., 2001; Hagen et al., 1994; Revuelta et al., 2000; Welbergen et al., 1991). A squat lift was defined as a lift in which the knees were bent (flexed) and the back was straight (erect). A stoop lift was defined as a lift where the knees were straight and the back was bent (flexed). Those subjects who were considered not to adopt a clearly defined squat or stoop posture were classified as having a mixed lifting technique. A pilot study of inter-observer assessment of posture yielded good reliability between therapists (Kappa statistic of 0.723 (P<0.05).

Statistical analyses

To establish whether there were differences across lifting versus lowering phases, the data for divisions of these phases were collapsed. A mixed model repeated measures analysis of variance (ANOVA) was utilised to assess motion measures and muscle activity. The between subject factor was initial lifting posture (squat, mixed or stoop), and the within subject factors were phase (lift vs lower) and time (start vs end). Significant main effects were assessed using pairwise comparisons with Bonfonneri adjustments.

In order to determine differences in the kinematic and trunk muscle activation patterns across the lifting and lowering phases for the different postures, a 2 (start vs end) by 4 (divisions) ANOVA was used for each variable of interest. One ANOVA examined the differences in the divisions during the lifting phase: i.e. the second half of box contact, and initial, middle and final third of lifting. A second ANOVA examined differences in the divisions associated with the lowering phase: the second half of the hold period, and initial, middle and final thirds of lowering. Statistical analyses were conducted using SPSS for windows version 15 software package (SPSS Version 15, SPSS Inc., 2006). The alpha level was set to $P < 0.05$.

RESULTS

Lumbosacral range of motion and postural index

There was no significant difference between groups (i.e. squat, mixed and stoop) for total range of lumbosacral flexion (upright standing to full flexion) recorded prior to lifting. This range of lumbosacral flexion (54.2 degrees) was comparable with previous studies that have validated lumbar spine range of motion against x-ray findings (Dolan and Adams, 1993a; Pearcy, Portek, and Shepherd, 1984).

Of the total of 31 subjects who participated in the study, 12 used a squat lifting technique, 11 used a mixed lifting technique and 8 used a stoop lifting technique. The therapist classification of lifting posture was comparable to measures of the postural index. Squat lifters had a significantly higher postural index than those using a mixed lift (P<0.05), who in turn had a postural index greater than those employing a stoop lift (P<0.05). No significant difference was found between the postural index measured at the beginning and end of the lifting task across the three groups (See Table 6.1).

Table 6.1. The mean (SD) Postural Index for the three lifting classification groups at the start and end of the task.

	Lifting posture					
	Squat N=12		Mixed N=11		Stoop N=8	
	Start	End	Start	End	Start	End
Postural Index	0.44 (0.07)	0.47* (0.13)	0.30 (0.10)	0.33* (0.11)	0.13 (0.12)	0.14 * (0.11)

Data are means and standard deviations
* indicates a significant difference between lifting posture (P<0.05)

Lumbosacral motion and velocity

Peak lumbosacral flexion and total range of motion during lifting and lowering were significantly less for those individuals using a squat technique compared to those using a mixed and stoop technique (P<0.05). Stoop lifters also exhibited significantly faster peak and average lumbosacral angular velocity than mixed or squat lifters during lifting and lowering (P<0.05).

With respect to lumbosacral motion patterns throughout the lifting and lowering squat lifters showed little change in lumbosacral extension until the final third of the lifting phase (P<0.05), where lumbosacral extension velocity was at its maximum (Figures 6.2A and 6.3A). During lowering, squat lifters displayed a rapid increase in velocity during the initial third of lowering (*P*<0.05), gradually decreasing velocity through the latter stages of lowering (Figure 6.3A**).**

In contrast to the squat lifters, mixed and stoop lifters displayed substantial increases in lumbosacral extension velocity between the initial and middle thirds of the lift (*P*<0.05) and continued to extend at higher rates in the final third of the lift (Figure 6.3). During lowering, lumbosacral flexion velocity of mixed and stoop lifters peaked in the middle phase (*P*<0.05), with substantial reductions in lumbosacral velocity in the final third of the lowering phase (Figure 6.3).

Fatigue affected only lumbosacral kinematic variables for mixed and stoop lifters who had significant reductions in lumbosacral extension and total range of lumbosacral motion (*P*<0.05) (Figure 6.2). The mixed and stoop lifters mean lifting and lowering lumbosacral velocity also decreased when fatigued (*P*<0.05) and this decrease was most prominent during the middle third of the lifting and lowering phases (*P*<0.05) (Figure 6.3).

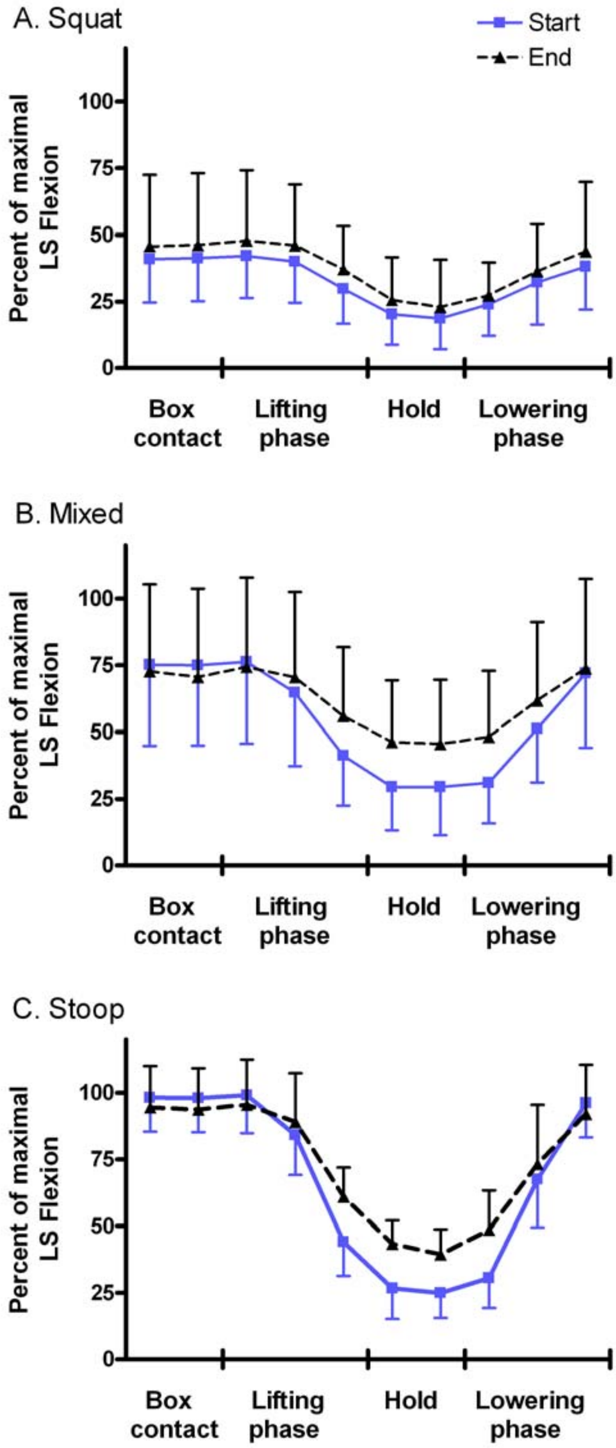


Figure 6.2. Mean percent of maximal lumbosacral (LS) flexion for A) squat, B) mixed, and C) stoop lifting postures at the start and end of the lifting and lowering task.

