NEUROMUSCULAR AND BIOMECHANICAL
ADAPTATIONS ASSOCIATED WITH CHRONIC
LOW BACK PAIN

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Abstract

Neuromuscular adaptations are a hallmark presentation of chronic low back pain (CLBP). People with CLBP have demonstrated trunk muscle co-contraction, trunk muscle weakness and decreased lumbar range of motion (ROM). The relationship between traditional neuromuscular assessments (e.g., lumbar ROM or trunk muscle strength) and self-reported disability is low. Arguably, the measurement of trunk ROM and trunk muscle strength are non-specific and cannot adequately explain the variances in disability in CLBP. Perhaps novel assessments assessing different domains of neuromuscular adaptations can better explain variances in self-reported disability in people with CLBP.

Novel assessment of knee muscle force control demonstrated an impairment in the quadriceps ability to produce an accurate force in anterior cruciate ligament reconstructed (ACLR) population compared to healthy cohort. The assessment of muscle force control has not been performed in the CLBP cohort. Similarly, the assessment of inter-joint coordination using relative phase angle analysis was originally developed to identify safe lifting technique in manual workers but has not been performed in CLBP cohort. Therefore the aims of this thesis were i) to compare CLBP and healthy participants on novel trunk muscle force control assessment, ii) to compare CLBP and healthy participants on the assessment of lifting inter-joint coordination and iii) to investigate the relationship between self-reported disability and the aforementioned novel neuromuscular assessments in people with CLBP.

To address these aims, four cross-sectional studies were conducted. Forty three participants with CLBP (mean Oswestry Disability Index (ODI) = 22.1 ± 13.2, mean pain Numerical Rating
Scale (NRS) = 3.6 ± 1.9 and 29 matched healthy control participants were recruited. Inter-session reliability were assessed on 17 CLBP participants and 16 healthy participants.

Study 1 compared and described the participant characteristics with respect to pain and self-reported disability quantified using NRS and ODI respectively. The findings demonstrated that people with CLBP reported significantly higher level of pain and disability than healthy participants. Incidentally, CLBP participants were nine years older than healthy people.

Study 2 compared lumbar extensor maximal voluntary isometric contraction (MVIC) and lumbar extensor muscle force control in people with CLBP and healthy people using a novel force matching protocol adapted from studies in the quadriceps. The assessment protocol utilised a force target varied between 20%-50% MVIC \( \text{(i.e., increasing and decreasing force output)} \). No significant differences in lumbar extensor MVIC were evident between the groups. CLBP group demonstrated increased force matching error compared to healthy people, suggestive of impairment in lumbar extensor muscle force control. People with CLBP demonstrated more error when increasing force production compared to when decreasing force production.

Study 3 compared kinematic and vertical ground reaction force assessment during a lifting task in people with and without CLBP. Participants with CLBP were characterised based on their disability phenotype \( \text{(i.e., higher and lower disability)} \). In addition to the measurement of trunk and lower limb ROM and angular velocity, trunk and lower limb joint coordination was assessed using relative phase angle analysis. There was no significant difference in trunk and lower limb ROM and angular velocity between all groups. Both groups demonstrated similar lifting trunk and lower limb joint coordination pattern and vertical ground reaction force
pattern. However, people with CLBP took longer to lift a light load compared to healthy controls.

Study 4 investigated the relationship between the assessments outlined in Studies 2 and 3 and ODI. There was no significant correlation between lumbar extensor MVIC, trunk and lower limb ROM and angular velocity during lifting and ODI. One unit increase in error when decreasing force production was associated with 47% increase in ODI. Similarly, one second increase in coordinated movement duration between the thorax and lumbar spine was associated with 47% increase in ODI. The amount of error when decreasing force output and duration of coordinated movement between the thorax and lumbar spine explained 27% of ODI variance. The findings of this thesis suggest that the assessment of lumbar extensor muscle force control and inter-joint coordination during lifting explain a significant portion of self-reported disability; thus, these measures are valid for CLBP patients. Future research is required to investigate whether improvement in the novel neuromuscular variables are associated with a decrease in self-reported disability.
Declaration

This is to certify that:

1. The thesis is my original work toward the PhD,
2. All acknowledgement has been made with respect to the materials used in this thesis,
3. All permissions with respect to the materials used in this thesis have been granted by relevant parties,
4. This thesis is fewer than 100,000 words exclusive of tables, figures, bibliographies and appendices,
5. All research conducted in this thesis has been approved by The University of Melbourne Human Research Ethics Committee.

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This thesis would not have been possible without the help of a group of supportive people. I would like to thank the following:

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Publications and conference papers

The publication and conference documents derived from the research in this thesis are listed below:


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Abbreviations

3D Three dimensional
ACLR Anterior cruciate ligament reconstructed
ANCOVA Analysis of covariates
ANOVA Analysis of variance
ASIS Anterior superior iliac spine
AUC Area under curve
BMI Body mass index
CLBP Chronic low back pain
cm Centimetres
Dev Deviation
EMG Electromyography
GRF Ground reaction force
Hz Hertz
ICC Intraclass-correlation coefficient
kg Kilogram
LE Lumbar extensor
LBP Low back pain
MARP Mean absolute relative phase
MCID Minimally clinically important difference
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
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<tbody>
<tr>
<td>mm</td>
<td>millimetres</td>
</tr>
<tr>
<td>MVIC</td>
<td>Maximal voluntary isometric contraction</td>
</tr>
<tr>
<td>N</td>
<td>Newton</td>
</tr>
<tr>
<td>NR</td>
<td>Not reported</td>
</tr>
<tr>
<td>NS</td>
<td>Not significant</td>
</tr>
<tr>
<td>Nm</td>
<td>Newton meters</td>
</tr>
<tr>
<td>NRS</td>
<td>Numerical Rating Scale</td>
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<tr>
<td>ODI</td>
<td>Oswestry Disability Index</td>
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<tr>
<td>PPI</td>
<td>Present Pain Index</td>
</tr>
<tr>
<td>QBPDS</td>
<td>Quebec Back Pain Disability Scale</td>
</tr>
<tr>
<td>RMDQ</td>
<td>Roland Morris Disability Questionnaire</td>
</tr>
<tr>
<td>RMSE</td>
<td>Root mean squared error</td>
</tr>
<tr>
<td>RMSE_{A}</td>
<td>Root mean squared error ascending phase</td>
</tr>
<tr>
<td>RMSE_{D}</td>
<td>Root mean squared error descending phase</td>
</tr>
<tr>
<td>RMSE_{T}</td>
<td>Root mean squared error total</td>
</tr>
<tr>
<td>ROM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>SD</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>SEE</td>
<td>Standard error of the estimate</td>
</tr>
<tr>
<td>SEM</td>
<td>Standard error of measurement</td>
</tr>
<tr>
<td>VAS</td>
<td>Visual Analogue Scale</td>
</tr>
<tr>
<td>Vel</td>
<td>Velocity</td>
</tr>
<tr>
<td>VRS</td>
<td>Verbal Rating Scale</td>
</tr>
<tr>
<td>WBB</td>
<td>Wii Balance Board</td>
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1.1. Thesis overview

1.1.1. Research background

The research reported in thesis is an accumulation of work conducted within the Centre for Health, Exercise and Sports Medicine at The University of Melbourne in collaboration with Kieser Training Physiotherapy (South Melbourne and Brighton) between November 2014 and February 2016. Kieser is a large physiotherapy clinical group with multiple sites across Australia. Each Kieser centre is equipped with unique cable-based strength training equipment and, in addition, MedX (Ocala, FL) lumbar extensor dynamometry equipment.

All testing was conducted after consenting participants with chronic low back pain (CLBP) completed their initial physiotherapy screening session at Kieser. Participants received their treatment on their second physiotherapy session. Reliability studies were performed on willing participants prior to the second physiotherapy visit (i.e., prior to CLBP intervention). Pilot testing sessions were conducted on weekends at a suitable time for participants. Despite this research collaboration, Kieser provided no financial support and there were no conflicts of interest.

1.1.2. Thesis structure

The primary aim of this thesis was to elucidate relationships between neuromuscular variables derived from novel open and closed kinetic chain tests and self-reported disability of people with CLBP. To achieve this aim, four cross-sectional studies were performed. Novel open and
closed kinetic chain tests were developed (based upon previous published studies) and were subject to reliability testing. The thesis structure can be summarised as follows:

- **Chapter one** is an overview of the thesis and an introduction to the research question.

- **Chapter two** is a review of the literature pertaining to disability, open kinetic chain assessments of trunk muscle force control and closed kinetic chain assessment of trunk and lower limb joint coordination during lifting. In addition, this chapter also examines the current evidence pertaining to the relationship between CLBP-related neuromuscular adaptations and self-reported disability.

- **Chapter three** is a descriptive study comparing CLBP and healthy control participants with respect to pain and disability.

- **Chapter four** incorporates a study that investigated lumbar extensor muscle force control in CLBP and healthy control participants. Specifically, sub-maximal force control was assessed using a novel, open kinetic chain protocol. Participants were required to increase and decrease their level of isometric lumbar extensor muscle force (i.e., 20 and 50% maximal voluntary contraction) in response to a fluctuating force ‘target’. Lumbar extensor muscle force control was calculated by quantifying the disparity between participant-generated force and the target force using root mean square error.

- **Chapter five** incorporates a study that examined coordination between the lumbar and thoracic segments and joints of the lower limb during lifting in CLBP and healthy control participants. Inter-joint coordination data were interpreted with respect to lifting-related vertical ground reaction force (GRF) data.
- **Chapter six** addresses the main aim of the thesis; that is, to investigate the relationship between open and closed kinetic chain neuromuscular variables and self-reported disability in people with CLBP using multivariate regression analysis. This chapter is divided into three parts. Part one aimed to investigate the association between novel open kinetic chain neuromuscular variables (discussed in Chapter 4) with self-reported disability of CLBP participants. Part two investigated the relationship between novel closed kinetic chain variables (discussed in Chapter 5) with self-reported disability. Part three investigated the predictive capacity of variables derived from both open and closed tests with respect to self-reported disability. In doing so, this final study addresses the validity of these novel test outcomes in CLBP participants.

- **Chapter seven** summarises the key findings; addresses the strengths and limitations; and outlines potential directions for future research.
1.1.3. Nomenclature within the thesis

A number of terms are used consistently within this thesis. Namely:

- **Neuromuscular adaptation variables** refers to motor control variables associated with CLBP in open and closed kinetic chain conditions.

- **Trunk** refers to the thoracic and lumbar region of the spine.

- **Trunk extensor muscles** refer to muscles which attach to the posterior aspect of the spine that could exert compressive load and sagittal extension torque on the trunk.

- **Open kinetic chain assessments** refer to motor control assessments whereby the distal body parts are not constrained. This allows for isolated muscle or limb assessment.

- **Closed kinetic chain assessments** refer to motor control assessments where the lower limbs are affixed to the ground. This allows for complex interaction between distal and proximal joints.

- **Self-reported disability** refers to self-perceived activity limitations assessed using validated questionnaires such as the ODI.
1.2. Introduction

Low back pain (LBP) is one of the most common musculoskeletal conditions that creates considerable personal and global burden (Buchbinder et al., 2013; Walker, 2000). The prevalence of LBP is high amongst young adults (i.e., 74.4%) (Jeffries, Milanese, & Grimmer-Somers, 2007) and the risk of LBP cumulatively increases over adult life (Docking et al., 2011). The natural progression of LBP is extremely variable with symptoms lasting from days to several years (Costa et al., 2012). However, the prognosis of those who present with a recent (i.e., acute) LBP episode is typically not favourable due to the slow recovery process. As such, nearly a third of patients don’t recover from their presenting condition after one year (Henschke et al., 2008). People with persisting back symptoms are often defined in the literature as suffering from CLBP (Deyo et al., 2014). Despite an enormous amount of funding allocated to the treatment of CLBP, the disability related to this condition continues to increase (Deyo, Mirza, Turner, & Martin, 2009).

Disability is defined as restriction in activities of daily living and capacity to participate in social activities (WHO, 2002). Disability is an inter-relation between personal factors such as self-perceived capacity, physical impairments, personal beliefs and environmental factors (Jette, 1994) and is usually assessed using validated questionnaires such as the ODI (Fairbank & Pynsent, 2000). Self-reported disability is associated with care seeking behaviour and a decline in work-related productivity (Buchbinder et al., 2013; Ferreira, Machado, et al., 2010). Thus, the main aim of clinical management of CLBP is a reduction in disability level (Beattie & Maher, 1997).

Along with an increase in self-reported disability, changes in the sensorimotor system are also commonly reported in people with CLBP (Hides, Stanton, Mendis, & Sexton, 2011; Hodges
& Richardson, 1996; Laird, Gilbert, Kent, & Keating, 2014; van Dieen, Cholewicki, & Radebold, 2003). These changes include: impairment of trunk muscle activation (van Dieen et al., 2003), motor planning (Hodges, 2001), range of motion (ROM) (Laird et al., 2014), speed of movement (Sanchez-Zuriaga, Lopez-Pascual, Garrido-Jaen, de Moya, & Prat-Pastor, 2011), movement coordination (Esola, McClure, Fitzgerald, & Siegler, 1996), proprioception (Willigenburg, Kingma, Hoozemans, & van Dieen, 2013), lumbar extensor muscle size (Hides, Gilmore, Stanton, & Bohlscheid, 2008) and lumbar extensor muscle strength (Steele, Bruce-Low, & Smith, 2014). Collectively known as neuromuscular adaptations, these often become targets for conservative treatment (Crow, Pizzari, & Buttifant, 2011; Laird, Kent, & Keating, 2012). However, it is interesting to note that treatment strategies designed to specifically address these neuromuscular adaptations only result in modest improvements in disability and no single treatment appears to be more effective than another (Ferreira et al., 2007; Hayden, Van Tulder, Malmivaara, & Koes, 2005; Maher, 2004; Van Dillen et al., 2016).

One of the main reasons for this null effect is the low to non-existent relationship between self-reported disability and traditional neuromuscular variables in people with CLBP (Al-Obaidi, Nelson, Al-Awadhi, & Al-Shuwaie, 2000; Deyo, 1988; Mannion, Junge, et al., 2001; Parks, Crichton, Goldford, & McGill, 2003; Renkawitz, Boluki, & Grifka, 2006; Waddell, 1987). Traditional neuromuscular variables such as lumbar ROM or lumbar extensor strength may not assess the appropriate domain of self-reported disability. Given the weak relationship between traditional neuromuscular variables and self-reported disability, targeted management of CLBP sufferers has recently shifted towards the psychological aspects of disability such as the treatment of pain-related fear (Nicholas & George, 2011). However, this treatment approach also has limited clinical application as it is currently unclear who would benefit from
this form of intervention (Kamper et al., 2010; Pincus, Smeets, Simmonds, & Sullivan, 2010). Similarly, interventions that specifically target psychological impairments (e.g., pain-related fear) are not superior to those that target physical impairments (Macedo, Smeets, Maher, Latimer, & McAuley, 2010; van Middelkoop et al., 2011). For this reason, researchers agree that a considerable amount of research is still required in all aspects of CLBP assessment (Fairbank et al., 2011; Ford & Hahne, 2013), including neuromuscular impairments (Maher, 2004).

Recently, several studies involving anterior cruciate ligament reconstructed (ACLR) and matched healthy cohorts incorporated a novel open kinetic chain test in order to derive quadriceps muscle force control – defined as the ability to produce accurate force (Perraton et al., 2016; Telianidis et al., 2014). The results of these studies suggest that ACLR patients exhibit impaired accuracy of quadriceps force output compared to healthy controls (Perraton et al., 2016; Telianidis et al., 2014) which has been associated with decreased hop performance (Bryant, Pua, & Clark, 2009). To date, analysis of muscle force control has not been performed in people with CLBP. Hence, rather than simply examining the quantity (i.e., maximum voluntary isometric contraction; MVIC) of trunk muscle force output like previous studies, this novel analysis could yield new insight into the quality of CLBP-related trunk muscle force production.

The closed kinetic chain assessment of inter-joint coordination between the trunk and lower limb was first developed to assess safe lifting techniques of healthy, manual workers (Burgess-Limerick, Abernethy, & Neal, 1993). Assessment of lifting kinematics is often combined with measurement of vertical GRF’s in order to describe lifting kinetics (Sanchez-Zuriaga et al., 2011). Relative phase angle analysis is a methodology used to assess movement coordination
between the trunk and lower limb joint by describing the phase lag (i.e., the deviation from synchronous movement) of two joints at any point in time throughout lifting movements (Burgess-Limerick et al., 1993; Burgess-Limerick, Abernethy, Neal, & Kippers, 1995). Interestingly, there has been no study to date that utilised a combination of relative phase angle analysis and vertical GRF assessment to assess lifting techniques of people with CLBP. Therefore, it is currently unknown whether people with CLBP demonstrate different relative phase angles between the trunk and lower limb and vertical GRF during lifting compared to healthy controls.

Moreover, the relationships between neuromuscular variables derived from novel open and closed kinetic chain tests and self-reported disability have not been explored in the literature. Therefore, these neuromuscular variables have, to this point, not been validated in people with CLBP. Thus, this thesis aims to:

1. Describe and compare CLBP and matched healthy participants with respect to pain and self-reported disability,
2. Compare CLBP participants with matched healthy controls using a novel open kinetic chain neuromuscular assessment test of lumbar extensor force accuracy,
3. Compare CLBP participants with matched healthy controls using a novel closed kinetic chain neuromuscular assessment test of inter-joint coordination during lifting, and
4. Investigate the relationship and predictive capacity of neuromuscular variables derived from open and closed kinetic chain tests and self-reported disability.
Chapter 2

Literature Review

2.1. Introduction

In order to provide the foundational background and rationale for the studies included in this thesis, it was necessary to review the literature pertaining to the relevant aspects of CLBP namely the definition of CLBP, the impairments in neuromuscular control and disability associated with CLBP and how they are assessed. The ultimate aim of the research reported in this thesis was to investigate the relationship between impairments in neuromuscular control measured during open and closed kinetic chain trunk assessments and self-reported measure of disability in people with CLBP. In open kinetic chain assessment, distal body parts are typically unconstrained allowing isolated assessment of a body segment or muscle (Augustsson & Thomee, 2000). Alternatively, neuromuscular assessment of the trunk could also involve complex interactions of multiple joints or in closed kinetic chain condition. During closed kinetic chain assessment, lower limbs are usually constrained because the sole of the foot is in contact with the ground allowing movements in the distal and proximal joints to occur contemporaneously (Augustsson & Thomee, 2000).

Although the prevalence of CLBP peaks at the age of 80 (Hoy et al., 2014), there have been reported differences in neuromuscular characteristics in younger (i.e., 18-65 years old) and older adults (i.e., above 65 years old) (Burgess, Hillier, Keogh, Kollmitzer, & Oddsson, 2009; Enoka et al., 2003). Tissue mechanics in children (<18 years) and the older adults (>65 years) are quite different to adults aged 18 to 65 (Frost & Schonau, 2000; Pereira et al., 2013; Volpi, Nazemi, & Fujita, 2004). Additionally, functional disability in older adults is influenced by
chronic co-morbidities (e.g., arthritis and diabetes) in addition to CLBP (Stewart Williams et al., 2015), whereas in children disability is typically associated with school-related activities (Watson et al., 2002). Thus, this literature review only focuses on individuals with CLBP aged between 18 and 65 years old.

2.2. Definition of chronic low back pain

The definition of LBP in the literature is highly variable. Most commonly, LBP is defined by a combination of its topography (i.e., the area of the body where the pain is ascribed) and temporality (i.e., the duration of LBP since onset) (Dionne et al., 2008; Hoy, Brooks, Blyth, & Buchbinder, 2010). By its topography, LBP has often been defined as pain in the area between the inferior angle of the twelfth rib (T12) to the inferior gluteal fold with or without associated leg pain (Walker, 2000) which usually is described using pain drawings (Uden, Astrom, & Bergenudd, 1988). By its temporality, LBP is most commonly sub-grouped as short-term LBP (acute LBP) and long-term LBP (CLBP) (Von Korff & Saunders, 1996).

Acute LBP is typically defined as LBP that has persisted for less than three months (Dionne et al., 2008). CLBP is defined as LBP that has persisted for at least three months and has resulted in pain on at least half the days in the past six months (Deyo et al., 2014). No standardised measure exists in the literature to define either acute or CLBP; thus, these definitions have been largely derived by collective agreement of expert panels (Deyo et al., 2014; Stanton, Latimer, Maher, & Hancock, 2010, 2011).
2.3. Anatomical and physiological considerations in people with chronic low back pain

It has been long postulated that the stability of the spine when controlling movements is controlled by three interdependent sub-systems: control (neural), passive (osteo-ligamentous) and active (muscular) (Panjabi, 1992a; Sjølander, Johansson, & Djupsjobacka, 2002) (see figure 2.1).

2.3.1. Trunk osteo-ligamentous system

The osteo-ligamentous sub-system (e.g., the lumbar zyapophyseal joints, the intervertebral discs and the vertebral bodies) provide support for axial loading and trunk movement (Denis, 1984). The lumbar zygapophyseal joint facets are angled towards the sagittal plane (ranging from 82°-86° with increasing sagittal orientation towards the lower lumbar segment) that primarily promote movements in flexion and extension (Jaumard, Welch, & Winkelstein, 2011).

Injuries to the osteo-ligamentous system (e.g., posterior longitudinal ligament strain during heavy lifting (Cholewicki & McGill, 1992)) have been hypothesised to disrupt the afferent input to the central nervous system leading to abnormal signals to the muscular system, which, in turn, could lead to excessive loading on the zygapophyseal joints and ligaments, perpetuating pain in people with CLBP (Panjabi, 2006; Sjolander et al., 2002).
2.3.2. Trunk muscular system

The osteo-ligamentous system alone has been proven unable to sustain loads associated with activities of daily living (Crisco, Panjabi, Yamamoto, & Oxland, 1992). Thus, to maintain posture during daily activities such as bending forward or lifting, the lumbar spine requires the assistance of the complex trunk muscular system (Bergmark, 1989; Cholewicki, Panjabi, & Khachatryan, 1997). Morphologically, the lumbar muscles can be divided into three groups: i) the short intersegmental muscles (interspinales and the intertransversarii mediales, the polysegmental muscles), ii) the multifidus (lumbar components of the longissimus and iliocostalis), and iii) the long polysegmental muscles (the thoracic components of the longissimus and iliocostalis which attach to the ilium and sacrum) (Bogduk, 2012).

The erector spinae muscle group consists of longissimus thoracis pars thoracis, iliocostalis lumborum pars thoracis, longissimus thoracis pars lumborum and iliocostalis lumborum pars
lumborum (Macintosh & Bogduk, 1987) (see figure 2.2). The two thoracic components originate from the thoracic transverse processes and ribs at the level of T5 and insert into the erector spinae aponeurosis; whereas, the two lumbar components arise from the accessory processes and L1 to L4 transverse processes and insert to the ilium (Macintosh & Bogduk, 1987). The erector spinae have been established as primary movement generators of the trunk during extension movement and eccentrically controlling trunk flexion (Bogduk, Macintosh, & Pearcy, 1992).

The lumbar multifidus consists of five distinct fascicles which originate from the spinous processes of the lumbar vertebrae and insert to the mammillary processes, the caudo-medial aspect of the posterior superior iliac spine and the deep surface of erector spinae aponeurosis on the sacrum (Macintosh, Valencia, Bogduk, & Munro, 1986). The multifidus fibres that span across two spinal levels and attach to the lamina, mammillary process and zygapophyseal joint capsules are referred as deep multifidus whereas the multifidus fibres which span over two vertebral levels are termed superficial multifidus (see figure 2.2) (Jemmett, MacDonald, & Agur, 2004). The multifidus serves as a posterior sagittal rotator and due to its short moment arm, it also serves as the stabiliser of lumbar extension and rotation (Bogduk, 2012). The role of multifidus as a stabiliser of the lumbar spine is supported by the high number of muscle spindles (see section 2.3.3) in its deep fibres (Brumagne, Cordo, Lysens, Verschueren, & Swinnen, 2000) and how the multifidus is activated early prior trunk perturbations (Briggs, Greig, Bennell, & Hodges, 2007).
The activity of the trunk extensor muscles are critical in maintaining lumbar posture during activities which require generation of a lumbar spine extensor moment during lifting (Bogduk, 2012). Given the attachments of trunk extensor muscles across the lumbar and thoracic spine, contraction of these muscles during lifting pull the thoracolumbar fascia taut and increase trunk stiffness (Cholewicki & VanVliet, 2002; Neumann, 2010). The forces generated by these muscles are transmitted through the thoracolumbar fascia, which has fibres spanning from the cervical region to the sacrum, erector spinae aponeurosis and sacrotuberous ligament, to the lower extremities (i.e., gluteus maximus and hamstrings) (Willard, Vleeming, Schuenke, Danneels, & Schleip, 2012).
2.3.3. Neural system

In order to maintain trunk posture, the neural system receives afferent information (i.e., sensory input) from the muscular and osteo-ligamentous systems (Panjabi, 1992a, 2006). The afferent information is projected to supraspinal structures through the dorsal column, spinothalamic, spinoreticular and spinocerebellar pathways to the cerebellum, the brain stem, the retinacular formation, the thalamus and the somatosensory cortex which in turn evokes descending signals involved in the reflex loops (i.e., α-motoneurones and γ-muscle spindles) (Sjolander et al., 2002). The initiation of α-motoneurones results in an automatic motor response for regulation of quick movements, whereas the initiation of γ-muscle spindle system results in regulation of muscle tone (Sjolander et al., 2002).

The presence of CLBP could affect motor output in any level of the nervous system, which could result in changes in mechanical properties of trunk muscles (Hodges & Tucker, 2011). In people with CLBP for instance, this is associated with increased stiffness of the spine through trunk flexors (i.e., rectus abdominis, internal and external obliques) and trunk extensor (i.e., erector spinae and multifidus) co-activation (van Dieën, Selen, & Cholewicki, 2003). If this trunk stiffness is associated with CLBP and/or disability, it could be considered mal-adaptive; that is, deleterious in the long-term for tissue healing, quality of movement and functional performance (Mok, Brauer, & Hodges, 2011; O'Sullivan, 2005; Radebold, Cholewicki, Panjabi, & Patel, 2000).

2.4. The impact of chronic low back pain on functional status

The main aim of clinical management of lower back disorders is the restoration of the patient’s functional status (Beattie & Maher, 1997). However, before this can be achieved, it is
important to understand the nature of CLBP. Waddell (1987) was among the first authors who described the multidimensional nature of CLBP by proposing three distinct domains (see figure 2.3).

Figure 2.3. Three distinct domains that underlie the nature of CLBP (adapted from Waddell (1987), 2002), Beattie and Maher (1997)). Note: the degrees of overlap are not to scale.

Waddell (1987) explained that although these three domains could be interrelated, the relationship between pain, disability and physical impairments can be disproportionate. The relationship between disability and physical impairments will be discussed in section 2.7. It is common that one element of CLBP could be poorly associated with another. Disability is determined by the inter-relation of self-perceived functional capacity, physical impairments (e.g., lumbar movement or strength impairment), personal beliefs and environmental factors (Jette, 1994; WHO, 2002). This view of disability has often been referred to as the
‘biopsychosocial model’; see figure 2.4). Hence, two individuals may present with similar physical impairments but very different disability levels. Because of this, disability has consistently been recommended as a core outcome domain to be measured in people with CLBP (Chiarotto et al., 2015). Therefore, it is paramount to assess pain, disability and physical impairments in people with LBP using validated outcome measures. It is also important to consider the actual or potential relationships between these domains and the effect of these relationships on the individual.

Figure 2.4. Relational model of disability which incorporates both personal and societal factors from International Classification of Functioning, Disability and Health (WHO, 2002).

2.5. Quantification of the impact of chronic low back pain on the individual

The impact of CLBP is typically assessed using a combination of patient self-reported measures (i.e., questionnaires for pain and disability) and physical performance tests conducted by clinicians (Deyo et al., 2014). Whilst many modes of assessments have been developed to
measure the impact of CLBP, currently there is no single assessment mode which evaluates both patients’ and clinicians’ perspectives (Longo, Loppini, Denaro, Maffulli, & Denaro, 2010). Thus, a comprehensive evaluation of CLBP impact on patients’ life should consist of both patient self-reported measures and physical performance measures (O'Sullivan, 2012).

2.5.1. The measurement of pain intensity in people with chronic low back pain

Pain intensity is the most significant predictor of reoccurring episodes and disability five years later (Campbell, Foster, Thomas, & Dunn, 2013). Therefore, it is important to use valid and reliable measures of pain intensity for individuals with CLBP in primary care (Campbell et al., 2013). The three most common tools to measure self-reported pain intensity are the Visual Analogue Scale (VAS), Verbal Rating Scale (VRS) and the NRS (Bolton & Wilkinson, 1998; Mannion, Balague, Pellise, & Cedraschi, 2007). The VAS consists of a 10-centimeter line, the ends of which are labeled as “0 (no pain)” and “10 (pain as bad as it could be)” with the rest of the line left blank. When descriptive terms are included along the line (e.g., mild (0-4 out of ten), moderate (4.5-7.4 out of ten) and severe (7.5-10 out of ten)), the scale is referred to as a graphical rating scale (Bombardier, 2000; Hawker, Mian, Kendzerska, & French, 2011). Patients are usually instructed to place a mark along the line as a representation of their pain intensity.

The VRS lists adjectives to describe pain intensities such as, but not limited to, none, very mild, mild, moderate, severe and very severe. Finally, the NRS usually involves verbally asking patients to rate their pain intensity by selecting a number between zero (no pain) to ten (pain as bad as it could be).
Out of the three tools, the NRS has been consistently recommended to quantify pain intensity in people with CLBP (Dionne et al., 2008; Mannion et al., 2007; Strong, Ashton, & Chant, 1991). The NRS is more reliable, is easier to understand (Ferraz et al., 1990) and more responsive to change (Bolton & Wilkinson, 1998) than both VAS and VRS in people with CLBP. In addition to being quick and simple to administer, the NRS has been demonstrated to be responsive to change across time in people with CLBP (Chapman et al., 2011).

2.5.2. The measurement of disability in people with chronic low back pain

Over the years, a substantial amount of CLBP research has been performed in an effort to develop outcome measures of disability (Longo et al., 2010). By far the two most commonly utilised disability outcome measures for CLBP are the Roland-Morris Disability Questionnaire (RMDQ) and the ODI (Chapman et al., 2011; Grotle, Brox, & Vollestad, 2005). The psychometric properties the RMDQ and the ODI have been studied extensively, with both outcome measures possessing good test-retest reliability ($\text{ICC}_{\text{RMDQ}} = 0.88-0.71$, $\text{ICC}_{\text{ODI}} = 0.84-0.95$), internal consistency (Cronbach’s $\alpha_{\text{RMDQ}} = 0.91$, Cronbach’s $\alpha_{\text{ODI}} = 0.87$), acceptable content and construct validity ($r_{\text{RMDQ}} = -0.37 - -0.60$, $r_{\text{ODI}} = -0.54 - -0.77$ with Short Form-36) and responsive to change ($D_{\text{RMDQ}} = 0.64$, $D_{\text{ODI}} = 0.76$) (Davidson, 2008, 2009; Davidson & Keating, 2002; Grevitt, Khazim, Webb, Mulholland, & Shepperd, 1997; Grotle, Brox, & Vollestad, 2003; Kopec, 2000; Monticone et al., 2012; Roland & Fairbank, 2000). The domains, administration and psychometric properties of both questionnaires are summarised in table 2.1.
### Table 2.1. Summary of the domains, administration and psychometric properties of the RMDQ and the ODI.

