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DOCLINE: Journal Copy EFTS Participant

Title:	Journal of biomechanics
Title Abbrev:	J Biomech
Citation:	2018 Mar 21;70():255-261. doi: 10.1016/j.jbiomech.2017.06.032. Epub 2017 Jul 5
Article:	Mechanical demands on the lower back in patients with non-ch
Author:	Shojaei I;Salt EG;Hooker Q;Bazrgari B
NLM Unique ID:	0157375 Verify: PubMed
PubMed UI:	28712541
ISSN:	0021-9290 (Print) 1873-2380 (Electronic)
Fill from:	**Any format**
Publisher:	Elsevier Science, New York ;|Oxford :
Copyright:	Copyright Compliance Guidelines
Authorization:	kir1
Need By:	N/A
Maximum Cost:	**$12.00**
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Journal of Biomechanics 70 (2018) 255–261

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Mechanical demands on the lower back in patients with non-chronic low back pain during a symmetric lowering and lifting task

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ARTICLE INFO

Article history:
Accepted 20 June 2017

Keywords:
Non-chronic low back pain
Lowering and lifting task
Lower back mechanical demand

ABSTRACT

There is limited information in the literature related to the lower back loading in patients with LBP, particularly those with non-chronic LBP. Toward addressing such a research gap, a case-control study was conducted to explore the differences in lower back mechanical loads between a group of females (n = 19) with non-chronic, non-specific LBP and a group of asymptomatic females (n = 19). The differences in lower back mechanical loads were determined when participants completed one symmetric lowering and lifting of a 4.5 kg load at their preferred cadence. The axial, shearing, and moment components of task demand at the time of peak moment component as well as measures of peak trunk kinematics were analyzed. Patient vs. asymptomatic group performed the task with smaller peak thoracic rotation and peak lumbar flexion. While no differences in the moment component of task demand on the lower back between the patients and controls were found, the shearing (40–50 age group) and axial components of task demand were, respectively, larger and smaller in patients vs. controls. Whether alterations in lower back loads in patients with non-chronic LBP are in response to pain or preceded the pain, the long-term exposure to abnormal lower back mechanics may adversely affect spinal structure and increase the likelihood of further injury or pain. Therefore, the underlying reason(s) as well as the potential consequence(s) of such altered lower back mechanics in patients with non-chronic LBP should to be further investigated.

1. Introduction

Low back pain (LBP) is a leading cause of disability with substantial direct and indirect cost (Balagué et al., 2012; Driscoll et al., 2014; Hoy et al., 2010; Maher et al., 2017). Complexity and multidimensional nature of LBP's risk factors pose a significant challenge for risk management strategies aimed at minimizing the level of exposure. Knowledge of the underlying mechanism(s) responsible for the development and/or persistence of LBP may open new avenues for managing this problem, via interventions that specifically target the underlying malfunctioning mechanism(s) rather than simply reducing generic risk factor exposures. Mechanical loads, specifically forces and deformations, in the lower back tissues can instantaneously or cumulatively exceed the tissues' injury/pain threshold and directly or indirectly lead to LBP (Adams, 2004; Adams et al., 2013; Coenen et al., 2014; Van Dieën et al., 1999). Therefore, a further understanding of this construct in patients with LBP could provide important insights into this health condition.

Mechanical loads experienced in the lower back tissues are directly related to mechanical equilibrium and stability of the lumbar spine (Arjmand et al., 2009; Kingma et al., 2007). Spine equilibrium requires that forces in the lower back tissues, at a minimum level, to balance the mechanical demand of the task (i.e., due to body weight, external loads, and inertia forces). Forces in the lower back tissues maybe larger than the minimum required force for equilibrium in response to stability requirement of spine (i.e., the capacity to maintain mechanical equilibrium at presence of perturbation). Therefore, spinal loads are the resultant of two sets of forces that balance each other around the spine: (1) body weight, external loads, and inertia forces (i.e., collectively known as the mechanical demands of the task on the lower back) and (2) the active muscle forces as well as the passive forces in the connective tissues attached to the spine (i.e., collectively known as the internal tissue responses) (Adams et al., 2013; Bazrgari et al., 2008 b; Bazrgari et al., 2009a; Bazrgari et al., 2009 b; Reeves and

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http://dx.doi.org/10.1016/j.jbiomech.2017.06.032
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Cholewicki, 2003). Potential injury mechanisms in the lower back due to mechanical loading have been shown in cadaveric studies (Adams, 2004; Adams et al., 2013). Lower back tissues can be injured due to excessive loads in the lumbar spine including compression force (e.g., vertebral body damage followed by internal disc disruption), bending moment in the sagittal plane (e.g., posterior ligaments and annulus damage), axial twist and shearing force (e.g., facet joints damage), and combined bending moment and compression force (e.g., annulus and nucleus damage) (Adams, 2004; Adams et al., 2013; Harris and Macnab, 1954; Osti et al., 1990; Roaf, 1960; Van Dieën et al., 1999).