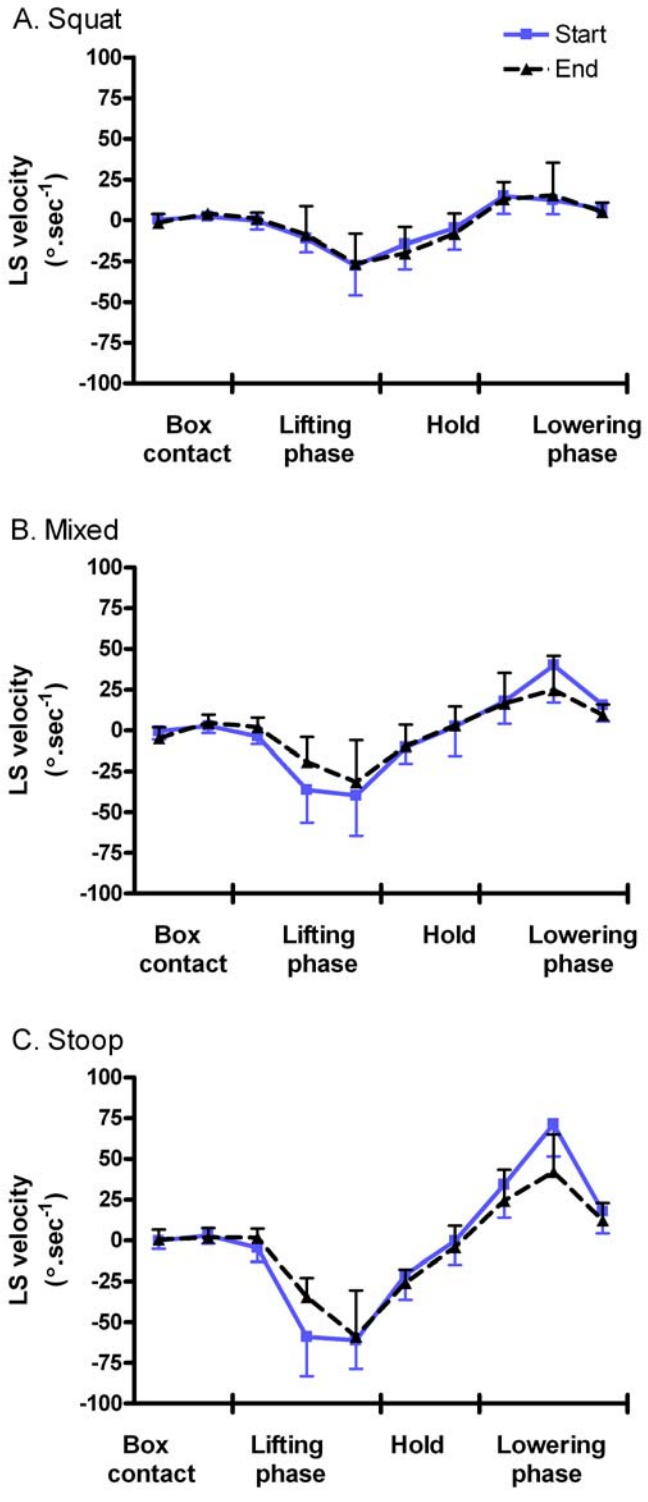


Figure 6.3. Mean lumbosacral (LS) velocity data for A) squat, B) mixed, and C) stoop lifting postures at the start and end of the lifting and lowering task. Negative values represent extension. Positive values represent flexion.

Muscle activation

Due to technical issues, EMG data from two stoop and two mixed lifters could not be processed. The mean percentage of MVC activation throughout the lifting and lowering for the UES and LES was significantly greater than any of the abdominal muscles irrespective of fatigue ($P<0.05$). UES and LES in mixed and stoop lifters exhibited greater average levels of activation during lifting when compared to the lowering phase ($P<0.05$). The IO muscle was the most active of the abdominal muscles ($P<0.05$).

Squat lifters tended to produce greatest levels of UES activity prior to initiating the lift (Figure 6.4A). In contrast, mixed and stoop lifters produced significantly less UES muscle activation prior to the lift when compared to the initial third of the lift ($P<0.05$). Throughout the remainder of the lifting phase all postures progressively decreased levels of UES activity ($P<0.05$) (Figure 6.4). During the lowering phase, all postures displayed lowest levels of UES during the hold phase, then progressively increased levels of activation before reaching a peak in the final stage of lowering ($P<0.05$).

With respect to the LES muscle group, squat lifters displayed similar levels of activation prior to and throughout each third of the lifting phase. During lowering, LES EMG levels of activation for squat lifters progressively increased from the second half of the hold phase until the middle third of the lower ($P<0.05$), maintaining the levels of activation at the end of lowering.

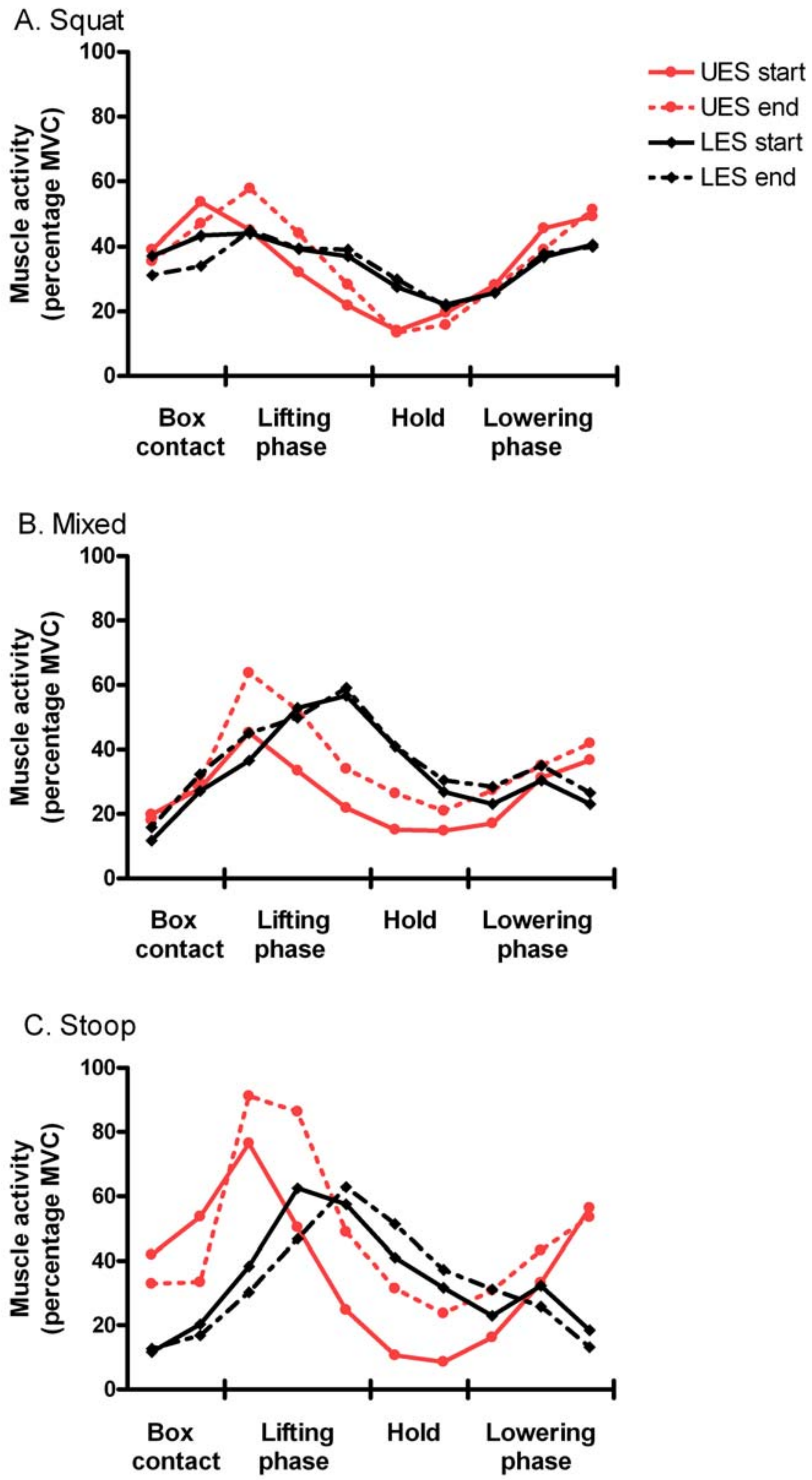


Figure 6.4. UES and LES levels of activation for A) squat, B) mixed, and C) stoop lifting postures at the start and end of the lifting and lowering task. Error bars have been removed for clarity.

