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Trunk and lower limb coordination during lifting in people with and without chronic low back pain

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Abstract

Differences in synchronous movement between the trunk and lower limb during lifting have been reported in chronic low back pain (CLBP) patients compared to healthy people. However, the relationship between movement coordination and disability in CLBP patients has not been investigated. A cross-sectional study was conducted to compare regional lumbar and lower limb coordination between CLBP (n = 43) and control (n = 29) groups. The CLBP group was divided into high- and low-disability groups based on their Oswestry Disability Index (ODI) score. The mean absolute relative phase (MARP) angles and mean deviation phase (DP) between the 1) lumbar spine and hip, and 2) hip and knee were measured. The relationship between MARP angle and DP and ODI were investigated using linear regression. The higher-disability CLBP group demonstrated significantly greater lumbar-hip MARP angles than the lower-disability CLBP group (mean difference = 12.97, % difference = 36, p = 0.041, 95% CI [2.97, 22.98]). The higher-disability CLBP group demonstrated significantly smaller hip-knee DP than controls (mean difference = 0.11, % difference = 76, p = 0.011, 95% CI [0.03, 0.19]). There were no significant differences in lumbar-hip and hip-knee MARP and DP between the lower-disability CLBP and control groups. Lumbar-hip MARP was positively associated with ODI ($R^2 = 0.92$, $\beta = 0.30$, $p = 0.048$). High-disability CLBP patients demonstrated decreased lumbar-hip movement coordination and stiffer hip-knee movement during lifting than low-disability CLBP patients and healthy controls.

1. Introduction

Lifting is a complex activity that requires coordination of the lower limbs and trunk (van Dieen et al., 1999). Poor movement coordination between the trunk and lower limb during lifting has been associated with the development of chronic low back pain (CLBP) given the increased
loading of bony and soft tissues (Coenen et al., 2014; Nelson et al., 1995). The kinematics of CLBP-related symmetrical lifting (i.e., lifting where the load is placed anteriorly about the body’s mid-sagittal plane (Lavender et al., 2003)) have been quantified by measuring the lumbar and hip range of motion (ROM) and angular velocity (Lariviere et al., 2002; McGregor et al., 1997; Sanchez-Zuriaga et al., 2011).

Studies assessing lifting-related lumbar ROM during symmetrical lifting in people with CLBP report inconsistent findings including increased (McGregor et al., 1997), decreased (Sanchez-Zuriaga et al., 2011) and no difference (Lariviere et al., 2002) in ROM compared to healthy people. Likewise, compared to healthy people, people with CLBP take longer to perform lifting tasks (Sanchez-Zuriaga et al., 2011). However, assessment of peak joint ROM and angular velocity does not provide any indication of inter-joint coordination during lifting. Thus, more sensitive and sophisticated analysis of lifting techniques are required to accurately assess trunk and lower limb coordination deficits in people with CLBP.

An alternative method for quantifying CLBP-related lifting kinematics is relative phase angle analysis – a technique used most commonly within the ergonomics literature for analyzing coordination between the trunk and lower limb joints during lifting (Burgess-Limerick et al., 1993). This approach provides continuous spatial and temporal measurement throughout the movement cycle given that phase angles are derived from joint displacement and joint velocity (Hamill et al., 1999; Stergiou et al., 2001). A previous study utilized this technique to compare inter-joint coordination of people with and without CLBP during a trunk extension movement (Mokhtarinia et al., 2016). Mokhtarinia et al. (2016) found that in people with CLBP, lumbar movement was more ‘in-phase’ with hip movement compared to healthy people during trunk extension – denoting stiffer lumbopelvic movement. Whilst interesting, this study has numerous
limitations. For instance, there was no measurement of vertical ground reaction force (GRF) which, in conjunction to trunk kinematics, has been used to identify compensatory movement strategies during functional tasks in people with CLBP (Shum et al., 2007b). Furthermore, CLBP participants were almost completely free of pain and disability at the time of testing. People with CLBP with higher disability levels have been found to exhibit more pronounced kinematic and kinetic mal-adaptations during lifting – as per reduced trunk ROM, lower limb ROM and vertical GRF’s through each leg compared to those with lower disability levels and healthy people (Sanchez-Zuriaga et al., 2011). Importantly, the association between lifting-related inter-joint coordination, vertical GRF’s and self-reported disability – commonly measured using the Oswestry Disability Index (ODI; (Fairbank and Pynsent, 2000)) – has not been previously investigated. Thus, it is currently unknown whether CLBP individuals with higher disability levels demonstrate different lifting-related trunk and lower limb inter-joint coordination and vertical GRF’s compared to those with lower disability levels and healthy people.

