Estimation of Lumbar Spinal Loading and Trunk Muscle Forces during Asymmetric Lifting Tasks: Application of Whole-body Musculoskeletal Modelling in OpenSim

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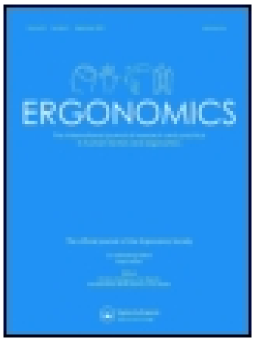

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Estimation of lumbar spinal loading and trunk muscle forces during asymmetric lifting tasks: application of whole-body musculoskeletal modelling in OpenSim

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ABSTRACT
Large spinal compressive force combined with axial torsional shear force during asymmetric lifting tasks is highly associated with lower back injury (LBI). The aim of this study was to estimate lumbar spinal loading and muscle forces during symmetric lifting (SL) and asymmetric lifting (AL) tasks using a whole-body musculoskeletal modelling approach. Thirteen healthy males lifted loads of 7 and 12 kg under two lifting conditions (SL and AL). Kinematic data and ground reaction force data were collected and then processed by a whole-body musculoskeletal model. The results show AL produced a significantly higher peak lateral shear force as well as greater peak force of psoas major, quadratus lumborum, multifidus, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum and external oblique than SL. The greater lateral shear forces combined with higher muscle force and asymmetrical muscle contractions may have the biomechanical mechanism responsible for the increased risk of LBI during AL.

Practitioner Summary: Estimating lumbar spinal loading and muscle forces during free-dynamic asymmetric lifting tasks with a whole-body musculoskeletal modelling in OpenSim is the core value of this research. The results show that certain muscle groups are fundamentally responsible for asymmetric movement, thereby producing high lumbar spinal loading and muscle forces, which may increase risks of LBI during asymmetric lifting tasks.

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Biomechanics; injury risks; manual handling; musculoskeletal disorders; back pain

1. Introduction

Lower back injury (LBI) is the most common cause of pain and disability in many workplaces and it leads to high costs associated with clinical diagnosis, treatment and lost work days (Murphy and Volinn 1999; Punnett et al. 2005). In the United States, approximately two-thirds of the adult population has experienced an LBI in their lifetime (Lawrence et al. 1998) and the annual health care cost on the LBI exceeded $100 billion (Katz 2006).

Large disc compression has been viewed as a risk factor and a good predictor of LBI (National Institute for Occupational Safety, Health. Division of Biomedica and Behavioral Science 1981; Norman et al. 1998). However, occupational LBI is unlikely to be caused by static compression alone, especially during lifting involving a twisting movement. An early epidemiologic study conducted in the United States (Kelsey et al. 1984) indicated that lifting while in a twisting posture was highly associated with the prevalence of LBI because of its high spinal shear force. This asymmetrical lifting posture also requires a high level of muscle contraction (Granata and Marras 1995b). Although increased muscle force is necessary to provide spinal stability by adding stiffness during twist lifting (Cholewicki, Panjabi, and Khachatryan 1997; Gardner-Morse and Stokes 1998), it may also contribute to an increase in lumbar spinal loading (Granata and Marras 1995b, 2000). Crisco et al. (1992) also reported that an inappropriate amount and timing of deep muscle recruitment might cause spinal instability and spinal column failure subsequently. Therefore, biomechanical analysis of lumbar spinal loading along with determination of the trunk muscle forces may provide insights into LBI.

Over the past few decades, many biomechanical models have been developed to estimate lumbar spinal loading during diverse lifting activities. Some studies employed static models for estimating the L4/L5 and L5/S1 compression force (Arjmand and Shirazi-Adl 2005, 2006; Arjmand, Shirazi-Adl, and Bazrgari 2006; Arjmand et al. 2015; Bruno, Bouxsein, and Anderson 2015; Chaffin 1969; Gagnon et al. 2011; Hajihosseinali, Arjmand, and Shirazi-Adl 2015; Morris, Lucas, and Bresler 1961; Rajaee et al. 2015; Schultz et al. 1982), whereas others used dynamic models such as electromyography (EMG)-assisted spine models (Faber et al. 2009a; Ferguson et al. 2002; Jia, Kim, and Nussbaum

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2011; Marras, Jorgensen, and Davis 2000; McGill, Marshall, and Andersen 2013), three-dimensional finite element analysis models (Bazrgari and Shirazi-Adl 2007; Bazrgari, Shirazi-Adl, and Arjmand 2007; Schmidt et al. 2007), dynamic linked segment models (Kingma et al. 2004) and three-dimensional geometric torso models (Nussbaum and Chaffin 1996). Musculoskeletal models allow for carrying out advanced biomechanical analyses relatively easily, determining the whole-body kinematics and kinetics, and simulating the musculoskeletal forces, which may be difficult to measure using traditional experimental methods, especially during dynamic movements. Recently, a three-dimensional, multi-segment musculoskeletal whole-body model (Zhu 2015) has been developed for estimating lumbar spinal loading and individual trunk muscle force during free-dynamic lifting tasks based on previous established musculoskeletal models (Christophy et al. 2012; Hamner, Seth, and Delp 2010; Thelen, Anderson, and Delp 2003), consisting of 23 body segments, 258 muscle fascicles, 27 joints and 49 degrees of freedom and allowing movement in the sagittal, coronal and transverse planes.

The aim of this research was to apply a whole-body musculoskeletal modelling approach to estimate lumbar spinal loading at the L5/S1, L4/L5 and L3/L4 joints and the trunk muscle forces during symmetric lifting (SL) and asymmetric lifting (AL) tasks. It was hypothesised that both lifting types and lifting loads would affect lumbar spinal loading and trunk muscle force.

2. Methods

2.1. Participant

Thirteen healthy male participants (age: 22.5 ± 3.8 years; body mass: 73.1 ± 8.7 kg; height: 1.8 ± 0.1 m; BMI 23.8 ± 1.5 kg/m^2) were recruited for the study. Participants who had acute or chronic LBI, relevant surgical history or leg injuries or who engaged in habitual power lifting or heavy strength training were excluded. Body mass index (BMI, kg/m^2) was also used to minimise the differences between participants; those with a BMI outside of a range of 18.5–24.9 were excluded. All participants signed an informed consent form approved by the University of Auckland Human Participants Ethics Committee (reference number: 012110).

