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Effects of chronic low back pain on local dynamic stability and intersegment coordination during repetitive lifting

by

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Abstract

Background: Despite advancing developments in modern medicine and growing knowledge pertaining to spinal disorders, chronic low back pain (CLBP) remains the leading cause of disability worldwide. This is partly due to varying movement dysfunctions observed in people with CLBP. Although previous research has shown that people with CLBP exhibit altered motor control strategies, the literature is characterized by inconsistent findings. Thus, improved empirical assessment methods, such as intersegment coordination and local dynamic stability, have the potential to improve measurement of and detection of neuromuscular deficiencies exhibited in people with CLBP. While intersegment coordination and local dynamic stability measures have been used extensively in healthy people, there is a scarcity of research that has applied these measures to assess people with CLBP. Therefore, this thesis primarily aimed to compare movement and stability between CLBP and healthy people using intersegment coordination and local dynamic stability assessment.

Methods: Twelve participants with CLBP and 12 healthy participants performed one set of repetitive deadlifts for 35 repetitions with a dowel rod and one set with a barbell loaded with 15% of their bodyweight for 35 repetitions. Inertial Measurement Units (IMU) were used to measure trunk and lower limb kinematic parameters. Local dynamic stability and intersegment coordination were then calculated from the kinematic data and compared between the groups using a 2-factor repeated measures ANOVA.

Results: Significantly greater local dynamic stability of the hip and knee was observed in the CLBP group compared to the healthy control group. The CLBP group also reported increased low back pain immediately after completion of the loaded and un-loaded lifting trials. No differences in intersegment coordination or coordination variability were observed between

two groups. There were no differences observed in local dynamic stability and intersegment coordination when lifting with a load compared to lifting with no-load for both groups.

Conclusion: Our findings suggest that the significantly greater local dynamic stability of the hip and knee exhibited by the CLBP group may be mediated by pain avoidance. Local dynamic stability assesses both spatial and temporal characteristics of a lifting cycle, thus potentially providing greater insight into the differences in motor control of people with CLBP than intersegment coordination analysis. This thesis has shown that local dynamic stability is a measure which has potential clinical utility for monitoring changes in stability over time.

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Statement of Authentication

The work presented in this thesis is, to the best of my knowledge and belief, original except where due acknowledgement is made in text. I hereby declare that I have not submitted this material, either in full or in part, for a degree at this or any other institution.

John Marquez

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Abbreviations

APDM	Ambulatory Parkinson’s Disease Monitoring
ANOVA	Analysis of Variance
ASIS	Anterior superior iliac spine
BMI	Body mass index
CI	Confidence interval
CLBP	Chronic low back pain
CM	Centimetre
COV	Coefficient of variation
CRP	Continuous relative phase
CSV	Comma separated values
DP	Deviation phase
EMG	Electromyography
FABQ	Fear Avoidance Belief Questionnaire
FE	Flexion-extension
IC	Intersegment coordination

ICC	Intra class correlations
IMU	Inertial measurement unit
Kg	Kilogram
LBP	Low back pain
LDS	Local dynamic stability
MARP	Mean absolute relative phase
MVC	Maximum voluntary contraction
NY	New York
ODI	Oswestry Disability Index
OR	Oregon
PCS	Pain Catastrophizing Scale
ROM	Range of motion
SD	Standard deviation
SPSS	Statistical Package for Social Sciences
TSK	Tampa Scale of Kinesiophobia
VAS	Visual Analog Scale

Chapter 1: Introduction

1.1 Introduction

Chronic low back pain (CLBP) is the leading cause of activity limitation worldwide [1]. The direct economic cost of CLBP in Australia is estimated to be 4.8 billion dollars annually [2]. Additionally, CLBP is the primary reason for lost work productivity and early retirement in Australia [3]. Despite trends in increasing research and medical expenditure on CLBP, there has been a concurrent increase in chronicity and disability in people with CLBP [4]. Growing evidence suggests that current clinical assessment of CLBP is inconsistent with current literature, and consequently, poor diagnoses is prevalent amongst patient populations [5]. Considering that clinical practice is discordant with contemporary evidence, and the socio-economic burden caused by CLBP, there is a crucial need to improve clinical assessment to facilitate better treatment decisions and outcomes [4, 5].

Previously, it has been determined that individuals with CLBP alter their motor activity and control strategies to avoid painful movements and postures [6-10]. Such adaptive and protective strategies may affect spinal loading and compromise spinal control stability by decreasing damping and increasing the stiffness of trunk [7]. Currently, the mechanisms underpinning these observed differences in motor control in CLBP populations are not well understood [9]. This is in part due to the large variation in subjective pain ratings across individuals and that many disabling disorders are considered non-specific, i.e., there is no valid and objective diagnostic tool and no accurate or precise diagnosis for CLBP [10-12]. Given that clinicians currently rely on subjective pain ratings when assessing the effectiveness of their interventions, an improved empirical method of measuring spine stability and motor control

would allow for an objective assessment of impairment, which could then lead to enhanced treatment effectiveness in clinical and workplace settings.

Traditional methods of kinematic analysis are somewhat limited in their utility in assessing movement and movement control because they can only analyse kinematic data based on the temporal and spatial aspects of the coordinate data. As such, they do not provide an assessment of how coordination between body segments changes or develops throughout an entire movement cycle. The use of a dynamical systems approach could address this issue and allow for the analysis of both the spatial and temporal aspects of the coordinated movement patterns and thus, may help to identify and quantify neuromuscular deficiencies exhibited in CLBP patients [14]. Because dynamical systems analyses provide insight into how movement occurs or changes over time, and current trends in occupational life involve repetitive movement over extended periods of time, it appears appropriate to apply dynamical systems analysis to CLBP patients during repetitive lifting tasks [15].

To date, there is considerable evidence demonstrating differences in muscle activation, trunk alignment, posture, and movement in individuals with a history of CLBP compared to healthy individuals [6-10]. However, the literature is characterized by inconsistent findings [6-13]. This partly due to variations in experimental study design and the assessment methods used. In addition, the subjective and individualised nature of pain is problematic, with each patient possessing different structural, histochemical, and neuromuscular changes resulting in their level of pain [6-8]. Therefore, the diverse array of potential dysfunction within CLBP populations makes it difficult to determine if pain is causative or reflective of motor control impairments.

Changes in muscle activity of the spine have been proposed as a plausible mechanism for underlying CLBP [10]. Hodges et al. [10] examined the changes in mechanical properties of the trunk using an electromyography (EMG) driven model. The effective trunk stiffness, mass and damping were estimated using trunk kinematics and cable force. Equal weights (12–15% body weight) were attached to the front and back of the trunk via pulleys such that the trunk could move freely, and no muscle activity was required to hold the weights. The trunk was then perturbed by the unexpected release of one of the weights. Results indicated there was no difference in trunk displacement between both groups for forward perturbations. However, trunk stiffness was significantly greater in recurrent CLBP patients who also demonstrated significantly lower damping than healthy controls [10]. Contrary to clinical belief, trunk stiffness increased, most likely due to augmented trunk muscle activity and changes in reflex control of the trunk muscles [10].

This increase in trunk stiffness may explain why other researchers have demonstrated a delay in initiation of lumbar spine flexion following an unexpected perturbation in CLBP patients. Mok et al. [12] investigated the response to a sudden load dropped into the participant's hands in CLBP patients. Centre of pressure movement was measured to determine the compensatory postural adjustments that CLBP patients use to respond to sudden loads. No difference was observed in the amplitude of centre of pressure movement between groups. However, people with CLBP had a delayed initiation of lumbar spine flexion and took significantly longer to regain postural stability. This reduced efficiency of postural stability exhibited by CLBP patients is likely mediated by the increased stiffness of the trunk, thus favouring rotation around the ankle in CLBP patients [10].

Delayed initiation of the lumbar spine exhibited by CLBP populations may be explained by the increased activation of the shortened muscles of the antagonists. This has been supported by Marras et al. [11], who used an EMG-assisted model to evaluate spine loading in CLBP patients and asymptomatic individuals. Participants lifted various weights from five different origins varying in horizontal distance and vertical height from the spine. Patients with CLBP experienced significantly greater spine compression and shear forces when performing lifting tasks in comparison to asymptomatic individuals. These increases in spine loading were mediated by greater levels of antagonistic muscle coactivation [11].

Traditionally, kinematic analysis of the alterations in motor control of CLBP populations have been quantified using discrete measures such as the average of spatiotemporal measures and peak joint mechanics. These studies have not shown any differences between the two population groups when using the kinematic data alone [6-10]. Courbalay et al. [9] determined the extent that load expectations modulate neuromechanical adaptations in individuals with and without CLBP when lifting and lowering various loads. EMG analysis showed significantly lower vastus lateralis activity in the CLBP group during both the concentric and eccentric phase of the lift, and when lifting with a load. This difference may be explained by guarding behaviours used by people with CLBP to limit low back movement during repetitive lifting [9]. Despite demonstrating significantly lower muscle activity, kinematic displacement data did not reveal any significant difference between groups during the lifting phase.

Similarly, Larivière et al. [13] performed a 3D analysis involving the assessment of L5/S1 loading, posture of segments, inertial parameters, and EMG. No differences between the groups were observed for trunk and limb angular rotations, velocity, and acceleration. However, significant differences in the activation of the paraspinal muscles were observed. This finding

suggests that the altered motor control strategy used by CLBP patients may be a protective mechanism to avoid movement-evoked pain.

Collectively, the findings from these studies and the broader literature show that while there are significant differences in muscle onset times, discrete kinematic measures have shown inconsistent differences between healthy and CLBP patients [9-13]. The lack of observable difference in the present studies may be due to a lack of sensitivity of the discrete kinematic variables to differentiate between patient populations and controls. Therefore, it is evident that more sensitive methods of analysis of kinematic data are required to reveal movement impairments in CLBP populations during lifting [13].

A relatively new motor control and movement assessment approach that may enhance the diagnosis and assessment of CLBP patients is based on dynamical systems theory. The following section outlines dynamical systems theory and its potential use for assessing movement in CLBP.

1.1.1 Dynamical systems theory

Dynamical systems theory is a mathematical approach that is used to explain the behaviour of complex systems, including biological systems such as the human motor control system [16]. The human motor control system has multiple degrees of freedom which can produce an infinite number of coordinated movements to complete a given functional exercise. Dynamical systems theory operates on the premise that the number of degrees of freedom of the motor control system can be dramatically reduced through a process termed “self-organization” [14-17, 19]. Self-organization refers to the human motor control system’s ability to spontaneously organize itself into a coordination pattern which enables a functional movement to occur [18].

The dynamical systems approach describes these coordination patterns in terms of "attractor states". An attractor state refers to the specific stable states to which movement trajectories converge over time. In human movement, this refers to an individual's coordination tendency [17]. For example, every time human movement is initiated, our body organizes itself into an attractor state which allows the functional movement to occur. Therefore, attractor states allow humans to generate coordinated, and stable movements. This reduced dimensionality of the motor system encourages the development of functionally preferred attractor states to support human movement (i.e., the motor control system will use the best solution given the constraints on the system and the functional movement task) [17]. Traditional kinematic analysis cannot provide information on how human movement or motor control develops over time. Thus, using a dynamical systems approach could allow us to investigate if and how the motor control system of CLBP populations differ from healthy populations with respect to time.

Further, changes in motor control occur through transition from one stable attractor state to another [20]. These changes can be initiated by alterations in a control parameter (e.g., lifting with a load). Therefore, alterations in a control parameter can produce significantly different movement outcomes when using dynamical systems analyses.

1.1.2 Local dynamic stability

One measure within the dynamical systems theory paradigm that can explore how attractor states evolve with time is local dynamic stability (LDS). Quantified by the maximum finite-time Lyapunov exponent, LDS measures the sensitivity of a dynamic system to infinitesimally small perturbations that occur naturally, such as mechanical disturbances and neuromuscular control errors [22]. To explain this further, consider the trunk displacement (or its derivatives) during consecutive repetitions of a lifting task. During repetitive lifting, internal perturbations

such as variability in neuromuscular control and pain affect movement when performing a functional task. In ideal conditions without fatigue or injury, repetition by repetition variability is negligible. However, because of multiple sources of internal perturbation, such as neuromuscular control errors or pain, the trunk movement after each consecutive repetition is slightly different from that of the preceding one. In the presence of low LDS (indicated by a large positive maximum Lyapunov exponent), this difference increases exponentially with each subsequent repetition [19]. The maximum Lyapunov exponent estimates how fast a dynamic system (e.g., angular rotation of the trunk during lifting.) diverges after an infinitesimal internal perturbation [19]. Therefore, when applied to a repetitive lifting task, LDS can quantify the ability of the person to attenuate and recover from small perturbations arising from internal factors. As a more robust measure of stability during movement tasks, and due to it's potential to be introduced into clinical practice, there is growing interest and measurement of LDS amongst researchers [19].

1.1.3 Intersegment coordination

Another method of assessing attractor state stability is to assess intersegment coordination variability. Intersegment coordination (IC), quantified by the continuous relative phase (CRP), can provide a comprehensive description of a movement as it describes the interactions of segments that move a person through complex repetitive movements, including lifting tasks [15]. The mean absolute relative phase (MARP) is an average of a CRP curve over the duration of a movement [16, 17]. Lower values are interpreted as representing more in-phase associations while higher values are interpreted as indicating more anti-phase relationships. The deviation phase (DP), given by the mean standard deviation of the CRP, quantifies coordination variability. The DP provides information on the stability of an executed coordination pattern, as dynamical systems theory suggests that transitions between states of

stable coordination patterns are preceded by increased variability [16]. When using the relative phase approach, stable attractor state behaviour during repetitive lifting would be characterized by a low deviation phase of the thigh-shank, lumbar-thigh, or sternum-lumbar coupling [16].

Together, LDS and IC can be used as supplementary measures. While evaluating the attractor state's stability through the use of MARP and DP provides analysis of the spatial aspects of the data, LDS explores the temporal dynamics of an attractor state. By combining IC analyses with LDS assessment, data can be gathered on both the average coordination pattern (spatial) and how the coordination pattern changes with time (temporal).

1.2 Significance of the study

To date, LDS and IC have been used to successfully differentiate between biomechanically different lifting techniques, movement paces and directions, fatigued versus non-fatigued conditions, lifting heavy versus light loads, and experimentally induced-low back pain [18, 20-25]. While these studies have demonstrated that dynamical systems analyses are effective in explaining the behaviour of the human motor control system during repetitive lifting, the effectiveness of dynamical systems analysis to differentiate between healthy and CLBP populations remains to be observed. Therefore, the application of dynamical systems analysis to objectively explore and quantify any differences in movement in CLBP populations is warranted.

Knowledge obtained from this thesis has the potential to direct future research by determining the ability of dynamical systems analyses to discriminate between CLBP and healthy populations during repetitive lifting tasks. The results of this research may also translate to

clinical practice whereby dynamical systems analyses can be implemented to assess patient function and monitor the efficacy of treatment strategies.

1.2.1 Research Aims

The primary aim of this research was to assess whether LDS and IC can differentiate between CLBP populations and healthy populations when performing repeated lifting. A secondary aim was to examine the effect of increasing the load lifted during repeated lifting on LDS and IC in both CLBP and healthy populations. It was hypothesized that chronic LBP patients would demonstrate altered movement and control (lower IC and higher LDS) during lifting compared to healthy people.

1.3 Research hypotheses

1.3.1 Primary null hypotheses

(i) No difference in LDS and IC between people with and without CLBP would be observed during repetitive deadlifts.

(ii) No change in LDS and IC when increasing the load lifted for the CLBP and Control groups would be observed during repetitive deadlifts.

Chapter 2: A review of local dynamic stability of the trunk and spine

2.1 Introduction

Abnormal trunk and spine motor function is considered a risk factor for CLBP and injury [26]. Motor functioning can also be influenced by repetitive lifting through altered muscle recruitment and co-contraction patterns [27]. As such, an objective method of assessing motor function in CLBP populations during repetitive lifting is important for improving our knowledge and understanding of movement control in CLBP populations. One such method capable of assessing motor function in CLBP is local dynamic stability (LDS). Through the use of LDS, the time-dependant behaviour of the human motor control system can be measured. Therefore, the use of LDS to identify potential alterations in the motor function of CLBP populations due to internal perturbations appears appropriate [28].

To date, LDS has been used in gait research to predict fall risk in elderly people, and differentiate between people with osteoarthritis, multiple sclerosis, and fall-prone adults [29-33]. While LDS has shown to be an effective tool to monitor the aforementioned neurological pathologies, there is limited research on its use in CLBP populations.

2.2 Method

2.2.1 Rationale

To date, the mechanism of pain and its interaction with motor function of the spine and trunk in CLBP has not been well established [10]. It has been suggested that the altered motor functioning exhibited by CLBP populations is mediated by increased motor variability to reduce the effects of pain [34, 35]. Motor variability alters the strategies of movement organization in response to perturbation [34]. Local dynamic stability assessment enables the quantitative analysis of the time-dependant characteristics of the human motor control system due to the inherent motor variability associated with repetitive movement [34]. Therefore, LDS may have utility for assessment of stability in CLBP populations as it may provide a more accurate evaluation of motor variability and neuromuscular adaptation under perturbation. Thus, the purpose of this review was to explore the effectiveness and applicability of using LDS analysis in CLBP.

2.2.2 Search strategy

For this review, an electronic search of the Western Sydney University Library and PubMed databases were conducted (Jan 2022). The search terms used were combined and resulted in the following string: back OR hip OR spin* OR trunk OR torso AND dynamic stability AND repetitive. The purpose of the search was to find all articles in which LDS was used to measure stability in repetitive trunk flexion/extension tasks and repetitive lifting tasks.

2.2.3 Study selection

Studies were considered relevant for this review if: (1) the assessment of dynamic stability parameters was based on a repetitive task which included flexion/extension of the trunk; and

(2) LDS analysis was applied to the trunk and/or hip. Only full-texts in English indexed from 1970 to 2022 were included in the search. Based on the inclusion criteria, the selection process resulted in collation of 18 relevant articles. A manual search for relevant articles was also performed based on the reference lists of the retrieved articles. The manual search resulted in one additional study being added. Thus, a total of 19 experimental studies were included for review.

2.3 Discussion

2.3.1 Local dynamic stability methodology and methodological issues

Local dynamic stability, as quantified by the maximum finite-time Lyapunov exponent, measures the ability of an individual to attenuate local perturbations during repetitive movement [23]. These local perturbations may be attributed to neuromuscular noise, pain, or non-uniform kinematics from movement cycle to cycle [14]. The maximum finite-time Lyapunov exponent quantifies the rate of divergence of a cyclic kinematic trajectory such as trunk flexion or trunk rotation [36].

Previous research has suggested that increased LDS is typically associated with a lower risk of injury due to an increased ability of the motor control system to attenuate perturbations during repetitive movement [29-33]. Although LDS has gained popularity as method of biomechanical analysis in gait, there is limited research that focusses on the use of LDS in CLBP research [29-33]. This may be due to several methodological issues that need to be considered before LDS analysis can be used in CLBP populations.

It has been demonstrated that a higher number of repetitions during repetitive flexion/extension tasks increases the precision of LDS measurement [36]. However, a higher number of repetitions may cause muscle fatigue and an exacerbation of pain in CLBP populations, which would affect the LDS outcome as a confounding variable. While early research in repetitive lifting tasks used an arbitrary number of repetitions to calculate LDS [37], Dupeyron et al. [38] investigated the minimum amount of repetitions needed to obtain precise estimates of LDS during repetitive trunk flexion/extension. Participants performed 100 repetitions of a freestyle rhythmic trunk flexion/extension task [38]. Starting in an upright posture, participants were required to touch a target positioned at knee and shoulder height. Intra class correlations (ICC) and coefficient of variation (COV) analyses were used to quantify precision as a function of the number of repetitions analysed [38]. The authors determined that a minimum of 30 repetitions are needed to obtain precise estimates of LDS during repetitive flexion/extension tasks. This brings forth another methodological consideration when applying LDS analysis to CLBP populations – whether CLBP populations are able to perform 30 repetitive flexion-extension cycles of the trunk without fatigue and an exacerbation of pain. Therefore, when applying LDS analysis to CLBP populations, fatigue, pain, and load may act as confounding variables to the LDS measurement and should be considered during research study design.

2.3.2 Fatigue and local dynamic stability of the trunk

Muscle fatigue is defined as a progressive phenomenon which results in a decrease in maximal force or muscle power output in response to contractile activity [39]. Thus, muscle fatigue is the inability of recruited motor units to generate maximal force output [40, 41]. Mechanisms of muscle fatigue come from peripheral factors associated with maintaining muscle contraction (e.g., blood flow, oxygen delivery, contraction efficiency), and factors associated with maintaining central motor output to the muscle from the nervous system (e.g. cortical and

motoneuron output) [42]. Fatigue is a critical factor in CLBP patients as it can directly impact the rate of sensorimotor control of movement [44]. This can cause decreased proprioception, decreased kinesthesia, altered reflexes, increased muscle response time, and increased central processing [42]. During repetitive lifting, fatigue results in changes in force output and muscle recruitment patterns, which may lead to alterations in movement kinematics [41]. Therefore, continuous or prolonged execution of a dynamic task may result in the impairment of LDS. If fatigue impairs LDS, then small kinematic disturbances or neuromuscular control errors may cause brief uncontrolled intervertebral movement and subsequently increase the risk of tissue strain injury in CLBP populations [28-31]. Additionally, because current trends in occupational life include repetitive movement over long periods of time, muscle fatigue is an unavoidable short-term outcome of such activity [43]. This demonstrates the importance of investigating the effects of fatigue on LDS in CLBP populations.

A study by Asgari et al. [40] investigated the effect of fatigue on LDS in a CLBP population. They required 14 healthy participants and 14 participants with CLBP to perform repeated dumbbell lifting using a squatting technique with a dumbbell in each hand [40]. The dumbbells were loaded with a weight equivalent to 15% of the participants' body weight and were lifted from the floor with straight arms until the participant's hips were fully extended before being lowered to the floor [40]. The lifting task was repeated until the participant reported a score of 17 (very difficult) on the Borg rating scale of perceived exertion [40]. This was considered the highest safe level of fatigue and the task stopping point to reduce the risk of possible injury to the CLBP participants [40]. The results demonstrated that in the presence of self-reported muscle fatigue, as indicated by a high rate of perceive exertion, a significant decline in the LDS of the trunk and hip were observed for both groups [40]. This finding suggests that following substantial fatigue, both groups adopted an alternate movement strategy in which the nearest

trajectories of the trunk and hip increasingly moved apart, resulting in increased instability [40]. The authors propose that the decreased LDS caused by increasing fatigue may be the result of increased trunk flexion and leg extension or changes in hip velocity [44-46].

While the findings of Asgari et al. [40] showed that performing a repetitive lifting task to near exhaustion has a significant effect on LDS, the findings are limited in that the reduction in the force-producing capacity of the back muscles was not measured. Because of this, it is not possible to determine whether the findings were largely or entirely the result of muscle fatigue [40]. Therefore, due to methodological issues in their study, it is not possible to ascertain whether the decline in stability was caused by muscle fatigue. It is likely that objective measures of fatigue in which force output is measured are more appropriate for evaluating the effect of muscle fatigue on LDS.