The potential causal mechanism for LBP via excessive mechanical load in lower back tissues (Adams, 2004; Adams et al., 2013; Coenen et al., 2014; Van Dieën et al., 1999) has motivated many research to investigate whether exposure to certain physical factors increases mechanical loads in the lower back. For instance, muscle forces and spinal loads under dynamic lifting tasks (Fathallah et al., 1998; Granata et al., 1997), whole body vibrations (Bazrgari et al., 2008a; Kitazaki and Griffin, 1997; Kong and Goel, 2003), sudden forward perturbations (Bazrgari et al., 2009a; Shahvarpour et al., 2015), and sudden release loading (Bazrgari et al., 2009b) have been estimated for asymptomatic individuals. Though the level and the type of association between exposure to physical factors and occurrence of LBP has been a source of disagreement in the literature (Adams et al., 2013; Maher et al., 2017; Roffey et al., 2010; Waddell and Burton, 2001; Wai et al., 2010), collectively these studies suggest increase in mechanical loads under exposure to physical factors. Similarly, investigation of spinal loads in patients with LBP may help verifying whether treatments offered for LBP should also improve the lower back biomechanics.

The published research on spinal loads in patients with LBP has mainly focused on persons with chronic condition. For lifting and lowering tasks from the floor to the hip level, Larivière et al. (2002) did not find any difference in peak moment demand and compression forces on the spine in patients with chronic LBP vs. controls. They used link-segment models to estimate mechanical demands of the task on the lower back and polynomial equations to estimate spinal loads (Larivière et al., 2002). Using a two-dimensional link-segment model and a single equivalent extensor muscle, Norman et al. (1998) reported larger peak and mean moments as well as larger compression and shearing forces on the spine of workers with chronic LBP vs. controls during regular work duties on the work site. Marras et al. (Marras et al., 2001) reported larger peak moment and compression as well as larger mean compression and shearing forces on the spine of patients with chronic LBP vs. asymptomatic controls using an EMG-assisted model during lifting tasks in the sagittal plane. Shahvarpour et al. (2016) reported similar muscle forces and spinal loads for patients with chronic LBP and asymptomatic controls using a detailed finite element model of spine during unstable sitting on a wobble chair. Notwithstanding the impact of experimental setup and modeling assumptions on findings of earlier studies, it is plausible to postulate differences in lower back loading between patients with chronic LBP and asymptomatic individuals; differences that are task dependent. To our best knowledge, there are only two studies of lower back loading in patients with non-chronic LBP. Using a link-segment model, Shum et al. (2007, 2010) calculated the lower back moment during trunk forward bending and backward return as well as sit-to-stand and stand-to-sit tasks. The lower back moment was smaller in patients at the end range of trunk forward bending but was larger at smaller bending angles (i.e., 15, 30, and 45 degrees). For sit-to-stand and stand-to-sit activity, the lower back moment was smaller in the main plane of movement (the sagittal plane) but larger in frontal and transverse planes among patients with non-chronic LBP compared to asymptomatic controls. Similar to studies of patients with chronic LBP, differences in lower back loads between patients with non-chronic LBP and asymptomatic individuals appears to be task dependent. The limited number of studies on lower back loading in patient with LBP, particularly those with non-chronic LBP, along with task dependency of change in lower back loading call for further investigation of this important construct in patients with LBP.

The objective of this study was set to investigate differences in mechanical demands of a task involving lowering and lifting a load in the sagittal plane on the lower back between a group of females with non-chronic LBP and a control group of asymptomatic females. Given that for the same two groups of participants, we have observed similar trunk range of rotation but smaller trunk angular acceleration in the patient vs. control group during free trunk forward bending and backward return (Shojaei et al., 2017), we hypothesized that the moment demand on the lower back would be smaller for patients vs. controls. However, since patients adopted a larger pelvic rotation during the free trunk bending and return (Shojaei et al., 2017), we further hypothesized that the shearing and axial components of the task demand will, respectively, be larger and smaller in patients with non-chronic LBP versus controls (Shojaei et al., 2016c).