2.2. Apparatus

2.2.1. Kinematics and kinetic measurement

An 8-camera Vicon motion analysis system (Oxford Metric, Oxford, UK) was used to collect kinematic data during lifting tasks at a sampling rate of 100 Hz. Two force platforms with a sampling rate of 1,000 Hz (Bertec Corporation, Worthington, Ohio, USA) measured ground reaction force (GRF) and moment.

2.2.2. Marker placement

Using adhesive, double-sided tape, 39 spherical reflective markers (20 mm in diameter) were attached into the upper and low extremities (Figure 1). Eight of the 39 markers were for the static trial and were placed on the lateral femoral epicondyle (RKNE, LKNE) and medial femoral epicondyle of the knee (RKNM, LKNM), and the medial malleolus (RANM, LANM) and lateral malleolus of the ankle (RANK, LANK). The other 31 markers were placed on the muscle belly of the triceps brachii (RTRI, LTRI), lateral epicondyle of the humerus (RELB, LELB), the tip of the acromion at the shoulder joint (RSHO, LSHO), between the styloid process of the radius and ulna (RWRI, LWRI), at the clavicular notch (CLAV), at the sternum (STRN), at the 7th cervical shoulder (C7) and at the 10th thoracic vertebrae (T10). In the pelvis, both the anterior superior iliac spine (RASI, LASI) and the superior aspect of the L5–sacral (SACR) interface were attached. In the lower extremity, four triads, which consisted of three markers each, were positioned on the lateral side of the thigh (RT1, RT2, RT3, LT1, LT2, LT3) and shank segment (RS1, RS2, RS3, LS1, LS2, LS3), and between the first and third metatarsal heads (RTOE, LTOE) and at the posterior aspect of the calcaneus (RHEE, LHEE) at the same height as the metatarsal heads marker.

2.3. Experimental procedures

Four lifting tasks were performed in a within-subject experimental design to investigate the effects of two lifting types (SL and AL) and two lifting loads (7 and 12 kg) on lumbar spinal forces, moments and muscle forces.

The object lifted was a wooden crate with handles. The crate had dimensions of 40 cm × 35 cm × 25 cm (width × depth × height) and was filled with barbells to achieve the desired weight (7 and 12 kg). The selected load weights were within the range of most lifting tasks, according to an assessment of over 1,000 lifting tasks in workplaces in the United States (Dempsey 2003).

The crate was positioned in the mid-sagittal plane. The lifting movement was started from the initial movement of the picking up the crate to the point of lowering the crate on the table. The SL and AL conditions were as follows: (1) SL: the participant lifted the crate from the floor (the handles are 25 cm above the floor) in front (mid-sagittal plane, 0 degree) and placed it onto a 72-cm high table in front, which was approximately the height of person's knuckles when his arms hung vertically in a standing position; (2) AL: the participant lifted the crate from the same place as the SL and placed it onto a table located in the

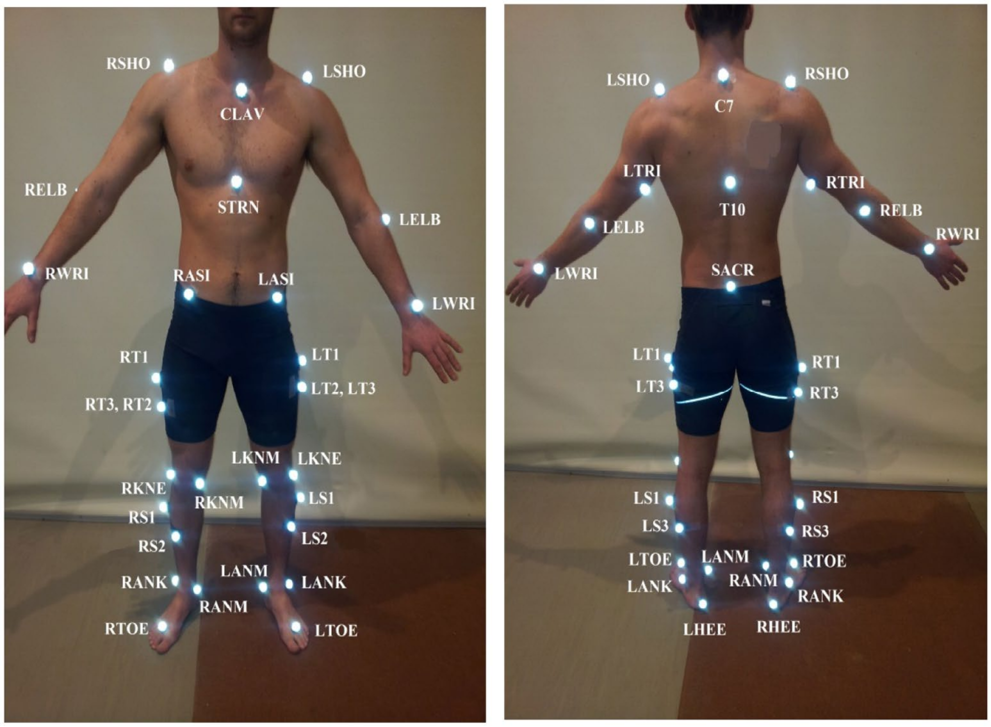

Figure 1. The placement of full marker positions on the participant.

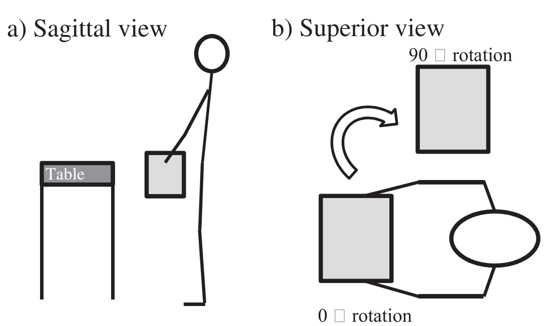

Figure 2. Participants lifted a 7-kg or a 12-kg crate from the floor in mid-sagittal plane and placed it onto a 72-cm high table either in the mid-sagittal plane or in the mid-frontal plane without any movement of their feet. The lifting movements started and finished in an upright standing position.