The LES activation pattern of stoop and mixed lifters differed from that exhibited by the squat lifters. When lifting the box, lowest levels of LES activity were evident in the second half of box contact and increased at a relatively linear rate throughout the first two thirds of the lifting phase ($P<0.05$), and continued to maintain similar levels of activation at the end stage of the lift (Figure 6.4). During box lowering the level of LES activation of mixed and stoop lifters remained similar throughout all divisions, with the exception of stoop lifters who significantly decreased LES muscle activation in the final third of the lowering phase ($P<0.05$).

The only abdominal muscle to display clear patterns of activity throughout the lifting and lowering cycle was IO. For all postures IO activity increased at a relatively linear rate throughout the three portions of the lifting phase ($P<0.05$). IO tended to remain at this level of activity during the hold phase and decreased levels of activity only in the final third of the lowering phase ($P<0.05$).

With respect to fatigue, all muscles with the exception of LES increased levels of muscle activity during the lifting and lowering cycle at the end of the task when compared to the beginning ($P<0.05$). For UES, this increase in activity at the end of the task was mainly due to increases in activation levels during the lifting phase (Figure 6.4). Although average activation levels of RAB and EO significantly increased at the end of the lifting and lowering task these increases were relatively minor, whereas, IO activity at the end of the task increased throughout the lifting, holding and lowering phases across all postures (Figure 6.5).

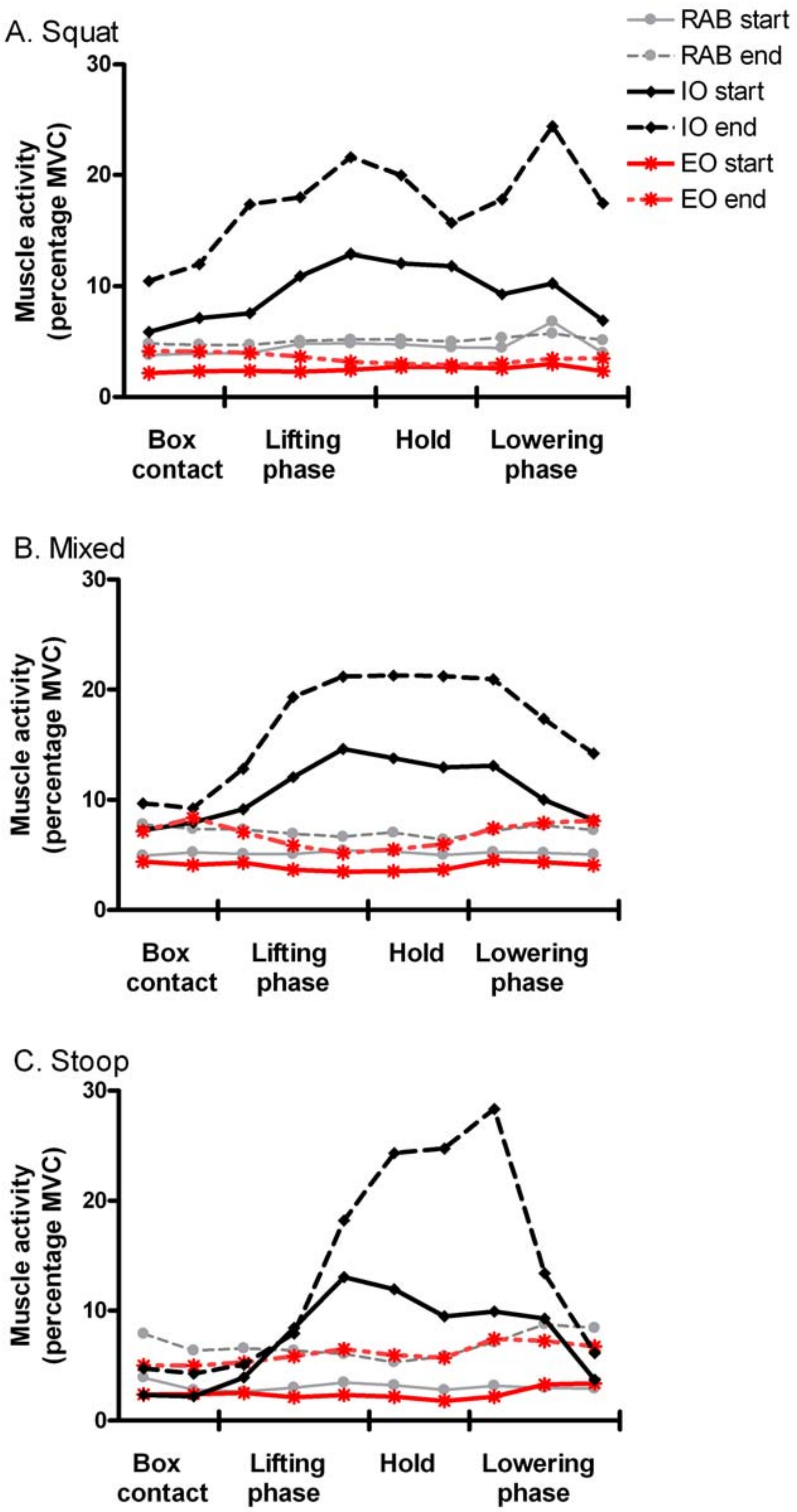


Figure 6.5. Abdominal muscle levels of activation for A) squat, B) mixed, and C) stoop lifting postures at the start and end of the lifting and lowering task. Error bars have been removed for clarity.

All subjects terminated the lifting task due to subjective exhaustion, with a median peak RPE of 19. In addition to subjective fatigue, for all of the groups there was a significant decrease in the MDF of UES and LES during the performance of the MVC immediately after the lifting task (Table 6.2).

Table 6 2. Median frequency (MDF) measures of UES and LES taken during maximum isometric voluntary contraction before and immediately after the repetitive lifting task.

	Lifting posture						ANOVA		
MDF	Squat		Mixed		Stoop		Posture	Fatigue	Interactions
(Hz)	Start	End	Start	End	Start	End			
UES	102.23 (19.1)	90.74 (24.56)	101.70 (14.65)	87.95 (14.98)	107.16 (10.93)	87.64 (14.98)	NS	*P*<0.5	NS
LES	108.68 (12.88)	101.46 (15.80)	110.20 (12.77)	96.21 (19.19)	114.23 (15.4)	98.8 (22.11)	NS	*P*<0.5	NS

Data are means and standard deviations

DISCUSSION

This study found that during high frequency lifting and lowering subjects using a self-selected squat lifting technique displayed different lumbosacral kinematics and trunk muscle activation patterns from those individuals who naturally adopted a mixed or stoop lifting technique. The findings of greater lumbosacral flexion shown by subjects using a stoop lift compared to those using a squat lift was similar to other research (Dolan, Earley et al., 1994; Potvin et al., 1991). Based on lumbosacral range of motion measurements and *in vitro* conversion calculations provided by Adams and Dolan (1991), subjects using a stoop and mixed lifting technique would have flexed their lumbar spine up to 40% and 21% of its elastic limit respectively, whereas the squat lifters flexed their spine to less than 2% of their predicted elastic limit (Adams and Dolan, 1991). Though subjects employing the stoop lifting technique were at 100% of their maximal lumbosacral flexion range they would have not exceeded predicted elastic limits for the ligaments and discs of the spine (Adams and Dolan, 1991), as passive tension of muscle and thoracolumbar fascia provides a safety margin in the *in vivo* spine (Adams and Hutton, 1986). However, this degree of lumbosacral flexion has been shown to substantially increase anterior shear forces acting on the lumbar spine (Bazrgari, Shirazi-Adl, and Arjmand, 2007; McGill et al., 2000; McGill and Kippers, 1994; Potvin et al., 1991) and when repeated at high frequency can potentially lead to vertebral damage over relatively short periods of time (Gallagher et al., 2005).

