The Effect of Lumbar Posture and Pelvis Fixation on Back Extensor Torque and Paravertebral Muscle Activation

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Glossary of Abbreviations

BE	Back Extensor
BW	Body Weight
CoM	Centre of Mass
CoR	Centre of Rotation
EMG	Electromyography
ES	Erector Spinae
LBI	Low Back Injury
LBP	Low Back Pain
LES	Lower Erector Spinae
MVIC	Maximal Voluntary Isometric Contraction
NME	Neuro Muscular Efficiency
RMS	Root Mean Square
ROM	Range Of Movement
UES	Upper Erector Spinae

Attestation of Authorship

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person (except where explicitly defined in the acknowledgements), nor material which to a substantial extent has been submitted for the award of any other degree or diploma of a university or other institution of higher learning.

Candidate: ________________________ Date: 12th Nov 2013

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Abstract

Lumbar postures using a high degree of flexion are considered a risk factor associated with manual handling related back injuries. The lumbar posture adopted during lifting is thought to alter the force capabilities of the trunk and the active contribution of the paravertebral muscles. To date, most studies examining back extensor (BE) force and muscle activation have been undertaken in non-functional positions with the pelvis fixated. Furthermore, paravertebral muscle activation has focused on a single spinal level, despite known anatomical and functional differences of upper erector spinae (UES), lower erector spinae (LES) and multifidus. An understanding of how lumbar posture can affect the relationships between trunk extensor torque and paravertebral muscle activity in functional lifting positions may provide important knowledge for vocational manual handling training. Thus, the aim of this study was to determine how lumbar posture and pelvis fixation affect BE torque and muscle activation of 3 divisions of the paravertebral muscles.

This study involved 26 healthy participants performing a simulated static lift in 3 lumbar postures (lordotic, flexed and mid-range). A maximal isometric lift was undertaken with and without the pelvis fixated. Participants were provided with real-time feedback of their lumbar posture using a three dimensional (3D) electromagnetic motion monitor. A chest harness connected to a 3D floor mounted force gauge via a metal chain, provided a measure of maximum isometric voluntary BE force. A 9-camera motion analysis system recorded body position, which was used in conjunction with kinetics, to estimate BE torque. Muscle activity of the 3 paravertebral muscles (UES, LES, and multifidus) was recorded using electromyography (EMG). EMG signal for each muscle during peak back extension in each posture were normalised to that produced during maximal exertion in the Biering-Sorensen position.

Lumbar posture was found to have a significant effect on BE torque, with torque increasing from a lordotic to flexed posture ($P<0.0001$). In contrast, a flexed lumbar posture produced significantly less muscle activation for all BE muscle groups than a mid and lordotic posture ($P<0.001$). As a result, the flexed posture showed a higher

neuromuscular efficiency (NME) ratio (torque/activation) when compared to the mid and lordotic posture ($P<0.001$). Each paravertebral muscle responded to a change in posture in a similar manner. Pelvis fixation was found to have no effect on torque or the intensity of muscle activation.

Initiating a lift using a lumbar posture similar to the mid position used in this study may help to avoid loading on the passive structures of the spine, which is thought to occur when using a fully flexed lumbar posture. The mid posture also offers higher levels of NME compared to the lordotic lumbar posture, which may be useful in situations when exerting high BE torque is necessary. Given that pelvis fixation did not influence BE torque and paravertebral muscle activation, it would suggest that functional lifting tasks should not be concerned with restraining the pelvis.

Chapter 1: Introduction

1.1 Overview

A flexed lumbar posture is considered a risk factor associated with low back injury (LBI) (Hoogendoorn et al., 2000; Punnett, Fine, Keyserling, Herrin, & Chaffin, 1991). Reducing the risk of LBI during occupations involving manual handling is one of the leading priorities in preventative medicine and a major concern for health professionals (Waters, Putz-Anderson, & Garg, 1994). Twenty percent of all injuries in the workplace affect the back and typically occur during lifting, especially tasks involving high force exertions and flexed lumbar postures (Hoogendoorn et al., 2000; Waters et al., 1994). Furthermore, LBI is known to be one of the most common and disabling musculoskeletal conditions, with the economic cost thought to exceed $500million in New Zealand each year (McBride, Begg, Herbison, & Buckingham, 2004). The National Institute for Occupational Safety and Health (NIOSH) have published international guidelines on manual handling (Waters et al., 1994) in order to minimise the risk of LBI, yet specific parameters regarding lumbar posture are not given.

Parameters considered important for the prevention of LBI during lifting include the magnitude of back extensor (BE) torque the spine generates to lift a load, and the level of paravertebral muscle activation occurring to control the spine (Roy, Keller, & Colloca, 2003; Tan, Parnianpour, Nordin, Hofer, & Willems, 1993). When generating BE torque, lumbar posture has been highlighted as an important factor known to influence the magnitude of the torque, as well as influencing paravertebral muscle recruitment. Research has shown that when the spine is fully flexed, BE torque is at its greatest, whilst paravertebral muscle activation levels are at their lowest (Dolan, Mannion, & Adams, 1994). At the end range of lumbar flexion, high loads on spinal structures occur (Gallagher, Marras, Litsky, & Burr, 2005) and this is when there is minimal muscle activation to protect the spine, leading to a high risk of LBI when lifting using a flexed lumbar posture.

The majority of the literature that has investigated the relationship between lumbar posture and muscle activity during lifting has focused on erector spinae (ES) muscle activation at one spinal level (usually lower ES at L3) and assumed this to be similar for all divisions of ES across the lumbar spine (Kaigle, Wessberg, & Hansson, 1998; Roy et al., 2003; Tan et al., 1993). However, anatomical and EMG studies suggest that the paravertebral muscles of the lumbar spine have three distinct portions (upper ES, lower ES and multifidus) that are anatomically and functionally quite different and each can be identified using EMG (Bojadsen, Silva, Rodrigues, & Amadio, 2000; Ng, Kippers, Parnianpour, & Richardson, 2002).

The majority of the literature investigating the relationship between lumbar posture and BE torque while lifting has been undertaken in non-functional positions where the pelvis and lower limbs have been restrained. The rationale behind using fixation is to isolate the BE muscles by preventing de-rotation of the pelvis and minimising the contribution of the pelvis and hip musculature. During lifting the spine does not work alone but is an important link in the biomechanical chain connecting the upper and lower body. Less is known about how the BE muscles generate torque and activate in functional lifting tasks or how pelvis fixation influences the biomechanics of the lumbar spine during lifting.

The purpose of this study is to investigate the effect how various lumbar postures during lifting affect BE torque. It also aims to determine muscle activity responses when lumbar posture changes during lifting in three divisions of the paravertebral muscles. The final aim is to determine the effect of pelvis fixation on BE torque production and muscle activation in functional lifting postures to assist in interpreting preceding research which has used non-functional positions.

The outcome from this study may have benefits for developing manual handling programmes aimed at reducing the risk of LBI during lifting. It will provide objective evidence of how the lumbar posture adopted during lifting influences torque and activation of the BE muscles from a functional lifting perspective. The outcome of this study will be of significance to health care professionals, health and safety personnel and those regularly involved in manual handling by providing information appropriate to the prevention and management of LBI. Finally, this study may contribute to the understanding and implications of lumbar posture during lifting to assist health professionals educate clients who undertake manual handling tasks in their occupations or daily activities.

Chapter 2: Literature Review

2.1 Introduction

This chapter begins by outlining how the literature was searched and relevant articles were included in this review. The relationship between lumbar posture and epidemiology of LBI is discussed, along with the effect lumbar posture has on the functional anatomy of the paravertebral muscles. This is followed by discussing how lumbar posture influences BE torque, paravertebral muscle activation and neuromuscular efficiency. The effect of pelvis fixation on BE torque and muscle activation are then discussed. The chapter ends with stating the research aims, objectives and purpose of the study.

2.2 Literature Review Search Process

A literature search was undertaken in August 2010 and repeated in September 2012. The electronic databases searched were Scopus, Cochrane, Pedro and EBSCOHost health database, which includes CINAHL, MEDLINE and SPORTdiscus. Searching of the literature was undertaken in themes to cover different sections of the literature review. These sections included epidemiology and associated biomechanical risk factors of LBP, the effect of lumbar posture on torque and muscle activation, and the effect of pelvis fixation on torque and muscle activation. The search method used different combinations of search terms to identify potential literature and varied according to the databases, which are outlined in Table 2.1.

Relevant articles were identified by title and if the main author deemed them appropriate to this study, the abstract read. Further literature was identified from a manual search of the in-text citations of these articles. Articles were excluded if they were not published in English or were related to spinal surgery. Any articles published before and up until December 2012 were included.

Table 2.1 Literature search themes and search terms

Key search themes		
Epidemiology	Posture, torque, activation	Pelvis Fixation, torque, activation
Risk factor*	Lumbar activat*	Pelvi* fixation
Epidemiolog*	Erector spinae	Pelvi* stabilization
Aeiteolog*	Back extens*	Pelvi*restraint
Prevalence	multifidus	Muscles
Lifting	Electromyograph*	Back exercise
Biomechanic*	Torque	Torque
Kinematic*	Moment	Activation
Low back	Isometric*	Electromyograph*
Lumbar	Lumbar range	Lumbar

2.3 Epidemiology

2.3.1 Prevalence

Low back pain (LBP) affects one in two New Zealanders and is considered one of the most common, costly and disabling musculoskeletal conditions (McBride et al., 2004). The cost of LBI to Accident Compensation Corporation (ACC) in 2008/9 exceeded $110 million. In 2008/9 a third of ACC claims were associated with carrying or lifting tasks (ACC, 2010). Also, 80% of all reported injuries sustained in manual handling lifting tasks affected the lumbar spine (Cole & Grimshaw, 2003).

The high prevalence of LBP has led to a number of epidemiological and biomechanical studies to identify risk factors associated with LBI in manual handling occupations. Over 100 risk factors have been identified that relate to the occurrence of

LBI in manual handling occupations, but some of the more common mechanical risk factors include the intensity and frequency of lifting, lifting distance and fatigue (Adams, Green, & Dolan, 1994; Cole & Grimshaw, 2003; Waters et al., 1994). However, both epidemiological and biomechanical studies indicate that a key risk factor associated with LBI is using a lumbar posture with a high degree of flexion (Hoogendoorn et al., 2000; McGill, Hughson, & Parks, 2000; Punnett et al., 1991).

Most epidemiological studies to date have quantified lumbar posture through the measurement of trunk flexion by using video analysis or devices capable of measuring lumbar motion (Hoogendoorn et al., 2000; McGill et al., 2000; Punnett et al., 1991). These studies have shown that the risk of LBI during lifting is increased when using a lumbar posture with a high degree of flexion. Punnett et al. (1991) used video recordings to analyse the lumbar posture of car assembly factory workers. Manual handling tasks performed in severe lumbar flexion (flexion greater than 45^{o}) were significantly correlated to a higher incidence of LBI compared to tasks performed in neutral or mild trunk flexion ($0 – 45^{o}$ of lumbar flexion). Hoogendoorn et al. (2000) also analysed work place video footage of manual handling lifting tasks and found if flexed postures of more than 60^{o} were sustained for more than 5% of the working day, there was a high risk factor for LBI. Marras et al. (1995) assessed 400 workers during their manual handling jobs and determined lumbar flexion using an tri-axial electrogoniometer worn by the subjects. It was found that even a small amount of lumbar flexion poses a risk for LBI, but working in higher degrees of flexion, comes with a higher degree of risk of injury. Marras et al. concluded that postures that involved lumbar flexion that exceeded 20^{o} were considered to be a risk factor for LBI.

Biomechanical studies also support epidemiological studies on the negative effects of using a flexed lumbar posture. Flexed lumbar postures during lifting are thought to increase the load on the passive systems of the lumbar spine (Bogduk, Macintosh, & Pearcy, 1992; Dolan et al., 1994) leading to potential injury of the facet joints, intervertebral discs and/or the posterior ligament complex of the spine (Gallagher & Marras, 2012; Potvin, McGill, & Norman, 1991). Loading of the passive tissues when moving from the upright neutral spine to approximately 80% of lumbar flexion is considered minimal, but beyond 80% of available flexion, there is an exponential increase in the passive tissue resisting the bending moment acting upon the spine (Dolan et al., 1994). Specifically, it is the posterior ligamentous complex which is considered to exert a large anterior shear on the spinal segments when end range of flexion is approached (Gallagher & Marras, 2012). Interestingly, in contrast to shear forces, compressive forces of the lumbar region have been shown to increases only five percent from an upright posture to that of full lumbar flexion (Macintosh, Bogduk, & Pearcy, 1993).

2.3.2 Functional anatomy of the paravertebral muscles

The erector spinae (ES) and multifidus are the main BE muscles which exert up to 80% of the extension forces across the lumbar spine (Bogduk, 2012). ES are a muscle group consisting of several divisions, which have distinct attachments and innervations. In the lumbar spine, the ES muscle is a complex structure consisting of two main parts; *Longissimus Thoracis* and *Iliocostalis Lumborum*, where the latter is positioned more lateral than the former (Bogduk, 2012; Macintosh & Bogduk, 1991). Each of these muscles has two divisions; *pars lumborum* and *pars thoracis*, with the *thoracic* component sitting more superficially than *lumborum*. However, clinical examination of the individual components of erector spinae is not possible because of their close proximity. This has resulted in a functional classification of the paravertebral muscles according to location of surface anatomy, and thus they have become commonly referred to as Upper Erector Spinae (UES) and Lower Erector Spinae (LES), with multifidus being recognised as a separate muscle (Bogduk et al., 1992; Macintosh et al., 1993; McGill, 1991).

The fascicles of the UES arise from all the thoracic transverse processes and lower seven ribs merging into the erector spinae aponeurosis, which spans across the lumbar spine to insert on the sacrum and pelvis (Bogduk, 2012). The ES aponeurosis does not have any direct attachment onto the lumbar spine. This results in the creation of a bowstring effect and gives the UES a large mechanical advantage to resist bending moments (Macintosh & Bogduk, 1991; Macintosh et al., 1993). In the upright posture, the UES is thought to produce 74% of the total back extensor moment (Bogduk et al., 1992).

It has been shown that towards the extreme of lumbar flexion the moment arm of the UES significantly reduces by up to 39% (Macintosh et al., 1993).

LES attaches to the accessory and transverse processes of each lumbar vertebra and converges to form the lumbar intermuscular aponeurosis which inserts onto the ilium near to the posterior superior iliac spine (Bogduk, 2012). The intermuscular aponeurosis crosses multiple spinal segments but has a shorter lever arm than UES. It is estimated that in an upright posture, the LES contributes to 15% of total BE torque (Bogduk, 2012). Lumbar flexion also reduces the ability of the LES to produce extensor torque, although to a lesser extent than the UES (Macintosh et al., 1993).

Multifidus is the largest, deepest and most medial of the paravertebral muscles. It consists of short fascicles arranged segmentally, arising from one lumbar vertebra and descends to attach to one or two lumbar segments caudally, where eventually the lower lumbar fascicles attach onto the sacrum. Multifidus descends in an oblique orientation, fanning more lateral as it goes caudally (Bogduk, 2012). The overlying more superficial ES muscles make clinical examination of multifidus challenging, but below the third lumbar vertebrae only the ES aponeurosis exist over the top of multifidus making it accessible for palpation and surface EMG (Bojadsen et al., 2000). It is estimated that 11% of total BE torque in a neutral lumbar spine originates from multifidus (Bogduk, 2012). Unlike the UES and LES, an increase in lumbar flexion has a positive effect on increasing the moment arm of multifidus by up to 30%, which is attributed to the more vertical orientation of the muscle fibres that span between adjacent segments (Macintosh et al., 1993).

Another key factor that influences back extensor torque of the paravertebral muscles, is sarcomere length and the position of the overlap of the actin and myosin filaments (Bogduk, 2012; Brown & Gerling, 2012; Macintosh & Bogduk, 1991; Raschke & Chaffin, 1996). Flexed lumbar postures elongate the paravertebral muscle fibres and position the overlap of actin and myosin filaments more optimally for generating extensor torque (Brown & Gerling, 2012). This is supported by biomechanical studies that found peak isometric torque capacity and optimal length-tension of the BE muscles to occur at 45^{o} of lumbar flexion (Raschke & Chaffin, 1996). Others consider optimal position for torque to be generated by contractile tissue at up to 80% of lumbar flexion (Dolan et al., 1994).

While length-tension relationships of the ES muscles may be more optimal near end range of lumbar flexion, muscle pennation angle and torque generating capacity of the LES may be reduced in fully flexed lumbar postures. Studies utilising diagnostic ultrasound have shown that when the spine moves from an upright posture to full flexion, the pennation angle of the LES reduces from 28^{o} to 10^{o} (McGill et al., 2000; Singh, Bailey, & Lee, 2011). The change in muscle fascicle obliquity reduces the mechanical advantage of muscle fibres that generate extensor torque and resist shear forces (McGill et al., 2000) (Figure 2.1).

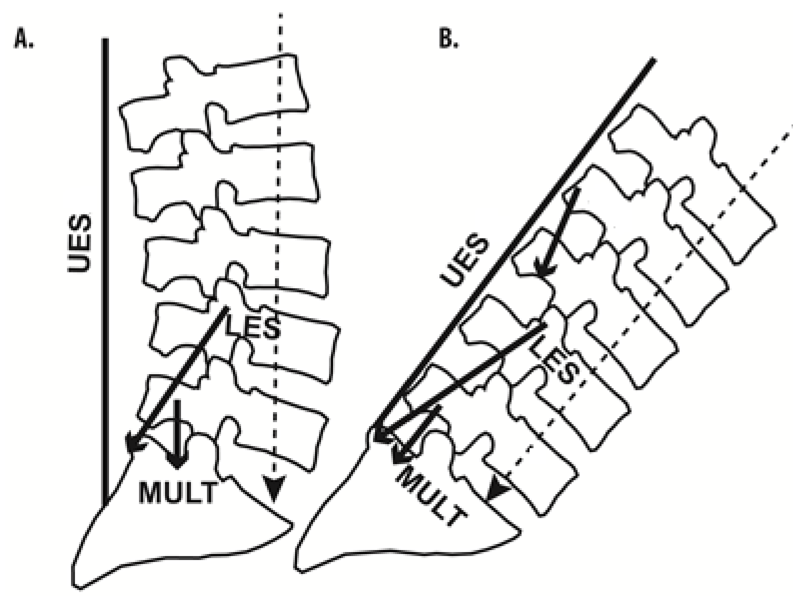


Figure 2.1 The change in muscle fibre obliquity with change from an upright (A) to a fully flexed (B) posture (adapted from Mawston & Boocock (2012))

In summary, while an increase in lumbar flexion improves the ability of the paravertebral muscles to generate extensor torque because of optimal length-tension and sarcomere positioning, a reduction in moment arm length and fascicle obliquity of muscle fibres has the potential to reduce ES muscle extensor torque production and to resist anterior shear forces towards maximal lumbar flexion. The combination of increased torque with flexion and an inability of muscles to control a rise in shear may present a risk for LBI.

2.4 Lumbar Posture

2.4.1 Lumbar posture definitions

An array of methods is used to measure lumbar posture, each having its own classification using different anatomical parameters. Definitions often vary according to body segments, which can include two or more lumbar vertebrae, incorporation of the thoracic spine and/or including the sacrum and pelvis (Table 2.2). Trunk inclination can also be used as an indicator of lumbar posture even though it may predominantly involve hip flexion and minimal lumbar spine movement. The definitions of lumbar posture include; lumbosacral angle, pelvisacral angle, trunk inclination, lordotic curvature, lumbar vertebral angle and lumbar curvature. The definition of these terms is outlined in Table 2.2.

Table 2.2 Definitions of lumbar posture

Author, year	Terminology	Definition	Measurement method
Bogduk, (2012)	Lumbosacral angle	The wedge shape of the L5/S1 disc and described by the angle of the intersection of the line through the inferior surface of L5 with the line through the upper surface of the sacrum	X-ray
Lavender, Trafimow, Andersson, Mayer, & Chen, (1994)	Trunk inclination	The motion produced from the lumbar and lower thoracic vertebrae measured by inclining the trunk forward in the sagittal plane measured from T7	Inclinometer
During, Goudfrooij, & Keessen (1985)	Lordotic curvature	The shape of the lower lumbar vertebral column created by constructing a circle through the frontal vertices of the vertebral bodies, defined by three points, L4, L5 and S1	X-ray
Burgess-Limerick & Abernethy (1997)	Lumbar vertebral angle	The angle subtended by the lines joining T1, PSIS and the ASIS to the PSIS	Photographs with retro-reflective markers
During et al. (1985)	Pelvisacral angle	The angle between the line through the centre of the hip joints and the centre of the superior surface of sacrum and the tangent line through the upper vertices of the sacrum	X-ray
Roy, Keller, & Colloca (2003)	Sagittal posture	The anterior or posterior inclination of the trunk that requires flexion or extension movement of the lumbar and thoracic spine without the involvement of pelvic movement	Isostation,B200 (isokinetic machine)
Adams & Dolan (1991); Dolan & Adams (1993); Mannion & Troke (1999)	Lumbar curvature:	The movement that occurs in the sagittal plane between S1 and L1 associated with and/or without pelvic movement	Electromagnetic Fastrak device

For the purpose of this thesis, one of the more encompassing definitions for lumbar posture that incorporates the whole of the lumbar spine was considered to be, the vertebral movement that occurs in the sagittal plane between the S1 and L1 vertebra associated with or without pelvic movement (Dolan & Adams, 1993). Vertebral movement in the sagittal plane produces the physiological movement of flexion and extension and is the main movement the lumbar spine performs. Concurrent with the physiological movement of lumbar flexion is forward vertebral translation (Bogduk, 2012).