Therefore, the primary aim of this study was to compare lifting-related kinematics (i.e., lumbar ROM, lower limb ROM, angular velocity, lumbar-lower limb inter-joint coordination) and kinetics (i.e., vertical GRF) in CLBP with lower and higher disability levels and healthy control participants. The secondary aim was to investigate the relationship between lifting-related kinematic and kinetic variables and self-reported disability level in CLBP participants. We hypothesized that compared to healthy controls, people with CLBP would demonstrate impaired lifting-related kinematics and kinetics ($H_1$) (i.e., increased trunk-lower limb joint coordination (Mokhtarinia et al., 2016) and decreased vertical GRF (Sanchez-Zuriaga et al., 2011)). Moreover, significant positive associations would be observed between lifting-related kinematics and kinetics and self-reported disability in people with CLBP ($H_2$).
2. Methods

2.1. Participants

Forty-three participants ($n_{\text{female}} = 23$) aged 25-60 years with CLBP were recruited from a large Physiotherapy clinic in Melbourne, Australia. These participants were new patients of the clinic and, as per diagnostic criteria of non-specific CLBP (Von Korff et al., 1993), reported pain between the level of the twelfth thoracic vertebra (T12) and the gluteal fold that had persisted for >3 months. Participants were excluded if they presented with overt neurological signs such as muscle weakness, previous spinal surgery, systemic or inflammatory conditions such as rheumatoid arthritis, malignancy, unstable spondylolisthesis (i.e., specific diagnosis of CLBP (Maitland et al., 2005)) or inability to understand written or spoken English. In addition, a group of 29 healthy control participants (age-, gender- and BMI-matched) with no history of CLBP were recruited from the community.

All participants completed the ODI (rated from 0-100% disability) and rated their pain out of 10 using the Numerical Rating Scale immediately before and after testing. The CLBP cohort was divided into low (ODI ≤ 20%) and moderate-high disability (ODI > 20%) sub-groups based upon their level of self-reported ODI disability (Fairbank and Pynsent, 2000). Ethics approval was obtained from The University of Melbourne’s Behavioural and Social Sciences Human Ethics Committee (ethics ID: 1340715). All participants provided written informed consent prior to entering the study.

2.2. Experimental procedures

Twenty-one retro-reflective markers of 13 mm diameter were attached to anatomical landmarks of each participant using double sided tape. The thorax, pelvis, thigh and lower leg segments were formed using three retro-reflective markers per segment (see Figure 1). The thorax marker
configuration used was similar to Christe et al. (2016) and is valid for investigating lumbar movement relative to the pelvis in the sagittal plane (Burgess-Limerick et al., 1993; Kippers and Parker, 1989). Hip (pelvis to thigh segments) and knee (thigh to shank) flexion angles were defined using the longitudinal axes of each segment. Participants were instructed to step onto two Wii Balance Boards (WBB; Nintendo, Kyoto, Japan), one under each of the feet (Right and Left) as described previously (Clark et al., 2014). An 8-kg kettlebell was then placed between the WBB’s, 5 cm in front of the participant’s toes. Participants were then instructed to lift the kettlebell up to the level of their abdomen using a self-selected pace and technique (see Figure 1). This task was repeated twice, the first being a familiarisation trial which was not analysed. Kinematic data were collected using a 12-camera Optitrack Flex 13 motion analysis system (NaturalPoint, Corvallis, OR) at a sampling rate of 120 Hz. The WBB’s were connected to a laptop computer via a wireless Bluetooth connection. Use of WBB’s for measurement of kinetic variables has been shown to be valid and reliable (Clark et al., 2010).