2. Methods

2.1. Participants

Nineteen females (aged 40–70 years) with health-care provider diagnosed non-specific LBP were included in this case-control study design to complete a set of experimental procedures that had already been used in a baseline study involving asymptomatic individuals between 20 and 70 years old (Shojaei et al., 2016a, 2016b; Vazirian et al., 2016b). Patients were excluded if their LBP had lasted more than 3 months as well as if they had significant cognitive impairment, intention to harm themselves or others, evidence of substance abuse, or did not have access to a telephone (Borson et al., 2000; Brown and Rounds, 1995; Ewing, 1984; Radloff, 1977). Upon completion of data collection from the patient group, the data from female participants in the baseline study who were within the same age range (i.e., 40–70 years old) of the patients in this study were extracted for comparison. Asymptomatic controls were recruited via advertisement and excluded if they had a recent (i.e., during the past year) history of LBP or musculoskeletal disorders (Shojaei et al., 2016a; Vazirian et al., 2016a, 2016b). Independent-samples t-tests indicated no differences in age, stature, body mass, or body mass index (BMI) between the two groups (Table 1). Prior to data collection, all participants completed an informed consent procedure approved by the Medical University of Kentucky Institutional Review Board.

2.2. Experimental procedures

Straps were used to attach wireless Inertial Measurement Units (IMUs; Xsens Technologies, Enschede, Netherlands) superficial to the T10 vertebral process, sacrum (S1), right thigh (superior to lateral femoral epicondyle), and right shank (superior to lateral malleolus) (Shojaei et al., 2016c).[1] IMUs placed at the T10 and the S1 levels were assumed to measure rotations of the thorax and pelvis as rigid bodies, while the difference between these rotations was considered to represent lumbar flexion/extension (Shojaei et al., 2016c) (Fig. 1). During the data collection, participants were instructed to complete one symmetric lowering and lifting task while standing in the center of a force platform (AMTI, Watertown, MA). Participants were asked to lower a 4.5 kg load from an upright posture to their knee height, pause for 5 s at this flexed posture, and then extend back to the initial upright standing posture. No more instruction was provided and the task was performed at the participants preferred cadence. The participants completed the task without practice, but if the proper way of performing the task was violated (for example, target height was not achieved) the task was repeated. The kinematics data tracked by IMUs and ground reaction forces collected from the force platform were sampled at the respective rates of 50 and 1000 Hz. Raw kinematics and kinetics data were low-pass filtered (cutoff frequencies of 6 Hz and 50 Hz, respectively) using a fourth order, bidirectional, Butterworth filter.

[1] IMUs were attached by student researchers. The first author of this manuscript was present in data collection of all participants and particularly assured the consistency of sensors locations between patients and controls.

Table 1
Mean (SD) participant characteristics.

	Patients	Controls	t-value	p-values
Age (years)	58 (9)	56 (9)	0.723	0.474
Stature (cm)	163 (7)	164 (5)	−0.592	0.557
Body mass (kg)	76 (17)	70 (12)	1.553	0.13
BMI	27.5 (4.6)	25.7 (4.1)	1.608	0.117
Level of pain*	3.84 (2.09)	–	–	–
Level of disability*	6.16 (4.54)	–	–	–

* The level of pain is based on the pain intensity construct of Wisconsin Brief Pain Inventory (Daut et al., 1983) and the disability is based on Roland Morris Disability Scale (Stroud et al., 2004).