<table>
<thead>
<tr>
<th></th>
<th><strong>RMDQ</strong></th>
<th><strong>ODI</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>(Roland &amp; Morris, 1983)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Domains</strong></td>
<td>24 yes/no statements which encompass walking, bending over, sitting, lying down, dressing, sleeping, self-care and daily activities</td>
<td>10 domains which include pain intensity, personal care, lifting, walking, sitting, standing, sleeping, sex life, social life and travelling</td>
</tr>
<tr>
<td><strong>Scoring</strong></td>
<td>Patients tick the most suitable statements. The score is the total items ticked from a minimum of zero to a maximum of 24 (maximum disability)</td>
<td>Patients select the most appropriate statement (out of six statements of increasing severity level) for each domain. The total is converted to percentage where 100% = maximum disability</td>
</tr>
<tr>
<td><strong>Clinical utility</strong></td>
<td>MCID = 4.87 (Monticone <em>et al.</em>, 2012) No interval level properties, no interpretation of results. Floor effects with low level disability patients (Davidson, 2009).</td>
<td>MCID = 9.50 (Monticone <em>et al.</em>, 2012) Interval level properties (<em>i.e.</em>, each domain is rated from no difficulty to the most challenging), provides score interpretation (<em>i.e.</em>, minimal, moderate, severe disability, crippled and bed bound). Some evidence that the ODI predicts functional task performance (Fisher &amp; Johnson, 1997)</td>
</tr>
</tbody>
</table>

ICC = Intraclass-correlation coefficient, SEM = Standard error of measurement, MICD = Minimally clinically important difference

Although both self-reported disability questionnaires appear to have adequate and comparable psychometric properties, the design of the outcome measures need to be appropriately considered before selecting them for research purposes (Smeets, Koke, Lin, Ferreira, & Demoulin, 2011). The RMDQ was originally designed to be used in research but was later adapted for clinical use (Roland & Morris, 1983). Although psychometrically sound, one of
the biggest shortfalls of the RMDQ is that there is insufficient scope of items representing activities from easy to challenging (Davidson, 2009). Moreover, there are not many items which assess higher difficulty activities (e.g., lifting heavy objects) which makes the RMDQ inadequate for assessing CLBP patients with mild disability level (Davidson, 2009) which are far more prevalent in the CLBP community than those with higher disability levels (Hoy et al., 2014). Moreover, research also identified that the RMDQ included items of poor fit to the definition of disability (e.g., item pertaining to appetite; (Garratt, 2003; Stratford & Binkley, 1997)). Due to these limitations, some authors recommend careful examination of RMDQ score distributions before analysing the results statistically (Smeets et al., 2011). Another limitation of the RMDQ is the lack of score interpretation on the varying degrees of disability (other than 0 = no disability and 24 = maximum disability).

The ODI on the other hand, has been recognised as the most commonly used condition-specific outcome measure to assess the impact of CLBP on disability levels and is frequently used as a comparator when evaluating other CLBP outcome measures in research (Chapman et al., 2011; Clement et al., 2015; Fairbank, Couper, Davies, & O'Brien, 1980; Fairbank & Pynsent, 2000). The ODI has undergone numerous revisions with concomitant validation in the last 30 years (Davidson, 2008; Fairbank, 2014). The ODI is a 10-domain questionnaire that assesses CLBP-related ADL limitations (i.e., pain intensity, personal care, lifting capacity, walking, sitting, standing, sleeping, sexual activity, social life and travelling capacity). The patient then marks the statement which accurately reflects his or her condition (see Appendix 2). One major advantage of the ODI compared to the RMDQ is the provision of interpretations of scores (Fairbank & Pynsent, 2000):
- **0-20% - minimal disability:** the patient can cope with most living activities. No treatment is usually required apart from advice on lifting and sitting activities;

- **21-40% - moderate disability:** the patient experiences difficulties with sitting, lifting and standing. Travel and social life are difficult and may be disabled from work. Personal care and sleeping are not grossly affected. This population can usually be managed by conservative means;

- **41-60% - severe disability:** the patient’s ADL are affected and may require detailed investigation;

- **61-80% - crippled:** CLBP impinges on all aspects of life;

- **81-100% - bed bound:** this patient is bed bound or exaggerating symptoms.

In addition, there is evidence that some domains of the ODI correlate with physical performance; for instance, the ODI lifting domain has been shown to correlate with lifting task performance (*e.g.*, weight lifted and/or time carrying the weight; (Fisher & Johnson, 1997). Additionally, ODI score categories have been demonstrated to correlate well with functional tasks such as bending forward (*r* = 0.65) (Ruiz, Bohl, Webb, Russo, & Grauer, 2014; Sanchez-Zuriaga *et al.*, 2011). However, the same task only modestly correlated with RMDQ scores (*r* = 0.31) (Simmonds *et al.*, 1998). Due to the potential limitations of the RMDQ, and the relationship between the ODI and physical performance that intuitively relates to back pain status, the focus of this thesis and the remainder of this literature review will be on the ODI.
2.5.3. The measurement of physical impairments in people with chronic low back pain

Although self-reported outcome measures provide a valid and reliable summary of the patient’s perspectives of their CLBP status (i.e., pain and disability), they do not provide a demonstrable or objective measure of physical impairment (Waddell, 1987). Therefore, in addition to subjective quantification of patients’ CLBP status, clinicians often include physical assessments to quantify physical impairments related to CLBP (Waddell, Somerville, Henderson, & Newton, 1992). Physical assessment aims to measure the capacity of the neuromuscular system and typically involves measurement of trunk muscle strength, endurance, ROM and velocity under isokinetic, isometric and/or isodynamic conditions (Davarian, Maroufi, Ebrahimi, Farahmand, & Parnianpour, 2012; Marras et al., 1999; Marras et al., 1995; Sanchez-Zuriaga et al., 2011).

2.6. Open kinetic chain neuromuscular assessment in people with chronic low back pain

2.6.1. Trunk muscle strength assessment in people with chronic low back pain

Trunk neuromuscular impairments (e.g., strength, control and activation) can be assessed in isolation to the upper and lower limb during open kinetic chain movements. In open kinetic chain assessment, distal body parts are typically unconstrained allowing isolated assessment of a body segment or muscle (Augustsson & Thomee, 2000). One of the most common methods to assess neuromuscular impairments used in the open kinetic chain in people with CLBP is the assessment of trunk extensor muscle (i.e., lumbar multifidus and erector spinae;
see section 2.3.2) strength with a dynamometer (Steele et al., 2014). A dynamometer refers to a portable or fixed device that provides an objective measurement of muscle force output. Trunk extensor muscle strength is typically quantified as torque during MVIC. However, trunk extensor MVIC measurement is controversial with studies reporting decreased strength (Mooney et al., 1997; Robinson, Cassisi, O'Connor, & MacMillan, 1992) or no difference in trunk extensor strength (Lariviere et al., 2010; Steele, Bruce-Low, Smith, Jessop, & Osborne, 2013) between people with CLBP and healthy people. In addition to the small number of studies directly comparing trunk extensor strength in healthy and CLBP populations, available studies often did not exclude people with previous lumbar surgeries in their CLBP group (Mooney et al., 1997; Robinson, Cassisi, et al., 1992; Steele et al., 2013). Previous surgery has been demonstrated to leave long-term injury and significant atrophy of the erector spinae and lumbar multifidus muscle (Gejo, Matsui, Kawaguchi, Ishihara, & Tsuji, 1999; Pourtaheri et al., 2016) which may confound trunk muscle strength assessment. Moreover, the assessment of MVIC in people with CLBP could be affected by the volition of the participants and fear of pain provocation (Al-Obaidi et al., 2000; Lariviere et al., 2010; Smith, Nelson, Sadoff, & Sadoff, 1989). Due to these limitations, it is currently unclear if people with CLBP possess weaker trunk extensor muscle compared to healthy people (Verbunt et al., 2003; Verbunt, Smeets, & Wittink, 2010).

2.6.2. Trunk muscle force control assessment in people with chronic low back pain

Rather than only assessing trunk extensor muscle strength (e.g., MVIC), some researchers have developed and validated assessments of trunk extensor muscle function during submaximal tasks that focus on the control of force (Larivière et al., 2009; Reeves et al., 2014). The ability
of the neuromuscular system to produce coordinated, accurate and/or smooth (i.e., less variable) force output is defined as muscle force control (Tracy & Enoka, 2002). Muscle force control can be considered as having two main components: (i) force accuracy, and (ii) force steadiness (Hortobagyi, Tunnel, Moody, Beam, & DeVita, 2001; Rice, McNair, Lewis, & Mannion, 2015). The measurement of force accuracy is expressed as muscle force error relative to a submaximal target, usually quantified as the root-mean-square-error between a target force and the participant’s actual force (Rice et al., 2015). On the other hand, the measurement of force steadiness is a measure of force fluctuation within submaximal or maximal force-time traces (Hortobagyi et al., 2001). Hence, an impairment of muscle force steadiness is typically characterised by increased standard deviations of one’s maximal or submaximal muscle force output over time (Enoka et al., 2003).

Previous studies have assessed submaximal muscle force control of the intrinsic hand muscles given the importance of precise finger movement for manual dexterity (Enoka et al., 2003). More recently, studies have assessed muscle force control of the muscles of the neck (Muceli, Farina, Kirkesola, Katch, & Falla, 2011), upper (Bandholm, Rasmussen, Aagaard, Jensen, & Diederichsen, 2006) and lower (Bryant et al., 2009; Rice et al., 2015) limbs in pain conditions (i.e., experimental and clinical pain). All of these studies proposed that the control of submaximal force is affected by pain. Moreover, conditions such as CLBP, which are known to be associated with aberrant changes in trunk neuromuscular adaptation (e.g., trunk muscle co-contraction (van Dieën et al., 2003)), could potentially be associated with impairments in trunk muscle force control.

The evidence pertaining to trunk muscle force control generally is scarce in the literature with only three studies identified (Descarreaux, Blouin, & Teasdale, 2004; Descarreaux, Lalonde,
& Normand, 2007; Miura & Sakuraba, 2014). The studies are summarised in table 2.2. Due to inconsistencies in the definition of trunk muscle force control in the literature, the experimental protocols and analytical methods used in its quantifications are diverse. The results of these studies suggest that people with CLBP display some differences in trunk muscle force control compared to healthy people.

In particular, Miura and Sakuraba (2014) reported that during a submaximal trunk extensor steadiness task, people with CLBP were 24%-28% less steady than healthy people. In this study, steadiness was quantified as the coefficient of variation (i.e., standard deviation of force over mean force) between the target force and the participant’s force (i.e., the higher coefficient of variation = less steady). In the study series by Descarreaux et al. (2004; 2007), trunk muscle extensor force control impairment was indicated by a 29%-42% delay in producing trunk extensor force output compared to healthy people during submaximal force accuracy tasks. In these studies, participants were instructed to produce a single force impulse (rather than sustaining force output) at 50% and 75% MVIC.

In the three relevant studies (Descarreaux et al., 2004; Descarreaux et al., 2007; Miura & Sakuraba, 2014), varying assessment protocols were identified. However, these trunk muscle force control protocols (i.e., trunk extension steadiness (Miura & Sakuraba, 2014) and trunk extension force accuracy tasks (Descarreaux et al., 2004; Descarreaux et al., 2007)) incorporated submaximal and non-variable force targets. It is important to note that many daily activities typically require the trunk motor control system to control submaximal and dynamic (i.e., variable) and static (i.e., non-variable) forces (Marras, 2008). In this respect, it is important to assess trunk muscle force control using a non-variable force target.
Table 2.2. Summary of trunk muscle force control studies in healthy and CLBP individuals

<table>
<thead>
<tr>
<th>Study</th>
<th>Participants</th>
<th>Methods</th>
<th>Outcome measures</th>
<th>Results</th>
<th>Group difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Healthy (n)</td>
<td>CLBP (n)</td>
<td>Assessment</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Feedback</td>
<td>None</td>
<td>Absolute error (Nm)</td>
<td>14.1</td>
<td>11.9</td>
</tr>
<tr>
<td></td>
<td>Pain (VAS)</td>
<td>4.3 ± 9.0</td>
<td>Position</td>
<td>Neutral standing, pelvis stabilised</td>
<td>Peak force variability (Nm)</td>
</tr>
<tr>
<td></td>
<td>Disability (ODI)</td>
<td>26.0% ± 9.3%</td>
<td>Protocol</td>
<td>Neutral standing, pelvis stabilised</td>
<td>Time to reach peak force (ms)</td>
</tr>
<tr>
<td>Descarreaux, <em>et al.</em>, 2004</td>
<td>15</td>
<td>7</td>
<td>Feedback</td>
<td>None</td>
<td>Absolute error (Nm)</td>
</tr>
<tr>
<td></td>
<td>Pain (VAS)</td>
<td>2.0 ± 1.4</td>
<td>Position</td>
<td>Neutral standing, pelvis stabilised</td>
<td>Peak force variability (Nm)</td>
</tr>
<tr>
<td></td>
<td>Disability (ODI)</td>
<td>22.6% ± 11.3%</td>
<td>Protocol</td>
<td>Neutral standing, pelvis stabilised</td>
<td>Time to reach peak force (ms)</td>
</tr>
<tr>
<td>Miura &amp; Sakuraba, 2014</td>
<td>14</td>
<td>14</td>
<td>Feedback</td>
<td>Visual</td>
<td>Absolute error at 30% MVIC (%COV)</td>
</tr>
<tr>
<td></td>
<td>Pain (VAS)</td>
<td>≥3</td>
<td>Position</td>
<td>Prone, pelvis stabilised</td>
<td>Absolute error at 50% MVIC (%COV)</td>
</tr>
<tr>
<td></td>
<td>Disability</td>
<td>NR</td>
<td>Protocol</td>
<td>Trunk extension steadiness Static</td>
<td></td>
</tr>
</tbody>
</table>

VAS = Visual Analogue Scale, ODI = Oswestry Disability Index, MVIC = Maximum voluntary isometric contraction, COV = Coefficient of variation, Nm = Newton.meter, ms = milliseconds, NS = Not significant, NR = Not reported. *Dynamic/static defined as possessing an intended variable component to the force (e.g., matching a moving/variable force target) for dynamic and an intended constant force target for static.
2.6.3. Clinical implications of trunk muscle force control assessment

Altered functional performance (e.g., lifting) in people with CLBP is generally associated with increased co-contraction of trunk agonistic and antagonistic muscles compared to individuals without CLBP (Reeves, Cholewicki, Milner, & Lee, 2008; Reeves, Everding, Cholewicki, & Morrisette, 2006). Increased trunk muscle co-contraction may have a protective role after an acute back injury; however, in those with CLBP increased co-contraction may potentially be maladaptive by reducing quality of movement and increasing compressive load on the spine (Marras, Davis, Ferguson, Lucas, & Gupta, 2001). Moreover, because some movement is required to evenly distribute loads on bony and soft tissues, increased trunk stiffness could be detrimental to overall tissue health (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Panjabi, 1992a, 2006). As poor neuromuscular strategies continue to persist in CLBP patients even during symptom remission (MacDonald, Moseley, & Hodges, 2009; Macdonald, Dawson, & Hodges, 2011), an assessment of trunk muscle force control is likely to be relevant in those with CLBP. In the short term, inaccurate force production, combined with altered muscle strength and activation, may result in soft tissue injury (i.e., ligamentous or muscular) during heavy lifting (Cholewicki & McGill, 1992). In a knee model, Herzog et al. (2003) demonstrated that inaccurate knee extensor force output contributed to changes in internal knee joint forces that manifested in reddening and fibrillation of articular surfaces (i.e., the first sign of joint degeneration). Thus, in the longer term, altered motor output may be associated with degenerative joint changes due to abnormal loading of chondral surfaces (Herzog et al., 2003).

Although data pertaining to trunk muscle force control in CLBP patients could be clinically relevant, current research has not identified any association between trunk muscle force control and clinical outcome measurements such as pain or disability level. At the knee, impaired hamstring
muscle force control following anterior cruciate ligament reconstruction has been associated with decreased functional performance such as hop length (Bryant, Clark, & Pua, 2011). Similarly, impaired muscle force control in the finger flexors has been associated with impaired precision grip performance in older adults (Cole, 1991). Thus, impairment in trunk muscle force control may be associated with functional impairment or self-reported disability in people with CLBP. This notion will be explored in Chapter 6.

2.7. Closed kinetic chain neuromuscular assessment in people with chronic low back pain

Physical assessments can also incorporate tasks of great functional relevance involving closed kinetic chain manoeuvres. During closed kinetic chain assessment, the lower limbs are usually constrained because the feet are in contact with the ground, allowing movements at the distal and proximal joints to occur simultaneously (Augustsson & Thomee, 2000). Therefore, closed kinetic chain spinal neuromuscular assessments tend to involve functional tasks such as walking (Crosbie, de Faria Negrão Filho, Nascimento, & Ferreira, 2013), sit-to-stand (Shum, Crosbie, & Lee, 2005), bending down (Williams, Haq, & Lee, 2012) or lifting objects from the ground (Sanchez-Zuriaga et al., 2011; Shum, Crosbie, & Lee, 2007a). Lifting in particular, is a highly complex task that has been linked to high work-related LBP rates resulting in significant medical cost and productivity loss (Hanney, Kolber, & Beekhuizen, 2009). More recently, lifting at work has been proposed to be a primary risk factor for developing future LBP (Coenen et al., 2014; Oliveira et al., 2015). Clinically, physical assessment involving lifting has been shown to be responsive to changes in CLBP symptoms (Strand et al., 2011) and improvement in lifting kinematic parameters (e.g., lumbar ROM and velocity) are proposed to be a valid criteria for CLBP recovery (Ferguson &
Marras, 2013). In light of these considerations, biomechanical assessment of lifting tasks in CLBP and healthy people will be the focus of the subsequent section.

The kinematics of symmetrical lifting (i.e., lifting where the load is placed anteriorly about the body’s mid-sagittal plane with both hands used to grip the load (Lavender, Andersson, Schipplein, & Fuentes, 2003)) is typically divided into two phases: flexion (i.e., bending) phase and extension (i.e., lifting) phase (Waters, Putz-Anderson, Garg, & Fine, 1993). The flexion phase begins at the onset of trunk movement and ends when the peak angle of lumbar flexion has been reached (Sanchez-Zuriaga et al., 2011) or when the hands have reached the object being lifted (Burgess-Limerick et al., 1995) (see figure 2.5A). This is followed by the extension phase which is initiated when the load is lifted from the ground and ends when the lumbar spine or trunk returns to its initial upright position (Bouilland, Loslever, & Lepoutre, 2002) (figure 2.5B). For brevity, the extension phase of lifting will be simply referred to as lifting from this point onwards.

Figure 2.5. Lifting phases. Flexion phase ends when the hands reach the load and the trunk is fully flexed. This marks the beginning of the extension phase (A). The extension (i.e., lifting) phase ends when the trunk is upright (B).
The lifting phase is of clinical relevance as it is associated with three types of mechanical loads or forces imposed on the spine: (i) compression forces through back muscle activity which could result in vertebral endplate fractures, (ii) anterior shear forces as a result of gravity acting on the trunk and muscle activity which could damage the neural arch, and (iii) trunk flexion forces which could impose stresses on posterior ligamentous and disc structures (van Dieen, Hoozemans, & Toussaint, 1999). Therefore, research over the years has been focused on assessing the lifting technique that imposes the least amount of load on the spine by controlling multiple lifting variables (Faber, Kingma, & van Dieen, 2011; Greenland, Merryweather, & Bloswick, 2013; Hoozemans, Kingma, de Vries, & van Dieen, 2008; Kingma, Bosch, Bruins, & van Dieen, 2004; Lavender et al., 2003; Lavender, Li, Andersson, & Natarajan, 1999). These lifting variables and their impact on spinal loads are summarised in figure 2.6.

### Spinal Load

<table>
<thead>
<tr>
<th>Low</th>
<th>High</th>
</tr>
</thead>
<tbody>
<tr>
<td>• Light mass lifted</td>
<td>• Heavy mass lifted</td>
</tr>
<tr>
<td>• Mass lifted close to body</td>
<td>• Mass lifted far from body</td>
</tr>
<tr>
<td>• High lifting height</td>
<td>• Low lifting height</td>
</tr>
<tr>
<td>• Normal lifting speed</td>
<td>• Fast or slow lifting speed</td>
</tr>
<tr>
<td>• No trunk rotation when lifting</td>
<td>• Lifting with trunk rotation</td>
</tr>
</tbody>
</table>

**Figure 2.6.** Summary of lifting variables and their association with spinal load.

### 2.7.1. Lifting biomechanics assessment in people with chronic low back pain

Previous biomechanical assessments of symmetrical lifting has incorporated three-dimensional (3D) movement analysis involving measurements of kinematic parameters such as ROM and angular velocity of the trunk and lower limb joints in the sagittal plane and the measurement of vertical GRF’s exerted by the lower limbs (Kingma et al., 2004; Marras et al., 1993; Sanchez-
Previous research has focused mainly on sagittal plane variables because they reflect the interactions across the lower limb joints, pelvis and spinal joints (e.g., Granata and Sanford (2000) and Lariviere, Gagnon, and Loisel (2000)). Additionally, the natural angle of inclination of the lumbar zygapophyseal joints, which are oriented in the sagittal plane, promote the movements of flexion and extension (see section 2.3.1). Studies comparing lifting kinematics parameters in the sagittal plane in people with CLBP and healthy controls is summarised in table 2.3.
Table 2.3. Summary of studies comparing lifting kinematics of people with and without CLBP.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Participants</th>
<th>Method</th>
<th>Joints</th>
<th>Healthy ROM (°)</th>
<th>Healthy Vel (°/s)</th>
<th>CLBP ROM (°)</th>
<th>CLBP Vel (°/s)</th>
<th>Between group difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sanchez-Zuriaga et al., 2011</td>
<td>16健康的(16)</td>
<td>CLBP 39</td>
<td>3D Thorax motion analysis</td>
<td>67.5 ± 19.2</td>
<td>68.5 ± 7.8</td>
<td>43.8 ± 18.8</td>
<td>53.4 ± 10.2</td>
<td>35</td>
</tr>
<tr>
<td></td>
<td>Pain (ODI) 33 %</td>
<td>13.2%</td>
<td>Lumbar</td>
<td>Healthy ROM (°)</td>
<td>Healthy Vel (°/s)</td>
<td>CLBP ROM (°)</td>
<td>CLBP Vel (°/s)</td>
<td>Between group difference (%)</td>
</tr>
<tr>
<td>Shum et al., 2007a</td>
<td>20疼痛(20)</td>
<td>CLBP 30</td>
<td>Accelero Hip motion analysis</td>
<td>48.0 ± 13.0</td>
<td>30.0 ± 11.0</td>
<td>34.0 ± 10.0</td>
<td>17.0 ± 8.5</td>
<td>29</td>
</tr>
<tr>
<td></td>
<td>Disability (VAS) 5.9 ± 1.9</td>
<td>10.6 ± 4.65</td>
<td>Lumbar</td>
<td>Healthy ROM (°)</td>
<td>Healthy Vel (°/s)</td>
<td>CLBP ROM (°)</td>
<td>CLBP Vel (°/s)</td>
<td>Between group difference (%)</td>
</tr>
<tr>
<td>Lariviere et al., 2002</td>
<td>18疼痛(18)</td>
<td>CLBP 15</td>
<td>2D Vertebral motion analysis</td>
<td>44.0 ± 7.0</td>
<td>86.5 ± 24.6</td>
<td>42.0 ± 7.0</td>
<td>99.7 ± 39.0</td>
<td>NS</td>
</tr>
<tr>
<td></td>
<td>Disability 2.6 ± 2.5</td>
<td></td>
<td>Hip</td>
<td>Healthy ROM (°)</td>
<td>Healthy Vel (°/s)</td>
<td>CLBP ROM (°)</td>
<td>CLBP Vel (°/s)</td>
<td>Between group difference (%)</td>
</tr>
<tr>
<td>McGregor et al., 1997</td>
<td>201疼痛(201)</td>
<td>CLBP 138</td>
<td>Triaxial Lumbar motion potentiometer</td>
<td>32.3 ± 13.7</td>
<td>NR</td>
<td>43.2 ± 18.0</td>
<td>NR</td>
<td>25</td>
</tr>
</tbody>
</table>

ROM = Range of motion, Vel = Angular velocity, NS = not significant, NR = Not reported, VAS = Visual Analogue Scale, ODI = Oswestry Disability Index, RMDQ = Roland-Morris Disability Questionnaire. *3D and 2D motion analyses were conducted using optoelectric motion capture system.
Surprisingly, only four studies directly comparing lifting biomechanics in people with CLBP and healthy people were identified. The results of these studies however were not uniform. In particular, Sanchez-Zuriaga et al. (2011) and Shum et al. (2007a) reported that people with CLBP lifted with less lumbar ROM compared to healthy controls (mean difference = 29%-34%). In contrast McGregor, McCarthy, and Hughes (1997) and Shum et al. (2007a) reported increased lumbar ROM (mean difference = 25%) in their CLBP group compared to healthy group during lifting, while Lariviere et al. (2002) found no significant differences between groups. Similarly, lumbar angular velocity study results during lifting are equivocal. Although between-study lifting kinematic inconsistencies could be attributed to different data collection methods, only one study (Sanchez-Zuriaga et al., 2011) provided validation of the kinematic protocol against the ODI in people with CLBP. Moreover, only Sanchez-Zuriaga et al. (2011) reported the reliability of their lifting protocol (inter-session ICC = 0.91, SEM = 2.52). Thus, Sanchez-Zuriaga et al. (2011) lifting protocol was utilised in Study 3.

Lifting is a biomechanically complex task which requires an interaction of multiple joints within the trunk and lower limb (Burgess-Limerick & Abernethy, 1997). Most studies of trunk movements have only focused on measurement of the lumbar spine relative to the hip movement (see table 2.3); however, only one study (Sanchez-Zuriaga et al., 2011) has acknowledged the contribution of the thoracic spine to the overall trunk ROM during lifting. Thoracic spine mobility determines the amount of knee flexion during lifting; that is, less thoracic flexion results in larger knee flexion during lifting which, in turn, results in lower spinal load (List, Gulay, Stoop, & Lorenzetti, 2013). Impairment in thoracic spine kinematics has recently been shown to affect functional task performance which requires trunk movements in people with CLBP (Christe, Redhead, Legrand, Jolles, & Favre, 2016; Crosbie, Nascimento, Filho Rde, & Ferreira, 2013). In
particular, when compared to healthy controls, people with CLBP demonstrated restricted overall thoracic ROM which was accompanied by increased hip ROM when tasked to reach down to put on a sock (Crosbie, Nascimento, et al., 2013). Similarly, during sit-to-stand-to-sit, people with CLBP demonstrated restriction of both thoracic and lumbar ROM compared to healthy controls (Christe et al., 2016). Collectively, these findings suggest that CLBP affects both the thoracic and lumbar spine which could result in compensatory strategies in the lower limb during functional task performance. However, the relationship between the thoracic spine, lumbar spine, hip and knee during lifting has not been explored in the literature.

It is also worth noting that the measurement of absolute joint ROM during a lifting task is problematic because trunk ROM has been demonstrated to be dependent on lifting variables such as the mass lifted and the initial position of the mass (see figure 2.6) (Burgess-Limerick & Abernethy, 1997; Faber et al., 2011). Thus, the differences in trunk ROM reported in each study may result from different lifting protocols rather than true trunk ROM. Therefore, other kinematics parameters are required to quantify lifting technique in both CLBP and healthy individuals.

### 2.7.2. Lifting inter-joint coordination in people with chronic low back pain

The kinematics of lifting can also be quantified using inter-joint coordination techniques. Trunk extension movement is accomplished by movements in the trunk (i.e., thorax and lumbar spine) and lower limb (e.g., hip) joints (Nelson, Walmsley, & Stevenson, 1995). Multiple studies have attempted to quantify and compare trunk and lower limb inter-joint coordination in people with CLBP and healthy people during a trunk extension task. The summary of these studies is outlined in table 2.4. Overall, the quantification of trunk-lower limb inter-joint coordination in the literature could be divided into two main methods: (i) discrete point measurement and (ii) continuous measurement of inter-joint coordination.
<table>
<thead>
<tr>
<th>Studies</th>
<th>Participants</th>
<th>Analysis type</th>
<th>Outcome measure</th>
<th>Results</th>
<th>Group difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Healthy (n)</td>
<td>CLBP (n)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Moktharinia et al., 2016</td>
<td>22</td>
<td>22</td>
<td>Relative phase angle analysis</td>
<td>Lumbar MARP (°)</td>
<td>18.70 ± 6.20  16.21 ± 7.23</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Hip MARP (°)</td>
<td>5.09 ± 3.27       5.16 ± 3.08</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>13</td>
</tr>
<tr>
<td>Kim et al., 2013</td>
<td>16</td>
<td>17</td>
<td>Lumbar-hip ROM ratio</td>
<td>Lumbar-hip ratio at 25% extension</td>
<td>0.6 ± 0.4      1.2 ± 0.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 50% extension</td>
<td>1.0 ± 0.5      1.5 ± 0.7</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 75% extension</td>
<td>1.2 ± 0.2      1.4 ± 0.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 100% extension</td>
<td>1.0 ± 0.4      1.6 ± 0.7</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>50</td>
</tr>
<tr>
<td>Shum et al., 2007a</td>
<td>20</td>
<td>30</td>
<td>Cross-correlation ratio</td>
<td>Lumbar-hip ROM ratio</td>
<td>0.53 ± 0.13    0.46 ± 0.18</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip cross-correlation ratio</td>
<td>0.99 ± 0.01    0.98 ± 0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>NS</td>
</tr>
<tr>
<td>Wong &amp; Lee, 2004</td>
<td>20</td>
<td>24</td>
<td>Cross-correlation ratio</td>
<td>Lumbar-hip ratio</td>
<td>1.36 ± 1.45    1.43 ± 1.52</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip cross-correlation ratio</td>
<td>0.96 ± 0.05    0.93 ± 0.10</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip mean time lag (s)</td>
<td>-0.02 ± 0.07   -0.05 ± 0.08</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>NS</td>
</tr>
<tr>
<td>Lariviere et al., 2002</td>
<td>18</td>
<td>15</td>
<td>Postural Index *</td>
<td>Postural Index</td>
<td>0.82 ± 0.47    0.71 ± 0.32</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>NS</td>
</tr>
</tbody>
</table>

Table 2.4. Summary of trunk-lower limb inter-joint coordination studies during trunk extension task in healthy and CLBP individuals.
<table>
<thead>
<tr>
<th>Studies</th>
<th>Participants</th>
<th>Analysis and task type</th>
<th>Outcome measure</th>
<th>Results</th>
<th>Group difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Healthy (n)</td>
<td>CLBP (n)</td>
<td></td>
<td>Healthy</td>
<td>CLBP</td>
</tr>
<tr>
<td>Lariviere et al., 2000</td>
<td>18</td>
<td>15</td>
<td>Lumbar-hip ROM ratio</td>
<td>NR</td>
<td>NR</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio</td>
<td>NR</td>
<td>NR</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Pain</td>
<td>NR</td>
<td>NR</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Disability (QBPDS)</td>
<td>12.7 ± 2.6</td>
<td>NR</td>
</tr>
<tr>
<td>McClure et al., 1997</td>
<td>12</td>
<td>12</td>
<td>Lumbar-hip ROM ratio</td>
<td>Lumbar-hip ratio at 25% extension</td>
<td>0.16 ± 0.14</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 50% extension</td>
<td>0.64 ± 0.24</td>
<td>0.59 ± 0.35</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 75% extension</td>
<td>0.94 ± 0.39</td>
<td>0.69 ± 0.35</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lumbar-hip ratio at 100% extension</td>
<td>3.04 ± 4.10</td>
<td>1.82 ± 0.90</td>
</tr>
<tr>
<td>Pacquet et al., 1994</td>
<td>10</td>
<td>10</td>
<td>Lumbar-hip ROM ratio</td>
<td>Qualitative description</td>
<td>NR</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Pain (PPI)</td>
<td>≤3/5</td>
<td>NR</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Disability</td>
<td>NR</td>
<td>NR</td>
</tr>
</tbody>
</table>

NR= Not Reported, NS = Not Significant, MARP = Mean Absolute Relative Phase, VAS = Visual Analogue Scale, NS = Not Significant, ROM = Range of motion, RMDQ = Roland-Morris Disability Questionnaire, ODI = Oswestry Disability Index, QBPDS = Quebec Back Pain Disability Scale, PPI = Present Pain Index. * Postural Index = ratio between knee ROM and the sum of lumbar, hip and ankle ROM
Discrete point measurement of inter-joint coordination

The discrete point measurement method quantifies inter-joint coordination by calculating the ratio of proximal to distal joint ROM (e.g., lumbar ROM to hip ROM ratio) at one, or several, measurement points throughout the trunk extension movement (Kim et al., 2013; Lariviere et al., 2000, 2002; McClure, Esola, Schreier, & Siegler, 1997). For example, during the first 25% of trunk extension from full trunk flexion, healthy people demonstrated a lumbar-hip ROM ratio of 0.16 (McClure et al., 1997) which suggests that the hip joint contributes more extension ROM than the lumbar spine at this measurement point. The results of the studies comparing inter-joint coordination using discrete point measurements are conflicting; with studies reporting an increase (Kim, Yi, & Cynn, 2015; McClure et al., 1997) or no difference (Lariviere et al., 2000; Paquet, Malouin, & Richards, 1994) in lumbar-hip ROM ratio during trunk extension in people with CLBP compared to healthy people. Lariviere et al. (2002) compared the inter-joint coordination between CLBP and healthy groups using the postural index. The postural index is defined as the ratio of the knee ROM to the sum of thoracic, lumbar and ankle ROM during lifting; a high index suggests that one lifts with the knees flexed whereas a low index suggests one lifts with the knees extended (i.e., flexed trunk) (Burgess-Limerick & Abernethy, 1997). Lifting with the knees flexed has been associated with decreased trunk flexion which resulted in lower L5/S1 compressive forces (Kingma, Faber, & van Dieen, 2010). Lariviere et al. (2002) however, demonstrated no significant differences in postural index in both CLBP and healthy individuals. Additionally, only Lariviere et al. (Lariviere et al., 2002) study assessed the thoracic spine and the knee during trunk extension tasks. In relevance to lifting (see section 2.6.1), the involvement of the trunk and lower limb joints are important in the modulation of spinal load (Kingma et al., 2010).