Granata and Gottipati [47] examined the effect of trunk extensor fatigue on LDS of the trunk. Participants performed a repetitive dynamic trunk extension movement on a 45° Roman chair with a load cell attachment. Local dynamic stability was assessed after performing a maximal voluntary contraction (MVC) and when the participant could only sustain 60% of their MVC. They found fatigue significantly reduced trunk LDS. The observed decrease in LDS at 60% MVC suggests that the trunk was less stable in the presence of fatigue compared to the non-fatigued MVC condition. The protocol used within this study was more appropriate for assessing the effect of muscle fatigue than the aforementioned study because muscle fatigue was assessed as a function of MVC. These results demonstrate that the ability of the neuromuscular system to attenuate local perturbations is impaired by fatigue of the trunk extensor muscles during dynamic trunk movement [47].

Using a similar fatigue and movement protocol to Granata and Gottipatti [47], Larson et al. [48] examined the effect of muscle fatigue on trunk stability. In this study, dynamic stability trials were separated by 24 hours [48]. In contrast to the findings of Granata and Gottipatti [47], Larson et al. [48] found no significant difference in trunk LDS following the fatigue protocol [48]. The researchers performed further analysis on the data on a person-by-person basis. This post-hoc grouping revealed three distinct responses amongst the population in which the participant either stabilized (stability increases), destabilized (stability decreases) or no changers (stability is maintained) following the fatiguing protocol. Therefore, the researchers proposed that despite showing non-significant findings, the mean response of the sample population did not represent the true meaningful response of the individuals within the population because each group responded differently to the fatiguing protocol. Between these groups, there were no significant differences in pain (measured by Visual Analog Scale [VAS] scores), catastrophic thinking related to pain (measured by Pain Catastrophizing Scale [PCS] scores), or fear of movement (measured by Tampa Scale of Kinesiophobia [TSK] scores). The authors suggest that this finding demonstrates that the differing responses to fatigue were more likely dependent on differences in motor control strategies than the perception or presence of pain. Furthermore, the authors conclude that it appears that the perceived pain (measured by VAS) associated with muscle fatigue and its recovery could play an important role in determining how people control the motion of their backs.

The contrasting results between these two studies may have occurred due to different times in which the stability assessment was implemented. In the study by Granata and Gottipati [47], LDS was assessed before and immediately following a fatigue protocol (16 minutes apart), whereas in the latter study, the LDS assessments took place 24 hours apart [48]. In addition, the contrasting results between these studies may be due to improvements in the methodologies

used to compute LDS. Dupeyron et al. [38] previously determined that a minimum of 30 repeated trunk flexion/extension cycles were needed to calculate precise estimates of dynamic trunk stability. In the study by Granata and Gottipatti [47], only 20 cycles were analyzed as opposed to the 30 cycles in the study by Larson et al. [48].

Graham et al. [49] assessed the changes in trunk LDS resulting from 1.5 hours of repetitive automotive industry work. Assembly line tasks frequently involve forward trunk flexion. To assess LDS, directly before and after completing 1.5 hours of manufacturing work, workers performed 30 continuous trunk flexion/extension movements with a constrained pelvis. No significant difference in trunk LDS was observed before and after the work shift. Thus, 90 min of repetitive work involving static trunk flexion did not impair responses to local perturbations that occur naturally during movement. While hours spent on repetitive tasks during work shifts has been associated with the prevalence and development of CLBP, the authors propose that trunk instability is likely not a mechanism by which long-term repetitive industry work contributes to risk of developing CLBP [49, 50].

The findings by Graham et al. [49] are contrary to the findings by Granata and Gottipati [47]. One reason for this may be the method in which fatigue was induced. In the study by Granata and Gottipati [47], a very high level of muscle fatigue was induced in a localized muscle group, the trunk extensors. Conversely, due to the nature of the automotive assembly work in the study by Graham et al. [49], lower levels of non-localized global muscle fatigue were likely induced over longer periods of time. Further, the findings in the study by Graham et al. [49] are confounded as the authors were unable to directly measure the amount of muscle fatigue developed during the 90 min of work due to the manufacturing company-imposed 10-min for testing time constraint. Due to the limited amount of research on the effect fatigue on LDS, and

the conflicting nature of the findings in the existing literature, it is unknown if fatigue influences LDS during repetitive flexion/extension tasks.

2.3.3 External Load – Heavy versus light loads and local dynamic stability of the trunk

To facilitate stable trunk movement during lifting, humans must generate the appropriate motor patterns to control muscle force and to effectively respond to biomechanical perturbations and neuromuscular control errors [51]. One such factor that can influence trunk stability during lifting is the load being lifted [51]. Thus, it is important to understand how external loads affect the production of stable spine and trunk movements.

Graham et al. [23] assessed how altering the load lifted over 30 consecutive repetitions affected LDS of the trunk. Thirty healthy participants (15 male, 15 female) performed two trials of 30 continuous box lifts from a target positioned at half their standing height to a target on the floor. In the loaded trial, the box was filled with a weight equivalent to 10% of the participant's maximum back strength. There was no load inside the box for the un-loaded trial. The results showed a significant increase in trunk LDS when lifting a load compared to the un-loaded condition.

A follow-up study by Graham and Brown [24] also assessed the effect of load variation on LDS of the trunk. They required healthy participants (12 male) to perform repetitive box lifts between shoulder and knee height for 30 repetitions across 3 load conditions (0%, 5%, and 10% of the participant's maximum back strength). The box was placed on a platform located at knee height and lifted to a platform at shoulder height. This study differed to the earlier study by Graham et al. [23] in that the movement protocol was performed at a different range of motion. In the study by Graham and Brown [24], the repetitive box lifts were performed

between shoulder and knee height. Whereas in the study by Graham et al. [23], the box lifts were performed at half their standing height to a target on the floor. Muscle activity of the spine were also recorded during the trials via EMG. Despite the variance in lifting task range of motion, Graham et al. [23] and Graham and Brown [24] both found significant increases in trunk LDS when lifting with a load.

The results of both studies showed that the neuromuscular control of trunk stability was significantly affected by load, with an increase in LDS demonstrated when lifting a load equivalent to 10% of the participants maximum strength. This finding is supported by previous mechanical stability models which show that during movement with loads, mechanical spinal stability increased due to augmented muscular and moment demands and joint compression force [23]. Due to the increase in moment demands when lifting with heavier loads, there is a concordant increase in muscular activation. This causes a consequential increase in muscle stiffness due to the increase in number of activated cross-bridges [23], which translates into greater trunk stiffness [23]. Therefore, because the trunk is in a more mechanically stable state when lifting with a load, there would also be a decreased need for feedback-induced muscular contraction following a perturbation [23]. Further, the kinematic response of the trunk to a perturbation is determined through both the mechanical stability of the trunk prior to loading, as well as the reflex response of the muscles after loading [30]. Therefore, when lifting with a load, the increased mechanical stability prior to an internal perturbation decreases the divergence of the lifting cycle trajectory (i.e., greater LDS) immediately after that perturbation. This demonstrates that these findings are justified by previous static and quasi-static mechanical stability models that suggest the trunk may be more stable when lifting with heavier loads.

In a subsequent study, Graham and Brown [51] re-analysed the EMG data collected from their earlier study [24]. In this study, they attempted to apply LDS analysis to the spine EMG muscle activity. However, it is unknown whether applying the methods used to characterize the dynamic stability of kinematics to EMG signals is a valid measure of stability. They found that an increase in load lifted was found to have no significant effect on LDS of spine muscle activations during repetitive lifting [51]. An explanation for the contradictory findings between these two studies is that the measurement of the LDS from spine muscle activation data is not the same as LDS measured from kinematic data. More research is needed before LDS of muscle activations can be used to quantify neuromuscular control of stability. Additionally, within these studies, LDS was computed from a relatively low number of cycles (n=25), which is lower than the reported minimum number of repetitions required to achieve acceptable levels of precision for LDS analysis (n=30) [38]. The first 5 repetitions were not analysed in these studies to ensure steady-state movement behaviour. The authors state that use of a low number of cycles was implemented to limit the effects of fatigue. However, as suggested earlier in this review, more research is needed to determine whether fatigue has a statistically significant effect on LDS.

2.3.4 External Load - Unstable Loading and local dynamic stability of the trunk

Goal directed occupational tasks often require the movement or lifting of unstable loads (e.g., a pail of water), or the movement of stable loads under unstable support conditions (e.g., lifting on ice). Therefore, humans must generate the appropriate motor patterns to control joints in the presence of these unstable external conditions.

Within the scope of this review, only one study was found that investigated LDS during the lifting of an unstable load. Beaudette et al. [52] examined the effect of lifting a stable load (box

loaded with 8 kg) in comparison to an unstable load (3.25 kg of the 8kg box was replaced with water). Each lifting scenario consisted of a total of 23 consecutive lifts and involved raising and lowering a box from shelf heights based on anatomical landmarks (lower shelf at the level of tibial tuberosity, and upper shelf at the level of the anterior superior iliac spine (ASIS). Relative to the stable control lifting trial, the unstable load had no significant effect on trunk LDS [52]. When adding an unstable surface (performing the task while standing on a Bosu ball) in addition to the unstable load, there was still no significant effect on LDS [52]. The use of EMG within the study allowed the authors to determine that when the external level of instability is increased, individuals contracted their muscles to stiffen the lumbar spine and to therefore maintain a consistent level of trunk LDS [52]. Due to the stiffening effects during the unstable support condition, there was no observable trend towards a decrease in LDS between the stable and unstable load conditions.

The non-significant main effect when lifting with an unstable load suggests that the LDS of the trunk is conserved by trunk muscle stiffening effects. Thus, the authors propose that it is possible LDS is monitored by the central nervous system during a lifting task to facilitate completion of the task and that external instability-induced perturbations are managed by the central nervous system. Because only one study was found for the effect of unstable loading on trunk LDS, a reliable conclusion cannot be drawn.

2.3.5 Speed of movement and local dynamic stability of the trunk

The pace at which a task is performed at is a relevant occupational factor and a control parameter which has been shown to influence temporal movement strategies during repetitive tasks [53]. More specifically, higher pace has been associated with more variability and errors during repetitive assembly work [54].

Granata and England [55] investigated the effect of movement speed on trunk LDS. In this study, 20 healthy participants (8 male, 12 female) were required to touch two targets with their hands held together under two different speed conditions. The slow condition consisted of 20 repetitions per minute for 90 seconds, and the fast condition consisted of 40 repetitions per minute for 45 seconds, for a total of 30 repetitions in both conditions [55]. Targets were located at two pre-specified locations. One target was placed at shoulder height in the anterior midline so that it could be reached when standing upright with the arms horizontally extended and the second target was placed 50 cm anterior to the knee. Participants were required to touch the upper target followed by the lower target repeatedly for the duration of each trial. The results showed that the participants had significantly lower LDS when they performed the repetitive trunk flexion/extension task at 40 repetitions per minute in comparison to when they performed the task at 20 repetitions per minute. The authors proposed several mechanisms for the decrease in LDS with an increase in speed. First, torso muscle activity and co-contraction increased with trunk velocity and acceleration [55]. Henneman's size principle dictates that modulation of muscles when muscle activity is high requires the recruitment of large motor units, thereby limiting fine motor control during fast paced movements. Second, momentum increases with velocity. Therefore, more neuromuscular effort was required to control and attenuate kinematic disturbances. Third, fast dynamic movements reduce the allowable time for neuromuscular corrections, which suggested increased delay in the active recruitment and neural feedback relative to movement trajectory [55].

Asgari et al. [56] replicated the study by Granata and England [55] with the inclusion of a third speed condition in which participants were allowed to perform the lifting task at a self-selected speed. Similarly, they found a significant decrease in LDS when performing a repetitive trunk flexion/extension task at progressively increasing speed [55, 56].

In contrast to the studies listed above, Graham and Brown [24] found that increasing speed did not affect trunk LDS. In their study, three experimental trials were performed in which participants performed 30 continuous freestyle box lifts (lifting 5% of their maximum back strength) at three different rates (6, 12, and 18 repetitions per minute). The box was lifted from knee height to shoulder level. A possible explanation for the contrasting findings may be due to the lower number of repetitions performed per minute (6, 12, and 18/min), and thus, the slower the speed of movement [24]. Granata and England [55], Asgari et al. [56], and Graham and Brown [24], all used 30 repetitions for their movement protocol, however, the time to complete the 30 repetitions varied. It is possible that the effect of movement speed on LDS may only become significant at faster cadences as demonstrated by Asgari et al. [56] and Granata and England [55]. However, it is important to note that the findings by Graham and Brown [24] may be better extrapolated to activities of daily living that have repetition akin with lower frequencies and movement speeds.

In a subsequent study, Graham and Brown [51] reanalysed their data and applied LDS analysis to spine muscle activations. They found that with an increase in lifting rate with a constant load there was a significant decrease in the LDS of muscle spine activations. These results contrast those from their previous study, where the increase in lifting rate with a constant load did not change the LDS of trunk kinematics [24]. The authors concluded that this showed that under the changing rate condition, participants were less able to maintain stable spine muscle activity [24]. However, as previously stated in this review, it is unknown whether calculation of LDS of spine muscle activations is a valid measure of stability.

The literature identified in this review suggests that there may be a linear relationship between LDS of the trunk and movement speed, with increases in movement speed resulting in decreases in LDS. At lower movement speeds, which are comparable to those of natural movements that would occur during daily living activities, the decline in LDS with movement speed is not significant.

2.3.6 Task asymmetry and local dynamic stability of the trunk

Asymmetrical movement and loading patterns have been proposed as contributing factors to the onset of CLBP symptoms [57, 58]. Spinal movement asymmetries during walking have also been hypothesised as possibly contributing to CLBP [59]. Additionally, workers involved in repetitive lifting, bending, asymmetrical postures, and manual handling are considered at high-risk of developing CLBP [60]. Occupational tasks and activities of daily living often involve asymmetrical lifting (i.e., movements that include components in both the sagittal and transverse planes).

Granata and England [37] examined the effect of asymmetric lifting on trunk LDS. They compared an asymmetric lifting task (each movement included rotation from left twist to right twist as well as flexion-extension) with a symmetric lifting task (flexion-extension). In the symmetric condition, participants were required to touch targets at shoulder and knee height in the anterior midline. Whereas in the asymmetric condition, the shoulder target was moved to the right and the knee target was moved to the left to induce a 45° axial rotation of the torso at the upper and lower targets. A lower-limb constraint was imposed in all experimental conditions by strapping the subject's legs and pelvis to a rigid structure to restrict movement of the lower limbs. Each movement cycle consisted of both the eccentric and concentric phase of the movement. Participants performed 30 movement cycles per trial. The results

demonstrated that the symmetric movements in the mid-sagittal plane were associated with significantly lower LDS than asymmetric trials [37].

Granata and Gottipati [47] investigated the effect of task asymmetry and fatigue on LDS of the trunk. Participants performed a repeated flexion/extension task in which they touched a target placed near knee level with their hands then returned to an upright posture at 30 flexion/extension cycles per minute until the participant had fatigued to 60% of their MVC. During the symmetric condition, the participants touched the target with both hands. Whereas during the asymmetric trials, they touched the target with their dominant hand only. There was no significant main effect of asymmetry on LDS of the trunk. It is possible that no main effect was observed for asymmetry in this study because the movement cycles predominantly remained in the sagittal plane, hence the dependant variable of task asymmetry was not appropriately assessed. Specifically, the asymmetric movement protocol used by Granata and Gottipati [47] required each participant to touch a target in the mid-sagittal plane using only their dominant hand, whereas in the study by Granata and England [37], the target was placed to the left or right of the mid-sagittal plane. Additionally, a lower limb constraint was not used in the study by Granata and Gottipati [47] which would result in additional movement of the hips and knees when performing the trunk flexion/extension task as opposed to the study by Granata and England [37] where a lower limb constraint was used.

When movement of the lower limbs was appropriately asymmetrical, there was a trend wherein asymmetric tasks resulted in greater trunk dynamic stability than symmetric tasks. The authors propose that this may be mediated by increased recruitment and coactivation of the internal and external oblique muscle groups during asymmetric lifting [37, 47]. The recruitment of these muscles is critical to control asymmetric tasks, whereas activation of these muscles is less

important when lifting in the mid-sagittal plane. No effect of asymmetry was observed when subjects were free standing in the latter study [47]. The authors propose that this interaction may indicate that dynamic coupling between the legs and torso contributes to the control of stability in asymmetric movements [47].

Dupeyron et al. [38] investigated the effect of a symmetric lifting task compared to an asymmetric lifting task. Participants performed 100 repetitions of trunk movements in flexion, trunk rotation, and in a combination of flexion and trunk rotation. Participants were required to perform a freestyle rhythmic trunk flexion task between an upright standing position into a flexed position in which both index fingers touched horizontal targets positioned at knee and shoulder height. For the trunk rotation task, participants performed a rhythmic pointing task alternating with the right and left hand in upright stance moving between vertical targets positioned bilaterally at shoulder height and one arm length laterally. For the combined task, participants were asked to successfully touch four targets in the following consecutive order: knee height on the left, shoulder height on the right, shoulder height on the left, and knee height on the right. There was a main effect of task and a significant interaction with the segment analysed. Movements in the sagittal plane were significantly less stable than combined movements in the sagittal and horizontal plane and in the horizontal plane only [38]. Results from this study align with the previous findings by Granata et al. [37, 47]. Dupeyron et al. [38] proposed that the effect observed in their study may be due to greater trunk muscle co-contraction in tasks involving twisting moments. Similarly, Granata and England [37] had previously emphasized the stabilizing effect of the activity of the oblique muscles in asymmetrical tasks.

Graham et al. [61] investigated the effect of task asymmetry in varsity athletes with and without CLBP. To match previous studies, the movement protocol used within this study was the same as the one used by Granata and England [37]. Participants performed 30 repetitions at a rate of 15 repetitions per minute. They found no significant difference in LDS between the healthy and CLBP groups in both symmetric and asymmetric lifting tasks. In agreement with previous studies, trunk LDS was significantly greater when moving asymmetrically. The use of EMG within this study confirmed the previous hypothesis posed by Dupeyron et al. [38] and Granata and England [37], that the increased stability during asymmetric tasks is due to increased trunk muscle co-contraction in tasks involving twisting and lateral bending movements due to oblique muscle activation. Trunk twisting coincided with higher levels of trunk muscle co-contraction, due to the lateral effects that the oblique abdominal muscles had in other planes [38]. The findings by Graham et al. [61] demonstrated that co-contraction was significantly increased during the asymmetrical tasks, corresponding to increased dynamic stability of kinematics in both healthy and CLBP participants.

Lee and Nussbaum [62] investigated the effect of task asymmetry on trunk LDS between experienced manual handling workers (minimum of 3 years recent experience in frequent lifting tasks) and novice workers. Participants completed a set of flexion/extension tasks in two conditions (0° symmetric and 60° asymmetric) [62]. A box was lifted and lowered (loaded with 10% of the participant's body weight) 20 times in each set [62]. The lifting origin and destination heights were adjusted to individual knee and elbow heights [62]. Experienced workers had significantly greater LDS than non-experienced workers during the symmetric trial [62]. This finding suggests that LDS in symmetric repetitive lifting tasks can be 'trained' or improved over time. There were no differences between experienced workers and novices during the asymmetric trial [62]. Therefore, the balance maintenance and torso movement

stability among novice workers may be dependent on the task conditions (asymmetry versus symmetry). The authors proposed that the explanation for this difference between task conditions was that experienced workers seemed to adjust torso kinematics/kinetics to maintain stable balance and torso movement, whereas novices may have de-emphasized balance and stability to obtain relatively consistent torso kinematic exposures [62].

2.3.7 Low back pain and local dynamic stability of the trunk

The above studies and findings demonstrate that the assessment of LDS has utility for evaluating the effects of load, speed, and fatigue in individuals without pathology during lifting. However, there is limited research that has applied this approach on patients with CLBP. As indicated above, Graham et al. [61] compared varsity athletes with and without CLBP during a repetitive flexion/extension task. No significant difference was observed in trunk LDS between the healthy and CLBP groups. However, this finding cannot be extrapolated to the general population with CLBP pathology because the sample comprised young varsity athletes. Despite meeting criteria for having CLBP, the authors report that these athletes were reportedly not challenged by activities of daily living or the repetitive movement protocol within the study. Additionally, the results cannot be extrapolated to occupational settings or daily living activities which involve lifting because no load was used in the trunk flexion/extension task.

Asgari et al. [40] also examined the effect of CLBP on LDS of the trunk, hip, and knee. This study required 28 participants (14 with CLBP, and 14 healthy participants) to hold two-dumbbells in their hands (loaded with 15% of their bodyweight) and lift the dumbbells from the floor with straight arms until the participant's hips were fully extended before being lowered to the floor. Lift cycles were repeated until participants reached and reported a score of 17 on the Borg scale. As stated earlier in this review, it is possible that confounding factors,

such as fatigue, affected trunk stability and coordination in this study [19]. The results showed that the CLBP group had significantly more stable hip movement in the frontal and transverse planes in comparison to healthy controls. There were no significant differences in trunk, knee, or ankle dynamic stability between groups. It is possible that when performing repetitive lifts until a score of 17 on the Borg scale is reached, a reduction in the force-producing capabilities of the trunk extensors may have affected both groups. Thus, there is a scarcity of research that has been performed on CLBP groups which use LDS as a criterion measure [40, 61].

2.3.8 Section Conclusion

This review aimed to provide an analysis of the application of LDS assessment in the trunk and hip regions. The review findings have shown that LDS is affected by movement speed, loads, fatigue, and task asymmetry during repetitive lifting. This review also provided methodological considerations for future studies on CLBP populations including: the total number of repetitions and its influence on the accuracy of the LDS measurement, the effect of load lifted during lifting tasks and the ability of CLBP populations to perform the lift, the effect the speed at which participants perform repetitive lifting, and the effect of task asymmetry during repetitive lifting movements. From this review, it is evident that LDS research in the context of the trunk and hip is limited, and more research needs to be directed towards investigating the differences in LDS between healthy and CLBP populations.

Chapter 3: A review of intersegment coordination of the trunk and lower limbs

3.1 Introduction

Functional movements, such as repetitive lifting, are complex activities that require the coordination of the trunk as well as the upper and lower limbs. Because repetitive lifting involves multiple body segments, intersegment coordination (IC) analysis can provide useful information about movement and motor variability, which reflects the consistency of the intersegmental relationship between segments during lifting [16, 17, 63]. Previously, the variability of trunk motion kinematics during lifting in CLBP populations has been studied as one segment acting in isolation [63, 64]. These studies have quantified the kinematics of CLBP populations by measuring lumbar and hip range of motion and angular velocity. However, these studies report inconsistent findings including increased, decreased and no difference in ROM between CLBP and healthy people [13, 63, 64]. Thus, an alternate method of analysis of lifting in CLBP populations may be needed to determine potential deficits in motor control.

Considering that repetitive lifting tasks involve multiple segment coordination throughout movement, the application of IC analysis could provide more accurate assessment of trunk and lower limb lifting deficits in CLBP populations. In the past, IC analysis has been utilized to incorporate angular position and angular velocity information over an entire motion cycle to compliment kinematic analyses [14-16]. While some investigators have examined the IC between the trunk and lower limbs during active movement of the trunk, there is limited research applying this analysis to examine the aberrant patterns of coordination in CLBP

populations [65, 66]. While discrete kinematic analysis provides insight into peak postures, IC aids in understanding how these postures change with time, and how segments move in relationship to one another over time [16, 17].