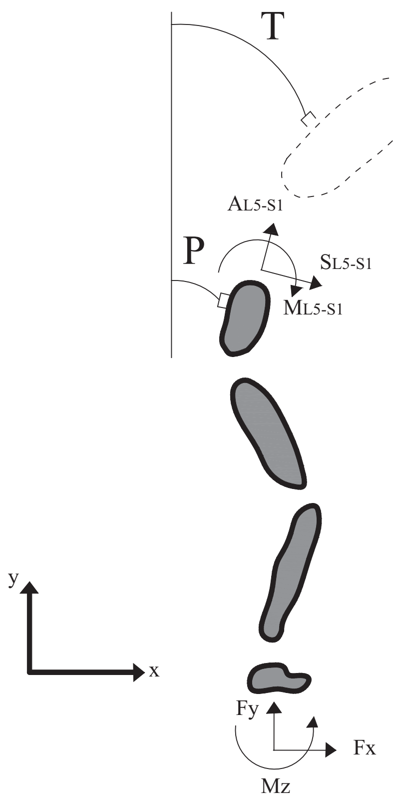

Fig. 1. Lateral view of the linked-segment model. Pelvic (P) and thoracic (T) rotations are shown in the figure and F_x, F_y and M_z denote ground reaction forces. Segments with solid lines were included in the "bottom-up" inverse dynamics approach. A_{L5-S1} (axial), S_{L5-S1} (shearing), and M_{L5-S1} (moment) represent the mechanical demands of task on the lower back.

2.3. Data analysis

A previously developed linked-segment model of the lower extremities and pelvis was used to estimate the net reaction forces and moments at the lower back (Shojaei et al., 2016c). Briefly, the model, developed in MATLAB (The MathWork Inc., Natick, MA, USA, version 8.6), included rigid bodies of seven segments (bilateral feet, shanks, and thighs as well as the pelvis) that were connected using frictionless point-contact joints (Fig. 1).

Using existing regression equations (Winter, 2009), anthropometric and inertial properties of each segment were estimated from participant characteristics (i.e., height and mass). Rotation matrices were then extracted from IMUs to calculate angular rotation of segments, whereas angular velocity and acceleration were obtained using a successive numerical differentiation procedure (Fig. 2).

The mean (SD) accuracy of IMUs (i.e., rotation measure), when used to measure a known rotation in our lab, was found to be 0.55 (0.32) deg and their reliability of repeated measurements (between-day) quantified using intra-class correlation coefficients was excellent (e.g., 1.000). Linear velocity and acceleration were found using the relationship between linear and angular velocity under the assumption that the position of ankle joint did not change throughout the entire task (Shojaei et al., 2016c). Considering the symmetrical nature of the task, equivalent kinematics were assumed for right and left lower extremity limbs. A "bottom-up" inverse dynamics approach (stepwise estimates at the ankle proceeded by knee and hip joints) was used to estimate reaction forces and moments at the lower back which was considered to be the superior level of the pelvis (Freivalds et al., 1984; Song and Qu, 2014) (Fig. 1). Projections of the lower back reaction forces perpendicular (axial) and parallel (shearing) to the L5-S1 intervertebral discs were calculated to represent the contribution of task demand to total axial and shearing forces (i.e., task demand plus the response from internal tissues). The standing orientation of the L5-S1 intervertebral disc, with respect to the gravity direction, was considered to be 50 degrees for 40–50 and 50–60 age groups and 54 degrees for the 60–70 age group (Schwab et al., 2006) for both patient and control groups. The axial and shearing demand as well as the moment demand on the lower back throughout the entire task are shown in Fig. 3 for a typical subject. Estimated forces and moments were normalized to individual body mass and body mass∗stature, respectively. To be able to present the kinetics measures in a more clinically-meaningful sense, the normalized values were multiplied by the mean body mass and mean body mass∗stature across participants (multiplying the measures by a constant value will not affect the results of statistical analyses).