As discrete assessment method only considers joint ratio at single occurrences during trunk extension, this assessment method could only assume that the movement of the trunk and lower
limb joints occurs simultaneously (i.e., perfectly synchronous) throughout trunk extension movement (Potvin, McGill, & Norman, 1991). However, deviation from perfectly synchronous movement between the trunk and lower limb during lifting has been observed when movement coordination of the trunk and lower limb was assessed continuously (Burgess-Limerick et al., 1993). With respect to lifting, unlike discrete assessment variables (e.g., lumbar-hip ROM ratio), continuous assessment variables (e.g., those derived from relative phase angle analysis; see section below) have been demonstrated to be responsive to changes in load and speed of movement in both CLBP and healthy people (Burgess-Limerick et al., 1995; Mokhtarinia, Sanjari, Chehrehrazi, Kahrizi, & Parnianpour, 2016) – which makes continuous assessment of joint coordination appropriate to assess the lifting technique in these groups.

**Continuous measurement of inter-joint coordination**

An alternative method to quantify inter-joint coordination is to continuously measure the relationship between proximal-distal joint couples throughout trunk extension movement (Nelson et al., 1995). Traditionally, this can be achieved by plotting the angular position of the proximal joint (e.g., lumbar) as a function of the distal joint (e.g., hip) (Nelson et al., 1995). Inter-joint coordination has also been quantified using cross-correlation coefficient analysis (Shum et al., 2007a; Wong & Lee, 2004). When continuously measuring movement-related inter-joint coordination, joint couples can move in-phase (e.g., trunk and hip move in the same direction) or out-of-phase (e.g., trunk and hip move in the opposite direction) (Burgess-Limerick et al., 1993). Imprecise timing (i.e., increased or decreased in-phase movement time) results in disorganised limb movements (Sainburg, Poizner, & Ghez, 1993). The cross-correlation coefficient provides two indicators relative to coordination: how similar the underlying pattern of the two signals are (the correlation score) and how synchronous the patterns are in time (the phase shift in the peak correlation score). The higher the coefficient, the more similar the patterns in the two joints are. While this similarity in the pattern may be
important, of potentially greater interest is how synchronous the two joint movements are which can be determined by assessing when the peak cross correlation value is obtained. If the peak occurs at a time point greater than zero this indicates a positive phase lag, and suggests that the reference joint (e.g., lumbar spine) moves earlier than the joint of interest (e.g., hip). The opposite occurs when the peak cross correlation occurs at a negative time point (negative phase lag), and a lag phase of zero indicates that the joint couple movement is perfectly in-phase (Shum et al., 2007a; Wong & Lee, 2004). However, studies utilising this method did not find any significant difference between the inter-joint interaction of the trunk and the hip in both healthy and CLBP groups during trunk extension tasks (Shum et al., 2007a; Wong & Lee, 2004). Whilst this method may be useful in determining the time delay between adjacent joint movements, cross-correlation analysis assumes that proximal-distal movement pattern is constant throughout movement. However, in the presence of pain, joint movements may not be uniform throughout task performance (Hodges & Smeets, 2015). Thus, the assumption of constant delay of a joint movement relative to another throughout performance of a task may not be an accurate representation of inter-joint movement relationships.

Finally, inter-joint coordination can also be assessed using relative phase angle analysis (Burgess-Limerick et al., 1993; Mokhtarina et al., 2016). Relative phase angle analysis will be explained in detail in section 5.4.4. In short, the relative phase angle analysis aims to quantify in-phase (i.e., synchronous) movement between the distal and proximal joints (Burgess-Limerick et al., 1993). In the analysis by Mokhtarina et al. (2016), the value closer to 0° indicates more in-phase movements and the value closer to 180° indicates more out of phase movements between the proximal (lumbar) and distal (hip) joints. Mokhtarina et al. (2016) demonstrated that people with CLBP move more synchronously through the lumbar spine and hip during trunk extension. Although interesting, the study findings may not be generalisable
to the larger CLBP sufferers due to their small sample size and their pain free CLBP participants.

2.7.3. Kinetic assessments in people with chronic low back pain

Measurement of GRF in people with CLBP traditionally involves assessing the ability to control vertical force exerted by lower limbs on a force platform (Byl & Sinnott, 1991). Most studies of kinetics in people with CLBP involves measurements of centre of pressure excursions during non-functional tasks such as during static bipedal standing with rapid arm movements (Frost & Brown, 2016), static single legged standing (Sung & Leininger, 2015) or unstable sitting (van Daele et al., 2009). These studies collectively suggest that people with CLBP demonstrated impairment in the ability to control vertical GRF (i.e., centre of pressure excursion) during standing compared to healthy controls (Ruhe, Fejer, & Walker, 2011). Only a handful of studies have measured kinetic variables during functional activities such as sit-to-stand or lifting (Sanchez-Zuriaga et al., 2011; Shum, Crosbie, & Lee, 2007b). Similarly, these limited data suggest that people with CLBP demonstrate decreased GRF through the lower limbs during daily activities (Sanchez-Zuriaga et al., 2011; Shum et al., 2007b). Although GRF has been studied in relation to trunk and lower limb ROM during functional task, GRF measures have not been studied in conjunction to inter-joint coordination parameters during lifting in people with CLBP. Having this information could provide information on CLBP related compensation mechanisms during lifting.
2.7.4. Clinical implications of inter-joint coordination and kinetic assessments in people with chronic low back pain

The collective aim of previous studies that have incorporated inter-joint coordination and kinetic assessments was to identify aberrant movement and kinetic patterns in people with CLBP compared to healthy people during trunk extension tasks. Although these studies are heterogenous in terms of their methodology, there are indications that compared to healthy people, people with CLBP demonstrate more synchronous lumbar-hip movement and decreased vertical GRF during functional tasks. Past research suggests that an in-phase pattern or phase-locked coordination pattern may be related to increased trunk muscle co-contraction (van Dieen et al., 2003) and may be associated with increased spinal loading (Tafazzol, Arjmand, Shirazi-Adl, & Parnianpour, 2014). Additionally, people with CLBP demonstrated decreased GRF’s during functional tasks such as lifting (Sanchez-Zuriaga et al., 2011). Impaired load sharing strategies (i.e., decreased GRF) has been associated with more load imposed on the trunk (Shum et al., 2007b). Unfortunately, none of the studies investigated the inter-joint coordination pattern in conjunction with GRF data during lifting. Having this information may provide better understanding of mal-adaptive biomechanical adaptations during lifting (O'Sullivan, 2005). Impairment in lifting strategies (i.e., maladaptive biomechanical adaptation) could result in abnormal tissue loading which in turn could perpetuate pain in people with CLBP (Cholewicki & McGill, 1992). None of the studies investigated the relationship between self-reported disability in people with CLBP and inter-joint coordination and vertical GRF parameters. Thus, it is currently unclear whether CLBP individuals with different disability levels would demonstrate different kinematic and kinetic pattern during a lifting task.
2.8. The relationship between self-reported disability and open and closed kinetic chain neuromuscular assessment in people with chronic low back pain

Open and closed kinetic chain neuromuscular impairments and self-reported disability level are inter-related domains of CLBP (Dubois, Abboud, St-Pierre, Piche, & Descarreaux, 2014; van Rooij et al., 2015). It is critical to understand the relationship between disability level and neuromuscular impairments because neuromuscular impairments are modifiable (Crow et al., 2011; Laird et al., 2012) and are often targeted in the management of CLBP to decrease disability (Grotle et al., 2005; Pengel et al., 2004). However, changes in open and closed neuromuscular adaptations following CLBP interventions have been shown to be associated with only a modest decrease in self-reported disability at short term (Ferreira, Ferreira, et al., 2010) and long term (Van Dillen et al., 2016) follow-up.

This may be due to evidence that the relationship between traditional open (e.g., trunk muscle strength) and closed kinetic chain neuromuscular measures (e.g., trunk ROM) and disability level is highly variable; ranging from non-existent (Al-Obaidi et al., 2000; Nattrass, Nitschke, Disler, Chou, & Ooi, 1999; Parks et al., 2003) to moderate ($r = 0.37-0.50$) (Dubois et al., 2014; Mannion, Taimela, Müntener, & Dvorak, 2001; Ruiz et al., 2014). Arguably, traditional neuromuscular impairment assessments such as trunk muscle strength and trunk ROM assessments may not assess the appropriate aspects of functional disability in people with CLBP (Caporaso, Pulkovski, Sprott, & Mannion, 2012; Deyo, 1988).

Moreover, the relationship between traditional physical impairment measures and self-reported disability was often assessed separately using bivariate correlations (Pfingsten, Lueder, Luedtke, Petzke, & Hildebrandt, 2014). When open and closed kinetic chain neuromuscular variables were assessed together using multivariate linear regression, they explained more variance in self-reported disability (Mannion, Junge, et al., 2001). In a large study ($n = 148$),
Mannion et al. (2001) utilised a combination of seven open and closed kinetic chain assessments to predict self-reported disability which included a combination of electromyographic parameters (trunk muscle activation, trunk muscle initial median frequency, isometric fatigue, dynamic fatigue and flexion-relaxation phenomenon), trunk muscle strength and trunk ROM. In combination, these open and closed kinetic chain variables explained 25% of the variance in self-reported disability. However, this battery of tests are of limited clinical applicability as they are time consuming and have to be conducted in a laboratory setting. In contrast, the measurement of trunk muscle force control (section 2.5.1) and inter-joint lifting coordination (section 2.6.2) are novel and have not been performed in a clinical setting. It is unknown whether the novel tests of trunk muscle force control and inter-joint lifting coordination are more strongly associated with self-reported disability in people with CLBP.

2.9. Conclusion

This review of the literature highlights the relationship between three distinct outcome domains in people with CLBP: pain, physical impairments and disability. Pain and disability are individual-specific domains that can be measured using patient-reported questionnaires. Physical impairments are numerous in individuals with CLBP and this literature review focussed on impairments which were most likely to be associated with clinically-important differences in disability. Assessments for physical impairments were categorised as (i) isolated open kinetic chain tasks such as the assessment of trunk muscle strength or trunk muscle force control, or (ii) complex multi-arthrodial closed kinetic chain functional assessment tasks such as lifting. Lifting technique can be quantified using trunk and lower limb kinematics (ROM, angular velocity, inter-joint coordination) and kinetics (vertical GRF).

The relationship between traditional neuromuscular assessment such as trunk muscle strength and trunk ROM and self-reported disability ranges from non-existent to moderate. The
relationship between the novel tests of trunk muscle force control and inter-joint lifting coordination and self-reported disability is still unknown, but there is potential for more sophisticated measures of trunk neuromuscular control to explain a greater amount of the variance in disability in those with CLBP than more simple measures of strength and ROM. Therefore, the ultimate aim of this thesis is to investigate the relationship between self-reported disability and novel open kinetic chain assessment of trunk muscle force control and closed kinetic chain assessment of inter-joint lifting coordination in people with CLBP.
Chapter 3

Study 1

Participant characteristics and self-reported disability in people with chronic low back pain

3.1. Introduction

In people with CLBP, the assessment of disability and pain is of paramount importance as they are the main targets of clinical management (Beattie & Maher, 1997; Smeets et al., 2011). Self-reported disability is commonly assessed using validated questionnaires such as the ODI (see section 2.4.2) whereas pain is typically measured using validated questionnaires such as the NRS (see section 2.4.1). Description of study population with respect to disability and pain is of importance to ensure accurate implementation of research results to the appropriate CLBP population (Eldredge, Weagel, & Kroth, 2014).

However, only some open and closed kinetic neuromuscular assessment studies (see sections 2.6 and 2.7) described their CLBP groups with respect to disability and pain. These studies reported ODI score ranging from 22.6% to 26.8% – indicating moderate disability (Fairbank & Pynsent, 2000) and NRS/VAS score between 2.0 to 6.0 – indicating mild to moderate pain (Hawker et al., 2011). To frame the results of the studies within this thesis in the context of prior research, and to allow for an informed assessment of the generalisability of the results in this thesis to a general CLBP population, it is important to outline the characteristics of the participants.
3.2. Aim

Thus, based on the abovementioned rationale, this study aims to:

1. Describe and compare both CLBP and healthy participant characteristics in Studies 2 (force control) and 3 (lifting inter-joint coordination) of this thesis with respect to self-reported disability.

2. Describe and compare both CLBP and healthy participant characteristics in Studies 2 (force control) and 3 (lifting inter-joint coordination) of this thesis with respect to pain.

3.3. Hypotheses

Based on the previous equivocal literature (see sections 2.6 and 2.7), the following hypotheses were adopted for the present study:

1. CLBP participants would be more disabled as quantified by the ODI than healthy participants.

2. CLBP participants would report more pain as quantified by the NRS than healthy participants.

3.4. Materials and methods

3.4.1. Participants

Forty-three participants (n=23) aged 25-60 years old with CLBP were recruited from two large Kieser physiotherapy clinics located in South Melbourne and Brighton, Melbourne (Victoria), Australia. These participants reported pain in the region between T12 and the gluteal fold for more than three months (Von Korff, Deyo, Cherkin, & Barlow, 1993) at the time of their first physiotherapy session; which consisted of client history-taking and physical assessment. During this session, treating physiotherapists screened potential participants for study eligibility as outlined in table 3.1. using a screening tool described in detail previously by
Maitland (Maitland, Hengeveld, Banks, & English, 2005). After obtaining informed consent, the CLBP participants were tested within the same week (prior to their second physiotherapy session) by the PhD candidate. A subgroup of 17 CLBP participants underwent a second test session as part of a reliability study, five to seven days following their initial assessment. Reliability testing will be discussed in each respective study chapters.

In addition, a control group of 29 healthy matched participants with no history of CLBP were recruited from the general community and The University of Melbourne staff members through the university newsletter. Healthy participants were tested at Kieser South Melbourne on the weekends. Similarly, a subgroup of 16 healthy participants underwent a second session five to seven days apart for a reliability study. Ethics approval was obtained from The University of Melbourne’s Behavioural and Social Sciences Human Ethics Committee (ethics ID: 1340715). All participants provided written informed consent prior to entering the study. The recruitment process is summarised using a CONSORT-like flow diagram in figure 3.1.
### Table 3.1. Participant inclusion and exclusion criteria.

<table>
<thead>
<tr>
<th></th>
<th>Inclusion criteria</th>
<th>Exclusion criteria</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>CLBP</strong></td>
<td>- Men/women aged between 25-60 years old</td>
<td>- Overt neurological signs related to LBP (e.g., lower limb weakness) (Maitland et al., 2005)</td>
</tr>
<tr>
<td></td>
<td>- Pain between the T12 region to the gluteal fold with or without radiation to lower limbs for more than three months (Von Korff et al., 1993)</td>
<td>- Malignancy</td>
</tr>
<tr>
<td></td>
<td>- Ability to understand spoken/written English</td>
<td>- Pregnancy</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Systemic or active inflammatory conditions (e.g., rheumatoid arthritis)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Unstable spinal fractures (e.g., unstable spondylolisthesis)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Previous spinal or lower limb surgery</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Musculoskeletal, cardiovascular or neurological conditions that may influence balance</td>
</tr>
<tr>
<td><strong>Control</strong></td>
<td>- Men/women aged between 25-60 years old</td>
<td>- History of CLBP (Von Korff et al., 1993).</td>
</tr>
<tr>
<td></td>
<td>- Ability to understand spoken/written English</td>
<td>- History of spinal or lower limb surgery</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Musculoskeletal, cardiovascular or neurological conditions that may influence balance</td>
</tr>
</tbody>
</table>

**Sample size calculation for studies two and three**

In order to achieve a moderate effect size for force-matching error and relative phase angle studies, a Cohen’s $d$ of 0.5 with significance level, $\alpha = 0.05$ and power, $\beta = 0.90$ were chosen. Thus, 28 participants per group were required (Faul, Erdfelder, Buchner, & Lang, 2009). However, to minimise the risk of subject dropouts related to clinical logistics (e.g., equipment and room usage by clinicians and participant availability), 46 CLBP and 33 healthy controls were approached.
Figure 3.1. The CONSORT-like flow diagram of CLBP and control groups.
3.4.2. Overview of data collection process

All data were collected by the PhD candidate between December 2014 and November 2015 at two Kieser physiotherapy clinics (South Melbourne and Brighton, Melbourne, Australia). Kieser is a physiotherapy clinical group that specialises in strength training, in particular lumbar extensor strengthening. Thus, each facility is equipped with an identical lumbar dynamometer (MedX, Ocala, Florida). This allowed testing at multiple sites. The assessment protocols for each study will be discussed in detail in future chapters. The assessment protocol for each study was refined by a series of pilot studies involving ten healthy participants and five people with CLBP to determine the appropriate amount of loading, total testing time, rest periods and target movement frequency (see Appendix 3). Participants included in the pilot study meet the selection criteria but were not included in the study proper.

Specifically for Study 3, CLBP participants were phenotyped based on their disability level – as determined by their ODI scores (Fairbank & Pynsent, 2000). Those who scored 20% or below were grouped under “low” disability (CLBP\textsubscript{low}) whereas, CLBP participants who scored above 20% were grouped under “moderate-high” disability (CLBP\textsubscript{high}).

3.4.3. Experimental procedure

Self-reported disability and pain

Prior to testing, all CLBP and healthy controls completed the ODI and NRS for the assessment of disability and pain, respectively. Ten CLBP participants could not participate in force control testing due to an increase in self-reported pain level (NRS >5) prior to strength assessment.

3.4.4. Statistical analysis

Comparisons between chronic low back pain and control groups

Participant characteristics (i.e., age, weight, height, BMI, CLBP duration, pain NRS and ODI scores) were tested for normality and equality of variances using Shapiro-Wilk and Levene median tests, respectively. Additionally, quartile-quartile and probability plots were also
generated for these variables. After confirming normality, these variables were expressed in means and standard deviations (mean ± SD) with α set at 0.05.

Characteristics of healthy and CLBP participants in Study 2 were compared using independent t-tests. Chi-squared (χ²) was used to compare the number of males and females in both groups. In Study 3, participant characteristics of healthy, CLBP_high and CLBP_low groups were compared using one-way analysis of variance (ANOVA). Normality, homoscedasticity and linearity of residual for ANOVA tests were assessed using Levene’s test and scatter graphs (Osborne & Waters, 2002; Williams, Grajales, & Kurkiewicz, 2013). If variables were normally distributed, they were presented as mean ± SD. All analyses were conducted using SPSS Version 21.0 (IBM, Inc., Chicago, IL).

3.5. Results

3.5.1. Study 2: force control participant characteristics

Thirty-three individuals with CLBP and 20 healthy control participated in this study. Descriptive data for participant characteristic variables (mean ± SD) together with the results of statistical comparisons are presented in table 3.2. The CLBP group was significantly older than the control group (mean difference = 7.0 years, % difference = 17, p = 0.01, 95% CI [1.5, 12.5]). CLBP group was significantly more disabled than the control group (mean difference = 20.6%, % difference = 99.5, p < 0.001, 95% CI [16.3, 25.0]). Similarly, the CLBP group reported significantly more pain than the control group (mean difference = 3.3, % difference = 100, p < 0.001, 95% CI [2.7, 3.9]). Finally, CLBP group reported significantly longer CLBP duration than control group (mean difference = 107.5 months, % difference = 100, p < 0.001, 95% CI [2.7, 3.9]). There were no significant differences in weight, height, BMI and gender between participant groups.
Table 3.2. Descriptive data (mean ± SD) pertaining to participant characteristics of CLBP and control groups for Study 2.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP (n = 33)</th>
<th>Control (n = 20)</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
<td>41.8 ± 10.8</td>
<td>34.8 ± 8.8</td>
<td>0.01</td>
</tr>
<tr>
<td>Gender (female, %)</td>
<td>18 (54.5%)</td>
<td>10 (50%)</td>
<td>0.10</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.70 ± 0.1</td>
<td>1.66 ± 0.1</td>
<td>0.17</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>75.3 ± 17.7</td>
<td>76.6 ± 20.2</td>
<td>0.31</td>
</tr>
<tr>
<td>BMI (m/kg²)</td>
<td>25.2 ± 4.7</td>
<td>27.0 ± 6.7</td>
<td>0.80</td>
</tr>
<tr>
<td>CLBP duration (months)</td>
<td>107.5 ± 119.0</td>
<td>0.0 ± 0.0</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>ODI (%)</td>
<td>20.7 ± 12.3</td>
<td>0.1 ± 0.4</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>NRS (/10)</td>
<td>3.30 ± 1.8</td>
<td>0.0 ± 0.0</td>
<td>&lt;0.001</td>
</tr>
</tbody>
</table>

Values indicate mean ± standard deviation, n = number of participants, BMI = Body Mass Index, ODI = Oswestry Disability Index, NRS = Numerical Rating Scale

3.5.2. Study 3: lifting inter-joint coordination participant characteristics

Forty-three participants with CLBP and 29 healthy control participants completed the study. Twenty-five CLBP participants with who scored 20% or less on the ODI were grouped under CLBP with “low” disability (CLBP_low; n = 25). Eighteen CLBP participants who scored higher than 20% were grouped under CLBP “moderate-high” disability (CLBP_high; n = 18). Descriptive data for participant characteristic variables (mean ± SD) together with the results of statistical comparisons are presented in table 3.3. The CLBP_high group was significantly older than the control group (mean difference = 9.0 years, % difference = 24, F_2,69 = 3.5, p = 0.04, 95% CI [0.8, 17.2]). There were no significant differences in age between CLBP_high and CLBP_low groups and between CLBP_low and control groups. The mean overall ODI and NRS scores of the CLBP group in this study were 21.4 ± 13.6% and 3.5 ± 1.7 out of ten respectively. The CLBP_high group reported significantly more disability than CLBP_low group (mean difference = 21.3, % difference = 62, p < 0.001, 95% CI [15.6, 27.0]) and control group (mean difference = 34.4%, % difference = 100, p < 0.001, 95% CI [30.7, 38.1]). Moreover, the CLBP_high group reported significantly more pain than CLBP_low group (mean difference = 1.5, % difference = 51, p = 0.00, 95% CI [0.4, 2.7]) and control group (mean difference = 4.5, %
difference = 100, \( p < 0.001 \), 95% CI [3.7, 5.3]). Finally, both CLBP_{high} (mean difference = 155.2 months, % difference = 100.0, \( p < 0.001 \), 95% CI [91.2, 219.2]) and CLBP_{low} groups (mean difference = 110.2 months, % difference = 100.0, \( p < 0.001 \), 95% CI [52.0, 168.5]) reported significantly longer CLBP duration than control group. There was no significant difference in CLBP duration between CLBP_{high} and CLBP_{low} groups. Similarly, there were no significant differences between all participant groups with respect to weight, height, BMI and gender.

Table 3.3. Descriptive data (mean ± SD) pertaining to participant characteristics of CLBP and control groups for Study 3.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP_{low} (n = 25)</th>
<th>CLBP_{high} (n = 18)</th>
<th>Control (n = 29)</th>
<th>( p ) values</th>
</tr>
</thead>
<tbody>
<tr>
<td>\text{Age (years)}</td>
<td>42.3 ± 11.1</td>
<td>46.7 ± 11.8</td>
<td>37.8 ± 11.5</td>
<td>0.04*</td>
</tr>
<tr>
<td>\text{Gender (female, %)}</td>
<td>11 (27.5%)</td>
<td>12 (30%)</td>
<td>17 (42.5%)</td>
<td>0.31</td>
</tr>
<tr>
<td>\text{Height (m)}</td>
<td>1.74 ± 0.1</td>
<td>1.70 ± 0.1</td>
<td>1.67 ± 0.1</td>
<td>0.053</td>
</tr>
<tr>
<td>\text{Mass (kg)}</td>
<td>81.6 ± 19.0</td>
<td>71.5 ± 14.0</td>
<td>73.4 ± 17.6</td>
<td>0.12</td>
</tr>
<tr>
<td>\text{BMI (m/kg}^2)\text{)}</td>
<td>26.0 ± 5.0</td>
<td>24.8 ± 3.8</td>
<td>25.9 ± 5.6</td>
<td>0.70</td>
</tr>
<tr>
<td>\text{CLBP duration (months)}</td>
<td>110.2 ± 107.8</td>
<td>155.2 ± 173.2</td>
<td>0.0 ± 0.0</td>
<td>&lt;0.001*</td>
</tr>
<tr>
<td>\text{ODI (%)}</td>
<td>13.2 ± 4.9</td>
<td>34.4 ± 10.9</td>
<td>0.1 ± 0.4</td>
<td>&lt;0.001**</td>
</tr>
<tr>
<td>\text{NRS (/10)}</td>
<td>3.0 ± 1.6</td>
<td>4.5 ± 1.9</td>
<td>0.0 ± 0.0</td>
<td>&lt;0.001**</td>
</tr>
</tbody>
</table>

Values indicate mean ± standard deviation, n = number of participants, BMI = Body Mass Index, ODI = Oswestry Disability Index, NRS = Numerical Rating Scale. *significant difference between CLBP_{high} and healthy groups. **significant difference between healthy group and both CLBP_{high} and CLBP_{low} groups. ***significant difference between CLBP_{high} and both CLBP_{low} and healthy groups.

3.6. Discussion

3.6.1. Overview of results

As anticipated, the results of this study suggest that CLBP participants were significantly more disabled than healthy control participants. Moreover, CLBP participants reported significantly more disability and pain than control participants. In Study 2, the CLBP group was significantly older than the control group. In Study 3, CLBP_{high} group reported higher levels of pain and
disability than CLBP_{low} and control groups. The CLBP_{high} group was also significantly older than the control group.

### 3.6.2. Oswestry Disability Index

Participants with CLBP reported higher ODI scores than healthy controls (accepted H\textsubscript{1a} and H\textsubscript{1b}). The ODI scores of the CLBP participants in this study (mean ODI = 21.4 ± 13.6\%; \textit{i.e.}, moderate disability) is comparable to the ODI scores reported in related studies incorporating open and/or closed kinetic chain tests in people with CLBP (see tables 2.2, 2.3 and 2.4). Only five studies (see sections 2.6 and 2.7) utilised the ODI to describe self-reported disability of their CLBP participants. The ODI scores in these studies ranged from 22.6 to 26.8 – \textit{i.e.}, moderate disability. A comparison between the ODI characteristics for CLBP participants in this study and other studies are depicted in figure 3.2. Although other studies utilised alternative disability outcome measures such as the RMDQ and QBPDS (Kopec, 2000; Kopec \textit{et al.}, 1996; Roland & Fairbank, 2000; Roland & Morris, 1983), these outcome measures correlate poorly with the ODI (r < 0.5) (Morris, Hee, Stallard, Underwood, & Patel, 2015). Thus, it is not recommended to convert results between different disability outcome measures (Morris \textit{et al.}, 2015). Therefore, only studies utilising the ODI were compared with this study.
3.6.3. Numerical Rating Scale

Participants with CLBP reported higher NRS scores than healthy controls (accepted H$_{2a}$ and H$_{2b}$). The pain scores of the CLBP participants in this study (mean NRS = 3.5 ± 1.7 out of 10 – *i.e.*, moderate pain) are comparable to the pain scores reported in related CLBP studies (see tables 2.2, 2.3 and 2.4). Most studies presented in Chapter 2 utilised the VAS and NRS when assessing the pain intensity of their CLBP groups. Both of these pain outcome measure scores are comparable as both VAS and NRS are highly correlated in people with CLBP ($r = 0.90$) (Hjermstad *et al*., 2011; Lee *et al*., 2015). The pain scores in the literature reviewed in Chapter 2 range from 2.0 to 6.0 out of ten – *i.e.*, mild to moderate pain. A comparison of pain intensity for CLBP participants between this study and related studies is depicted in figure 3.3. One study (Paquet *et al*., 1994) utilised the Present Pain Intensity – a subscale of McGill Pain Questionnaire which asked the participant to rate their present pain out of five categories (1 = mild, 2 = discomforting, 3 = distressing, 4 = horrible and 5 = excruciating) (Melzack, 1975).
which may not be comparable with VAS and NRS. Thus, the study by Paquet et al. (1994) was not compared with this study.

Figure 3.3. The pain intensity scores of people with CLBP in this study (yellow bar) and other studies (light blue bars).