The aim of the review in this section was to examine the IC literature and establish the effectiveness and applicability of using IC analysis in CLBP populations.

3.2 Method

3.2.1 Rationale

Motor variability is fundamental to human movement and is essential to musculoskeletal health in CLBP populations [10]. While it has been suggested that people with CLBP alter their motor variability to reduce the effect of pain, the evidence to date has been inconclusive [13, 63, 64]. Additionally, the mechanism of pain and its interaction with motor variability in CLBP has not been established [10]. While traditional kinematic analyses of CLBP movement have considered the trunk as one segment, most functional movements and daily living activities involve multiple segment movement and coordination [66]. Intersegment coordination, quantified by continuous relative phase (CRP), can provide important information on how segments interact relative to each other, as well as the consistency of the intersegmental relationship. It has been proposed by previous authors that IC may be useful in determining impairments in movement coordination and pattern stability in CLBP populations [66]. Therefore, IC analysis could be useful for showing aberrant coordination patterns in CLBP populations and may provide better evaluation of motor variability than traditional kinematic analyses.

3.2.2 Search strategy

In this review, an electronic search was conducted in the Western Sydney University Library database, and PubMed databases (Jan 2022). The search terms were combined and resulted in the following string: back OR hip OR spin* OR trunk OR torso OR pelv* AND intersegment coordination OR relative phase OR segment* coordination OR inter* coordination OR phase angle AND repetitive. The purpose of the search was to find all articles in which IC was used to measure stability in repetitive trunk flexion/extension tasks.

3.2.3 Study selection

Studies were considered relevant for inclusion in this review if: (1) the assessment of IC was based on a task which included repetitive flexion/extension of the trunk; and (2) continuous relative phase analysis was applied to the trunk and/or hip. Only full-texts in English indexed from 1970 to 2022 were included in the search. A manual search for relevant articles was also performed based on the reference lists of the retrieved articles. The manual search resulted in 3 additional studies being added, resulting in a total of 13 experimental studies.

3.3 Discussion

3.3.1 Intersegment coordination methodology and methodological considerations

Intersegment coordination, as measured by continuous relative phase, quantifies the coordination pattern and the variability of the coordination pattern [23]. This approach provides continuous spatial measurement throughout the entire movement cycle and are derived from segment rotation and velocity [23]. Continuous relative phase shows the phase relationship between two segments [16]. During in-phase or rigid movements, bilateral homologous muscle groups contract synchronously, resulting in the segments moving in synchronization with one

another [67]. Whereas in anti-phase movements, the muscles will contract and relax at opposite times [67]. For example, a distal to proximal thigh and trunk coordination pattern during repetitive lifting would indicate that the distal segment (thigh) is leading the proximal segment (trunk) throughout the movement.

Previous research has demonstrated aberrant coordination patterns and variability in CLBP populations during repetitive movement such as walking, running, forward reaching, and axial rotation [68-71]. Although IC analysis has gained popularity as method of biomechanical analysis in gait, there is limited research that focusses on the use of IC in CLBP research [15, 16, 68]. This may be due to several methodological limitations that need to be considered before applying IC analysis to a repetitive lifting task.

During repetitive lifting, movement variability could be viewed as healthy and essential for optimal flexibility and stability [8]. However, significantly increased variability could result from an individual's inability to use a stable motor patterning to execute the repetitive task and therefore represent functional deficits in motor control [68, 69]. Similarly, significantly decreased variability could also represent a pathological state with limited movement options [68, 69]. Therefore, it would be difficult to determine whether the changes in IC are beneficial or representative of a deficit in motor control.

Another methodological consideration for using IC analyses during lifting is the effect of lifting with a load. It has been established that repetitive lifting with a load can result in increased movement variability due to muscle fatigue [72]. Several papers have shown that as the load lifted is increased, lumbar spine motion tends to lag further behind the lower limb joints [72-

74]. However, in CLBP populations, repetitive lifting with a load may cause an exacerbation of pain, which may also affect IC [40]. Therefore, load and muscle fatigue may act as confounding variables and make it difficult to determine whether observed differences in IC are caused by pain.

Intersegment coordination patterns also appear to vary between sexes during repetitive lifting [75, 76]. It has been demonstrated that males produce higher spine loads than females when performing identical lifting tasks [75]. To date, CLBP motor control research has mostly included both sexes and have not adjusted loads relative to the participant's strength [75]. While repetitive lifting with absolute loads produces greater external validity regarding occupational settings and activities of daily living (i.e., males and females are required to lift the same loads in the work setting and during daily living activities), adjusting the load lifted for the participant's relative strength results in greater internal validity regarding controlling for the biological strength differences between sexes.

Therefore, when applying IC analysis to CLBP populations, load, sex, and fatigue may act as confounding variables to the measurement of IC and should be carefully considered during research study design. This review will now explore the literature surrounding the effect of load, sex, and fatigue on IC during repetitive trunk flexion/extension tasks in greater detail.

3.3.2 External load, lifting origin/destination and intersegment coordination

Current literature has shown conflicting findings on the effect of load on IC. Burgess-Limerick et al. [74] investigated the effect of increasing load on IC in healthy populations. Thirty-nine healthy participants were required to lift a load on to a shelf 25 times [74]. Five different loads,

starting from 2.5 kg with increasing increments of 2 kg up to 10.5 kg, were examined. The variability of IC was found to significantly increase with each progressive increase in load. The authors concluded that the observed changes in coordination are a functional adaptation to reduce the muscular effort required to complete the lifting task [74].

Hu et al. [77] investigated the influence of load on lumbar-pelvis coordination during repetitive box lifts in a healthy population. Twelve male subjects performed repetitive box lifts from the floor under 2 load conditions: with 20 lb inside the box, and no load inside the box. The results showed that lifting the 20 lb load resulted in more in-phase lumbar-pelvic coordination compared to lifting with no load. The authors proposed that when lifting with a load, healthy individuals adopt a more in-phase and guarded lumbar-pelvis motion patterns to protect themselves from injury [77]. As previous studies have shown CLBP populations tend to demonstrate more protective motions to reduce the risk of injury [8-12], the findings from Hu et al. [77] suggest that when lifting with a load, altered lumbar-pelvic coordination could be expected to be more pronounced in CLBP populations.

Mokhtarinia et al. [65] investigated the effect of load variation on IC of the lumbar-pelvis and pelvis-thigh. Fifteen participants with CLBP, and 18 healthy participants performed a repetitive trunk flexion-extension task from a standing position and touched a target at knee height with and without wearing an 8 kg uniformly loaded vest. Further demonstrating the inconsistency in the current literature, the authors found no difference in lumbar-pelvis coordination when lifting with a load compared to a no-load condition [65]. The non-significant difference in lumbar-pelvis coordination opposes the findings by Hu et al. [77] and Burgess-Limerick et al. [74]. This may be explained by the varying experimental design for trunk loading. In the studies by Hu et al. [77] and Burgess-Limerick et al. [74], a load was held in the participants hand

rather than wearing a weighted vest. Despite no difference in lumbar-pelvis coordination when lifting with a load, there was a significant reduction in in-phase movement of the thigh and pelvis when lifting with a load.

The conflicting nature of the existing literature investigating the effect of load lifted on IC may also be explained by the different relative loads used between studies. Scholz [73] examined the effect of lifting loads, prescribed as a percentage of the participant's MVC, on IC of the knee, hip and lumbar during repetitive flexion/extension lifting [73]. MVC was assessed through a maximum-effort isometric squat lift [73]. During experimental trials participants lifted a weighted box starting with an initial load of 15% of their maximum lifting capacity. Progressive trials were performed with the load increasing by 15% increments up to 75%. They found continuous significant decreases in knee-lumbar and knee-hip IC as the load increased from 15% to 75% of MVC. As relative load increased, movement of the knee-lumbar IC was less in-phase, and knee extension led back extension at a faster rate.

A later study by the same author demonstrated similar findings [78]. When maximum lifting capacity was determined by maximum-effort lifting against a load cell, an increase in relative load from 15% to 75% resulted in more distal to proximal lower-limb and trunk coordination patterns, despite no significant changes in peak displacement angles of the trunk and hip during the lifts [78]. These findings suggest that individualising load based on the individual capacity of each participant may be needed to elicit an effect on IC.

Considering that there is limited research on the effect of load on IC of the trunk, another possible explanation for the conflicting research is the variance in lifting origin between studies. Intersegment coordination patterns during repetitive lifting are shown to vary

depending on the origin that the object is lifted from, and the height that the objected is lifted. Splittstoesser et al. [79] investigated the effect of altering the lifting origin and destination on IC of the trunk. Lower lifting origins were found to be associated with more in-phase lifting patterns between the trunk and hip [79]. During repetitive lifting, as the height lifted from the floor increased (at 0 cm, 19 cm, 38 cm, 57 cm, and 76 cm from the floor), IC was shown to significantly decrease [79]. The significantly more in-phase lifting pattern at lower lifting heights may serve as a protective mechanism, due to the increased peak L5/S1 resultant moments at lower origin lifting heights [79]. Previous literature has supported the idea that increased peak L5/S1 moments are a significant risk factor for the development of CLBP [7-12]. Consequently, future studies that apply IC analysis to CLBP populations should control for lifting origin if there are multiple task conditions or trials. Additionally, when applying IC analysis to a repetitive lifting task, a lifting origin closer to the floor will result in the most in-phase coordination pattern and will reduce the confounding effect of the origin height of the load.

3.3.3 Fatigue and intersegment coordination

Fatigue is another factor that has been shown to influence IC [44, 80]. Sparto et al. [44] investigated the effect of fatigue on IC of the trunk, hip, and knee. Twelve healthy participants performed a repetitive box lift from the floor to full extension of the hips. Prior to performing the lifting task, each participant's maximum lifting capacity was tested. Experimental lifting trials were then performed at 25% of their maximum lifting capacity at a self-selected pace until the participant could no longer continue [44]. Results demonstrated a decrease in hip and knee ROM, an increase in spine peak flexion angle, and a decrease in hip-lumbar inter-joint coordination with increasing fatigue. This change in coordination may be a functional adaptation towards a more physiologically demanding movement pattern. As the participant's

fatigue, they may adopt a less efficient movement pattern and place less emphasis on the muscles of the lower limbs and increase back extensor demand to allow for continuation of the task.

Van Dieën et al. [80] also examined the effect of fatigue on IC during repetitive lifting. Five healthy men performed 75 repetitions of the deadlift with an 8 kg barbell (15 repetitions/minute for 5 minutes). IC was measured at the lumbosacral joint, hip, knee, and ankle. The participants performed the deadlift using two different types of techniques. The first was a squat-style deadlift which minimized trunk flexion, and the second was stoop-lift that minimized flexion of the knee joint. The findings demonstrated that the IC did not change across the repetitive lifts, except between the knee and hip in the squat-style deadlift. The nonsignificant findings may be explained by a lack of fatigue induced by the movement protocol. While the authors relied on the subjective reports of the participants to confirm fatigue was present, muscle fatigue was not objectively measured. Additionally, due to the small sample size (n = 5), it is likely that the study was statistically underpowered. If muscle fatigue was induced across all participants, these findings suggest that the inter-joint coordination pattern is adaptable and can change to maintain a consistent performance level to meet the demands of the task [80].

A possible explanation for the varying results between the study by van Dieën et al. [80] and Sparto et al. [44], may be due to the different loads used to induce fatigue. Sparto et al. [44] used a heavier load, equivalent to 25% of each participant's maximum lifting capacity, whereas in the van Dieën et al. [80], a standard load of 8 kg was used across all participants [49, 50]. It is plausible that fatigue produced by heavier load (high intensity) short-duration motor tasks will result in a reduction IC. Whereas, in less strenuous tasks, the neuromuscular system can adapt to the demands of the task which would result in no change in IC.

The current literature suggests that fatigue is likely to result in modified IC during repetitive lifting [44]. High intensity fatiguing lifting protocols are shown to result in declines in force output, and results in an increase in perceived effort, which can then lead to an increase in movement complexity [44]. It is possible that the changes in IC due to muscle fatigue may follow a pattern like that observed when individuals learn new motor tasks as people begin to organize multiple degrees of freedom in a new fatigued state [81]. Since high-intensity fatiguing protocols may be a confounding variable in IC, and to reduce the risk of injury to the CLBP group, the study in this thesis will limit the effect of fatigue by using a task with a lighter load comparable to those of past literature.

3.3.4 Sex and intersegment coordination

Differences in body size and physical capacity between males and females has the potential to influence lifting strategies and performance during repetitive lifting. Sex-related differences in IC patterns have been observed during lifting tasks [82]. Lindbeck and Kjellberg [82] investigated the difference in IC patterns between males and females in a lifting task. When compared to males, female IC was more synchronous and had less variability when lifting 12.8 kg and 8.7 kg boxes from the floor to 61% of stature, or chest height [82]. The more in-phase coordination pattern exhibited by females may be caused by inherent biological differences in strength compared to males. It is possible that the lifting task was not physically demanding enough for the males to use a rigid coordination pattern in the study by Lindbeck and Kjellberg [82]. While the findings by Lindbeck and Kjellberg [82] suggest that females adopted a more efficient coordination pattern for this specific task, the biological differences in strength between males and females were not controlled for in this study. Therefore, it is possible that the confounding effect of strength affected their findings.

In contrast to Lindbeck and Kjellberg [82], Plamondon et al. [75] showed that experienced female manual workers displayed significantly decreased IC of the hip and trunk during lifting than experienced males and novices in a repetitive manual handling task. The task consisted of lifting 24 boxes weighing 15 kg from one pallet to another at a self-determined pace and then at a pace of 9 lifts per minute. Due to a lack of comparison to novice females, it is unknown if women tend to adopt a distal-proximal coordination pattern in general, or if this was a learned adaptation to manual handling experience. Using the same protocol, a follow-up study by Plamondon et al. [76], controlled for the biological differences in strength to investigate the differences in IC using relative loads (10 kg for females, 15 kg for males). Significantly reduced IC exhibited by the females was again observed compared to males, According to Plamondon et al. [76], this finding supports an influence of factors intrinsically linked to sex, such as the biological differences in strength and the anatomical differences in the pelvis. They concluded that it is possible that the sex difference in the strength ratio of the hip extensors/back extensors between men and women may explain the reduced IC exhibited by females.

To date, most motor control research on CLBP populations has included males and females. However, the IC literature suggests that there may be intrinsic differences between males and females that affect their lifting performance and kinematics. Therefore, when assessing intersegment coordination during repetitive lifting tasks, controlling for biological differences between males and females to reduce the confounding effects of sex on IC appears prudent. This can be achieved by adjusting the load lifted based on sex or by each participant's individual capacity.

3.3.5 Low back pain and intersegment coordination

The severity of self-reported disability due to CLBP has been shown to affect IC of the trunk and lower limbs during lifting. Pranata et al. [66] compared patients with lower disability and patients with high-disability CLBP while lifting an 8 kg kettlebell. Participant disability was measured through self-reported Oswestry Disability Index (ODI) scores. Patients with higher disability demonstrated significantly decreased lumbar, hip and knee inter-joint coordination in comparison to people with lower disability and healthy controls. This indicates less synchronous coordination patterns by the higher disability group. These findings suggest that it is possible that CLBP patients with high disability have a neuromuscular system that is adaptable to task/environmental demands and load and can alter IC in order to perform functional movement with less pain [66]. Alternatively, the findings could also reflect more inflexible motor behaviour that is less adaptable to task/environmental demands and load [66]. Due to the standardized load (8 kg) used across all patients, it is likely that the confounding effect of load may have affected the findings in this study.

Similarly, Silfies et al. [70] found that lumbo-pelvic IC was less synchronous in CLBP patients than healthy controls during a repetitive flexion/extension task. Thirty CLBP patients and 35 healthy people performed repeated flexion-extension trials in which they were to touch a stationary target at their shoulder height followed by a return to a standing position. The target was placed anterior to their body at a distance equivalent to 50% of their maximum functional reach to allow for trunk flexion and extension to occur. Trunk flexion and extension motion was standardized to 6 seconds (3 seconds of flexion, and 3 seconds of extension). The authors proposed that a plausible explanation for the pattern change is an adapted motor plan of co-contraction that attempts to restrict motion at the lumbar spine. Additionally, they found that CLBP patients exhibited significantly greater trial-to-trial pattern variability (indicated by

deviation phase) during the eccentric portion of the movement. The authors state that the increased trial-to-trial pattern variability may have been caused by trunk extensor muscle dysfunction. The reduced pattern repeatability exhibited by CLBP populations is consistent with previous research findings that have shown reduced cross-sectional area, strength, and endurance of trunk extensors in CLBP populations (7-12). Additionally, the findings of this study are limited in that the movement protocol lacked external validity. This is because movement during occupational settings and daily living activities are not performed at the cadences used within this study (3 seconds of flexion, 3 seconds of extension).

Mokhtarinia et al. [65] reported no significant difference in lumbar-pelvis, or pelvis-thigh IC during repetitive flexion-extension tasks in CLBP patients in comparison to healthy controls over 30 cycles of a repetitive trunk flexion/extension task. This finding contradicts the earlier findings by Silfies et al. [70] and Pranata et al. [66]. The non-significant findings may be explained by the differences in flexion/extension task between the studies. In the studies by Silfies et al. [70], and Pranata et al. [66], the load was held in the participant's hands, whereas in the latter study, a weight vest was worn which affixed the load closer to the participant's centre of mass [65]. When performing a repetitive lifting task with a load in the hands, there is a larger moment arm between the object that is being lifted and the participant's centre of mass, which would change the dynamics of the exercise. Therefore, direct comparison of findings between these studies is limited. During occupational tasks and daily living activities, objects are usually held in a person's hands, therefore the movement protocol used by Mokhtarinia et al. [65] limits generalisation of the findings to occupational settings and daily living activities.

Seay et al. [83] investigated the variability in IC during lifting between a group with a history of low back pain and a group with no history of CLBP. The participants lifted an 11 kg box from ankle height in front to a shelf at waist height for 10 minutes at 12 cycles per minute. They found no significant difference in trunk-pelvis IC between groups. However, the validity of these findings is limited as the group with a history of low back pain were pain free for > 6 months prior to data collection.

In summary, the literature in this review suggests that CLBP populations exhibit decreased IC during repetitive lifting tasks [66, 70]. Despite confounding variables affecting outcomes in the present literature, IC appears to be sensitive enough to differentiate between CLBP and healthy populations [66, 70]. These findings also suggest that the degree of CLBP-related disability, as measured by the Oswestry Disability Index (ODI), can affect IC of the trunk and lower limbs [66]. As such, when applying IC analysis to a CLBP population, the level of disability within the CLBP population may act as confounding variable and should be considered during study design.

3.3.6 Section Conclusion

This review aimed to provide an analysis of the current literature investigating IC in the context of the back and hip regions. The findings of this review have demonstrated that IC can detect subtle neuromuscular control impairments in CLBP patients. Additionally, this review has provided methodological considerations for future studies which apply IC analysis to a CLBP population including: the confounding effect of sex, load, fatigue, and disability. It is concluded that IC analysis is a valuable method of movement analysis during lifting in CLBP populations. Further, its utility could aid in the methods of clinical assessment and improve the prescription and assessment of rehabilitation programs for CLBP patients. However, the current literature

investigating IC in CLBP populations demonstrates poor methodological design and lacks external validity. Therefore, more research investigating the utility of IC analysis in CLBP is warranted.

Chapter 4: Methods

4.1 Aims

The primary aim of this study was to compare measures of LDS and IC during repeated lifting between people with and without CLBP. A secondary aim was to examine the effect the load lifted has on LDS and IC. It was hypothesized that CLBP patients would demonstrate less in-phase movement, greater IC variability, and greater LDS (indicated by lower maximum finite-time Lyapunov exponent values) during lifting compared to healthy people. It was also hypothesized that both CLBP patients and healthy people would demonstrate increased LDS, increased IC, and decreased coordination variability when lifting with a load.

4.2 Methods

4.2.1 Study design

This was an observational case-control study which included participants with and without CLBP. Participants with CLBP were matched for sex to a healthy participant without CLBP. Insufficient research examining differences in LDS and IC between people with CLBP and healthy controls meant that it was not possible to perform a priori statistical power analyses for this study [40, 61]. Therefore, a convenience sample was used. This sample size was determined by matching previous literature which has investigated differences in kinematics between people with CLBP and healthy controls during a repetitive lifting task [7-13].

4.2.2 Ethical considerations

Ethical approval was granted by the Western Sydney University Human Research Ethics Committee (HREC Number: H14357 [Appendix A]). All participants provided informed written consent prior to data collection (Appendix B) and were free to withdraw participation at any time without reason or consequence.

4.2.3 Participant recruitment

A variety of strategies were used to recruit potential participants. These included advertisement via social media (Facebook, Twitter, Instagram), word of mouth, university campus noticeboards and university course vUWS sites. A participant recruitment flyer was posted to social media sites, university noticeboards and course vUWS sites (Appendix C). The word-of-mouth approach was also used by the research team in a similar way by identifying potential participants via their extended network connections. Existing connections with potential participants were also individually contacted by the primary researcher via email or phone.

Once an individual had registered their interest in participating, the primary investigator provided the participant with an information sheet, which further explained the aims and methods of the study (Appendix D). Additionally, prospective participants were allocated with sufficient time (at least 24-48 hours) to consider the information before deciding to participate or not. Once an individual had registered their interest in participating, screening was performed prior to consent to ensure that participant met the eligibility criteria.

4.2.4 Inclusion and exclusion criteria

Twelve healthy adults (6 males and 6 females), and 12 adults with chronic low-back pain (6 males and 6 females) were recruited to participate in this study. Participants were matched

based on their sex. All included participants were between 18 and 55 years of age with the CLBP group reporting pain symptoms between T12 to the gluteal folds that was not from a specific origin (as confirmed from previous back surgical history, spondylolisthesis, spinal stenosis, persistent referred pain symptoms into the lower leg). Participants with a history of spinal surgery, spinal column abnormality, daily symptoms of pain that go down into their leg, a diagnosis of a mental health condition or any neuromuscular or metabolic disease were excluded.

4.2.5 Laboratory set up and instrumentation

Experimental testing was conducted in a Sport and Exercise Science Laboratory at the Campbelltown Campus of Western Sydney University. Each participant attended the laboratory on three separate occasions. This consisted of one familiarisation session of 30 minutes duration, and two testing sessions of approximately one hour duration. All sessions were separated by a minimum of 24 hours. The order in which the two testing sessions took place was pseudo-randomized based on each participants preference at session one. Of the 24 participants, 10 out of 12 from the CLBP group, and 9 out of 12 from the Control group opted to perform the un-loaded testing session first.