mid-frontal plane (at approx. 90°) (Figure 2). Participants were instructed to adopt a semi-squat lifting technique (slightly flexed knees and trunk while lifting). Although the squat technique is commonly instructed in workplaces to reduce lumbar spinal loading, it has been documented that losing balance during squatting is more likely to occur and it eventually causes high lumbar loading (Toussaint, Commissaris, and Beek 1997). Participants in the present study, therefore, were asked to pick up the crate while squatting with natural posture of the trunk to prevent losing balance. The initial horizontal distance between the feet and the placement table was selected by the participant, with a maximum distance of 40 cm. The most common distance chosen in workplaces is about 40 cm (Dempsey 2003), and an experimental study indicated that distances of greater than 40 cm may lead to changes in the lifting technique, which can eventually affect lumbar spinal loading (Schipplein et al. 1995). Hence, all participants were asked to choose a comfortable horizontal distance, provided it did not exceed 40 cm. The participants were not allowed to move their feet while lifting because the direction of the pelvis might influence the lateral flexion and torsional torques (Plamondon, Gagnon, and Gravel 1995). In addition, all participants were asked to complete lifting in approximately two seconds (from initial lifting movement to placing the crate onto the table) in order to minimise the time effect on lumbar spinal loading (Dolan et al. 2001). Under all lifting conditions, the time was given verbally by the primary investigator. Five repetitions were performed for each lifting condition (20 lifts in all). The four lifting conditions were in random order, and participants were allowed two minutes' rest in a seated position between different lifting conditions.

2.4. Musculoskeletal model

A whole-body musculoskeletal model (Figure 3) with 23 body segments, 258 muscle fascicles, 27 joints and 49 degrees of freedom (Zhu 2015) was used to simulate lumbar spinal loading and muscle forces. The model involved components for the trunk, upper extremities and lower

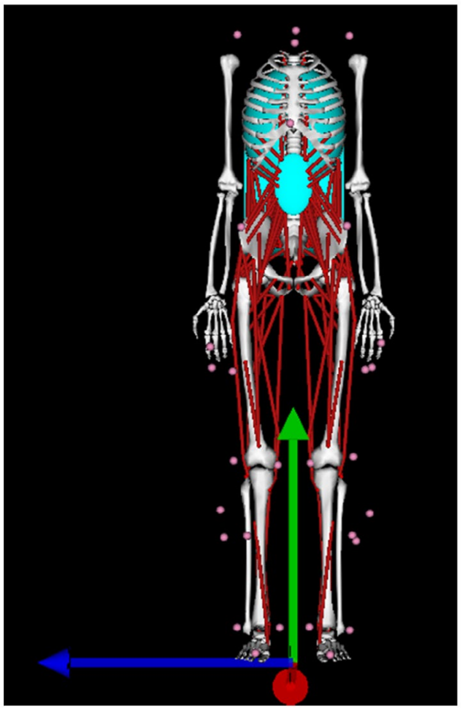

Figure 3. Individualised musculoskeletal model with marker positions.
Notes: The *X*-axis represents the anterior direction. The *Y*-axis represents the superior direction. The *Z*-axis represents the right lateral direction.

extremities. The trunk model consisted of rigid bodies for the five lumbar vertebrae, sacrum, pelvis and torso with bone geometries for the thoracic spine and ribcage, which were modified from a base lumbar model (Christophy et al. 2012). The degrees of freedom for the lumbar were six: flexion and extension, lateral bending, axial rotation and three translational directions. Also, each lumbar vertebral body includes an independent coordinate system and thereby contributes to the overall motion of the lumbar spine. The lower extremities contained rigid bodies of the pelvis, thighs, legs and feet, which were sourced from a Gait2354 model developed in OpenSim software (Stanford University, Stanford, US). The kinematic relationship between the 5th lumbar vertebral body and the sacrum was referred from a study by Anderson and Pandy (1999). The upper extremities contained only skeletal structures which were modified based on the Gait2354 (Hamner, Seth, and Delp 2010). The muscle groups in the trunk were the following: the psoas major, rectus abdominis, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum, quadratus lumborum, multifidus, external oblique and internal oblique. Additionally, seven lower limb muscles were included; gluteus, hamstrings, quadriceps, soleus, gastrocnemius, tibialis posterior and tibialis anterior. The muscles were based on a Hill-type muscle tendon model (Hill 1938).

2.5. Data processing in OpenSim

The marker's data and GRF data were input to OpenSim (www.simtk.org). The processing procedure followed a series of standard simulation steps in OpenSim. The first step was scaling the musculoskeletal model to the anthropometrics of the participant. The scaling tool allowed the creations of a subject-specific model by matching the size of body segments of a particular subject as closely as possible while comparing experimental marker data (from a static trial) to virtual markers placed on the model. After scaling, inverse kinematics was performed to calculate the joint kinematics so that three-dimensional joint angles were produced. In the next step, inverse dynamics was applied to determine joint moments at each joint, which include the L5/S1, L4/L5 and L3/L4 moments. Static optimisation was then conducted as an extension of the inverse kinematics for decomposing the net joint moments into individual muscle forces. Two hundred and fifty-eight muscle forces were quantified under a given cost function, which was a minimisation of the sum of the square of the muscle activations. Finally, joint reaction analysis was conducted to calculate resultant forces that were transferred between joined bodies which derived from loads acting on the model. Lumbar spinal loading was calculated by solving the dynamics equations with the input of muscle forces, gravity and joint moment. Moreover, in order to attenuate the noise contained within the raw marker data, a filtering process was also subjected during static optimisation, using a low-pass sixth-order Butterworth digital filter at a cut-off frequency of 4 Hz, which was determined based on the residual analysis.

2.6. Statistical analysis

Three best-recorded trials were included out of the five trials for each lifting condition for the statistical analysis as in a few trials, reflective markers on the skin were not captured well, which were obscured by the participant's other segments. The average value was obtained from those three lifts. Dependent variables were peak values of 3-D lumbar spinal loading (compression force, anterior shear force and lateral shear force) at the L5/S1, L4/L5 and L3/L4 joints, 3-D moments (extension, lateral bending and axial rotation moment) at the L5/S1, L4/L5 and L3/L4 joints and 3-D trunk angles (flexion, lateral bending and axial rotation). In addition, force values in eight muscle groups in the lumbar spine region were estimated during the lifting movement. A paired *t*-test was performed to compare the difference between the right and left sides of each peak muscle force for all lifting conditions.