3.7. Conclusion

In conclusion, CLBP participants reported significantly more disability and pain than healthy controls. Disability levels and pain intensity reported by CLBP participants were comparable to the related studies in the literature; thus, this thesis’ CLBP participants are representative of moderate disability level and pain intensity.
Chapter 4
Study 2

Lumbar extensor muscle force control in people with and without chronic low back pain

4.1. Introduction

It is well known that impairments in trunk muscle function are a characteristic of CLBP (Radebold et al., 2000; van Dieen et al., 2003). Traditional open kinetic chain neuromuscular assessments such as lumbar extensor strength tests are commonly performed to quantify trunk muscle function in people with CLBP (Steele et al., 2014). However, the methods used to compare lumbar extensor strength of people with or without CLBP – typically quantified as torque during MVIC in CLBP and healthy individuals are not consistent in the literature (see section 2.6.1). Therefore, it is unclear whether people with CLBP have weaker lumbar extensor muscles compared to healthy people (Verbunt et al., 2003; Verbunt et al., 2010). Changes in trunk muscle function have also been proposed to occur as a result of CLBP; possibly due to alterations in motor planning, changes in spinal reflexes (Hodges, 2001; Indahl, Kaigle, Reikeras, & Holm, 1997) and changes to the mechanical properties of passive structures such as the ligaments and discs of the spine (Dreyer & Dreyfuss, 1996; Roberts, Eisenstein, Menage, Evans, & Ashton, 1995; Sjolander et al., 2002) (see section 2.4). Collectively, these changes may impact on the ability of people with CLBP to control trunk muscle force output. Muscle force control is defined as the ability of the neuromuscular system to produce coordinated, accurate and/or smooth (i.e., less variable) force output (Enoka, 2002; Tracy, Mehoudar, & Ortega, 2007). In general, trunk muscle force control is determined by measuring
the trunk muscle submaximal force steadiness or submaximal force accuracy (see section 2.6.2). People with CLBP have demonstrated to be 24%-28% less steady in producing constant submaximal trunk extensor force than healthy people (Miura & Sakuraba, 2014). However, in most functional tasks the trunk extensor muscles produce dynamic force (i.e., cyclical loading) (Marras, 2008). Hence, one may question the external validity of assessing trunk extensor force control using a constant force target. In this context, incorporation of a variable or fluctuating submaximal force target may be more appropriate when assessing trunk muscle force control. Moreover, utilising fluctuating force target allows the assessment of force production accuracy during the ramp up (i.e., increasing force production) or ramp down (i.e., decreasing force production) phases of the test. The ability to control trunk extensor muscle force output accurately could be an important risk factor in the development of work related musculoskeletal disorders, which are known to involve cyclic phases of increasing and decreasing force (Srinivasan & Mathiassen, 2012).

A recent study reported on the development of a submaximal lumbar extensor force-matching task which was found to possess good inter-session reliability in a group of healthy people without history of CLBP (Reeves et al., 2014). However, to the author’s knowledge, no previous studies have assessed the reliability of a lumbar extensor force-matching task in people with CLBP. Moreover, no studies have compared trunk extensor muscle force control utilising a fluctuating submaximal force target in people with CLBP and healthy individuals. Therefore, it is unclear whether lumbar extensor muscle force control is impaired during such a task.
4.2. Aims

Thus, based on the abovementioned rationale, this study aimed to:

1. Assess the reliability of a novel submaximal lumbar extensor muscle force control protocol in healthy controls and people with CLBP,
2. Compare lumbar extensor MVIC in people with CLBP and healthy controls (the same participants reported in Study 1),
3. Compare isometric lumbar extensor muscle force control in people with CLBP and healthy controls using the aforementioned force-matching task. Specifically, by assessing the amount of force control error:
   a. Throughout the force-matching task (i.e., total error).
   b. During the ramp-up part of the test. (i.e., increasing force production)
   c. During the ramp-down part of the test (i.e., decreasing force production).
4. Compare the differences between people with CLBP and healthy controls during the ramp-up and ramp-down parts of the test.

4.3. Hypotheses

Based on the previous equivocal literature (see sections 2.6.1 and 2.6.2), the following hypotheses were adopted for the present study:

1. The submaximal lumbar extensor muscle force control protocol would have good inter-session reliability in:
   a. CLBP group (H_{1a}) (Descarreaux et al., 2004)
   b. Control group (H_{1b}) (Reeves et al., 2014).
2. The CLBP group would demonstrate significantly less lumbar extensor MVIC than the control group (Steele et al., 2014) (H_{2}).
3. Compared to the control group, the CLBP group would demonstrate:
   a. Significantly greater total force-matching error (Miura & Sakuraba, 2014; Willigenburg et al., 2013) (H₃ᵃ).
   b. Significantly greater force-matching error during the ramp up phase of the test (Miura & Sakuraba, 2014; Willigenburg et al., 2013) (H₃ᵇ).
   c. Significantly greater force-matching error during the ramp down phase of the test (Radebold, Cholewicki, Polzhofer, & Greene, 2001) (H₃ᶜ).

4. Within the groups it was hypothesised that:
   a. People with CLBP would demonstrate significantly more error during the ramp-down compared to the ramp-up part of the test (H₄ᵃ) (van Dieen et al., 2003).
   b. Control group would demonstrate similar amount of error during the ramp-up and ramp-down parts of the test (H₄ᵇ).

4.4. Materials and methods

4.4.1. Participants

The CLBP group consisted of 33 men and women aged 25-60 years with CLBP. In addition, a control group consisting of 20 healthy control participants (matched to the CLBP group according to age, gender and BMI) with no history of CLBP were recruited. See section 3.4.1 for detailed participant recruitment process and sample size calculation and table 3.1 for eligibility criteria.

A subgroup of the CLBP (n = 17) and control groups (n = 16) were invited to attend a second session 5-7 days apart prior to their second physiotherapy session (i.e., prior to intervention) to complete a reliability study for the lumbar extensor muscle force control and lumbar extensor strength assessment protocols. This sample size is larger than a recently published reliability
study (n = 10) employing similar assessment of a force-matching task – albeit only in a healthy population (Reeves et al., 2014).

4.4.2. Overview of experimental procedure

The PhD candidate developed the lumbar extensor muscle force control protocol founded on recently published studies on the quadriceps (Perraton et al., 2016; Telianidis et al., 2014) with the assistance of the research team. The additional analysis of force-matching accuracy during increasing and decreasing force production was a novel feature of this study. The assessment protocol was refined by a pilot study and reliability testing prior to commencing the main study proper. The pilot study involved ten healthy controls and five people with CLBP to determine the appropriate amount of loading, total testing time, rest periods and target movement frequency (see Appendix 3). Participants included in the pilot study meet the eligibility criteria for the healthy control group but were not included in the study proper or the reliability testing. The inter-session reliability of lumbar extensor muscle force control protocol was assessed in both control and CLBP participants who were willing to be re-assessed within 5-7 days of initial testing.

4.4.3. Experimental procedure

Self-reported disability and pain

All CLBP participants completed the ODI (see section 2.4.2) and NRS (see section 2.4.1) prior to testing. Additionally, participants were also asked to rate their pain using the NRS before and after testing.

Isometric test preparation

Participants were instructed to sit as far back as possible on the lumbar dynamometer (MedX, Ocala, Florida) with their back and head supported by the backrest and headrest, respectively. The dynamometer moving arm (i.e., the back/head rest part) was locked in a neutral spine position (i.e., 12° flexion, 0° being full extension; (Graves et al., 1994)). In this position, a belt
was fastened around the participant’s lap. The dynamometer foot rest height was continuously decreased and locked when participants were not able to move their feet. Similarly, a femoral restraint which provided a downward pressure on participants’ thigh was adjusted. These steps were necessary to prevent the pelvis from posteriorly rotating, limiting movements of the lower limb and, in turn, minimising the contribution of lower limb musculatures (specifically the hip extensors) during isometric lumbar extension testing (Graves et al., 1994; Smith, Bissell, Bruce-Low, & Wakefield, 2011). A diagram of the MedX restraint system and picture of the setup on the MedX are presented in figure 4.1A and B respectively.

![Diagram of MedX restraint system and setup on MedX](image)

**Figure 4.1.** The MedX restraint system (adapted from Graves *et al.*, 1994) (A). Full extension is defined as 0° lumbar flexion and full flexion is 72° lumbar flexion. A participant is seated on MedX Lumbar Dynamometer at 12° lumbar flexion (B).

After the steps involving pelvic stabilisation, participants’ pain-free (NRS = 0/10) lumbar ROM was measured. To do so, the dynamometer moving arm was unlocked and participants’ trunk was moved comfortably from flexion (maximum of 72°) to extension (minimum of 0°). This flexion-extension “rocking” motion were repeated for 30 seconds as a warm-up for lumbar soft tissues and at the end of which, the pain-free lumbar ROM measurement was taken. Following this, participants were instructed to close their eyes and were again rocked into flexion-
extension at approximately 50% of their lumbar ROM for five seconds and were instructed to find their upright position. Doing so allowed a counterweight to be measured to minimise the contribution of the upper body resultant force (i.e., the head, torso and upper limbs) during lumbar extension MVIC measurement (Graves et al., 1990; Pollock, Graves, & Leggett, 1991). This counterweight force was deducted from the total isometric lumbar extension force generated. All the aforementioned settings for each participant were saved automatically on the MedX computer. For participants involved in reliability testing (n_{CLBP} = 17, n_{Healthy} = 16), these settings were used during re-testing. The participant testing position is illustrated in figure 4.2.

Preceding the lumbar extensor MVIC testing, the participants were given a warm-up session to accustom them to the strength testing procedure. The participants were locked at 12° lumbar flexion (neutral spine) and instructed to press their lower back against the backrest increasing isometric force production (ramp up) to approximately 50% MVIC over four seconds. This contraction was sustained over two seconds. Participants were allowed a minimum of 30 seconds of rest prior to MVIC testing.

**Isometric lumbar extensor strength**

Lumbar extensor MVIC was measured to derive submaximal values for the target-matching task. The MedX has demonstrated validity (Pollock et al., 1991) and reliability in assessing isometric strength and range of motion in healthy (r = 0.81-0.97, SEE = 24.0 Nm) (Graves et al., 1990) and CLBP (r = 0.57-0.93, SEE = 12.0 – 44.5 Nm) populations (Robinson, Greene, O'Connor, Graves, & MacMillan, 1992). Only one MVIC trial was performed in order to minimise the risks of muscle fatigue and cramps which could impede force control testing. Participants were instructed to ramp-up their isometric force production over four seconds and hold their MVIC for two seconds. During this two-second hold, the MVIC measurement was taken.
Lumbar extensor muscle force control assessment

A novel method of assessing muscle force control utilising a variable force target - recently developed in our laboratory – tested patients following ACLR (Perraton et al., 2016; Telianidis et al., 2014). In this study, ACLR and healthy control participants were instructed to contract their quadriceps to increase force production from 5%-20% MVIC fluctuating at the frequency of 0.128 Hertz (Hz). The ACLR group demonstrated increased force-matching error compared to the control group indicating an impairment of quadriceps muscle force control (Telianidis et al., 2014). This particular assessment technique was adapted to the lumbar extensor muscles in this study.

Participants adopted the same position as lumbar extensor MVIC assessment described above. Visual feedback was provided via a tablet (HP Stream 8 5801TW, running Microsoft Windows 8.1) placed approximately 1.5 meter in front of the participant slightly below eye level without affecting head or upper torso placement on the MedX (see figure 4.2). An analogue output was used to extract the raw data from the MedX at 2000 Hz. The output was connected to a CompactDAQ with 9125 modules (National Instruments U.S.A.). A customised LabVIEW software (National Instruments U.S.A.) program was then used to analyse the data (see figure 4.3A and B).

On the tablet, participants were shown two coloured cursors: yellow represented the participants’ torque and red represented the target torque (see figure 4.3). When testing began, the red target torque oscillated at a frequency of 0.16 Hz for 60 seconds. This frequency is equivalent to ten sinusoidal cycles per minute; or five ascending and five descending cycles. The upper and lower limits for the red target cursor were set at 20% and 50% participants’ lumbar extensor MVIC respectively, determined from isometric lumbar extensor MVIC testing above. The value limits of 20% and 50% lumbar extensor MVIC for the varying force target were selected based upon estimated lumbar extensor contraction intensities during activities of
daily living (e.g., pushing and (unexpected) turning a 200 kg cart ≈2.89-62.79% maximal voluntary contraction (Lee, Hoozemans, & van Dieën, 2012)) and from a separate pilot testing (see Appendix 3). Participant feedback during pilot testing confirmed that when testing was conducted longer than 60 seconds or at higher load limits or at higher frequency, it was associated with lower back discomfort post-testing and worse performance. Additionally, when two or more trials were performed, five healthy pilot participants reported increased lower back discomfort or cramping. Hence, the test upper and lower load limits, frequency, type of target pattern and duration of testing were selected based upon a combination of the most challenging test the pilot participants were able to complete safely that had the least adverse effects on task performance. Prior to commencing the test, participants were provided with a standardised explanation and instruction of the test and were given an opportunity to ask questions.

In order to familiarise participants with the testing protocol as well as to minimise learning effects (and pain provocation in CLBP participants), participants completed a single practice trial of 60 seconds duration. To minimise the effect of pain anticipation during force control testing (Al-Obaidi et al., 2000), participants were reassured that only a small fraction to half of the maximum force was required for the trial. During this trial, participants were instructed to match the moving red torque target as accurately as possible by increasing (ramping up) isometric torque production from 20-50% lumbar extension MVIC and decreasing (ramping down) isometric torque production from 50-20% lumbar extension MVIC (figure 4.3B). As torque differences between target and participants’ torques could appear accentuated in participants with lesser strength, the target torque was normalised between 20% and 50% MVIC (limits of target matching task). Participants were instructed to focus on the software on the tablet and to remain silent during testing. No verbal encouragement was given and the testing environment was kept quiet and free of visual distractions. Following familiarisation,
participants were allowed 30 seconds of rest prior to testing proper. The testing proper was identical to the practice trial.

Figure 4.2. Participant set-up for lumbar extensor muscle force control assessment.

Figure 4.3. Participant’s view during lumbar extensor force control assessment (A). A screenshot of lumbar extensor muscle force control assessment process (B).
4.4.4. Data analysis

Isometric lumbar extensor strength and force control

Lumbar extensor isometric strength data were obtained from the MedX, filtered using a low-pass Symlet-8 undecimated wavelet filter (62.5 Hz) and converted to torque in Newton meters (Nm). Calibration for the custom data acquisition system was performed by applying a series of loads to the machine, recording the results from the MedX software and the raw data from the data acquisition system and creating a calibration factor for the raw data with the MedX results as the criterion reference using linear regression.

Lumbar extensor muscle force control was calculated by quantifying the disparity between participant-generated force and the target force using root mean squared error (RMSE). The calculation of RMSE is preferred to the calculation of COV as differences between participants’ torque output and target torque could appear accentuated in participants with lesser strength when using COV calculations. RMSE calculation provides positive numbers with higher RMSE indicating greater error or greater deviation from target torque. To account for potential errors caused by participants adjusting their sitting position, the results of the first and last repetition of the sine wave were not used for calculation. The average value for the remaining two repetitions were used for statistical analysis. A representative plot of lumbar extensor muscle force versus target force is presented in figure 4.4.

In addition to the measurement of the average total RMSE (RMSET) described above, sub-regional analyses were also performed to measure RMSE during the ascending (ramp up; RMSEA) phase of the test (between 20%-50% lumbar extensor MVIC) and the average RMSE during the descending (ramp down; RMSED) phase of the test (between 50%-20% lumbar extensor MVIC; figure 4.4). This approach was implemented to measure the phase of the test at which the error is the greatest.
4.4.5. Statistical analysis

Reliability testing

In order to measure inter-session test-retest reliability, Intra-class Correlation Coefficient (ICC$_{3,k}$) with 95% Confidence Intervals (CI) was used. The ICC quantifies degree of which individual variables are related to themselves between the two measurement points and thus, are used to evaluate the consistency or reproducibility of outcome measures (Shrout & Fleiss, 1979). ICC model 3 and form k were chosen because average measures were taken for each variables of interest (Portney & Watkins, 2009). ICC values range from 0 to 1 with values closer to 1 represents stronger reliability; values less than 0.40 = poor, 0.40 – 0.59 = fair, 0.60 – 0.74 = good and >0.75 = excellent (Bruton, Conway, & Holgate, 2000). Although there is no consensus of the acceptable value of ICC, Chinn (1991) recommended a value above 0.6 for an outcome measure to be considered reliable. Standard error of measurement (SEM) was calculated by multiplying the standard deviation of the sample variable and the square root of one minus the ICC (Tighe, McManus, Dewhurst, Chis, & Mucklow, 2010).
Comparisons between chronic low back pain and control groups

Normality and equality of variances were analysed using Shapiro-Wilk and Levene median tests respectively for participant characteristic variables (age, weight, height, BMI, CLBP duration, NRS and ODI scores) and open kinetic chain variables (lumbar extension MVIC, RMSET, RMSEA and RMSED). Additionally, quartile-quartile and probability plots were also observed for these variables. The plots were used to determine normality if there was a discrepancy between normality test results and the plots. If the variables were normally distributed they were expressed in mean ± SD with significance level, α, set at 0.05. Between-group participant characteristics were compared using a series of independent t-tests. \( \chi^2 \) was used to compare the number of males and females in healthy and CLBP groups.

Analysis of covariance (ANCOVA) was performed to control for age differences between CLBP and healthy groups (see section 3.5.1). \( H_1 \) was assessed using group (CLBP and healthy controls) x RMSE (RMSET, RMSEA and RMSED) factorial ANCOVA to compare the main effects, group simple effects and interactions between group and RMSE. Within group analyses were conducted using one-way ANCOVA to compare RMSEA and RMSED. The effect size of significant ANCOVA test results were quantified using partial eta squared (\( \eta^2_p \)) with values around 0.01 is classified as small, around 0.06 as medium and around 0.14 as large effect size (Cohen, 1988). Normality, homoscedasticity and linearity of residual for ANCOVA tests were assessed using Levene’s test and scatter graphs (Osborne & Waters, 2002; Williams et al., 2013). Pair-wise comparisons were conducted using Fisher’s Least Significant Difference test.

All analyses were conducted using SPSS Version 21.0 (IBM, Inc., Chicago, IL).
4.5. Results

4.5.1. Participant characteristics

Participant characteristics (mean ± SD) have been summarised in section 3.5.1 and table 3.2. There was a significant difference in age between CLBP and control group (mean difference = 7.0 years, % difference = 17, \( p = 0.01 \), 95% CI [1.53, 12.48]). There were no significant differences between participant groups for weight, height, BMI and gender.

4.5.2. Reliability

All the variables tested have demonstrated moderate to high reliability evident from the ICC ranging from 0.71 to 0.98. The ICC and SEM of the variables tested derived from open kinetic chain assessment is summarised in table 4.1.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP (n = 17)</th>
<th>Control (n = 16)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>LE strength (Nm)</td>
<td>0.83</td>
<td>2.33</td>
</tr>
<tr>
<td>RMSE_T</td>
<td>0.88</td>
<td>0.04</td>
</tr>
<tr>
<td>RMSE_A</td>
<td>0.85</td>
<td>0.01</td>
</tr>
<tr>
<td>RMSE_D</td>
<td>0.83</td>
<td>0.04</td>
</tr>
</tbody>
</table>

ICC = Intraclass-correlation coefficient, SEM = Standard error of measurement in the unit of the variable, LE strength = Isometric lumbar extensor strength, Nm = Newton.meter, RMSE_T = total average root mean square error, RMSE_A = average root mean square error during ascending phase, RMSE_D = average root mean square error during descending phase

4.5.3. Comparisons between chronic low back pain and control groups

Pain during testing

There were no reported adverse events during or following testing. CLBP and control participants did not report any increase in pain during and after testing.
Isometric lumbar extensor strength

There were no significant differences in lumbar extension MVIC between CLBP group (180.00 ± 12.61 Nm) and control group (182.03 ± 16.42 Nm, F_{1,50} = 0.009, p = 0.93). Age was not a significant covariate for this general linear model (p = 0.53).

Isometric lumbar extensor muscle force control

Descriptive data (mean ± SD) for the force control variables together with the results of statistical comparisons for CLBP and control groups are presented in table 4.2. Age was a significant covariate (p < 0.001) for this analysis. Controlling for age, there was a significant main effect of group (F_{1,152} = 11.2, \eta^2_p = 0.07, p = 0.001) and main effect of RMSE (F_{2,152} = 3.3, \eta^2_p = 0.04, p = 0.039). The group and RMSE interaction was approaching significance (F_{2,152} = 2.7, \eta^2_p = 0.03, p = 0.069). The CLBP group produced significantly more RMSE_T than the control group (% difference = 30, p = 0.022, 95% CI [0.18, 2.21]). Similarly, the CLBP group also produced significantly more RMSE_A than the control group (% difference = 45, p = 0.001, 95% CI [0.76, 2.80]). However, there was no significant differences in RMSE_D between both groups (p = 0.80). A sample muscle force control test outcome for a participant with good and impaired ability to control force output is illustrated in figure 4.5.
Figure 4.5. A sample outcome of lumbar extensor muscle force control assessment for someone with good (i.e., a control participant) (A) and impaired (i.e., a CLBP participant) ability to control force output (B). Red trace = target force, blue trace = participant’s force.

Table 4.2. Descriptive data (mean ± SD) pertaining to open kinetic chain variables together with results of statistical analyses in CLBP and control groups.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP Mean ± SD</th>
<th>Control Mean ± SD</th>
<th>Mean Difference [95% CI]</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td>LE strength (Nm)</td>
<td>180.00 ± 12.61</td>
<td>182.03 ± 16.42</td>
<td>2.02 [-40.63, 44.67]</td>
<td>0.93</td>
</tr>
<tr>
<td>RMSE_T</td>
<td>5.21 ± 1.90</td>
<td>3.60 ± 1.01</td>
<td>1.60 [0.78, 2.43]</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>RMSE_A</td>
<td>5.87 ± 1.90</td>
<td>3.68 ± 1.25</td>
<td>2.19 [1.01, 3.37]</td>
<td>0.001</td>
</tr>
<tr>
<td>RMSE_D</td>
<td>4.13 ± 1.35</td>
<td>3.59 ± 1.30</td>
<td>0.54 [-0.22, 1.30]</td>
<td>0.16</td>
</tr>
</tbody>
</table>

SD = Standard deviation, CI = Confidence interval, LE strength = Lumbar extensor strength, Nm = Newton.meter, RMSE_T = Average total root mean square error, RMSE_A = Average root mean square error of ascending phase, RMSE_D = Average root mean square error of descending phase
Within group RMSE comparison demonstrated that in the CLBP group, there was a main effect of RMSE ($F_{2,95} = 5.83, \ p = 0.004$). There were large and statistically significant differences between RMSE$_D$ and RMSE$_A$ (mean difference = 1.74, % difference = 43, $\ p < 0.001$, 95% CI [0.87, 2.61]) and between RMSE$_D$ and RMSE$_T$ (mean difference = 1.08, % difference = 21, $\ p = 0.015$, 95% CI [0.21, 1.95]) but the difference between RMSE$_A$ and RMSE$_T$ was not significant ($\ p = 0.14$). There was no significant within group differences in all RMSE variables in healthy controls ($\ p = 0.87 - 0.98$). Between- and within-group RMSE variables of CLBP and control groups are shown in figure 4.6.

**Figure 4.6.** Root mean squared error (RMSE) values in CLBP and control groups. In the CLBP group, RMSE$_A$ (ascending) is significantly larger than RMSE$_D$ (descending). No within group differences in RMSE values were found in healthy controls. $^a p < 0.01$, $^* p < 0.05$. Error bars denote standard deviation.
4.6. Discussion

4.6.1. Overview of results

This study found that (i) the assessment of lumbar extensor muscle force control can be performed reliably in CLBP and healthy individuals. There was (ii) no significant differences in lumbar extensor strength between CLBP and healthy individuals. People with CLBP demonstrated (iii) increased force-matching error compared to healthy controls indicating impaired lumbar extensor muscle force control. A detailed discussion of each finding follows.

4.6.2. Reliability of the lumbar extensor muscle force control protocol

The assessment of submaximal lumbar extensor muscle force demonstrated moderate to high reliability in healthy people (accepted H1a) and people with CLBP (accepted H1b). The lumbar extensor muscle force control task demonstrated excellent inter-session reliability (i.e., ICC between 0.83 and 0.88) in healthy people and moderate-excellent inter-session reliability (i.e., ICC between 0.71 and 0.98) in people with CLBP. The differences in reliability found between the groups were surprising and could be attributed to a number of factors such as variability in their neuromuscular systems (Hodges, Coppieters, MacDonald, & Cholewicki, 2013), variable function and expectations of the test, expectations of pain (Al-Obaidi et al., 2000) resulting in more variable performance – as evidenced by the higher SEM in the CLBP group (SEM = 2.33). Recent study demonstrated that the assessment of target matching task using a similar protocol in the trunk has excellent inter-session reliability (ICC = 0.95) in healthy people (Reeves et al., 2014). This study confirms Reeves et al. (2014) findings that lumbar extensor muscle force control can be assessed reliably in healthy controls, and now in CLBP populations.

4.6.3. Lumbar extensor strength in people with chronic low back pain

The results of this study may seem contrary to its primary hypothesis (reject H2,) with CLBP from healthy controls. Lumbar extensor weakness is believed to be prevalent in people with CLBP (Steele et al., 2014). Steele et al. (2014) argued that isolation of the lumbar extensor
muscle is paramount for a valid assessment of lumbar extensor strength. Therefore, the use of a lumbar dynamometer that adequately restrains the hip extensor muscles has long been advocated to test for lumbar extension strength (Graves et al., 1990; Graves et al., 1994). However, the results of the current study reporting no significant difference in lumbar extension strength between CLBP and control groups despite utilising Graves et al. (1990, 1994) assessment protocol. Lumbar extensor strength is a widely debated topic in CLBP with studies reported evidence of multifidus deconditioning (Hides et al., 2008; Hides, Stokes, Saide, Jull, & Cooper, 1994) in the absence of lumbar extensor strength (Verbunt et al., 2003; Verbunt et al., 2010). However, it has been postulated that the assessment of lumbar extensor MVIC was not only solely affected by the presence of pain but also by fear of re-injury, anticipation of pain during testing and sincerity of effort (see section 2.6.1). This study’s CLBP cohort pain level was comparable to those of Lariviere et al. (2010) Lariviere et al. (2010) (VAS = 3.1-3.6) who also found no significant difference in lumbar extensor strength in people with CLBP to healthy people. If pain anticipation explains a large variance (up to 25%) of lumbar extensor MVIC (Al-Obaidi et al., 2000), the pain level reported in the CLBP group in this study may not be high enough to show significant differences in lumbar extensor MVIC. Thus as some researchers suggest, lumbar extensor MVIC assessment, albeit reliable, may be of little utility to differentiate people with CLBP from healthy people (Robinson, Greene, et al., 1992).

4.6.4. Lumbar extensor muscle force control in people with chronic low back pain

This study demonstrated that people with CLBP are impaired with respect to controlling variable, submaximal lumbar extensor force output compared to healthy controls. In particular, people with CLBP demonstrated elevated RMSET (accepted H3a) and RMSEA (accepted H3b) but not RMSED (rejected H3c) compared to healthy people. The difference between force control error in this study were considerably larger than those reported in a previous study (Miura &
Sakuraba, 2014) (i.e., 24%-28% vs. 30%-45%) which used a non-variable, sub-maximal force target (see section 2.6.2).

The higher amount of total error in this study could be explained by a several factors. Firstly, the central nervous system increases the level of agonist (i.e., the trunk extensors) muscle activity proportionally as the amount of load lifted increases (Reeves, Pathak, Popovich, & Vijayanagar, 2013); in this case as the force target increases from 20% to 50% MVIC. As the force target decreases from 50% to 20% MVIC, activation of antagonist muscles (i.e., trunk flexors) (Reeves et al., 2013) and/or a proportional decrease in lumbar extensor muscle activity is required to decrease force output to maintain an accurate target matching. There is evidence to suggest that the ability of the central nervous system to regulate agonist-antagonist muscle activity is impaired in people with CLBP (D'Hooge et al., 2013). This impairment could be explained by the ‘motor adaptation to pain theory’ (Hodges & Tucker, 2011). Specifically, pain tends to increase net trunk muscle activation and, in turn, the stiffness of trunk musculature (i.e., trunk muscle co-contraction) (Hodges et al., 2013; van Dieën et al., 2003) – adaptations that may persist well after the resolution/attenuation of symptoms (Hodges, van den Hoorn, Dawson, & Cholewicki, 2009). Increased trunk muscle stiffness has been associated with decreased anticipatory lumbar movement in response to perturbations (Mok, Brauer, & Hodges, 2007) and delayed anticipatory trunk muscle agonist-antagonist coordination time in response to changes in external loads (Radebold et al., 2000). Therefore, it is not surprising that augmented force-matching error is observed when the target force is constantly varied in people with CLBP.

Secondly, increased total error could be attributed to the isolation of lumbar extensor muscle during testing. In order to maximise lumbar extensor muscle activity, adequate pelvic restraint is required to minimise the involvement of larger and more powerful hip extensors during lumbar extension (Smidt et al., 1983). This study replicated the set-up suggested by Graves et
al. (1994) where a restraint system at 12° lumbar flexion (pelvic, femur, thigh restraints and a foot board) was used to stabilise the pelvis and a counterweight was utilised to neutralise the effect of gravity on upper body mass. This set-up has been demonstrated to minimise gluteal muscle activation during isometric submaximal muscle testing (Udermann et al., 1999). Moreover, in a past study, when the lumbar extensor muscles were fatigued, the hamstring musculature was recruited as a compensatory strategy which could confound lumbar extensor muscle force assessment (San Juan et al., 2005). In this study, lumbar extensor fatigue was minimised by utilising single MVIC and force-matching trials and frequent rests. Therefore, possible compensatory strategies adopted by the hip extensors were minimised by the study design. Isolated lumbar extensor assessment perhaps also contributes to more error being exhibited during our force-matching task compared to past studies where lower limb contributions during testing were not being adequately controlled for.

4.6.5. Within-group lumbar extensor muscle force control phase analysis

Interestingly, in this study people with CLBP demonstrated more impairment in force control during the ramp-up phase than during the ramp-down phase of the test despite no change in pain level during testing (rejected H_{4a}). In contrast, control participants demonstrated almost identical amount of errors during both test phases (accepted H_{4b}). Although it is difficult to generalise this finding to previous studies due to the novelty of this study’s methodology, Descarreaux et al. (2004) reported slower rate of isometric force production (i.e., ramp-up rate) compared to healthy controls (see section 2.6.2) which could result in increased error in the ramp-up phase in the CLBP group in the present study. During the force-matching test utilised in this study, the lumbar extensors contracted concentrically during ramp-up phase and relax during ramp-down phase of the test. Increased error in concentric contraction (ramp-up phase) is supported by the findings of MacDonald et al. (2011) who reported over-recruitment of multifidus (superficial and deep fibers) muscles – denoted by increased muscle thickness on
ultrasonography, during gentle leg lifting activity in people with recurrent LBP. Over-recruitment of lumbar extensor muscles could result in CLBP participants to overshoot resulting in augmented error in the ramp-up phase of the force-matching test.

4.6.6. Clinical implications

Based on the study findings, lumbar extensor strength assessment may not be justifiable for people with CLBP. Although lumbar extensor muscle atrophy has been reported in people with CLBP (Fortin & Macedo, 2013), it is not associated with lumbar extensor strength (Käser et al., 2001). Moreover, lumbar extensor strength is not associated with self-reported disability in people with CLBP (Al-Obaidi et al., 2000) (see section 2.8). (Boucher, Preuss, Henry, Dumas, & Larivière, 2016)

Motor control impairment has been suggested to be a mechanism contributing to the pathogenesis of CLBP (Cholewicki et al., 2005; Panjabi, 1992b). Some activities of daily living (e.g., lifting) are highly variable and exert high compression forces on the spine (Hoozemans et al., 2008). Such activities require an optimal trunk motor system (central and peripheral) to accurately modulate muscle activity in order to control internal and external forces imposed on the trunk (McGill, Grenier, Kavcic, & Cholewicki, 2003). Inaccurate force generation to perform sub-maximal tasks may lead to excessive forces applied to areas of the spine, thereby increasing spinal tissue load causing pain (Marras et al., 2001). This in turn has been associated with increased risk of developing future work related CLBP (Griffith et al., 2012). Thus, the assessment of lumbar extensor muscle force control is potentially important in clinical practice.