The amount of sagittal plane movement between L1 and S1 varies greatly within the literature (Intolo et al., 2009). In a study using electromagnetic sensors (Polhemus Fastrak device) attached to L1 and S1 on 149 younger healthy adults, lumbar flexion was performed with participants lower limbs and pelvis secured in a standing frame. Lumbar range varied from 34^{o} to 75^{o} with a mean of 55^{o} (Dolan et al., 1994). Other studies that have also measured lumbar curvature between L1 and S1 using a Fastrak lumbar motion device, where the pelvis and lower limbs were unrestrained, reported a mean range of lumbar flexion in young healthy adults of 67^{o} (Hindle, Pearcy, Cross, & Miller, 1990) and 71^{o} (Pearcy & Hindle, 1989). A systematic review on lumbar range of motion found a significant reduction in lumbar flexion with increase in age (Intolo et al., 2009). Gender differences also exist with females having more range than their male counterparts (Intolo et al., 2009). It would seem that even when posture is clearly defined and measured using the same device, the amount of lumbar motion can vary greatly between individuals.

2.4.2 Posture and back extensor torque

Lumbar posture is known to impact upon the ability of the BE muscles to generate torque. When the lumbar spine becomes more flexed, the paravertebral muscle elongate and this alter their length-tension and consequently, the torque generating capacity of the paravertebral muscles (Ward et al., 2009). Several studies have investigated the relationship between lumbar posture and BE torque and it is consistently shown that when the spine assumes an increasingly flexed posture the magnitude of BE torque also increases (Graves et al., 1990b; Holmes et al., 1996; Parnianpour, Campello, & Sheikhzadeh, 1991; Roy et al., 2003; Smidt et al., 1983; Tan et al., 1993). Conversely, when adopting a more lordotic posture the capacity to generate BE torque diminishes. The described lumbar posture-torque relationship appears in, males and females (Graves et al., 1990b; Roy et al., 2003; Smidt et al., 1983), younger participants (Roy et al., 2003; Tan et al., 1993), older adults (Holmes et al., 1996), healthy individuals (Graves et al., 1990b; Parnianpour et al., 1991; Roy et al., 2003) and those with LBP (Holmes et al., 1996; Smidt et al., 1983; Smith, Bissell, Bruce-Low, & Wakefield, 2011). The same finding also occurs irrespective of method used to measure torque.

There is a gender difference in the torque generating capacity of the BE muscles, with males being stronger than females (Graves et al., 1990a; Keller & Roy, 2002; Smidt et al., 1983). Smidt et al. (1983) investigated maximal isometric back extension torque on 12 females and 12 males using a trunk dynamometer in a seated position from 20^{o} of extension to 40^{o} of lumbar flexion in 20^{o} increments. Men were found to be significantly stronger than females, on average by 57%. These findings also concur with Graves and colleagues (1990a) who undertook a similar protocol using 56 men and 80 women, in which males were found to be stronger than females in all lumbar postures. These gender differences still remained when torque was normalised per unit of body weight. Keller & Roy (2002) also found a gender difference in maximal isometric BE torque, but when torque was normalised according to body weight and height, no gender differences existed in torque production. The importance of normalising data in order to compare genders was further highlighted by Roy et al. (2003) who tested the maximal isometric torque of 10 male and 10 female participants from 20^{o} of extension to 50^{o} of flexion while standing with pelvis and lower limbs fixated in a B200 isostation device. When torque in each lumbar posture was expressed as a percent of torque in the upright standing, no gender differences existed.

The magnitude of BE torque is not only dependent upon gender but also the test position. Previous studies tend to have used either standing or sitting positions. In standing, the BE muscles have been shown to produce maximum torques between 164 Nm and 265 Nm at the end flexion test position in young adults (Roy et al., 2003; Tan et al., 1993). In contrast, studies using a sitting test procedure have found far greater back extensor torques of between 320 Nm and 345 Nm at end flexion (Graves et al., 1990b; Smidt et al., 1983). These extensor torque values are based upon 0^{o} of flexion to be the upright standing or sitting positions. Arguably, when assuming upright sitting, 90^{o} of hip flexion will induce posterior rotation of the pelvis which will place the lumbar spine into more flexion compared to upright standing (Yasukouchi & Isayama, 1995). Therefore, when testing BE torque in sitting, subjects are more likely to adopt postures where the lumbar spine is more flexed compared to standing (Callaghan & McGill, 2001). The alteration in low back joint loading and kinematics during standing compared to unsupported sitting may partly explain why higher torques are recorded in sitting than upright standing.

The angle-torque relationship is also noteworthy. Studies performed in standing showed a linear increase in torque from 0^{o} of flexion to 50^{o} (Parnianpour, Li, Nordin, & Kahanovitz, 1989; Roy et al., 2003; Tan et al., 1993) (Figure 2.2). In studies testing BE torque in sitting the torque-angle relationship is more variable between studies with Graves et al. (1990b) and Holmes et al. (1996) finding only a small increase in torque from neutral upright sitting to end range of flexion (Figure 2.3). In contrast Smidt et al. (1983) found the biggest change in torque output in the early range of flexion with similar levels of torque in 20^{o} and 40^{o} of lumbar flexion. Interestingly, at the end test position in sitting, torque production appears to reduce in some cases (Graves et al., 1990b; Smidt et al., 1983).

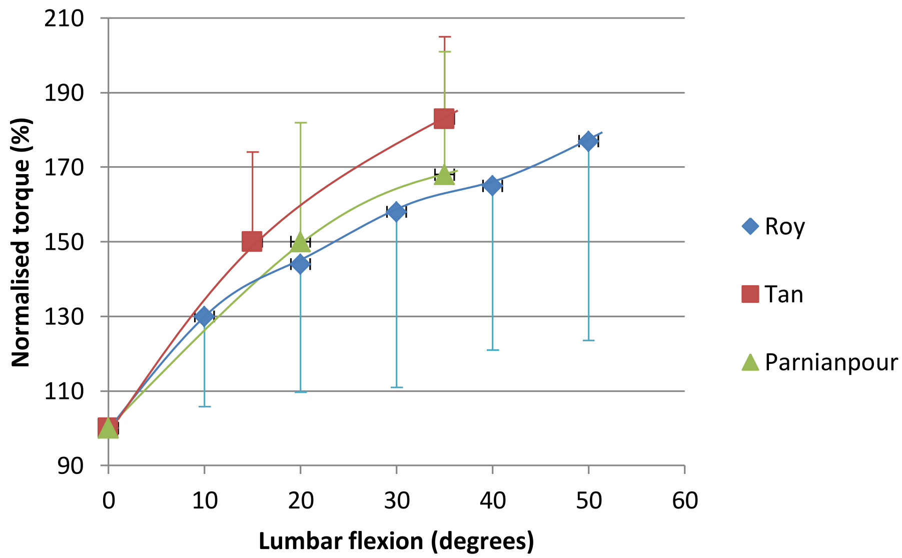


Figure 2.2 Normalised torque for posture in studies undertaken in standing

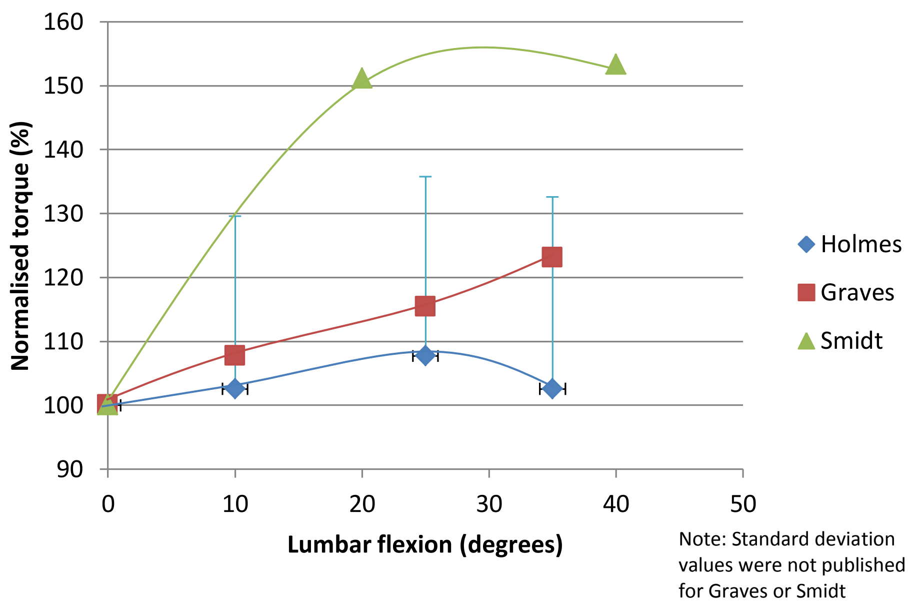


Figure 2.3 Normalised torque for posture in studies undertaken in sitting

There have been few studies that have assessed extensor torque at the extreme of lumbar flexion. Previous studies have measured torque production at absolute angles of lumbar flexion up to 50^{o} of lumbar flexion. For example, the end test position used by Tan et al. (1993) and Parnianpour et al. (1989) was 35°, which equates to only 50% percent of total lumbar flexion (Pearcy & Hindle, 1989). The most flexed test position adopted in a study was 50° (Roy et al., 2003), which was approximately 70% of maximal lumbar flexion, according to Pearcy & Hindle (1989). Further research is needed to assess extensor torque in end range flexion as this posture is often adopted during lifting (Hoogendoorn et al., 2000) and this is when there is significant recruitment of passive structures of the spine, which is associated with risk of injury (Dolan et al., 1994; Hoogendoorn et al., 2000).

2.4.3 Posture and paravertebral muscle activation

Lumbar posture is known to influence paravertebral muscle activation (Kaigle et al., 1998; Kippers & Parker, 1984; Roy et al., 2003; Tan et al., 1993; Toussaint et al., 1995). Muscle activation is commonly quantified using amplitude measures of surface or fine wire EMG (Soderberg & Knutson, 2000). Only a small number of studies have examined activation of the paravertebral muscles during maximal isometric voluntary contraction at different points of lumbar flexion.

One of the first groups to investigate the effects of lumbar posture on ES muscle activation was Marras, King & Joynt (1984) who used fine wire EMG to examine levels of the left and right ES activation during maximal isometric back extension, with minimal fixation, when standing in neutral (0°), mid (22.5°) and near-end range (45°) lumbar flexion. EMG data were normalised against the maximum value recorded across all lumbar postures. Despite not stipulating the exact lumbar spinal level where ES activation was recorded, activation patterns were similar for left and right ES. ES were found to be less activated in mid and near end-range postures when compared to upright standing (neutral posture). However, ES activation levels did not differ significantly between mid and near-end range postures where it was thought an increase in latissimus dorsi recruitment, due to its superior mechanical advantage, may have compensated for a reduction in ES activation with flexion.

Tan et al. (1993) also undertook a similar study to that of Marras et al. (1984), looking at the effect of lumbar posture on maximum isometric BE torque in three lumbar postures (0°, 15° and 35° flexed). Thirty-one healthy men were positioned with their pelvis and lower limbs fixated into in a B200 isostation dynamometer with muscle activation being recorded bilaterally form the medial and lateral ES, level with L3. No difference in activation levels were found from the medial and lateral portions of ES. In contrast to Marras and colleagues, Tan et al. found activation levels of ES to significantly increase from upright standing to 15^{o} flexion. Activation levels continued to rise at a slower rate thereafter and no significant difference between the mid and end test point was found (Figure 2.4).

More recently, Roy et al. (2003) examined male and female participants maximum isometric back extensions while recording activation bilaterally from the LES (level with L3) using surface EMG. Participants were placed in an isokinetic machine with their pelvis and lower limbs restrained. Lumbar motion was tested from 20^{o} of extension to 50^{o} of flexion in 10^{o} increments, with 0^{o} being upright standing. No difference was found between the activation levels of left and right LES. From 20^{o} of extension to 30^{o} of flexion the activation levels between males and females were similar. From 40^{o} onwards, men continued to increase muscle recruitment, while females exhibited a decline in activation. This gender difference was not statistically significant resulting in a net reduction in activation of the LES at 50^{o} compared to the peak at 40^{o} (Figure 2.4).

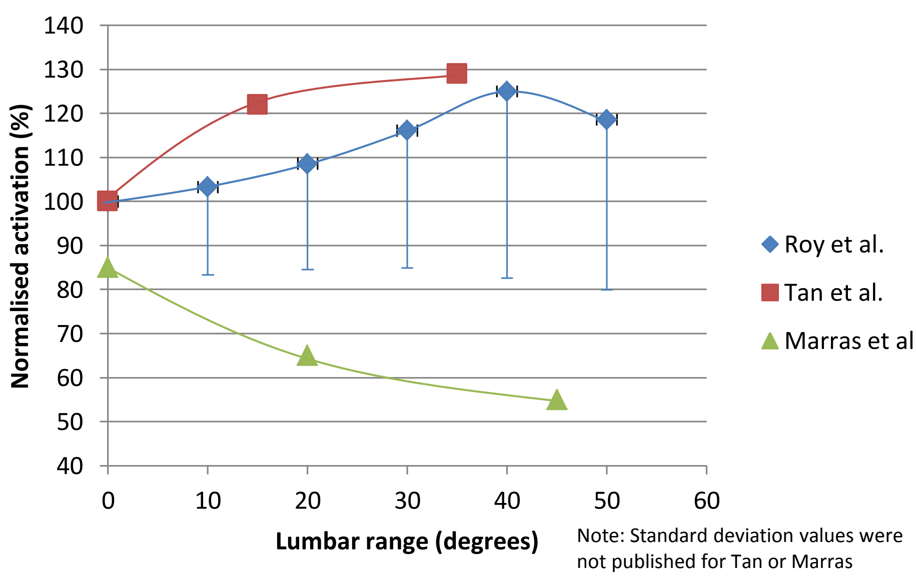


Figure 2.4 The effect of lumbar posture on the mean activation of LES

As shown in Figure 2.4, the most flexed test position in the three studies examining MIVC of ES was 50^{o}. Based on average range of motion data from Pearcy & Hindle (1989), who validated measuring lumbar range of motion with the Fastrak device against x-ray, it is unlikely that subjects performed MIVC in full lumbar flexion. While other studies have recorded ES muscle activation in end range lumbar flexion, their findings were not published because activation levels were not the main focus of their study (Dolan & Adams, 1993; Dolan et al., 1994). Therefore, it remains unclear as to how activation levels of the paravertebral muscle may behave at the extreme of lumbar flexion during maximum isometric contractions.

Previous studies have investigated EMG activation levels at the end range of lumbar flexion using submaximal efforts and during dynamic back extension tasks (Dolan & Adams, 1993; Dolan et al., 1994; Kaigle et al., 1998; Kippers & Parker, 1984; Marras, Rangarajulu, & Wongsam, 1987; Toussaint et al., 1995). In these studies it has been found that towards the end range of lumbar flexion ES muscle recruitment significantly diminishes (Dolan et al., 1994; Kippers & Parker, 1984; Toussaint et al., 1995). Dolan et al. (1994) found that when more than 80% of available lumbar motion had occurred, the LES activation sharply declined. Others have shown a noticeable decline in activation of the LES beyond 90% of flexion (Kippers & Parker, 1984). At the extreme of lumbar flexion it has been frequently reported that the paravertebral muscles become electrically silent even when carrying a load of 10-15kg (Dolan et al., 1994; Kippers & Parker, 1984; Toussaint et al., 1995). Electrical silence of the paravertebral muscles when approaching full flexion has come to be known as the 'flexion-relaxation phenomenon' (Dolan et al., 1994). EMG fidelity is lost during dynamic movement (De Luca, 1997; Hof, 1984) and an increase rate of change in lumbar curvature alters activation patterns of ES (Dolan et al., 1994; Marras et al., 1984). Therefore, it remains unknown if the 'flexion-relaxation phenomenon' would continue to exist at the extremes of flexion during maximal isometric contractions.

While there is good evidence for lumbar posture influencing the level of muscle activation of LES, less is known about the influence of lumbar posture on other divisions of the paravertebral muscles such as the UES and multifidus. Dolan et al. (1994) recorded activation of the UES as well as the LES during maximal isometric extensions from “slight” lumbar flexion to the extreme of flexion. The results found the UES to be more activated than the LES throughout all lumbar postures, although the difference was not statistically significant. Toussaint et al. (1995) recorded activation of the UES, LES and also multifidus, for six male participants while they performed a lowering and lifting tasks of a 15kg weight. While the LES and multifidus displayed the ‘flexion-relaxation phenomenon’, the UES remained consistently activated throughout the lifting cycle with no evidence of relaxing at the extreme of flexion, irrespective of carrying a heavy load or not. It was suggested that the different divisions of paravertebral muscles appear to function independently of each other, indicating that recording activation at a single spinal level may not be sufficient to represent all the different divisions of the paravertebral muscles acting upon the lumbar spine. Whilst Toussaint et al. illustrated a different response between the UES and LES with change in lumbar posture, no study to date appears to have specifically reported on the influence of lumbar posture on levels of activation of the different divisions of the paravertebral muscles when performing simulated lifting tasks at maximal effort. Such information would assist in clarifying if the ‘flexion-relation phenomenon’ continues to exist at end range of lumbar flexion during maximal lifting or whether the spine remains under active muscular control.

2.4.4 Posture and neuromuscular efficiency

Expressing the amount of torque that can be produced per unit of muscle activation (torque/activation), in any given lumbar posture can provide information on the active (muscles) and passive (discs and ligaments) tissue contributions to extensor moment (Tan et al., 1993). NME can be estimated from several studies where concurrent measurements of maximal isometric torque and activation are reported (Marras et al., 1984; Roy et al., 2003; Tan et al., 1993). In the study by Marras et al. (1984) maximal isometric BE torque was measured in 0^{o}, 22.5^{o} and 45^{o} of trunk inclination (the angle between the trunk and thigh) concurrent with EMG activation of the lumbar ES. Data were normalised against the maximum value found for each variable. Even though NME was not directly calculated by the authors, using the available data, NME was found to rise uniformly with an increase in flexion (Figure 2.5). NME was directly calculated by Tan and co-workers (1993) using maximal BE torque and muscle activation of the medial ES adjacent to L3, at 0°, 15° and 35° of lumbar flexion, with data normalised to the upright posture (0°). As with Marras et al. a very similar NME trend was found (Figure 2.5). Roy and colleagues (2003) also normalised maximal isometric BE torque and activation levels of the LES to the upright posture to calculate NME between 0° and 50° of lumbar flexion. There was an overall upward trend in NME with an increase in lumbar flexion, but after 10° of flexion the rate of change became less. Nonetheless, there appears to be a clear pattern linking an increase in lumbar flexion with a rise in NME with end test point being the most efficient (Figure 2.5).

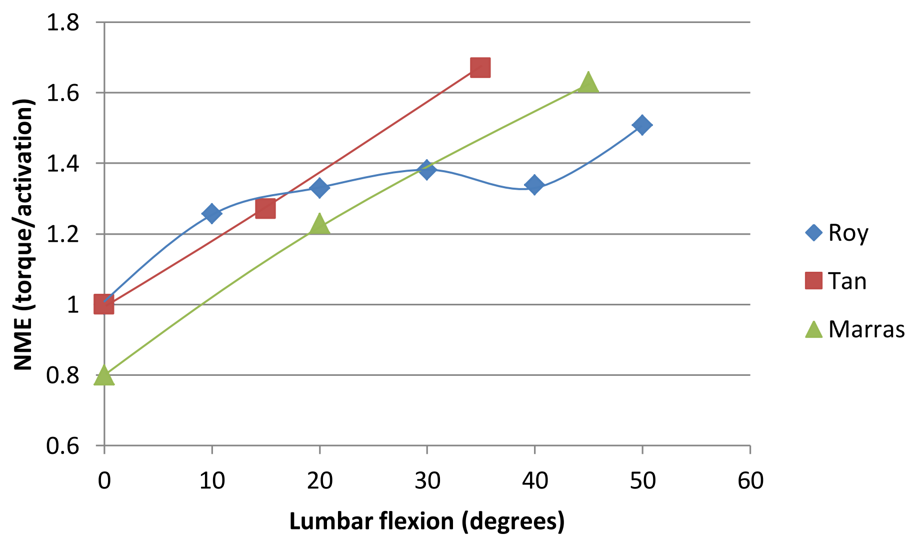


Figure 2.5 Neuromuscular efficiency of LES

A number of theories have been developed to explain why neuromuscular efficiency improves as lumbar spine flexion increases. While the relationship between muscle activation and torque is considered to be linear (Dolan & Adams, 1993; Hof, 1984) if torque production was proportional to activation in each posture, NME would remain relatively unchanged. One theory giving rise to improved NME relates to muscle elongation reducing the volume of muscle units below the electrode which diminishes the amount of signal detected (De Luca, 1997; Tan et al., 1993). Other groups (Hof, 1984; Tan et al., 1993; Ward et al., 2009) suggested that with an increase in lumbar flexion the paravertebral muscles approach their optimal muscle length-tension, which is likely to create more efficient torque production.