A two-way repeated measure analysis of variance (ANOVA) was employed for parametric data to assess the main effects and interaction effects of the lifting loads and lifting types on lumbar spinal forces, lumbar net moments, trunk angles and trunk muscle forces. SPSS Version 20 (IBM Corporation, Armonk, NY, USA) was used for data analysis. All variables have passed normality test using the Kolmogorov–Smirnov and Shapiro–Wilk tests. The significance was set at $p \leq 0.05$.

3. Results

In general, peak compression and anterior shear forces and extension moment at the L5/S1, L4/L5 and L3/L4 joints occurred immediately after lifting-off (Figures 4 and 5). On the other hand, peak lateral shear force, lateral bending moment and axial rotation moment appeared during the lowering of the crate to the table, with the first peak being the greater of the two (Figures 4 and 5).

Increased load (12 kg) resulted in greater values in the spinal forces and moments than a 7-kg load at all joints. However, the effects of lifting types were not always statistically significant on the 3-D lumbar spinal forces or 3-D moments. As shown in Table 1, there was a significant effect of lifting loads on compression and anterior shear forces at all three joints, but the lifting types did not statistically influence them. The lateral shear force only showed significant effect on both lifting types and loads at the L5/S1 and at the L4/L5. Moreover, a significant interaction between lifting types and loads was found in the lateral shear force at the L5/S1. Twisting movement with heavier load also created higher lateral bending and rotation moments (Table 1) and this may influence developing greater lumbar lateral shear force.

For the peaks in the muscle force production (Table 2), the contralateral of psoas major, quadratus lumborum, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum, and bilateral of multifidus and external oblique were significantly higher during AL than SL. All these same muscle groups also produced greater peak force when lifting a 12-kg load than a 7-kg load. However, there was no effect of either lifting types or loads on rectus abdominis and internal oblique.

Additionally, asymmetrical muscle contractions were observed by comparison of the right and left sides of muscles (Figure 6). The contralateral of the psoas major, quadratus lumborum, multifidus, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum and external oblique presented significantly higher peak muscle force than those of ipsilateral during AL. However, the rectus abdominis, multifidus and internal oblique always showed symmetric muscle contractions, regardless of lifting types or loads.

4. Discussion

This research applied a whole-body musculoskeletal model to estimate biomechanical quantities such as joint reaction forces, moments and muscle forces during symmetric lifting and asymmetric lifting tasks. Compared with most previous models, the model presented in this study includes more detailed trunk musculature and spine kinematics than earlier models and can be viewed as an advanced and accessible tool for investigating lumbar spine loading during lifting activities. In particular, compared with the most recent model (Bruno, Bouxsein, and Anderson 2015), the current model has the following differences: (1) this model is a whole-body model with detailed trunk musculature and spine kinematics and more degrees of freedom which are more suitable for conducting a whole-body lifting movement study, while Bruno's model only adopted trunk and upper limb segments. (2) Bruno's model was used to simulate the static and isometric activities, whereas the current model was used to simulate dynamic lifting tasks, which was a more complicated situation. (3) A customised set was developed for the current study, which allows the biomechanical model to be scaled to match the anthropometry of each participant based on measured anatomical marker positions and the participant's body weight, thereby generating subject-specific simulations. In comparison, Bruno's model provides a scaling method based on the height and weight of the participant.

The results of the present study show that AL produced significantly higher peak lateral shear force as well as greater peak force of psoas major, quadratus lumborum, multifidus, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum and external oblique than SL. The magnitudes of 3-D lumbar spinal forces generally agree with results from previously published studies, which showed 3-D lumbar spinal forces at the L5/S1 were influenced by box loads (Davis, Marras, and Waters 1998; Faber et al. 2009a,b) and body postures (Granata and Marras 1995a; Granata, Marras, and Davis 1999; McGill, Norman, and Cholewicki 1996).

Overall, lifting with a twisting movement of the spine increased 3-D lumbar spinal forces and moments at the L5/S1, L4/L5 and L3/L4 joints. The amount of increasing forces and moments was also greater with heavier lifting loads. For the muscle force in the trunk, six out of the eight muscle groups showed significantly higher muscle force in AL than SL, indicating that AL requires more muscle force than SL. More asymmetrical muscle contractions also occurred during AL than SL.

Axial torsion of the spine has been presumed to be a critical LBI risk factor and this is closely related to high compression and shear force at lumbar joints (Kelsey et al.