A recent study (Boucher et al., 2016) demonstrated that an 8-week lumbar muscle strengthening exercise program (e.g., side planks, bridge and quadruped leg extension/birddog) did not improve movement control or proprioception in people with CLBP. Similarly, the prescription of the same lumbar strengthening exercises did not result in acute changes in the ability of people with CLBP to activate transversus abdominis (Hall, Tsao, MacDonald, Coppieters,
Hodges, 2009) thereby suggesting the need for a specific motor control exercise to target motor control-related impairments. However before an evidence-based lumbar extensor muscle force control intervention is developed, the relationship between lumbar extensor muscle force control and self-reported disability has to be established (see section 2.6.3). This will be explored in Chapter 6.

4.7. Strength and limitations

The strengths of this study are as follow:

1. The study explored a new domain of neuromuscular assessment: precision of force production in the lumbar extensor muscles which could be used to differentiate people with CLBP from healthy people. To the author’s knowledge, this is the first study that has compared the ability of people with CLBP and healthy people to accurately match a variable submaximal force target; a task which may reproduce some of the demands of some types of work and activities of daily living. By utilising a relatively large sample size compared to some previous studies, the finding of this study have the potential to be better generalised to the wider population of people with CLBP.

2. This study developed a novel lumbar extensor muscle force control assessment to help produce new knowledge to the field of CLBP research. Importantly, the force control tests underwent a rigorous development process with input from clinicians, researchers and participants of a pilot study and underwent test-retest reliability testing. This rigorous approach allowed greater understanding of potential contributors to altered lumbar extensor muscle force output in this population.

3. The lumbar extensor force-matching protocol utilised a reproducible set-up and included a representative sample of patients with CLBP from a large physiotherapy clinic. The protocol provides insight into the relative contribution of the various systems involved in
producing accurate force, namely the muscular system, osteo-ligamentous system, central and peripheral nervous systems (see section 2.3) (Panjabi, 1992a; Reeves, Narendra, & Cholewicki, 2011; Sjolander et al., 2002).

4. The rigorous process in the development of the lumbar extensor muscle force control test ultimately led to a test which was safe and did not result in adverse events within the study population. This, would in turn, lead to the development of lumbar extensor force control assessment technique utilising commercially viable and cost effective equipment (e.g., gaming technology).

Albeit novel, this study also has several limitations:

1. This study results only pertain to a relatively younger CLBP cohort (mean age 41.8 ± 10.8) and as CLBP prevalence peaks at the age about 60 (Meucci, Fassa, & Faria, 2015), the results may not be generalisable to older individuals with CLBP.

2. The CLBP group exhibited quite low pain levels (NRS = 3.30 ± 1.8). Although this may be representative of the population attending physiotherapy clinics (Hoy et al., 2010), the study results may not be generalisable to CLBP individuals with higher levels of pain.

3. Almost 25% of the eligible participants were deemed ineligible by screening physiotherapists for muscle force control testing on the MedX lumbar dynamometer as they reported pain NRS ≥ 5/10. Although the study remained well powered, this may show that not every CLBP patient could be tested using this study methodology on the MedX lumbar dynamometer due to the position and restrain system applied on participants. Moreover, the MedX is a costly equipment and not widely available, thus the assessment protocol outlined in this study may not be easily generalisable to clinical practice.

4. Due to the isometric nature of the test, the results may not be transferable to isotonic muscle force control assessment. Similarly, the utilisation of continuous visual feedback in the test is not representative of real-life situations.
5. As there was no superficial or indwelling electromyography (EMG) utilised in this study, the existence of trunk muscle co-contraction of superficial trunk muscles (e.g., rectus abdominis and erector spinae) or deep trunk muscles (e.g., transversus abdominis and deep lumbar multifidus) (van Dieën et al., 2003) as a contributing factor to impairment trunk muscle force control in people with CLBP could not be confirmed. Thus, EMG assessment can provide the mechanism of trunk muscle force control; i.e., accuracy of trunk muscle force production in people with CLBP may occur as a result of trunk muscle co-contraction or in the absence of co-contraction, a result of visual feedback; i.e., participant saw his/her force on the tablet and then adjusted force production (Willigenburg et al., 2013; Willigenburg, Kingma, & van Dieen, 2010).

6. The cross-sectional nature of this study could not confirm whether or not the impairment of lumbar extensor muscle force control is a cause or effect of CLBP. Future prospective studies should be conducted to answer this question. Similarly, the relationship between lumbar extensor muscle force control variables and self-reported disability level has not been investigated and limits the utility of the test. This relationship will be investigated in chapter six.

7. This study did not analyse the psychosocial components CLBP (e.g., pain related fear). For instance, catastrophising behavior and kinesiophobia have been associated with lumbar muscle co-contraction and increased trunk stiffness in CLBP population (Karayannis, Smeets, van den Hoorn, & Hodges, 2013; van der Hulst, Vollenbroek-Hutten, Schreurs, Rietman, & Hermens, 2010) which could affect lumbar extensor muscle force control test results.

Future directions pertaining to this study will be discussed in section 7.2.
4.8. Conclusion

This study demonstrated that people with CLBP exhibit more error during a lumbar extensor force-matching task than healthy individuals. In particular, people with CLBP demonstrated difficulty producing an accurate increase in force output. This indicates that lumbar extensor muscle force control is impaired in people with CLBP. Interestingly, this phenomenon occurred independent of lumbar extensor muscle strength deficit. Further research is required to determine the functional relevance of diminished lumbar extensor muscle force control in CLBP population. The open kinetic chain variables used in this chapter will form the foundation for the analyses conducted in Chapter 6.
Chapter 5

Study 3

Coordination between the trunk and lower limb joints during symmetrical lifting in people with and without chronic low back pain

5.1. Introduction

Symmetrical lifting (see section 2.7) is a complex activity that requires sagittal plane restricted joint coordination that has been linked to initiation of CLBP (Coenen et al., 2014; Nelson et al., 1995). In clinical practice, deficits in lifting kinematics and kinetics can persist for many months (i.e., in CLBP > 3 months; see section 2.7.1). Most commonly, lifting kinematics are quantified by comparing the lumbar and hip ROM and angular velocity of people with and without CLBP (see section 2.7.1). However, previous studies assessing lumbar ROM during lifting in people with CLBP report inconsistent findings such as increased (McGregor et al., 1997), decreased (Sanchez-Zuriaga et al., 2011) or no difference (Lariviere et al., 2002) in lumbar ROM compared to healthy people. Likewise, comparisons of angular velocity are also equivocal (see section 2.7.1). Thus, a different type of kinematic analysis techniques may be required to accurately assess CLBP-related deficits during lifting.

A potential method to quantify lifting kinematics is to analyse the coordination between the trunk and lower limb joints during a standardised lifting task using relative phase angle analysis (Burgess-Limerick et al., 1993). This approach provides continuous spatial and temporal measurement throughout a movement cycle (see section 5.4.5 for detailed description). This is
due to the fact that phase angles are derived from joint displacement and joint velocity throughout the movement cycle (Hamill *et al.*, 1999). The relative phase angles are derived from subtracting the distal joint phase angle from the proximal joint phase angle. Positive values indicate that the proximal joint precedes the distal joint and vice versa. A value of zero indicates the joint couple movements are perfectly in-phase (i.e., synchronous) with each other (Burgess-Limerick *et al.*, 1993).

To the author’s knowledge, only one previous study has utilised this technique to compare inter-joint coordination of people with CLBP and healthy controls during trunk extension movement (Mokhtarinia *et al.*, 2016). Mokhtarinia *et al.* (2016) found that in people with CLBP, lumbar movement was more in-phase with hip movement compared to healthy people during trunk extension. However, several issues arose from this study. For instance, there were no measures of vertical GRF which, in conjunction to trunk kinematics, has been used to identify compensatory movement strategies during functional tasks in people with CLBP (Shum *et al.*, 2007b). Moreover, CLBP participants were almost completely pain free at the time of testing. Hence, it is unknown whether the novel assessment of joint coordination using relative phase angle analysis was reliable in people with CLBP. Additionally, the CLBP group were not disabled by their back symptoms. People who are more disabled by CLBP have demonstrated more severe kinematic and kinetic maladaptation’s during lifting – evident by reduced trunk ROM, lower limb ROM and vertical GRF’s through each leg – compared to those with lower disability level and healthy people (Sanchez-Zuriaga *et al.*, 2011). Thus, it is currently unknown whether CLBP individuals with higher disability levels would demonstrate different lifting-related trunk and lower limb joint coordination and vertical GRF’s to those with lower disability level and healthy people.
5.2. Aims
Thus, based on the aforementioned rationale, this study aimed to:

1. Investigate the reliability of lifting kinematic and kinetic assessment protocols (i.e., ROM, angular velocity, inter-joint coordination and vertical GRF assessments) in CLBP and healthy control participants during lifting.

2. Compare lifting-related kinematics (i.e., trunk ROM, lower limb ROM, angular velocity, trunk-lower limb inter-joint coordination) and kinetics (i.e., vertical GRF’s exerted by left and right legs) in CLBP and healthy control participants.

3. Compare lifting-related kinematics (i.e., trunk ROM, lower limb ROM, angular velocity, trunk-lower limb inter-joint coordination) and kinetics (i.e., vertical GRF’s exerted by left and right legs) in CLBP participants with higher and lower disability levels.

5.3. Hypotheses
Based on the previous equivocal literature (see sections 2.7.1, 2.7.2 and 2.7.3), the following hypotheses were adopted for the present study:

1. The kinematic (i.e., ROM, angular velocity and inter-joint coordination) and kinetic (i.e., vertical GRF) protocols would demonstrate good inter-session reliability in:
   a. CLBP group (H1a) (Mokhtarinia et al., 2016)
   b. Healthy control group (H1b) (Burgess-Limerick et al., 1993; Mokhtarinia et al., 2016; Sanchez-Zuriaga et al., 2011).

2. Compared to healthy controls, CLBP participants would demonstrate:
   
   With respect to trunk-lower limb lifting ROM and angular velocity-
   a. Smaller trunk and lower limb ROM (H2a) (Laird et al., 2014; Sanchez-Zuriaga et al., 2011)
b. (Sanchez-Zuriaga et al., 2011) Smaller trunk and lower limb angular velocity (H2b) (Laird et al., 2014; Sanchez-Zuriaga et al., 2011)

(Sanchez-Zuriaga et al., 2011) *With respect to trunk-lower limb inter-joint coordination variables* (see section 5.4.5)-

c. Different thorax-lumbar, lumbar-hip and hip-knee relative phase peak deviation (Dev) (H2c) (Kim et al., 2013; McClure et al., 1997).

d. Different thorax-lumbar, lumbar-hip and hip-knee coordination time (H2d) (Kim et al., 2013; McClure et al., 1997; Mokhtarinia et al., 2016).

e. Different thorax-lumbar, lumbar-hip and hip-knee relative phase area under curve (H2e) (Kim et al., 2013; McClure et al., 1997)

*With respect to kinetic variables-*

f. Less vertical GRF’s (H2f) (Sanchez-Zuriaga et al., 2011).

3. Within the CLBP group, compared to participants with lower disability, participants with higher disability would demonstrate-

*With respect to trunk-lower limb lifting ROM and angular velocity:*

a. Smaller trunk and lower limb ROM (H3a) (Laird et al., 2014; Sanchez-Zuriaga et al., 2011)

b. (Sanchez-Zuriaga et al., 2011) Smaller trunk and lower limb angular velocity (H3b) (Laird et al., 2014; Sanchez-Zuriaga et al., 2011)

(Sanchez-Zuriaga et al., 2011) *With respect to trunk-lower limb inter-joint coordination variables* (see section 5.4.5)-

c. Different thorax-lumbar, lumbar-hip and hip-knee relative phase peak deviation (H3c) (Kim et al., 2013; McClure et al., 1997).

d. Different thorax-lumbar, lumbar-hip and hip-knee coordination time (H3d) (Kim et al., 2013; McClure et al., 1997; Mokhtarinia et al., 2016).
e. Different thorax-lumbar, lumbar-hip and hip-knee relative phase area under curve (H3e) (Kim et al., 2013; McClure et al., 1997).

With respect to kinetic variables:

f. Less vertical GRF’s (H3f) (Sanchez-Zuriaga et al., 2011).

5.4. Materials and methods

5.4.1. Participants

The CLBP group consisted of 43 men and women aged 25-60 years with CLBP. In addition, a group of 29 healthy control participants (matched to the CLBP group for age, gender and BMI) with no history of CLBP were recruited. The CLBP cohort was divided into low and moderate-high disability sub-groups based upon their degree of self-reported ODI disability (Fairbank & Pynsent, 2000). Twenty-five CLBP participants had an ODI score of 20% or less and grouped under “low” disability group (CLBPlow). Similarly, 18 CLBP participants had an ODI score of above 20%; therefore, they were grouped under “moderate-high” disability group (CLBP_{high}). See section 3.4.1 for detailed participant recruitment process and sample size calculation and table 3.1 for eligibility criteria.

A subgroup of the CLBP (n = 17) and control groups (n = 16) were invited to attend a second session 5-7 days apart prior to their second physiotherapy session (i.e., prior to intervention) to complete a reliability study for the symmetrical lifting task protocol. This sample size is larger than a recently published reliability study in both CLBP and healthy controls (n = 12 per group) employing similar analytical methods for determination of inter-joint coordination of a symmetrical lifting task (Mokhtarinia et al., 2016).

5.4.2. Overview of experimental procedure

The protocol for the lifting task was developed by the PhD candidate with the assistance of research supervisors and was based on a previously published study (Sanchez-Zuriaga et al.,
2011). The protocol was refined by a pilot study involving ten healthy participants and five people with CLBP to determine the appropriate amount of weight lifted, repetitions and length of testing. Participants included in the pilot-testing meet the inclusion criteria for the study proper but were not included (see Appendix 3). Inter-session (between-day) reliability of lifting assessment protocol was assessed in both healthy and CLBP participants who were willing to be re-assessed within 5-7 days of initial testing prior to receiving treatment.

### 5.4.3. Experimental procedure

#### Self-reported disability and pain

All CLBP participants completed the ODI (see section 2.4.2) and NRS (see section 2.4.1) prior testing. Additionally, participants were also asked to rate their pain using the NRS before and after testing.

#### Instrumentation

Kinematic data were collected using a 12-camera Optitrack (Flex 13) (see figure 5.1A) motion analysis system (NaturalPoint, Corvallis, OR). Cameras were mounted on the ceiling of an 8 m x 4 m room (see figure 5.1B) and sampled at a frequency of 120 Hz. These 12 cameras were attached to two synchronised hubs connected to a laptop computer (HP Pavilion, Intel® Core™ i7, 2.20 GHz, 8 GB RAM) running Microsoft Windows 8.1.

Prior to each kinematic test session, each camera’s 3D position was calibrated. To do so, a T-wand with three reflective markers attached was waved within the motion capture area in a figure of eight pattern. After sufficient frame samples had been collected, the Optitrack Motive software constructed the 3D capture volume. This was done by simultaneously synchronising two dimensional images captured by each camera and associating the position of known calibration markers by triangulation. The mean calibration error was observed to determine the amount (in millimeters) the amount of re-projection error by the cameras. Based upon the mean error, the software rated the calibration results from poor to exceptional. The motion capture
system was re-calibrated when the mean calibration error exceeded 1 mm (Meldrum, Shouldice, Conroy, Jones, & Forward, 2014). The mean calibration error during this study was consistently below 0.5 mm. Following this, the motion capture system’s global coordinate was calibrated using an L-frame with three reflective markers attached to it placed in the motion capture area. This step calibrated the system’s sagittal, coronal and axial plane coordinate which would orientate the cameras to this reference plane (Optitrack users manual, http://wiki.optitrack.com/index.php?title=Calibration).

Kinetic data was collected using two 45 cm x 26.5 cm Wii Balance Boards (WBB’s; Nintendo, Kyoto, Japan). WBB’s were connected to the kinematic laptop computer via Bluetooth connection and interfaced using custom LabView 8.5 software (National Instruments). Prior to data collection, WBB’s were calibrated by placing several known loads at different positions on the WBB’s as described by Clark et al. (2010). Use of WBB’s for measurement of kinetic variables has been shown to be valid and reliable (Clark et al., 2010). Kinetic variables derived from the WBB’s included mean, minimum and maximum vertical GRF’s applied by the participants’ feet on each WBB (one foot on each board). The motion capture area is depicted in figure 5.1B.
Figure 5.1. Optitrack Flex 13 portable motion capture camera (A) and the motion capture setup (B) with three of the twelve cameras circled in red. Participants placed a foot on each of the two Nintendo WBB’s during the lifting task.

**Participant preparation**

All participants were instructed to wear shorts, males were topless and females wore a sports tank top to ensure that the skin overlying the key anatomical landmarks were exposed. Participants’ loose clothing was secured using non-reflective tape and footwear was removed before participants stepped on the WBB’s. A total of 24 retro-reflective markers of 13 mm diameter were attached to anatomical landmarks of each participant using double sided tape (see figure 5.2 and Appendix 4).

The trunk (*i.e.*, thorax and lumbar) segments were created using three markers per segment. The thorax segment was triangular in shape with its apex at the spinous process of T1 and the base spanning 4 cm; 2 cm from either side of T7 spinous process). Similarly, the lumbar segment was a triangle with its apex at the spinous process of L1 and the base length of 4 cm; 2 cm from either side of L3 spinous process. The thoracic-lumbar joint (thorax) was formed by the bisecting lines formed by T1-T7 and L1-L3. The lumbar-pelvis joint (lumbar) was formed by the bisecting lines of L1-L3 and a midpoint formed in between the ilia markers-S2. Similar
marker triads for the thoracic and lumbar segments have been previously utilised to assess inter-segmental spinal movement during complex trunk movements (Preuss & Popovic, 2010).

The Helen Hayes marker-set (bilateral anterior superior iliac spines (ASIS) and sacrum) was used on the pelvis to calculate the hip joint (Davis, Ounpuu, Tyburski, & Gage, 1991). An additional two markers on the ilia, on the mid-axillary lines were added for pelvis modelling due to a likely occlusion of the ASIS markers during dynamic tasks. As such, the ASIS markers were removed after the static trial (Collins, Ghoussayni, Ewins, & Kent, 2009) leaving the three markers: ilia and sacrum markers for dynamic testing. The head segment was a triangle with the base points at the temples and its apex at the greater occipital protuberance. However, analysis of the head segment is outside the scope of this thesis.

The thigh and shank segments each had three tracking markers on the lateral sides. Two markers on bilateral medial and lateral femoral epicondyles and medial and lateral malleoli were placed during the static trial to determine the joint centers and removed during the dynamic trial. The knee and ankle joint centers were defined as the midpoints between the medial and lateral femoral epicondyles and medial and lateral malleoli respectively (Collins et al., 2009; Schache, Baker, & Lamoreux, 2006).

Anatomical landmarks were carefully inspected and manually palpated prior to marker attachment (Schmid et al., 2015). Marker placement via palpation of anatomical landmarks has been demonstrated to be valid and reliable (Hidalgo, Gilliaux, Poncin, & Detrembleur, 2012). Marker placements are illustrated in figure 5.2 (a detailed anatomical description of these marker placements is summarised in Appendix 4).
Figure 5.2. Anatomical marker placements used in this study. Head markers are attached via a headband. Data related to the head segment is not discussed in this thesis. Markers attached to each of the body segments are colour-coded: red = head, green = thorax, yellow = lumbar spine, blue = pelvis, purple = thighs and orange = lower leg (see Appendix 4)

Static trial

A static trial was performed at the beginning of each assessment session. The aim of the static trial was to determine the axes of relevant body segments and joints based upon the marker placement. During the static trial, participants stood quietly with their arms folded in front of their chest and their hands touching the shoulders. Kinematic data was collected for two seconds and kinetic data were not recorded during static trial.

Symmetrical lifting protocol

The lifting task utilised in this study was based on a previously published study by (Sanchez-Zuriaga et al., 2011). Participants in the study by Sanchez-Zuriaga et al. (2011) were required to lift a 10 kg load off the ground to the level of their abdomen, turn to the right and place it on a table (i.e., asymmetrical lifting). Unlike Sanchez-Zuriaga et al. (2011), participants in the current study were required to lift an 8 kg weight from the floor to the abdomen level from an
upright standing position (i.e., symmetrical lifting; see section 2.7). Pilot testing revealed that 50% of participants with CLBP reported pain NRS ≥ 5/10 when lifting weights in excess of 8 kg (see Appendix 3). Moreover, a weight of 8 kg was selected because this is the average weight of a bag of groceries (Silvetti et al., 2015).

Participants were shown a video of the lifting task and were given a standardised instruction, “When you hear the signal, lift the weight from the floor as naturally as you normally would with both hands to the abdomen level with the elbows held high by your sides.” Following the explanation and video demonstration of the lifting task, participants were set up for a single practice trial. Prior to test commencement, participants were screened for baseline pain level and were instructed to inform the PhD candidate if pain severity increased during testing. If pain provocation occurred, the test ceased immediately.

Participants were instructed to step on the WBB’s, with the toes of each foot at the edge of the boards. Then, participants were asked to stand up relaxed, looking forward with their arms crossed in front of their chest with their hands touching their shoulders to minimise the influence of the upper limbs on bending. An 8 kg kettlebell was then placed between the WBB’s, 5 cm in front of the participants’ toes. The practice trial was not recorded and was used to familiarise the participants to the lifting task. None of the CLBP or control participants reported any changes in pain during or after the practice trial. Participants rested for thirty seconds after the practice trial prior to commencing the lifting trial proper. The symmetrical lifting setup was identical to the practice trial setup (i.e., one trial). Once the starting position had been adopted, the signal for the participant to commence lifting was given three seconds after data recording had begun. The lifting task is demonstrated in figure 5.3.
Figure 5.3. Participant performing symmetrical lifting task. 1: starting position; relaxed standing with arms crossed in front of the chest and hands touching the shoulder. 2: bending and lifting phase; participants were instructed to lift an 8 kg kettlebell as naturally as they could. 3: finished posture; the kettlebell was held in front of the abdomen with the elbows held high to prevent disruption to ilium markers data acquisition (Note: The table on the right of the participant was used for an asymmetrical lifting task that is not included in this thesis).

5.4.4. Data analysis

Cleaning of kinematic data

All kinematic data were manually cleaned. Each marker for each trial was individually named using the codes listed in Appendix 4. Gaps (i.e., missing frames) ≤ 10 frames in kinematic data resulting from marker dropouts were filled by selecting an adjacent marker belonging to the same body segment (i.e., similar trajectory) using the Optitrack Motive gap-filling algorithm. There were no data gaps of > 10 frames for any trials.

Kinematic data modelling, filtering and processing

Kinematic data was collected using Optitrack Motive (NaturalPoint, Corvallis, OR) software. The cleaned and gap-filled data from the Optitrack Motive software were then passed through a custom written analysis pipeline (Visual3D v5.01.6, C-Motion, Inc., Germantown, MD). The Visual3D custom pipeline (written by Mr. Benjamin Mentiplay, PhD candidate, Australian
Catholic University, Melbourne) extracted marker angular and velocity information. Visual3D was used to create a 3D model of the head, thorax, lumbar, pelvis, thigh and shank segments. Each segment was denoted by at least three tracking markers used during the dynamic trials (see section 5.4.3). Joint angle (i.e., ROM; °) and velocity (i.e., Vel = angular displacement divided by time; °/s) data were then derived using custom written LabVIEW 8.5 (National Instruments) software.

The joint angle data were then filtered using a fourth order zero-phase shift low-pass Butterworth filter. Filtering is required to remove noise from kinematic data (Kristianslund, Krosshaug, & van den Bogert, 2012). A frequency of 6 Hz has been used as a standard cut-off Butterworth filter frequency for joint angle (Bartlett, 2007). In this study however, significant noise still existed after filtering using a frequency of 6 Hz. When joint angle data was filtered using a 1 Hz cut-off frequency, it resulted in the least amount of noise and, in turn, resulted in a data profile similar to a previously published study with identical data analysis methodology (Burgess-Limerick et al., 1993). Thus, a 1 Hz filter was chosen as the cut-off frequency. Detailed steps on the selection of Butterworth filter cut-off frequency for velocity data is discussed in Appendix 5. Data were filtered using custom written software (Microsoft Visual Basic 7.1 for Microsoft Excel 2013) based on the calculations by Erer (2007). Kinematic variables from the data processing were angular displacement and angular velocity of:

- Thorax segment relative to lumbar segment (thorax ROM and thorax Vel),
- Lumbar segment relative to pelvis segment (lumbar ROM and lumbar Vel),
- Pelvis segment relative to femur segment (hip ROM and hip Vel) and
- Femur segment relative to shank segment (knee ROM and knee Vel).

The angular positions (°) of each joint (i.e., thorax, lumbar, hip and knee) at each frame of capture (1 frame = 1/120 seconds) were obtained and used for relative phase angle analysis. Additionally, the total time to complete the lifting task was calculated. As a convention, ROM
and velocity values decreased during flexion and increased during extension movements (i.e., during erect standing, hip angle is approximately 180°).

**Kinetic data filtering and processing**

Kinetic data from the WBB’s was sampled at 40 Hz and filtered using an eight order Butterworth filter with a low-pass cut-off frequency of 12 Hz which was adequate for the measurement of maximum, minimum and average body weight (Clark et al., 2010). Filtered data were then processed using a custom written LabVIEW 8.5 (National Instruments) software to derive maximum (F\text{Max}), minimum (F\text{Min}) and mean (F\text{Mean}) vertical GRF’s for the left (L) and right (R) foot in Newton (N). All vertical GRF data were then normalised to participant’s mass (kg).

**Relative phase angle analysis**

The methods of Burgess-Limerick et al. (1993; 1995) were used to calculate the degree of movement coordination between the trunk and joints of the lower limb joint during the lifting task(1993; 1995). Phase plane analysis involves plotting the angular position of a joint as a function of its angular velocity. The angular position and velocity are normalised to have a minimum value equal to -1 and maximum value equal to 1. Figure 5.4 illustrates a plot of normalised lumbar angular position as a function of normalised lumbar angular velocity. A positive clockwise convention was used in this study. Bending begins at the far right of the graph and ends on the far left where angular velocity is 0; i.e., the bottom half of the graph. Symmetrical lifting phase, the phase analysed in this study, begins at full flexion (far left where velocity = 0) and ends at full extension (far right where joint angle = 1, velocity = 0); i.e., the top half of the graph (see figure 5.4).
Figure 5.4. Lumbar angular velocity as a function of lumbar angular position during bending (flexion) and lifting (extension). The phase angle $\theta$ at any point of flexion and extension can be calculated using the formula $\tan^{-1}(\text{angular velocity/angular position})$ (Note: Only the extension phase is analysed in this study).

The phase angle ($\theta$) at any point in the movement (see figure 5.4), can be calculated using the formula $\tan^{-1}(\text{normalised angular velocity/normalised angular position})$. The phase angles of the thorax, hip and knee joints were calculated using the same methodology.

The term relative phase angle in this study is defined as the ‘phase angle of the proximal joint minus the phase angle of the distal joint.’ Analysis pairs the four joints (i.e., thorax, lumbar, hip and knee) into three joint couples (i.e., thorax-lumbar, lumbar-hip and hip-knee). A positive relative phase angle indicates that the proximal joint leads the distal joint during a movement phase and negative relative phase angle indicates that the distal joint leads the proximal joint. When the movement of a joint couple is perfectly in-phase, the relative phase angle is zero. Relative phase angles were plotted as a function of time (figure 5.5).

Clinically, it is important to differentiate the trunk into lumbar and thoracic segments (see section 2.6.1). Arguably, the measurement of the lumbar vertebral-hip joint couple by Burgess-Limerick et al. (1995) was more reflective of the measurement of relative phase angles between
the thorax and pelvis segments. To demonstrate this notion, a comparison between the analysis performed by Burgess-Limerick et al. (1993, 1995) and the current study was conducted (figures 5.5A and B).

Figure 5.5. Relative phase angle pattern between the proximal and distal joints in a healthy individual. Calculating the relative phase angles of the thorax relative to the hip (i.e., assuming the trunk moves a single segment on the pelvis) produces graph (A) which is comparable to Burgess-Limerick et al. (1993, 1995) studies. Separating the trunk into two segments (thoracic and lumbar segments) in the same individual results in graph (B). This suggests that the trunk does not behave as a single segment. The lumbar and thoracic spine behave differently and thus, their inter-joint relationship should be analysed separately.
Quantification of inter-joint coordination during lifting

As shown in figure 5.5B, during extension the relative phase angle of the thorax-lumbar joint couple (blue trace) is negative, indicating that the distal joint (i.e., lumbar) leads the proximal joint (i.e., thorax). Similarly, the knee (distal) leads the hip (proximal) joint (see grey trace). By contrast, the relative phase angle of the lumbar-hip joint couple (red trace) is positive, indicating that the lumbar spine leads the hip during lifting (i.e., proximal leads distal joint). This analysis can be simplified by calculating the peak deviation (Dev) from in-phase movement (i.e., the furthest degree of separation between the joints during movement; thorax-lumbar Dev, lumbar-hip Dev and hip-knee Dev) for each joint couple (Burgess-Limerick et al., 1995). Thus, the peak values for each joint couple can be positive or negative.

In this study, the time to reach the peak deviation for each joint couple was also calculated. This measurement indicates how long the joints of each couple stay in-phase with each other (Burgess-Limerick et al., 1995). In this study, inter-joint timing is quantified as thorax-lumbar time, lumbar-pelvis time and hip-knee time in seconds. This timing is normalised into percentage of total lifting time for each participant.

Lastly, unique to this study, the total deviation from in-phase movement throughout the extension phase was quantified by calculating the area under curve (AUC) of the relative phase angle curve for each joint couple. Asynchronous movements are characterised by a marked increase in angular movement path length over time (Sainburg et al., 1993). Thus, the measurement of AUC allows quantification of deviations from synchronous joint movements over time. AUC for each joint couple (thorax-lumbar AUC, lumbar-hip AUC and hip-knee AUC) was calculated using the integration formula:

\[ A = \int_{0}^{n} f(x) \, dx \]
Where:

\[ A = \text{area under the curve (°.frame)} \]

\[ 0 = \text{the first frame of extension phase (frame)} \]

\[ n = \text{the final frame of the extension phase (frame)} \]

\[ f(x) = \text{relative phase angle at time } x (°) \times \text{time (frame) as time difference approaches 0 (dx)} \]

Thus, as the AUC is the product of relative phase angle (°) and time (frame of capture; 1 frame = 1/120 seconds), its unit measurement is °.frame. Figure 5.6 illustrates inter-joint coordination variables utilised in this study (with the exception of the AUC). The AUC is normalised to total lifting time for each participant.

**Figure 5.6.** Inter-joint coordination variables of interest. The red circles are the peak (maximum/positive in the case for the lumbar-hip couple or minimum/negative for the thorax-lumbar and hip-knee couples) deviation from in-phase joint couple movements. The black arrows are the time from the start of extension to the points of maximum and minimum peak deviation (*Note: Area under curve (AUC) for each joint couple is not shown*).