While torque improves with optimal muscle length-tension, torque is not solely produced by active tissue alone, but also from tension within passive tissues (disc, ligaments and connective tissues within muscles) (Bogduk, 2012; Dolan et al., 1994; Hill, 1938; Potvin et al., 1991). Dolan et al. (1994) investigated the passive tissue contribution to BE torque throughout the lumbar flexion range. Measurements of maximal isometric back extensions and activation of ES were recorded from a 'slight' flexed to fully flexed lumbar posture. A series of equations was used to calculate the gradient of the relationship between torque and activation in each posture and where the gradient crossed the Y axis (no muscle activity), was considered the passive tissue contribution. They concluded passive tissue contribution to be small from early to mid lumbar flexion and then it becomes exponentially larger when more than 80% of available lumbar flexion is exceeded. Passive tissue contribution at the extreme of lumbar flexion is also supported by anatomical studies showing engagement of resistance of the ligaments and disc towards the end of lumbar flexion (McGill, 1988).

With an exponential increase in passive tissue contribution towards the end of lumbar flexion, a sharp increase in NME could be expected to be seen. In the studies where NME has been estimated the highest degree of lumbar flexion used was 50^{o} (Roy et al., 2003). However, the NME at the extreme of flexion is unknown. At the end range of lumbar flexion, high NME efficiency appears advantageous as more torque can be produced with less muscle activation, leading to low metabolic demands. However, it is the loading of the passive structures, in particular the interspinous ligament complex, which increases anterior shear forces within the spine and this has the potentially to cause tissue damage (McGill, 1988; Potvin et al., 1991).

The NME ratio of trunk extension has been calculated based on the LES muscle activation with no known studies to date having determined the NME of the UES and multifidus muscles in different lumbar postures. Dolan et al. (1994) have measured the UES and LES activation levels at the extreme of flexion and have found the UES to remain more activated than the LES at submaximal and maximal effort. The higher levels of activation in the UES at the end of flexion would result in a lower NME than the LES, and would indicate that the active contribution of the ES muscles to extensor torque in extreme of lumbar flexion may be greater than previous thought. Considering the UES is thought to contribute up to 74% of total BE torque (Bogduk, 2012), knowing its NME could have a significant bearing of the overall NME of the lumbar spine.

2.5 Pelvis Fixation

Pelvis fixation is often used during measurements of BE torque and paravertebral muscle activation to isolate trunk function from that of the pelvis (Dolan et al., 1994; Graves et al., 1990a; Kaigle et al., 1998; Kippers & Parker, 1984; Parnianpour et al., 1991; Roy et al., 2003; Smidt et al., 1983; Tan et al., 1993; Toussaint et al., 1995). Pelvis fixation is proposed to isolate the trunk by preventing de-rotation of the pelvis and minimising the contribution of the hip extensors (Graves et al., 1994; Walsworth, 2004). Without fixation, it is thought the paravertebral muscles will only work isometrically to stabilise the spine and it would be the hamstrings and gluteal muscles producing the prime movement of extension (Graves et al., 1994; San Juan et al., 2005). However, the rationale for pelvis fixation requires closer scrutiny and will be explored further in this section.

2.5.1 Pelvis fixation and back extensor torque

Considering the widespread use of pelvis fixation during BE torque studies (Dolan et al., 1994; Roy et al., 2003; Smith et al., 2011; Tan et al., 1993), only one study was identified testing the effect of pelvis fixation on BE torque (Smidt et al., 1983). Smidt et al. (1983) tested 13 young participants and compared the effect of full pelvis fixation, partial fixation (feet and thighs restrained but not the pelvis) and minimal fixation (only the feet fixated) on BE torque at varying degrees of lumbar flexion, while seated on the Iowa trunk dynamometer. While no significant difference was found between partial and full pelvis fixation, a significant reduction in torque output was found when minimal fixation was used.

Pelvis fixation and torque production of the BE muscles has been examined in studies investigating optimal BE muscle strengthening regimes (Da Silva, Lariviere, Arsenault, Nadeau, & Plamondon, 2009; Graves et al., 1994; Smith et al., 2011; Udermann et al., 1999). These studies tend to use isokinetic apparatus requiring participants to be seated with and without pelvis fixation. Maximal BE torque was measured before and after undertaking a strengthening regime using isokinetic machines where the level of pelvis fixation was manipulated (Graves et al., 1994; Smith et al., 2011). Some participants trained on the MedX™ whilst others were allocated to alternative machines considered to give less pelvis fixation. Each group only trained once a week performing 8-12 repetitions. The group that trained on the MedX™ significantly increased BE torque compared to participants who trained on other devices. It was concluded the MedX™ was more specific in targeting the paravertebral muscles due to pelvis fixation (Graves et al., 1994; Smith et al., 2011). The infrequency of training undertaken made it difficult to determine whether BE torque gains were due to pelvis fixation or if the MedX™ group improved due to a training effect of using the MedX™ apparatus. Methodological issues with these studies leave it unknown as to the effect pelvis fixation has on strengthening and torque production of the BE muscles.

Udermann et al. (1999) also investigated the effect of pelvis restraints on BE function and noted that it was virtually impossible to test lumbar extension torque in sitting without some degree of pelvis fixation. This was also recognised in other studies that compared full and partial fixation to no fixation (Da Silva et al., 2009; Walsworth, 2004). It is likely the apparatus used to measure BE torque dictated the need to fixate the pelvis for its effective use, rather than fixation specifically targeting the paravertebral muscles.

2.5.2 Pelvis fixation and muscle activation

A direct method for investigating the influence of pelvis fixation on BE muscles is through muscle recruitment and amplitude measures using surface EMG. The effect of pelvis fixation on recruitment of the paravertebral muscles has been examined in several studies (Benson, Smith, & Bybee, 2002; Da Silva et al., 2009; San Juan et al., 2005;

Udermann et al., 1999; Walsworth, 2004), although only under concentric or eccentric conditions, with no known studies specifically using isometric protocols. Within these studies using dynamic protocols the results are conflicting on the effect of pelvis fixation on ES activation. Two studies examined the effect of pelvis fixation on ES muscle activation during dynamic back extension tasks performed at 40-50% of MVC in a seated position (Da Silva et al., 2009; San Juan et al., 2005). Fixation was found to significantly increase muscle activation in the UES and LES. However, partial and full pelvis fixation resulted in similar activation levels, which in turn, were greater than those produced with no fixation (San Juan et al., 2005). Conversely, in another study, the MedX™ was used to measure activation of paravertebral muscles adjacent to L1 and L5 during maximal isometric and dynamic back extension tasks. ES activation levels were similar for pelvis fixation and no fixation conditions (Udermann et al., 1999; Walsworth, 2004). Walsworth (2004) found no difference in levels of the LES activation during seated back extensions on the MedX™ (thought to give 'rigorous' pelvis fixation) compared to the Cybex (which only fixates the lower limbs and not the pelvis). It remains unclear as to the affect pelvis fixation has on paravertebral muscles, but it does appear that activation levels do not reduce with pelvis fixation.

Another purpose of pelvis fixation is to minimise the contribution of the hip extensors in order to facilitate the recruitment of the BE muscles. Two studies used surface EMG over the hip extensors to investigate the effect of pelvis fixation on hip extensor recruitment during dynamic back extension tasks, using 40-50% of maximum effort (Da Silva et al., 2009; San Juan et al., 2005). Results showed no significant differences in activation levels of the hip extensors between pelvis fixation, partial fixation or when the pelvis was free to move. Udermann et al. (1999) also used surface EMG on the gluteal and hamstring muscle groups during dynamic back extensions using a load of 80% of body weight. They found no effect of fixation on hamstring muscle activation, but the gluteals were found to be more active with the pelvis restrained compared to no fixation. It remains unclear whether pelvis fixation can minimise the contribution of the hip extensors during back extension tasks in order to facilitate the recruitment of the paravertebral muscles.

2.6 Summary

From the literature review there are apparent gaps within the research pertaining to the effect lumbar posture and pelvis fixation have on BE torque and paravertebral muscle activation. The majority of studies investigating maximal BE torque and the effect of lumbar posture have used a maximum of 50^{o} of lumbar flexion. Considering that the lumbar spine can have up to 70^{o} of flexion in young healthy adults, less is known about BE torque at the extremes of lumbar flexion. Furthermore, the vast majority of research investigating BE torque is based upon experiments where participants are fixated in isokinetic machinery and not while performing functional tasks such as lifting, leaving it unknown if what we presently know about BE torque is transferable to functional tasks.

Muscle activation of ES has been shown to increase with lumbar flexion although Marras et al. (1984) found contrasting results where ES activation reduced with flexion. Akin to the research on extensor torque, maximum activation beyond 50^{o} of flexion appears to be less understood. The majority of research has focused on ES activation at a single spinal level, targeting LES. There is some evidence to suggest that the UES may respond differently to lumbar postural changes compared to the LES but there is no consensus within the literature, with little consideration given to the contribution of multifidus.

Pelvis fixation is frequently used when examining the effect of lumbar posture on BE torque and muscle activation in an attempt to isolate trunk function from that of the pelvis and lower limbs. However, the ability of pelvis fixation to reduce the hip extensor contribution to BE torque and/or to increase activation levels of the paravertebral muscles appears to be conflicting between studies. Furthermore, fixating the pelvis is not replicating functional positions or the actions of how the back and its muscle systems would normally work together.

The disparities within the literature, as outlined above, create the purposes of this study with the aim to extend present knowledge on the biomechanics of the lumbar spine.

2.7 Aims & Objectives

The aim of this study was to determine how lumbar posture and pelvis fixation affected BE torque and level of muscle activation of three divisions of the paravertebral muscles.

The objective of this study was to measure isometric BE torque and muscle activation of three divisions of the paravertebral muscle (UES, LES and multifidus) in three lumbar postures (fully lordotic, fully flexed and mid way between the former postures), with and without the pelvis fixated, using a 3D force gauge and 3D motion analysis system to measure BE torque, electromagnetic sensors to assess lumbar posture and surface EMG to record muscle activation, respectively.

2.8 Purpose of the Study

The purposes of this study were based on its aims and objectives in order to investigate the following hypotheses:

1. The effects of lumbar posture on BE torque. It was hypothesised that back extensor torque would increase with lumbar flexion.
2. The effects of lumbar posture on paravertebral muscle activation. It was hypothesised that muscle activation would initially increase with torque production but reduce towards the end range of flexion.
3. The effects of lumbar posture on the UES, LES and multifidus muscle activity. It was hypothesised that the relative muscle activity of each division of the paravertebral muscles (UES, LES and multifidus) would increase with an increase in lumbar flexion
4. The effect of neuromuscular efficiency (NME) of the paravertebral muscles. It was hypothesised that NME would increase from the lordotic posture, to the fully flexed posture.
5. The effect of pelvis fixation on isometric BE torque and paravertebral muscle activation. It was hypothesised that pelvis fixation would increase torque production and muscle activation.

Chapter 3: Methods

3.1 Introduction

This chapter describes the methods used to investigate the hypotheses of this study. The methods section will begin by outlining the study design and describing the recruitment process of participants, which includes the criteria used for participant selection. Following this, independent variables will then be defined and a detailed description of kinetic, kinematic motion analysis and electromyography (EMG) measures will be presented. The experimental procedure will be described along with data processing methods and statistical analyses.

3.2 Methods

3.2.1 Study design

A repeated measures experimental design was used to investigate the effect of lumbar posture (flexed, mid and lordotic) and pelvis fixation (fixed and free) on the magnitude of BE torque and the level of three paravertebral muscle activation (adjacent to L4/5, L3 and T10) during a maximal isometric trunk extension in a simulated lifting position. A total of six conditions were performed. Ethical approval was gained from the Auckland University of Technology Ethics Committee (AUTEC) (Appendix A).

3.2.2 Design overview

In this study participants were required to exert a maximal isometric back extension, lasting three to five seconds, in three different lumbar postures. Postures assumed were full lumbar flexion, maximal lumbar lordosis and the midpoint between these extremes. The lumbar posture was determined using a Polhemus Fastrak device (Polhemus Navigation, Kaiser Aerospace Inc., Vermont USA) with sensors attached to the skin overlying L1 and S1 spinous processes. The Polhemus Fastrak device has been shown to be a highly reliable measure of lumbar flexion and has been validated against x-ray measurements of the lumbar spine (Adams & Dolan, 1991). Each posture was tested with the pelvis free to move and with the pelvis restrained. A harness worn around the pelvis with four straps anchored to the floor was used to fixate the pelvis (Montgomery, Boocock, & Hing, 2011).

Estimates of BE torque were collected using two independent methods. The first method utilized a three dimensional floor mounted force gauge that connected via a chain to a chest harness worn by the participant. Maximal isometric back extension was exerted through the chain to the force gauge. The second method, which occurred concurrently with the first method, required the participant to stand with each foot on a separate force plate while exerting the maximal back extension. The force plates measured ground reaction forces for each lower limb independently.

Kinematic analysis was used to determine the position of the trunk, pelvis and lower limbs during the maximal back extension. Thirty six retro-reflective markers were placed over the body and equipment and their position in 3D space were captured by a nine-camera motion analysis system (Qualysis Medical AB, Gothenburg, Sweden).

The kinetic and kinematic data were combined to estimate BE moment acting about the lower back. The two methods used to calculate BE torque involved 1) force measurements from the force gauge, hereafter referred to as the external moment, and 2) force measurements from two force plates, forming an eight segment rigid-link biomechanical model, hereafter referred to as the internal moment.

Surface electromyography (EMG) of the UES, LES and multifidus muscles on the right side of the spine were measured and synchronized with the kinetic and kinematic data for the six different maximal isometric back extension test conditions. The level of back muscle activation recorded during back extensions was normalized to maximum isometric voluntary contraction performed in the Biering-Sorensen position.

3.2.3 Participants

Twenty-six healthy volunteers (13 males and 13 females) participated and completed this study. The inclusion and exclusion criteria for the study were:

Inclusion criteria:

- Aged 18-35 years old.
- Spoke and understood English.
- Self-reported to be medically fit and well.
- Able to attend one, two hour laboratory session at the Health Rehabilitation Research Institute (HRRI) at AUT University.

Exclusion criteria:

- Previous spinal surgery.
- Previous abdominal surgery in the last 12 months.
- Medically diagnosed with any neurological or rheumatological condition.
- History of low back pain (LBP) in the previous 12 months that required absence from work, LBP requiring medical treatment, or LBP lasted longer than three days.
- Pregnant or given birth in the last 12 months.

The recruitment process is outlined in Figure 3.1. Participants were recruited via word of mouth, poster advertisements on student and staff notice boards and announcements on electronic notice boards at the university's North Shore campus (Appendix B). Individuals who expressed interest in participating in the study were contacted by phone or by email to outline the requirements of the study and the risks involved. Eligible volunteers were sent a copy of the participant information sheet to read prior to testing (Appendix C). All participants who met the requirements of the study were invited to attend the HRRI where the test procedure was explained in more depth and a demonstration given, followed by an opportunity to ask questions. If the participants were satisfied with the test procedure, a consent form was completed (Appendix D) and an individual identification number allocated to each participant to conceal identity.

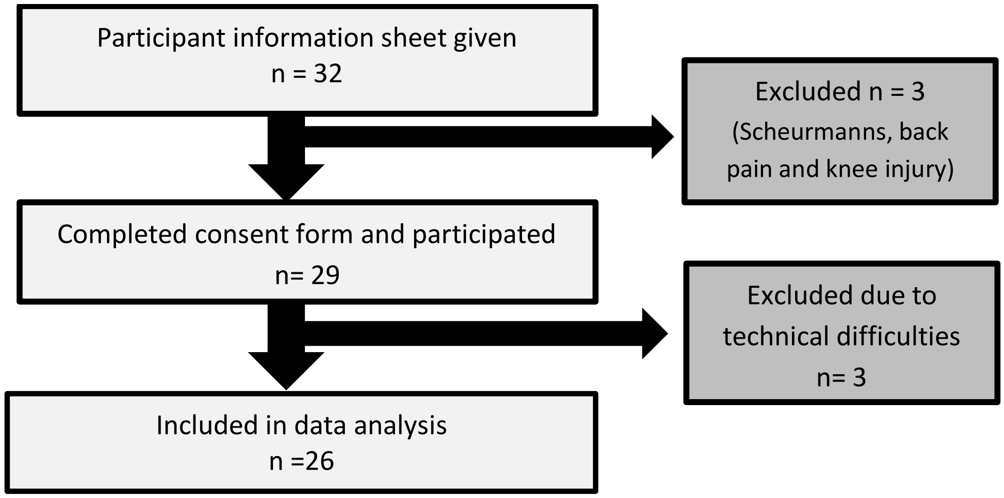


Figure 3.1 Recruitment process

3.2.4 Randomisation process

Each participant was randomly assigned to the initial level of fixation (fixed versus free) and the order in which each posture would be tested. There were six possible posture sequence combinations available (Table 3.1). Microsoft Excel software package was used to generate the random number sequences assigned to each pelvis fixation condition and posture sequence.

Table 3.1 Pelvis fixation and posture sequence

Pelvis fixation	Number	Posture sequence 1 2 3	Number
Fixed	1	Flexed, mid, lordotic	1
Free	2	Flexed, lordotic, mid	2
		Mid, flexed, lordotic	3
		Mid, lordotic, flexed	4
		Lordotic, mid, flexed	5
		Lordotic, flexed, mid	6

3.3 Experimental Measures

This study involved collection of kinetic data, kinematic data, measurement of lumbar posture and recording EMG measures of the BE muscles. The following section explains how these were undertaken.

3.3.1 Kinetic data

Two methods were used to measure force exertions:

A 3D force gauge (Advanced Mechanical Technology Inc., MC3A-100, USA) was used to measure maximal isometric force exerted by the trunk when extending the back. The force gauge was securely bolted to the floor located directly in front of the participant's feet. A custom-made metal plate with a loop attachment was bolted to the top of the force gauge to allow a chain to be attached. A chain was attached between the loop on the force gauge and the chest harness worn by the participant. A carabineer was attached to the harness end of the chain so it could be detached in between trials to allow the participant to return to an upright posture. The harness was made of woven non-stretch material that had an adjustable Velcro™ chest belt and adjustable shoulder straps to ensure the harness was located level with the sixth thoracic vertebrae (Figure 3.2). It was worn as a close fit with micro fleece padding for participant comfort. The force gauge data was collected at a sampling rate of 1200Hz and low pass filtered using a Butterworth second order filter with a cut off frequency of 6Hz.

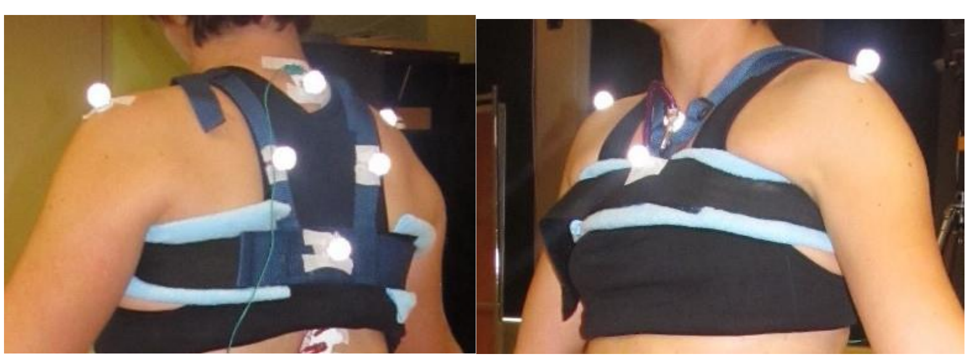


Figure 3.2 Chest harness marker position

Ground reaction forces when performing back extensions were also measured using two floor mounted force plates (Advanced Medical Technology Inc., USA). These recorded 3D ground reaction forces independently for the right and left lower limbs of each participant during maximal back extension exertions. These data were collected at a sampling rate of 1200Hz, smoothed and passed through a Butterworth second order filter with a cut off frequency of 6Hz due to the static nature of the test procedure.