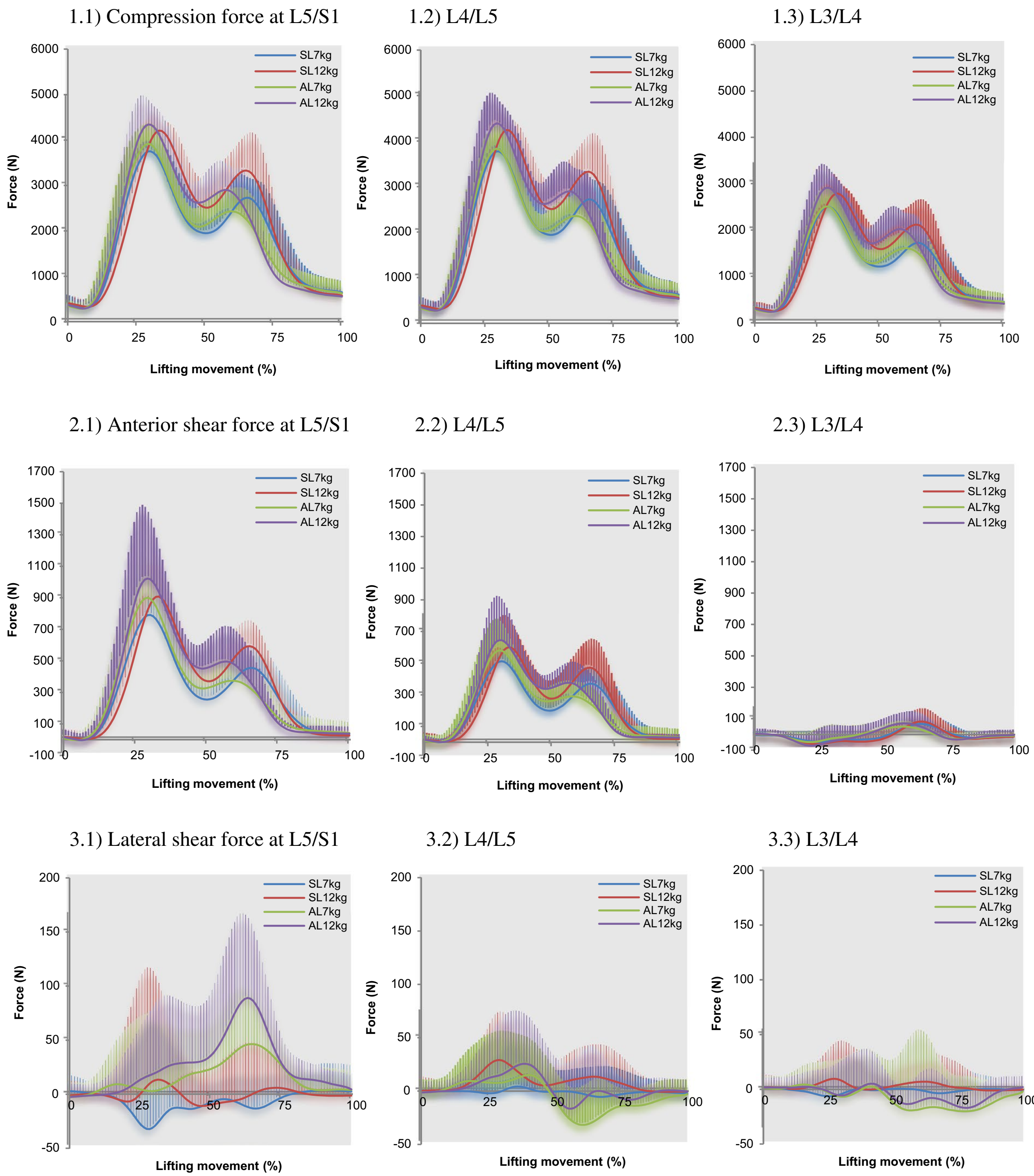

Figure 4. Mean lumbar spinal force (N) pattern after time-normalised at the three joints for (1) compression force (2) anterior shear force and (3) lateral shear force (13 participants and 1 repetition).

Notes: Positive values indicate superior, anterior and right lateral directions. The curve represents lifting movement from the standing position just before initial movement for lifting to completion of lifting.

1984; Schmidt et al. 2007). In the current study, the average of compression force values was higher than the NIOSH Action Limit of 3,400 N of compression on the disc, but below the Maximum Permissible Limit of 6,400 N (National Institute for Occupational Safety, Health. Division of Biomedica and Behavioral Science 1981). However, the anterior shear force at the L5/S1 in the current study exceeds 1,000 N for all lifting conditions. Although the tolerance limits of anterior–posterior shear force for spine are not as well documented as those for compression force, approximately 1,000 N has been recommended as a maximum permissible limit for single exertions (McGill et al. 1998). A recent review determined appropriate limits for anterior–posterior shear exposure (Gallagher and Marras 2012), and it was suggested that a 1,000 N shear limit would be recommended for infrequent shear loading

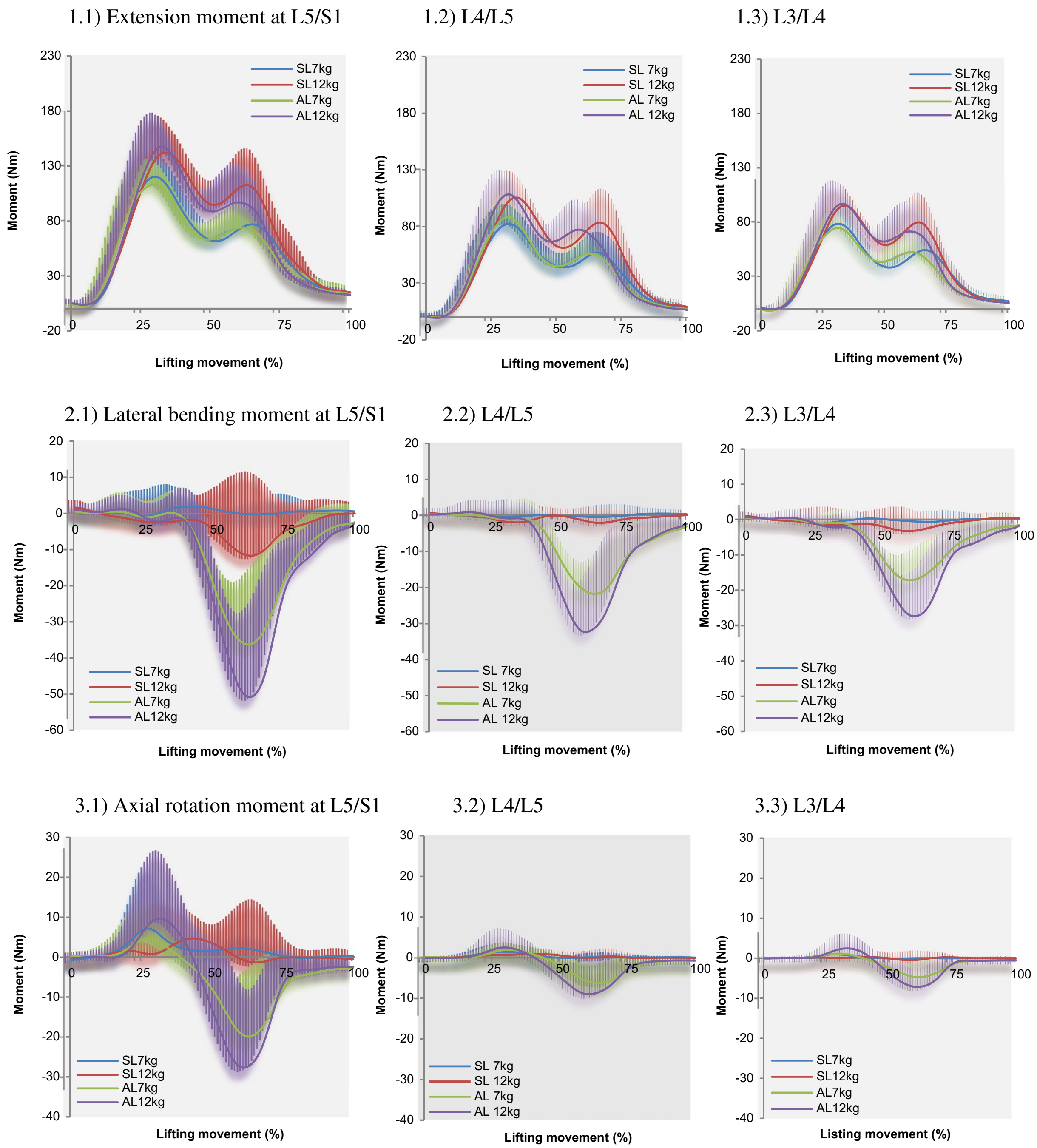

Figure 5. Mean moment pattern (Nm) at the three joints for (1) extension (2) lateral bending and (3) axial rotation moment (13 participants and 1 repetition).