2.3. Data analysis

2.3.1. Kinematic data

The kinematic data was cleaned and gap-filled using Optitrack Motive software (NaturalPoint, Corvallis, OR) and then passed through a custom written analysis pipeline (Visual3D v5.01.6, C-Motion, Inc., Germantown, MD). Angular displacement and angular velocity data were then derived using custom written LabVIEW 2009 (National Instruments, Austin, TX) software. Joint angle data were filtered using a fourth order zero-phase shift low-pass Butterworth filter with a 6 Hz cut-off frequency (Bartlett, 2007). The start and end positions for symmetrical lifting were maximal lumbar flexion and lumbar extension (within participant comfort), respectively (see
Figure 3). As per Hamill et al. (1999), angular displacement ($\theta_i$) and angular velocity ($\omega_i$) data were normalized to -1 to +1 intervals using the equation:

$$Normalized \theta_i = \left( \frac{[\theta_i - \min(\theta)]}{[\max(\theta) - \min(\theta)]} \right) \times 2 - 1$$

$$Normalized \omega_i = \frac{\omega_i}{\max(|\omega|)}$$

Where $i$ is each iteration in the data array, $\min(\theta)$ and $\max(\theta)$ are minimal and maximal points in angular displacement data array respectively and $\max(|\omega|)$ is the maximal value in the absolute angular velocity data array.

Phase plane of each segment was obtained by plotting the normalized $\omega_i$ (vertical axis) as a function of normalized $\theta_i$ (horizontal axis). As per Stergiou et al. (2001), the phase angles ($\phi_i$) for each segment were derived using the equation:

$$\phi_i = \tan^{-1} \frac{Normalized \omega_i}{Normalized \theta_i}$$

A two-quadrant (0-180°) inverse tangent was used for this equation (Hamill et al., 1999). The continuous relative phase curve for each segment couple was plotted by subtracting the phase angles of the proximal from the distal joints at each data point (i.e., $\phi_{\text{hip}} - \phi_{\text{lumbar}}$ and $\phi_{\text{knee}} - \phi_{\text{hip}}$).

To quantify and compare relative phase curves between the groups, the mean absolute relative phase angle (MARP) values for each joint couple (i.e., lumbar-hip, hip-knee) were obtained by averaging the continuous phase angle values over the total data collection points using the formula of Stergiou et al. (2001):

$$MARP = \frac{\sum_{i=1}^{p} |\phi_{\text{relative phase}}|}{p}$$
Where $p$ is the number of points in each of the two periods. A MARP angle closer to $0^\circ$ indicates more in-phase movement coupling whereas values closer to $180^\circ$ suggests more out-of-phase movement coupling.

Deviation phase (DP) or coordination variability for each joint couple was also analyzed by averaging the standard deviation (SD) of the CRP curve over the total data collection points using the formula of Stergiou et al. (2001):

$$DP = \frac{\sum_{i=1}^{p} SDi}{p}$$

Where $p$ is the number of points in each of the two periods. A DP value closer to $0^\circ$ indicates less variable relationship between the segment couples. All lifting kinematic analyses were normalized to total lifting time.

**Kinetic data**

Kinetic data were filtered using an eighth order Butterworth filter with a low-pass cut-off frequency of 12 Hz (Clark et al., 2010). Filtered data were then processed using custom written LabVIEW 2009 (National Instruments, Austin, TX) software to derive maximum, minimum and mean vertical GRF’s for each foot in Newtons (N). All vertical GRF data were normalized to body weight. In our unpublished reliability testing in 17 CLBP and 16 healthy people, all testing variables demonstrated moderate to excellent reliability with ICC’s ranging from 0.64 to 0.98.

2.4. **Statistical analysis**

Descriptive data, kinematic, kinetic variables were normally distributed – confirmed with Shapiro-Wilk tests, and presented as means and standard deviations. Between-group comparisons of the dependent variables (i.e., ROM, angular velocity, MARP, DP, vertical GRF and lifting time) were assessed using ANCOVA with age as the covariate given that the CLBP$_{\text{high}}$ group was older than the control group. Vertical GRF were assessed using 2 (left and right sides)
(CLBP
\textsubscript{low}, CLBP
\textsubscript{high} and control groups) factorial ANCOVA. Significant main effect size was quantified using $\eta^2_p$. Normality, homoscedasticity and linearity of residual for ANCOVA were assessed using Levene’s test and scatter graphs (Osborne and Waters, 2002; Williams et al., 2013). When indicated, pairwise comparisons were performed \textit{post-hoc} using Fisher’s Least Significant Difference. The relationship between the independent variable (i.e., ODI) and the dependent variables (i.e., ROM, angular velocity, MARP, DP, vertical GRF) were investigated using linear regression. Average values of the left and right side vertical GRF’s were used for regression analyses. Variables that exhibited a significant correlation with the ODI were included in a linear regression model. All statistical analyses were conducted using SPSS Version 21.0 with $\alpha$ set at 0.05 (IBM, Inc., Chicago, IL).