3.3.2 Kinematic data

3.3.2.1 Motion analysis

Kinematic data of the trunk, pelvis and lower limbs were captured using a nine-camera ProReflex motion analysis system (Qualysis Medical, Sweden) using a collection rate of 60Hz. The motion analysis system was used to detect 36 retro-reflective markers attached to the participant. The cameras were positioned to allow best visual coverage of the experimental set up. The x, y, and z co-ordinate data from each retro-reflective marker were subsequently stored in Qualysis software prior to further analysis using Visual 3D biomechanics software package (C-motion Inc, version 4.0, USA).

3.3.2.2 Retro-reflective marker placement

Twenty retro-reflective markers (19mm diameter) were attached by an experienced physiotherapist to the anatomical landmarks on the participant. Five markers were also attached to the equipment, chest harness and force gauge, to identify its relative 3D position within the laboratory (Figure 3.2 and Table 3.2). The retro-reflective markers were attached with double backed adhesive tape (3M St.Pauls, MN, USA). Reinforcement strips of Fixamull™ were applied to each marker to prevent dislodgement during testing. An additional 11 tracking marker were also attached to the thigh, shank and trunk to assist in the identification and biomechanical modelling of each body segment. These were chosen based on recommendations by Cappozzo, Della Croce, Leardini, & Chiari (2005). While errors can arise from the placement of retro-reflective markers on the skin surface with body movement, passive marker systems are non-invasive, do not restrict natural movement patterns and have been shown to have a high degree of accuracy (Cappozzo et al., 2005).

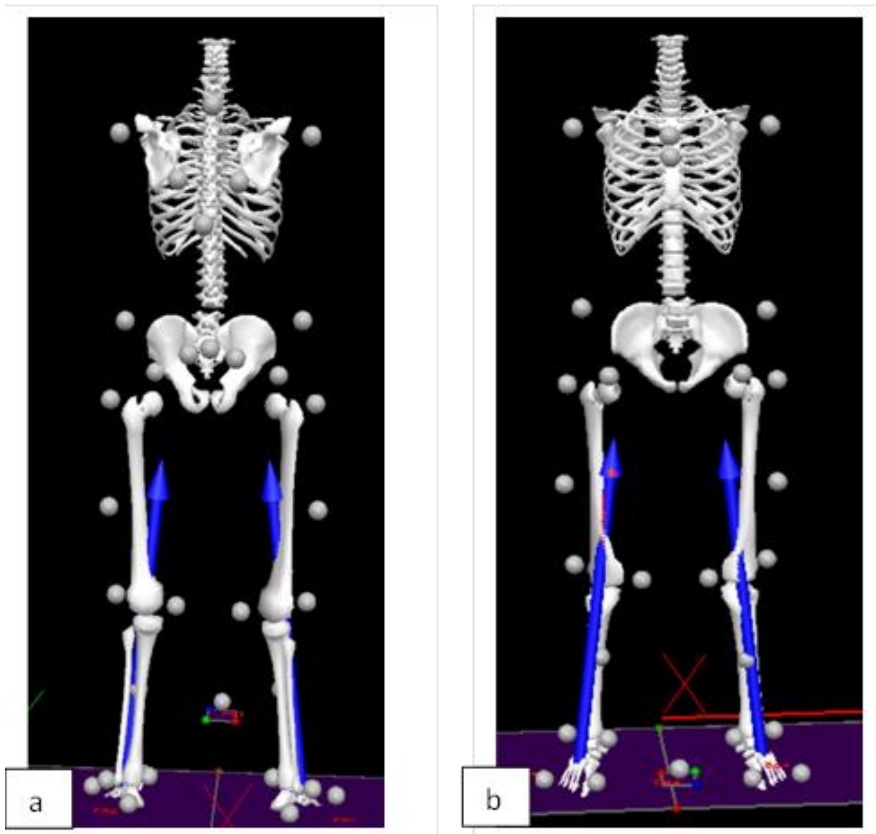


Figure 3.3 Adapted from Visual 3D; view of marker locations a. Posterior view b. Anterior view

Table 3.2 Anatomical location of retro-reflective markers and body segments formed

Retro-reflective markers	Type	Body Segment formed
Medial joint line of first metatarsal phalangeal joint	B	
Lateral joint line of fifth metatarsal phalangeal joint	B	
Posterior heel, three centimeters from floor	B	Foot
Distal tip of medial malleolus	B	
Distal tip of lateral malleolus	B	
Mid spine of tibia	T	Leg
Lateral knee joint line	B	
Medial knee joint line	T	
Mid lateral thigh	T	Thigh
Greater trochanter	B	
Posterior superior iliac spine	B	Pelvis
Apex of iliac crest	B	
Fifth lumbar spinous process	T	
Tip of acromion	B	Thorax
Sternal notch	T	
Posterior chest belt	E	
Anterior chest belt	E	
Anterior superior iliac spine	T	
Top of the force gauge	E	Additional markers
First thoracic spinous process	T	

B = body marker T = tracker marker E= equipment marker

3.3.2.3 Biomechanical modelling

Markers were initially identified, tracked in the Qualysis software package (Qualysis Track Manager) and exported in C3D format into Visual 3D (C-Motion Inc, USA, version 4) for further biomechanical analysis. Visual 3D was used to create an eight segment rigid-link, dynamic biomechanical model of the lower limbs, pelvis and trunk, each body segment being represented as a known geometric objects scaled to each participant (Figure 3.3) (Hanavan, 1964). For each body segment, the origin was positioned at the proximal end of each body segment, the only exception being the trunk which used the caudal end of the segment. Anthropometric data collected from each participant ensured the model was as accurate as possible. Measurements used in the construction of the model included body weight, height, pelvis depth level with the anterior superior iliac spine, chest depth below the manubrium and the diameter of each proximal thigh. BW estimates of segment masses and CoM were based on Dempster's (1955) data. The C3D files were exported from Visual 3D in ASCII format and saved in a Microsoft™ Excel file for subsequent statistical analysis.

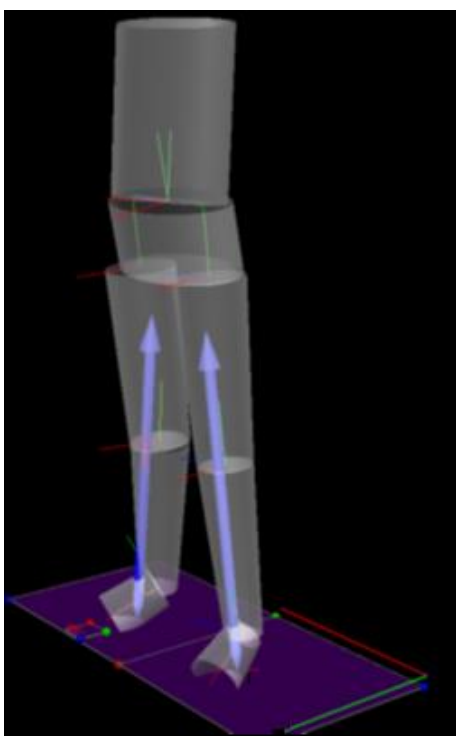


Figure 3.4 The eight segment, rigid-link biomechanical model (from Visual 3D)

3.3.2.4 Axes orientation

The force gauge, force plates and laboratory axes were orientated to match the global movement co-ordinates in Visual 3D. The global movement co-ordinate system was defined by co-ordinates in the three orthogonal axes:

- The anterior to posterior axis (X) was towards the midline of the two force plates
- The medial to lateral axis (Y) was towards the direction the participant was facing
- The vertical axis (Z) was directly upwards

The localised joint co-ordinate system for the feet, knees, hips, pelvis and trunk in Visual 3D were aligned to the recommendations of the International Society of Biomechanics Joint Co-ordinate systems (Wu et al., 2002). The local coordinate axes were located at the proximal end of each segment, with the exception of the trunk where it was located at caudad end of the segment. The three orthogonal axes were defined by:

- The anterior to posterior axis (X), directed forward from the participant
- The positive vertical axis (Y), directed upwards
- The medial to lateral axis (Z), directed to the right of the participant

3.3.2.5 Extensor moment calculation

The parameters used to estimate the external (extensor) moment are shown in Figure 3.5. The extensor moment (EM_X) was calculated after the addition of a force vector to the trunk segment. The mass of the head and the upper body was based on the data presented such that it was positioned 0.374 along the trunk segment axis. Its relative position was based on the work of Winter (2009). EM_X was estimated using the formula adapted from Dolan & Adams (1993) and can be expressed as:

$$EM_X = \mathcal{X}_1 F_z + \mathcal{X}_2 B_z + \mathcal{X}_3 H_z$$

- F_Z is the force applied to the force gauge.
- B_Z is the mass of the trunk (0.355xBW (Dempster (1955)).
- H_z is the mass of the upper limbs and head.
- $\mathcal{X}_1$ is the horizontal distance between the anterior chest belt and the centre of rotation (CoR) of the lumbar spine. The CoR of the lumbar spine was located 7.5 centimeters anteriorly from the tip of the spinous process of the fifth lumbar vertebrae (Dolan & Adams, 1993).
- $\mathcal{X}_2$ and $\mathcal{X}_3$ are the horizontal distances from the CoM of the trunk and the CoM of the head and arms to the CoR of the lumbar spine, respectively (Figure 3.5).

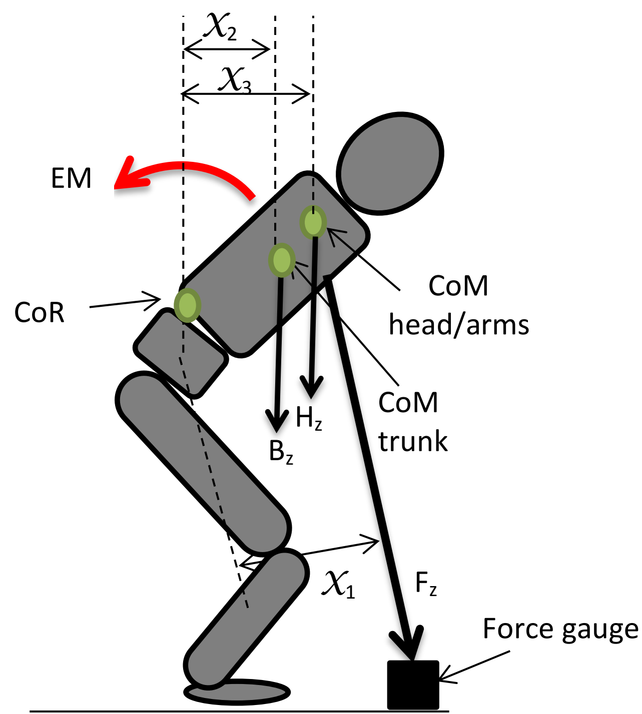


Figure 3.5 Parameters used to calculate extensor moment

Centre of mass of the upper body was calculated to be 37.4% inferior along the line from the first thoracic vertebra spinous process to the line dissecting the greater trochanters of the hip (Figure 3.6) (Dempster 1955). The CoR of the trunk segment was located 0.5 along the segment length and determined by the default setting in Visual 3D.

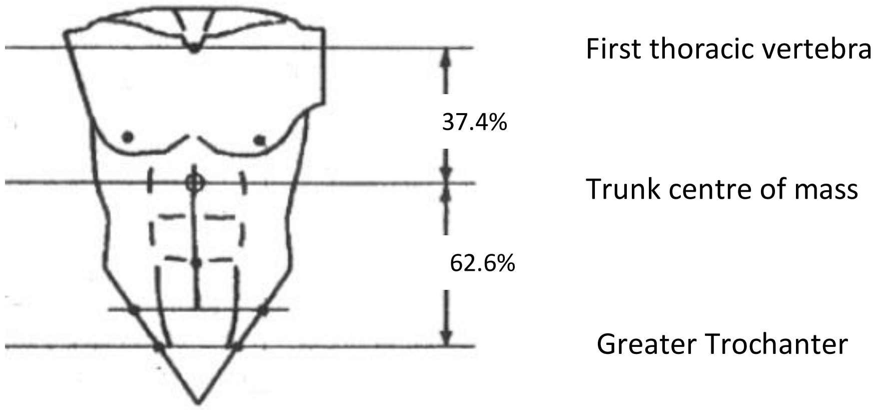


Figure 3.6 Upper body centre of mass position

The internally derived moments, using the force measurements from the force plates, were estimated using inverse dynamics in the unrestrained pelvis condition only. Ground reaction forces were used to estimate joint reactions forces and muscle moments acting about the ankles, knees, hips, and pelvis. The net muscle moment acting about the pelvis represented the internally derived BE moment.

3.3.2.6 Lumbar posture

Lumbar posture was measured using the Polhemus Fastrak system (Polhemus Navigation, Kaiser Aerospace Inc., Vermont USA). Fastrak is a real time, six degree-of-freedom tracking system that detects rotation of a sensor located within the electromagnetic field of a receiver unit. In the standing position, Fastrak 3D movement sensors were secured using doubled sided adhesive tape (3M St.Paul, MN, USA) to the skin surface superficial to the spinous process of the first lumbar vertebra (L1) and the first sacral vertebra (S1). The L1 and S1 vertebral spinous processes were referenced from locating the intervertebral space between L4 and L5, level with the top of the iliac crest and cross checked by palpating up from S2 level located level with the PSISs (Snider, Snider, Degenhardt, Johnson, & Kribs, 2011). Fixomull™ tape was applied over the base of the sensors to minimise their movement.

Changes in lumbar posture were sampled at a frequency of 30Hz with LabVIEW™ (National Instruments, Austin, Texas, USA, 2009, version 9.0). Lumbar curvature during lift initiation was the relative angle between the L1 sensor relative to the receiver and between the S1 sensor and the receiver (Dolan & Adams, 1993). A graphical representation of lumbar curvature was available in real time for the participant to view.

3.3.2.7 Lumbar fixation

A custom-built harness was used to fixate the pelvis which enabled participants to stand and adopt a lifting posture (Montgomery et al., 2011). The harness was made of woven non-stretch material and was worn around the participant's pelvis below the level of the posterior superior iliac spines. The device was adjustable with a Velcro™ waist belt and with a rubber inlay to prevent slippage. Four adjustable straps extended off the waist belt to attach to anchor points on the floor. To provide the anchor points at the desired locations, a wooden board was bolted to the laboratory floor with hooks attached at each corner (Figure3.7). Weights were placed near the hooks to prevent the board lifting. The central section of the board was removed to expose the force plates.

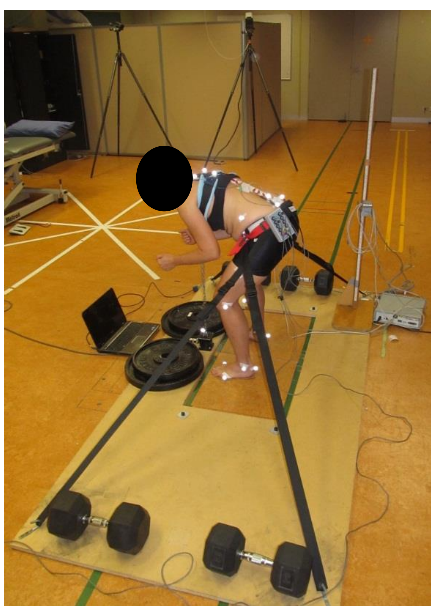


Figure 3.7 Fixation device and set up

3.3.3 Electromyography

Surface electrodes were used to collect muscle EMG signals from the UES, LES and multifidus. The following section describes how the skin was prepared, the electrode placement determined and EMG was collected and processed.

3.3.3.1 Skin preparation

Prior to electrode placement L4/5 intervertebral space and L3 and T10 spinous processes were located and marked with a pen on the skin surface by an experienced physiotherapist. The skin surface superficial to an area two to three centimeters to the right of midline was shaved and rubbed with alcohol wipes. The skin over the bony prominence of the second thoracic vertebral spinous process was also prepared and an earth electrode applied.

3.3.3.2 Electrodes and placement

Nortrode 20 silver/silver chloride 20mm bi-polar self-adhesive electrodes (Nortrode 20™, Myotronics, Inc, WA, USA) were applied to the prepared skin overlying the UES, LES and multifidus. Skin resistance was measured for each bi-polar electrode and a level of resistance less than 10kΩ was considered acceptable (De Luca, 1997).

Electrode placements were established according to the SEMTIL guidelines (Hermans & Freriks, 2012) and the anatomical landmarks were located by an experienced physiotherapist. The multifidus electrode was placed level with L4/5 intervertebral space, on a 30^{o} oblique orientation fanning more lateral, distally. the UES and LES electrodes were attached to the skin surface 2-3 cm lateral and parallel to the spinous process of T10 and L3 spinous processes, respectively.

3.3.3.3 EMG collection and processing

EMG data were collected using a Bortec AMT8-channel differential amplifier (Bortec Biomedical ltd, Canada). Each of the bi-polar electrodes were connected to a preamplifier that plugged into the portable receiver located on a belt around the participant's waist. Pre-amplified signals from all the muscles of interest were transmitted, via a cable, from the receiver unit to a Bortec Biomedical amplifier with variable gains of 1-3 continuous, 10GOhm input impedance and common mode rejection ratio of 115dB (De Luca, 1997). The amplifier gain was set at 500 and the hardware high and low pass filters were set at 10Hz and 500Hz, respectively. It is widely accepted within the literature that most of the power spectrum of the EMG signal lies within 5-500Hz (Soderberg & Knutson, 2000). EMG signals from the amplifier were converted to digital data via the analogue to digital conversion board and collected in the Qualysis software at a sampling frequency of 1200Hz. The EMG data were synchronized with the kinetic and kinematic data in the Qualysis software.

The EMG data files were converted into text files and exported in text format for analysis using LabVIEW™ (National Instruments, Texas, USA, version 9.0, 2009) software programme. A visual cursor routine was used to establish a one second epoc where the peak maximum force from the force gauge occurred during the three MVIC for each posture. Visual inspection and power spectrum analysis of the epoc were undertaken to ensure that no movement artifacts were evident. From the three trials for each posture, the file with the highest mean force produced over the one second period was selected for further data analysis in LabVIEW™. The raw EMG data were demeaned and bandpass filtered between 10 and 500Hz using a Butterworth 4^{th} order filter.

The root mean square (RMS) method was used to assess the level of activation of each muscle during rest, MIVC of back extension tasks. Briefly this method initially squares all the data points within the epoc of EMG data in order to gain positive values for all data points, and the final RMS value is estimated by calculating the square root of the mean of the squared values. In order to compare levels of muscle activation across muscles and individuals, RMS for MVIC were normalised and expressed as a percentage of MVIC obtained during the Biering-Sorensen test. The normalization method of the EMG (EMG_N) subtracts the RMS of the EMG signal during rest from that during the lift initiation task. The resulting value is divided by the difference in RMS recorded during MVIC and rest, then multiplied by 100.

$$EMG_N = \frac{\text{Lifting RMS - resting RMS}}{\text{Biering-Sorensen RMS - resting RMS}} \quad \text{X } 100$$

3.4 Experimental Procedures

3.4.1 Equipment calibration

At the start of each testing session the cameras were calibrated according to the manufacturer's instructions. The average movement residue of the markers during calibration was less than two millimeters. The wooden board was bolted in situ around the force plates and the location of the force plates and the force gauge were defined so that the force vectors and laboratory coordinates were synchronized within Qualysis software. Both force plates and the force gauge were zeroed prior to data collection.

3.4.2 Participant training

At the start of the test session all participants were familiarized with the testing procedure and given training on how to assume the different lift initiation postures (lordotic, mid and flexed positions) (Figure 3.8) and how to generate smooth, sustained forces during the lift initiation task.

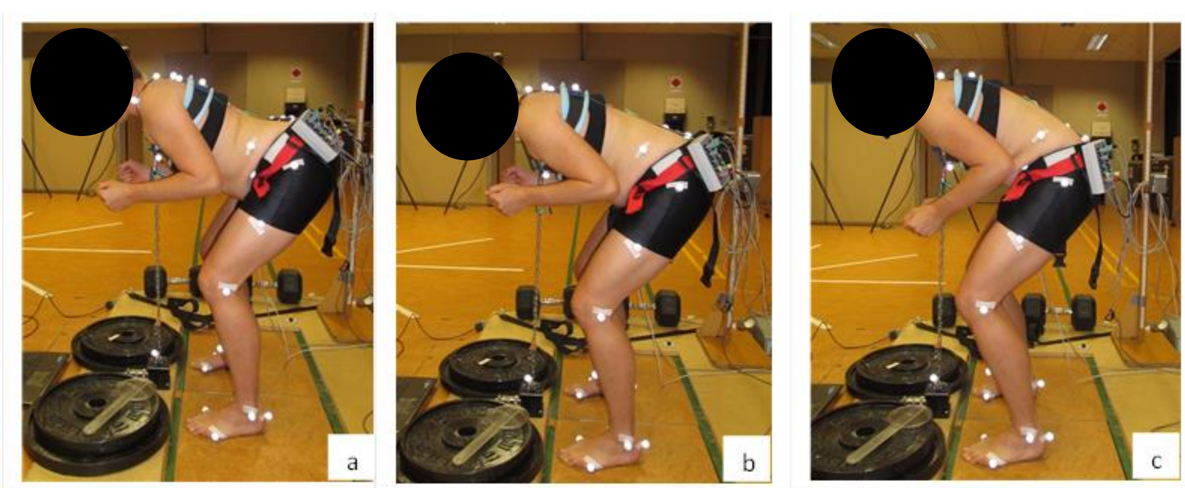


Figure 3.8 Lift initiation postures a. lordotic b. mid c. flexed.