Notes: Positive values indicate extension, right lateral rotation and left axial rotation. The curve represents lifting movement from the standing position just before initial movement for lifting to completion of lifting.

(≤100 loading per day) and 700 N shear limit for frequent loading for up to 1,000 shear loading per day. In this regard, the anterior shear force in the current study may be excessively high for all lifting conditions at the L5/S1. The mean values of L4/L5 showed relatively lower anterior shear force than the L5/S1, but still it went beyond 700 N.

Moreover, the current study showed that the lateral shear force was significantly influenced by both lifting types and lifting loads. The changes in magnitude of lateral shear force within lifting types and loads can be clearly observed in Figure 4. The force patterns fluctuated during the whole lifting movement, but the force significantly increased as load increased and also as AL was performed during the time of placing the crate. Unlike the compression or anterior shear forces, the lateral shear force only reached its peak when the crate was lowered onto

Table 1. Peak mean values of 3-D lumbar spinal forces and 3-D lumbar joint moments at the L5/S1, L5/L4 and L4/L3 joints and lumbar angles.

	Lumbar Spinal Forces (N)									Lumbar Joint Moments (Nm)									Lumbar Angles (degree)		
	Compression force			Anterior shear force			Lateral shear force			Extension			Lateral bending			Axial rotation			Flexion	Lateral bending	Axial rotation
	L5/S1	L4/L5	L3/L4	L5/S1	L4/L5	L3/L4	L5/S1	L4/L5	L3/L4	L5/S1	L4/L5	L3/L4	L5/S1	L4/L5	L3/L4	L5/S1	L4/L5	L3/L4			
SL7 kg	4777.70	4818.68	3265.90	1105.06	722.12	87.49	52.90	43.18	39.30	146.13	104.94	95.59	−7.12	−4.94	−4.32	−0.69	1.01	0.56	−37.63	1.86	−15.21
	(445.99)	(462.30)	(371.48)	(343.79)	(204.36)	(30.05)	(40.99)	(24.28)	(39.08)	(29.41)	(18.25)	(16.33)	(3.94)	(2.23)	(2.20)	(0.79)	(2.92)	(2.14)	(8.34)	(1.70)	(17.16)
SL12 kg	5472.85	5534.08	3722.45	1304.89	849.04	131.08	73.41	60.13	50.00	181.42	126.74	116.46	−19 (25)	−6.06	−5.49	−1.49	0.34	−0.33	−39.33	2.22	−15.50
	(383.91)	(425.70)	(339.63)	(405.57)	(251.76)	(42.08)	(53.36)	(34.90)	(43.15)	(24.14)	(14.11)	(12.86)		(2.29)	(2.17)	(1.81)	(4.30)	(3.27)	(10.26)	(2.75)	(21.72)
AL7 kg	4878.34	4887.34	3274.37	1176.05	758.68	82.26	96.59	49.96	38.64	138.98	101.23	92.00	−51.33	−31.91	−25.57	−3.20	−9.93	−8.11	−39.93	13.75	−113.79
	(462.84)	(408.73)	(323.09)	(319.45)	(192.86)	(23.48)	(65.99)	(24.50)	(34.81)	(24.51)	(18.75)	(15.97)	(20.02)	(9.47)	(7.86)	(1.51)	(4.81)	(3.49)	(11.39)	(6.76)	(39.08)
AL12 kg	5546.89	5619.37	3783.84	1380.62	884.79	105.92	153.41	84.46	51.97	178.95	128.22	118.03	−75.38	−49.25	−39.96	−4.79	−14.64	−12.04	−42.34	15.64	−122.85
	(426.88)	(473.10)	(418.48)	(411.05)	(249.75)	(29.64)	(72.15)	(28.45)	(29.93)	(19.71)	(16.15)	(14.89)	(20.74)	(13.31)	(10.98)	(2.84)	(6.54)	(4.14)	(11.28)	(5.63)	(35.88)
Two-way repeated measure ANOVAs for the effect of lifting types and loads																					
Lifting types (*T*)	.28	.28	.38	.15	.23	.06	**.001**	**.03**	.90	.19	.55	.51	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**	**.02**	**<.001**	**<.001**
Lifting loads (*L*)	**<.001**	**<.001**	**<.001**	**.001**	**.002**	**<.001**	**<.001**	**<.001**	.006	**<.001**	**<.001**	**<.001**	**.001**	**<.001**	**<.001**	**.005**	**<.001**	**<.001**	0.11	.13	.30
$T \times L$	.82	.90	.79	.68	.99	**.02**	**.04**	.12	.72	.50	.24	.21	.23	**<.001**	**<.001**	.31	**.003**	.05	.74	.36	.39

Notes: Two-way repeated measure ANOVA's within-subject results (p-values) are presented. Significant effects ($p < 0.05$) are indicated by bold values. SL: symmetric lifting; AL: asymmetric lifting.

Table 2. Peak mean values of the peak muscle forces of 13 participants in 8 trunk muscle groups.