3. Results

3.1. Participants

Participant characteristics are outlined in Table 1. Twenty-five CLBP participants had an ODI score of $\leq 20\%$ and grouped as low disability (CLBP
\textsubscript{low}). Eighteen CLBP participants had an ODI score $> 20\%$; therefore, they were grouped as moderate-high disability (CLBP
\textsubscript{high}).

The CLBP
\textsubscript{high} group was significantly older than the control group (mean difference = 9.0 years, $F_{2,69} = 3.5, p = 0.04, 95\% \ CI [0.8, 17.2]$). There was no change in pain level after assessment in all groups. There was a significant main effect of group for total time to complete the lifting task ($F_{2,69} = 9.67, \eta^2_p = 0.22, p < 0.001$). \textit{Post-hoc} comparisons demonstrated that the CLBP
\textsubscript{low} (mean difference with control = 0.74 seconds, % difference = 39.57, $p = 0.003, 95\% \ CI [0.22, 1.26]$) and CLBP
\textsubscript{high} (mean difference with control = 0.94 seconds, % difference = 50, $p = 0.001, 95\%\ CI [0.37, 1.51]$) groups took longer to complete the lifting task compared to the healthy group.
There was no significant difference between CLBP\textsubscript{low} and CLBP\textsubscript{high} in total lifting time ($p = 0.71$).

3.2. Joint range of motion and angular velocity

After adjusting for age, ANCOVA analyses demonstrated no significant between-group differences in joint ROM and angular velocities during the lifting task (see Table 2).

3.3. Inter-joint coordination

Between-group inter-joint coordination variable comparisons are summarised in Table 3. CLBP\textsubscript{high} group demonstrated significantly more lumbar-hip MARP than the CLBP\textsubscript{low} group (mean difference = 12.97, % difference = 36, $p = 0.041$, 95% CI [2.97, 22.98]). There were no significant differences in lumbar-hip MARP between CLBP\textsubscript{low} and control groups and CLBP\textsubscript{high} and control groups. The CLBP\textsubscript{high} group demonstrated significantly less hip-knee DP than the control group (mean difference = 0.11, % difference = 76, $p = 0.011$, 95% CI [0.03, 0.19]). There were no significant differences in hip-knee DP between CLBP\textsubscript{low} and control groups and CLBP\textsubscript{low} and CLBP\textsubscript{high} groups.

3.4. Vertical ground reaction forces

There were no statistically significant between-group differences for any of the vertical GRF variables (see Table 4). Moreover, there were no significant side-to-side differences for any vertical GRF variables.

3.5. Relationship between lifting kinematic and kinetic variables and disability

There was a significant correlation between ODI and lumbar-hip MARP ($r = 0.30$, $p = 0.048$). There was no significant correlation between other kinematic and kinetic variables and ODI. Lumbar-hip MARP predicted 9.2% of ODI ($B = 0.43$, $\beta = 0.30$, $p = 0.048$). The relationship between ODI and lumbar-hip MARP is depicted in Figure 4.
4. Discussion

CLBP patients with high-disability demonstrated decreased lifting-related lumbar-hip coordination compared to CLBP patients with lower disability. High-disability CLBP patients also demonstrated decreased hip-knee movement variability during lifting compared to healthy controls. There was no significant difference in lifting-related movement coordination patterns and movement variability between low-disability CLBP patients and controls. This is the first study examining lifting-related inter-joint coordination using relative phase angle analysis in people with and without CLBP across disability levels. In people with CLBP, decreased lumbar-hip movement coordination during lifting was associated with increased disability.