Although peak lumbosacral flexion differed between postures, all postures exhibited minimal change in lumbosacral angle prior to and on initiation of the lift. These findings concur with those reported by van Dieen et al. (1998), and McGill and Norman (1986). During the initiation of the lift large moments are generated to counter inertial forces arising from the mass of the box and trunk (Delitto and Rose, 1992; McGill and Norman, 1986). When loading demands are high, maintenance of a constant ES muscle length is advantageous as ES torque production is greater at low levels of lumbosacral velocity (McGill and Norman, 1986; Raschke and Chaffin, 1996). However, the mechanism by which lumbar posture was maintained may have differed between postures. Squat lifters produced similar levels of ES activity prior to and during the initiation of the lift. This was probably because their lumbar spines were in a degree of flexion where there would be minimal ligamentous recruitment (Adams and Dolan, 1991) and most of the moment demand would have to be met by the ES muscles. In contrast, mixed and stoop lifters showed considerably lower activation of the ES muscles prior to the lift when compared to the initiation of the lift. Stoop and mixed lifters were in a range of lumbar flexion that would be able to utilise the posterior ligamentous system and stretch of the ES muscles to resist the bending moment of the trunk (Adams and Dolan, 1991; Potvin et al., 1991). Therefore, prior to lifting in the stoop posture less ES activity would be required to resist bending moment imposed on the spine.

In addition to the increased lumbar flexion, lumbosacral velocity during lifting and lowering was greater for stoop lifters than squat and mixed lifters. These differences were most notable during the middle third of lifting and lowering. High rates of lumbosacral velocity during lifting may have implications for LBI as high inertial forces, increased moments (Adams and Dolan, 1996), and compressive loading and shear forces (Granata and Marras, 1995) have been associated with rapid changes in lumbosacral velocity. The higher lumbosacral velocities exhibited during lowering by stoop lifters may also have implications for LBI injury. Cadaver studies involving modelling indicate that increases in loading rate of the spine increases disc and ligament resistance to flexion when compared to slower loading rates (Adams and Dolan, 1995, 1996; Wang et al., 2000).

Lumbosacral velocity data provided some insight into the different patterns of muscle activation displayed by the UES and LES muscles throughout the lifting cycle when adopting different lifting postures. Irrespective of posture, during the lift phase UES activity was at its greatest levels during the initiation of the lift where velocity was low and decreased at a relatively linear rate through the remainder of the lift cycle. Because of its long lever arm, UES is at a greater mechanical advantage to resist the high bending moments compared to the LES (Callaghan and McGill, 1995; Daggfeldt and Thorstensson, 2003; Macintosh and Bogduk, 1987).

Therefore, maximal activation of the UES would be advantageous during the initial stages of the lift. The reduction in UES activity throughout the remainder of the lift would be expected as the moment arm of the trunk and box are reduced as the trunk becomes more upright and the box is brought closer to the body (Keyserling, 2000). Furthermore, the lever arm for the UES aponeurosis would increase as the spine became less flexed (Tveit, Daggfeldt, Hetland, and Thorstensson, 1994).

The LES muscle group in the mixed and stoop lifters showed quite a different pattern of activation from that of the UES throughout the lifting phase and tended to follow the lumbosacral velocity profile. For these postures, the LES was at its lowest level of activation prior to the lift where lumbosacral velocity was low and progressively increased to peak near the end of the lift where lumbosacral velocity was high (Figures 6.3 and 6.4). de Looze et al. (1993) also reported that subjects using a “back” (stoop) technique produced lowest levels of ES activation at the beginning of the lifting phase. LES is more advantaged mechanically than the UES to generate sagittal rotation and posterior shear (Macintosh and Bogduk, 1987, 1991). Therefore, given the positive relationship between ES muscle activation levels and lumbosacral velocity (Dolan and Adams, 1993b; Granata and Marras, 1995) it would seem that a progressive increase in the LES muscle activity through the lifting phase may have been produced to generate rapid changes in lumbar spine curvature (velocity) during lifting in the mixed and stoop postures that utilised a large range of lumbar motion.

During lowering, UES activation patterns were similar between postures, with progressively increasing levels of activation throughout this phase. During box lowering, mixed and stoop lifters activated their LES less than during lifting. Kumar and Davis (1983) also reported lower levels of LES activity during lowering when compared to lifting when using a stoop lifting technique. Decreased levels of LES activity during lowering have been attributed to the reduced EMG activation to torque ratio produced during eccentric muscle action (Tesch et al., 1990).

Furthermore, stoop lifters significantly decreased the levels of LES activity in the final third of lowering. These lower levels of LES muscle activation at the end of the lowering phase corresponded to near maximal flexion for the stoop lifters. It has been shown that when at near maximal lumbosacral flexion during lifting and lowering the LES may exhibit minimal activation (Potvin et al., 1991; Toussaint et al., 1995) without reductions in activity of UES (Toussaint et al., 1995). At end range lumbar flexion decreased fascicle obliquity (Macintosh et al., 1993) and reduced lever arm length (Tveit et al., 1994) make the LES less mechanically advantaged than the UES to resist bending moments without large compressive costs (Toussaint et al., 1995). It has therefore been suggested that in high degrees of lumbar flexion decreasing LES activity, and utilising the UES and ligamentous system to resist bending moments would be advantageous as there would be less compressive cost for given extensor torque production (Toussaint et al., 1995). However, if the spine is less flexed (as shown by the squat lifters) the increased obliquity and lever arm of the LES would allow greater mechanical advantage to resist shear and bending moments (Macintosh et al., 1993; McGill et al., 2000). Thus, the increased activation levels of the LES of squat lifters observed during lowering may reflect a better mechanical efficiency of the LES to aid the UES in resisting bending moments, as the trunk and box were lowered.

Overall, all abdominal muscles were less active than the ES muscles throughout the lifting and lowering cycle, with very low amounts of RAB and EO activity observed. This finding has been documented in other studies of lifting (McGill and Norman, 1986; Potvin et al., 1991). The most active abdominal muscle during lifting and lowering was IO. According to a study by Daggfeldt and Thorstensson (1997), which used a modelling approach, the relatively transverse orientation of the IO muscle fibres gives it a mechanical advantage to aid in the production of extensor moment via intra-abdominal pressure (IAP). Previous work by Cresswell (1993) supports this contention. Cresswell found that during resisted trunk extensions there was negligible activity of EO and RAB, whereas IO muscle activation was associated with increases in IAP.

In the current study, IO was most active during the latter stages of the lift phase and tended to maintain this level of activation during the hold and initial lowering phase where activity of the ES muscles was minimal. These findings were similar to some authors (Cappozzo et al., 1985; de Looze et al., 1999; McGill and Norman, 1986) but in contrast to others (Delitto and Rose, 1992; Vakos et al., 1994). It has been shown that in a more upright position abdominal muscle activation is most advantageous in terms of spinal 'stability-load trade off' (Cholewicki et al., 1997; Granata and Marras, 2000). That is in an upright posture trunk stability (Cholewicki and McGill, 1996) and flexion moments (McGill and Norman, 1986) are low compared to when the trunk is flexed forward (Granata et al., 2005).

Effects of fatigue

Decreases in the median frequency of the UES and LES muscle groups at the end of the task indicated that irrespective of posture subjects were fatigued. Despite showing evidence of ES muscle fatigue, there was no significant change in the postural index or maximal lumbosacral flexion for each posture at the end of the lifting task. Some authors have found that the postural index and lumbosacral flexion of those individuals using freestyle and squat lifting techniques increases when fatigued (Bonato et al., 2003; Hagen, Sorhagen, and Harms-Ringdahl, 1995). In contrast, Marras and Granata (1997) found that experienced manual handlers reduced peak lumbar flexion throughout a five hour period. These contrasting results may be attributed to a methodological difference between studies, where initial self-selected lifting posture was not categorised, subjects were provided with lifting technique instruction, and lifting and lowering height were variable. For example, objects lifted directly from the floor tend to produce a greater degree of lumbosacral flexion than when lifted from a handle height 15 cm above the floor (Dolan, Mannion et al., 1994; Potvin et al., 1991)

Whilst peak lumbosacral flexion was unchanged at the end of the task for each group, a reduction in lumbosacral extension, total ROM and velocity during the middle third of the lift and lower was found in the mixed and stoop lifting groups. Reductions in lumbosacral velocity and range of motion in the sagittal plane have also been found during repeated trunk extensions (Parnianpour et al., 1988). A reduction in lumbosacral velocity may be related to a loss of contractile speed that has been reported in fatigued muscle (Bigland-Ritchie, Johansson, Lippold, and Woods, 1983). At the end of the lifting and lowering task the fatigued ES muscles may not have been able to generate force as quickly, compromising lumbosacral velocity in the postures that utilise a large range of lumbosacral motion (mixed and stoop lifters). Therefore, given the inverse relationship between lumbosacral velocity and lumbar spine torque production that has been observed previously (Marras, Rangarajulu, and Wongsam, 1987), a reduction in lumbosacral velocity may have been used as a compensatory mechanism by the ES muscles to maintain the extensor moment requirements. In addition, having the lumbar spine more flexed during lifting and lowering may have allowed more optimal torque production of the ES muscles in relation to EMG output when compared to a more extended position (Raschke and Chaffin, 1996; A. L. Roy et al., 2003) and may have permitted greater recruitment of the passive elastic properties of the muscle through a larger proportion of the lifting cycle (McGill et al., 1994).