Visual feedback from a laptop connected to the Polhemus Fastrak device was used to educate participants on how to assume each of the lumbar postures. A measurement was taken from Fastrak of the range of motion of the lumbar spine from upright standing to maximal forward flexion, whilst the participants maintained full knee extension. This was recorded manually.

Participants were asked to connect the chain to the chest harness and were guided into 45^{o} of knee flexion to simulate lifting a box located in front of their feet. Once in this position, the investigator used a combination of verbal and visual instructions along with feedback from Polhemus Fastrak device to show participants how to anteriorly and posteriorly tilt their pelvis. Participants practised until the extremes of lumbar flexion and lordosis did not vary more than two or three degrees (as displayed visually on the laptop). The range of motion from full lordotic to fully flexed lumbar posture in the lift initiation position was recorded manually. Once achieved, the participant returned to the upright standing posture to rest and repeated the process again to ensure consistency of range of motion.

When the participant was proficient in assuming the three lumbar positions (maximum lumbar lordosis, maximum lumbar flexion and the mid point of these two extremes), they were asked to use approximately 50% to 75% of their maximum effort to exert an upward tension from the chest harness on the chain connected to the force gauge, while maintaining the different lumbar postures. During these sub-maximal exertions participants were instructed to increase force over a three second period and then to maintain a steady and constant force for a further three seconds. Participants were given visual feedback of the force trace via a computer screen, allowing correction or affirmation of the technique used. This was repeated a minimum of five times and until the technique of maintaining lumbar curvature while sustaining a constant smooth force was achieved. While the purpose of this session was to train the participant, it also served as a warm-up exercise for the BE muscles.

3.4.3 Participant preparation

At the start of the test session, participants were asked to change into close fitting shorts and vest. The age of the participant was recorded and the anthropometric data measured.

Anthropometric data included:

- Chest depth at mid sternum
- Right and left thigh width at the greater trochanter
- Height
- Body weight

The participant was required to lie prone on a plinth and surface electrodes were applied to the skin surface superficial to the UES, LES and multifidus muscles on the right side of the trunk. The participant was then asked to return to standing and the chest and pelvis harness were fitted. The 36 retro-reflective markers were also applied to the skin surface on the anatomical landmarks of the trunk and lower limbs.

Participants were asked to stand towards the front of the force plates, one foot on each plate, equidistance from midline, shoulder width apart, to simulate a lifting a box located in front of their feet. When foot position was established a line was drawn on the floor around the participant's feet to permit consistent foot position for each trial. The Fastrak sensors were then attached to the L1 and S1 spinous processes.

Next, the chain length between the chest harness and force gauge was determined. The participant was required to flex their knees to 45^{o} (as determined by a manual goniometer). From this position the participant was instructed to flex forward and simulate lifting the box from its handles, which were located 35cm above the floor (Figure 3.9). With the knees remaining in 45^{o} of flexion, the lumbar spine posture was set to the mid-position (as determined by the Fastrak), the participant applied light tension to the chain, and the length of the chain connecting the chest harness to the force gauge was determined.

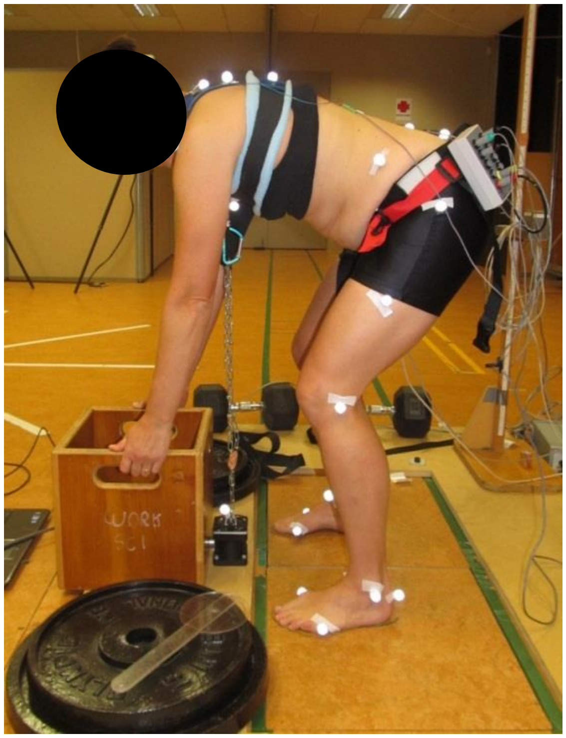


Figure 3.9 Method to establish chain length

3.4.4 Testing session

The first data file to be collected was a 10 second resting EMG file with the participant lying prone on a plinth. A kinematic data file of the participant standing in the anatomical position with each foot on a force plates was captured to act as the reference 'static' file to apply to subsequent dynamic files in Qualysis motion analysis.

The randomised sequence of posture tasks predetermined the initial fixation condition and lumbar posture the participant was to adopt. The participant was positioned with their feet on the designated foot marks, established during the practice session, and instructed to connect the chain and carabineer from the force gauge to the chest harness. Participants were asked to assume the required lumbar posture with the aid of the Fastrak lumbar curvature angle displayed on the laptop screen. The participant was instructed to exert light tension on the chain while keeping their knees flexed to 45^o. Once the participant confirmed they were in the correct lumbar posture an active motion file within Qualysis was captured during the maximal isometric trunk extension. During the maximal exertions participants were instructed to increase force over a three second period and maintain a steady and constant force for a further three seconds. Verbal encouragement was given throughout the maximal isometric contraction. At the end of the each trial, the participant unclipped the chain from the chest harness, disconnected the pelvis harness from the floor anchors and returned to an erect standing position. A minimum rest period of two minutes was given between trials to allow sufficient time for recovery and to minimise the effect of fatigue. The participant performed three maximal isometric trunk extensions for each of the three lumbar postures (lordotic, mid and flexion) in both the fixed and free pelvic conditions.

Following maximal isometric trunk extensions in all the lifting postures, participants performed three maximum isometric back extensions in the Biering-Sorensen position (Figure 3.10). The participant was positioned prone with their legs and pelvis securely strapped to the plinth and their ASIS's in line with the top of the plinth, leaving the trunk unsupported. The chain from the chest harness was attached to the floor mounted force gauge. The force and EMG data acquired during the MVIC performed in the Biering-Sorensen position were used to normalize the data collection during the active simulated lift trials. This position was chosen for normalisation because it has been shown to be a reliable and commonly used method to normalise low back EMG data (Coorevits, Danneels, Cambier, Ramon, & Vanderstraeten, 2008). Furthermore, data from the Fastrak device indicated that lumbar curvature in the Biering-Sorensen position was similar to that in upright standing.

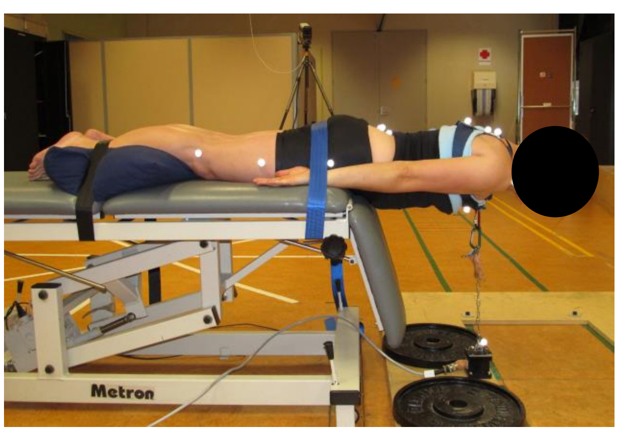


Figure 3.10 Biering-Sorensen position

3.5 Statistical Analysis

All data were analysed for statistical significance using IBM SPSS version 19.0 (SPSS Inc. Chicago) software package. Data for each trial were initially inspected using box plots to statistically identify any potential outliers and also checked for normality. A two factor analysis of variance (ANOVA) with repeated measures was used to test the main effect of the two independent variables, posture (3) and pelvis fixation (2) on the dependent variables of BE moment and muscle. Subsequent, 3x2x3 repeated ANOVA was conducted for each of the three independent muscle divisions. The statistical significance was set to an alpha level $p<0.05$. Significant main effects and interactions were subjected to post hoc Bonferroni tests.

Chapter 4: Results

4.1 Introduction

This chapter presents the findings from this study. It is divided into six sections starting with the participants' information, followed by how the data were managed. The next section describes the findings related to the effect of posture on torque values and muscle activation. This is followed by reporting the influence of posture on the neuromuscular efficiency of the paravertebral muscles. The chapter is concluded by discussing the effect of pelvis fixation on torque and muscle activation.

4.2 Participant Information

4.2.1 Demographic data

Table 4.1 displays the demographic data of the participants included in this study. While the mean age of males and females were closely matched, the mean weight and height were significantly greater for males compared to females ($P>0.01$).

Table 4.1 Participant demographics

Gender	Mean age(yrs)	Mean weight (kg)	Height (cm)
Male (n=13)	24.2	78.50	176.6
Female (n=13)	22.6	64.35	163.8
Total	23.4	71.43	170.2

4.2.2 Lumbar range of motion

The Fastrak device measured the range of motion (ROM) of the lumbar spine from relaxed upright standing to maximum forward flexion with knees extended (standing ROM). Measurements of lumbar ROM in the lifting posture were also recorded between the maximal lumbar lordotic position (end ROM of extension) and the maximal lumbar flexion position, which is referred to as the lifting ROM (Figure 4.1).

Lumbar angle measurements were similar during maximal flexion in standing and in the lift initiation position. However, when participants attempted to assume the maximal lordotic posture in the lift initiation position the lumbar spine was flexed to 22° when compared to the upright standing posture.

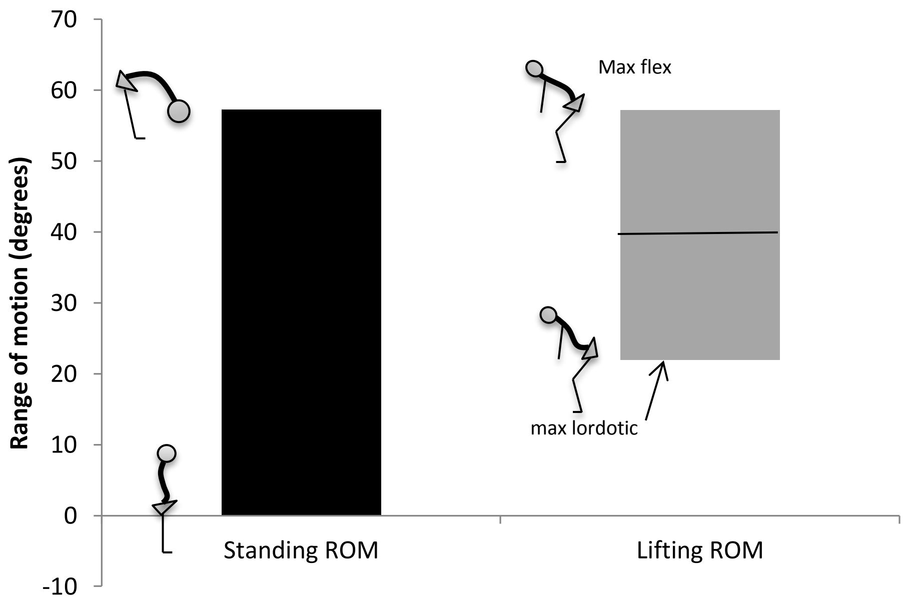


Figure 4.1 Mean range of motion of the lumbar spine in standing and during lift initiation position

4.3 Data Management

4.3.1 Back extensor torque calculation and validation

Descriptive statistics indicated that the mean peak torque of the externally derived moment (force gauge) was slightly lower for all conditions than the internally derived moment (force plates). However, independent t-tests showed no significant difference in peak torque between the two methods of measurements (Table 4.2). Therefore, all subsequent analysis of data used in this results section are expressed as torque values calculated using the externally derived method (force gauge) as this method can be used for calculation of torque in both free and fixated lifting postures.

Table 4.2 T-test results comparing the two methods of torque calculation

Test condition	Peak torque (Nm) (External derived) Mean (SD)	Peak torque (Nm) (Internal derived) Mean (SD)	Sig level
Free Lord	159.67 (55.5)	178.23 (48.4)	0.286
Free Mid	170.77 (48.6)	185.03 (46.9)	0.205
Free Flex	191.91 (51.7)	204.16 (46.4)	0.372

4.3.2 Torque and gender

ANOVA revealed main effects for gender and posture (P = 0.001), but no significant gender by posture interactions. Subsequently post hoc analysis revealed that in all test conditions males produced the highest mean torque irrespective of posture. The means and standard deviations of the mean peak torque values collected for each gender in each posture and pelvis fixation condition are displayed in Table 4.3. Flexed postures produced the largest torque, followed by mid postures, with the least torque being produced in lordotic postures for both genders.

Table 4.3 Mean (SD) peak torque for gender in each test condition

Condition		Torque (Nm)	
	Males Mean (SD)	Females Mean (SD)	All participants Mean (SD)
Fix Lord	210.23 (25.7)	136.25 (22.0)	173.24 (44.4)
Fix Mid	214.32 (30.8)	152.18 (25.6)	183.25 (42.11)
Fix Flex	220.18 (31.8)	176.37 (43.2)	198.28 (43.4)
Free Lord	212.84 (40.6)	143.61 (25.4)	178.23 (48.4)
Free Mid	220.09 (33.8)	149.97 (27.9)	185.03 (46.4)
Free Flex	234.33 (29.8)	174.00 (40.3)	204.16 (46.4)

To take into account body mass differences between genders, peak torque was expressed in terms of torque (Nm) per unit of body weight (kg) (Figure 4.2). Data analysis revealed there were no significant differences (sig= 0.348) between males and females when torque was normalised to body weight.

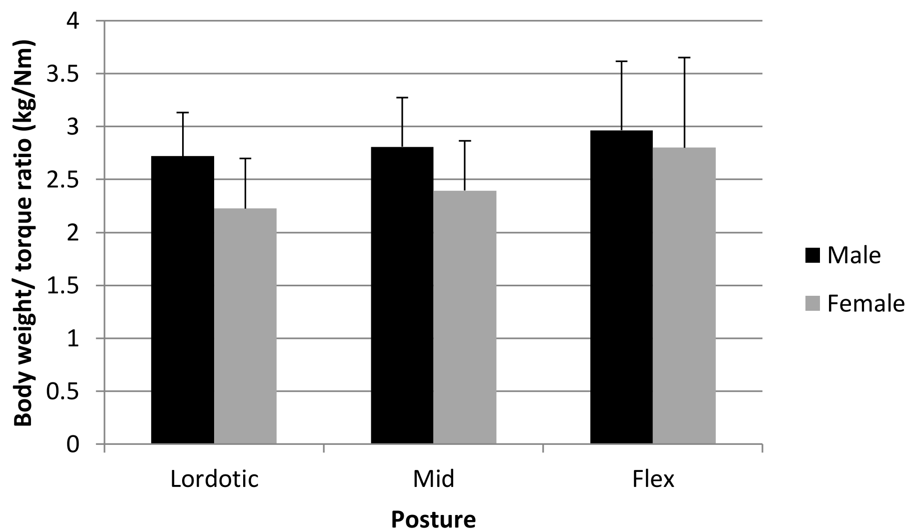


Figure 4.2 Torque expressed per unit of body weight for each posture and gender

Torque was also normalised by expressing torque as a percentage of maximum torque produced in the Biering-Sorensen position (Figure 4.3). Again, data analysis revealed there to be no gender differences when the data were normalised in this way. Descriptive analysis showed when torque was normalised using the Biering-Sorensen method compared to expressing torque as a percentage of body weight, males and female torque data were similar. Therefore, male and female normalised torque data, expressed as a percentage of MVIC during the Biering-Sorensen position are, hereafter, pooled for all subsequent analysis of main effects and interactions.

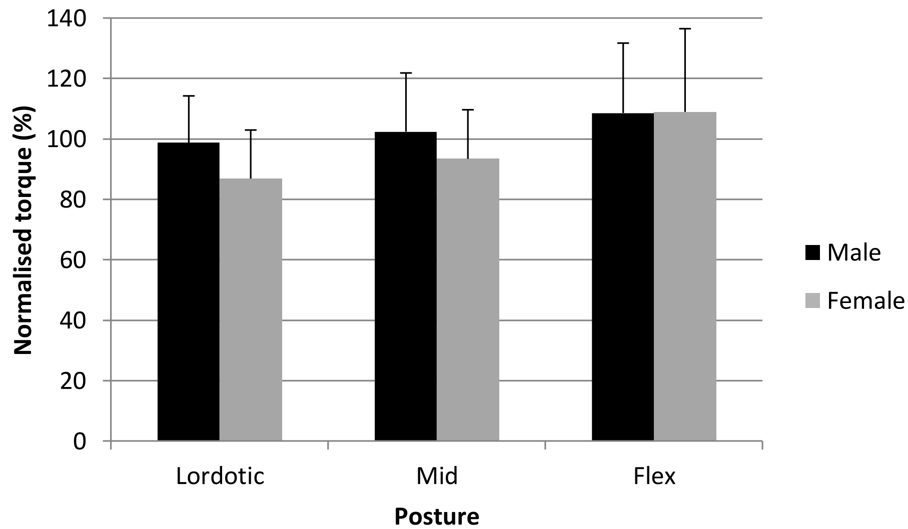


Figure 4.3 Torque expressed as a percentage of MVIC during the Biering-Sorensen for each lumbar posture and gender

4.3.3 EMG and gender

The RMS value of the EMG data collected in each of the lifting postures was normalised to the EMG recorded during MVIC in the Biering-Sorensen position. Although males tended to activate the UES and multifidus at a greater percentage of MVIC than females, the difference in the level of muscle activation was not significant between genders (Figure 4.4). ANOVA also showed there was no significant interaction effect between gender and posture, or between gender and level of fixation. Therefore, all normalised EMG data presented, hereafter, are analysed collectively for male and females.

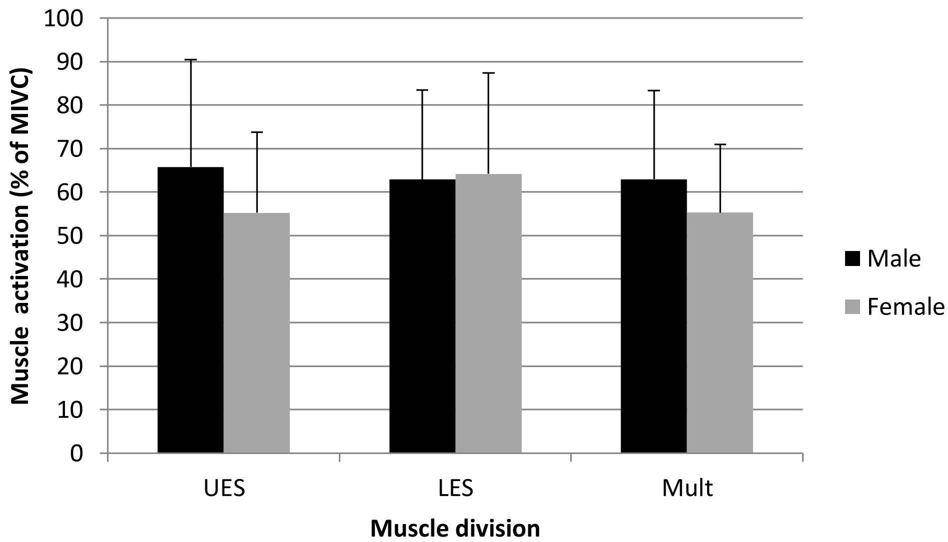
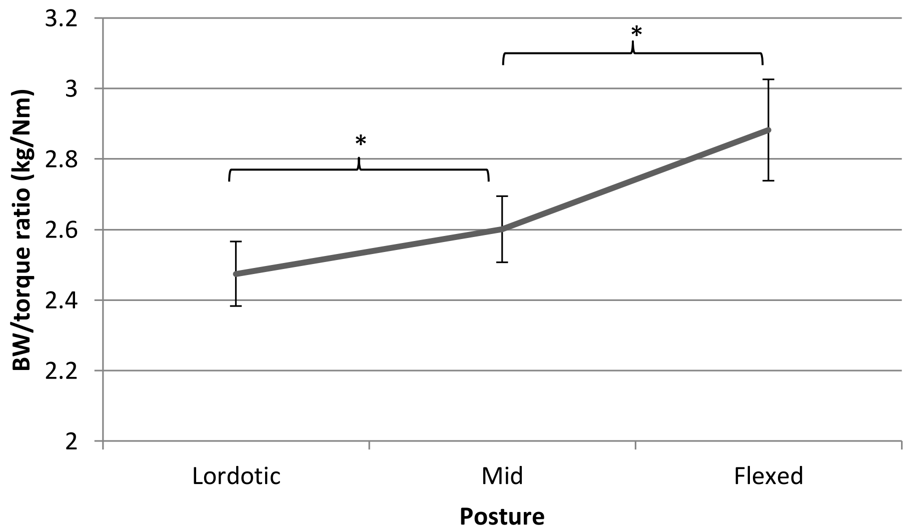


Figure 4.4 Normalised EMG by muscle division for each gender

4.4 Lumbar Posture

4.4.1 Posture and torque/body weight ratio

Figure 4.5 shows that torque/body weight (BW) ratio increased from lordotic to mid lifting posture and also from mid to flexed lifting postures. When torque was normalised as Nm/Kg of BW, posture had a significant main effect on torque (p=0.001). Post hoc analysis of comparing the three postures revealed that significantly higher torque production occurred in the fully flexed position when compared to mid-range posture (p=0.013), which in turn had significantly greater torque levels than when participants adopted a lordotic lifting posture (p=0.014) (Table 4.4).