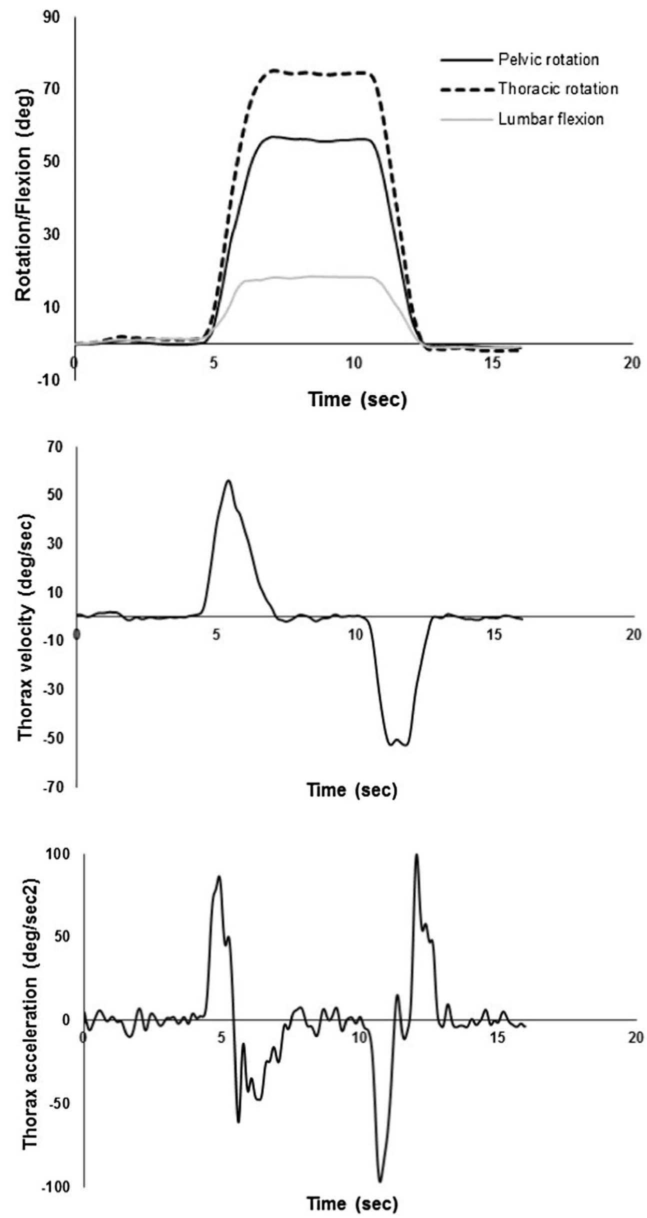

Fig. 2. A typical example of pelvic and thoracic rotations as well as lumbar flexion (top) during the lowering and lifting task. Thorax angular velocity (middle) and acceleration (bottom) were obtained using a successive numerical differentiation procedure.

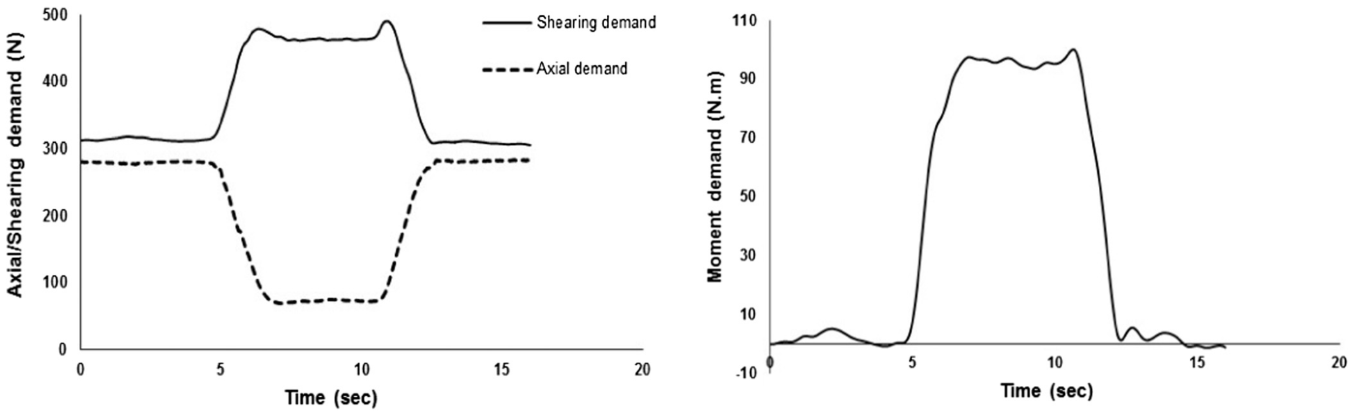

Fig. 3. A typical example of axial and shearing demand (left) and the moment demand (right) on the lower back throughout the entire lowering and lifting task.

2.4. Statistical analysis

The dependent measures included the axial, shearing, and the moment components of task demand as well as several measures of trunk kinematics. Specifically for each phase of task, the values of components of task demand at the *time of peak moment component* (TPMC) as well as the peak pelvic and thoracic rotations along with the corresponding values of lumbar flexion were used for statistical analyses. Mixed-model analysis of variance (ANOVA) tests were conducted on the task demand variables with group (with and without LBP) and age (40–50, 50–60, and 60–70) as the between-subjects factors and task phase (lowering and lifting) as the within-subjects factor. Furthermore, univariate ANOVA tests were used to determine effects of group and age and their interaction on the kinematics variables. Mixed-model and univariate ANOVA assumptions were verified, and significant ANOVA tests were followed by post hoc tests using Tukey's procedure. All statistical analyses were performed using SPSS (IBM SMSS Statistics 23, Armonk, NY, USA), and summary values are reported as means (SD). A p value $\leq$0.05 was considered as statistically significant for all measurements.