Analyses of joint ROM and angular velocity demonstrated no significant differences between CLBP and control groups for all joints (rejected H1). These findings are in agreement with past lifting studies (Lariviere et al., 2000, 2002) but in contrast with others (McGregor et al., 1997; Sanchez-Zuriaga et al., 2011; Shum et al., 2007a) that demonstrated that CLBP patients exhibit decreased lifting-related trunk ROM compared to controls. Laird et al. (2016) demonstrated a large between-day variability – up to 19° in lumbopelvic extension ROM, in CLBP patients during lumbar flexion-extension task. The large variation in lumbar ROM suggests that these fundamental kinematic variables may not be sensitive in differentiating between people with and without CLBP during symmetrical lifting tasks. Moreover, we did not phenotype (i.e., sub-group) our CLBP participants based on their pain-provoking movements (O'Sullivan, 2005; Sahrmann, 2002) but rather on disability level. When phenotyped based on painful movement directions, significant differences in ROM may be observed – for instance, CLBP patients who reported pain with lumbar flexion – classified under “flexion pattern” (O'Sullivan, 2005), demonstrated >10% more lumbar flexion during cycling than controls (Van Hoof et al., 2012).
High-disability CLBP patients demonstrated decreased lumbar-hip joint coordination (i.e., more lumbar-hip MARP) and decreased hip-knee movement variability (i.e., decreased hip-knee DP) (H1 rejected). Decreased lumbar-hip coordination is consistent with Silfies et al. (2009) who also reported increased lumbar-hip MARP during loaded forward reaching task in CLBP patients. Lifting, like forward reaching, requires a substantial contribution from the lumbar extensor muscles (e.g., erector spinae and multifidus) to control spinal movement (Burgess-Limerick et al., 1995; Silfies et al., 2009). However, in people with CLBP, lumbar extensor muscle function is impaired – as evident by decreased multifidus cross-sectional area (Hides et al., 2008), decreased lumbar proprioception (Brumagne et al., 2000; Willigenburg et al., 2013), delayed lumbar muscle activation in response to perturbation (Radebold et al., 2001) and, decreased lumbar extensor muscle force control (Pranata et al., 2017). Conceivably, these CLBP-related maladaptations could contribute to impaired lumbar-hip movement coordination during lifting. Decreased movement variability between the hip and knee (i.e., more rigid/guarded lower limb movement) could represent a strategy to maintain spinal posture in order to compensate for impaired lumbar-hip coordination associated with rapid trunk extension movement (McGregor and Hukins, 2009).

Our findings differ from those of Mokhtarinia et al. (2016) in CLBP and Shojaei et al. (2017) in acute LBP patients – who reported increased lumbar-hip coordination during flexion-extension tasks in CLBP and acute LBP patients, respectively. In our study and that of Silfies et al. (2009), participants used their arms to lift a load (i.e., 8 and 4.5 kg, respectively); however, in the study by Mokhtarinia et al. (2016), the 8 kg load was affixed to the trunk of the participants via a load vest whilst no load was lifted by participants in the Shojaei et al. (2017) study. Thus, loaded upper limb movements may have contributed to variations in lumbar-hip coordination between
studies. Assuming that our control group demonstrated the ‘ideal’ lumbar-hip coordination profile during lifting, findings of this study support the notion that CLBP patients can exhibit trunk-lower limb movement coordination patterns that are either too variable (i.e., decreased coordination) or too rigid (i.e., increased coordination) (Stergiou et al., 2006).

This study demonstrated a relationship between decreased lifting-related lumbar-hip coordination and increased self-reported disability (H2 partially accepted). A healthy neuromuscular system is characterized by the ability to adapt to internal and external perturbations (Stergiou and Decker, 2011). Decreased lifting-related lumbar-hip coordination with increasing disability suggests that CLBP severity contributes to a progressively more predictable neuromuscular system (i.e., more inflexible motor behavior) that is less adaptable to task/environmental demands and load (Stergiou and Decker, 2011). In turn, this may contribute to an increased risk of injury or exacerbate pain in CLBP patients due increased trunk bending moment, erector spinae fatigue (Mehta et al., 2014) and initiation of intervertebral disc degeneration cycle (Adams et al., 2000; Vergroesen et al., 2015). There are lifting strategies that have been demonstrated to attenuate spinal load – such as, ‘load splitting’ (Faber et al., 2009; Kingma et al., 2016). This may be achieved by i) distributing the load on each side of the body rather than lifting a single load anterior to the body (reduction of L5/S1 compression forces by 8-32%) (Faber et al., 2009) or, ii) lifting load with one hand leaning on a thigh (reduction of L5/S1 compression forces up to 28%) (Kingma et al., 2016).