*significant difference (P=0.001)

Figure 4.5 Torque expressed as a ratio of BW for each posture

Table 4.4 Post hoc analysis of torque/BW between different postures

ALL			**95% CI**	
Posture1	**Posture2**	**Sig. level**	**Lower bound**	**Upper bound**
Flex	Mid	0.013*	0.051	0.511
Flex	Lordotic	0.001*	0.182	0.635
Mid	Lordotic	0.014*	0.022	0.232

4.4.2 Posture and normalised torque

Figure 4.6 shows torque expressed as a percentage of the torque in Biering-Sorensen position. A significant main effect of posture on normalised torque was found (P>0.0001). Post hoc analysis found a significant difference between the flexed and mid postures (p=0.004), mid and lordotic postures (p=0.014) and flexed and lordotic postures (p=0.0001). Participants in the lordotic lifting posture produced 93% of the torque generated during the Biering-Sorensen test, with a significant increase to 97% in the mid posture, with a further increase of 11% from mid to fully flexed positions.

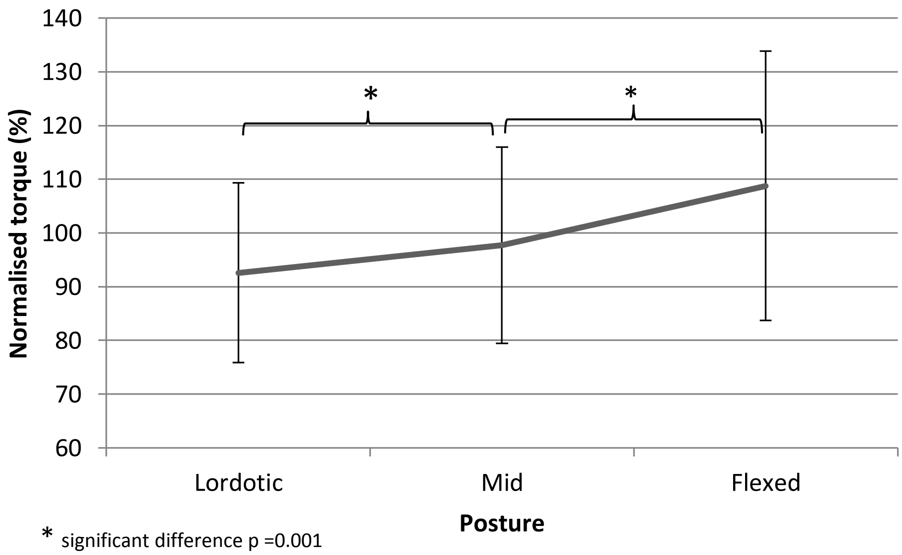


Figure 4.6 Normalised torque and posture

4.4.3 Posture and normalised EMG

Figure 4.7 shows EMG expressed as a percentage of the EMG activity in the Biering-Sorensen position. ANOVA showed a significant main effect for posture (p=0.001) and post hoc analysis (Table 4.5) revealed that there was a significant difference between lordotic and mid postures (p=0.001). A 13% drop in activation occurred between the lordotic and mid postures. A significant difference was also found between mid and flexed postures (p=0.001), with a 41% decline in activation from the mid to flexed lifting posture. The difference in activation between lordotic and flexed postures was also significant (p=0.001), with lordotic postures producing the greatest amount of normalised activation (Figure 4.7).

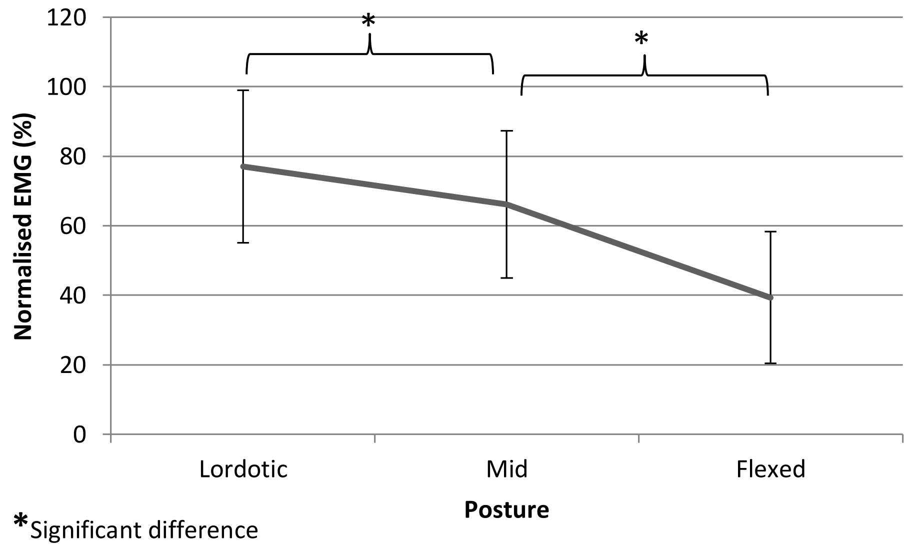


Figure 4.7 Normalised EMG for each lifting posture

Table 4.5 Post hoc analysis significance levels and CI of normalised EMG

Posture1	Posture2	Sig. level	95% CI Lower bound	Upper bound
Flex	Mid	0.0001*	-36.28	-17.42
Flex	Lordotic	0.0001*	-45.53	-29.83
Mid	Lordotic	0.001*	-17.68	-3.98

4.4.4 Posture and normalised EMG by muscle division

Figure 4.8 displays the EMG for the UES, LES and multifidus in each lifting posture, expressed as a percentage of the EMG produced in the Biering-Sorensen position. All divisions of muscles exhibited a similar amount of activation for each posture, thus there was no statistical significance between activation levels of each muscle division. All muscle divisions showed a significant difference in the amount of activation between flexed and mid postures, mid and lordotic postures and flexed and lordotic postures. All muscle divisions showed the greatest decline in activation between the mid to flexed postures.

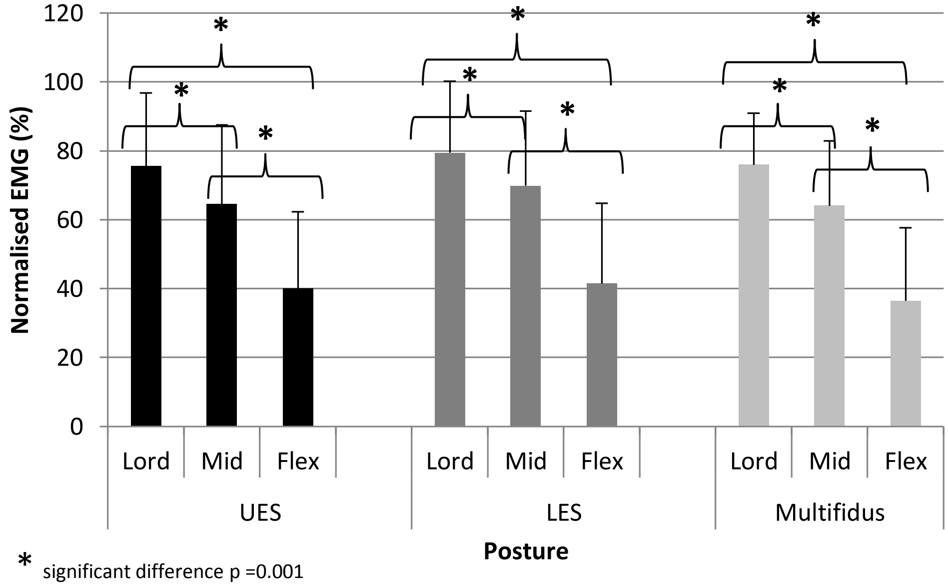


Figure 4.8 Mean and SD for normalised EMG for posture and muscle division

4.5 Neuromuscular Efficiency

Neuromuscular efficiency ratio was calculated by dividing normalised torque by average normalised EMG of all three divisions (T_N/EMG_N). Normalised values were calculated as a percentage of that produced in the of Biering-Sorensen position. There was a significant main effect of posture on neuromuscular efficiency ($p>0.0001$). Post hoc analysis revealed a significant difference between lordotic and mid postures ($p=0.014$), mid and flexed postures ($p=0.013$), and subsequently, a difference between flexed and lordotic postures ($p=0.0001$) (Table 4.6). NME was shown to increase by 23% between the lordotic and mid lifting postures and by 87% from the mid to fully flexed lifting postures (Figure 4.9).

Table 4.6 Post hoc analysis of posture on neuromuscular efficiency

Postures compared	Sig. level	95% CI Lower bound	Upper bound
Flex - Mid	0.013	0.051	0.511
Flex - Lordotic	0.0001	0.182	0.635
Mid - Lordotic	0.014	0.022	0.232

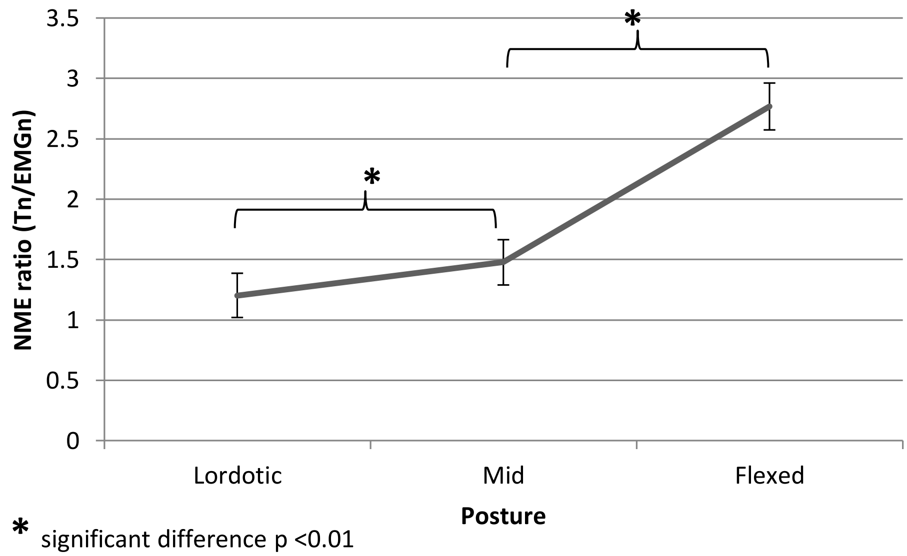


Figure 4.9 The mean neuromuscular efficiency in each posture

Normalised EMG found each muscle division responded in the same way to each other with alterations in posture. Consequently, the NME of each muscle division was also comparable to the overall NME resulting in no significant effect of muscle division on NME (Figure 4.10).

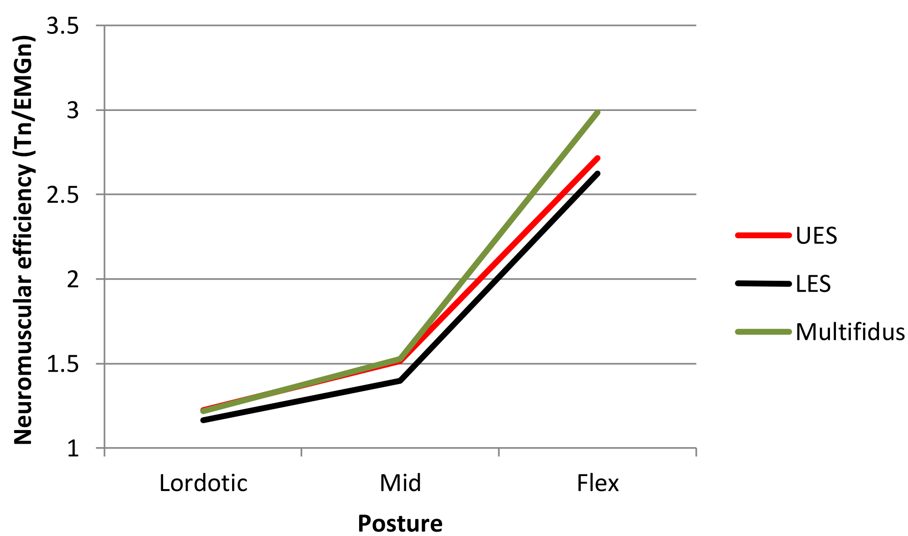


Figure 4.10 NME of each division of muscle in each posture

4.6 Pelvis Fixation

4.6.1 Fixation and normalised torque in each posture

Figure 4.11 shows the effect of pelvis fixation on torque for each posture. Whilst the free pelvis condition generated more torque in each posture, the difference was small therefore pelvis fixation was shown to have no significant effect.

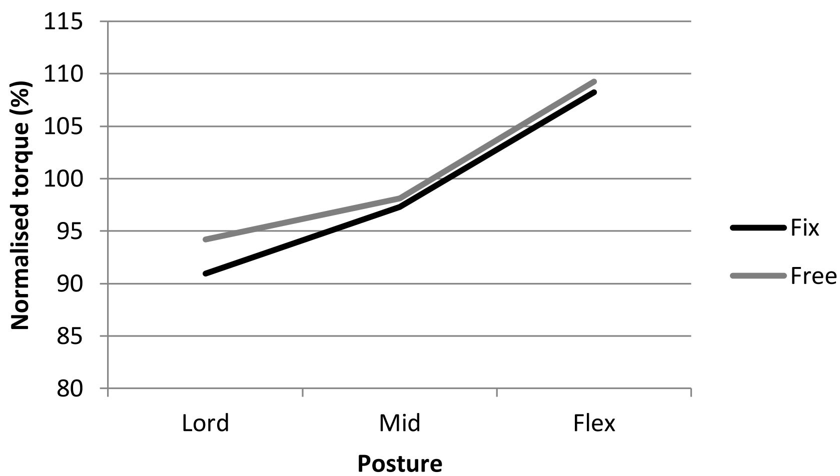


Figure 4.11 The effect of pelvis fixation condition in each posture

4.6.2 Fixation and normalised EMG for each muscle division

Figure 4.12 displays the normalised EMG for each muscle division under the fixed and free pelvis conditions. Descriptive statistics showed that all muscles in the free conditions produced slightly less activation than when the pelvis was free to move. ANOVA revealed that there were no significant differences ($p=0.38$) in activation levels between any of the muscle divisions when the pelvis was fixated or free to move.

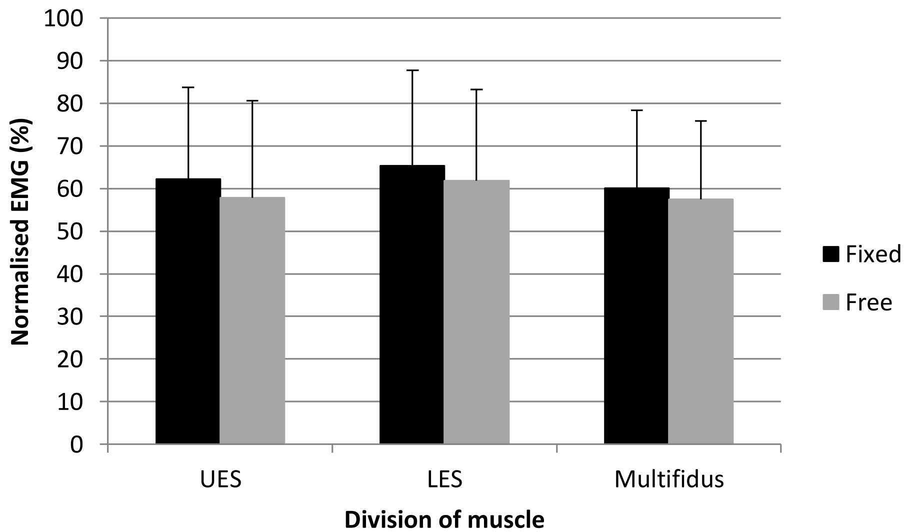


Figure 4.12 Normalised EMG and division of muscle comparing fixation

Chapter 5: Discussion

5.1 Introduction

The purpose of this study was to investigate the effect of lumbar posture and pelvis fixation on BE torque and the level of muscle activation of three separate divisions of the paravertebral muscles during a simulated isometric lift initiation task. This chapter aims to discuss the research findings and to compare it to the existing literature on this topic. It will first begin by discussing the findings related to lumbar posture and its influence on back extension torque and muscle activation, followed by NME. Pelvis fixation and its effect on torque and muscle activation will be discussed. This section will conclude by outlining the clinical implications of the study and its limitations.

5.2 Lumbar Posture

5.2.1 Posture and torque

The results of this study showed that posture had a significant effect on torque production, with the flexed lumbar posture significantly producing more torque than mid or lordotic postures. These findings concur with existing research that has found BE torque to be greater in the flexed posture compared to other lumbar postures (Dolan et al., 1994; Parnianpour et al., 1991; Raschke & Chaffin, 1996; Roy et al., 2003; Tan et al., 1993; Toussaint et al., 1995). However, in contrast to previous research the current study found that the greatest change in torque production occurred when moving from the mid to flexed posture. Previous studies have found rapid increases in torque from a neutral upright standing position to 20-30^{o} of lumbar flexion (Parnianpour et al., 1991; Roy et al., 2003; Tan et al., 1993), beyond which the rate of change decreases.

A possible reason for the contrasting findings between this study and those in the literature may stem from differences in the methodological approaches used. Previous research protocols have often been undertaken using isokinetic devices to test maximal isometric BE torque with participants standing with their legs and pelvis fixated. The upright standing position equated to the neutral posture (0^{o} of flexion). However, the current study used a more functional lifting posture with only the pelvis fixated and the lower limbs free to move. When participants were asked to maximally extend their lumbar spine in the functional lifting position their lumbar spine remained at approximately, 22^{o} of flexion (approximately one third of the way into flexion). While no known research has specifically investigated lumbar range of motion in simulated lifting postures, it is known that when performing activities that require hip flexion, there is associated posterior pelvic rotation, which subsequently causes lumbar flexion (Yasukouchi & Isayama, 1995). This flattening of the lumbar spine has also been noted during dynamic lifting tasks (Arjmand, Plamondon, Shirazi-Adl, Lariviare, & Parnianpour, 2011; Potvin et al., 1991). The amount of lumbar flexion during simulated lifting using the maximal lordotic posture in the current study was comparable to a mid-flexion position used in previous studies. This also highlights that care should be taken when interpreting findings from studies that have investigated the relationship between torque and lumbar posture where the lumbar spine is extended or in a position of less than 20^{o} of flexion. Such postures are rarely seen during lifting tasks, particularly when lifting objects from below knee height.

Another possible reason why the current study found a significant increase in torque towards the end range of lumbar flexion may also relate to how posture influences the length-tension relationship of ES muscle fascicles. Delp and colleagues (2001) undertook an *in vitro* study to calculate optimal sarcomere length of Iliocostalis lumborum and longissimus thoracis. A laser diffraction technique was used to establish muscle sarcomere lengths in ES muscle fascicles, which were then normalised to the previously established optimal sarcomere length of human muscle (Delp et al., 2001). It was predicted that optimal fascicle length for the ES muscle group ranged from 16-20% beyond that of resting length in supine. It was concluded that the ES muscles have an improved ability to generate muscle force in more elongated (flexed) positions than in a neutral (supine) posture. Optimal length-tension relationships in more flexed positions have also been found in the multifidus muscle (Ward et al., 2009). Ward and colleagues (2009) determined sarcomere length and the length-tension relationships of multifidus fascicles that were dissected from patients undergoing surgery in flexed and lordotic postures. Single fibre samples were attached to an ultrasensitive force transducer and a micromanipulator that stretched the muscle fibre in 250 μm increments. They found, unlike other muscles within the body, the length-tension curve for multifidus continued to increase as the spine became more flexed, reaching greatest tension at approximately 40^o of lumbar flexion. This suggests that multifidus has the potential to generate considerably more tension when using postures in higher degrees of lumbar flexion.