To measure spatiotemporal parameters, Ambulatory Parkinson's Disease Monitoring (APDM, Portland, OR, United States) Opal IMU (inertial measurement unit) sensors, consisting of tri-axial linear accelerometers, gyroscopic, and magnetometers were used. A Lumbar IMU was firmly attached using straps with a quick-release buckle clip to the participant on the superior aspect of the posterior sacral surface, positioned 1 cm below the L5 spinous process (Figure 1). An IMU was also placed on the sternum, centred over the manubrium (Figure 1). Additional IMUs were placed on the lateral side of the upper right thigh (centrally and halfway between

the greater trochanter and lateral epicondyle of the knee) using a Velcro strap, and the right lower leg (anterior to the medial surface of the tibia so that the Velcro strap wrapped around the widest part of the gastrocnemius) (Figure 1). IMU signal outputs were wirelessly transmitted to a Lenovo laptop (ThinkPad E14, Hong Kong) to be automatically processed and calculated via the corresponding Moveo Explorer software. Data were collected from the IMU sensors at 128 Hz.

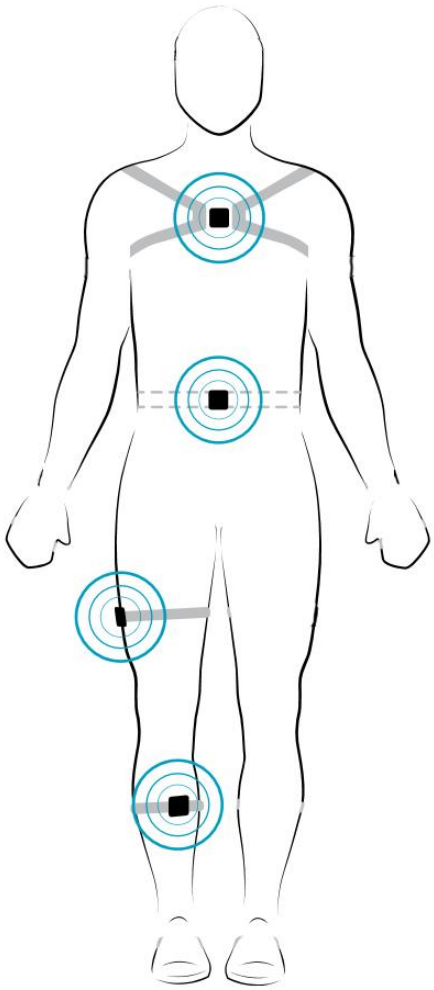

Figure 1. A visual representation of IMU sensor placement over the four segments including: Sternum, Lumbar, Right Thigh, and Right Shank.

4.2.6 Procedures

Prior to the commencement of data collection, all participants were familiarised with the experimental procedure. During the familiarisation session, participants completed the participant consent form and their age (years), sex, height (m), weight (kg), and duration of symptoms (years) were recorded. Participants also completed the Fear-Avoidance Beliefs Questionnaire (FABQ) (Appendix E), and the Oswestry Disability Index (ODI) (rated from 0 to 100% disability) during the familiarisation session (Appendix F). The ODI is a 10-item questionnaire examining how a patient's low back pain affects different activities during their life (0-100% scoring; 0% no disability). Participants were fitted with the IMU sensors to allow them to get accustomed to movement whilst the IMU sensors were attached. The repetitive lifting task was demonstrated to the participant, and they were given an opportunity to practice the repetitive lifting task whilst the IMU sensors were attached. Motion data were not recorded during the familiarisation session.

Following the completion of a familiarisation session, two testing sessions separated by a minimum of 24 hours were conducted by each participant. Upon arrival, participants were given the option to choose between performing the loaded or un-loaded condition. Experimental data collection commenced with participants being fitted with the IMU sensors. Once the wireless IMU sensors were detected by the computer, the participant was asked to stand still until baseline position in space data was established. During the loaded testing session, participants performed a repetitive lifting task (the deadlift) for 35 repetitions using a barbell that was weighed equivalent to 15% of the participant's bodyweight. The lifting task was performed at 20-25 repetitions/minute to match the speed at which work-related lifting tasks are performed [84, 85]. Lifting performance was monitored visually by the primary researcher with verbal feedback given to speed up or slow down if necessary. One complete

lift cycle was defined as movement from a standing position, in which the hips were fully extended, and into a flexed position until the barbell touched the participant's toes, and back to an extended position. The barbell/s that were used did not have weight plates on the sleeves, and therefore allowed the participant to touch their toes. Participants were required to adopt a natural rhythm during lifting, without any abrupt motions. If participants did not touch their toes during any of the repetitions in the trial, or the lifting trial was performed at a cadence faster than 20-25 repetitions/minute, verbal feedback was given. Following the verbal feedback, if there was no correction in cadence or if there were ≥ 5 repetitions in the trial in which the participant did not touch their toes, the lifting trial was to be repeated. To simulate the natural lifting movement used in some occupational tasks, a metronome was not used in this study to strictly confine cadence. This was done to increase the external validity of the study and allow for extrapolation of the findings to work-related lifting [84, 85]. Participants also rated their pain levels on a 10-cm VAS pain scale before and after the two lifting trials (Appendix G). During the un-loaded testing session, participants performed the same repetitive lifting task (the deadlift) for 35 repetitions under the same control conditions, except they used an un-weighted dowel rod.

4.2.7 Data analysis

Motion data from the IMU's were collected using Moveo Explorer (APDM Inc., Portland, OR, United States) from which angular acceleration, angular velocity and orientation quaternion data of each IMU sensor (sternum, lumbar, right upper leg, right lower leg) were computed. Joint angles of the back, hip, and knee were extracted automatically with APDM proprietary software. Segment angles of the sternum, lumbar, thigh, and shank were determined via a custom MATLAB algorithm (MATLAB 2022b, Mathworks, Massachusetts, United States) (Appendix H). The dependant variable of LDS was computed through a custom R algorithm

(RStudio, Posit, Massachusetts, United States) (Appendix I). Intersegment coordination was calculated via a custom MATLAB algorithm (Appendix H).

4.2.8 Local dynamic stability

Local dynamic stability was quantified by estimating the maximum finite-time Lyapunov exponent using Kantz's algorithm [89]. First, 3D Euler angles were exported from Moveo Explorer. The Moveo Explorer software output timestamped tuples of the 3D Euler angles of the back, hip, and knee, in comma separated values (CSV) format (Figure 2.A). Next, the Euclidian norm of the 3D Euler angles of the back, hip, and knee were calculated according to Equation 1 (Figure 2.B).

$$\text{Equation 1: Euclidean norm: } |x| = \sqrt{\theta_{saggital}^2 + \theta_{frontal}^2 + \theta_{transverse}^2}$$

To match previous LDS literature, kinematic data from the first five cycles of the lifting task were removed from the trial signal to ensure steady state motion [23, 24, 40, 61]. The remaining 30 cycles were analysed, which has been shown to be appropriate for data reliability and to obtain precise estimates of LDS [38]. To avoid sampling artifacts, other types of interference and accommodate different signal lengths, the remaining trial signal was time-normalized to 8000 samples (Figure 2.C) [23, 24, 40]. This was done to reduce the confounding effect of movement speed on LDS [37]. Using an algorithm developed in R, a 6D state space for the time series was reconstructed from the three-dimensional joint displacements using a time-delay of 2.6 samples (Appendix I). The exponential rate of divergence between nearest neighbour trajectories in the reconstructed state space was determined by estimating a line of

best-fit across the first 1.6 seconds of the average logarithmic divergence curve (Figure 2.D) [89]. The LDS measure was given by the slope of the line of best-fit.

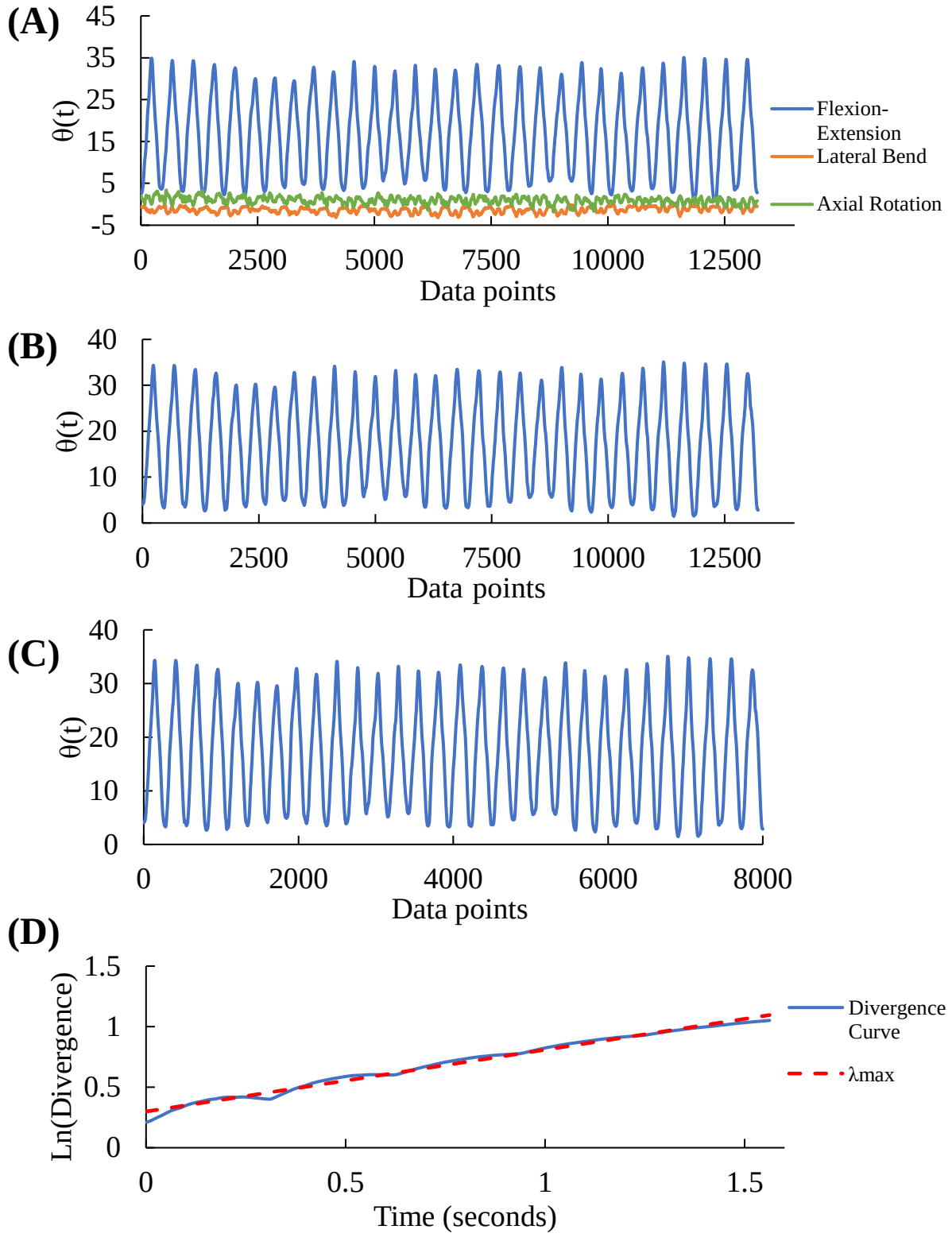

Figure 2. Steps of LDS calculation. Data represent a single participant trial. (A) Original 3-D Angular Time Series data. (B) The Euclidean norm of the three angles. (C) The Euclidian norm time-normalized to 8000 points. (D) Average logarithmic rate of divergence of all nearest neighbour pairs over 1.6 s. Where: Ln(Divergence) = Logarithmic Divergence; θ(t) = Angular rotation of joint; λmax = Maximum finite-time Lyapunov Exponent.

4.2.9 Intersegment coordination

Intersegment coordination was calculated using CRP curves on flexion-extension data. Raw IMU orientation quaternion data from the lifting trials were imported into MATLAB. A custom algorithm developed in MATLAB was used to convert the sternum, lumbar, thigh, and shank IMU quaternion data to Euler angles (Appendix H). Based on previous literature, Euler data were then filtered using a 6 Hz Butterworth filter [86-88]. The MATLAB program then resampled the Euler data into individual lift cycles (Appendix H).

Segment angular velocity was then calculated by taking the derivative of the IMU angular positions using the three-point central finite-differences method as expressed in Equation 2.

$$\text{Equation 2: } \omega_i = \frac{(\theta_{i+1})-(\theta_{i-1})}{2\Delta t}$$

$$\text{Where: } \theta = \text{segment angle; } i = \text{data point; } \Delta t = \text{difference in time between points } (\frac{1}{128})$$

All segment angles and velocities were then divided into individual flexion-extension cycles, as defined by successive maximum flexion angle, and interpolated to 101 data points corresponding to 0 - 100% of the flexion-extension cycle.

Using the following equations (Equations 3a and 3b), segment angular positions (Figure 3.A) and velocity (Figure 3.B) were then phase-normalized from -1 to 1 to minimize effects of signal amplitude and frequency on the calculation of the segment phase angle:

$$\text{Equation 3a: Horizontal axis (angle): } \theta_i = 2 \times \frac{[\theta_i - \min(\theta_i)]}{[\max(\theta_i) - \min(\theta_i)]} - 1$$

$$\text{Where: } \theta = \text{segment angle; } i = \text{data point}$$

Equation 3b: Vertical axis (angular velocity): $\omega_i = \frac{\omega_i}{\max(|\omega_i|)}$

Where: $\omega =$ segment angular velocity; $i =$ data point

Phase angles were then calculated at each time point of the flexion-extension cycle using a four-quadrant inverse tangent function (atan2) and defined as the angle from the right horizontal axis (Equation 4).

Equation 4: $\varphi = \tan^{-1}(\text{angular velocity}/\text{angular rotation})$

The CRP curve was obtained by subtracting the phase angle of the distal segment from the proximal segment for each data point through the entire cycle. The Sternum-Lumbar CRP curve was obtained by subtracting the sternum value from the corresponding lumbar value for each data point throughout the CRP curve (Equation 5a). The Lumbar-Thigh CRP curve was obtained by subtracting the thigh value from the corresponding lumbar value for each data point throughout the CRP curve (Equation 5b). The Thigh-Shank CRP curve was obtained by subtracting the shank value from the corresponding thigh value for each data point throughout the CRP curve (Equation 5c). To avoid discontinues, phase angles were unwrapped using a two-quadrant tangent inverse according to the following equation (Equations 5a, 5b and 5c) (Figure 3.C).

Equation 5a: $\text{Sternum} - \text{Lumbar Coupling: } \varphi lumbar - \varphi sternum$

Equation 5b: $\text{Lumbar} - \text{Thigh Coupling: } \varphi lumbar - \varphi thigh$

Equation 5c: $\text{Thigh} - \text{Shank Coupling: } \varphi thigh - \varphi shank$

Mean Absolute Relative Phase (MARP) and Deviation Phase (DP) were calculated over three time-bands: (1) 0 - 100% of the cycle, to evaluate IC throughout the entire lifting cycle (Equations 6a and 7a); (2) 0 - 50% of the cycle, to evaluate IC during flexion of the trunk and eccentric loading of the trunk extensors (Equations 6b and 7b); and (3) 50 - 100% of the cycle, to evaluate IC during extension of the trunk and concentric loading of the trunk extensors (Equations 6c and 7c).

Mean Absolute Relative Phase was calculated by taking the average of the CRP curve. Values closer to 0° indicate a more ‘‘in phase” coupling between the segments while values closer to 180° signify an ‘‘out of phase” coupling.

Equation 6a: MARP (0 - 100%) = $\sum_{i=1}^{101} |\varphi_{relative\ phase}|_i/101$

Equation 6b: Flexion MARP (0 - 50%) = $\sum_{i=1}^{50} |\varphi_{relative\ phase}|_i/50$

Equation 6c: Extension MARP (50 - 100%) = $\sum_{i=51}^{101} |\varphi_{relative\ phase}|_i/50$

Deviation Phase was calculated by taking the average of the mean standard deviation (SD) ensemble curve. DP values closer to 0° indicate less coordination variability or more coordination stability.

Equation 7a: DP (0-100%) = $\sum_{i=1}^{101} SD_i/101$

Equation 7b: Flexion DP (0-50%) = $\sum_{i=1}^{50} SD_i/50$

Equation 7c: Extension DP (50-100%) = $\sum_{i=51}^{101} SD_i/50$

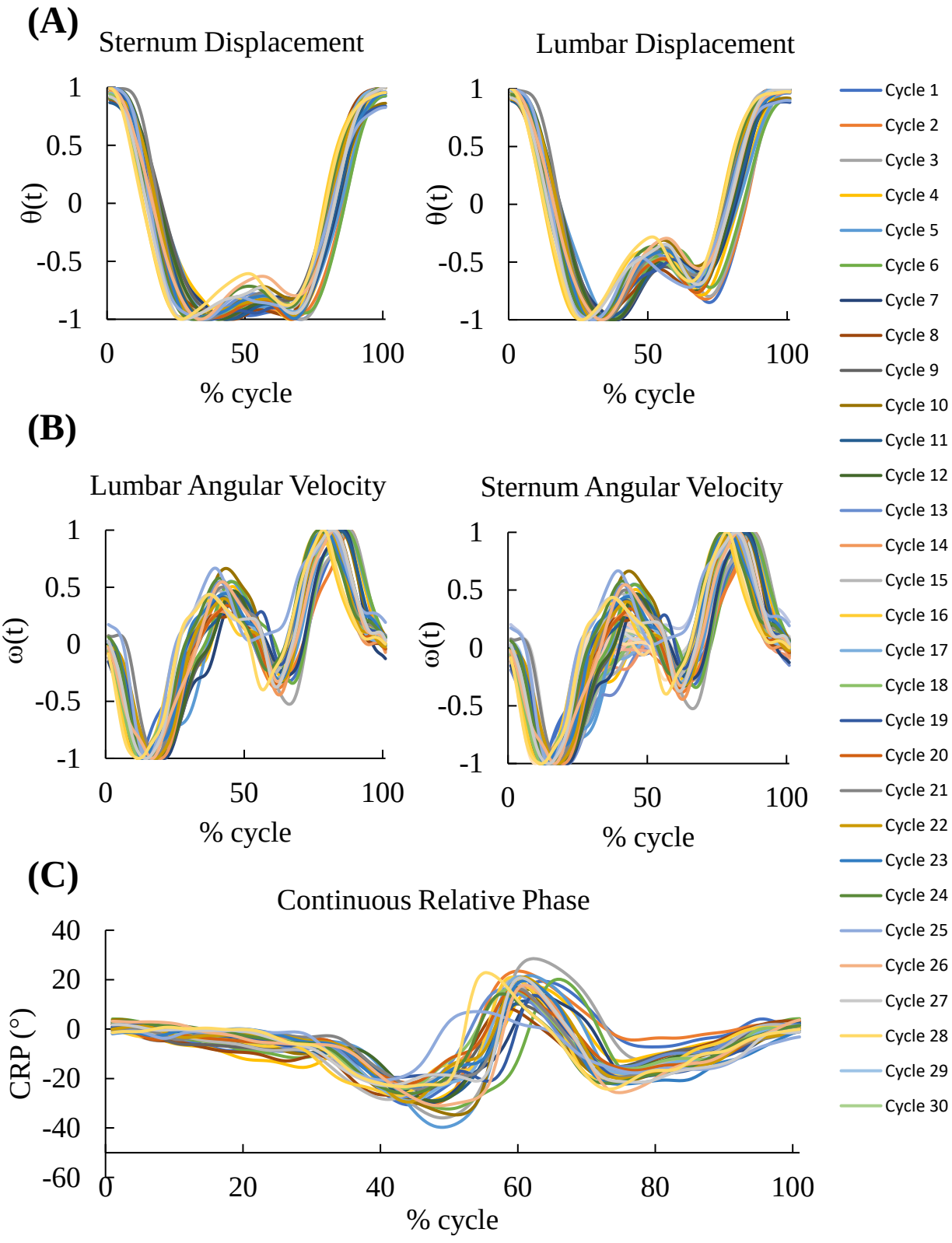

Figure 3. Calculation of CRP angle from a single participant trial. (A) Normalized sagittal plane Lumbar and Sternum IMU angular rotation. (B) Normalized sagittal plane Lumbar and Sternum angular velocity. (C) The CRP angle between the Sternum and Lumbar IMUs. Where: θ(t) = Angular rotation; ω(t) = Angular velocity.

4.2.10 Statistical analysis

All statistical analyses were performed using Statistical Package for Social Sciences (SPSS) (IBM SPSS Statistics 28, International Business Machines Corporation, NY, United States). Independent t-tests were performed to determine baseline differences in age (years), sex, height (cm), and weight (kg) between participants with CLBP and healthy participants. Statistical significance was accepted at $p < 0.05$.

A 2-factor repeated measures analysis of variance (ANOVA) with between (2 levels – CLBP and healthy controls) and within (2 levels – load and no load) group factors were used to examine differences in the dependant variables. Mean differences were expressed with 95% confidence intervals (CI). The significance level for this study was set at $p \leq 0.05$ and Bonferroni corrections were performed to reduce the risk of type 1 error. Data is presented as mean ± SD. The independent within subject variables for load were dowel and 15% bodyweight, and the between subject variables for condition were CLBP and Healthy. The dependent variables were LDS, MARP, and DP.

Separate one-way ANOVAs were performed on both the CLBP group and the Control group to examine for differences in pain before and after completing the lifting trial. The dependant variable was self-reported pain, as indicated by the VAS score, and the independent variable was time (pre and immediately post-trial).

Data were assessed for normality using the Shapiro-Wilk test, and Mauchly's test of sphericity was performed and if significant, a Greenhouse-Geisser correction procedure was used. If there was a statistically significant a main effect observed, post hoc independent t-tests were performed to determine the differences for each main effect. If statistically significant within

group differences were observed, post hoc 1-way ANOVA were performed for both the CLBP and healthy participant groups to determine at which loading condition within each group there were statistically significant differences for the dependant variables. Where significant F values were observed for group, post hoc tests with Bonferroni's correction were used to determine the differences.

Chapter 5: Results

5.1 Participant characteristics

Participant characteristics are outlined in Table 1. No significant differences were observed between people with and without CLBP for any variable.

Table 1. Descriptive data (mean ± SD) pertaining to participant characteristics of CLBP and control groups.

Variable (units)	**CLBP (n = 12)**	**Control (n = 12)**	***t* Value**	***p* Value**
Age (years)	21 ± 1.8	24 ± 6.7	1.9	0.146
Sex (female, %)	6 (50%)	6 (50%)	0	1.0
Height (cm)	173 ± 9.7	169.8 ± 9.3	-0.84	0.411
Mass (kg)	76.8 ± 12.4	70.4 ± 11	-1.34	0.195
BMI m/kg²	26 ± 5	24 ± 2	-0.93	0.361
CLBP duration (years)	3.08 ± 1.8	-	-	-
ODI (%)	15.9 ± 5.8	-	-	-
FABQ - Physical Activity	11.8 ± 7.4	-	-	-
FABQ - Work	5.8 ± 8.4	-	-	-

n = number of participants, BMI = Body Mass Index, ODI = Oswestry Disability Index, FABQ = Fear Avoidance Beliefs Questionnaire.