There are several possible reasons why this study may not have found between-group differences in lifting-related ROM and vertical GRF. Firstly, the load lifted in this study was relatively low (i.e., 8 kg). Burgess-Limerick et al. (1995) demonstrated increased asynchronous movement between the trunk and hip with increasing loads (i.e., up to 12.5 kg) in healthy participants.
Therefore, it is likely that CLBP individuals may demonstrate different lifting patterns to those that are healthy with heavier loads. Secondly, this study assessed CLBP participants in a non-fatigued state. Lumbar extensor muscle fatigue has been demonstrated to increase lifting-related synchronous movement between the trunk and lower limb joints in healthy people – a compensatory response thought to decrease the risk of injury (Hu and Ning, 2015). As increased lumbar extensor fatigability has been reported in people with CLBP (Kankaanpaa et al., 1998), testing CLBP patients in a fatigued state could contribute further to our understanding of changes in trunk-lower limb lifting coordination. This study involved analysis only in a single plane (i.e., sagittal) during a symmetrical lifting task. Impairment in lumbo-pelvis ROM in the coronal plane have been reported during walking in CLBP people compared to healthy controls (Crosbie et al., 2013). Similarly, asymmetrical lifting tasks (i.e., lifting with trunk rotation) have been shown to differentiate between people with and without CLBP with respect to kinematic and kinetic variables (Sanchez-Zuriaga et al., 2011). Thus, between-group kinematic and kinetic differences may be augmented when multi-planar tasks are utilized. Additionally, this study only incorporated a single test lift in order to minimize the risk of pain provocation and burden of assessment on the participant. However, a single lift may not have the predictive capacity of multiple lifts partly due to the variations and consistencies within each individual (Burgess-Limerick et al., 1995; Granata et al., 1999). Lastly, no electromyography (EMG) was used in this study. Recent studies suggest that although participants with CLBP are able to complete lifting tasks as well as healthy controls, there were significant differences in EMG activation and loading patterns (Nelson-Wong et al., 2012). Hence, utilisation of EMG may explain the lack of difference in kinematic variables between the groups (Ferguson et al., 2004). As lifting inter-joint coordination is associated with disability in people with CLBP, lifting re-education may
therefore decrease disability. Interventions that target lumbar-hip coordination and CLBP disability should therefore be the focus for future studies.

The findings of this study suggest that higher-disability CLBP patients demonstrate impaired trunk and lower limb joint coordination during lifting compared to lower-disability CLBP patients and healthy individuals. Moreover, impaired lumbar-hip coordination during lifting is associated with increased disability in people with CLBP. Thus, people with CLBP should consider compensatory lifting strategies that decrease spinal load and, in turn, likely decrease self-reported disability.

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Conflict of interest:

ALB and RC are supported by a National Health and Medical Research Council R.D. Wright Biomedical Fellowships (#1053521 and #1090415, respectively). This played no role in any aspect of the study.

References


Figure 1.

Figure 2.
Figure 3.

The relationship between lumbar-hip movement coordination during lifting and disability

Figure 4.
**Table 1.** Descriptive data (mean ± SD) pertaining to participant characteristics of CLBP and control groups.

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

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

**Table 2.** Range of motion and angular velocity (mean ± SD) of individuals with CLBP with low and high level of disability and control group.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP\text{low}</th>
<th>CLBP\text{high}</th>
<th>Control</th>
<th>( F_{2,68} )</th>
<th>( p ) values*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lumbar ROM (°)</td>
<td>35.05 ± 12.01</td>
<td>33.37 ± 10.72</td>
<td>36.17 ± 13.48</td>
<td>0.04</td>
<td>0.15</td>
</tr>
<tr>
<td>Hip ROM (°)</td>
<td>87.06 ± 21.23</td>
<td>87.90 ± 6.23</td>
<td>83.77 ± 21.26</td>
<td>0.21</td>
<td>0.81</td>
</tr>
<tr>
<td>Knee ROM (°)</td>
<td>40.00 ± 27.80</td>
<td>50.17 ± 21.23</td>
<td>41.76 ± 28.08</td>
<td>0.57</td>
<td>0.57</td>
</tr>
<tr>
<td>Lumbar Vel (°/s)</td>
<td>71.72 ± 27.29</td>
<td>59.21 ± 20.42</td>
<td>77.90 ± 44.11</td>
<td>0.83</td>
<td>0.44</td>
</tr>
<tr>
<td>Hip Vel (°/s)</td>
<td>168.21 ± 62.90</td>
<td>133.59 ± 38.70</td>
<td>177.01 ± 57.33</td>
<td>0.39</td>
<td>0.15</td>
</tr>
<tr>
<td>Knee Vel (°/s)</td>
<td>89.67 ± 50.97</td>
<td>85.25 ± 30.58</td>
<td>101.24 ± 60.53</td>
<td>1.94</td>
<td>0.68</td>
</tr>
</tbody>
</table>