Irrespective of posture, there was an increase in abdominal muscle co-contraction at the end of the lifting task. Increased abdominal co-activation has also been noted during fatigue when performing static lumbar extensions (Reeves et al., 2008; Sparto, Parnianpour, Marras et al., 1997). This increase in abdominal co-contraction has been suggested to be a mechanism to stiffen the spine and aid stability (Reeves et al., 2008). All of the lifting groups in the current study showed evidence of local ES muscle fatigue at the end of the lifting task. It has been shown that when ES muscles are fatigued, the ability of the ES muscle group to control force production (Reeves et al., 2008; Sparto, Parnianpour, Marras et al., 1997) and posture within the primary plane of motion is reduced (Parnianpour et al., 1988; Sparto, Parnianpour, Reinsel, and Simon, 1997b). Deficits associated with these altered force producing capabilities of the ES have been suggested to reduce spinal stability (Reeves et al., 2008). As increased abdominal muscle co-contraction has been shown to improve spinal stability (Cholewicki et al., 1997; Gardner-Morse and Stokes, 1998; Granata and Marras, 2000), the additional abdominal muscle activation observed when fatigued may have been a compensatory mechanism to improve trunk stability with the onset of ES muscle fatigue (Reeves et al., 2008).

CONCLUSION

The findings of this study showed that during a high frequency lifting and lowering task individuals using a naturally selected mixed and stoop technique produced differing lumbosacral kinematics and LES muscle activation patterns from those subjects who used a squat lift. In particular, the end range of lumbar flexion and the relatively high lumbosacral velocity displayed by those individuals who adopted a stoop lift may potentially place additional loading on the inert structures of the lumbar spine. Furthermore, lumbosacral kinematics in individuals using mixed and stoop lifting techniques were compromised by lifting-induced fatigue to a greater extent than in those individuals using a squat lifting technique. Irrespective of posture, the role of co-activation of the abdominal musculature may be important in spinal stability as an individual becomes more fatigued. These findings highlight that lumbosacral kinematics and associated trunk muscle activity during repetitive lifting should be taken into consideration when assessing lifting posture during repetitive manual handling tasks in the work place.

Chapter Seven

Summary and Conclusions

Although sudden loading and high frequency repetitive lifting have been identified as key risk factors for LBI, there has been relatively little research examining neuromuscular and postural responses to these activities when individuals adopt different postures or during the onset of physical fatigue. For example, few studies have investigated differences in postural responses to sudden loading in a stoop and upright posture, or following exposure to repetitive manual handling resulting in fatigue. Furthermore, there have been no studies that have investigated postural, neuromuscular, cardiovascular and psychophysical responses (for example, lumbar motion, EMG, RPE and HR), to high frequency repetitive lifting in individuals who adopt different self-selected lifting techniques (squat, mixed or stoop). Consequently this thesis has sought to address some of these issues.

SUDDEN LOADING

A key finding from those studies which investigated the effects of sudden loading was that the posture adopted prior to the loading event influenced neuromuscular responses, and lower limb and spinal kinematics. Sudden loading in an upright posture led to relatively simultaneous trunk and lower limb joint initiation and co-activation of oblique abdominal and ES musculature. In contrast, sudden loading in the stoop posture produced a relatively small contribution from the abdominal musculature and exhibited a sequential joint initiation strategy, with less motion of the lumbar spine and joints of the lower limb. The most probable explanation for these differences is the degree of recruitment of passive and active tissues of the lumbar spine. In an upright standing posture the lumbar spine is relatively unstable as there is minimal passive resistance arising from the posterior ligamentous system (Cholewicki and McGill, 1996). Therefore, in this position co-activation of the abdominal and ES trunk musculature would potentially increase lumbar spine stability (Cholewicki et al., 1997; Gardner-Morse and Stokes, 1998; Granata and Marras, 2000). In contrast, the stoop posture places the lumbar spine near its end range of flexion which is likely to facilitate the recruitment of the posterior ligamentous system and the passive elastic properties of the stretched ES muscles (Macintosh et al., 1993; McGill et al., 2000; McGill et al., 1994). This recruitment of passive tissues appears to aid stability and may allow forces to be transmitted in a sequential manner to the lower limbs through ligamentous connections thereby lessening the inertial loads on individual joints (Cordo and Nashner, 1982).

Prior exposure and forewarning influenced postural responses to sudden loading to a greater extent in the upright position when compared to a stoop posture. When subjects were exposed to the first unexpected loading trial in the upright standing posture, all muscle response times were delayed and joint excursion was greater when compared to subsequent exposures. The larger joint excursions during the first trial may have led to a greater excursion of the body’s centre of gravity, potential loss of equilibrium, and greater forces imposed on the lumbar spine. This highlighted the importance of reducing the risk of exposure to sudden, unexpected loading events within the work environment (Magnusson et al., 1996).

Exposure to a warning signal enhanced neuromuscular and postural responses to sudden loading when in the upright posture by increasing preparatory levels of LES muscle activity and enabling more global muscles to activate more quickly in response to the loading event. This type of strategy is likely to increase stability (Cholewicki and McGill, 1996) at an intersegment level in the lumbar spine in preparation for loading whilst allowing global muscles more time to contract in response to the perturbation (Allison and Henry, 2002). Although muscle onset latencies decreased with warning, the relative muscle onset sequencing remained similar for all trials. These findings would support the "postural set theory" that postural responses are centrally derived (Cordo and Nashner, 1982; Horak and Nashner, 1986). Therefore, it would seem that the centrally derived set of commands relating to muscle selection and timing (the postural set) was executed more rapidly when warning was providing, subsequently reducing joint range of motion.

Warning in the upright posture led to earlier onset of knee joint extension, hip and trunk flexion, which has been shown to be an efficient means of maintaining the body's centre of gravity within the base of support (Alexandrov et al., 1998). In a stoop posture, warning was shown only to shorten the onset latencies of the UES and LES, although, as was observed for upright standing, hip and knee motion were reduced. Hence, it would seem that irrespective of posture, warning allowed earlier activation of key muscles which would stiffen joints and enable equilibrium to be maintained more efficiently. The implementation of warning systems into the physical work environment is difficult. However, training the identification and awareness of key warning stimuli prior to sudden disturbances in equilibrium may be an intervention approach that can be incorporated into worker training schemes.

Although the effects of warning were clearly evident, postural and neuromuscular responses following lifting-induced fatigue were less apparent. There was no evidence of increased levels of preparatory muscle activity following fatigue, but LES and EO onset latencies decreased following both a gentle exercise programme and lifting-induced fatigue. The reduced onset latencies may reflect a learning or exercise effect rather than a specific response to fatigue. Interestingly, even with earlier onset of the LES and EO it would seem that fatigue induced by repetitive lifting inhibited the ability of an individual to reduce joint motion when warning was provided. This may indicate that lifting-induced fatigue may affect the contractile properties of the muscle as opposed to the ability of the central nervous system to initiate muscle activation. This thought was supported by evidence of local fatigue of the ES muscles following the repetitive lifting task. The influence of fatigue on the contractile component of the muscular response to sudden loading was not specifically examined in the current thesis and warrants further investigation in order to fully understand the neuromuscular control associated with a postural disturbance when physically fatigued.