Muscle groups	PM		QL		MF		IO		RA		IL		LTpL		EO	
	R	L	R	L	R	L	R	L	R	L	R	L	R	L	R	L
Peak values (mean ± SD)																
SL7 kg	4.21 (0.92)	4.34 (1.15)	61.70 (14.44)	48.07 (9.81)	51.09 (15.53)	43.86 (9.80)	34.20 (24.46)	46.47 (22.93)	11.29 (13.79)	10.74 (13.89)	1863.94 (186.96)	1969.55 (169.97)	115.77 (13.73)	122.53 (18.19)	27.26 (30.55)	12.17 (7.99)
SL12 kg	4.13 (0.87)	6.57 (5.32)	67.70 (13.06)	66.03 (21.65)	57.83 (13.93)	52.48 (7.85)	37.99 (22.85)	49.93 (17.84)	7.33 (6.32)	7.61 (6.93)	2228.35 (221.67)	2304.69 (230.93)	134.64 (10.89)	137.61 (15.56)	19.73 (11.89)	28.19 (39.25)
AL7 kg	5.21 (2.64)	46.57 (23.90)	63.50 (8.30)	142.06 (53.42)	54.74 (13.31)	51.40 (12.58)	32.23 (13.67)	35.99 (11.13)	13.74 (13.28)	14.47 (14.54)	1937.72 (158.22)	2138.75 (243.85)	117.22 (11.75)	126.30 (15.56)	31.48 (17.12)	132.34 (60.85)
AL12 kg	6.01 (2.83)	63.90 (30.53)	76.76 (12.77)	184.37 (54.03)	68.80 (20.65)	66.82 (18.37)	37.71 (17.66)	50.73 (19.67)	13.47 (12.23)	14.19 (13.08)	2260.34 (340.19)	2525.52 (263.27)	143.08 (18.59)	154.44 (22.56)	51.42 (29.03)	211.81 (82.91)
Two-way repeated measure ANOVAs for the effect of lifting types and loads																
Lifting types (*T*)	.06	**<.001**	.1	**<.001**	**.02**	**.001**	.85	.42	.15	.12	.30	**<.001**	.10	**.01**	**.003**	**<.001**
Lifting loads (*L*)	.51	**.001**	**.002**	**<.001**	**.001**	**.002**	.24	.06	.40	.50	**<.001**	**<.001**	**<.001**	**<.001**	.13	**<.001**
$T \times L$	2.28	**.003**	.17	.21	.11	.32	.67	.14	.54	.62	.74	.62	.28	.11	**.03**	**.03**

Notes: Two-way repeated measure ANOVA's within-subject results (*p*-values) are presented. Significant effects ($p < 0.05$) are indicated by bold values. R: right, L: left, PM: psoas major, RA: rectus abdominis, IL: iliocostalis lumborum pars lumborum, LTpL: longissimus thoracis pars lumborum, QL: quadratus lumborum, MF: multifidus, EO: external oblique and IO: internal oblique.

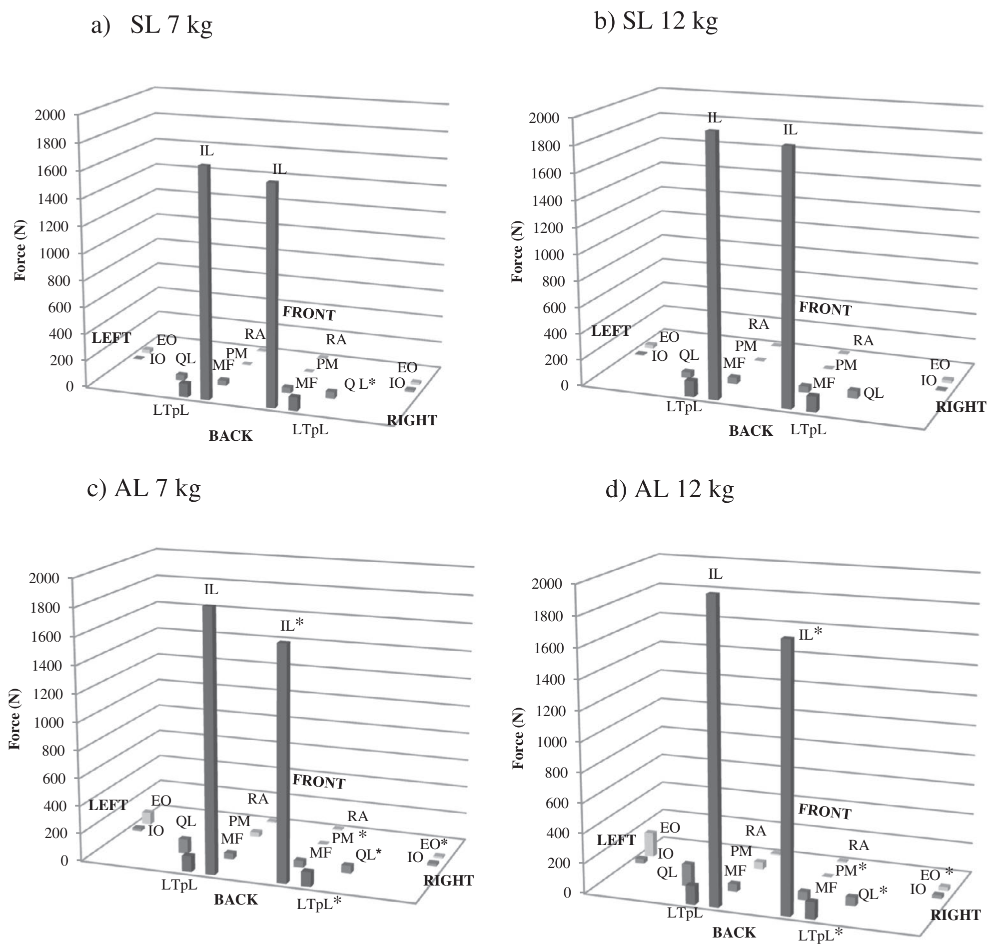

Figure 6. The peak forces in eight trunk muscle groups during (a) SL 7 kg (b) SL 12 kg (c) AL 7 kg and (d) AL 12 kg (Include 13 participants and a single repetition).

Notes: * indicates a significant ($p < 0.05$) difference between the left and right sides of muscles (compared using a paired *t*-test), representing an asymmetrical muscle contraction. PM: psoas major, RA: rectus abdominis, IL: iliocostalis lumborum pars lumborum, LTpL: longissimus thoracis pars lumborum, QL: quadratus lumborum, MF: multifidus, EO: external oblique and IO: internal oblique.

the table with twisting movement, whereas during SL, the value was relatively lower and even hard to define particular force patterns. As only lateral shear force reached a peak when placing the crate down, it may help explain how AL might increase lumbar spinal loading.

In line with a previous study (Marras and Granata 1995), the present study found that torsion of the spine during AL required more trunk muscle contractions compared with SL. A high level of force generation of the trunk muscles may add stiffness and protection of the spine by improving biomechanical spinal stability, defined as maintenance of equilibrium in the presence of small kinematic disturbances or control disturbances (Granata, Lee, and Franklin 2005). In the present study, significantly higher force production of the contralateral psoas major, quadratus lumborum, multifidus, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum and external oblique during AL may play an important role in stabilising the spine or maintaining the asymmetrical lifting posture by generating great muscle forces.