The length-tension relationship of ES has also been estimated through biomechanical modelling of participants undertaking back extension tasks and these studies agree with the anatomical studies on how lumbar flexion influences torque production (Mannion & Dolan, 1996; Raschke & Chaffin, 1996). For example, Raschke & Chaffin (1996) recorded the EMG of ES at L3/4 whilst measuring dynamic trunk extensor torque at 0^o, 22.5^o and 45^o of trunk flexion. A standing frame was used to fixate the legs and pelvis of participants while performing a lifting task which involved pulling up on a handle attached to a floor mounted force gauge. A detailed anatomical biomechanical model of the lumbar spine was used to estimate the length-tension of the ES. It was concluded that the optimal length-tension of ES occurred at 45^o of lumbar flexion. Unfortunately, no measurements were taken beyond 45^o of lumbar flexion, so it is unclear what changes occur in the length-tension relationships for ES towards the end ranges of lumbar flexion.

To date, there appears to be no studies published on maximal isometric extensor torque for the extremes of lumbar flexion. The current study indicated that full lumbar flexion was the most optimal position for extensor torque production. In full lumbar flexion, the role of the passive tissues and their contributions to resisting bending moments of the lumbar spine also needs further consideration. Biomechanical models would suggest that there is significant loading on passive tissues of the lumbar spine at the end range of spinal flexion, generated primarily from the iliolumbar ligaments and intervertebral discs. Dolan et al. (1994) conducted a series of *in vivo* and *in vitro* experiments to determine the contribution of the passive tissues of the lumbar spine when resisting bending moments throughout lumbar flexion. They found that beyond approximately 80% of full lumbar flexion there was an exponential rise of passive tissue resistance to the bending moment. Dolan et al. concluded that the recruitment of the passive structures of the lumbar spine may provide the potential energy to contribute to the extensor moment at the end range of lumbar flexion.

Even though the current study found a significant effect of lumbar posture on BE torque, the overall increase in torque in the lift initiation position when moving from a fully lordotic to a fully flexed position was only 18%. Other studies have found 68% (Parnianpour et al., 1991), 77% (Roy et al., 2003) and 83% increase (Tan et al., 1993). These studies normalised torque to that measured in upright standing, which produced considerably less torque than that measured in the Biering-Sorensen position used in this study, even though the lumbar curvature was the same as that of upright standing. For example, Roy et al., (2003) found a mean extension torque in standing of 101 Nm for a group of young males and females (age range of 21-39 years), while the mean extension torque in the present study was 182Nm. The lower levels of torque measured in an upright standing posture may be associated with the postural stability demands required during upright standing. For example, McGill (1991) measured maximal isometric extension torque in standing and in the Biering-Sorensen position and found standing postures to be an inferior method of testing optimal back performance. It was proposed, in more complex tasks, such as in standing, the trunk muscles are responsible for maintaining equilibrium in all three axes of movement of the spine and not solely in producing torque in the desired direction. Thus, muscle exertion may be assigned to stabilise the spine resulting in a net loss in torque in the desired direction. This demonstrates the complexity of the spine and there is need for caution when applying findings from non-functional measurements of BE torque to functional lifting tasks.

5.2.1.1 Gender and torque

The effect of gender on torque was found to be in agreement with previous studies, with males being capable of generating higher peak torques than females (Dolan et al., 1994; Graves et al., 1990b; Mannion & Dolan, 1996; Roy et al., 2003). However, as others have reported (Keller & Roy, 2002), when torque was normalised to BW, the current study found that peak torque did not significantly differ between genders. Consideration of body dimension in torque production is also supported by anatomical studies. For example, Mannion et al. (1997) obtained muscle biopsies from the UES and LES of healthy young adults and found a positive correlation between body size and muscle fibre size in both the UES and LES. This suggests that larger individuals have higher ES muscle mass and therefore, greater potential to generate more extensor torque than participants of a smaller size.

The gender findings of the current study differed to that of Graves et al. (1990b) who found when BE torque was normalised against BW, males produced significantly more BE torque per kilogram of BW than females. Differences in findings between the present study and Graves et al. may relate to strength characteristics of male subjects in each study. The mean maximum torque for males reported by Graves et al. was twice that produced by the male participants in the current study, whereas females in each of these studies produced similar extensor torque. A possible explanation for Graves et al. findings may be related to the physical activity levels. The level of physical activity and training of the BE muscles has been shown to be positively correlated to muscle performance and ability to produce torque (Graves et al., 1994). Unfortunately, neither the current study nor that of Graves et al. (1990b) measured physical activity levels amongst their participants.

The current study found no differences between gender when BE torque was normalised to that produced during maximal isometric trunk extension in the Biering-Sorensen position. This test position was selected because it is commonly used for strength testing of the trunk musculature (Biering-Sorensen, 1984; Coorevits et al., 2008) and in present study the lumbar curvature was similar to that in upright standing. The Biering-Sorensen position has also been shown to demonstrate a high level of consistency for measures of BE torque (McGill, 1991). The present study's findings are similar to Roy et al. (2003), who normalised extension torque against maximum exertion in upright standing and found this method to eliminate gender differences.

5.2.3 Posture and EMG

Lumbar posture was found to have a significant effect on paravertebral EMG, with greatest levels of activation in the lordotic posture and the lowest levels of activation occurring when adopting the mid and flexed postures. Increases in paravertebral muscle activation in the lordotic posture and diminishing activation towards the mid posture (which equates to 22^o to 40^o of lumbar flexion respectively) are similar to those reported by Marras et al. (1984) who found a relatively linear decline in the LES muscle activation during maximal isometric lumbar extension from upright standing (0^o) to 45^o of flexion. However, muscle activity findings in the current study were contrary to the majority of literature, which have shown a linear increase in muscle activity during MVIC of the BE muscles in lumbar postures ranging from 0^o (upright standing) to 50^o lumbar flexion (Dolan et al., 1994; Roy et al., 2003; Tan et al., 1993). These contrasting findings, may in part, be explained by the positioning and fixation methods used in other studies. Roy et al. (2003) and Tan et al. (1993) used the B200 isostation apparatus where participants were required to stand with their pelvis and lower limbs fixated in a rigid frame with knees fully extended. However, in the current study participants were fixated using a belt system attached to the pelvis with no restraints placed on the lower limbs and allowing the knees to flex to 45^o.

An alternative explanation for a reduction in paravertebral muscle activation levels from early in the range of flexion to $40^o$of lumbar flexion may be related to the level of trunk support. The present study gave no external support to the trunk in attempt to simulate a functional lifting task. In contrast, preceding research secured participants upper torso to an isokinetic device for torque measurements. The additional support and fixation of the torso may have resulted in altering the activation of the paravertebral muscle as the trunk becomes more flexed. To some extent, this could explain why the functional lifting task used in the present study resulted in a different activation pattern to the findings reported in other studies that attempted to isolate the function of the lumbar spine from the rest of the biomechanical chain.

Another factor that should be taken into consideration when interpreting the effects of lumbar posture on BE torque and paravertebral muscle activation is lumbar stability. McGill (1991) investigated the muscle activation of the BE and abdominal muscles during maximal isometric exertion with participants pelvis and trunk fixated in a Cybex dynamometer whilst standing. It was found that in some postures there were low levels of torque but high levels of ES and abdominal muscle co-contraction. It was concluded that in some postures stabilising the lumbar spine during maximal exertions was more important than the generation of torque. It is plausible that the recruitment pattern of the paravertebral muscles during back extensions from flexed postures is aimed at stabilising the spine when using low levels of lumbar flexion. Biomechanical models have predicted that during functional tasks high stability demands occur in neutral or the mid-range of spinal postures (McGill, 1991; Panjabi, Abumi, Duranceau, & Oxland, 1989).

A neurophysiological rationale for a reduction in muscle activation may be related to the change in muscle length and its effect on the EMG signal. As muscles elongate there is a reduction in the cross section area of the muscle fibres (Brown & Gerling, 2012). EMG amplitude is proportional to the number of detectable firing motor units below the electrode and therefore when a muscle elongates there is less volume below the electrode, thus a reduction in activation (De Luca, 1997). This is unlikely to be the main explanation for the decline in muscle activity with flexion in the current study as other similar studies have shown EMG to increase in more flexed lumbar postures where the paravertebral muscles have been elongated (Roy et al., 2003; Tan et al., 1993).

Further explanation for the increase in torque and a decrease in paravertebral muscle activation between the lordotic and mid-range of lumbar flexion could be the generation of torque by BE muscles other than those recorded in this study (Marras et al., 1984; Phillips, Mercer, & Bogduk, 2008). Other muscles that have the potential to exert an extension moment on the lumbar spine include quadratus lumborum, the hip extensors and latissimus dorsi (LD). Phillips et al. (2008) investigated fascicle attachment and orientation of quadratus lumborum and predicted that its capacity to generate BE torque was limited to less than 10% of that of the paravertebral muscles. Conversely, the hip extensors have a large torque generating capacity to assist in posteriorly rotating the pelvis when raising the trunk and load being lifted, but they do not act directly across the lumbar spine (Bogduk, 2012). Further evidence to support the role of LD and its contribution to BE torque when lifting comes from anatomical studies. These have revealed that up to two thirds of the force generating capacity of LD resides within the lumbar region (Bogduk, 2012; Potvin et al., 1991). LD is considered to have only a modest force generating capacity due to low number of sarcomeres arranged in parallel, but has high numbers of sarcomeres arranged in series, allowing it to accommodate large changes in length (Gerling & Brown, 2013). This sarcomere arrangement resulted in near maximal force production across a wide range of lumbar motion including the end range of lumbar flexion (Gerling & Brown, 2013; Raschke & Chaffin, 1996).

Several studies investigating the effect of posture on BE torque have investigated the activation of LD (Marras et al., 1984; Raschke & Chaffin, 1996; Tan et al., 1993). Marras et al. (1984) found that as the paravertebral muscle activation declined with flexion, there was a concurrent increase in LD muscle activation. It was proposed that the decline in activation of ES with lumbar flexion was compensated for by an increase in LD recruitment. Furthermore, LD has been shown to be recruited more commonly when high levels of trunk exertion are required (Tan et al., 1993) and particularly when performing isometric contractions (Marras et al., 1984).

This study found a significant reduction in paravertebral muscle activation with lumbar flexion. However, there was only a 14% reduction between the lordotic and mid postures, compared to a 41% reduction in activation from mid to the full flexion. The large reduction in ES muscle activation and increase in extensor torque between the mid and the fully flexed posture suggest that towards end range of lumbar flexion torque production is not generated solely by active tissue (muscle), but also through tension within the passive spinal structures of the lumbar spine. Studies using biomechanical modelling, which have assumed a linear relationship between torque and activation, have predicted that a major contribution to BE torque during lumbar flexion comes from the passive tissues of the spine (Dolan et al., 1994; Kippers & Parker, 1984). Findings from these studies indicate that in the final 20% of lumbar flexion there is an exponential increase in loading of the passive tissues (Dolan et al., 1994). It is possible that in the fully flexed lumbar spine the forces arising from the passive tissue may result in inhibition of muscle activation (Indahl, Kaigle, Reikeräs, & Holm, 1997), allowing a more metabolic efficient and passive method of generating extensor torque.

The loading of passive tissues provides a likely explanation for the reduction in activation between mid and fully flexed postures, but there is a lack of literature that has investigated BE muscle activity during MVIC at the extremes of lumbar flexion. Studies have been conducted where ES muscle activation has been recorded at end range of lumbar flexion (Dolan & Adams, 1993; Dolan et al., 1994; Toussaint et al., 1995). However, these studies have used submaximal isometric exertions or had participants perform dynamic lifting tasks at submaximal effort. These studies showed ES muscle activation to decline sharply in the last 20% of lumbar flexion, to such an extent that there was total electrical silence of muscles at terminal flexion (Dolan et al., 1994; Kippers & Parker, 1984; Toussaint et al., 1995). The present study also found a significant decrease in paravertebral muscle activation at the end range of flexion but with recruitment remaining at 39% of their maximum. These findings are similar to those of submaximal lifting studies that have required participants to lift moderate weights between 10 and 15kg. (Dolan et al., 1994; Kippers & Parker, 1984; Toussaint et al., 1995). Toussaint et al. (1995) noted ES muscle activation levels of up to 20% of maximum during the dynamic lifting of a 15kg weight towards end range flexion.

5.2.4 Posture and EMG for each muscle division

All paravertebral muscle divisions demonstrated a similar reduction in activation between lordotic to mid postures and from mid to flexed postures, differing by a maximum of 5%. This was an interesting finding given that anatomical evidence would suggest that the UES, LES and multifidus muscle architecture differs considerably, and it has been implied that these muscle divisions may have varying roles when performing functional tasks in different lumbar postures (Panjabi et al., 1989; Toussaint et al., 1995). Whilst fibre orientation and location of the different division of ES may differ (Macintosh et al., 1993), distribution of muscle fibre types in the UES and LES are similar (Mannion et al., 1997). It has been argued that the similar fibre type distribution of the UES and LES may reflect their similar function during a number of tasks.

Only one other study (Dolan et al., 1994) is known to have compared activation levels of the UES and LES during a static lifting task. Dolan et al. (1994) recorded EMG of the UES and LES during MVIC of BE muscles at 12 points of lumbar flexion while the participant was pulling up on a floor mounted force gauge with their pelvis and lower limbs restrained in a standing frame. Similar to the current study, no significant differences were found between activation levels of the UES and LES in each lumbar posture. Dynamic, submaximal lifting tasks have been used to investigated the effect of posture on paravertebral muscle activation (Toussaint et al., 1995). It was found that activation of the UES remained above 60% of MVC during the lifting task, with the greatest activation occurring at full flexion. In contrast, muscle activation of the LES and multifidus diminished in more flexed postures (Toussaint et al., 1995). It was concluded that towards the end range of flexion the extensor torque produced by the LES is, in part, taken over by the UES, mainly due to the greater mechanical advantage that the UES displays in this position (Toussaint et al., 1995). However, care should be taken when interpreting results from dynamic lifting tasks, because muscle activation and recruitment patterns are often influenced by changes in angular velocity of the lumbar spine (Dolan et al., 1994; Hof, 1984).

The similar activation pattern of all three divisions, irrespective of their different function, attachment and innervation indicates that during relatively static tasks the UES, LES and multifidus are influenced by changes in posture in a similar way. This is also supported by anatomical studies which have suggested that the overlapping arrangement of the paravertebral muscle fascicles allow consistent forces to be generated as the spine flexes (Macintosh et al., 1993). It is thought that a decrease or increase in mechanical leverage of one fascicle is compensated by a reciprocal change in another (Macintosh et al., 1993). This has implications for the rationale often used to justify the isolation of individual ES muscles during exercise strengthening and functional rehabilitation programmes (O'Sullivan, Twomey, & Allison, 1997).

5.2.5 Neuromuscular efficiency

This study found the neuromuscular efficiency (NME) ratio to increase from lordotic to mid posture, with a greater increase occurring from mid to flexed postures. Previous research also found a rise in NME with an increase in lumbar flexion. Both Roy et al. (2003) and Tan et al. (1994) calculated the NME ratio of the LES in a neutral spine posture to be 1.0, rising linearly to 1.6 at 50^{o} of flexion and 1.67 at 35^{o} of flexion. The lordotic and mid postures in the present study represented approximately 20^{o} and 40^{o} of lumbar flexion, with a NME ratio of 1.2 and 1.5 for the respective positions, which is analogous to previous research. Unlike previous studies, which omitted to test the extreme of lumbar flexion, this study found a much higher NME ratio of 2.8 in full flexion. This represented an 87% increase in NME from the mid to end range of flexion, compared to only 23% increase from lordotic to mid posture.

Tan et al. (1993) suggested improvements in the NME ratio of the LES, was in part, the result of greater mechanical advantage of ES in mid-range flexion (35^{o}) where maximum torque was produced. However, Tan et al. only tested up to 35^{o} and other studies have suggested that 45^{o} is the position of optimal length-tension of ES (Raschke & Chaffin, 1996). While improvements in NME between the lordotic and mid posture could be the result of improved mechanical advantage due to optimal length-tension of ES, anatomical studies have demonstrated that at end range of lumbar flexion, the decrease in ES muscle moment arm length has a reduced mechanical advantage by up to 39% (Macintosh et al., 1993). These findings raises questions as to the mechanisms contributing to a sharp increase in NME at end range of flexion, as found in this study.

The most likely reason for this study finding a large increase in NME ratio at the end range of lumbar flexion can be explained by the change in the torque production from the active to the passive tissues of the lumbar spine as it flexes. As discussed earlier in this section, both biomechanical and anatomical studies have shown at end range of flexion there is less torque produced from active tissues of the spine and there is greater contribution to torque from the passive spinal structures, leading to relatively lower metabolic cost (Bogduk, 2012; Dolan et al., 1994; Raschke & Chaffin, 1996; Tan et al., 1993). While this study has shown end range of lumbar flexion to be the most neuromuscular efficient, high torque production combined with low muscle activation can potentially pose a high risk of back injury, suggesting high NME may not necessarily be advantageous.

5.3 Fixation

5.3.1 Fixation and torque

This study found there was no effect on BE torque when the pelvis was restrained compared to when the pelvis was free to move. Despite the wide spread use of pelvis fixation in studies investigating BE torque and posture (Dolan et al., 1994; Graves et al., 1990a; Kaigle et al., 1998; Kippers & Parker, 1984; Parnianpour et al., 1991; Roy et al., 2003; Smidt et al., 1983; Tan et al., 1993; Toussaint et al., 1995), only one other study appears to have investigated the efficacy of pelvis fixation on isometric BE torque. Smidt et al. (1983) compared the effect of minimal fixation (only the feet restrained), partial fixation (use of restraints across the thighs and lower leg but not the pelvis) and full fixation on BE torque at varying degrees of lumbar flexion, while seated on the Iowa trunk dynamometer. A significant reduction in extensor torque was found when using minimal fixation compared to partial and full fixation. However, no difference in extensor torque production was found between partial fixation and full fixation. These findings suggest that providing some type of fixation to the thighs, and not specifically to the pelvis, is important when using a trunk dynamometer. The present study did not fixate the lower limbs, only the pelvis. Consequently, it is difficult to compare the results of this study with those of Smidt et al., but the different forms of pelvis and lower body fixation may explain why in this study no influence of pelvis fixation on BE torque was found.

There is a scarcity of research into the efficacy of using pelvis fixation, although fixation has been widely used during studies investigating BE strength training regimes (Graves et al., 1994; Smith et al., 2011). In these studies, it was noted that it was virtually impossible to perform lumbar extension in sitting without some degree of fixation

(Udermann et al., 1999). This was also recognised in other studies comparing different fixation configurations, where some degree of fixation was required as opposed to using no restraints (Da Silva et al., 2009; Walsworth, 2004). It would appear that seated isokinetic devices are more effective when using some degree of pelvis or lower limb fixation and without fixation the ability to generate BE torque may be compromised.

The present study found no effect of pelvis fixation, which may be the result of using a different and more functional position that does not require fixation in order to perform the back extension task. Using isokinetic devices to measure BE torque appear to require pelvis fixation and it is plausible that this method is chosen for the convenience of testing and the experimental design. Unfortunately these devices are expensive, often not accessible in the clinical setting and do not assess trunk extensor torque in functional positions where the pelvis and lower limbs are free to move. In contrast, this study used a functional task that could be performed equally as well with or without fixation. Testing of the BE in functional positions seems important in order to study how the lumbar spine works in conjunction with the pelvis, not just in isolation (Yasukouchi & Isayama, 1995).

5.3.2 Fixation and EMG

In this study, the level of pelvis fixation did not influence the level of paravertebral muscle activation. No studies appear to have investigated the effect of pelvis fixation on ES muscle activation during maximal isometric exertions. Research that has investigated the effect of pelvis fixation on the activation of the BE muscles have often used dynamic protocols. Findings from these dynamic studies are inconclusive, with some studies finding increased levels of ES activation with fixation (Da Silva et al., 2009; San Juan et al., 2005) while others have found no effect on activation levels of the paravertebral muscle (Udermann et al., 1999; Walsworth, 2004).

The studies using dynamic protocols to investigate paravertebral activation, tend to restrain the pelvis and lower limbs, which arguably gives rise to a higher degree of stability than the method of fixation used in the present study; restraining only the pelvis. McGill (1991) proposed that when spinal stability is challenged, an alteration in ES recruitment pattern occurs in order to stabilise the spine in preference of generating extensor torque. It was suggested that when more external stability is given in the form of pelvis or lower limb fixation, this could potentially lead to alterations in ES activation levels, although this was not found in the present study.

5.4 Clinical Implications

The findings of this study contribute to the body of knowledge involving biomechanics of the lumbar spine. More specifically it has provided an understanding of how lumbar posture influences BE torque and the activation of the paravertebral muscles in a functional lifting position. This study has implications for people undertaking manual handling lifting tasks, for organisations that provide manual handling guidelines as well as health professionals involved in the management and prevention of LBI.