REPETITIVE LIFTING

A key finding concerning the exposure to high frequency lifting was that RPE and HR responses were similar across self-selected lifting postures. Even when expressed as a percentage of predicted maximum HR there was high subject variability. Furthermore, peak HRs recorded during the task were only two thirds of the predicted maximal HR. This indicated that excessive cardiorespiratory stress was probably not the limiting factor for termination of the task and highlights the difficulty in determining safe absolute or relative limits for HR during the performance of manual handling tasks. However, HR patterns throughout the task were similar for all individuals and may provide a better indication of fatigue than peak or average HR measures. The slow rise in RPE and HR measures throughout the task showed that subjects were unable to maintain subjective and physiological steady state, and this may be an important indicator of, or a prerequisite to fatigue (Herman et al., 2003). These findings would suggest that HR and RPE patterns may be important when attempting to assess and establish physiological fatigue limits for individuals involved in repetitive manual handling tasks (Asfour et al., 1988; Herman et al., 2003).

It would also seem that perception of local back discomfort and MDF measures of the ES muscle EMG signal may be important indices of fatigue during the performance of highly repetitive lifting activities (Bonato et al., 2003; Hagen and Harms-Ringdahl, 1994). All subjects in the current study terminated the repetitive lifting and lowering task due to subjective back muscle fatigue, and MDF measures provided evidence of local ES fatigue at the end of the task. Whilst the implementation of EMG measures in the working environment is impractical, establishing a relationship between these objective and subjective measures of muscle fatigue in the laboratory setting may eventually allow the utilisation of subjective measures as an indicator of local muscle fatigue in the working environment (Bonato et al., 2003).

Whilst subjective and HR responses were similar across postures, lumbar spine kinematics and associated trunk muscle activity differed between self-selected lifting postures. During repetitive lifting and lowering, the stoop lifters flexed their spine to maximum levels and moved their spine at speeds up to five times greater than those individuals who adopted a squat lifting technique. High levels of lumbar flexion and velocity have been considered risk factors for LBI and biomechanical evidence would indicate that both of these factors also increase the loading of the inert structures of the spine (Adams and Dolan, 1991, 1996; Dolan, Mannion et al., 1994; Marras et al., 1995; Wang et al., 2000). Based on these findings, it would appear that individuals who adopt a stoop lift during a manual handling task put themselves at greater risk of LBI than those adopting a squat lifting technique. As the current study investigated lumbar kinematic responses to high frequency lifting and lowering, it should not be assumed that similar responses would occur during manual handling activities performed at different loads, lifting heights, and lifting rates. However, these findings highlight the importance of taking into consideration the speed and extent of lumbar spine motion in tasks that require repetitive manual handling in working environments. To enable the practical implementation of measures of lumbar posture in the workplace, relationships between the current lab-based assessment of lumbar posture and work-based assessment techniques need to be established.

The kinematic profiles exhibited by each posture may provide some insight into the functional roles of the UES and LES muscles. The UES has a longer lever arm than the LES and is more mechanically advantaged than LES to resist bending moments, even when the spine is flexed (Callaghan and McGill, 1995; Daggfeldt and Thorstensson, 2003; Macintosh and Bogduk, 1987). This is probably why the highest levels of UES muscle activity across all postures were observed at the beginning of the lift and towards the end of the lowering phase, where the lumbar spine was most flexed, lumbosacral velocity was low, and bending moments would have been at their highest. In contrast, mixed and stoop lifters displayed the lowest levels of LES activity when the spine was most flexed and lumbosacral velocity was low. Also, the LES tended to be most active during the middle and end stages of the lift where lumbosacral velocity was relatively high. The LES is more mechanically advantaged than UES to produce posterior shear and sagittal rotation at a segmental level when the spine is less flexed (Macintosh and Bogduk, 1986, 1991). This may indicate that the LES may have an important role to play in the rate of change of lumbar curvature (velocity) in less flexed lumbar spine positions where the external bending moments are not maximal.

During lifting and lowering, unlike the LES, abdominal muscles showed similar activation patterns for all postures. The RAB and EO muscles had consistently low activation levels throughout all phases of the lifting cycle, and IO was considerably more active than the other abdominal muscles. It has been suggested that the orientation of the IO fibres may give it a better mechanical advantage to aid spinal stability by increasing IAP when compared to other superficial abdominal muscles (Cresswell, 1993; Daggfeldt and Thorstensson, 1997). IO also produced the highest levels of muscle activity when subjects approached a more upright posture. This may be related to fact that the relative total compressive and shear cost of additional abdominal co-contraction associated with increased stability of the lumbar spine is considerably less in upright standing compared to positions where the trunk is inclined forward (Granata and Marras, 2000; Granata et al., 2005).

Trunk muscle activation and lumbar kinematic findings from chapter six provide insight into the neuromuscular control of the lumbar spine during repetitive manual handling activities. This has implications for the muscular rehabilitation of individuals with low back injury who are returning to repetitive manual handling tasks. The findings indicated that during repetitive lifting and lowering, the muscular control of lumbar motion may vary for different lifting postures, and may be dependent on the speed and extent of lumbar motion. In order to advance current trunk muscle rehabilitation techniques for low back injury, the relationship between activation of trunk muscular and lumbar spine kinematics during manual handling tasks need to be taken into consideration.

Despite evidence of subjective and ES muscle fatigue, subjects maintained their preferred lifting style throughout the lifting task and no change in peak lumbar flexion was evident for any of the postures. However, when fatigued, mixed and stoop lifters showed reductions in the amount of lumbar extension and lumbosacral velocity during the middle third of the lifting and lowering phases. This may have been a compensatory mechanism for the loss of ES contractile speed due to local ES muscle fatigue (Bigland-Ritchie et al., 1983) but could potentially decrease spinal stability (Parnianpour et al., 1988). In addition, a more flexed lumbar spine may have optimised the ratio of ES muscle activation to torque production (Raschke and Chaffin, 1996; A. L. Roy et al., 2003). Fatigue also led to increased co-activation of the abdominal muscles and additional involvement of the UES, particularly during the lifting phase. This increased co-activation of abdominal musculature has been suggested as a mechanism to increase spinal stability when the ES muscles are fatigued (Reeves et al., 2008).

RECOMMENDATIONS FOR FUTURE RESEARCH

The findings from the current thesis identified a number of areas for future research.

1. In the current study, sudden loading was performed in a relatively static upright and stoop posture, yet vocational tasks in which sudden loading injuries often occur are typically dynamic. Implementation of sudden loading during the movement phase of a manual handling task may provide greater insight into injury mechanisms associated with sudden unexpected disturbances to the body's equilibrium.

2. The current study indicated that irrespective of lifting posture, a failure to sustain subjective and physiological steady state and local ES muscle fatigue seemed to be indicators of fatigue during repetitive manual handing. However, it is not clear how adjusting work load to maintain subjective steady state during highly repetitive manual handling activities would influence HR, VO_2 patterns and local muscle fatigue when adopting different lifting postures. Furthermore, the influence of rest breaks on physiological and biomechanical responses to repetitive manual handling tasks requires further investigation.

3. The current study found distinct differences in lumbosacral kinematics and trunk muscle activation for different lifting postures in response to the handling task. The handling task controlled the height of the lift and lower, as well as the lifting frequency. Further investigations into self-selected lifting posture, lumbosacral kinematics and muscle activation for a range of different lifting heights and frequencies of lifting would help to determine whether self-selected lifting posture and associated lumbosacral biomechanics are task specific or change as a result of these task parameters.

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Appendices

Appendix 1

Subject Information

AUCKLAND INSTITUTE OF TECHNOLOGY
Te Whare Takiura O Tamaki Makau Rau

Kinematic (movement) and electromyographic (muscular) responses to sudden loading of the trunk: effects of posture and fatigue.

Investigators: Grant Mawston and Peter McNair

You are invited to participate in a research study performed as part of the requirements for completion of a MHSc thesis for Grant Mawston. Participation is completely voluntary and you may withdraw from the study at any time prior to the completion of data collection without giving a reason or being disadvantaged.

<u>Aim</u>

The aim of this study is to determine the effect of standing posture and fatigue on postural responses during a sudden loading event. Sudden loading events and fatigue have been associated with low back pain in industry. This study will allow us to gather information about how the muscles of the body respond to unpredictable loading when fatigued. There has also been controversy as to what is the ideal lifting posture. Findings from this study will give us an idea of movement and muscular responses to sudden loading in normal standing and stooped positions.