5.2 Effect of chronic low back pain on local dynamic stability

Lower maximum finite-time Lyapunov exponent values indicate greater LDS. The maximum finite-time Lyapunov exponent of the hip was significantly lower in our CLBP group in both the un-loaded (mean difference = - 0.41, standard error = 0.16, p = 0.018, CI - 0.74 to - 0.08), and loaded condition (mean difference = - 0.47, standard error = 0.2, p = 0.03, CI = - 0.88 to - 0.05). The maximum finite-time Lyapunov exponent of the knee was significantly lower in the CLBP group in both the un-loaded (mean difference = - 0.39, standard error = 0.16, p = 0.02, CI = 0.73 to 0.06), and loaded condition (mean difference = - 0.41, standard error = 0.18, p = 0.04, CI = - 0.79 to - 0.03) (Table 2). These findings demonstrate that the CLBP group had significantly greater LDS in the hip and knee under both load conditions. There were no significant between group differences in LDS of the back in both lifting conditions (Table 2). All computed maximum finite-time Lyapunov exponents were positive, suggesting exponential divergence for reconstructed trajectories.

Table 2. Independent post-hoc analyses of short-term maximum finite-time Lyapunov exponents (λmax).

Loading condition	**Joint**	**CLBP (Mean ± SD)**	**Control (Mean ± SD)**	**F ratio**	***p* Value**
No Load	Back	0.75 ± 0.32	0.81 ± 0.53	0.091	0.765
	Hip	1.73 ± 0.26	2.14 ± 0.47	6.551	0.018*
	Knee	1.08 ± 0.41	1.48 ± 0.46	4.938	0.037*
Load	Back	0.78 ± 0.37	0.77 ± 0.36	0.001	0.974
	Hip	1.69 ± 0.34	2.16 ± 0.58	5.400	0.03*
	Knee	1.03 ± 0.35	1.42 ± 0.41	6.103	0.022*

Units are in λmax. Values indicate mean ± standard deviation. * Indicates significant differences between CLBP and Control groups ($p < 0.05$). Note: Lower values indicate higher LDS.

5.3 Effect of chronic low back pain on intersegment coordination

There were no statistically significant between group differences in IC for MARP of the Sternum-Lumbar ($F_{(1,22)} = 0.05$, $p = 0.82$, $1-\beta = 0.06$), Lumbar-Thigh ($F_{(1,22)} = 1.26$, $p = 0.27$, $1-\beta = 0.19$), and Thigh-Shank ($F_{(1,22)} = 0.15$, $p = 0.7$, $1-\beta = 0.07$).

There were no statistically significant between group differences for MARP during flexion (50 - 100% of the lifting cycle) for the Sternum-Lumbar ($F_{(1,22)} = 0.08$, $p = 0.93$, $1-\beta = 0.05$), Lumbar-Thigh ($F_{(1,22)} = 1.82$, $p = 0.19$, $1-\beta = 0.25$), and Thigh-Shank ($F_{(1,22)} = 0.05$, $p = 0.82$, $1-\beta = 0.06$).

There were no statistically significant between group differences for MARP during extension (50 - 100% of the lifting cycle) for the Sternum-Lumbar ($F_{(1,22)} = 0.03$, $p = 0.87$, $1-\beta = 0.05$), Lumbar-Thigh ($F_{(1,22)} = 0.84$, $p = 0.37$, $1-\beta = 0.14$), and Thigh-Shank ($F_{(1,22)} = 0.54$, $p = 0.47$, $1-\beta = 0.11$).

There were no statistically significant between group differences in IC for DP of the Sternum-Lumbar ($F_{(1,22)} = 1.82$, $p = 0.19$, $1-\beta = 0.25$), Lumbar-Thigh ($F_{(1,22)} = 0.67$, $p = 0.42$, $1-\beta = 0.12$), and Thigh-Shank ($F_{(1,22)} = 0.41$, $p = 0.53$, $1-\beta = 0.09$).

There were no statistically significant between group differences for DP during flexion (0 - 50% of the lifting cycle) for the Sternum-Lumbar ($F_{(1,22)} = 1.79$, $p = 0.2$, $1-\beta = 0.25$), Lumbar-Thigh ($F_{(1,22)} = 0.15$, $p = 0.7$, $1-\beta = 0.07$), and Thigh-Shank ($F_{(1,22)} = 1.25$, $p = 0.28$, $1-\beta = 0.19$).

There were no statistically between group differences for DP during extension (50 - 100% of the lifting cycle) for the Sternum-Lumbar ($F_{(1,22)} = 1.88$, $p = 0.18$, $1-\beta = 0.19$), Lumbar-Thigh ($F_{(1,22)} = 1.25$, $p = 0.28$, $1-\beta = 0.14$), and Thigh-Shank ($F_{(1,22)} = 0.11$, $p = 0.75$, $1-\beta = 0.06$).

5.4 Effect of load on local dynamic stability

There were no statistically significant within-group differences for local dynamic stability of the back ($F_{(1,22)} = 0.16$, $p = 0.89$, $1-\beta = 0.05$), hip ($F_{(1,22)} = 0.01$, $p = 0.93$, $1-\beta = 0.05$), and, knee ($F_{(1,22)} = 0.77$, $p = 0.39$, $1-\beta = 0.14$) when lifting with a load compared to the no-load condition (Figure 4).

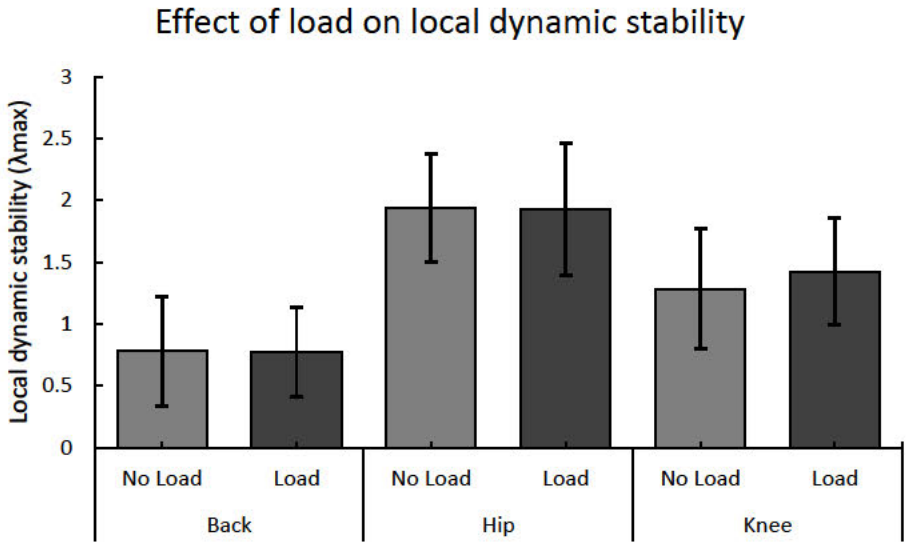

Figure 4. Effect of load on local dynamic stability. Error bars represent 95% confidence intervals.

There were no significant load by group interaction, with no difference in LDS of the back ($F_{(1,22)} = 0.349$, $p = 0.56$, $1-\beta = 0.09$), hip ($F_{(1,22)} = 0.05$, $p = 0.83$, $1-\beta = 0.06$), and knee ($F_{(1,22)} = 0.01$, $p = 0.93$, $1-\beta = 0.05$).

5.5 Effect of load on intersegment coordination

There were no statistically significant within-group differences for MARP of the Sternum-Lumbar ($F_{(1,22)} = 0.014$, $p = 0.91$, $1-\beta = 0.05$), Lumbar-Thigh ($F_{(1,22)} = 0.56$, $p = 0.46$, $1-\beta = 0.11$), and, Thigh-Shank ($F_{(1,22)} = 0.34$, $p = 0.86$, $1-\beta = 0.05$) for load.

There were no statistically significant within-group differences for the Sternum-Lumbar ($F_{(1,22)} = 3.96$, $p = 0.06$, $1-\beta = 0.48$), Lumbar-Thigh ($F_{(1,22)} = 0.43$, $p = 0.52$, $1-\beta = 0.97$), and Thigh-Shank ($F_{(1,22)} = 0.15$, $p = 0.7$, $1-\beta = 0.07$) for load during flexion of the trunk MARP (0 - 50% of the lifting cycle).

There were no statistically significant within-group differences for MARP of the Sternum-Lumbar ($F_{(1,22)} = 2.84$, $p = 0.1$, $1-\beta = 0.36$), Lumbar-Thigh ($F_{(1,22)} = 0.61$, $p = 0.44$, $1-\beta = 0.12$), and, Thigh-Shank ($F_{(1,22)} = 0.01$, $p = 0.98$, $1-\beta = 0.05$) for load during extension of the trunk (50 - 100% of the lifting cycle).

There were no significant load by group interaction, with no difference in MARP of the Sternum-Lumbar ($F_{(1,22)} = 0.63$, $p = 0.44$, $1-\beta = 0.12$), Lumbar-Thigh ($F_{(1,22)} = 0.19$, $p = 0.67$, $1-\beta = 0.07$), and Thigh-Shank ($F_{(1,22)} = 0.02$, $p = 0.91$, $1-\beta = 0.05$) when lifting with a load.

There were no statistically significant within-group differences for DP of the Sternum-Lumbar ($F_{(1,22)} = 0.57$, $p = 0.46$, $1-\beta = 0.11$), Lumbar-Thigh ($F_{(1,22)} = 0.05$, $p = 0.83$, $1-\beta = 0.05$), and Thigh-Shank ($F_{(1,22)} = 2.38$, $p = 0.14$, $1-\beta = 0.14$) when lifting with a load.

There were no statistically significant within-group differences for DP of the Sternum-Lumbar ($F_{(1,22)} = 0.31$, $p = 0.58$, $1-\beta = 0.08$), Lumbar-Thigh ($F_{(1,22)} = 2.24$, $p = 0.15$, $1-\beta = 0.3$), and Thigh-Shank ($F_{(1,22)} = 2.45$, 0.13, $1-\beta = 0.06$) for load during flexion of the trunk.

There were no statistically significant within-group differences for DP of the Sternum-Lumbar ($F_{(1,22)} = 2.24$, $p = 0.15$, $1-\beta = 0.29$), Lumbar-Thigh ($F_{(1,22)} = 0.02$, $p = 0.88$, $1-\beta = 0.05$), and Thigh-Shank ($F_{(1,22)} = 1.97$, $p = 0.174$, $1-\beta = 0.26$) for load during extension of the trunk.

There were no significant load by group interaction, with no difference in DP of the Sternum-Lumbar ($F_{(1,22)} = 0.71$, $p = 0.1$, $1-\beta = 0.13$), Lumbar-Thigh ($F_{(1,22)} = 2.02$, $p = 0.17$, $1-\beta = 0.28$), and Thigh-Shank ($F_{(1,22)} = 0.21$, $p = 0.21$, $1-\beta = 0.23$) when lifting with a load.

5.6 Pain provocation from performing the repetitive lifting task

Pain, as measured by VAS significantly increased after performing the lifting trial in the CLBP group in both the un-loaded ($F_{(1,22)} = 4.62$, $p = 0.04$) and load condition ($F_{(1,22)} = 7.62$, $p = 0.01$). There were no significant differences in VAS scores in the Control group in both the un-loaded ($F_{(1,22)} = 2.01$, $p = 0.171$) and load condition ($F_{(1,22)} = 1.18$, $p = 0.29$).

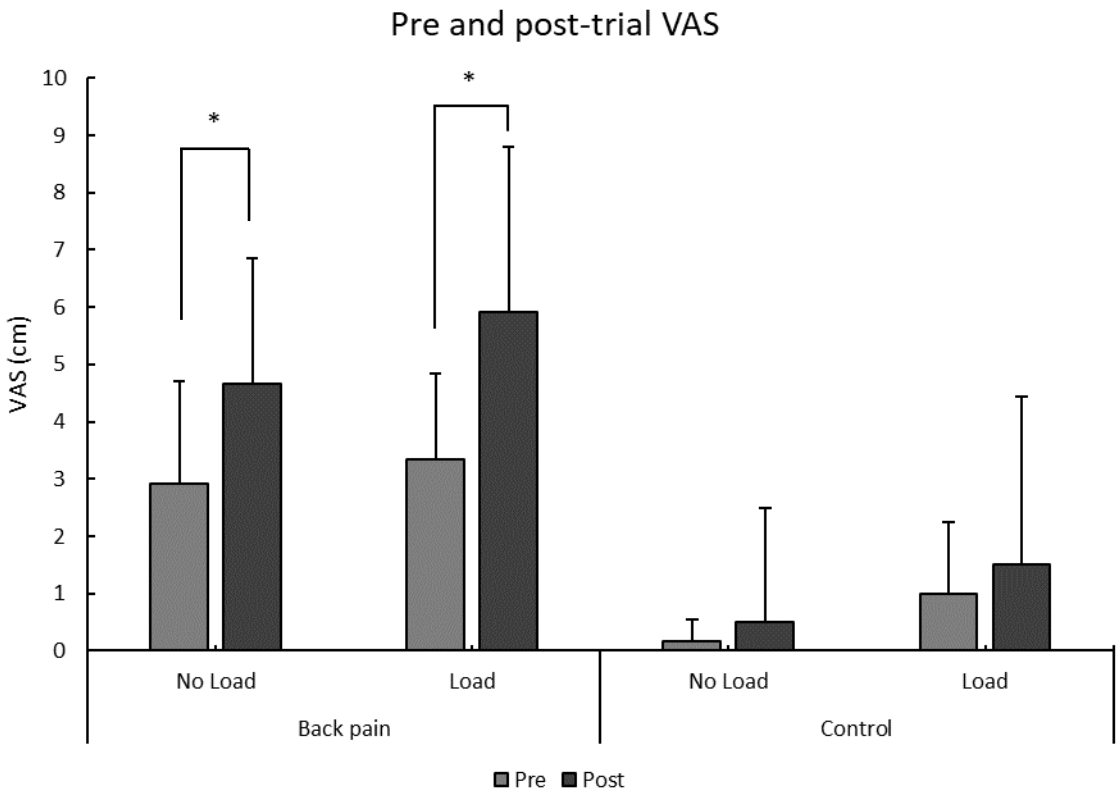

Figure 5. VAS pain rating scores before, and immediately after completing the lifting trials. Error bars represent 95% confidence intervals (* indicates $p < 0.05$). Data are Mean + SD.

Chapter 6: Discussion

The current study aimed to compare the measures of LDS and IC during repeated lifting between people with and without CLBP. A secondary aim was to examine the effect the load lifted has on LDS and IC. We hypothesized that the CLBP group would demonstrate higher LDS (indicated by a lower λmax), less in-phase movement (indicated by a higher MARP), and greater IC variability (indicated by a higher DP) than the healthy group. We also hypothesized that both CLBP patients and healthy people would demonstrate increased LDS, increased IC, and decreased coordination variability when lifting with a load.

Local dynamic stability was significantly higher in the CLBP group at the hip and knee in both the loaded and un-loaded conditions. Contrary to our hypothesis, no difference was observed in IC variability and IC between the CLBP and healthy groups. There were also no differences in IC or LDS for both the CLBP group and healthy group when lifting with a dowel in comparison to lifting with a barbell. Only the CLBP group reported a significant increase in pain immediately after completing each lifting trial.

6.1 Effect of chronic low back pain on local dynamic stability

Our study showed that only hip and knee LDS were affected by CLBP. This may be the result of the neuromuscular system of the CLBP group imposing greater spatiotemporal constraints as increased spasticity of the hamstrings and hip flexors have been reported as symptoms of CLBP [90]. These muscles play a major role in controlling trunk movement at the hip during repetitive lifting tasks. Additionally, previous literature has shown increased trunk antagonist

coactivation in people with CLBP when lifting loads [11-13, 91]. Because individuals with CLBP often have tight hamstrings and hip flexors [92], and have increased trunk antagonist coactivation during lifting, this may create an increase in muscle stiffness that could directly translate into greater trunk rotational stiffness [25]. Therefore, because people with CLBP may be in a more mechanically stable state at the hip and knee during repetitive lifting tasks, there may be a decreased need for feedback-induced muscular contraction following a perturbation.

Previous literature suggests the kinematic response of the trunk to a perturbation is partially determined through the mechanical stability of the trunk prior to lifting [25]. Therefore, because CLBP exhibit tighter muscles than healthy individuals [92], the increased mechanical stability prior to a perturbation could decrease the divergence of trajectories immediately after that perturbation, resulting in higher LDS [24]. The increased activation of these muscles might be due to a deficiency of the neuromuscular system in stabilising hip movement; subsequently restraining movement of the trunk in people with CLBP.

During repetitive lifting, the trunk will follow an intended trajectory that represents the desired kinematic path of the vertebrae [61]. A stable attractor would be characterised by a motor control system that is better able to maintain movement within its intended trajectory [17]. The motor control system attenuates internal perturbations by manipulating actuators and effectors (tendons and muscles) to allow movement to return to their intended trajectories [17]. The greater LDS of the hip and knee exhibited by our CLBP group suggests that the neuromuscular system manipulates muscle activation in the CLBP group to direct their joint movement to a more stable attractor [17]. The increased stability of the attractor may be achieved by an adapted motor plan of co-contraction that increases activation of the muscles surrounding the hip and knee. This notion is supported by previous literature which has shown LDS of the

lumbar spine to be modulated by the rotational stiffness of the muscles surrounding it [24, 25]. Our findings, alongside previously reported findings of reduced strength, endurance, and cross-sectional area of the trunk extensors in people with CLBP, demonstrates the importance of the trunk extensors in CLBP treatment strategies [94-96].

6.2 Effect of chronic low back pain on intersegment coordination

In this study IC was used to complement LDS assessment and help explain any observed differences. Contrary to our hypothesis, we observed no difference in any IC measures (MARP or DP) between both groups, and in both lifting trials. Our results may be explained by several reasons. First, during a repetitive lifting trial with load, the physical demand of completing the lifting task increases with time [40]. These changes are mediated by increases in fatigue or pain over the duration of performing a repetitive lifting trial [40]. Decreased IC over time would suggest that people with CLBP have a less predictable neuromuscular system (i.e., more flexible motor behaviour) that is less adaptable to increasing task demands and load [66]. Whereas a greater IC over time would indicate that people with CLBP are executing the task with a more rigid lifting strategy and more predictable neuromuscular system [66]. Previous authors have proposed that due to musculoskeletal dysfunction and poor proprioception in people with CLBP, they would have a reduction in their ability to monitor and incorporate timely sensory feedback into their movement adjustments during increasing task demands [70, 97, 98]. It was also suggested that these musculoskeletal changes may cause an adaptation of the neuromuscular system into altered intersegmental dynamics, in an attempt to optimize LDS [70]. Contrary to this, our CLBP group were able to produce adequate IC to perform the lifting task, despite significantly greater LDS of the hip and knee. Because we observed no difference in IC between groups, it is possible that the significantly greater LDS of the hip and knee that

we observed in the CLBP group was an adaptive mechanism of the neuromuscular system to maintain sufficient coordination patterns to execute the lifting task. It is also possible that deterministic properties of the motor control system may provide the neuromuscular and musculoskeletal system with the necessary mechanisms to adapt to these changing task conditions (i.e., increasing task demands) [99-103]. These deterministic behavioural properties may allow for the generation of effective movement patterns that reduce variability in their coordination patterns [99-103]. Therefore, our findings may show that during repetitive lifting, as the task conditions change, neuromuscular adaptation may allow for coordination and movement patterns to be maintained.

An alternative explanation for our non-significant findings may be due to the varying motor control strategies demonstrated by people with CLBP in previous literature. The effect of CLBP on IC and inter-joint coordination has been conflicting to-date. Several studies have shown that people with CLBP demonstrate decreased inter-joint and IC during repetitive lifting, while other studies have shown an increase [65, 66, 70, 83]. Assuming our control group demonstrated the ideal and average IC during lifting, our lack of significant findings may be explained by our CLBP group exhibiting a range of either increased or decreased IC. This is supported by previous literature that has shown people with CLBP demonstrate significantly increased IC of the trunk-hip, mediated by a pain avoidance lifting strategy during a repetitive lifting task [65, 83]. In contrast, other studies have shown that people with CLBP demonstrate significantly decreased IC of the trunk-hip during repetitive lifting, mediated by muscle dysfunction in people with CLBP [66, 70]. In our study, it is also possible that elements of the lifting task such as load, speed, number of trials, fatigue were not sufficient to produce a difference in IC between the groups. That is, the lifting task did not challenge the CLBP group enough to elicit alterations in IC.

6.3 Effect of loading condition on local dynamic stability and intersegment coordination

In contrast to our hypothesis and a majority of the previous literature, repetitive lifting in our study with a load compared to no load had no effect on LDS [23, 24]. In many previous studies, the load lifted was based on 10% of the participant's maximum strength [23, 24]. In the present study, participants lifted a weight equivalent to 15% of their bodyweight. This normalised load was chosen to ensure completion of the lifting trial by the CLBP group. However, as this approach did not control for each participant's individual strength levels, it is likely that the prescribed load task demand was not high enough to sufficiently challenge participants and cause altered LDS and IC.

Considering increased co-contraction levels during repetitive flexion-extension movements [24, 59, 104-106], we hypothesized that there exists a more in-phase and less variable coordination pattern during our loaded trials. The confounding effect of participant strength variability may also explain of our lack of significant findings for IC measures. Previous research has found that when lifting with a load, the coordination variability, indicated by DP, would be significantly lower during repetitive lifting tasks [65]. This is reportedly due to the increased demands of the task when lifting with a load, which would require participants to execute the task with a more rigid lifting strategy [65]. However, our findings suggest that both the coordination pattern, and the stability of the coordination pattern remained unaffected by load. This may indicate that the load increase between lifting trials was not sufficient for participants to use a rigid lifting strategy.

Another possible reason for the difference in findings between our study and previous literature is the difference in lifting tasks. In previous investigations, loads were lifted in boxes from knee-height to shoulder-height [23, 24]. These lifts likely forced a more "stoop-like" lifting movement, whereby the load is held anterior to the body. Supporting the load anterior to the body generates a moment arm between the individuals centre of gravity and the load, thus resulting in greater dynamic spine movement throughout the trial. In contrast, participants in our study performed full range deadlifts. During this movement, the load can be maintained closer to the centre of gravity, thus allowing participants to lift with greater lower limb contribution and reduced spine flexion. The notable differences in joint and segment excursions between the lifting techniques limits the ability to directly compare findings between this and previous studies.

6.4 Effect of pain provocation on motor control in people with CLBP

Previous literature suggests there is a causal relationship between CLBP and altered motor control [107]. In response to pain, people with CLBP may alter motor control strategies as a guarding mechanism whereby they modify their joint control strategies to protect themselves from a biomechanical perturbation that can intensify pain [108]. Additionally, to avoid or minimise pain and fatigue, increased motor variability (cycle-to-cycle variation) due to changes in muscle activation have been observed people with CLBP [109]. For example, an observed increase in LDS of the hip has been attributed in people with CLBP as a possible a pain-avoidance function [40]. The CLBP group in the present study reported a significant increase in pain immediately after both lifting trials. Therefore, it is possible that pain avoidance contributed to the significantly greater LDS of the hip and knee in the CLBP group.