3. Results

3.1. Interaction effects

There was a significant interaction effect of group by age on the shearing component of task demand (Table 2). Specifically, for individuals in 40–50 age group the shearing component was larger ($F = 7.85$, $p = 0.026$) in patients (457.9 N ± 23.0 N) vs. controls (384.2 N ± 31.6 N).

3.2. Main effects

3.2.1. Group

There were no differences in the moment component of task demand between patients with non-chronic LBP and asymptomatic controls, whereas the axial component at TPMC was smaller in patients vs. controls (Tables 2 and 3). Moreover, the patient group adopted a smaller peak thoracic rotation as well as a smaller peak lumbar flexion (Tables 2 and 3).

3.2.2. Age

There were no age-related differences in any of the kinetics and kinematics outcome measures (Tables 2 and 3).

3.2.3. Task phase

Larger moment and smaller axial components of task demand at TPMC were observed during lowering vs. lifting phase of the task (Tables 2 and 3).

4. Discussion

The purpose of this study was to investigate differences in the mechanical demands of a lowering and lifting task in the sagittal plane on the lower back between a group of females with non-chronic LBP and a group of asymptomatic females. We did not find any differences in the peak moment component of task demand between the patients and controls, however, the shearing (40–50 age group) and axial components of task demand at TPMC were, respectively, larger and smaller in patients vs. controls. These between group differences rejected our hypothesis on moment demand of task, but confirmed our hypothesis on the shearing and the axial components of task demand.

While several studies have investigated the differences in the mechanical demand of physical tasks on the lower back between patients with chronic LBP and controls, only a few studies investigated such differences between patients with non-chronic LBP and controls (Danneels et al., 2002; Shum et al., 2007, 2010). For a trunk forward bending and backward return task, Shum et al. (2010) reported larger moment demand at smaller flexion angle

Table 2
Summary of statistics results for the effects of group (patients with non-chronic LBP and controls), age (40–50, 50–60, and 60–70), and task phase (lowering and lifting) on the components of task demand as well as trunk kinematics for the lowering and lifting task. TPMC: Time of peak moment component.

	Task Demands at TPMC						Peak Kinematics					
	Moment		*Shearing*		*Axial*		*Thoracic Rotation*		*Pelvic Rotation*		*Lumbar Flexion*	
	F	*p*	*F*	*p*	*F*	*p*	*F*	*p*	*F*	*p*	*F*	*p*
Group (G)	0.06	0.806	6.10	**0.020**	8.10	**0.008**	7.91	**0.009**	3.60	0.068	18.06	**<0.001**
Age (A)	0.21	0.812	1.59	0.222	2.40	0.110	0.10	0.903	2.34	0.114	1.68	0.203
Phase (P)	4.32	**0.047**	3.41	0.076	5.46	**0.027**	–	–	–	–	–	–
G X A	1.48	0.247	3.53	**0.043**	0.11	0.894	2.25	0.124	0.37	0.692	1.14	0.335
G X P	0.75	0.395	0.12	0.737	3.31	0.080	–	–	–	–	–	–
A X P	1.39	0.268	1.58	0.224	2.98	0.068	–	–	–	–	–	–
G X A X P	0.61	0.552	0.10	0.904	0.71	0.501	–	–	–	–	–	–

Boldface indicates significant effect.

Table 3
Summary of outcome measures including mean (SD) for the effects of group (patients with non-chronic LBP and asymptomatic controls) and age (40–50, 50–60, 60–70), and task phase (lowering and lifting) on the components of task demand as well as trunk kinematics for the lowering and lifting task. TPMC: Time of peak moment component.