Procedure

Electrode recording devices will be taped to the skin surface covering several muscle groups of your arm, leg and trunk. These measure muscle activity and have been specifically designed so there is no risk of any electrical shock. Once the electrodes have been attached, you will be asked to perform maximum contractions from several muscle groups against the researchers manual resistance. ***Shiny surface markers will be attached to several bony landmarks on your foot, knee, hip, back and arm.***

You will be assigned randomly to one of two groups. You will be required to stand in an upright standing and stoop position, holding a box. An unexpected load will be added to the box. If you are assigned to the fatigue task group, you will be asked to carry out the fatigue task, which requires you to lift a box weighing ten kilograms to the beat of a metronome, until you are fatigued, or you are advised by the researcher to stop. During the fatigue task your heart rate will be measured and investigator

will ask you to rank how exerted you feel using a visual exertion scale. You can withdraw from the fatiguing task at any stage.
Video footage will be taken of you during the loading and lifting tasks. The camera is set up so that the markers attached to you are illuminated. The researchers will be the only people who will see your video footage.

If you have any musculoskeletal or cardiovascular conditions that may affect your performance in the tasks, you will be excluded from participating in this project.

All measurements will be undertaken at the School of Physiotherapy, Akoranga Campus, Auckland University of Technology. You will be required to undertake the tasks on one occasion, lasting approximately two hours.

Risks

There is some risk associated with lifting and sudden loading of the trunk, and you may experience minor low back and buttock discomfort after the fatigue task. However, other studies have used similar procedures to the current investigation and the current study uses weights, rate of lifting and postures that are commonly used in the workplace. If you feel that you have excessive discomfort during the lifting and loading tasks, please advise the researcher and you will be withdrawn from the experiment. ***If you feel any discomfort following the experiment, please tell the investigator who will provide you with advise and options with respect to managing any discomfort.***

Results

Privacy and confidentiality is of utmost importance. You will be identified by number only and access to data is restricted to the researchers. Results will be published in a peer-reviewed journal and presented at a national conference. Results of the study will be made available to you at the completion of the study upon request by contacting Peter McNair (see next paragraph).

If you have any concerns regarding the nature of this project, then you should contact the Project Supervisor, Peter McNair (peter.mcnair@aut.ac.nz), ph 307-9999 Ext. 7146. Any concerns regarding the conduct of the research should be made to the Executive Secretary, AUTEC, Madeline Banda (madeline.banda@aut.ac.nz), ph. 307-9999 Ext. 8044.

Appendix 2

Consent to Participation in Research

Title of project: **Kinematic and electromyographic responses to sudden loading in the trunk: effects of posture and fatigue.**

Project Supervisor: **Peter McNair**

Researcher: **Grant Mawston**

- I have read and understood the information provided about this research project.
- I have had an opportunity to ask questions and to have them answered.
- I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. If I withdraw, I understand that all relevant tapes and transcripts, or parts thereof, will be destroyed
- I agree to take part in this research.

Participant signature: ....
Participant name:

Date:

Approved by the Auckland Institute of Technology Ethics Committee on 11/12/2000

AITEC Reference number _00/79

Appendix 3.

Participant Information Sheet

Date Information Sheet Produced: 06th March, 2008

Do you need an interpreter?

English	I wish to have an interpreter	Yes	No
Maori	E hiahia ana ahau ki tetahi Kaiwhakamaori / Kaiwhakapakeha korero	Ae	Kao
Samoan	Oute mana'o ia iai se fa'amatala upu	Ioe	Leai
Tongan	Oku ou fiema'u ha fakatonulea	Io	Ikai
Cook Island	Ka inangaro au I tetai tangata uri reo	Ae	Kare
Niuean	Fia manako au ke fakaaoga e taha tagata fakahokohoko kupu	E	Nakai

Project Title

The effect of repetitive lifting on postural, psychophysical, and physiological responses of young and middle aged individuals.

An Invitation

I am Grant Mawston, a researcher at the Health and Rehabilitation Research Centre at Auckland University of Technology.

You are invited to participate in a research study. Participation is completely voluntary and you may withdraw from the study at any time prior to the completion of data collection without giving a reason or being disadvantaged.
Your participation in this study will be stopped should any harmful effects appear.

What is the purpose of this research?

This study aims to investigate how repetitive lifting affects perceptual effort, body motion, and physiological measures (oxygen consumption, heart rate, and muscle activity) in young and middle aged males.

Repetitive lifting has been associated with the increased risk of low back pain in industry. This study will allow us to gather information about how the body responds to repetitive lifting.

The results of this study will be analysed and written up for publication in a medical journal.

No material that could personally identify you will be used in any reports on this study unless your personal approval is given for the dissemination of results to specific persons (please see the section below titled "How will my privacy be protected?" for more information on privacy issues).

Are you eligible to participate in this project?

If you are a male between the age of 18 and 55 years old and have no musculoskeletal or cardiovascular condition that may affect your performance in the VO_2 max and lifting task (described below) you are eligible to participate in this study. You are not eligible to participate in this study if you:

- Have any medical conditions (eg. cardiovascular disease, neurological or psychological disease, cancer, respiratory disease).
- Have had a low back injury within the last 6 months.
- Have a chronic low back injury (low back pain for greater than 3 months).
- Are under 18 or over 55 years of age.
- Are female.
- Have any other severe musculoskeletal condition that may inhibit lower limb movement.

What will happen in this research?

This study involves two sessions of approximately 2 hours, one week apart.

Session 1: Lifting familiarisation and VO_2 max test using a cycle ergometer

In the first session you will be familiarised with the lifting task and perform a VO_2 max test. After giving your consent you will be asked to complete the Habitual Activity Questionnaire which provides an indication of your activity levels. On completion of the questionnaire your weight, height, resting heart rate, breathing rate and blood pressure will be recorded. You will then be asked to take part in the lifting familiarisation task which will require you to practice lifting a box from mid-shin to waist height. This will be performed to the beat of a metronome at a frequency of 10 lifts per minute.

Following the practice lifting session you will be asked to perform a VO_2 max test. Your VO_2 max will provide us with a measure of your cardiorespiratory fitness. The VO_2 max test will be performed on an exercise cycle (exercycle). The test will begin with a warm-up, and then the resistance of the exercycle will be gradually increased every two minutes until your VO_2 max is reached, which usually coincides with your maximal exertion. During the test you will wear a face mask and a heart rate monitor, and the researcher will ask you to rate your level of effort. It should be noted that this is a maximal test that stresses the heart and lungs and you can terminate the test at any time during the testing procedure.

Session 2: Lifting test

The lifting test session will begin with a general warm-up. Electrodes will then be placed over a number of muscles on your trunk and legs, and shiny markers will be placed on various bony prominences on your lower limbs and trunk. These are used to track your movements and posture. The lifting task will require you to lift a box weighing 10 kg to the beat of a metronome, until you are fatigued, or you are advised by the researcher to stop. During the lifting task your heart rate, oxygen consumption, muscle activity and lifting posture will be measured. The investigator will also ask you to rate how exerted you feel at regular intervals throughout the lifting task using a perceived exertion scale. You can withdraw from the lifting task at any stage.

A video camera will record the lifting task in order that 3 physiotherapists can assess your lifting posture. These therapists and the researchers will be the only people who will see the video recording.
If you have any musculoskeletal or cardiovascular condition that may affect your performance in both the VO_2 max and lifting tasks, you will be excluded from participating in this project.
All measurements will be undertaken at the Health and Rehabilitation Research Centre, Akoranga Campus, Auckland University of Technology. The two sessions will last for approximately two hours.

What are the discomforts and risks?

- There are some risks associated with both VO_2 max testing and repetitive lifting to fatigue. Both tests will require maximal effort, stressing the heart, lungs and musculoskeletal system and with increasing exercise intensity there is a risk of a cardiovascular incident or musculoskeletal injury.

- There is a risk of delayed onset muscle soreness. As the name suggests this is soreness of the muscles that begins one or more days after exercise. This can occur following exercise which you are not used to. While this can be uncomfortable the symptoms usually go away after one or two days.

How will these discomforts and risks be alleviated?