The iliocostalis lumborum pars lumborum, one of the erector muscles, produced the largest muscle force, which ranged from 1,800 to 2,500 N for four lifting conditions, accounting for approximately 32–45% of the total compression force at the L5/S1, while other muscles such as longissimus thoracis pars lumborum did not contribute much in balancing the external moment, producing force ranging from 115 to 155 N. Similar findings were observed from previous research. Gagnon, Larivière, and Loisel (2001) reported that thoracic rector spinae and latissimus dorsi produced approximately 500 and 100 N, respectively, while lumbar erector spinae produced much higher force (approximately 1,000–1,400 N) with a 12-kg external load. Cholewicki and McGill (1994) also reported that when the estimated compression force at the L4/L5 was 5,397–5,303 N, the estimated force of pars lumborum ranged from 68 to 237 N and longissimus thoracis ranged from 57 to 75 N. Danneels et al. (2001) also reported that the iliocostalis lumborum pars lumborum had a high muscle activation level during asymmetric lifting tasks. The greater muscle force from iliocostalis lumborum pars lumborum might also be associated with achieving the equilibrium of the body when an object was lifted, especially in front of the body. When the crate was lifted in front, the centre of mass (CoM) of the upper body could be shifted anteriorly, requiring greater forces through the iliocostalis lumborum pars lumborum to maintain balance of the CoM. The high force production of iliocostalis lumborum pars lumborum in the present study may eventually influence the great lumbar spinal loading. Previous research (Rohlmann et al. 2014) reported that when a backpack was carried, the spinal force was significantly decreased compared with carrying a weight in front of the body. This backpack may relieve an extensor moment of the extensor muscles and thereby lumbar spinal loading was even reduced.

The muscle force of quadratus lumborum was obviously increased during lowering the crate in AL. The contralateral quadratus lumborum force increased significantly, almost reaching 150 N when placing the crate onto the table. The quadratus lumborum is a wide and deep erector spinae, and this muscle acts to move and stabilise the pelvis and lumbar spine (Bogduk 2005). Hence, during axial rotation movement, quadratus lumborum may contribute to the lateral flexion of the trunk so that large forces may be produced in the contralateral quadratus lumborum. These results are also consistent with an early EMG study (McGill, Juker, and Kropf 1996), which observed that quadratus lumborum was heavily active during extension–lifting or standing–twist tasks compared with flexion task (eg curl-up). This confirms the current result that the force production of quadratus lumborum was increased as extension of the trunk after lifting-off with twist of body. Interestingly, unlike other muscle groups, the quadratus lumborum showed asymmetrical muscle contraction during one SL condition (7 kg). When the crate was heavier, the quadratus lumborum revealed symmetrical behaviours. This result may indicate that adequately heavy load may help stabilise the spine. One possible explanation is that the participants were more careful with the heavier load. To maintain body posture, they had greater activation of the quadratus lumborum through the lifting tasks.

Similarly, the contralateral psoas major also showed the peak muscle production only in AL condition during placing the crate. The function of the psoas major was reported to be the lateral flexor of the lumbar spine (Skandalakis 1988) and stabiliser of the lumbar spine (Crisco and Panjabi 1990). A previous study also reported that the psoas major may create a large anterior shear force at the L5/S1, assessed based on the anatomical data attained from cadavers and geometrical scaling data attained from magnetic resonance imaging scans of healthy living subjects (Santaguida and McGill 1995). Thus, the great shear force during AL in the present study may relate to the muscle production of psoas major.

In addition, the external oblique clearly appeared to be involved in AL in the present study. This finding was consistent with an early work by McGill (1991), which reported that myoelectric signal in external oblique was strongly activated during dynamic axial twist activities. The peak force production of external oblique was symmetric in SL condition, but as a consequence of AL condition, the muscle force of the bilateral external oblique significantly increased during twisted motion. Also, external oblique had significant interaction effect between lifting loads and types. This may indicate that the impact of lifting types depends on the lifting loads.

The multifidus also produced significantly higher muscle force during AL than SL. Danneels et al. (2001) observed that the contralateral multifidus showed the greater muscle activity compared with the ipsilateral multifidus during a one-handed lifting task. Moreover, Adams et al. (2007) demonstrated that the orientations of the multifidus may help resist anterior shear. In the present study, the peak force of the multifidus during placing the crate may participate in producing an anterior shear force.

In contrast, the force production of rectus abdominis and internal oblique was not significantly different between lifting types or loads (Table 2). The deep muscles such as internal oblique have been reported for increasing the intra-abdominal pressure and hence provide spine stability (Granata, Lee, and Franklin 2005). Thus, the consistent force production of internal oblique during lifting may act as a critical stabiliser and protector of the lumbar spine. Similar findings were reported from a previous EMG study (Danneels et al. 2001), which investigated muscle recruitment patterns during lifting an object either in two hands or one-hand tasks. They reported that the bilateral internal oblique showed symmetric co-contractions in all lifting conditions; in contrast, the significant differences between right and left were observed in the locomotors including external oblique. In the present study, the rectus abdominis did not show any particular force patterns within all lifting conditions. A previous study also found no significant differences in the rectus abdominis activation between a stable and an unstable trunk in the bench press (Norwood et al. 2007). Thus, in the present study, the rectus abdominis may not be influenced by the instability of spine during AL.

There are several limitations to the current study. First, a small range of lifting loads in this study may not allow observation of significant differences in 3-D lumbar spinal loading between lifting types. Second, the muscle force in the present study was estimated through simulation, which relies on certain assumptions. The model was considered as a single rigid structure and also spinal ligaments and abdominal pressure were not included.

5. Conclusion

This study estimated lumbar loading and muscle forces with a whole-body musculoskeletal model in OpenSim, consisting more detailed trunk musculoskeletal structure, upper extremities and lower extremities. The results show that the muscle groups of psoas major, quadratus lumborum, multifidus, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum and external oblique are fundamentally responsible for the twisting movement of the spine. The greater lateral shear forces combined with higher muscle force and asymmetrical muscle contractions may have the biomechanical mechanism responsible for the increased risk of lower back injury during asymmetric lifting tasks.

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