In summary, patterns of inter-joint coordination during lifting can be quantified by calculating the following variables pertaining to the relative phase angle analysis:
1. The peak relative phase angle (Dev): the furthest deviation (positive or negative values) from in-phase joint couple movement. A positive value indicates the proximal joint leads the distal joint during a movement phase. A negative value indicates the distal joint leads the proximal joint during a movement phase.

2. Time to reach the peak relative phase angle (time): duration of in-phase joint couple movement.

3. Area under the relative phase angle curve (AUC): sum of deviations from in-phase joint couple movement throughout the lifting phase.

To ensure all inter-joint coordination data were processed correctly, the PhD candidate devised a custom-written spreadsheet using Microsoft Excel 2013 and a custom written macro program using Microsoft Visual Basic 7.1. Each participant’s data stream was manually graphed and analysed prior to statistical analyses. All the kinematic and kinetic variables and their respective units are summarised in table 5.1.
Table 5.1. Summary of the kinematic and kinetic variables used in this study.

<table>
<thead>
<tr>
<th>Variables</th>
<th>Units</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Kinematic variables</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thorax ROM</td>
<td>°</td>
<td>Range of motion of thoracic joint</td>
</tr>
<tr>
<td>Lumbar ROM</td>
<td>°</td>
<td>Range of motion of lumbar joint</td>
</tr>
<tr>
<td>Hip ROM</td>
<td>°</td>
<td>Range of motion of hip joint</td>
</tr>
<tr>
<td>Knee ROM</td>
<td>°</td>
<td>Range of motion of knee joint</td>
</tr>
<tr>
<td>Thorax Vel</td>
<td>°/s</td>
<td>Angular velocity of thoracic joint</td>
</tr>
<tr>
<td>Lumbar Vel</td>
<td>°/s</td>
<td>Angular velocity of lumbar joint</td>
</tr>
<tr>
<td>Hip Vel</td>
<td>°/s</td>
<td>Angular velocity of hip joint</td>
</tr>
<tr>
<td>Knee Vel</td>
<td>°/s</td>
<td>Angular velocity of knee joint</td>
</tr>
<tr>
<td>Thorax-lumbar Dev</td>
<td>°</td>
<td>Peak relative phase angle of thorax-lumbar joint couple</td>
</tr>
<tr>
<td>Lumbar-hip Dev</td>
<td>°</td>
<td>Peak relative phase angle of lumbar-hip joint couple</td>
</tr>
<tr>
<td>Hip-knee Dev</td>
<td>°</td>
<td>Peak relative phase angle of hip-knee joint couple</td>
</tr>
<tr>
<td>Thorax-lumbar time</td>
<td>s</td>
<td>Time to reach peak relative phase angle of thorax-lumbar joint couple</td>
</tr>
<tr>
<td>Lumbar-hip time</td>
<td>s</td>
<td>Time to reach peak relative phase angle of lumbar-hip joint couple</td>
</tr>
<tr>
<td>Hip-knee time</td>
<td>s</td>
<td>Time to reach peak relative phase angle of hip-knee joint couple</td>
</tr>
<tr>
<td>Thorax-lumbar AUC</td>
<td>°.frame</td>
<td>Total deviation from in-phase movement of thorax-lumbar joint couple</td>
</tr>
<tr>
<td>Lumbar-hip AUC</td>
<td>°.frame</td>
<td>Total deviation from in-phase movement of lumbar-hip joint couple</td>
</tr>
<tr>
<td>Hip-knee AUC</td>
<td>°.frame</td>
<td>Total deviation from in-phase movement of hip-knee joint couple</td>
</tr>
<tr>
<td><strong>Kinetic variables</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LF&lt;sub&gt;Max&lt;/sub&gt;</td>
<td>N</td>
<td>Maximum ground reaction force through left foot</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Max&lt;/sub&gt;</td>
<td>N</td>
<td>Maximum ground reaction force through right foot</td>
</tr>
<tr>
<td>LF&lt;sub&gt;Min&lt;/sub&gt;</td>
<td>N</td>
<td>Minimum ground reaction force through left foot</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Min&lt;/sub&gt;</td>
<td>N</td>
<td>Minimum ground reaction force through right foot</td>
</tr>
<tr>
<td>LF&lt;sub&gt;Mean&lt;/sub&gt;</td>
<td>N</td>
<td>Mean ground reaction force through left foot</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Mean&lt;/sub&gt;</td>
<td>N</td>
<td>Mean ground reaction force through right foot</td>
</tr>
</tbody>
</table>

° = degrees, °/s = degrees/second, s = seconds, °.frame = degrees.frame, N = Newton
5.4.5. Statistical analysis

Reliability testing

The inter-session (between-day) reliability was measured using a convenience sample of healthy controls (n = 16) and CLBP participants (n = 17). The participants for this reliability study were the same as those involved in the reliability study of the trunk muscle force control protocol (see section 4.4.5). This sample size is larger than a recently published reliability study (n = 12 per group) employing similar analytical methods for determination of inter-joint coordination of a symmetrical lifting task in both healthy and CLBP people (Mokhtarinia et al., 2016). Participants were invited to return to the clinic in order to complete a second test session within seven days of the initial test. As per assessment of reliability in the previous chapter (section 4.4.5), ICC_{3,k} with 95% CI was used and SEM was calculated. Although there is no consensus of the acceptable value of ICC, Chinn (1991) recommended a value of 0.6 for an outcome measure to be considered reliable.

Comparisons between chronic low back pain and control groups

For all closed kinetic chain variables, normality and equality of variances were analysed using Shapiro-Wilk and Levene median tests respectively. Normality was also assessed by observing the quartile-quartile and probability plots for each variable. The plots were used to determine normality if there was a discrepancy between Shapiro-Wilk and Levene test results. If the variables were normally distributed, they were presented as mean ± SD with α set at 0.05. Between-group comparisons of kinematic variables (see table 5.1) and the total time to complete the lifting task were assessed using one-way ANCOVA to control for differences in age between groups (section 3.5.2). All analyses were conducted with α set at 0.05. Additionally, within-group comparison of time to reach peak deviation for each joint couple (i.e., thorax-lumbar time, lumbar-hip time and hip-knee time) was analysed using one-way
ANOVA. Between-group kinetic variables were assessed using 2 (side; left and right) x 3 (group; control, CLBP_{low} and CLBP_{high}) factorial ANCOVA. In the event of a significant main effect, the effect size of was quantified using $\eta_p^2$ with values around 0.01 is classified as small, around 0.06 as medium and around 0.14 as large effect size (Cohen, 1988). Normality, homoscedasticity and linearity of residual for ANCOVA and ANOVA were assessed using Levene’s test and scatter graphs (Osborne & Waters, 2002; Williams et al., 2013). When indicated, pairwise comparisons were performed post-hoc using Tukey’s Honestly Significant Difference. All statistical analyses were conducted using SPSS Version 21.0 (IBM, Inc., Chicago, IL).

5.5. Results

5.5.1. Participant characteristics

Participant characteristics were outlined in section 3.5.2 and table 3.3. The CLBP_{high} group was significantly older than the control group (mean difference = 9.0 years, % difference = 24, $F_{2,69} = 3.5$, $p = 0.04$, 95% CI [0.8, 17.2]). There were no significant differences in age between CLBP_{high} and CLBP_{low} groups and between CLBP_{low} and control groups. Similarly, there were no significant differences between all participant groups with respect to weight, height, BMI and gender. The CLBP_{high} group reported significantly more disability than CLBP_{low} group (mean difference = 21.3, % difference = 62, $p < 0.001$, 95% CI [15.6, 27.0]) and control group (mean difference = 34.4%, % difference = 100, $p < 0.001$, 95% CI [30.7, 38.1]). Moreover, the CLBP_{high} group reported significantly more pain than CLBP_{low} group (mean difference = 1.5, % difference = 51, $p = 0.00$, 95% CI [0.4, 2.7]) and control group (mean difference = 4.5, % difference = 100, $p < 0.001$, 95% CI [3.7, 5.3]).
5.5.2. Reliability

All variables tested demonstrated moderate to excellent reliability with ICC’s ranging from 0.64 to 0.98. Inter-session ICC and SEM values for the lifting-related variables are summarised in table 5.2.
Table 5.2. Reliability of the variables for closed kinetic chain assessment in CLBP and control groups.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>Healthy controls (n = 16)</th>
<th>CLBP participants (n = 17)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ICC</td>
<td>SEM</td>
</tr>
<tr>
<td>Thorax ROM (°)</td>
<td>0.76</td>
<td>6.37</td>
</tr>
<tr>
<td>Lumbar ROM (°)</td>
<td>0.88</td>
<td>3.03</td>
</tr>
<tr>
<td>Hip ROM (°)</td>
<td>0.88</td>
<td>8.08</td>
</tr>
<tr>
<td>Knee ROM (°)</td>
<td>0.70</td>
<td>14.32</td>
</tr>
<tr>
<td>Thorax Vel (°/s)</td>
<td>0.86</td>
<td>16.02</td>
</tr>
<tr>
<td>Lumbar Vel (°/s)</td>
<td>0.91</td>
<td>5.15</td>
</tr>
<tr>
<td>Hip Vel (°/s)</td>
<td>0.92</td>
<td>14.38</td>
</tr>
<tr>
<td>Knee Vel (°/s)</td>
<td>0.81</td>
<td>20.71</td>
</tr>
<tr>
<td>Thorax-lumbar Dev (°)</td>
<td>0.73</td>
<td>65.27</td>
</tr>
<tr>
<td>Lumbar-hip Dev (°)</td>
<td>0.64</td>
<td>46.49</td>
</tr>
<tr>
<td>Hip-knee Dev (°)</td>
<td>0.89</td>
<td>37.73</td>
</tr>
<tr>
<td>Thorax-lumbar time (s)</td>
<td>0.67</td>
<td>0.41</td>
</tr>
<tr>
<td>Lumbar-Hip time (s)</td>
<td>0.85</td>
<td>0.20</td>
</tr>
<tr>
<td>Hip-knee time (s)</td>
<td>0.69</td>
<td>0.37</td>
</tr>
<tr>
<td>Thorax-lumbar AUC (°.frame)</td>
<td>0.65</td>
<td>17427.23</td>
</tr>
<tr>
<td>Lumbar-Hip AUC (°.frame)</td>
<td>0.73</td>
<td>5977.82</td>
</tr>
<tr>
<td>Hip-knee AUC (°.frame)</td>
<td>0.71</td>
<td>6458.69</td>
</tr>
<tr>
<td>LFMax (N)</td>
<td>0.96</td>
<td>2.28</td>
</tr>
<tr>
<td>RFMax (N)</td>
<td>0.84</td>
<td>4.47</td>
</tr>
<tr>
<td>LFSmin (N)</td>
<td>0.96</td>
<td>1.99</td>
</tr>
<tr>
<td>RFmin (N)</td>
<td>0.81</td>
<td>4.46</td>
</tr>
<tr>
<td>LFmean (N)</td>
<td>0.98</td>
<td>1.47</td>
</tr>
<tr>
<td>RFmean (N)</td>
<td>0.83</td>
<td>4.09</td>
</tr>
</tbody>
</table>

ICC = intraclass-correlation coefficient, SEM = standard error of measurement in the unit of the variable, ° = degrees, °/s = degrees/second, s = seconds, °.frame = degrees.frame, N = Newton.
5.5.3. Comparisons between chronic low back pain and control groups

Pain during testing

There were no adverse events reported during or following testing. No participants reported increased levels of pain during or after testing.

Time to complete lifting task

There was a significant main effect of group for total time to complete the lifting task ($F_{2,69} = 9.67, \eta^2_p = 0.22, p = 0.000$). Post-hoc comparisons demonstrated that the CLBP$_{\text{low}}$ (mean difference with control = 0.74 seconds, % difference = 39.57, $p = 0.003$, 95% CI [0.22, 1.26]) and CLBP$_{\text{high}}$ (mean difference with control = 0.94 seconds, % difference = 50, $p = 0.001$, 95% CI [0.37, 1.51]) groups took longer to complete the lifting task compared to the healthy group. There was no significant difference between CLBP$_{\text{low}}$ and CLBP$_{\text{high}}$ in total lifting time ($p = 0.71$).

Joint range of motion and angular velocity

After adjusting for age, ANCOVA analyses demonstrated no significant between-group differences in joint ROM and angular velocities during the lifting task. Descriptive data for trunk and lower limb ROM and angular velocity (mean ± SD) together with the results of statistical comparisons are presented in table 5.3.
Table 5.3. Descriptive data (mean ± SD) pertaining to trunk and lower limb ROM and angular velocity together with results of statistical analyses in CLBP and control groups.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP&lt;sub&gt;low&lt;/sub&gt; Mean ± SD</th>
<th>CLBP&lt;sub&gt;high&lt;/sub&gt; Mean ± SD</th>
<th>Control Mean ± SD</th>
<th>F&lt;sub&gt;2.69&lt;/sub&gt;</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax ROM (°)</td>
<td>35.05 ± 12.01</td>
<td>33.37 ± 10.72</td>
<td>36.17 ± 13.48</td>
<td>0.04</td>
<td>0.15</td>
</tr>
<tr>
<td>Lumbar ROM (°)</td>
<td>20.87 ± 9.49</td>
<td>26.20 ± 11.26</td>
<td>21.20 ± 8.94</td>
<td>1.60</td>
<td>0.21</td>
</tr>
<tr>
<td>Hip ROM (°)</td>
<td>87.06 ± 21.23</td>
<td>87.90 ± 6.23</td>
<td>83.77 ± 21.26</td>
<td>0.21</td>
<td>0.81</td>
</tr>
<tr>
<td>Knee ROM (°)</td>
<td>40.00 ± 27.80</td>
<td>50.17 ± 21.23</td>
<td>41.76 ± 28.08</td>
<td>0.57</td>
<td>0.57</td>
</tr>
<tr>
<td>Thorax Vel (°/s)</td>
<td>71.72 ± 27.29</td>
<td>59.21 ± 20.42</td>
<td>77.90 ± 44.11</td>
<td>0.83</td>
<td>0.44</td>
</tr>
<tr>
<td>Lumbar Vel (°/s)</td>
<td>36.39 ± 14.84</td>
<td>42.84 ± 19.12</td>
<td>45.00 ± 17.13</td>
<td>1.64</td>
<td>0.20</td>
</tr>
<tr>
<td>Hip Vel (°/s)</td>
<td>168.21 ± 62.90</td>
<td>133.59 ± 38.70</td>
<td>177.01 ± 57.33</td>
<td>0.39*</td>
<td>0.15</td>
</tr>
<tr>
<td>Knee Vel (°/s)</td>
<td>89.67 ± 50.97</td>
<td>85.25 ± 30.58</td>
<td>101.24 ± 60.53</td>
<td>1.94</td>
<td>0.68</td>
</tr>
</tbody>
</table>

SD = standard deviation, ROM = range of motion, Vel = angular velocity, ° = degrees, °/s = degrees/second, *F<sub>2.68</sub> age is a significant covariate

**Inter-joint coordination**

Statistical analysis revealed a significant main effect of group for thorax-lumbar time (F<sub>2.69</sub> = 3.94, $\eta^2_p = 0.10, p = 0.02$). Post-hoc analyses revealed that the CLBP<sub>high</sub> took significantly longer to reach peak relative phase angle of thorax-lumbar joint couple (i.e., longer in in-phase movement) than the control group (mean difference = 0.62 seconds, % difference = 70, $p = 0.019, 95\%$ CI [0.087, 1.16]). There were no significant differences between the CLBP<sub>high</sub> and CLBP<sub>low</sub> ($p = 0.14$) and CLBP<sub>low</sub> and healthy groups in thorax-lumbar time ($p = 0.65$). There was no significant main effects for lumbar-hip time and hip-knee time ($p > 0.05$). However, when normalised to total lifting time, there was no significant difference in thorax-lumbar time ($p = 0.23$), lumbar-hip time ($p = 0.36$) and hip-knee time ($p = 0.45$) between groups. There was no significant main effect of group for thorax-lumbar Dev (F<sub>2.67</sub> = 0.88, $p = 0.42$), lumbar-hip Dev (F<sub>2.67</sub> = 0.02, $p = 0.99$) and hip-knee Dev (F<sub>2.68</sub> = 0.0006, $p = 0.99$). Upon inspection of the raw data, one CLBP<sub>low</sub> participant lumbar data was over two SD’s from the mean thorax-lumbar Dev and lumbar-hip Dev. Although an inclusion of this participant’s lumbar data in the Dev analyses did not alter the significance of the results, it would negatively skew the Dev data
distribution. Thus, the lumbar data for this participant was excluded from the Dev analyses.

There was no significant difference of thorax-lumbar AUC, lumbar-hip AUC and hip-knee AUC for both normalised and non-normalised datasets ($p > 0.05$). Descriptive data for inter-joint coordination variables (mean ± SD) together with the results of statistical comparisons are presented in table 5.4. A sample relative phase angle pattern for a participant representative of each group is illustrated in figure 5.7.
Table 5.4. Descriptive data (mean ± SD) pertaining to inter-joint coordination variables together with results of statistical analyses in CLBP and control groups.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP&lt;sub&gt;low&lt;/sub&gt;</th>
<th>CLBP&lt;sub&gt;high&lt;/sub&gt;</th>
<th>Control</th>
<th>F&lt;sub&gt;2,68&lt;/sub&gt;</th>
<th>p values&lt;sup&gt;*&lt;/sup&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax-lumbar Dev (°)</td>
<td>-6.82 ± 111.34</td>
<td>-48.96 ± 65.59</td>
<td>-31.70 ± 121.54</td>
<td>0.88&lt;sup&gt;#&lt;/sup&gt;</td>
<td>0.42</td>
</tr>
<tr>
<td>Lumbar-hip Dev (°)</td>
<td>0.82 ± 138.60</td>
<td>-6.81 ± 54.26</td>
<td>0.33 ± 75.10</td>
<td>0.02&lt;sup&gt;#&lt;/sup&gt;</td>
<td>0.99</td>
</tr>
<tr>
<td>Hip-knee Dev (°)</td>
<td>-22.07 ± 240.60</td>
<td>-28.27 ± 136.70</td>
<td>-25.67 ± 148.18</td>
<td>0.006</td>
<td>0.99</td>
</tr>
<tr>
<td>Thorax-lumbar time&lt;sup&gt;a&lt;/sup&gt; (%)</td>
<td>41.69 ± 25.05</td>
<td>55.20 ± 27.98</td>
<td>47.00 ± 25.22</td>
<td>1.50</td>
<td>0.23</td>
</tr>
<tr>
<td>Lumbar-hip time&lt;sup&gt;a&lt;/sup&gt; (%)</td>
<td>39.71 ± 27.11</td>
<td>52.06 ± 28.05</td>
<td>46.43 ± 24.54</td>
<td>1.04</td>
<td>0.36</td>
</tr>
<tr>
<td>Hip-knee time&lt;sup&gt;a&lt;/sup&gt; (%)</td>
<td>38.68 ± 26.08</td>
<td>47.19 ± 34.82</td>
<td>34.16 ± 32.20</td>
<td>0.80</td>
<td>0.45</td>
</tr>
<tr>
<td>Thorax-lumbar time (s)</td>
<td>1.07 ± 0.67</td>
<td>1.52 ± 1.03</td>
<td>0.89 ± 0.59</td>
<td>3.94</td>
<td>0.02</td>
</tr>
<tr>
<td>Lumbar-hip time (s)</td>
<td>1.07 ± 0.82</td>
<td>1.30 ± 0.88</td>
<td>0.84 ± 0.49</td>
<td>2.24</td>
<td>0.11</td>
</tr>
<tr>
<td>Hip-knee time (s)</td>
<td>1.04 ± 0.80</td>
<td>1.17 ± 0.97</td>
<td>0.68 ± 0.66</td>
<td>2.54</td>
<td>0.09</td>
</tr>
<tr>
<td>Thorax-lumbar AUC&lt;sup&gt;a&lt;/sup&gt; (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>0.82 ± 1.61</td>
<td>0.58 ± 0.65</td>
<td>0.58 ± 0.80</td>
<td>0.38</td>
<td>0.69</td>
</tr>
<tr>
<td>Lumbar-hip AUC&lt;sup&gt;a&lt;/sup&gt; (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>0.88 ± 1.91</td>
<td>0.19 ± 0.13</td>
<td>0.34 ± 0.53</td>
<td>2.20</td>
<td>0.12</td>
</tr>
<tr>
<td>Hip-knee AUC&lt;sup&gt;a&lt;/sup&gt; (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>1.02 ± 2.04</td>
<td>0.44 ± 1.00</td>
<td>0.72 ± 1.08</td>
<td>0.72</td>
<td>0.49</td>
</tr>
<tr>
<td>Thorax-lumbar AUC (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>2.21 ± 4.50</td>
<td>1.55 ± 1.78</td>
<td>1.25 ± 2.31</td>
<td>0.64</td>
<td>0.53</td>
</tr>
<tr>
<td>Lumbar-hip AUC (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>2.43 ± 5.27</td>
<td>0.54 ± 0.49</td>
<td>0.62 ± 0.86</td>
<td>2.77</td>
<td>0.07</td>
</tr>
<tr>
<td>Hip-knee AUC (x 10&lt;sup&gt;4&lt;/sup&gt;.frame)</td>
<td>3.11 ± 6.61</td>
<td>0.65 ± 0.93</td>
<td>1.27 ± 1.93</td>
<td>2.20</td>
<td>0.12</td>
</tr>
</tbody>
</table>

SD = standard deviation, AUC = Area Under Curve, s = seconds, °.frame = degrees.frame, N = Newton. ° normalised to total lifting time, *p values for group main effects

*F<sub>2,67</sub> as one CLBP<sub>low</sub> participant lumbar data was excluded.
Figure 5.7. A sample outcome of relative phase pattern from someone from the CLBP<sub>low</sub>, CLBP<sub>high</sub> and control groups. In all groups, during extension (lifting), the lumbar spine leads the thoracic spine (negative peak relative phase angle), the lumbar spine leads the hip (positive peak relative phase angle) and the knee leads the hip (negative peak relative phase angle).
Vertical ground reaction forces

Descriptive data pertaining to kinetic variables (mean ± SD) together with the results of statistical comparisons are presented in table 5.5. There were no statistically significant between-group differences for any of the vertical GRF variables. Moreover, there were no significant side (left and right) differences for any vertical GRF variables.

Table 5.5. Descriptive data (mean ± SD) pertaining to kinetic variables derived from CLBP and control groups together with results of statistical analyses.

<table>
<thead>
<tr>
<th>Variables</th>
<th>CLBP&lt;sub&gt;low&lt;/sub&gt; Mean ± SD</th>
<th>CLBP&lt;sub&gt;high&lt;/sub&gt; Mean ± SD</th>
<th>Control Mean ± SD</th>
<th>F&lt;sub&gt;1,138&lt;/sub&gt; (side)</th>
<th>p values (side)</th>
<th>F&lt;sub&gt;2,138&lt;/sub&gt; (group)</th>
<th>p values (group)</th>
</tr>
</thead>
<tbody>
<tr>
<td>LF&lt;sub&gt;Max&lt;/sub&gt;</td>
<td>0.61 ± 0.05</td>
<td>0.62 ± 0.04</td>
<td>0.63 ± 0.04</td>
<td>1.43</td>
<td>0.23</td>
<td>1.00</td>
<td>0.37</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Max&lt;/sub&gt;</td>
<td>0.62 ± 0.05</td>
<td>0.61 ± 0.05</td>
<td>0.62 ± 0.05</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LF&lt;sub&gt;Min&lt;/sub&gt;</td>
<td>0.45 ± 0.04</td>
<td>0.46 ± 0.04</td>
<td>0.46 ± 0.03</td>
<td>0.25</td>
<td>0.62</td>
<td>0.003</td>
<td>0.997</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Min&lt;/sub&gt;</td>
<td>0.46 ± 0.04</td>
<td>0.45 ± 0.05</td>
<td>0.45 ± 0.05</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LF&lt;sub&gt;Mean&lt;/sub&gt;</td>
<td>0.53 ± 0.04</td>
<td>0.54 ± 0.04</td>
<td>0.54 ± 0.03</td>
<td>0.05</td>
<td>0.82</td>
<td>0.07</td>
<td>0.94</td>
</tr>
<tr>
<td>RF&lt;sub&gt;Mean&lt;/sub&gt;</td>
<td>0.54 ± 0.04</td>
<td>0.53 ± 0.04</td>
<td>0.53 ± 0.03</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

SD = standard deviation, L = Left, R = Right, F = vertical force, Max = maximum, Min = minimum. Kinetic variables are percentage of body weight.
5.6. Discussion

5.6.1. Overview of results

The main findings of this study included (i) assessment of inter-joint coordination and vertical GRF during symmetrical lifting can be reliably performed in people with CLBP and healthy people, and ii) CLBP groups demonstrated similar lifting strategies albeit took longer time to lift compared to healthy controls as per similar synchronous movement duration and movement sequence between the trunk and lower limb and comparable lifting-related vertical GRF’s. Detailed discussion of the study findings follows.

5.6.2. Reliability of the kinematic and kinetic assessment protocols

The lifting kinematic and kinetic assessment protocols are reliable (ICC >0.6) in both CLBP and healthy groups (Chinn, 1991) (accepted H1a and H1b). The ICC’s relating to the assessment of trunk ROM, angular velocity and vertical GRF during lifting in CLBP participants and healthy controls are comparable to previously reported data (ICC_{ROM} = 0.82-0.91; ICC_{Vel} = 0.73; ICC_{Force} = 0.94) (Sanchez-Zuriaga et al., 2011). To the author’s knowledge, this is the first study that has evaluated the inter-session reliability of relative phase angle analysis variables during a lifting task in detail. The SEM of the relative phase angle analysis variables was high in all participant groups but may, in part, be the result of a single trial being performed during testing. Burgess-Limerick et al. (1995) reported high within-session kinematic variability in the first ten out of 100 lifting trials performed in healthy participants. Based on pilot testing, performing more than ten lifting trials was not feasible given the physical (i.e., pain aggravation from CLBP) and logistical (i.e., time burden) demand on study participants (see Appendix 3).
5.6.3. Joint range of motion and angular velocity in chronic low back pain and healthy control groups

Analyses of joint ROM and angular velocity demonstrated no significant differences between CLBP and control groups and between both CLBP groups for all joints (rejected H2a, b and H3a, b). These findings are in agreement with those of Lariviere et al. (2002) who also found no significant differences in ROM and angular velocity for the trunk and lower limb between CLBP participants and healthy controls. However, the findings contradict those of Sanchez-Zuriaga et al. (2011) who demonstrated that CLBP patients exhibit decreased trunk ROM and angular velocity during lifting compared to controls (see section 2.6.1). Interestingly, the lifting-related thoracic and lumbar ROM’s reported by Sanchez-Zuriaga et al. (2011) are comparable to those exhibited by CLBP groups in the present study (thoracic ROM = 43.8 ± 18.8° vs. 36.2 ± 13.5° respectively and lumbar ROM = 26.9 ± 8.3° vs. 21.2 ± 8.9° respectively), indicating that control participants in this study lifted with less thoracic and lumbar ROM than those in Sanchez-Zuriaga et al. (2011). The large variation in pain free thoracic and lumbar ROM confirms that these fundamental kinematic variables are not sensitive in differentiating between people with and without CLBP during symmetrical lifting task.

Additionally, this study did not group CLBP participants into movement-based phenotypes (e.g., Sahrmann (2002), O'Sullivan (2005)). When a heterogenous CLBP participants were grouped using pain-provoking movement directions, trunk ROM differences between CLBP and healthy controls were revealed in tasks such as sitting (Dankaerts, O'Sullivan, Burnett, & Straker, 2006) and forward bending (Kim et al., 2013). Therefore, phenotyping CLBP participants based on their pain-provoking movement directions may be able to differentiate CLBP and healthy individuals during lifting.
5.6.4. Lifting inter-joint coordination between the trunk and lower limb and vertical ground reaction forces in chronic low back pain and healthy control groups

There were no significant differences in the peak relative phase angles, in-phase movement duration and AUC for all joint couples between all groups (rejected H2c, d, e and H3c, d, e). These kinematic findings are supported by the kinetic data that all groups exerted similar vertical GRF’s during lifting (rejected H2f and H3f). This suggests that the movement sequence or trunk-lower limb movement pattern during lifting is similar in people with and without CLBP. In particular, the findings of this study are partially supported by Burgess-Limerick et al. (1995) who reported that during lifting, the knee leads the hip. However, in contrast to Burgess-Limerick et al. (1995), the present study found that the lumbar spine leads the hip during lifting. The disparity between this study and that performed by Burgess-Limerick et al. (1995) is related to the separation of the trunk into the thoracic spine and lumbar spine in the current study (see section 5.4.5). Specifically, when the spinal regions were analysed separately, the “lumbar-vertebral”-hip joint couple reported by Burgess-Limerick et al. (1995) behaved differently to the lumbar-hip couple in this study (i.e., proximal leads distal joint versus distal leads proximal joint; see figure 5.5). The relevance of differentiating the trunk into thoracic and lumbar segments has been discussed in section 2.7.1. More importantly, these study findings suggest that during lifting, distal joints do not appear to consistently lead the proximal joints as previous studies have suggested (Burgess-Limerick et al., 1995; McClure et al., 1997). Consistent with previous literature (Potvin et al., 1991), each joint couple (i.e., thorax-lumbar, lumbar-hip and hip-knee) reached maximum deviation from in-phase movement at approximately the same time, confirming the notion that during lifting, joint movements occur contemporaneously (Potvin et al., 1991). Interestingly, this pattern of joint coordination is demonstrated in each
research group; hence, it is apparent that CLBP – with varying degrees of severity – does not alter patterns of lifting-related movement.

There are several possibilities as to why this study failed to demonstrate between-group differences in lifting-related kinematics and kinetics. Firstly, the load lifted in this study was relatively low. In this respect, Burgess-Limerick et al. (1995) demonstrated increased asynchronous movement between the trunk and hip with increasing lifting load (i.e., up to 12.5 kg) in healthy participants. Therefore, it is highly plausible CLBP individuals would demonstrate a different lifting pattern with higher loads. Secondly, this study assessed CLBP participants in a non-fatigued state. Lumbar extensor muscle fatigue has been demonstrated to increase lifting-related synchronous movement between the trunk and lower limb joints in healthy people – a compensatory response thought to decrease the risk of injury (Hu & Ning, 2015). As increased lumbar extensor fatigability has been reported in people with CLBP (Kankaanpaa, Taimela, Laaksonen, Hanninen, & Airaksinen, 1998), testing in a fatigue condition could accentuate changes in trunk-lower limb lifting coordination in CLBP individuals. Finally, this study involved analysis only in a single plane (i.e., sagittal) during a symmetrical lifting task. Impairment in lumbo-pelvis ROM in the transverse plane have been reported during walking in CLBP people compared to healthy controls (Crosbie, de Faria Negrao Filho, Nascimento, & Ferreira, 2013). Similarly, asymmetrical lifting tasks (i.e., lifting with trunk rotation) have been shown to differentiate between people with and without CLBP with respect to kinematic and kinetic variables (Sanchez-Zuriaga et al., 2011). Thus, between-group kinematic and kinetic differences could be augmented when multi-planar tasks are utilised.