The significant increase in pain immediately after both lifting trials in the CLBP group may be explained by their lower level of disability (15.9% ± 5.8% ODI) relative to the broader CLBP population. It is plausible that in people with CLBP who have low-moderate disability, the neuromuscular system will increase LDS during a repetitive flexion-extension task to maintain a consistent coordination pattern, rather than to avoid pain. Conversely, we could expect in CLBP with higher levels of disability to exhibit reductions in LDS during repetitive lifting as the neuromuscular system prioritises a pain-protective lifting strategy. Consequently, the use of a pain-protective lifting strategy will likely result in increases in coordination variability as the neuromuscular system has reduced the importance of maintaining a consistent coordination pattern. This hypothesis suggests individuals with CLBP may allow for an exacerbation of pain when performing functional tasks until a critical pain threshold is reached and may explain why we observed no difference in IC measures between groups. This theory is supported by previous literature whereby high-disability CLBP patients demonstrated significantly decreased lumbar-hip inter-joint coordination compared to CLBP patients with lower disability when performing a repetitive flexion/extension task [66]. Similar to our study, Pranata et al. [66] showed no difference in lumbar-hip inter-joint coordination between healthy controls and the low-disability CLBP group. However, further studies are warranted to investigate and confirm this hypothesis.

Additionally, pain avoidance may only in part explain the difference in hip and knee LDS between the CLBP group and healthy group. Our CLBP group had low levels of pain prior to completing the trials (2.92 ± 1.78 un-loaded; 3.33 ± 1.5 loaded). Hence, because our CLBP group had a lower level of disability and had relatively low levels of pain prior to completing the trial, it is possible that their pain had limited effect on LDS and IC.

6.5 Interpretation of findings

The interaction between the neuromuscular and musculoskeletal systems may influence the stability and variability of human movement in the presence of different types of perturbations [99, 100]. In repetitive lifting tasks, perturbations can be external (i.e., happening in the environment) or internal (i.e., happening inside the body). A healthy neuromuscular system is characterized by the ability to adapt to internal and external perturbations [28]. In the human motor control system, constraints placed on the system would decrease variability by either reducing the degrees of freedom or by limiting the potential combination of movements that are needed to execute the lifting task [14]. During repetitive lifting, the trunk will follow an intended trajectory that represents the desired kinematic path of the vertebrae [61]. Internal perturbations, such as neuromuscular control errors, destabilize the human motor control system. These disturbances cause the attractor to deviate from the optimal kinematic path of the vertebrae and require a correction from the human motor control system to correct the movement back to the desired trajectory [61]. Local dynamic stability quantifies the ability of the motor control system to attenuate these perturbations. Therefore, our findings show that the CLBP group were better able to resist internal perturbations and disturbances to their motor control than the healthy group in their hip and knee.

Interpretation of our findings is challenging because it is unknown whether the observed differences between groups reflect the adaptability of the neuromuscular system to enable task completion with minimal pain and should therefore be considered beneficial [110]. For example, the observed differences between groups could reflect the ability of the neuromuscular system to stabilize the trunk in a pain-protective manner [111]. Alternatively, the differences we observed may represent a motor control system that is overly rigid and is

associated with abnormal motor development [95]. For example, the greater LDS of the hip and knee exhibited in CLBP may be associated with increased trunk stiffness, which therefore could lead to increased spinal loading while lifting and consequently, an increased risk of injury.

It has been previously suggested that variability and stability observe different aspects of motor control [66]. This is supported by our findings which showed no difference in IC between the CLBP group and healthy groups, despite significantly greater LDS of the hip and knee relative to the healthy group. While it can be expected that these variability indexes would not react in the same way under various conditions, utilizing IC to evaluate the variability of kinematic time-series data ignores the time-dependent attributes of variability by assuming that each repetition is independent and unrelated to previous repetitions and that repetition-to-repetition variations are random [99, 101-103]. Additionally, IC does not account for the presence of feedback loops in the motor control of repetitive lifting. During repetitive lifting, each repetition is connected, where each consecutive repetition in the lifting cycle is influenced by the previous repetition/s. While IC gives accurate measures of motor variability within the system, it may not be as useful for showing the underlying neural processes of human movement exhibited by people with CLBP. While IC analysis provides a general picture of the level of the motor variability, no information is apparent about how the variations evolve with time [99-103]. Thus, utilising IC can mask the true structure of motor variability within the neuromuscular system, since all repetitions are averaged to generate a mean of the participants lifting cycle [99-103]. In this averaging procedure, the temporal variations of neuromuscular control of stability are lost. In contrast, LDS illustrates both spatial and temporal aspects of the data [55, 99-103]. Hence, LDS overcomes the limitations of IC by considering the interdependence of repetitions and changes in kinematics over time, thus providing greater

insight into motor control. While IC was used to supplement LDS analysis in the present study, our findings suggest that analysis of both the spatial and temporal aspects of the kinematic time-series is important for revealing motor control alterations in people with CLBP.

6.6 Limitations

Because our CLBP group was controlled and matched on sex with strict inclusion criteria, we were limited in the number of available participants. Additionally, COVID-19 lockdown closures affected our laboratory testing opportunities and our ability to recruit participants Once capacity to test re-opened, participants were reluctant to volunteer due to ongoing concern about COVID-19 exposure and risk, thus limiting the recruitment of participants within the permitted period of candidature for this degree. Therefore, our sample population was not representative of the broader CLBP population and limits the ability to generalise the findings.

Our CLBP participants had lower levels of pain and disability (15.9% ± 5.8% ODI) compared to patients found in clinics or workplaces [112]. Thus, our CLBP group were highly functioning, and therefore may not represent a majority of the broader CLBP population. Therefore, the findings of the present study cannot be generalised to all people with the pathology.

This study was delimited to include both males and female participants. Previous literature has shown that females use different coordination strategies as opposed to males when performing repetitive lifting tasks [75, 82, 86]. However, collecting data from equal numbers of both males

and females was considered important to reduce an un-balanced sex effect on the measurement of IC.

Another limitation to the current study was the possibility of a change in lifting technique. Previous literature has demonstrated that muscle fatigue results in alterations in trunk kinematics, and therefore alterations in lifting technique [113, 114]. To increase external validity, and to allow for extrapolation of our findings to work-related lifting, our experimental protocol allowed participants to use desired movement patterns. People with CLBP have increased trunk extensor fatigability in comparison to healthy people [96]. Because we did not measure the force producing capabilities of the trunk extensors before and after the lifting trial, we cannot rule out fatigue of the trunk muscles as a source of the differences in LDS between groups. Moreover, it is also possible that the premature trunk extensor fatigability exhibited by people with CLBP is reflected by alterations in motor control and detected by the changes in hip and knee LDS that we observed.

Additionally, the order in which the participants performed the un-loaded and loaded lifting trials was pseudo-randomized. A majority of participants opted to perform the un-loaded condition first (10 in the CLBP group; 9 in the Control group). Therefore, a learning effect may have occurred after performing the un-loaded lifting trial which may have resulted in the loaded lifting trial being confounded by practice. This practice could have led to increased stability and coordination during the loaded lifting trial.

Finally, we did not test individual participant functional lifting capacity relative to the deadlift. Thus, we were unable to individually prescribe loads commensurate with ability that were

challenging enough to elicit change to the motor control system and effect neuromuscular output adaptation. Future studies need to assess and categorise lifting performance capacity and set challenging loads prior to IC and LDS assessment in a CLBP population, because changes in IC are more pronounced when lifting loads relative to a participant's maximum voluntary strength [78]. It is possible that the lack of significant findings is due to the varying individual difficulty of the lifting task between both groups.

This study showed that compared to a healthy group, the CLBP group demonstrated significantly greater LDS of the hip and knee during lifting in both the loaded and un-loaded conditions. Differences in movement between groups were identifiable when using LDS assessment, but not IC. Our results suggest that analyses which consider both spatial and temporal aspects of the data, such as LDS, may be needed to reveal differences is neuromuscular control between CLBP and healthy groups. Future research that addresses these limitations is needed to further observe the effect of CLBP on LDS and IC.

Chapter 7: Conclusion

7.1 Summary

This study compared the measures of LDS and IC of the lower extremities and trunk during a repetitive lifting task between people with and without CLBP. Results demonstrated that the CLBP group exhibited significantly greater LDS than the healthy group, which is reflective of alterations of lower limb motor control. We observed greater LDS in the hip and knee and a significant increase in pain provocation in the CLBP group following completion of the lifting trials. There were no differences in IC or coordination variability between groups. Lifting with a load versus no load had no effect on LDS and IC in both groups.

7.2 Originality of Research

This thesis work builds upon existing literature by examining the effect of CLBP on LDS and IC and provides important preliminary information to facilitate future research on the motor control impairments of people with CLBP. Previous studies have showed that LDS and IC can differentiate between biomechanically different lifting techniques, movement paces and directions, fatigued versus non-fatigued conditions, lifting heavy versus light loads, and experimentally induced-low back pain [22-24]. To our knowledge, this is the first study to show the measurement of LDS can differentiate between healthy and CLBP people during repetitive deadlifts with a load. Additionally, this is the first study to suggest that LDS may be more successful in differentiating the control stability strategies demonstrated by people with CLBP, compared to IC during repetitive lifting. The results of this thesis emphasize an

increased importance of examination of lower limb movements in people with CLBP for clinical assessment. Hence, the shift in knowledge from this thesis furthers our understanding of motor control and moves us closer to improved clinical assessment of the motor control in people with CLBP.

7.3 Practical Implications/Recommendations

Chronic low back pain is considered a complex condition with biological and psychosocial factors. While the findings of this dissertation support previous literature by showing altered motor control in people with CLBP, the mechanisms driving altered motor control remain to be elucidated. Our findings could be the direct effect of CLBP or be mediated by fear of movement, disabilities, or other potential consequences of CLBP. Future research should continue to explore the underlying mechanism of CLBP and their associated motor impairment to aid development of more effective rehabilitation programs and treatment methods.

Our findings support the use of LDS assessment to identify motor control anomalies in CLBP patients. Given that our results showed lower-limb joint control strategies are altered in people with CLBP, we recommend that clinicians assess trunk extensor muscle function with tests that challenge inter-joint movement control and include dynamic movement of the lower limbs. Exercise interventions may be improved by incorporating movements that involve a diverse range of trunk movement strategies which prioritise increasing flexibility of the lower limb muscles. Increasing lower limb muscle flexibility will allow the neuromuscular system of people with CLBP to regain greater flexibility in its response to perturbations within the system.

7.4 Concluding Statement

A time and cost-effective screening method for directly assessing the neuromuscular system response to perturbations and control errors that occur naturally during movements would be beneficial for practitioners and allow better monitoring of improvements in movement over time. This thesis has shown quantitative measurement of LDS during repetitive lifting tasks is to be a useful method of assessing motor control alterations in people with CLBP. Thus, this study's findings contribute to the goal of developing effective quantitative tools for motor control evaluation in people with CLBP that when applied, could enhance the clinical practice of assessing and treating CLBP.

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Appendices

Appendix A: Letter of Ethical Approval from WSU Ethics Committee

WESTERN SYDNEY
UNIVERSITY

HUMAN RESEARCH ETHICS COMMITTEE

13 September 2021
Doctor Peter Clothier
School of Health Sciences

Dear Peter,

Project Title: "Effects of chronic low back pain on local dynamic stability and intersegment coordination during repetitive lifting"

HREC Approval Number: H14357
Risk Rating: High

I am pleased to advise the above research project meets the requirements of the National Statement on Ethical Conduct in Human Research 2007 (Updated 2018).

Ethical approval for this project has been granted by the Western Sydney University Human Research Ethics Committee. This HREC is constituted and operates in accordance with the National Statement on Ethical Conduct in Human Research 2007 (Updated 2018).

Approval of this project is valid from 13 September 2021 until 13 July 2022.

This protocol covers the following researchers:

Peter Clothier, John Marquez, Paul Marshall, Michael Knox

Summary of Conditions of Approval

1. A progress report will be due annually on the anniversary of the approval date.

2. A final report will be due at the expiration of the approval period.

3. Any amendments to the project must be approved by the Human Research Ethics Committee prior to being implemented. Amendments must be requested using the HREC Amendment Request Form.

4. Any serious or unexpected adverse events on participants must be reported to the Human Research Ethics Committee via the Human Ethics Officer as a matter of priority.

5. Any unforeseen events that might affect continued ethical acceptability of the project should also be reported to the Committee as a matter of priority.

6. Consent forms are to be retained within the archives of the School or Research Institute and made available to the Committee upon request.

7. Approval is only valid while you hold a position or are enrolled at Western Sydney University. You will need to transfer your project or seek fresh ethics approval from your new institution if you leave Western Sydney University.

8. Project specific conditions:

There are no specific conditions applicable.

Please quote the registration number and title as indicated above in the subject line on all future correspondence related to this project. All correspondence should be sent to humanethics@westernsydney.edu.au as this email address is closely monitored.

Yours sincerely

[redacted]

Associate Professor Gabrielle Weidemann
Presiding Member,
Western Sydney University Human Research Ethics Committee

Western Sydney University
ABN 53 014 069 881 CRICOS Provider No. 00917K
Locked Bag 1797 Penrith NSW 2751 Australia
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Appendix B: Participant Consent Form

WESTERN SYDNEY
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Consent Form – General (Extended)

Project Title: Effects of chronic low back pain on local dynamic stability and intersegment coordination during a light load repetitive lifting task.

This study has been approved by the Human Research Ethics Committee at Western Sydney University. The ethics reference number is: H*14357*

I hereby consent to participate in the above named research project.

I acknowledge that:

- I have read the participant information sheet (or where appropriate, have had it read to me) and have been given the opportunity to discuss the information and my involvement in the project with the researcher/s
- The procedures required for the project and the time involved have been explained to me, and any questions I have about the project have been answered to my satisfaction.

I consent to:

☐ *Having my age, height and weight recorded.*

☐ *Providing information regarding my health and injury history.*

☐ *Having IMU sensors placed on my thorax, pelvis, arms, and legs.*

☐ *Having my joint and segment movements recorded.*

☐ *Performing 2x bouts of resistance exercise consisting of 35 lifting repetitions.*

I consent for my data and information provided to be used in this project and other related projects for an extended period of time.

I understand that my involvement is confidential and that the information gained during the study may be published and stored for other research use but no information about me will be used in any way that reveals my identity.

I understand that I can withdraw from the study at any time without affecting my relationship with the researcher/s, and any organisations involved, now or in the future.

Name:

Signed: **Date:**

What if I have a complaint?

If you have any complaints or reservations about the ethical conduct of this research, you may contact the Ethics Committee through Research Engagement, Development and Innovation (REDI) on Tel +61 2 4736 0229 or email humanethics@westernsydney.edu.au.

Any issues you raise will be treated in confidence and investigated fully, and you will be informed of the outcome.

Western Sydney University
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Appendix C: Participant Flyer

RESEARCH PARTICIPANTS NEEDED

Our team is investigating the differences in movement during repetitive lifting between chronic low back pain populations and people without low back pain.

We are looking for both males and females with and without chronic low back pain who:

- Do not have a history of spinal surgery, spinal column abnormality, daily symptoms of pain that go down into their leg, a diagnosis of a mental health condition or any neuromuscular or metabolic disease.
- Are aged between 18 and 55 years

You will be required to attend our laboratory for up to 60 minutes on three separate occasions. On the first occasion, we will discuss participation and collect informed consent if you agree to participate. During this session, we will familiarize you with the procedures that will take place during the study. For the experimental trials we will need to place small devices for motion capture of the trunk and pelvis. On another occasion, you will be asked to perform 35 repetitions of a lifting task with a light load. On the other occasion, you will be asked to perform 35 repetitions of a lifting task floor without any load. During these trials, we will record your joint movements. The order in which the loaded and unloaded trials take place will be randomized.

If you are interested in participating, ot to find out more about the study, please

e-mail John at: 18749449@student.westernsydney.edu.au

If you know someone else with low back pain, please forward this information on to them for their consideration.

WESTERN SYDNEY
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This project has been approved by the Western Sydney University Human Research Ethics Committee: H14357

Appendix D: Participant Information Sheet

WESTERN SYDNEY
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Research study: Effects of chronic LBP on local dynamic stability and intersegment coordination during a light load repetitive lifting task.

Participant Information Sheet – General (Extended)

Project Title: Effects of chronic low back pain on local dynamic stability and intersegment coordination during a light load repetitive lifting task.

Project Summary: Low back pain is the leading cause of disability worldwide, causing substantial financial burden. Considering the socioeconomic burden of low back pain, there is a need to improve clinical assessments to facilitate better treatment decisions. While previous research has found differences in the way people move with low back pain, the reasons for these differences in movement are not well understood. According to the Australian Physiotherapy Association (APA), the efficacy of physiotherapy intervention programs for low back pain is currently evidenced by qualitative measures such as reduced disability, reduced absenteeism, and faster return to work. This project will investigate potential measures that could be used by clinicians to help inform their exercise prescription and measure improvements in low back pain over time.

You are invited to participate in a research study being conducted by John Marquez, School of Health Sciences – under the Supervision of Dr Peter Clothier, Lecturer, School of Health Sciences, and Michael Knox, Tutor, School of Health Sciences, and Dr Paul Marshall, Research Fellow, Department of Exercise Sciences, University of Auckland. The research will be investigating the differences in trunk motor control during repetitive lifting between chronic low back pain populations and a healthy population without low back pain.

How is the study being paid for? This research is supported by the Australian Government under the Research Training Program.

What will I be asked to do? First, upon registering your initial interest, we will screen you for eligibility to participate. This will involve answering a series of questions about your health and injury history. If eligible we will then provide a comprehensive overview of the investigation, explain what is expected of you, answer any questions you may have. If you agree to participate we will explain the process and ask you to provide informed consent. Participation will involve you attending our laboratory for up to 60 minutes on three separate occasions. We would like you to bring comfortable, tight-fitting clothing. On the first occasion, we will discuss participation and collect informed consent if you agree to participate. During this session, we will familiarize you with the procedures that will take place during the study. For the experimental trials we will need to place small devices for motion capture of the trunk and pelvis. On another occasion, you will be asked to perform 35 repetitions of a lifting task from the floor with a load. On the other occasion, you will be asked to perform 35

repetitions of a lifting task from the floor with a dowel rod. During these trials, we will record your joint movements. The order in which the loaded and unloaded trials take place will be randomized.

How much of my time will I need to give? Each of the 3 visits to the laboratory will require your attendance for up to 60 minutes.

What benefits will I, and/or the broader community, receive for participating? The immediate impact of this research is that we will be able to determine whether there are differences in movement during the lifting task between healthy populations and those with chronic low back pain. This will help inform exercise prescription for practitioners and assist individuals to overcome exercise avoidance due to fear which may alleviate symptoms and improve recovery.

Will the study involve any risk or discomfort for me? If so, what will be done to rectify it? During the repetitive lifting trials, normal exercise related discomfort due to physical exertion and muscular effort may be experienced. This type of discomfort is usually resolved quickly following the stopping of exercise. It is not impossible that performing repetitive lifting may result in an exacerbation of pain. However, previous research in chronic LBP populations has shown performing lifting tasks with load to be safe with no participants withdrawing from these two studies (1, 2). Feedback regarding your wellbeing, with regards to current level of low back pain will be monitored during the practice session. If you feel uncomfortable during the trial, you can stop lifting at any time. All adverse incidents will be recorded and reported to appropriate personnel. The familiarization session will also provide you with an understanding of what will be expected during the experimental trials. Normal, mild discomfort following exercise does not require medication or treatment and will be alleviated within 24 to 48H. You are welcome to remain at the facility for as long as you like until any exercise related symptoms are alleviated. Passive stretching will be instructed and encouraged to reduce the likelihood of soreness post-exercise.

In the unlikely event that you experience an exacerbation of pain or new injury related to the study after returning home from the laboratory, you should contact a research team member ASAP for supportive advice, immediately cease activity that aggravates, engage comfortable rest and to seek medical assessment \ treatment. This treatment should include any or all of: taking standard pain medication; First Aid treatment such as Rest, Ice, Compression, Elevation (RICE), visiting your general practitioner; or attending an Emergency department.

1. Asgari N, Sanjari MA, Esteki A. Local dynamic stability of the spine and its coordinated lower joints during repetitive Lifting: Effects of fatigue and chronic low back pain. Human movement science. 2017 Aug 1;54:339-46.
2. Graham RB, Oikawa LY, Ross GB. Comparing the local dynamic stability of trunk movements between varsity athletes with and without non-specific low back pain. Journal of Biomechanics. 2014 Apr 11;47(6):1459-64.

How do you intend to publish or disseminate the results? It is anticipated that the results of this research project will be published and/or presented in a variety of academic forums. In any publication and/or presentation, information will be provided in such a way that the individual participant data or identity cannot be identified. Within any publication or presentation, only group averages and trends from the dataset will be presented.

Will the data and information that I have provided be disposed of? Only the researchers will have access to the information you provide and raw data collected. However, your de-identified data may be used in other related projects in future investigations. These projects may include low back pain and fatigue studies. Please note that the minimum retention period for participant data is seven years post publication. After this time, all data collected and the information you provide will be securely disposed of, all hard-copy data will be destroyed, and all electronic data will be permanently deleted.

Can I withdraw from the study? Participation is entirely voluntary and you are not obliged to be involved. If you do participate you can withdraw at any time without giving reason and with no consequence brought upon you for doing so. Should you choose to exercise your voluntary right and withdraw from the study at any time, all of your collected participant data will be destroyed.

Can I tell other people about the study? Yes, you can tell other people about the study. If they are interested in participating, please advise them to contact the principal investigators John Marquez or Peter Clothier via the contact details listed below.

What if I require further information? Please contact a member of the research team below should you wish to discuss the research further before deciding whether or not to participate

Mr. John Marquez, B.HSci
E-mail: 18749449@student.westernsydney.edu.au

Dr. Peter Clothier, PhD
Lecturer, Sport & Exercise Science
School of Health Sciences
E-mail: p.clothier@westernsydney.edu.au

What if I have a complaint? If you have any complaints or reservations about the ethical conduct of this research, you may contact the Ethics Committee through Research Engagement, Development and Innovation (REDI) on Tel +61 2 4736 0229 or email humanethics@westernsydney.edu.au.

Any issues you raise will be treated in confidence and investigated fully, and you will be informed of the outcome.

If you agree to participate in this study, you will be asked to sign the Participant Consent Form. The information sheet is for you to keep and the consent form is retained by the researcher/s.

This study has been approved by the Western Sydney University Human Research Ethics Committee. The Approval number is H [enter approval number once the project has been approved].

What will happen with my information if I agree to it being used in projects other than this one? Thank you for considering being a participant in a University research project. The researchers are asking that you agree to supply your information (data) for use in this project and to also agree to allow the data to potentially be used in future research projects.

This request is in line with current University and government policy that encourages the re-use of data once it has been collected. Collecting information for research can be an inconvenience or burden for participants and has significant costs associated with it. Sharing your data with other researchers gives potential for others to reflect on the data and its findings, to re-use it with new insight, and increase understanding in this research area.

You have been asked to agree to **Extended consent.** When you agree to extended consent it means that you agree that your data, as part of a larger dataset (the information collected for this project) can be re-used in projects that are:

- an extension of this project
- closely related to this project
- in the same general area of this research.

The researchers will only allow this data to be used by other members of the research team (John Marquez, Peter Clothier, Paul Marshall, and Michael Knox) in work they are a registered investigator and contributing author. Such work could involve collaboration with scientists from other institutions.

To enable this re-use, your data will be held at the University in its data repository and managed under a Data Management Plan. The stored data available for re-use will not have information in it that makes you identifiable. The re-use of the data will only be allowed after an ethics committee has agreed that the new use of the data meets the requirements of ethics review. The researchers want to keep the data for 5 years for possible re-use. After this time the data will be securely destroyed.