		Group		Age			Task Phase	
		Patients	*Controls*	*40–50*	*50–60*	*60–70*	*Lowering*	*Lifting*
Task demands at TPMC	*Moment (Nm)*	89.5 (19.0)	89.6 (26.6)	88.2 (24.8)	87.4 (20.0)	93.3 (24.8)	91.8 (20.6)	87.3 (25.0)
	Shearing (N)	446.7 (36.0)	415.5 (47.3)	409.8 (56.7)	449.0 (38.1)	429.3 (31.3)	424.1 (42.9)	438.5 (45.0)
	Axial (N)	74.2 (81.9)	159.1 (80.8)	176.3 (77.9)	96.9 (82.3)	88.7 (89.3)	103.4 (89)	127.1 (91.7)
Peak kinematics	*Thoracic Rotation (°)*	75.2 (10.3)	85.4 (11.3)	81.4 (13.4)	80.9 (7.5)	78.6 (14.7)	–	–
	Pelvic Rotation (°)	42.6 (10.2)	34.0 (11.9)	29.7 (10.1)	42.0 (11.6)	40.9 (10.7)	–	–
	Lumbar Flexion (°)	32.6 (11.0)	51.4 (13.4)	51.6 (16.2)	39.0 (12.4)	37.6 (15.6)	–	–

Table 4
Statistics results as well as outcome measures including mean (SD) for the effects of group (LBP patients or asymptomatic controls) on kinematics characteristics of the lowering and lifting task at the time of peak moment component (TPMC).

	Kinematics at TPMC							
	Thoracic rotation (°)		*Pelvic rotation (°)*		*Lumbar flexion (°)*		*Thorax angular acceleration (°/s^2)*	
	F	*p*	*F*	*p*	*F*	*p*	*F*	*p*
Group	0.14	0.709	8.84	**0.006**	9.45	**0.005**	0.35	0.562
	Mean (SD)		Mean (SD)		Mean (SD)		Mean (SD)	
Patients	68.8(10.6)		38.7 (10.6)		30.0 (9.8)		92.1 (60.5)	
Controls	69.7 (17.6)		26.9 (11.5)		42.9 (13.7)		82.1 (72.0)	

and smaller moment demand at the end range of forward bending between patients with non-chronic LBP and controls. Instead of point-by-point comparison, we compared peak moment demand between the groups which happened to occur at ~85% of trunk end range of flexion in both groups. Considering that the transition from larger to smaller differences in the reported differences in moment demand between patients and controls by Shum et al. (2010) occurred somewhere between the mid and the end range of trunk flexion, our results seem to be consistent with their findings. Danneels et al. (2002) reported similar electromyography (EMG) activity of the multifidus and iliocostalis lumborum pars thoracis in patients with non-chronic LBP and controls during coordination and strength exercises (Danneels et al., 2002). Our finding of similar moment mechanical demands on the lower back, though an indication of comparable total internal tissue responses to the task demand in both groups, doesn't suggest comparable active muscle response. Specifically, the observed smaller lumbar flexion in patients (Tables 2–4) suggests a smaller passive contribution of lower back tissues in offsetting the moment demand of task (Shojaei et al., 2016a), hence an indication of larger active muscle contribution. Participants were instructed to bend forward with a straightened back (i.e., controlled contribution of passive tissues in offsetting the task demand) in Danneels et al. (2002); an instruction that could be the reason for differences between our findings and those of Danneels et al. (2002). It is also notable that unlike the findings on similar EMG activity of the muscles in patients with non-chronic LBP vs. controls (Danneels et al., 2002), Danneels et al. (2002) reported lower EMG activity of the muscles in patient with chronic LBP vs. control.

Our hypothesis on smaller moment demand of task in patients was driven by our findings in an earlier study wherein we observed similar peak thorax rotation but smaller peak angular acceleration during free trunk forward bending and backward return in patients vs. controls (Shojaei et al., 2017). Smaller peak thorax rotation also was observed in patients in this study, hence further supporting our hypothesis on moment demand. However, we did not find any differences in the moment demand between the groups. The reason for such lack of difference was that the thoracic rotation as well as the thorax angular acceleration at TPMC were comparable between patients and controls (Table 4).

Furthermore, our hypothesis on larger shearing and smaller axial components of the task demand in patients with non-chronic LBP versus controls was based on our earlier observation of larger pelvic rotation in patients vs. controls during free trunk forward bending and return. In contrast to free motion, peak pelvic rotation was found to be comparable between the groups (Tables 2 and 3) in this study. Nevertheless, our hypothesis was approved as pelvic rotation at TPMC, where the statistical analyses for the task demands were performed, was larger in patients (Table 4). Additionally, the difference in pelvic rotation between patients and controls was larger (not statistically though) in 40–50 years old age group compared to the other two age groups (i.e., 14.5, 9.2, and 8 degrees in respectively 40–50, 50–60, and 60–70 age groups). Such an age by group difference in pelvic rotation may had a role in the observed differences in shearing demand of the task only in the 40–50 years old age group.