5.6.5. Clinical implications

Participants with CLBP in this study were able to complete the lifting task successfully with no adverse events or increase in pain intensity. Moreover, CLBP participants demonstrated a
similar lifting pattern to the control group, suggesting that lifting-related motor programming is unchanged with CLBP. However, people with CLBP took longer to lift an 8 kg weight than healthy controls. Although it is currently unknown whether the lifting pattern identified in this study was a result or the cause of CLBP, increased lifting duration in people with CLBP compared to healthy controls is indicative of a compensatory neuromuscular strategy to minimise the likelihood of pain provocation (O'Sullivan, 2005). Increased lifting duration has been associated with increased spinal loading and is potentially injurious in people with compromised spinal health such as in CLBP patients (see section 2.7). This speculation supports the concept that an “optimum” lifting technique is dictated by both individual (e.g., spinal health) and environmental factors (e.g., load lifted, load position) (Dreischarf, Rohlmann, Graichen, Bergmann, & Schmidt, 2016; Kingma et al., 2004; van Dieen et al., 1999). With respect to the current study findings, clinicians could aim to decrease lifting-related spinal load in people with CLBP by decreasing lifting duration (i.e., instruction to lift briskly), decrease synchronous movement duration between the hip and knee and increase peak force exerted by the legs (i.e., an instruction to lift with the legs) (Kingma et al., 2004).

It is currently unknown whether lifting-related kinematic and kinetic variables analysed in this study are associated with self-reported disability of individuals with CLBP. For instance, increased movement coordination between the trunk and lower limb or ‘phase-locked’ (i.e., more rigid) movement pattern (Schoner, 1990) may not be a positive feature and is likely related to decreased movement variability through the trunk (i.e., guarded trunk movement or dyskinesia) which, in turn, contributes to increased trunk displacement (Mok et al., 2007). A common contributing factor to decreased trunk movement variability is trunk flexor (i.e., rectus abdominis, internal and external obliques) and extensor (i.e., longissimus, iliocostalis and multifidus) co-contraction (Radebold et al., 2000; Radebold et al., 2001; van Dieen et al., 2003; van Dieën et al., 2003). This, in turn, could be associated with increased spinal compressive
load (Ferguson, Marras, Burr, Davis, & Gupta, 2004; Greenland et al., 2013) and may increase the risk of developing future low back disorders (Mehta, Lavender, & Jagacinski, 2014). Thus, the relationship between disability and lifting kinematic and kinetics will be the focus of the subsequent chapter.
5.7. Strengths and limitations

The strengths of this study include:

1. To the author’s knowledge, this is the first study to utilise in-depth analysis of inter-joint coordination using relative phase angle analysis in combination with vertical GRF data to quantify lifting patterns in people with CLBP during a symmetrical lifting task. The symmetrical lifting protocol and analyses utilised in this study were developed based on past studies (Burgess-Limerick et al., 1993; Sanchez-Zuriaga et al., 2011), pilot and reliability testing. Thus, this rigorous process contributed new knowledge on the inter-relationship between trunk and lower limb joint movement during lifting.

2. This is also the first study to incorporate assessment of inter-joint coordination between thoracic and lumbar spine using relative phase angle analysis. By differentiating the trunk into thoracic and lumbar spine, this study found that the inter-relationship between the trunk and lower limb joints is not uniform during lifting.

Albeit utilising several novel analysis strategies, this study had several limitations:

1. Results only pertain to a relatively young CLBP cohort (mean age ranging from 42.3 to 46.7 years) and as back pain prevalence peaks at the age of 80 (Fairbank & Pynsent, 2000; Hoy et al., 2014), the results may not be generalisable to older CLBP population.

2. Due to the cross-sectional nature of this study, it is not possible to confirm whether or not the lifting patterns exhibited in this study were a cause or effect of CLBP. Future prospective studies should be conducted to answer this question.

3. Although the relationship between basic kinematic measures such as joint ROM and self-reported disability level has been studied previously, the correlation between them was low (Parks et al., 2003). The relationship between kinematic and kinetic variables utilised in this study and disability level has not been studied. Therefore, this may limit the clinical applicability of the assessment. This will be investigated in Chapter 6.
4. The CLBP participants in this study were not phenotyped based on movement patterns. Past studies suggest that significant kinematic differences exist when CLBP participants were divided based upon their pain provoking movements (Dankaerts et al., 2006; O'Sullivan, 2005; Sheeran, Sparkes, Caterson, Busse-Morris, & van Deursen, 2012; Van Dillen et al., 2016). Not sub-classifying the CLBP participants using movement-based phenotypes could have resulted in the small number of kinematic variables reaching statistical significance.

5. No EMG was used in this study. Recent studies suggest that although participants with CLBP were able to complete lifting tasks as well as healthy controls, there were significant differences in the muscle activation and loading patterns based on EMG data (Nelson-Wong, Alex, Csepe, Lancaster, & Callaghan, 2012). The utilisation of EMG could therefore be able to explain the lack of discrepancies in kinematic variables between the clinical groups (Ferguson et al., 2004).

6. This study only utilised a single lift to minimise the risk of pain provocation and burden of assessment on the participant. However, a single lift may not have the predictive capacity of multiple lifts partly due to the variations and consistencies within each individual (Burgess-Limerick et al., 1995; Granata, Marras, & Davis, 1999). Additionally, a single lift trial may not be adequate to predict or screen for future lifting injuries (Faber et al., 2011; Latikka, Battie, Videman, & Gibbons, 1995). Thus, it remains unknown whether repetition of the lifting task may results in variable movement strategies or not.

7. This study did not assess the psychosocial factors associated with persistent pain in the CLBP group. Pain behaviour may also result in changes in trunk muscle activation which could be related to the changes in CLBP group kinematic and kinetic data (Karayannis et al., 2013).

Future directions pertaining to this study will be discussed in section 7.2.
5.8. Conclusion

A comprehensive investigation of lifting biomechanics in people with CLBP using relative phase angle analysis was performed in this chapter. The findings of this study suggest that both CLBP and healthy individuals exhibit a similar lifting profile as demonstrated by comparable trunk and lower limb joint inter-coordination, in-phase movement duration and vertical GRF’s. However, CLBP participants lifted slower than healthy controls. It is unknown whether the lifting-related kinematics and kinetics are associated with self-reported disability of CLBP participants. Thus, the relationship between disability level and lifting inter-joint coordination and vertical GRF variables will be addressed in the next chapter.
Chapter 6

Study 4

Relationships between open and closed kinetic chain neuromuscular variables and disability level in people with chronic low back pain

6.1. Introduction

A large proportion of people in industrialised countries – 3.7 million Australians (16% of the population) (AIHW, 2015), suffer from CLBP (Buchbinder et al., 2013). It has been estimated that 12% of those with CLBP in Western countries are disabled by it (Balague, Mannion, Pellise, & Cedraschi, 2012). Self-reported disability – usually assessed using a questionnaire such as the ODI (Fairbank & Pynsent, 2000) – has been shown to be a strong predictor of whether people seek medical intervention for CLBP, more so than pain intensity alone (Ferreira, Machado, et al., 2010). Thus, improvement in disability is often the aim of CLBP treatment (Sowden et al., 2012).

Related to self-reported disability, neuromuscular impairments also represent an important dimension of CLBP (Waddell, 2006). Impairments in neuromuscular function assessed via open kinetic chain movements present as alterations in trunk muscle strength (Steele et al., 2014) and/or force control (see section 2.6). Closed kinetic chain-related neuromuscular impairments of CLBP individuals are reflected in decreased ROM and angular velocity of trunk and lower limb joints (Sanchez-Zuriaga et al., 2011) together with changes in trunk and lower limb joint coordination and decreased ability controlling vertical GRF’s during lifting (see section 2.7).

The association between disability and trunk muscle strength and trunk ROM in people with
CLBP ranges from non-existent to moderate (see section 2.8) (Mannion, Junge, et al., 2001; Nattrass et al., 1999; Parks et al., 2003). It is possible that previous experimental procedures used to elucidate changes in muscle function and lifting mechanics are not sensitive enough detect subtle CLBP-related neuromuscular mal-adaptations. However, the validity of the novel open and closed kinetic chain neuromuscular assessment variables have not been investigated. Hence, it is important to quantify the relationship between self-reported levels of CLBP-related disability and novel measures of lumbar extensor muscle force control (Study 2) and trunk-lower limb joint coordination and vertical GRF’s during lifting (Study 3).

6.2. Aims

Thus, based on the abovementioned rationale, this study aimed to:

1. Investigate the relationship between open kinetic chain neuromuscular variables: lumbar extensor MVIC, RMSE_T, RMSE_A and RMSE_D (see section 4.4.4) and self-reported disability level (bivariate analyses) of CLBP participants.

2. Investigate the relationship between closed kinetic chain neuromuscular variables: trunk and lower limb ROM, angular velocity, trunk-lower limb inter-joint coordination variables and vertical GRF variables (see section 5.4.4) and self-reported disability level (bivariate analyses) of CLBP participants.

3. Investigate whether self-reported disability levels of those with CLBP could be predicted by a combination of open and closed kinetic chain variables (multivariate analysis).
6.3. Hypotheses

Based on the previous equivocal literature (see section 2.8), the following hypotheses were adopted for the present study:

1. Regarding the relationship between self-reported disability and open kinetic chain variables:
   a. Lumbar extensor MVIC would not be significantly associated with self-reported disability (H1a) (Al-Obaidi et al., 2000).
   b. RMSE variables (RMSEA, RMSED, RMSET) would be significantly and positively associated with self-reported disability (H1b) (Ferreira, Ferreira, et al., 2010).

2. Regarding the relationship between self-reported disability and closed kinetic chain variables:
   a. ROM and angular velocity would not be significantly associated with self-reported disability (H2a) (Parks et al., 2003; Poitras, Loisel, Prince, & Lemaire, 2000).
   b. Lifting inter-joint coordination variables would be significantly and positively associated with self-reported disabilty (H2b) (Sanchez-Zuriaga et al., 2011).
   c. Lifting kinetic variables would be significantly and negatively associated with self-reported disabilty (H2c) (Sanchez-Zuriaga et al., 2011).

3. Self-reported disability level of CLBP participants will be predicted using a combination of novel open (i.e., RMSE variables) and closed kinetic chain (i.e., inter-joint coordination and kinetic variables) neuromuscular variables (H3) (Mannion, Junge, et al., 2001).
6.4. Materials and methods

6.4.1. Participants

Thirty-three men and women aged 25-60 years with CLBP who completed studies two and three were enrolled in this study. See section 3.4.1 for detailed participant recruitment process and table 3.1 for eligibility criteria.

6.4.2. Experimental procedure

Details of the experimental protocol for assessing lumbar extensor strength and lumbar extensor muscle force control have been described previously (see section 4.4). Likewise, the methodology for the assessment of trunk and lower limb ROM, angular velocity, inter-joint coordination and vertical GRF’s during lifting was detailed in the respective chapter (see section 5.4). In summary, lumbar extensor strength and sub-maximal muscle force control assessment was conducted using the MedX dynamometer (Ocala, FL). Lumbar extensor muscle force control was assessed using a submaximal force target that oscillated between 20%-50% lumbar extensor MVIC. Lifting ROM and angular velocity data were obtained using marker-based assessment using optoelectric equipment (Optitrack Flex 13, NaturalPoint, Corvallis, OR). Vertical GRF data were obtained using two WBB’s (Nintendo, Kyoto, Japan). Trunk and lower limb inter-joint coordination was analysed using relative phase angle analysis.

6.4.3. Statistical analysis

Bivariate relationships between neuromuscular variables and disability level

Linearity and strength of bivariate relationships between candidate predictors (i.e., neuromuscular variables (see sections 4.4.4 and 5.4.4) and dependent variable (i.e., ODI) were analysed using the Pearson product-moment correlation coefficient and scatter graphs. A correlation is considered weak when the correlation coefficient (r) is between 0.00 and 0.25, fair when r is between 0.25 and 0.50, moderate when r is between 0.50 and 0.75 and excellent when r is larger than 0.75 (Portney & Watkins, 2009).
H1a, b (i.e., relationship between self-reported disability and open kinetic chain variables) were assessed using linear regression after quantifying the strength of bivariate relationships between the open kinetic chain neuromuscular variables (sections 4.4.4) and the ODI as a continuous scale. Similarly, H2a, b, c (i.e., the relationship between self-reported disability and closed kinetic chain variables) were investigated using linear regression after confirming the strength of bivariate relationships between the kinematic and kinetic variables (section 5.4.4 and the ODI scores as a continuous scale. As peak deviation of the relative phase angle curve for each joint couple could be positive or negative (Dev; see section 5.4.4), this data was converted into absolute values. As there were no significant left and right side differences for lifting-related GRF’s (see section 5.5.3), average bilateral values were used for statistical analyses. Results of the bivariate relationships were presented using correlation matrices. Open and closed neuromuscular variables that exhibited a significant correlation with the ODI were included within the multivariate linear regression model as candidate predictor variables to investigate H3 (Osborne & Waters, 2002). The number of candidate predictors in this study was determined using G*Power 3.1.9.2. Based on the acquired sample size of 33 and with α set at 0.05 and power (1-β) set at 0.80, a three predictor model could adequately detect an R-squared (R²) of 0.50 (Faul et al., 2009).

**Evaluation of the multiple linear regression model**

Collinearity of the model predictors was assessed using variance inflation factor and tolerance level. Variance inflation factor close to 10 and tolerance value close to 0.1 are suggestive of collinearity (Craney & Surles, 2002; Tabachnick & Fidell, 2001). Normality of the residuals were analysed using Shapiro-Wilk test, histogram, quartile-quartile and probability plots. Homoscedasticity and linearity of residuals were assessed using Levene’s test and scatter graphs (Osborne & Waters, 2002; Williams et al., 2013). An α level of 0.05 was used as the
level of significance in all analyses. All analyses were conducted using SPSS Version 21.0 (IBM, Inc., Chicago, IL).

6.5. Results

6.5.1. Relationships between open kinetic chain neuromuscular candidate predictors and disability

Results of the bivariate correlations between candidate predictors of open kinetic chain variables and disability level are summarised in table 6.1. Only RMSE_D was significantly correlated (moderately and positively) with ODI in CLBP participants ($r = 0.47, p = 0.006$). The other open kinetic chain neuromuscular variables including RMSE_A, RMSE_T and lumbar extensor MVIC were not significantly correlated with self-reported disability ($p > 0.05$). Not surprisingly, trunk muscle force control variables (RMSE_A, RMSE_D and RMSE_T) were significantly correlated with each other.

<table>
<thead>
<tr>
<th>Variables</th>
<th>ODI</th>
<th>LE Strength</th>
<th>RMSE_A</th>
<th>RMSE_D</th>
<th>RMSE_T</th>
</tr>
</thead>
<tbody>
<tr>
<td>LE strength</td>
<td>0.29</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>RMSE_A</td>
<td>0.27</td>
<td>-0.11</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>RMSE_D</td>
<td>0.47*</td>
<td>-0.08</td>
<td>0.59*</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>RMSE_T</td>
<td>0.28</td>
<td>-0.25</td>
<td>0.71*</td>
<td>0.60*</td>
<td>-</td>
</tr>
</tbody>
</table>

ODI = Oswestry Disability Index, LE strength = isometric lumbar extensor strength, RMSE_A = average root mean square error during ascending phase, RMSE_D = average root mean square error during descending phase, RMSE_T = total average root mean square error. *$p < 0.01$

As only RMSE_D was significantly correlated with ODI score, a single predictor model was used to predict self-reported disability level. In isolation, RMSE_D significantly predicted 19% of the variance in self-reported disability (adjusted $R^2 = 0.19$, standardised $\beta = 0.47$, $p = 0.006$). Thus, a one unit increase in RMSE_D is associated with an increase in the ODI score of 47%; that is,
less-accurate lumbar extensor force output was significantly associated with increasing
disability.

6.5.2. Relationships between closed kinetic chain neuromuscular
candidate predictors and disability

Results of the bivariate correlations between candidate predictors of ROM and angular velocity
and disability level are summarised in table 6.2. There were no significant correlations between
either joint ROM and angular velocity and ODI. Therefore, these closed kinetic chain
neuromuscular variables were not included in the multivariate model.

Results of the bivariate correlations between candidate predictors of inter-joint coordination
variables and disability level are summarised in table 6.3. There was a significant, moderate
positive correlation between thoracic-lumbar in-phase time and ODI (thorax-lumbar time $r = 0.47$, $p = 0.006$). The correlation between lumbar-hip in-phase time and ODI was approaching
significance (lumbar-hip time $r = 0.32$, $p = 0.072$).

Results of the bivariate correlations between candidate predictors of vertical GRF variables and
disability level are summarised in table 6.4. Two GRF variables demonstrated significant,
moderate negative correlation with ODI ($F_{\text{Max}} r = -0.43$, $p = 0.012$; $F_{\text{Mean}} r = -0.42$, $p = 0.014$)
whereas $F_{\text{Min}}$ was approaching statistical significance ($r = -0.31$, $p = 0.076$).

As the multivariate regression analysis was only powered to include three predictor variables,
the three closed kinetic chain variables with the strongest correlation with the ODI were
selected. These variables were thorax-lumbar time, $F_{\text{Max}}$ and $F_{\text{Mean}}$. The correlations between
thorax-lumbar time with $F_{\text{Max}}$ ($r = -0.49$, $p = 0.004$) and $F_{\text{Mean}}$ ($r = -0.47$, $p = 0.005$) were
significant, moderate and negative. Unsurprisingly, correlations between $F_{\text{Max}}$ and $F_{\text{Mean}}$ were
significant and strong ($r = 0.99$, $p < 0.001$). Correlation coefficients between the three candidate
predictor variables were higher than the a priori cutoff $r$ value of 0.40 derived from a study
utilising similar analysis (Sanchez-Zuriaga et al., 2011). When any two or three variables were
included in the multiple regression model, none of the predictors reached statistical significance. Hence, only the neuromuscular variable with the strongest correlation with the ODI was selected (i.e., thorax-lumbar time) for the linear regression model. Thorax-lumbar time alone predicted 19% the variance in the ODI score (adjusted $R^2 = 0.19$, standardised $\beta = 0.47$, $p = 0.001$). Thus, a one second increase in in-phase movement between the thorax and lumbar spine was associated with a 47% increase in the ODI score.
### Table 6.2. Bivariate relationships between range of motion and angular velocity and disability level of CLBP participants

<table>
<thead>
<tr>
<th>Variables</th>
<th>ODI</th>
<th>Thorax ROM</th>
<th>Lumbar ROM</th>
<th>Hip ROM</th>
<th>Knee ROM</th>
<th>Thorax Vel</th>
<th>Lumbar Vel</th>
<th>Hip Vel</th>
<th>Knee Vel</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax ROM</td>
<td>0.03</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Lumbar ROM</td>
<td>0.05</td>
<td>-0.29</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hip ROM</td>
<td>0.02</td>
<td>0.46**</td>
<td>0.13</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Knee ROM</td>
<td>0.05</td>
<td>0.04</td>
<td>0.06</td>
<td>0.60**</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Thorax Vel</td>
<td>-0.20</td>
<td>0.51**</td>
<td>-0.23</td>
<td>0.21</td>
<td>-0.004</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Lumbar Vel</td>
<td>-0.07</td>
<td>-0.29</td>
<td>0.85**</td>
<td>0.16</td>
<td>0.05</td>
<td>0.03</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hip Vel</td>
<td>-0.32</td>
<td>-0.16</td>
<td>0.07</td>
<td>0.40*</td>
<td>0.43*</td>
<td>0.48**</td>
<td>0.39*</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Knee Vel</td>
<td>-0.20</td>
<td>-0.25</td>
<td>-0.03</td>
<td>0.28</td>
<td>0.80**</td>
<td>0.08</td>
<td>0.12</td>
<td>0.64**</td>
<td>-</td>
</tr>
</tbody>
</table>

ODI = Oswestry Disability Index, ROM = Range of motion, Vel = Angular velocity. *p < 0.05, **p < 0.01

### Table 6.3. Bivariate relationships between inter-joint coordination variables and disability level of CLBP participants

<table>
<thead>
<tr>
<th></th>
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<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax-lumbar Dev</td>
<td>-0.06</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Lumbar-hip Dev</td>
<td>-0.14</td>
<td>0.92**</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hip-knee Dev</td>
<td>-0.11</td>
<td>0.05</td>
<td>0.23</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Thorax-lumbar time</td>
<td>0.47**</td>
<td>0.22</td>
<td>0.25</td>
<td>0.27</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Lumbar-hip time</td>
<td>0.32</td>
<td>0.21</td>
<td>0.22</td>
<td>0.37*</td>
<td>0.51**</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hip-knee time</td>
<td>0.17</td>
<td>-0.13</td>
<td>-0.10</td>
<td>0.24</td>
<td>0.40*</td>
<td>0.53**</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Thorax-lumbar AUC</td>
<td>-0.02</td>
<td>0.77**</td>
<td>0.67**</td>
<td>0.09</td>
<td>0.21</td>
<td>0.30</td>
<td>-0.10</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Lumbar-hip AUC</td>
<td>-0.13</td>
<td>0.73**</td>
<td>0.84**</td>
<td>0.33</td>
<td>0.29</td>
<td>0.29</td>
<td>-0.09</td>
<td>0.83**</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hip-knee AUC</td>
<td>-0.17</td>
<td>0.07</td>
<td>0.23</td>
<td>0.95**</td>
<td>0.22</td>
<td>0.38*</td>
<td>0.21</td>
<td>0.20</td>
<td>0.39*</td>
<td>-</td>
</tr>
</tbody>
</table>

ODI = Oswestry Disability Index, Dev = Absolute peak relative phase angle, Time = Time to reach absolute peak relative phase angle, AUC = Area under relative phase angle curve. *p < 0.05, **p < 0.01
Table 6.4. Bivariate relationships between kinetic variables and disability levels of CLBP participants

<table>
<thead>
<tr>
<th>Variables</th>
<th>ODI</th>
<th>F_{Max}</th>
<th>F_{Min}</th>
<th>F_{Mean}</th>
</tr>
</thead>
<tbody>
<tr>
<td>F_{Max}</td>
<td>-0.41*</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>F_{Min}</td>
<td>-0.31</td>
<td>0.93**</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>F_{Mean}</td>
<td>-0.39*</td>
<td>0.99**</td>
<td>0.97**</td>
<td>-</td>
</tr>
</tbody>
</table>

ODI = Oswestry Disability Index, F_{Max} = Maximum vertical ground reaction force, F_{Min} = Minimum vertical ground reaction force, F_{Mean} = Mean vertical ground reaction force. *p < 0.05, **p < 0.01

6.5.3. Relationships between open and closed neuromuscular variables and disability

A total of two candidate predictors of disability were derived from the previous analyses: i) RMSED and, ii) thorax-lumbar time. The bivariate relationship between the novel open (RMSED) and closed (thorax-lumbar time) neuromuscular predictors was fair and approaching significance (r = 0.34, p = 0.054). Together, these variables predict 28% the variance of the ODI score (adjusted R^2 = 0.28, p = 0.003). Within this model, a one unit increase in RMSED (i.e., impairment in the ability to decrease force production) is associated with a 35% increase in ODI score when the other variable is held constant (standardised \( \beta = 0.35, p = 0.037 \)). Similarly, a one second increase in in-phase movement between the thorax and lumbar spine is associated with a 35% increase in the ODI score when the other variable is held constant (standardised \( \beta = 0.35, p = 0.036 \)). The two-variable linear regression model to predict disability levels are summarised in table 6.5.
Table 6.5. The summary of the two-variable linear regression model to predict disability levels of CLBP participants

<table>
<thead>
<tr>
<th>Regression model</th>
<th>Adjusted R²</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0.28</td>
<td>0.003</td>
</tr>
<tr>
<td>Predictors</td>
<td>Standardized β</td>
<td></td>
</tr>
<tr>
<td>RMSE_D</td>
<td>0.35</td>
<td>0.036</td>
</tr>
<tr>
<td>Thorax-lumbar Time</td>
<td>0.35</td>
<td>0.037</td>
</tr>
</tbody>
</table>

RMSE_D = Average root mean square error during ramp down phase, Thorax-lumbar time = thorax-lumbar in-phase time duration.

6.5.4. Evaluation of regression models

Variance inflation factors and tolerance were within acceptable limits for all regression models suggesting that collinearity between the variables were acceptable. Homoscedasticity, normality and linearity of standardised residuals were demonstrated for all regression models (Osborne & Waters, 2002; Williams et al., 2013).

6.6. Discussion

6.6.1. Overview of results

The primary aim of this study was to investigate whether self-reported disability can be predicted using a combination of open and closed kinetic chain neuromuscular variables derived from novel tests. Of importance, both RMSE_D and thorax-lumbar joint interaction time were correlated moderately and positively with disability level. Collectively, the results of this final study answered the thesis’ main question and collectively, the two variables predicted 28% of the variance in disability. Thus, these results confirmed the face validity of the open and closed kinetic chain neuromuscular tests outlined in this thesis.
6.6.2. Relationships between open kinetic chain neuromuscular variables and disability

**Lumbar extensor muscle strength and disability**

Lumbar extensor MVIC was not associated with disability levels (accepted $H_{1a}$) in CLBP participants. This finding is in agreement with several previous studies (Al-Obaidi et al., 2000; Deyo, 1988; Waddell et al., 1992) which suggest that this measure may assess the wrong domain of muscle function in people with CLBP. In a small study ($n = 12$), improvement in lumbar extension strength following a strength-training program in people with CLBP was associated with changes in disability ($r = -0.45$ to $-0.52$, $p < 0.05$) (Steele et al., 2013). However, in a large scale study ($n = 148$), changes in disability following a lumbar strengthening exercise program were not associated with an increase in lumbar extensor strength but rather due to a decrease in distress, fear-avoidant behaviour and increased efficacy in controlling pain (Mannion, Junge, et al., 2001). Moreover, improvement in lumbar extensor strength was not associated with an increase in erector spinae and multifidus cross-sectional area (Käser et al., 2001). Collectively, results of this study and previous research suggest that utilising lumbar extension strength to predict self-reported disability of people with CLBP may not be well founded.

**Lumbar extensor muscle force control and disability**

Lumbar extensor force control was moderately and positively correlated with self-reported disability of CLBP participants (accepted $H_{1b}$). Additionally, the relationship between lumbar extensor MVIC and force control variables was not significant. This suggests that the ability to control submaximal strength is independent from the ability to produce maximum strength, a finding previously reported in the quadriceps following ACLR surgery (Bryant et al., 2009). Trunk flexor (e.g., rectus abdominis) and extensor (e.g., erector spinae) muscle co-contraction is a common feature of CLBP and could alter functional performance (Reeves et al., 2008;
Reeves et al., 2006). In people with CLBP, increased trunk muscle co-contraction (i.e.,
dyskinesia) may be mal-adaptive by reducing quality of movement and increasing compressive
load on the spine (Marras et al., 2001) (see section 2.6.3). This could be reflected in impaired
trunk muscle force control (see section 4.6.4) which, in turn, was associated with disability in
CLBP participants.

In this study, an impaired ability to accurately decrease muscle force output was correlated
moderately with self-reported disability level (r = 0.47). Streiner et al. (2014) suggest that in
order for two outcome measures to quantify the same underlying construct, they should
demonstrate a bivariate correlation coefficient between 0.40 and 0.80. Similarly, correlation
coefficients ranging from 0.40 to 0.60 have also been observed between physical function tests
such as walking and sit-to-stand and self-reported disability (Novy, Simmonds, & Lee, 2002;
Simmonds et al., 1998). Thus, findings of this study confirm the face validity of the novel open
kinetic chain variable of lumbar extensor muscle force control.

6.6.3. Relationships between closed kinetic chain neuromuscular variables
and disability

Joint range of motion and angular velocity

This study failed to demonstrate a significant correlation between self-reported disability and
either ROM or angular velocity during a symmetrical lifting task (accepted H2a). These findings
are in agreement with previous studies that have concluded that simplistic kinematic
measurements such as lumbar ROM on their own should not be used to predict disability in
people with CLBP (Mannion, Junge, et al., 2001; Parks et al., 2003; Sullivan, Shoaf, & Riddle,
2000). Sullivan et al. (2000) investigated multiple confounders that may influence the
relationship between disability and lumbar ROM such as participant characteristics (e.g., age,
gender, height and weight) and clinicians’ belief that lumbar ROM was related to disability
using multiple regression analysis. Sullivan et al. (2000) found that lumbar ROM explained
very little of the disability variance ($R^2 = 1\%$) when the confounding factors were controlled via statistical means. However, Sullivan et al. (2000) study only assessed lumbar ROM during trunk flexion – an arguably non-functional task. In this study, neither trunk ROM or trunk angular velocity (the first derivative of trunk ROM) during lifting were correlated with disability. Thus, as trunk ROM and angular velocity during lifting could not explain the variance in disability, on their own, they should not be used as surrogate measures for disability.

**Inter-joint coordination variables during lifting**

In-phase movement time (*i.e.*, *phase-locked* movement pattern; see section 5.6.5) between the thorax and lumbar spine during lifting demonstrated a significant, positive correlation with self-reported disability (accepted $H_{2b}$). Previous research suggests that when extending their lumbar spine from a fully flexed position, people with CLBP demonstrate increased lumbar to hip ROM ratio (*i.e.*, spinal strategy) (McClure et al., 1997). Additionally, during lumbar extension, people with CLBP have been found to demonstrate a more in-phase movement pattern through the lumbar spine and pelvis (Mokhtarinia et al., 2016). This CLBP strategy is accompanied by early activity of the lumbar extensor muscles relative to the gluteus maximus (Nelson-Wong et al., 2012). The findings of this study add to the previous research in that when extending from a fully flexed trunk position, decreased dissociation between the thorax and lumbar spine was correlated with increased disability in people with CLBP. Longer in-phase time during lifting may expose those with CLBP to augmented spinal load (Greenland et al., 2013) which, in turn, could perpetuate pain and reduce movement quality. Interestingly, the amount of deviation and peak deviation from in-phase joint couple movements were not correlated with disability. Movement variability is desirable to maintain posture in the presence of perturbations during daily activities (Mok et al., 2007) and is required to distribute load around spinal structures (*e.g.*, intervertebral discs, zygapophysial joints) as constant loading from trunk muscle co-contraction (see section 2.6.3) can be pathological for tissue health (Stergiou & Decker, 2011).
Decreased movement variability has been postulated to perpetuate pain and disability in people with CLBP (Hodges & Smeets, 2015; O'Sullivan, 2005). Thus, the findings of this study confirmed the face validity of lifting inter-joint coordination measurement using relative phase angle analysis in people with CLBP.

**Kinetic variables during lifting**

The findings of the present study revealed that vertical GRF variables are negatively correlated with disability level in people with CLBP (*i.e.*, decreased vertical GRF’s is associated with increased disability) (accepted H$_{2c}$). Compared to back lifting (*i.e.*, lifting with back bent with straight knees), leg lifting (*i.e.*, lifting with straight back and bent knees; see section 2.7) has long been recommended given smaller net pressures in L4-5 intervertebral discs and net muscle forces (Bazrgari, Shirazi-Adl, & Arjmand, 2007; Wilke, Neef, Hinz, Seidel, & Claes, 2001). Moreover, people with CLBP in the study by Sanchez-Zuriaga *et al.* (Sanchez-Zuriaga *et al.*, 2011) were found to generate lower vertical GRF’s during a series of lifting tasks compared to healthy people – a finding which suggests that those with CLBP adopt a back lifting technique, probably in an attempt to reduce pain in the lumbar spine but ultimately, could lead to further injury (see section 5.6.5). Lifting is a complex movement that involves multiple interactions of joints, muscles and passive structures (Bazrgari *et al.*, 2007). Thus, it may not be appropriate to evaluate the relationship between vertical GRF’s and disability level without considering other variables such as joint coordination and trunk muscle activity – both of which are directly related to spinal load (Marras *et al.*, 2001). Regardless, the findings of this study confirmed the face validity of the measurement of vertical GRF’s during lifting in people with CLBP.