You are welcome to discuss these issues further with the researchers before deciding if you agree. You can also find more information about the re-use of data in research in the National Statement on Ethical Conduct in Human Research – see Sections 2.2.14 - 2.2.18.

https://www.nhmrc.gov.au/about-us/publications/national-statement-ethical-conduct-human-research-2007-updated-2018

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Appendix E: Fear Avoidance Beliefs Questionnaire

Fear-Avoidance Beliefs Questionnaire (FABQ)
Waddell et al (1993) Pain , 52 (1993) 157 - 168

For each statement please circle any number from 0 to 6 to say how much physical activities such as bending, lifting, walking or driving affect or would affect *your* back pain.

	Completely disagree			Unsure			Completely agree
1. My pain was caused by physical activity……..	0	1	2	3	4	5	6
2. Physical activity makes my pain worse…….	0	1	2	3	4	5	6
3. Physical activity might harm my back……..	0	1	2	3	4	5	6
4. I should not do physical activities which (might) make my pain worse	0	1	2	3	4	5	6
5. I cannot do physical activities which (might) make my pain worse…...	0	1	2	3	4	5	6

The following statements are about how your normal work affects or would affect your back pain

	Completely disagree			Unsure			Completely agree
6. My pain was caused by my work or by an accident at work……	0	1	2	3	4	5	6
7. My work aggravated my pain……	0	1	2	3	4	5	6
8. I have a claim for compensation for my pain……	0	1	2	3	4	5	6
9. My work is too heavy for me……..	0	1	2	3	4	5	6
10. My work makes or would make my pain worse……..	0	1	2	3	4	5	6
11. My work might harm my back……..	0	1	2	3	4	5	6
12. I should not do my normal work with my present pain…….	0	1	2	3	4	5	6
13. I cannot do my normal work with my present pain……...	0	1	2	3	4	5	6
14. I cannot do my normal work till my pain is treated……..	0	1	2	3	4	5	6
15. I do not think that I will be back to my normal work within 3 months.	0	1	2	3	4	5	6
16. I do not think that I will ever be able to go back to that work……...	0	1	2	3	4	5	6

Scoring
Scale 1: fear-avoidance beliefs about work – items 6, 7, 9, 10, 11, 12, 15.
Scale 2: fear-avoidance beliefs about physical activity – items 2, 3, 4, 5.

Source: Gordon Waddell, Mary Newton, Iain Henderson, Douglas Somerville and Chris J. Main, A Fear-Avoidance Beliefs Questionnaire (FABQ) and the role of fear-avoidance beliefs in chronic low back pain and disability, *Pain*, 52 (1993) 157 – 168, 166.

Appendix F: Oswestry Disability Index

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 off 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

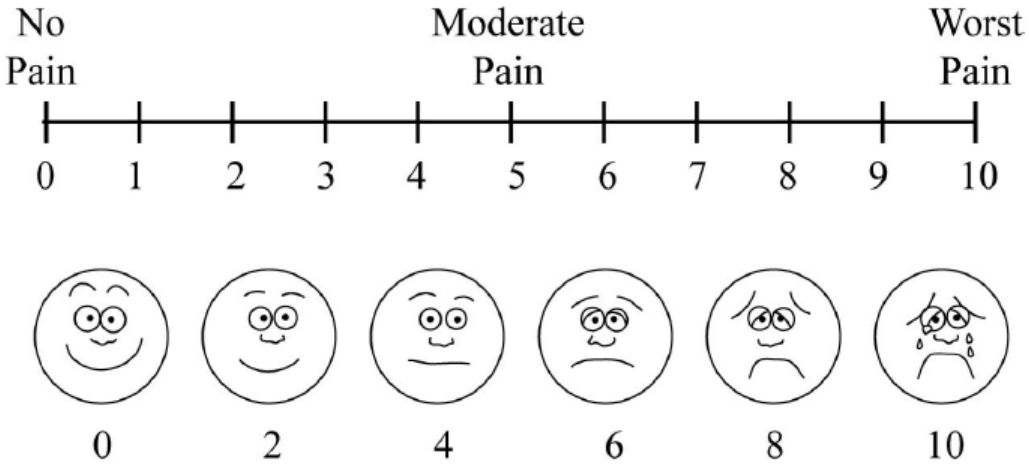
- ☐ My social life is normal and gives me no extra pain
- ☐ My social life is normal but increases the degree of pain
- ☐ Pain has no significant effect on my social life apart from limiting my more energetic interests eg, sport
- ☐ Pain has restricted my social life and I do not go out as often
- ☐ 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 G: VAS Pain Scale



Appendix H: MATLAB code for calculation of Intersegment Coordination

Data analysis script

```
clear; close all; clc; % Clears Workspace, Command Window, and closes Figures
```

Load Data

```
[File, Path] = uigetfile('.h5');
addpath(Path);
fileFormat = h5readatt(File,'/','FileFormatVersion');
info=h5info(File)
```

```
info = struct with fields:
      Filename: 'G:\My Drive\UNI\Masters Thesis\Data analysis\Control\Amy\AmyDowel.h5'
          Name: '/'
        Groups: [2×1 struct]
      Datasets: [1×1 struct]
     Datatypes: []
         Links: []
    Attributes: [1×1 struct]
```

```
if fileFormat < 5
error('TruncateHDF only works on fileFormat versions 5+');
end

sensors = h5info(File, '/Sensors');
processed = h5info(File, '/Processed');

for i = 1:size(sensors.Groups)
    label = h5readatt(File,[sensors.Groups(i).Name '/Configuration'], 'Label 0');
    if strfind(label, 'Sternum')
        Sternum_Quat = [h5read(File,[processed.Groups(i).Name '/Orientation'])]';
    else if strfind(label, 'Lumbar')
            Lumbar_Quat = [h5read(File,[processed.Groups(i).Name '/Orientation'])]';
    else if strfind(label, 'Right Upper Leg')
            Right_Thigh_Quat = [h5read(File,[processed.Groups(i).Name '/Orientation'])]';
    else if strfind(label, 'Right Lower Leg')
            Right_Shank_Quat = [h5read(File, [processed.Groups(i).Name '/Orientation'])]';
    end
    end
    end
    end
end
```

Create time vector

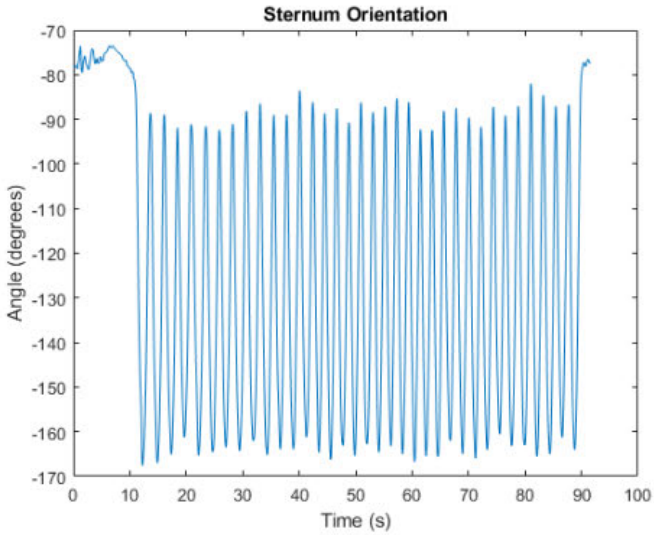
```
Hz = double(h5readatt(File,[sensors.Groups(1).Name '/Configuration'], 'Sample Rate'));
dt = 1/Hz;
t = 0:size(Sternum_Quat,1)-1;
tv = t*dt;
tt = tv(length(tv));

```

Convert from Quaternion to Euler Angles

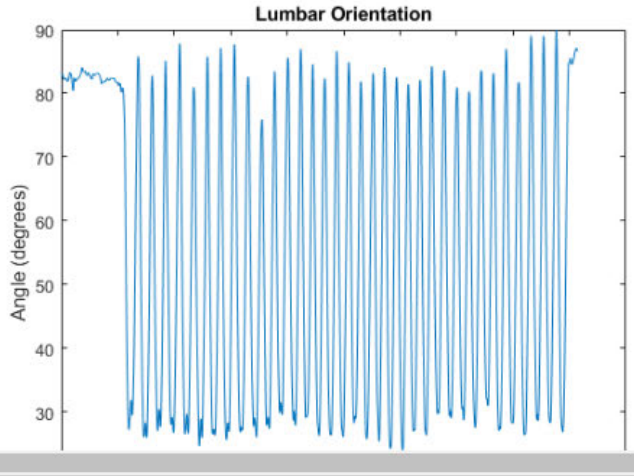
```
Sternum_angles = rad2deg(quat2eul(Sternum_Quat,'XYZ'));
Lumbar_angles = rad2deg(quat2eul(Lumbar_Quat,'XYZ'));
Thigh_angles = rad2deg(quat2eul(Right_Thigh_Quat,'XYZ'));
Shank_angles = rad2deg(quat2eul(Right_Shank_Quat,'XYZ'));

Sternum_angle = Sternum_angles(:,1);
Lumbar_angle = Lumbar_angles(:,1);
Thigh_angle = Thigh_angles(:,3);
Shank_angle = Shank_angles(:,3);

```

Plot Figures

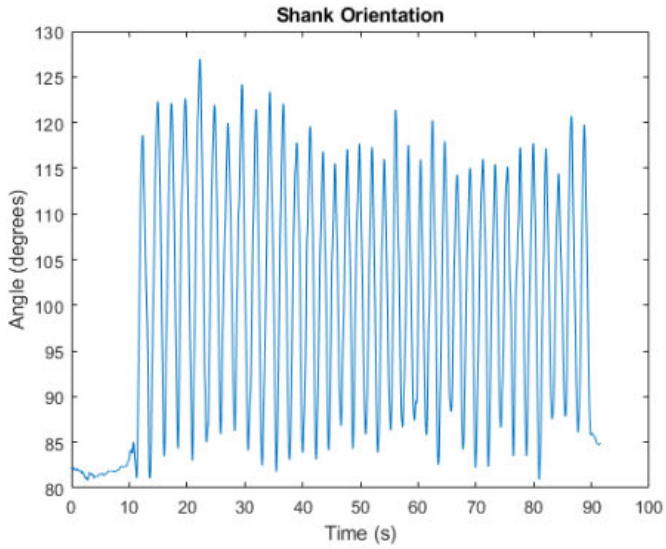
```
figure
plot(tv,Sternum_angle)
title('Sternum Orientation')
xlabel('Time (s)')
ylabel('Angle (degrees)')
hold on
```




```
figure
plot(tv,Lumbar_angle)
title('Lumbar Orientation')
xlabel('Time (s)')
ylabel('Angle (degrees)')
hold on
```

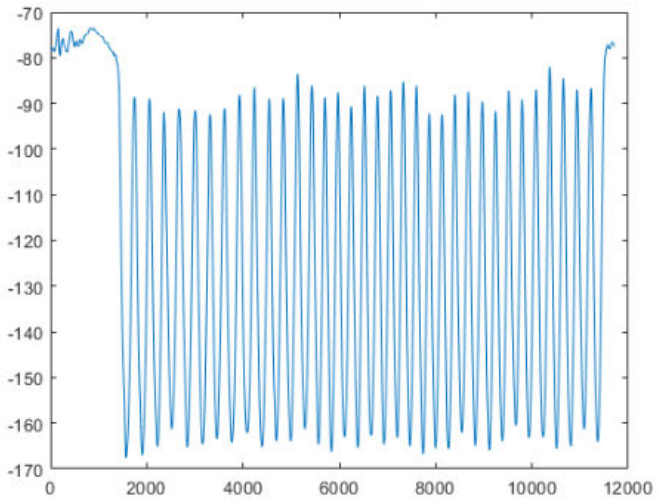


```
94
95  figure
96  plot(tv,Shank_angle)
97  title('Shank Orientation')
98  xlabel('Time (s)')
99  ylabel('Angle (degrees)')
100 hold on
```



Create vectors for analysis

```
101 figure
102 plot(Sternum_angle)
103 [Start_Stop,~] = ginput(2)
```

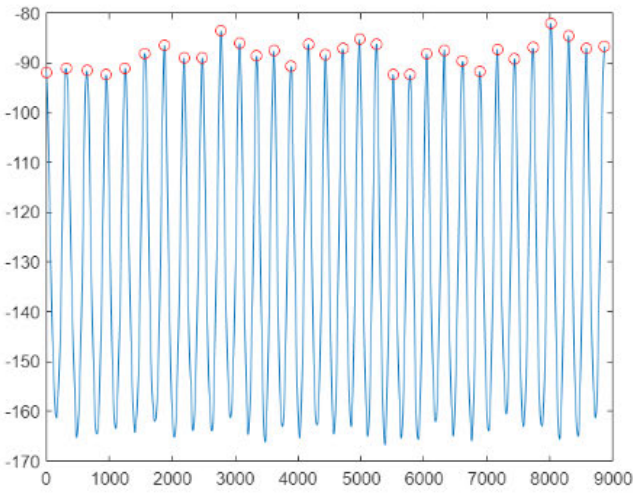



```
Start_Stop = 2×1
10^4 ×
      0.2392
      1.1184
```

```
close all

Start_Stop = round(Start_Stop);
Start_Stop(1) = find(Sternum_angle==max(Sternum_angle(Start_Stop(1)-200:Start_Stop(1)+200,1)));
Start_Stop(2) = find(Sternum_angle==max(Sternum_angle(Start_Stop(2)-200:Start_Stop(2)+200,1)));

Sternum_angle = Sternum_angle(Start_Stop(1)-1:Start_Stop(2)+1);
Lumbar_angle = Lumbar_angle(Start_Stop(1)-1:Start_Stop(2)+1);
Thigh_angle = Thigh_angle(Start_Stop(1)-1:Start_Stop(2)+1);
Shank_angle = Shank_angle(Start_Stop(1)-1:Start_Stop(2)+1);

for i = 2:size(Sternum_angle,1)-2
    Sternum_velocity(i) = (Sternum_angle(i+1)-Sternum_angle(i-1))/(2*dt);
    Lumbar_velocity(i) = (Lumbar_angle(i+1)-Lumbar_angle(i-1))/(2*dt);
    Thigh_velocity(i) = (Thigh_angle(i+1)-Thigh_angle(i-1))/(2*dt);
    Shank_velocity(i) = (Shank_angle(i+1)-Shank_angle(i-1))/(2*dt);
end

Sternum_angle = Sternum_angle(2:size(Sternum_angle,1)-1);
Lumbar_angle = Lumbar_angle(2:size(Lumbar_angle,1)-1);
Thigh_angle = Thigh_angle(2:size(Thigh_angle,1)-1);
Shank_angle = Shank_angle(2:size(Shank_angle,1)-1);
```

Calculate Cycles

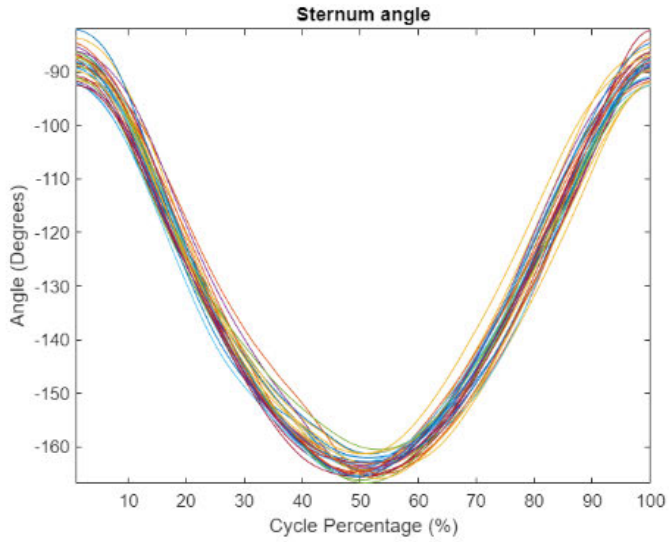
```
[Ster_pFind, Stern_pFind2] = findpeaks(Sternum_angle,'MinPeakHeight',mean(Sternum_angle),'MinPeakDistance'

Stern_pFind2 = [1; Stern_pFind2; length(Sternum_angle)];
Ster_pFind = Sternum_angle(Stern_pFind2);

figure
plot(1:length(Sternum_angle),Sternum_angle,Stern_pFind2,Ster_pFind,'ro')
hold on
```



Resample Cycles

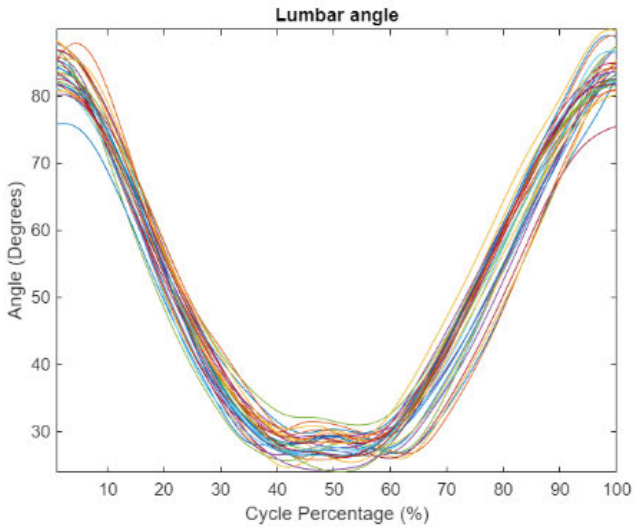
Resample Cycles

```
for i = 1:size(Ster_pFind,1)-1

    Temporary_SA_Sig = Sternum_angle(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_LA_Sig = Lumbar_angle(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_TA_Sig = Thigh_angle(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_ShA_Sig = Shank_angle(Stern_pFind2(i):Stern_pFind2(i+1));

    Temporary_SV_Sig = Sternum_velocity(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_LV_Sig = Lumbar_velocity(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_TV_Sig = Thigh_velocity(Stern_pFind2(i):Stern_pFind2(i+1));
    Temporary_ShV_Sig = Shank_velocity(Stern_pFind2(i):Stern_pFind2(i+1));

    SternumA_Cycles(:,i) = interp1(1:length(Temporary_SA_Sig),Temporary_SA_Sig,...
    (1:101)/101*length(Temporary_SA_Sig),'spline');
    LumbarA_Cycles(:,i) = interp1(1:length(Temporary_LA_Sig),Temporary_LA_Sig,...
    (1:101)/101*length(Temporary_LA_Sig),'spline');
    ThighA_Cycles(:,i) = interp1(1:length(Temporary_TA_Sig),Temporary_TA_Sig,...
    (1:101)/101*length(Temporary_TA_Sig),'spline');
    ShankA_Cycles(:,i) = interp1(1:length(Temporary_ShA_Sig),Temporary_ShA_Sig,...
    (1:101)/101*length(Temporary_ShA_Sig),'spline');

    SternumV_Cycles(:,i) = interp1(1:length(Temporary_SV_Sig),Temporary_SV_Sig,...
    (1:101)/101*length(Temporary_SV_Sig),'spline');
    LumbarV_Cycles(:,i) = interp1(1:length(Temporary_LV_Sig),Temporary_LV_Sig,...
    (1:101)/101*length(Temporary_LV_Sig),'spline');
    ThighV_Cycles(:,i) = interp1(1:length(Temporary_TV_Sig),Temporary_TV_Sig,...
    (1:101)/101*length(Temporary_TV_Sig),'spline');
    ShankV_Cycles(:,i) = interp1(1:length(Temporary_ShV_Sig),Temporary_ShV_Sig,...
    (1:101)/101*length(Temporary_ShV_Sig),'spline');
end

figure
plot(SternumA_Cycles)
title('Sternum angle')
xlabel('Cycle Percentage (%)')
ylabel('Angle (Degrees)')
axis([1 100 -inf inf])
hold on
```

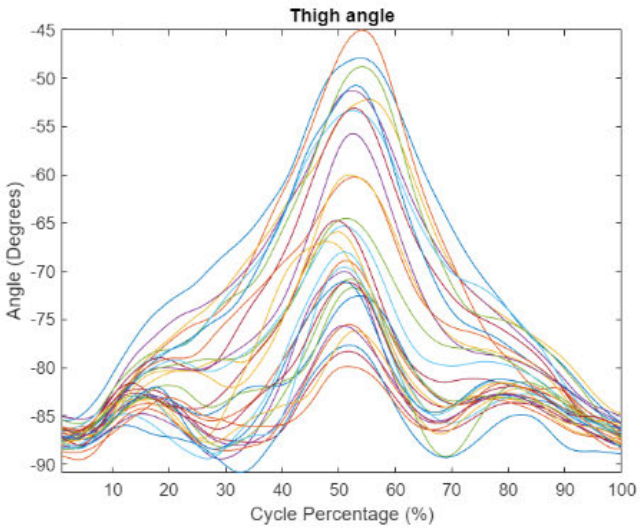


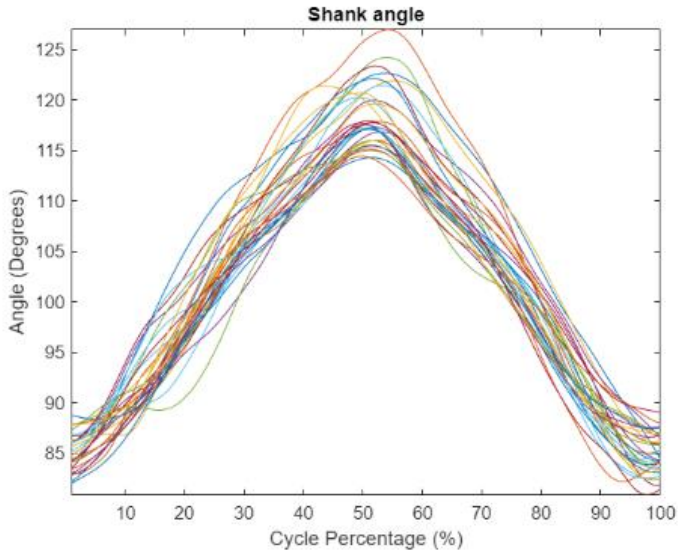

```
figure
```

```
figure
plot(LumbarA_Cycles)
title('Lumbar angle')
xlabel('Cycle Percentage (%)')
ylabel('Angle (Degrees)')
axis([1 100 -inf inf])
hold on
```

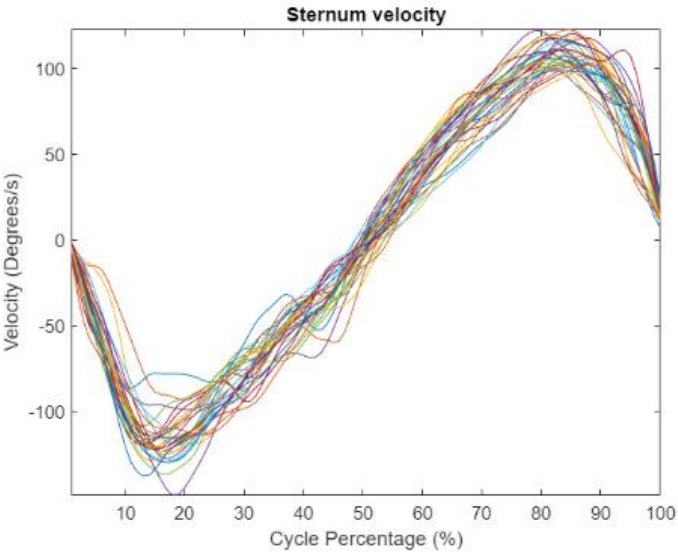



```
figure
plot(ThighA_Cycles)
title('Thigh angle')
xlabel('Cycle Percentage (%)')
ylabel('Angle (Degrees)')
axis([1 100 -inf inf])
hold on
```