As compared to controls, patients significantly changed their lumbo-pelvic kinematics from the free-style trunk motion to the lowering and lifting task considered in this study. Specifically, patients vs. control adopted a much smaller thorax range of rotation in the lowering and lifting task (i.e., 75.2 vs. 85.4) than in free-style forward bending (104.6 vs. 99.1). Such a reduction in the peak thoracic rotation in patients was achieved by a reduction in the lumbar contribution to the thoracic rotation from 43° to 32.6° (~24% reduction), while the reduction in the lumbar contribution to the thoracic rotation in the control group was from 55.7° to 51.4° (~8% reduction). The significant reduction of the lumbar contribution under the lowering and lifting task may be an overprotective neuromuscular strategy in patients, for instance, to avoid likely overstretching of pain sensitive tissues in the posterior elements of the ligamentous spine.

We found larger moment demand on the lower back under lowering (91.8 Nm) vs. lifting (87.3 Nm) phase of the task that is consistent with the reports on higher occurrence of musculoskeletal injuries (i.e., 67%) during lowering tasks (Lamonde, 1987). However, the literature on differences in mechanical loads on the lower back under lowering vs. lifting tasks is not consistent; there are reports of smaller (De Looze et al., 1993; Larivière et al., 2002), similar (Gagnon and Gagnon, 1992), and larger (Davis et al., 1998) mechanical loads on the lower back under lowering vs. lifting tasks. Such inconsistency in the reported mechanical loads can be due to the differences in task characteristics (e.g., the weight of load carried, lift origin and destination) and the lifting technique (e.g., a standardized lifting technique or motion pace vs. a free-style technique).

Our findings contribute to the current understanding of mechanical demands of a sagittally symmetric lowering and lifting task on the lower back in patients with non-chronic, non-specific LBP, however, there are study limitations. We only recruited female patients, therefore, generalizability of the study findings is limited. We did not asked the participants about their level of pain when performing the tasks, therefore, it remains unclear if and how the observed changes in trunk kinematics and the resultant kinetics were affected by their perception of pain during the experiment. Due to lack of reports on incidence and alignment of pelvis in patients with non-chronic LBP and also inconclusive results from the literature (Hanson et al., 2002; Jackson et al., 2000, 2003; Legaye et al., 1998; Marty et al., 2002) for patients with chronic LBP, same values of sacral orientation were used for both patients and controls when calculating axial and shearing projections of lower back reaction forces. While mechanical demand of physical tasks on the lower back constitutes a small portion of spinal load (i.e., ~20%), it directly influences internal muscle responses that constitute the major portion of spinal loads. Studying muscle response and the resultant spinal loads, however, requires detailed model-based studies (Arjmand et al., 2009; Bazrgari et al., 2008a) as well as electromyography-based measures of the trunk muscles (Callaghan and McGill, 2001).

In summary, we found patients with non-chronic LBP vs. controls adopt distinct trunk kinematics involving less lumbar flexion to perform lifting and lowering task, leading to our observation of differences in the shearing and axial demands of the task on the lower back between the two groups. Although such kinetics differences might have been driven by a neuromuscular effort to minimize lumbar flexion in patients, it directly affects equilibrium and stability of the spine, and hence, the load experienced in the lower back tissues. Regardless of the underlying source of such kinetics differences in patients with LBP, their impact on spine equilibrium and stability and lower back loading should be further investigated. Given the continuity of the spinal column, alterations in mechanical contributions to task demand in one area/component should be compensated by another area/component. The likelihood of further injury and/or structural changes in the lower back tissues that can lead to persistence of LBP increases if the tissue(s) offering compensatory mechanical contributions are not evolved for such response. Furthering knowledge of these biomechanical differences can positively impact the efficiency of present management paradigm for LBP and can help better match patient pathology with target treatments with the long-term goal of avoiding LBP recurrence and/or progression from a non-chronic to a chronic stage.

Conflict of interest statement

We declare that all authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented therein.

Acknowledgements

This work was supported in part by the National Center for Research Resources – United States and the National Center for Advancing Translational Sciences – United States [UL1TR000117] as well as an award (R21OH010195) from the Centers for Disease Control and Prevention – United States. The content of this manuscript is solely the responsibility of the authors and does not necessarily represent the official views of the National Institute of Health and the Centers for Disease Control and Prevention – United States.

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