- You will be required to complete a screening questionnaire (PAR-Q) to identify potential risk factors for exercise.
- The researchers involved in testing are trained in cardiopulmonary resuscitation (CPR) and have a set protocol for dealing with a cardiovascular event. Access to defibrillation equipment is available via the Student Health Clinic located adjacent to the HRRC.
- The tests undertaken in the current study have been used for research on healthy populations within your age group with no reported adverse effects.
- During either of the two test sessions should you feel any chest or arm pain, feel dizzy, faint, shortness of breath or nauseous, or have excessive discomfort please advise the researcher and you will be withdrawn from the experiment. If you feel any discomfort following the experiment, you should inform the investigator who will provide you with advice and options regarding the management of any discomfort.

What are the benefits?

Information from the VO_2 test will provide you with a reliable measure of your current cardiovascular fitness level as well as providing objective measures that correspond to your perceived level of effort during each stage of the test. Data from the lifting session will provide information about how repetitive lifting affects your work capacity.

What compensation is available for injury or negligence?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will my privacy be protected?

Any information we collect will not be able to be identified as belonging to you. All data collected will only be identified by a number. The researchers will be the only people who have access to this information. All information will be kept in a secure room and in a locked filing cabinet.

What are the costs of participating in this research?

There is no financial cost to you to participate in this research. It will take approximately 2 hours per session, with approximately a week between sessions.

What opportunity do I have to consider this invitation?

You have one week to decide whether you wish to take part in the study. You have a right to choose not to participate. If you agree to take part you are free to withdraw from the study at anytime, without having to give a reason.

How do I agree to participate in this research?

If you agree to participate in the study please complete the attached consent form.
Will I receive feedback on the results of this research?
If you wish to have a copy of the results of this research, please inform the supervisor, Grant Mawston. This will be available after the study is completed and published.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Grant Mawston, grant.mawston@aut.ac.nz, and a 921 9999 ext 7180.
Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Madeline Banda, madeline.banda@aut.ac.nz , 921 9999 ext 8044.

Whom do I contact for further information about this research?
Please feel free to contact the researcher if you have any questions about this study.
Researcher Contact Details:

Grant Mawston, grant.mawston@aut.ac.nz, 921 9999 ext 7180.

This study has received approval from the AUT Ethics Committee on 22nd February 2008
Reference number: 08/15

Appendix 4

Consent to Participation in Research

Title of project: **The effects of repetitive lifting on postural, psychophysical, and physiological responses in young and middle aged individuals.**

Researchers: **Grant Mawston, Mark Boocock, Peter McNair, and Jennifer Bassement**

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* I have read and understood the information provided about this research project.
* I have had an opportunity to ask questions and have them answered.
* I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way. If I withdraw, I understand that all relevant video tapes and data will be destroyed.
* I have not had a history of low back pain within the last six months; previous spinal surgery; any cardiovascular or neurological condition where aerobic stress is likely to cause harm; and any musculoskeletal injury that may affect performance.
* I agree to take part in this research.
* I would like to receive a summary of the research Yes No

Participant signature: ....

Participant name:

Participant Contact Details (if appropriate):

……………………………………………………………………………………..

……………………………………………………………………………………..

……………………………………………………………………………………..

Date:

Approved by the Auckland University of Technology Ethics Committee on 22nd February 2008 **AUTEC Reference number** 08/15_

Appendix 5

HABITUAL PHYSICAL ACTIVITY QUESTIONNAIRE (Baecke et al., 1982)

Questionnaire codes and method of calculation of scores on the habitual physical activity

		SCORE							
NAME :……………………………………………..									
1) What is your main occupation? …………………………………………………………		1	-	3	-	5			
2) At work I sit	NEVER / SELDOM / SOMETIMES / OFTEN / ALWAYS	1	-	2	-	3	-	4	5
3) At work I stand	NEVER / SELDOM / SOMETIMES / OFTEN / ALWAYS	1	-	2	-	3	-	4	5
4) At work I walk	NEVER / SELDOM / SOMETIMES / OFTEN / ALWAYS	1	-	2	-	3	-	4	5
5) At work I lift heavy loads	NEVER / SELDOM / SOMETIMES / OFTEN / VERY OFTEN	1	-	2	-	3	-	4	5
6) After working I am tired	VERY OFTEN / OFTEN / SOMETIMES / SELDOM / NEVER	5	-	4	-	3	-	2	1
7) At work I sweat	VERY OFTEN / OFTEN / SOMETIMES / SELDOM / NEVER	5	-	4	-	3	-	2	1
8) In comparison with others of my own age I think my work is physically	MUCH HEAVIER / HEAVIER / AS HEAVY / LIGTHER / MUCH LIGHTER	5	-	4	-	3	-	2	1

9) Do you play sport YES / NO

If yes:

- Which sport do you play most frequently? ....
- How many hours a week? <1 / 1-2 / 2-3 / 3-4 / >4
- How many months a year? <1 / 1-3 / 4-6 / 7-9 / >9

If you play a second sport

- Which sport is it? ……………………………………………………………………………………………..
- How many hours a week? <1 / 1-2 / 2-3 / 3-4 / >4
- How many month a year? <1 / 1-3 / 4-6 / 7-9 / >9

10) In comparison with others of my age I think my physical activity during leisure time is	MUCH MORE / MORE / THE SAME / LESS / MUCH LESS	5	-	4	-	3	-	2	1
11) During leisure time I sweat	VERY OFTEN / OFTEN / SOMETIMES / SELDOM / NEVER	5	-	4	-	3	-	2	1
12) During leisure time I play sport	NEVER / SELDOM / SOMETIMES / OFTEN / VERY OFTEN	1	-	2	-	3	-	4	5

13) During leisure time I watch television								
NEVER / SELDOM / SOMETIMES / OFTEN / VERY OFTEN	1	-	2	-	3	-	4	5
14) During leisure time I walk								
NEVER / SELDOM / SOMETIMES / OFTEN / VERY OFTEN	1	-	2	-	3	-	4	5
15) During leisure time I cycle								
NEVER / SELDOM / SOMETIMES / OFTEN / VERY OFTEN	1	-	2	-	3	-	4	5
16) How many minutes do you walk and/or cycle per day to and from work, school and shopping?								
<5 / 5-15 / 15-30 / 30-45 / >45	1	-	2	-	3	-	4	5

Quantification of Occupation level of physical activity

Level occupation physical activity	Low level	Middle level	High Level
Examples of occupations	clerical work, driving, shop keeping, teaching, studying, housework, medical practice, university educated occupations	factory work, plumbing, carpentry, farming,	dock work, construction work, sport
Score	1	3	5

Quantification and calculation of the simple sports score (question 9)

***Which sport do you play most frequently*?**

Level sporting physical activity	Low level	Middle level	High Level
Examples of sporting activities	billiards, sailing, bowling, golf	Badminton, cycling, dancing, swimming, tennis	Boxing, basketball, football, rugby, rowing
Average energy expenditure (MJ/h)	0.76	1.26	1.76
Intensity Score	1	3	5

How many hours a week?

Time	<1	-	1-2	-	2-3	-	3-4	-	>4
Score	0.5	-	1.5	-	2.5	-	3.5	-	4.5

How many months per year

Proportion	<1	-	1-3	-	4-6	-	7-9	-	>9
Score	0.04	-	0.17	-	0.42	-	0.67	-	0.92

If you play a second sport

How many hours a week?

Time	<1	-	1-2	-	2-3	-	3-4	-	>4
Score	0.5	-	1.5	-	2.5	-	3.5	-	4.5

How many months per year

Proportion	<1	-	1-3	-	4-6	-	7-9	-	>9
Score	0.04	-	0.17	-	0.42	-	0.67	-	0.92

Simple sports score (I_9):

A zero score for people who do not play sport

$$I_9 = \sum_{I=1}^{2} (\text{intensity x time x proportion})$$

=	0	-	<4	-	<8	-	<12
Score	0.01	-	4	-	8	-	12

Calculation of scores of the indices of physical activity:

Work index = $[I_1 + (6 – I_2) + I_3 + I_4 + I_5 + I_6 + I_7 + I_8] / 8$

Sport index = $[I_9 + I_{10} + I_{11} + I_{12}] / 4$

Leisure index = $[(6 – I_{13}) + I_{14} + I_{15} + I_{16}] / 4$