**6.6.4. Relationships between open and closed kinetic chain neuromuscular variables and disability**

Collectively, the findings of this study support the notion that neuromuscular variables are important predictors of self-reported disability (accepted H$_3$) (Dubois *et al.*, 2014; Mannion,
Specifically, open and closed kinetic chain variables derived from novel tests outlined in study two (trunk muscle force control) and study three (lifting inter-joint coordination and vertical GRF’s) are valid for use in people with CLBP. The two assessment variables (i.e., RMSED and thorax-lumbar in-phase time) reported in this study were slightly more predictive of disability variance than those used in the Mannion et al. (2001) study (see section 2.8). However, Mannion et al. (2001) utilised a variety of neuromuscular variables derived from open and closed tests including muscle activation, lumbar extensor strength, median frequency of EMG of bilateral erector spinae, isometric and dynamic erector spinae fatigue, lumbar ROM and erector spinae flexion relaxation time (i.e., the absence of EMG) activity during full lumbar flexion). These variables were grouped under “performance measures” using principal component analysis and predicted 25% of the variance of self-reported disability. Importantly, the findings of the present study suggest that the assessment of lumbar extensor muscle force control and inter-joint coordination during lifting could replace a battery of physical assessment tests. The novel biomechanical assessments outlined in this thesis were able to differentiate CLBP participants with lower disability level (mean ODI = 20.7%) from healthy controls. Therefore, the combination of this thesis’ novel open and closed kinetic chain assessments may direct future classification and treatment of CLBP sufferers with low disability levels.

There has been a recent shift in the related literature to move away from the assessment of neuromuscular and biomechanical impairments and instead, focus on assessing psychological and psychosocial factors (e.g., assessment of pain related fear) in order to explain disability in people with CLBP (Nicholas & George, 2011). On face value, this transition away from the assessment of physiological impairments seems reasonable, since a number of neuromuscular variables including trunk muscle activation and trunk ROM have not been able to differentiate people with CLBP from healthy people (Deyo, 1988; Pulkovski et al., 2012; Waddell et al.,
Moreover, several randomised controlled trials have demonstrated that CLBP-related treatment approaches which addressed specific neuromuscular deficits derived from traditional open and closed kinetic chain tests resulted in a modest improvement of disability; these interventions were not superior to general exercise (Ferreira et al., 2007; Vasseljen, Unsgaard-Tøndel, Westad, & Mork, 2012). Thus, more research is required across the physiological domains in order to ensure effective and targeted management of patients with CLBP (Fairbank et al., 2011). The research presented in this thesis has contributed to this notion.

6.6.5. Clinical implications

The ODI is the most commonly utilised outcome measure for CLBP that is easy and quick to administer (Chapman et al., 2011; Fairbank & Pynsent, 2000). The results of this study suggest that patients with CLBP who score higher on the ODI are likely to demonstrate more impairments in lumbar extensor muscle force control and impairments in lifting-related movement pattern than CLBP patients who have lower ODI scores. Considering that CLBP patients in the current study who relatively had low disability (mean ODI = 20.7%), still demonstrated adverse changes in the neuromuscular variables assessed by the novel tests, suggests that the novel tests should be conducted on CLBP patients across disability levels.

As lumbar muscle force control and inter-joint kinematic variables are predictive of CLBP disability level, interventions that target these variables may, in turn, contribute to improvements in disability level. In this study, an improvement of one unit in force-matching error (mean error = 4.13 ± 1.35) while decreasing lumbar extensor force output in people with CLBP, was related to a 47% improvement in ODI (MCID = 10% (Fairbank & Pynsent, 2000)).

Previous studies suggest that motor control-based training should be specific to the task of interest in order to induce changes at a cortical level and, in turn, improve task performance (Hall et al., 2009; Tsao, Galea, & Hodges, 2010). The lumbar extensor muscle force control protocol outlined in this thesis (see Study 2) could be adapted as a motor control training
modality. An example setup for this test may include: portable force platform such as a WBB (Nintendo, Kyoto, Japan) affixed to the backrest of a chair in lieu of a dynamometer. Participants would sit and pushes back against the WBB to match a target force displayed on a tablet/laptop computer located in front of him/her (see section 4.4.3). The use of WBB’s in the assessment of force outside the laboratory setting has been validated previously (Clark et al., 2010). Thus, the aforementioned setup could also be adapted for training purposes. Therefore, the trainability of trunk muscle force control should be the focus for future research.

This study findings confirmed recent postulations by Mokhtarinia et al. (2016) and results of Study 3 that increased in-phase motion between the thoracic spine and lumbar spine during lifting is deleterious in people with CLBP. Additionally, pertaining to lifting-related vertical GRF’s, results of this study also confirm that a leg lifting technique (see section 5.6.5) is preferred by people with CLBP. In relation to this study, a one second decrease in thoracolumbar coupled movement during lifting was associated with an improvement of ODI by 47%. Thus, interventions that promote thoracolumbar movement dissociation and/or decrease thoracolumbar stiffness may decrease lifting-related disability. The setup for lifting biomechanics assessment or intervention can be reproduced using a portable, low-cost accelerometers with data transmitted wirelessly to a computer (e.g., the Valedo system, Hocoma, AG) (Bauer et al., 2015) and portable gaming system as force platforms (e.g., WBB’s as force platforms). Future research should investigate the trainability of lifting biomechanics using portable technologies in people with CLBP.

In addition to confirming the face validity of the novel neuromuscular assessments in this thesis, the studies in this thesis also confirmed the external validity of these novel assessments. The established relationship between this thesis’ novel tests confirmed that the assessment of trunk muscle force control, lifting inter-joint coordination and lifting vertical GRF’s assess the appropriate domain of muscle function in CLBP population. Moreover, as the research
participants were physiotherapist-screened CLBP clients and were tested at physiotherapy clinics during a clinical session within a period of 30 minutes, the novel assessments outlined in this thesis is generalisable to wider CLBP population.

6.7. Strengths and limitations

The strengths of this study are as follow:

1. To the author’s knowledge, this is the first study to concurrently validate the assessment of lumbar extensor muscle force control in people with CLBP with a self-reported disability measure. Results suggest that open kinetic chain assessment of motor control can still play a role in the management of people with CLBP.

2. Similarly, this is the first study to evaluate trunk-lower limb coordination during lifting with self-reported disability of people with CLBP. Moreover, the study findings suggest that clinicians should not neglect the importance of the thoracic-lumbar movement coordination when correcting symptomatic lifting patterns in people with CLBP.

3. Overall, the findings suggest that the measurement of physical impairments still have a role in the management of CLBP. When the relevant physical impairment domains (i.e., precision of muscle force production and inter-joint coordination) are assessed, a significant portion of self-reported disability variance could be explained.

4. The study also established the external validity of these novel measures in CLBP population. As all assessments in this thesis were performed at Physiotherapy clinics involving physiotherapist-screened CLBP clients, the results of this study is generalisable to the clinical population.

5. The validation of the novel assessment strategies could lead to novel intervention strategies (e.g., training of force control and movement patterns using portable gaming technologies) to manage people with CLBP.
This study also has several limitations:

1. This study included a small sample size and thus, only three predictor variables could be included in the multivariate regression analyses. Therefore, this study could not take into account potentially important non-modifiable covariates such as age, gender or duration of back pain (Kim et al., 2015; Sung, Park, & Kim, 2012; van Abbema et al., 2011).

2. Similarly, the study variables explained 28% of disability variance; hence, 72% of the variance remains unexplained. Mannion et al. (2001) demonstrated that up to 36% of disability variance could be explained by psychological factors such as pain-related fear. A more recent study suggests that perceived functioning may be a better indicator of disability rather than functional tests in general (van Rooij et al., 2015). Moreover, preliminary evidence suggests that a significant but weak correlation exists between pain-related fear and motor control (Karayannis et al., 2013). These variables were not controlled for in this study. Having this information may assist in identifying which CLBP participants could benefit from the novel assessments reported in this thesis.

3. This study assumed a linear relationship between neuromuscular variables and self-reported disability in people with CLBP. Other forms of relationships may exist but could not be explored in this study; such as the curvilinear (U-shaped) relationship between physical activity and risk of CLBP (Heneweer, Vanhees, & Picavet, 2009).

4. Given the cross-sectional nature of this study, it is important to acknowledge the possibility of reverse causation. Specifically, it is unclear whether neuromuscular adaptations derived via open and closed kinetic chain assessment existed prior to CLBP onset. In future studies, this could be addressed, in part, via different statistical analyses (e.g., logistic regression analysis), longitudinal studies and randomised controlled trials.

Future directions pertaining to this study will be discussed in section 7.2.
6.8. Conclusion

This study demonstrated an association between lumbar extensor muscle force control – as derived using a novel open kinetic chain task, and self-reported disability level of CLBP individuals. The inability to accurately control decreasing muscle force production explained 19% of the variance of self-reported disability. Moreover, thorax-lumbar spine in-phase movement time during lifting was correlated with self-reported disability level of CLBP individuals. In isolation, this variable explained 19% of the variance of self-reported disability. Together, both open and closed kinetic chain neuromuscular assessment variables explained 28% of the variance in CLBP disability. These findings confirm the face validity of the novel open and closed kinetic chain assessments implemented in this thesis. Overall, the studies in this thesis confirm the notion that more advanced open and closed kinetic neuromuscular assessments are better predictors of self-reported CLBP disability level than traditional non-specific physical assessment tests.
Chapter 7
Summary and future directions

7.1. Summary of findings
The primary aim of this thesis was to investigate the relationship between trunk muscle force control, trunk and lower limb joint coordination during lifting and self-reported disability in people with CLBP. This research implemented novel open and closed neuromuscular assessment techniques in people with CLBP in order to address an important question; that is, is there a relationship between neuromuscular impairments, assessed using advanced biomechanical techniques, and self-reported disability in people with CLBP? Results of this thesis confirmed the validity of the neuromuscular variables derived from the advanced open and closed kinetic chain tests. A summary of the four studies included in this thesis follows:

7.1.1. Study 1: Participant characteristics and self-reported disability in people with chronic low back pain
The first study in this thesis (Chapter 3) described the characteristics of CLBP and healthy participants with respect to their self-reported disability and pain. The findings of this chapter included: CLBP participants reported more disability (higher ODI score) and more pain intensity than healthy controls. Importantly, the ODI and pain scores of the CLBP participants in this study were comparable to those included in previous related studies.

7.1.2. Study 2: Lumbar extensor muscle force control in people with and without chronic low back pain
The second study in this thesis (Chapter 4) incorporated a novel open kinetic chain protocol for assessment of lumbar extensor muscle force control. This protocol – adapted from an
established quadriceps force control test – assessed the accuracy of isometric lumbar extensor muscle force output with respect to a fluctuating, submaximal ‘target’ force in CLBP and healthy participants. Participants were required to increase and decrease lumbar extensor muscle force within 20% and 50% of the muscles MVIC.

The protocol for assessment of lumbar extensor muscle force control demonstrated moderate to excellent inter-session reliability in both healthy and CLBP participants. CLBP participant’s demonstrated significantly higher force-matching error compared to healthy controls. Additionally, CLBP participant’s demonstrated significantly higher lumbar extensor force-matching error when increasing submaximal force production from 20% and 50% MVIC. Augmented force-matching error occurred despite both healthy and CLBP groups possessing similar lumbar extensor strength.

The findings of Study 2 indicate that the protocol used to assess lumbar extensor muscle force control was capable of differentiating CLBP participants from control participants. The significance of this study was the implementation of a variable, fluctuating force target – reflective of the type of lumbar extensor muscle contraction utilised during daily activities. Importantly, this protocol demonstrated larger between-group differences compared to static force target protocols employed in previous studies.

7.1.3. Study 3: Coordination between the trunk and lower limb joints during symmetrical lifting in people with and without chronic low back pain

The third study in this thesis (Chapter 5) investigated lifting biomechanics using both kinematic (i.e., relative phase angle analysis) and kinetic (i.e., vertical GRF) analyses. With respect to the former, coordination between the thoracic, lumbar, hip and knee joints during lifting was assessed. Use of relative phase angles allowed for assessment of joint couple movements during the entire extension phase of lifting. This study sought to differentiate movement patterns of
two CLBP phenotypes based on their disability level; that is, higher (CLBP\textsubscript{high}) and lower disability (CLBP\textsubscript{low}), and healthy controls.

The lifting protocol used in this study demonstrated moderate to excellent inter-session reliability in both healthy and CLBP participants. No significant between-group differences were found for ROM and angular velocity for all trunk and lower limb joints during lifting. Similarly, all groups demonstrated a similar lifting pattern with respect to inter-joint coordination; that is, the knee led the hip, the lumbar spine led the hip and the lumbar spine led the thorax. Additionally, there were no significant differences in vertical GRF’s between the CLBP and control groups. However, people with CLBP took longer to lift a relatively light load than healthy controls. The findings suggest that during symmetrical lifting, trunk-lower limb joint coordination patterns and vertical GRF are not affected by CLBP.

7.1.4. Study 4: Relationships between open and closed kinetic chain neuromuscular variables and disability in people with chronic low back pain

The final study in this thesis investigated the relationship between neuromuscular variables derived from novel open and closed kinetic chain tasks and self-reported disability level of CLBP participants. The first part of this chapter investigated the relationship between lumbar extensor muscle force control variables and self-reported disability. The second part of this chapter investigated the relationship between inter-joint coordination and kinetic variables during symmetrical lifting and self-reported disability. The final part of this chapter addressed the main aim of this thesis: to investigate if both open and closed kinetic chain neuromuscular variables could predict self-reported CLBP-related disability.

The findings of this study indicate impaired capacity of the lumbar extensor muscles to accurately decrease isometric force production was positively associated with increased self-reported disability level of CLBP participants. This variable (\textit{i.e.}, RMSE\textsubscript{D}) significantly
predicted 19% of the variance in self-reported CLBP disability. A one point increase in error was associated with an increase in ODI score of 47%. The second part of the study demonstrated a significant positive correlation between thorax-lumbar spine in-phase movement time and ODI score. Similarly, this variable predicted 19% the variance in the ODI score and indicates that a one second increase in thorax-lumbar spine in-phase movement time was also associated with a 47% increase in ODI score. Finally, multivariate regression analysis showed that a combination of both RMSEd and thorax-lumbar in-phase movement time predicted 28% of the variance of the ODI score. A one unit increase in error (i.e., impairment in the ability to decrease force production) was associated with a 35% increase in ODI score when the other variable was held constant. Similarly, a one second increase in in-phase thorax-lumbar spine movement was associated with a 35% increase in the ODI score when the other variable was held constant. These findings are significant from a clinical perspective given that they confirm the disability-related importance of neuromuscular-biomechanical impairments in people with CLBP. Hence, improvement in neuromuscular-biomechanical function via clinical interventions will contribute to a reduction in disability levels of those with CLBP. In the context of the research findings, clinicians should move away from performing traditional neuromuscular-biomechanical assessments including lumbar flexion ROM and lumbar extension strength given that these measures are unable to account for any of the variance in self-reported CLBP disability.

7.2. Future research directions
The thesis strengths and limitations have been reported in each main study chapter (see sections 4.7, 5.7 and 6.7). In order to direct future research, the information outlined based on this thesis can be summarised as following:
1. As the research reported in this thesis pertain to adults aged 18-60, it is currently unknown whether the CLBP-related open and closed kinetic chain neuromuscular adaptations would also be observed in older adults. Thus, future research should specifically investigate the assessment of trunk muscle force control and trunk and lower limb inter-joint coordination during lifting in population aged > 60.

2. Future studies should include superficial or indwelling EMG to confirm the existence of trunk muscle co-contraction as a contributing factor to the impairment of trunk muscle force control and lifting inter-joint coordination in people with CLBP.

3. Prospective studies should be conducted to ascertain whether impairments in muscle force control, lifting inter-joint coordination and vertical GRF’s are a result or cause of CLBP.

4. The study did not assess the effect of fatigue on lumbar extensor muscle force control and lifting-related inter-joint coordination. Lumbar muscle fatigue was actually minimised in the studies herein due to the requirement for research participants (who were new clients of the physiotherapy clinics; see section 3.4.1) to participate in physiotherapy sessions. Testing in a non-fatigued or non-repetitive condition may not reflect real life situations as risk of injury increases when fatigued (Sparto, Parnianpour, Reinsel, & Simon, 1997). Trunk muscle fatigue has been shown to decrease trunk muscle force steadiness (Salomoni & Graven-Nielsen, 2012) and phase-locked movement pattern during lifting (Hu & Ning, 2015; Sparto et al., 1997).

5. Assessment of asymmetrical lifting may be more relevant in people with CLBP as it has been associated with much higher spinal load than symmetrical lifting (Marras, Ferguson, Burr, Davis, & Gupta, 2004). Moreover, this thesis only assessed inter-joint coordination in the sagittal plane. Future studies should assess inter-joint coordination
in the transverse and axial planes as movements in these planes exert torsional and shear forces on the lumbar spine (Kingma et al., 2010).

6. CLBP participants in the thesis studies were not sub-classified based on their pain provoking movement directions (e.g., flexion-based or extension-based groups). This might have resulted in small number of lifting kinematic variables to reach statistical significance. Therefore, future studies should assess trunk and lower limb joint coordination during lifting in CLBP individuals grouped under pain provoking movement-based phenotypes.

7. Although this thesis has established the relationship between trunk muscle force control variables with disability level, the relationships between trunk muscle force control variables and functional performance have not been investigated. Thus, it is unknown if improvement in trunk muscle force control is associated with improvement in functional activities (e.g., lifting performance). Thus, future studies should investigate the multivariate associations between lumbar extensor muscle force control variables with physical performance.

8. The assessment of kinesiophobia (i.e., fear of movement) was outside the scope of the thesis. Kinesiophobia has been associated with changes in trunk muscle activity and trunk kinematics (Karayannis et al., 2013) and thus, may affect this thesis’ open and closed kinetic chain test results. Thus, future studies should assess the effect of pain-related fear on trunk muscle force control and joint coordination and vertical GRF’s during lifting.

9. Future studies should focus on the trainability of trunk muscle force control and inter-joint coordination during lifting. Cost-effective portable gaming technology (i.e., Microsoft Kinect (Microsoft, Redmond, Washington) and WBB (Nintendo, Kyoto, Japan)) could be used to achieve this aim. Assessing trainability of these novel
neuromuscular variables is potentially important due to their significant relationships with self-reported disability. A future randomised controlled trial could be performed to investigate the efficacy of lumbar extensor force control training in people with CLBP in improving self-reported disability.

In conclusion, this thesis has provided contemporary evidence and possible new mechanisms of neuromuscular-biomechanical adaptations by introducing novel tests of trunk muscle force control and trunk-lower limb inter-joint coordination during lifting in people with CLBP. Moreover, this thesis confirmed the face validity of these novel tests by investigating the relationship between the neuromuscular variables derived from these tests and self-reported disability. This research will inform future studies with respect to relevant neuromuscular-biomechanical tests and outcome measures leading to novel interventions for people with CLBP.


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Stewart Williams, J., Ng, N., Peltzer, K., Yawson, A., Biritwum, R., Maximova, T., . . . Chatterji, S. (2015). Risk Factors and Disability Associated with Low Back Pain in Older Adults in Low- and Middle-Income Countries. Results from the WHO Study on Global AGEing and Adult Health (SAGE). *PLoS ONE*, 10(6), e0127880. doi:10.1371/journal.pone.0127880


Tracy, B. L., & Enoka, R. M. (2002). Older adults are less steady during submaximal isometric contractions with the knee extensor muscles. *J Appl Physiol (1985)*, 92(3), 1004-1012. doi:10.1152/japplphysiol.00954.2001


Appendices

1. Human Ethics Approval
2. Oswestry Disability Index
3. Pilot Testing
4. Reflective Marker Placements
5. Butterworth Filter Frequency Selection
Appendix 1

Human Ethics Approval

A letter of approval by the University of Melbourne Human Research Ethics Committee confirming this PhD research is included below:
Appendix 2

Oswestry Disability Index

The Oswestry Disability Index by Fairbank & Pynsent (2000) is included below:

Oswestry Low Back Pain Disability Questionnaire

Instructions
This questionnaire has been designed to give us information as to how your back or leg pain is affecting your ability to manage in everyday life. Please answer by checking ONE box in each section for the statement which best applies to you. We realise you may consider that two or more statements in any one section apply but please just shade out the spot that indicates the statement which most clearly describes your problem.

Section 1 – Pain intensity
☐ I have no pain at the moment
☐ The pain is very mild at the moment
☐ The pain is moderate at the moment
☐ The pain is fairly severe at the moment
☐ The pain is very severe at the moment
☐ The pain is the worst imaginable at the moment

Section 2 – Personal care (washing, dressing etc)
☐ I can look after myself normally without causing extra pain
☐ I can look after myself normally but it causes extra pain
☐ It is painful to look after myself and I am slow and careful
☐ I need some help but manage most of my personal care
☐ I need help every day in most aspects of self-care
☐ I do not get dressed, I wash with difficulty and stay in bed

Section 3 – Lifting
☐ I can lift heavy weights without extra pain
☐ I can lift heavy weights but it gives extra pain
☐ Pain prevents me from lifting heavy weights on the floor, but I can manage if they are conveniently placed eg. on a table
☐ Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned
☐ I can lift very light weights
☐ I cannot lift or carry anything at all

Section 4 – Walking
☐ Pain does not prevent me walking any distance
☐ Pain prevents me from walking more than 2 kilometres
☐ Pain prevents me from walking more than 1 kilometre
☐ Pain prevents me from walking more than 500 metres
☐ I can only walk using a stick or crutches
☐ I am in bed most of the time
Section 5 – Sitting
☐ I can sit in any chair as long as I like
☐ I can only sit in my favourite chair as long as I like
☐ Pain prevents me sitting more than one hour
☐ Pain prevents me from sitting more than 30 minutes
☐ Pain prevents me from sitting more than 10 minutes
☐ Pain prevents me from sitting at all

Section 6 – Standing
☐ I can stand as long as I want without extra pain
☐ I can stand as long as I want but it gives me extra pain
☐ Pain prevents me from standing for more than 1 hour
☐ Pain prevents me from standing for more than 3 minutes
☐ Pain prevents me from standing for more than 10 minutes
☐ Pain prevents me from standing at all

Section 7 – Sleeping
☐ My sleep is never disturbed by pain
☐ My sleep is occasionally disturbed by pain
☐ Because of pain I have less than 6 hours sleep
☐ Because of pain I have less than 4 hours sleep
☐ Because of pain I have less than 2 hours sleep
☐ Pain prevents me from sleeping at all

Section 8 – Sex life (if applicable)
☐ My sex life is normal and causes no extra pain
☐ My sex life is normal but causes some extra pain
☐ My sex life is nearly normal but is very painful
☐ My sex life is severely restricted by pain
☐ My sex life is nearly absent because of pain
☐ Pain prevents any sex life at all

Section 9 – Social life
☐ My social life is normal and gives me no extra pain
☐ My social life is normal but increases the degree of pain
☐ Pain has no significant effect on my social life apart from limiting my more energetic interests eg, sport
☐ Pain has restricted my social life and I do not go out as often
☐ Pain has restricted my social life to my home
☐ I have no social life because of pain

Section 10 – Travelling
☐ I can travel anywhere without pain
☐ I can travel anywhere but it gives me extra pain
☐ Pain is bad but I manage journeys over two hours
☐ Pain restricts me to journeys of less than one hour
☐ Pain restricts me to short necessary journeys under 30 minutes
☐ Pain prevents me from travelling except to receive treatment

*Note: Distances of 1 mile, ½ mile and 100 yards have been replaced by metric distances in the Walking section
Appendix 3

Pilot Testing

Upon human ethics approval and prior to data collection, a pilot testing was conducted with 5 participants with CLBP and 10 healthy participants who were colleagues at The University of Melbourne. These participants fulfilled the inclusion and exclusion criteria outlined in Study 1 (see table 3.1) but their data were not included in Study 2 and 3 analyses.

The aims of the pilot testing are to assess the data collection protocols and parameters pertaining to lumbar extensor muscle force control assessment and lifting task. The quality of the data obtained and participants’ feedback with respect to task difficulty, potential pain provocation, muscle fatigue and their confidence in performing the tasks were also assessed. For the assessment of lumbar extensor muscle force control (Study 2), multiple frequencies were used to determine the speed of movement of target torque (e.g., 1 Hz, 0.5 Hz, 0.2 Hz, 0.18 Hz, 0.16 Hz, 0.14 Hz and 0.10 Hz). Pilot testing was used to determine the upper or lower limits of the force target (e.g., 5%-30%, 20%-50% and 30%-70%). The length of the force control test (i.e., 3, 2 or 1 testing sets) were also determined. For Study 3, pilot testing was used to determine the amount of load lifted (e.g., 5 kg, 8 kg, 10 kg and 12 kg) and the number of lifting repetitions.

In order to objectively assess the appropriateness of the protocols, numeral rating scales (0-10) were used to quantify pain, fatigue and difficulty were administered after each test. For instance, in the assessment of muscle force control (Study 2), when the frequency of 0.5 Hz was used, healthy participants would rate 9/10 on the difficulty scale and would provoke 8/10 pain on the low back pain participants. However, a frequency of 0.16 Hz could be completed successfully by all participants without provoking any pain. This optimal test frequency was obtained with a combination of inspection of the numeral rating scales, successful completion of tasks and
discussion with participants. A copy of the numerical rating scales used for pilot testing is included below:

How much discomfort, if any, do you have now (before the task)?

0 1 2 3 4 5 6 7 8 9 10
No discomfort at all Extreme discomfort

How difficult was the task?

0 1 2 3 4 5 6 7 8 9 10
No difficulty at all Extremely difficult

How much discomfort, if any, did you have after the task?

0 1 2 3 4 5 6 7 8 9 10
No discomfort at all Extreme discomfort

How tiring was the task?

0 1 2 3 4 5 6 7 8 9 10
Not tiring at all Extremely tiring
Appendix 4

Reflective Marker Placements
<table>
<thead>
<tr>
<th>No.</th>
<th>Segments</th>
<th>Marker names</th>
<th>Marker locations</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Head</td>
<td>H1</td>
<td>External occipital protuberance</td>
</tr>
<tr>
<td></td>
<td></td>
<td>H2</td>
<td>Left temple</td>
</tr>
<tr>
<td></td>
<td></td>
<td>H3</td>
<td>Right temple</td>
</tr>
<tr>
<td>2</td>
<td>Thorax</td>
<td>T1</td>
<td>T1 spinous process</td>
</tr>
<tr>
<td></td>
<td></td>
<td>T2</td>
<td>5 cm left of T7 spinous process</td>
</tr>
<tr>
<td></td>
<td></td>
<td>T3</td>
<td>5 cm right of T7 spinous process</td>
</tr>
<tr>
<td>3</td>
<td>Lumbar</td>
<td>L1</td>
<td>L1 spinous process</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L2</td>
<td>5 cm left of L3 spinous process</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L3</td>
<td>5 cm right of L3 spinous process</td>
</tr>
<tr>
<td>4</td>
<td>Pelvis</td>
<td>P1</td>
<td>Mid-point between left and right posterior superior iliac spine</td>
</tr>
<tr>
<td></td>
<td></td>
<td>P2</td>
<td>Left iliac spine on midaxillary line</td>
</tr>
<tr>
<td></td>
<td></td>
<td>P3</td>
<td>Right iliac spine on midaxillary line</td>
</tr>
<tr>
<td></td>
<td></td>
<td>P4 *</td>
<td>Left anterior superior iliac spine</td>
</tr>
<tr>
<td></td>
<td></td>
<td>P5 *</td>
<td>Right anterior superior iliac spine</td>
</tr>
<tr>
<td>5</td>
<td>Left thigh</td>
<td>LT1</td>
<td>Proximal one third between left greater trochanter and left lateral femoral condyle on left iliotibial tract</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LT2</td>
<td>5 cm anterior of the midway line between left greater trochanter and left lateral femoral condyle on left rectus femoris</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LT3</td>
<td>Distal one third between left greater trochanter and left lateral femoral condyle on left iliotibial tract</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LT4 *</td>
<td>Lateral femoral epicondyle of the left femur</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LT5 *</td>
<td>Medial femoral epicondyle of the left femur</td>
</tr>
<tr>
<td>6</td>
<td>Right thigh</td>
<td>RT1</td>
<td>Proximal one third between right greater trochanter and right lateral femoral condyle on right iliotibial tract</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RT2</td>
<td>5 cm anterior of the midway line between right greater trochanter and right lateral femoral condyle on right rectus femoris</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RT3</td>
<td>Distal one third between right greater trochanter and right lateral femoral condyle on right iliotibial tract</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RT4 *</td>
<td>Lateral femoral epicondyle of the right femur</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RT5 *</td>
<td>Medial femoral epicondyle of the right femur</td>
</tr>
<tr>
<td>7</td>
<td>Left shank</td>
<td>LS1</td>
<td>Proximal one third between left tibial tuberosity and midpoint between left medial and lateral malleoli on left tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LS2</td>
<td>5 cm lateral of the midway line between left tibial tuberosity and midpoint between left medial and lateral malleoli on left peroneus longus</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LS3</td>
<td>Distal one third between left tibial tuberosity and midpoint between left medial and lateral malleoli on left tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LS4 *</td>
<td>Lateral malleolus of the left tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>LS5 *</td>
<td>Medial malleolus of the left tibia</td>
</tr>
<tr>
<td>8</td>
<td>Right shank</td>
<td>RS1</td>
<td>Proximal one third between right tibial tuberosity and midpoint between right medial and lateral malleoli on right tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RS2</td>
<td>5 cm lateral of the midway line between right tibial tuberosity and midpoint between right medial and lateral malleoli on right peroneus longus</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RS3</td>
<td>Distal one third between right tibial tuberosity and midpoint between right medial and lateral malleoli on right tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RS4 *</td>
<td>Lateral malleolus of the right tibia</td>
</tr>
<tr>
<td></td>
<td></td>
<td>RS5 *</td>
<td>Medial malleolus of the right tibia</td>
</tr>
</tbody>
</table>

*Markers used for static trials only and removed during testing*
Appendix 5

Butterworth Filter Frequency Selection

In order to analyse the kinematic data, in particular the angular velocity described in Study 3, data filtering was required to remove noise which may distort research findings (Kristianslund et al., 2007). A fourth-order Butterworth low-pass filter with 6 Hz cut-off frequency is commonly recommended to achieve this purpose (Bartlett, 2007). In the assessment of relative phase angle, significant amount of noise was still present when a 6 Hz filter was used. When a significant amount of noise is still present, usually a lower frequency is utilised to achieve a smoother kinematic data curve and this is usually left to the judgement of the researcher (Schreven et al., 2015). In this thesis, a 3 Hz and 1 Hz filters were trialled on the angular velocity. The angular velocity data were then plotted as a function of time. The comparison of which is shown below:

![Figure 1](image)

**Figure 1.** Angular velocity of the knee (blue), hip (grey), lumbar (yellow) and thorax (red) plotted as a function of time when a 3 Hz filter was used in a lifting task of a sample participant.
Figure 2. Angular velocity of the knee (blue), hip (grey), lumbar (yellow) and thorax (red) plotted as a function of time when a 1 Hz filter was used in a lifting task of a sample participant.

Figure 3. Normalised hip angular velocity plotted against normalised hip angular position using a 1 Hz filter (A) and a 3 Hz filter (B).

The utilisation of a 1 Hz filter on the angular velocity data would result in a relative phase angle curve identical to Burgess-Limerick et al. (1995) study. For this reason, a 1 Hz cut-off frequency was used to process the angular velocity data for the lifting task in Study 3.