ROM = range of motion, Vel = angular velocity, SD = standard deviation, *\( p \) values for group main effects.
Table 3. Relative phase angle variables (mean ± SD) of individuals with CLBP with low and high level of disability and control group.

<table>
<thead>
<tr>
<th>Variables (units)</th>
<th>CLBP&lt;sub&gt;low&lt;/sub&gt; Mean ± SD</th>
<th>CLBP&lt;sub&gt;high&lt;/sub&gt; Mean ± SD</th>
<th>Control Mean ± SD</th>
<th>F&lt;sub&gt;2,68&lt;/sub&gt;</th>
<th>p values*</th>
</tr>
</thead>
<tbody>
<tr>
<td>MARP lumbar-hip (°/frame)</td>
<td>23.72 ± 3.21</td>
<td>36.25 ± 3.88</td>
<td>27.91 ± 3.05</td>
<td>3.35</td>
<td>0.041*</td>
</tr>
<tr>
<td>MARP hip-knee (°/frame)</td>
<td>23.97 ± 5.90</td>
<td>11.44 ± 7.17</td>
<td>30.29 ± 5.60</td>
<td>2.05</td>
<td>0.14</td>
</tr>
<tr>
<td>DP lumbar-hip (°/frame)</td>
<td>0.06 ± 0.01</td>
<td>0.09 ± 0.01</td>
<td>0.09 ± 0.01</td>
<td>2.09</td>
<td>0.13</td>
</tr>
<tr>
<td>DP hip-knee (°/frame)</td>
<td>0.07 ± 0.03</td>
<td>0.03 ± 0.03</td>
<td>0.14 ± 0.03</td>
<td>3.90</td>
<td>0.026**</td>
</tr>
</tbody>
</table>

SD = standard deviation, s = seconds, °/frame = degrees per frame of capture (1 frame = 1/120 s), MARP = mean absolute relative phase angle, DP = deviation phase, *p values for group main effects, #significant differences between CLBP<sub>low</sub> and CLBP<sub>high</sub> groups, ##significant differences between CLBP<sub>high</sub> and control groups.

Table 4. Descriptive data (mean ± SD) pertaining to kinetic variables derived from CLBP and control groups together with results of statistical analyses.

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

SD = standard deviation, L = Left, R = Right, F = vertical force, Max = maximum, Min = minimum. Kinetic variables are percentage of body weight.
Figure 1. The marker configuration used in this study. Three markers are used to create a rigid body: the thorax (green), pelvis (blue), thighs (purple) and shanks (orange). The T7 markers are 5 cm distal of their respective spinous processes. Dashed lines represent the defined axes of rotation, with virtual markers created based on existing marker locations where the axis did not originate or terminate at a captured marker. Post processing was performed to negate data where needed to align the direction of rotation between limbs. SP = spinous process.

Figure 2. Participant performing symmetrical lifting task and retroreflective marker configurations. Three markers per body segment (thorax, lumbar, pelvis, thigh and shank). 1: starting position; relaxed standing with arms crossed in front of the chest and hands touching the shoulder. 2: bending and lifting phase; participants were instructed to lift an 8-kg kettlebell as naturally as they could. 3: finished posture; the kettlebell was held in front of the abdomen with the elbows held high to prevent disruption to ilium markers/data acquisition.

Figure 3. Lumbar angular velocity (ω) as a function of lumbar angular position (θ) during bending (flexion) and lifting (extension). The phase angle, φ, at any point of flexion and extension can be calculated using the formula $\tan^{-1} (\omega/\theta)$. Note: only the extension phase was analysed in this study.

Figure 4. The linear relationship between Oswestry Disability Index (ODI) and lumbar-hip lifting coordination in people with CLBP ($R^2 = 9.2\%$ and $\beta = 0.30$).