When a lordotic posture was maintained during a MVIC of the BE muscles, this study found low levels of torque and high levels of paravertebral muscle activation. This resulted in the lordotic posture having a low neuromuscular efficiency. In comparison, a flexed lumbar posture was found to produce high levels of torque and low levels of paravertebral muscle activation, showing that using a flexed lumbar spine when initiating lifting has a high neuromuscular efficiency. The high levels of NME in more flexed postures may provide some rationale for people choosing a flexed lumbar posture in preference to a lordotic lumbar posture when performing lifting activities (Straker & Duncan, 2000). The increased torque production in more flexed postures also has implications for health professionals and highlights the need to standardize lumbar posture when assessing BE torque.

The observed reduction in ES muscle activation concurrent to an increase in torque between the lordotic and mid-range of flexion, suggests that the mid-range posture has more optimal length-tension of the paravertebral muscles than a lordotic posture. In more flexed positions the passive properties of ES and the other soft tissues of the spine, as well as the recruitment of other muscles, may be important contributors to the production of extensor torque. This has importance in the development of future biomechanical models, which should consider the length-tension relationships and the viscoelastic properties of other passive tissues of the spine when modelling BE torque and ES activation.

Findings from the current study would indicate that using a mid-range posture during maximal lifting has the benefit of generating higher BE torque than when using a lordotic posture. The mid posture also has improved neuromuscular efficiency of the paravertebral muscles compared to the lordotic spine, yet it does not excessively load the passive spinal structures, which occurs at the end range of lumbar flexion. This study has also shown that when lifting a box from below knee height, it appears that one third of lumbar flexion is required, even when maintaining a maximally lordotic lumbar posture. Advising individuals to lift without flexing the lower back appears unrealistic.

Previous research has shown paravertebral muscles have the ability to increase their recruitment far into the flexion range when the participant is fixated to an isokinetic apparatus. However, this study showed paravertebral muscle activation to reduce with lumbar flexion, irrespective of pelvis fixation. The position of testing appears to be critical in how the paravertebral muscles respond to changes in lumbar posture. This highlights the importance of testing BE muscle strength in positions that simulate functional working activities, in order to improve understanding on how the BE muscles would normally be recruited. This also highlights the importance of using postures that simulate lifting activities for functional training and rehabilitation of the paravertebral muscles of clients involved in manual handling activities.

The findings of this study also have implications for the functional training of individuals involved in manual handling activities. Traditional physiotherapy rehabilitation techniques have focused on retraining individual divisions of ES and multifidus. The functional postures used in this study appear to activate all divisions of the paravertebral muscles equally and therefore, should be integrated into vocational and rehabilitation programmes involving the lumbar spine.

Previous research investigating BE torque have used non-functional positions and sought to isolate the lumbar spine by restraining the pelvis. The findings from this study showed pelvis fixation to have no effect on BE torque or muscle activation, which has implications for future studies into the biomechanics of lifting and questions the relevance of using pelvis fixation. This also has implication for the clinical setting and questions the justification for using fixation devices for functional strengthening and rehabilitating the paravertebral muscles.

5.5 Limitations of the Study

This study was designed to be methodologically sound, although a number of limitations have been identified.

Firstly, the study aimed to simulate maximal lifting ability in an initial lifting posture. In order to estimate the extensor force acting on the torso and eliminate variability in arm posture, a chain connected the chest harness to the force gauge. A more natural lifting technique would have involved a lifting posture that used the upper limbs.

The data from three participants were excluded from the analysis due to retro-reflective markers being hidden from the motion analysis cameras, or because markers became dislodged. Marker displacement due to skin movement, particularly at the pelvis, may have led to errors when determining body posture and therefore, should be considered a potential source of error.

A main aim of this study was to investigate BE torque and the activation of the paravertebral muscle group. However, the paravertebral muscles are not the only BEs, with latissimus dorsi, gluteus maximus and quadratus lumborum also having the capacity to generate an extensor moment on the lumbar spine. Determining the contribution of these muscles to the BE torque was not feasible in the current study due to the complexity of measuring EMG from these muscles. However, this was not seen as a major limitation as the paravertebral muscles have been estimated to produce up to 80% of the total extensor forces (Bogduk, 2012).

Pelvis fixation used in this study restricted pelvis movement through tension within the harness and its anchor points on the floor. This allowed unrestricted movement of the lower limbs and had the added advantage of enabling the retro-reflective markers to be visible to the motion capture cameras. Although this proved effective in restraining the pelvis, it was possible for the pelvis to move slightly during maximum back extension. A study by Montgomery et al. (2011) using a similar pelvis restraint system and found approximately 83% reduction of pelvic rotation during trunk flexion and rotation tasks.

Calculations of back extensor torque were calculated using two methods; internal derived moment and external derived moment. These methods used a slightly different centre of rotation for the lumbar spine: origin of the cylinder used to represent the trunk segment (Internal); and a point approximating to the centre of the L5/S1 disc (external). This may explain the slight systematic error in estimated torque resulting from the two methods. Also, the centre of mass of the upper body was estimated using Dempster (1955) data. It may have been preferable to estimate the centre of mass of the upper body for each participant by directly measuring kinematic data using a marker set to denote the upper body.

Finally, consideration should also be given to the fact that the task studied was static and attempted to simulate an initial lifting posture. Whilst there is likely to be high forces towards the initial stages of the lift, maximal effort may not occur during movement. Therefore, extrapolating these findings to a functional lifting task should be done with caution. It should also be borne in mind that this study was conducted on a group of healthy young adults and therefore, these findings may not transpose to an older population or those with a low back injury.

Chapter 6: Conclusion & Recommendations

6.1 Conclusion

The main findings of this study showed that lumbar posture had a significant effect on BE torque and paravertebral muscle activation. A posture simulating the initial stage of a lift showed that when maintaining a lordotic posture during a MVIC of the BE muscles, low levels of torque and high levels of paravertebral muscle activation occur. Consequently, the lordotic posture had a low neuromuscular efficiency when compared to mid-range and fully flexed lumbar postures. In contrast, performing the lift in a flexed lumbar posture resulted in high levels of torque and low levels of paravertebral muscle activation, showing that using a flexed lumbar spine has a high neuromuscular efficiency. Neuromuscular efficiency at the end range of flexion was approximately twice that reported in the literature, maybe because the majority of previous studies have not investigated BE torque at the extremes of lumbar flexion.

While end range of lumbar flexion has high efficiency, the literature would suggest that it is primarily achieved through the loading of passive spinal structures to resist bending moments rather than additional muscle activation. The loading of the passive tissues of the spine has been identified as a potential risk factor for LBI (Hoogendoorn et al., 2000). Using a lumbar posture similar to the mid position used in this study is likely to avoid the loading of the passive structures yet produce higher levels of neuromuscular efficiency compared to the lordotic lumbar posture. Adopting a mid lumbar posture may be useful in situations when exerting high BE torque is necessary.

During MVIC, the relative contribution of the UES, LES and multifidus were similar in each lumbar posture tested. Traditional physiotherapy rehabilitation programmes have often retrained a single division of the paravertebral muscles in non-functional positions that do not reflect the lumbar postures adopted during lifting activities (O'Sullivan et al., 1997). Findings from this study suggest that vocational and rehabilitation programmes should use functional tasks that involve the combined movements of the trunk and lower body in order to appropriately activate all divisions of the paravertebral muscles.

Pelvis fixation was not found to influence BE torque or muscle activation. This suggests that functional lifting tasks should not be concerned with restraining the pelvis when measuring BE torque and paravertebral muscle activation. This study has implications for the assessment of BE torque in the clinical setting, showing that pelvis fixation may not be required when assessing trunk extension torque.

6.2 Recommendations

This study identified five key areas for future research.

i. Additional studies should be undertaken to establish the effects of lumbar posture on torque and muscle recruitment patterns in an older population and in people who have sustained a low back injury to direct rehabilitation programmes more appropriately.
ii. Future studies should consider other BE muscles in addition to UES, LES and multifidus, which contribute to BE torque in different lumbar postures, as this may impact upon rehabilitation programmes.
iii. Further research should consider the response of BE muscles during the performance of dynamic functional lifting tasks that normally occur in everyday life.
iv. Additional research is required to establish the effect of asymmetrical trunk postures on torque production and paravertebral muscle activation during functional lifting tasks.
v. It would be of interest to health professionals for future research to establish clinical based measures of BE torque in a clinical environment, without the need for complex and expensive equipment.

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Appendices

Appendix A

M E M O R A N D U M

Auckland University of Technology Ethics Committee (AUTEC)

To: Grant Mawston
From: **Dr Rosemary Godbold and Madeline Banda** Executive Secretary, AUTEC
Date: 10 May 2011
Subject: Ethics Application Number 11/15 **The effect of lumbar spine position on back extensor torque and trunk muscle activation.**

Dear Grant

Thank you for providing written evidence as requested. We are pleased to advise that it satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC) at their meeting on 14 February 2011 and that on 19 April 2011, we approved your ethics application. This delegated approval is made in accordance with section 5.3.2.3 of AUTEC's *Applying for Ethics Approval: Guidelines and Procedures* and is subject to endorsement at AUTEC's meeting on 23 May 2011.

Your ethics application is approved for a period of three years until 19 April 2014.

We advise that as part of the ethics approval process, you are required to submit the following to AUTEC:

- A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/research/research-ethics/ethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 19 April 2014;
- A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/research/research-ethics/ethics. This report is to be submitted either when the approval expires on 19 April 2014 or on completion of the project, whichever comes sooner;

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are reminded that, as applicant, you are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

Please note that AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to make the arrangements necessary to obtain this.

When communicating with us about this application, we ask that you use the application number and study title to enable us to provide you with prompt service. Should you have any further enquiries regarding this matter, you are welcome to contact Charles Grinter, Ethics Coordinator, by email at ethics@aut.ac.nz or by telephone on 921 9999 at extension 8860.

On behalf of AUTEC and ourselves, we wish you success with your research and look forward to reading about it in your reports.

Yours sincerely

Dr Rosemary Godbold and Madeline Banda
Executive Secretary
Auckland University of Technology Ethics Committee
Cc: Laura Holder laura.holder@aut.ac.nz, Peter McNair

Appendix B

An investigation of back strength and muscle activation in different postures.

Volunteers required!

- This is part of a Physiotherapy research project investigating back muscle strength and activation in different postures.
- Participants must be aged between 18 and 35 years old.
- We welcome participants with or without a history of back pain.
- You should not have any other muscle, joint, or neurological disorder.
- You should not have received any spinal or abdominal surgery in the past.
- You will not be included in the study if you are pregnant.
- You are required to be able to speak and understand English.

For further information please contact Laura on laura.holder@aut.ac.nz or 021 488430.

Appendix C

Participant Information Sheet

Date Information Sheet Produced:

05. 07. 2011

Project Title

The effect of lumbar spine position on back extensor torque and trunk muscle activation.

An Invitation

My name is Laura Holder and I am a physiotherapist. I would like to invite you to take part in this study that is investigating the effect of lower back position on how different parts of the back muscles are activated and produce force. I am undertaking this research as part of a Masters of Health Science. Before you accept this invitation please read the following outline of the study. You will need to make a decision about taking part in this study prior to attending the testing session. Your participation is entirely voluntary and you can withdraw from the study at any time without giving a reason.

What is the purpose of this research?

This purpose of this research is to determine if different parts of the back muscles work differently when the back position is changed and also when the pelvis is restricted from moving compared to when it is free. It will mainly be studied in people who have no back pain, with four or five case studies on people who have low back pain. Since low back pain affects one in two New Zealanders and most injuries are associated with a forward bent position, knowing how muscles work may impact upon the rehabilitation that physiotherapist plan for their back pain clients. The successful completion of this study may lead to the information being presented at professional conference or within a professional journal. It will also give the principle researcher a Masters of Health Science qualification.

How was I identified and why am I being invited to participate in this research?

You as a participant are welcomed to take part in this research because you have responded to advertisement requesting volunteers for this study. We are seeking volunteers who are aged 18-35yrs old, whether or not you have back pain. You will be excluded if you have had previous spinal or abdominal surgery, generalized muscle joint or neurological disorders, have a medical condition that may preclude maximum effort strength testing, are pregnant or unable to speak and understand English.

Participants must also ‘fall’ into one of two categories:

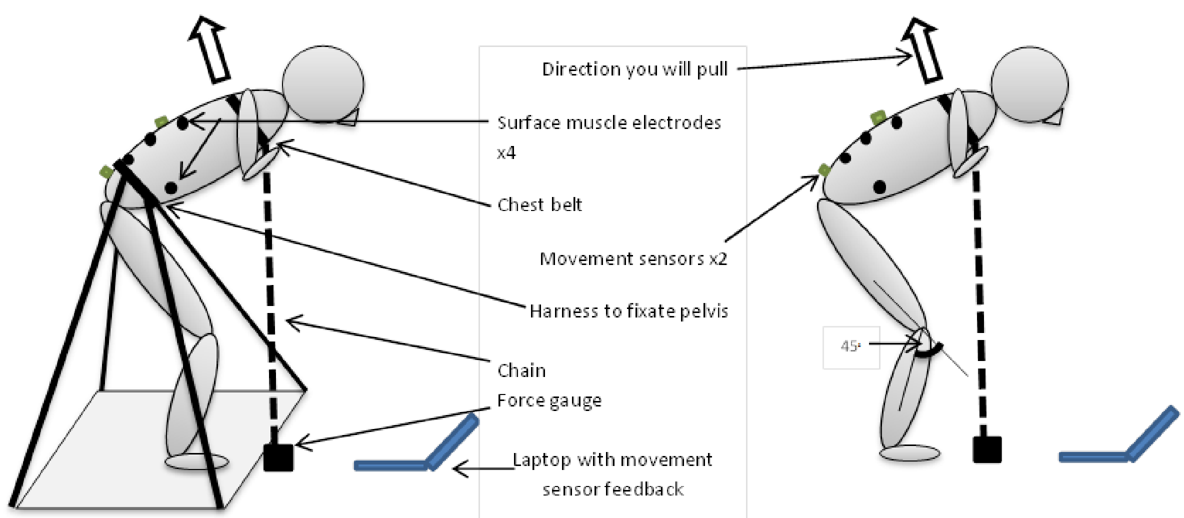
1. If you have not experience any pain originating from the back in the last 12 months.

or

2. If you have low back pain, your symptoms must have been present for at least three months in the last 12 months, have experienced pain in the last month but not have radiating leg symptoms.

What will happen in this research?

Your participation will involve attending the AUT University North Shore Campus for one session of data collection. It is advised to come dressed in loose fitting clothes and wear shorts. You will be first required to complete a consent form and there will be opportunity to ask me any questions about what is involved. You will then be required remove your top (a modesty gown will be provided so your front will remain covered but exposures the skin on your back). Next, you will be asked to lie on a bed on your front and have three pads/electrodes attached to the right side of your back muscles and one to the front of your tummy muscles, to measure their activity. You will then be required to undertake a 10 minute warm-up consisting of five minutes on a static bike followed by practice lifts at 50 percent of your maximum effort in the test position. Two small movement sensors will then be placed on your lower back to measure spinal motion (see figure 1) and this connects to the laptop. Approximately 30 small reflective markers will be placed on your legs, and trunk that will enable a nine-camera system to detect your body position. This will allow me to convert the reading on the force gauge in to torque measurement. Testing will then begin. With your knees semi-bent you will be asked to stoop over as though you are about to lift a box off the floor. The chest harness that I would like you to wear will have a small chain connecting it to the force gauge of the floor. Maintaining this position your back muscles will be tested statically in three different back positions; maximum bent, maximum arched and half way between. The laptop will give you feedback so you will be able to maintain the correct position. You will be tested under two different conditions. One is with the pelvis fixed (see figure 1a) and the other without pelvis fixation (see figure 1b). In each of these positions you will be required to pull up on the chain and sustain a maximum effort for five seconds. This will be repeated three times in each position with a two minute rest where you will be able to stand upright. In total, you will be required to perform 18 maximum contractions lasting five seconds each.



1a. Fixed test position 1b. Unfixed test position

Figure 1: Diagram of the test position.

What are the discomforts and risks?

The risks involved are minimal. However, when doing any unfamiliar activity it can result in mild muscle discomfort the following day but this tends to be short lived. Undertaking maximum muscle contractions poses a low risk of muscle strain. Back injury is not expected as similar methods have been used in a previous low back study with no incidence of back injury. If you have low back pain, your condition may or may not be slightly aggravated. The skin underlying the EMG electrodes may show some redness and minor irritation for several hours after the completion of the experiment.

How will these discomforts and risks be alleviated?

You will be able to terminate testing at any time. You will be required to perform a 10 minute standardised warm up which will include riding an exercise bike for five minutes followed by lifting a box 10 times from floor to mid-thigh. To further assist warming up, you will then perform a specific warm-up consisting of three back extensions at 50 percent of maximal effort in the different test positions. This warm-up will help to reduce the chance of any muscle strain or soreness occurring while making the testing procedure feels familiar.

A mandatory rest between test positions (two minutes) will also decrease the risk of strains occurring as a result of fatigue. For additional comfort, following each maximum effort you will be able to unclip yourself from the chain and stand upright.

I am a registered physiotherapist who is trained in cardiopulmonary resuscitation will be doing the testing therefore I will be able to identifying potentially hazardous situations that may occur during the testing. You will be able to communicate with me at all times during the testing procedures and may cease participation in the study at any time. You will be given my contact details should there be any discomfort following the testing session.

To address the potential skin irritation the skin will be cleaned before and after application of the electrodes. Aloe vera cream will be available in the laboratory to reduce any irritation if required. To minimise muscle soreness or strains you will be given a warm-up as outlined in 'What will happen in this research'. Aloe vera will also be available to sooth any minor skin irritation where the electrodes were placed.

What are the benefits?

You will receive no direct benefit from participation in the study. The study findings will help to determine how the different sections of the back muscles work in different ranges of lumbar flexion and under the different conditions of pelvis fixation. Knowing what normally occurs and comparing to case samples with people with lower back pain will provide necessary information on how back muscle muscles are affected by injury. This in turn will help guide the rehabilitation of back injuries. It will also benefit researchers in future investigations related to the optimal back muscle rehabilitation methods. The researcher will benefit by gaining a Masters of Health Science.

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?

No material which could personally identify you will be used in any reports in this study. For the analysis of the data, each participant will be given a confidential coding, so that their information can be linked. After the analysis, the data will be kept locked in a filing cabinet in the Health and Rehabilitation Research Institute, Auckland University of Technology. The consent forms for the study will be stored in a similar fashion but will be kept separate from the data collected. All results are pooled, so no names or participants or any material that could identify an individual will be published or presented.

What are the costs of participating in this research?

There are no monetary costs involved in taking part in this study. The only cost to you is the time in attending the testing session which will be no longer than two hours.

What opportunity do I have to consider this invitation?

Once you have received and read this information you will need to inform the researcher if you intend on participating within two weeks. If you are interested in participating, you will be given a verbal explanation of the study and any questions you have will be answered. An appointment to attend will be made at least one week away, so that you have a further week to consider whether you wish to take part. Please remember that your participation is voluntary and you are able to withdraw from the study at any time by informing the researcher.

How do I agree to participate in this research?

You need to contact Laura Holder, 09 921 937 or 021 488430, laura.holder@aut.ac.nz Before participating in this research project you will be required to sign a consent form to acknowledge that you understand what is involved and that you are happy to participate. This will be done on the day of testing and before testing begins.

Will I receive feedback on the results of this research?

You are given an opportunity on the Consent Form to indicate if you would like feedback on the research project. If you answer "yes" to this, a 1-page summary of the results of the project will be sent to the contact details that you provide on the Consent Form.

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, Dr Grant Mawston, grant.mawston@aut.ac.nz tel: 09 921 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?

Researcher Contact Details:

Laura Holder, laura.holder@aut.ac.nz tel: 09 921 9377, mob: 021 488430

Project Supervisors Contact Details:

Dr Grant Mawston, grant.mawston@aut.ac.nz tel: 09 921 9999 ext 7180

Associate Professor Mark Boocock, mark.boocock@aut.ac.nz tel: 09 921 9999 ext 7167

Approved by the Auckland University of Technology Ethics Committee on 19th April 2011, AUTEC Reference number 11 / 15.

Appendix D

Consent Form

Project title: The effect of lumbar spine position on back extensor torque and trunk muscle activation.

Project Supervisor: Dr Grant Mawston.

Researcher: Laura Holder.

○ I have read and understood the information provided about this research project in the Information Sheet dated **05.07.2011**.

○ 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.

○ I am not suffering from heart disease, high blood pressure, any respiratory condition (mild asthma excluded), any infection, any illness or injury, other than back pain, that impairs my physical performance.

○ I am not pregnant.

○ I agree to take part in this research.

○ I wish to receive a copy of the report from the research (please tick one):

Yes○ No○

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

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Date:

Approved by the Auckland University of Technology Ethics Committee on 19th April 2011 reference number 11 / 15.

Note: The Participant should retain a copy of this form.