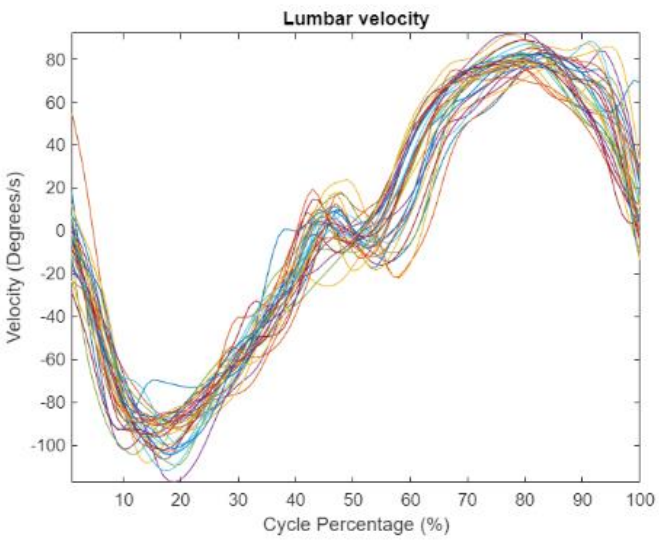




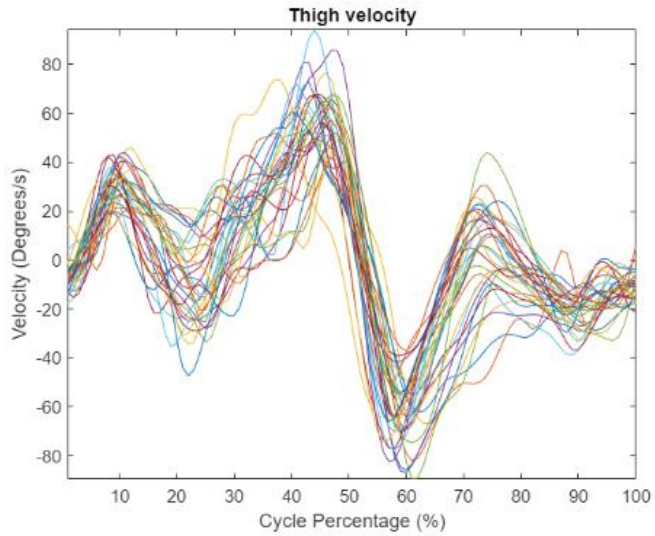

```
figure
plot(SternumV_Cycles)
title('Sternum velocity')
xlabel('Cycle Percentage (%)')
ylabel('Velocity (Degrees/s)')
axis([1 100 -inf inf])
hold on
```



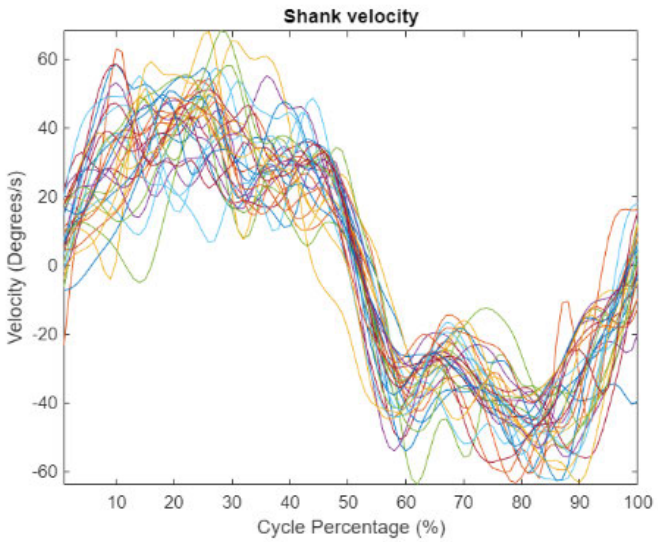

```
figure
plot(LumbarV_Cycles)
title('Lumbar velocity')
xlabel('Cycle Percentage (%)')
ylabel('Velocity (Degrees/s)')
axis([1 100 -inf inf])
hold on
```




```
figure
plot(ThighV_Cycles)
title('Thigh velocity')
xlabel('Cycle Percentage (%)')
ylabel('Velocity (Degrees/s)')
axis([1 100 -inf inf])
hold on
```

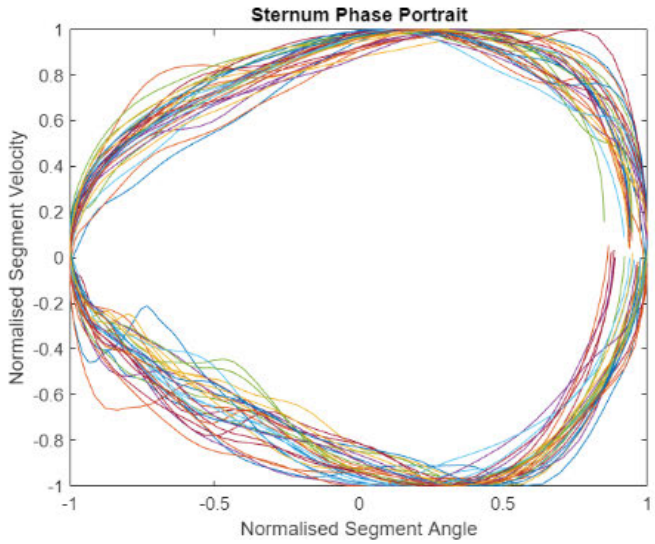



```
figure
plot(ShankV_Cycles)
title('Shank velocity')
xlabel('Cycle Percentage (%)')
ylabel('Velocity (Degrees/s)')
axis([1 100 -inf inf])
hold on
```

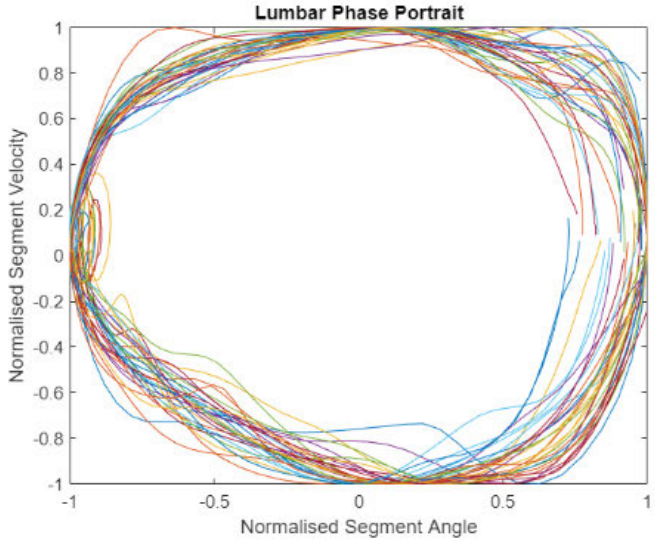


Normalised Signals

```
for j = 1:size(SternumA_Cycles,2);
    i = 1:101;

    SternumA_cNorm(i,j) = 2*(SternumA_Cycles(i,j)-min(SternumA_Cycles(:,j)))/...
        (max(SternumA_Cycles(:,j))-min(SternumA_Cycles(:,j)))-1;
    LumbarA_cNorm(i,j) = 2*(LumbarA_Cycles(i,j)-min(LumbarA_Cycles(:,j)))/...
        (max(LumbarA_Cycles(:,j))-min(LumbarA_Cycles(:,j)))-1;
    ThighA_cNorm(i,j) = 2*(ThighA_Cycles(i,j)-min(ThighA_Cycles(:,j)))/...
        (max(ThighA_Cycles(:,j))-min(ThighA_Cycles(:,j)))-1;
    ShankA_cNorm(i,j) = 2*(ShankA_Cycles(i,j)-min(ShankA_Cycles(:,j)))/...
        (max(ShankA_Cycles(:,j))-min(ShankA_Cycles(:,j)))-1;

        SternumV_cNorm(i,j) = 2*(SternumV_Cycles(i,j)-min(SternumV_Cycles(:,j)))/...
        (max(SternumV_Cycles(:,j))-min(SternumV_Cycles(:,j)))-1;
    LumbarV_cNorm(i,j) = 2*(LumbarV_Cycles(i,j)-min(LumbarV_Cycles(:,j)))/...
        (max(LumbarV_Cycles(:,j))-min(LumbarV_Cycles(:,j)))-1;
    ThighV_cNorm(i,j) = 2*(ThighV_Cycles(i,j)-min(ThighV_Cycles(:,j)))/...
        (max(ThighV_Cycles(:,j))-min(ThighV_Cycles(:,j)))-1;
    ShankV_cNorm(i,j) = 2*(ShankV_Cycles(i,j)-min(ShankV_Cycles(:,j)))/...
        (max(ShankV_Cycles(:,j))-min(ShankV_Cycles(:,j)))-1;

end


figure
plot(SternumA_cNorm,SternumV_cNorm)
title('Sternum Phase Portrait')
xlabel('Normalised Segment Angle')
ylabel('Normalised Segment Velocity')
hold on
```

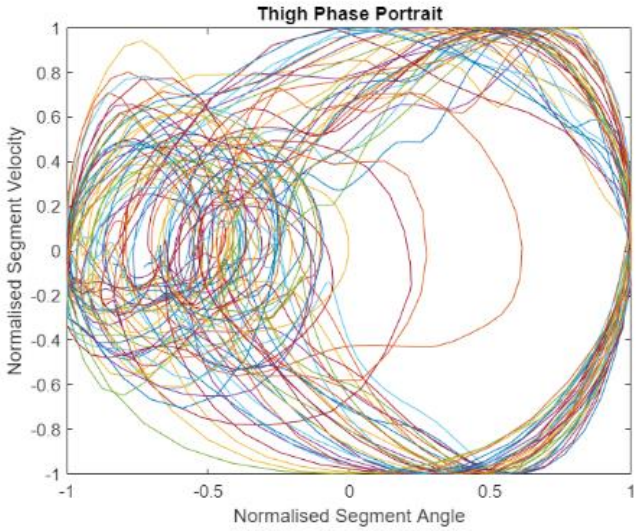



```
figure
plot(LumbarA_cNorm,LumbarV_cNorm)
title('Lumbar Phase Portrait')
xlabel('Normalised Segment Angle')
ylabel('Normalised Segment Velocity')
hold on
```

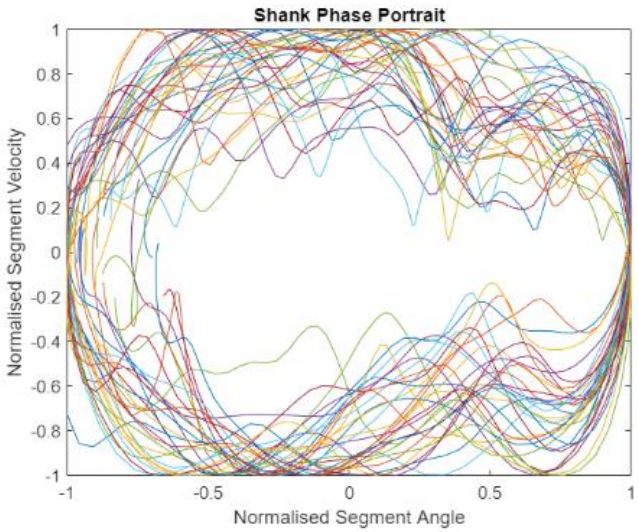



```
figure
plot(ThighA_cNorm,ThighV_cNorm)
title('Thigh Phase Portrait')
xlabel('Normalised Segment Angle')
ylabel('Normalised Segment Velocity')
hold on
```

```
figure
plot(ThighA_cNorm,ThighV_cNorm)
title('Thigh Phase Portrait')
xlabel('Normalised Segment Angle')
ylabel('Normalised Segment Velocity')
hold on
```



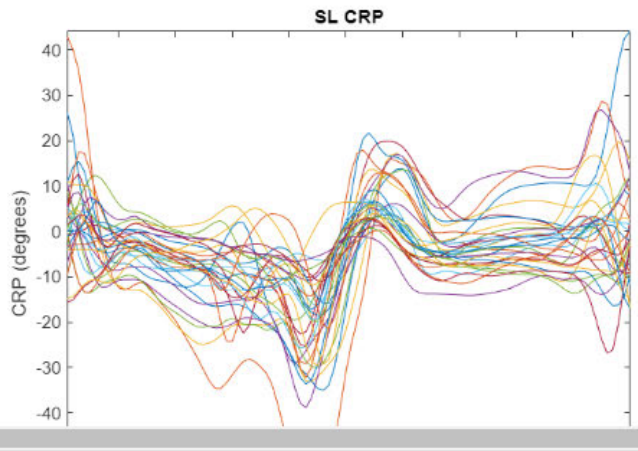

```
figure
plot(ShankA_cNorm,ShankV_cNorm)
title('Shank Phase Portrait')
xlabel('Normalised Segment Angle')
ylabel('Normalised Segment Velocity')
hold on
```



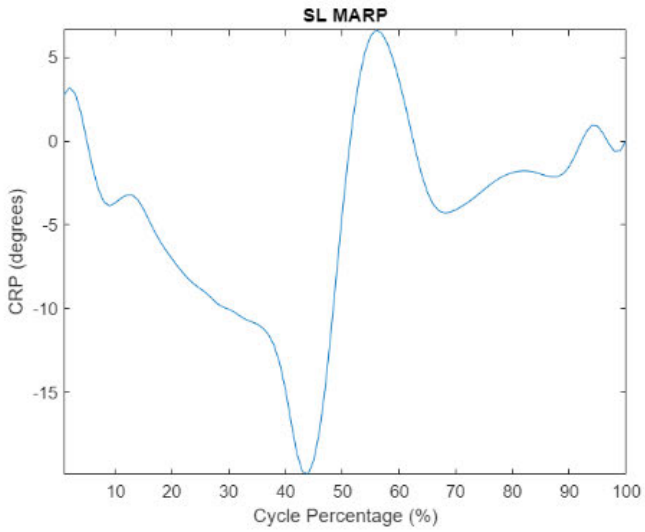
Phase Angle, CRP, MARP & DP Calculations

Phase Angle, CRP, MARP & DP Calculations

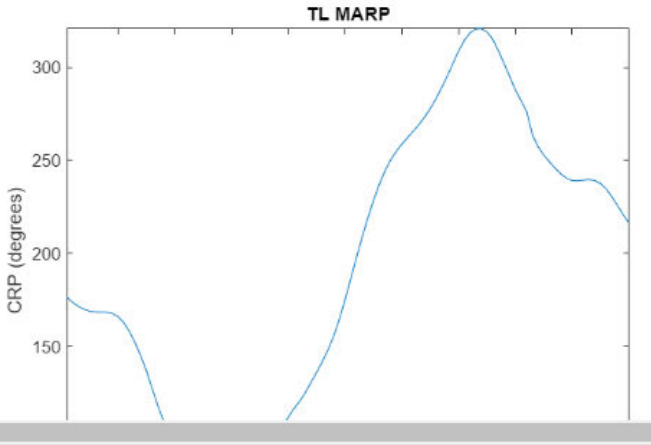
```
Sternum_PA = rad2deg(atan2(SternumV_cNorm,SternumA_cNorm));
Lumbar_PA = rad2deg(atan2(LumbarV_cNorm,LumbarA_cNorm));
Thigh_PA = rad2deg(atan2(ThighV_cNorm,ThighA_cNorm));
Shank_PA = rad2deg(atan2(ShankV_cNorm,ShankA_cNorm));

SL_CRP = Lumbar_PA-Sternum_PA;
TL_CRP = Lumbar_PA-Thigh_PA;
ST_CRP = Thigh_PA-Shank_PA;

SL_CRP2 = abs(Lumbar_PA)-abs(Sternum_PA);
TL_CRP2 = abs(Lumbar_PA)-abs(Thigh_PA);
ST_CRP2 = abs(Thigh_PA)-abs(Shank_PA);

for i = 1:size(SL_CRP,1)
    for j = 1:size(SL_CRP,2)
        if SL_CRP(i,j) > 180
            SL_CRP(i,j) = SL_CRP(i,j)-360;
        elseif SL_CRP(i,j) < -180
            SL_CRP(i,j) = SL_CRP(i,j)+360;
        else
            SL_CRP(i,j) = SL_CRP(i,j);
        end
    end
end

for i = 1:size(SL_CRP,1)
    for j = 1:size(SL_CRP,2)
        if SL_CRP(i,j) < -90
            SL_CRP(i,j) = SL_CRP(i,j)+360;
        else
            SL_CRP(i,j) = SL_CRP(i,j);
        end
    end
end



figure
plot(SL_CRP)
title('SL CRP')
xlabel('Cycle Percentage (%)')
ylabel('CRP (degrees)')
axis([1 100 -inf inf])
hold on
```



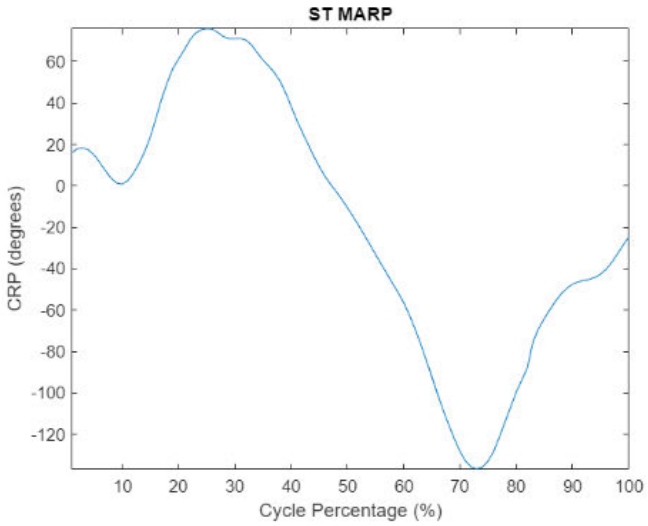
```
396
397  for i = 1:101
398  SL_MARP(i) = mean(SL_CRP(i,:));
399  SL_DP(i) = std(SL_CRP(i,:));
400  TL_MARP(i) = mean(TL_CRP(i,:));
401  TL_DP(i) = std(TL_CRP(i,:));
402  ST_MARP(i) = mean(ST_CRP(i,:));
403  ST_DP(i) = std(ST_CRP(i,:));
404  end
405  figure
406  plot(SL_MARP)
407  title('SL MARP')
408  xlabel('Cycle Percentage (%)')
409  ylabel('CRP (degrees)')
410  axis([1 100 -inf inf])
411  hold on
```




```
412  figure
413  plot(TL_MARP)
414  title('TL MARP')
415  xlabel('Cycle Percentage (%)')
416  ylabel('CRP (degrees)')
417  axis([1 100 -inf inf])
418  hold on
```



```
419
420  figure
421  plot(ST_MARP)
422  title('ST MARP')
423  xlabel('Cycle Percentage (%)')
424  ylabel('CRP (degrees)')
425  axis([1 100 -inf inf])
426  hold on
```




```
427
428  SL_MARP_FE = sum(SL_MARP)/101;
429  SL_MARP_F = sum(SL_MARP(1:50))/50;
430  SL_MARP_E = sum(SL_MARP(51:101))/51;
431
432  SL_DP_FE = sum(SL_DP)/101;
433  SL_DP_F = sum(SL_DP(1:50))/50;
434  SL_DP_E = sum(SL_DP(51:101))/51;
435
436  TL_MARP_FE = sum(TL_MARP)/101;
437  TL_MARP_F = sum(TL_MARP(1:50))/50;
438  TL_MARP_E = sum(TL_MARP(51:101))/51;
439
440  TL_DP_FE = sum(TL_DP)/101;
441  TL_DP_F = sum(TL_DP(1:50))/50;
442  TL_DP_E = sum(TL_DP(51:101))/51;
443
444  ST_MARP_FE = sum(ST_MARP)/101;
445  ST_MARP_F = sum(ST_MARP(1:50))/50;
446  ST_MARP_E = sum(ST_MARP(51:101))/51;
447
448  ST_DP_FE = sum(ST_DP)/101;
449  ST_DP_F = sum(ST_DP(1:50))/50;
450  ST_DP_E = sum(ST_DP(51:101))/51;
451
452  a = [SL_MARP_FE;SL_MARP_F;SL_MARP_E;SL_DP_FE;...
453      SL_DP_F;SL_DP_E;TL_MARP_FE;TL_MARP_F;TL_MARP_E;...
454      TL_DP_FE;TL_DP_F;TL_DP_E;ST_MARP_FE;ST_MARP_F;...
455      ST_MARP_E;ST_DP_FE;ST_DP_F;ST_DP_E];
456
457  % SL_CRP2 = rad2deg(atan2(SternumV_cNorm.*LumbarA_cNorm-LumbarV_cNorm.*SternumA_cNorm,...
458  %     SternumA_cNorm.*LumbarA_cNorm+SternumV_cNorm.*LumbarV_cNorm));
459
```

Appendix I: R code for computation of Local Dynamic Stability

```
setwd("C:/Users/User/Documents/R")

#load in the tseries package

library(tseriesChaos)

#input the raw data and convert

aknee=read.csv("interpolated knee.csv")
ahip=read.csv("interpolated hip.csv")
aback=read.csv("interpolated back.csv")

#adjust second reference value for the absolute time of the lifting trial
#use aa function to calculate length of interpolated data and correct for sampling frequency of 128Hz

aa=ts(aknee)

t=length(aa)/128

knee=ts(aknee, 0, t, 128)
hip=ts(ahip, 0, t, 128)
back=ts(aback, 0, t, 128)

#run the Lyapunov Exponent, d is 10% of cycle, s is average cycle length

dd=length(knee)
cycle=8000/30
cyclength=cycle*1/128

#Run script to here to obtain cycle length, divide by 10 for d, while s=cycle length

#m is the embedding dimension, d is the time delay (set to 10% of cycle length), k number of
#considered neighbours, eps radius to find neighbours, t is Theiler window, ref is number of points to take into account

output1=lyap_k(knee, m=6, d=26, t=2, k=8, ref=7800, s=267,  eps=3)
output2=lyap_k(hip, m=6, d=26, t=2, k=8, ref=7800, s=267,  eps=3)
output3=lyap_k(back, m=6, d=26, t=2, k=8, ref=7800, s=267,  eps=3)

#plot the outputs

max(output1)

plot(output1)

#Extract half the Lyapunov Exponent for Linear Curve Fitting in Excel

forplot1=window(output1, 0, cyclength/2)

forplot2=window(output2, 0, cyclength/2)

forplot3=window(output3, 0, cyclength/2)

library(zoo)

write.zoo(forplot1, "knee analyze.txt")
write.zoo(forplot2, "hip analyze.txt")
write.zoo(forplot3, "back analyze.txt")

#END
```