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The influence of external load configuration on trunk biomechanics and spinal loading during sudden loading

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ABSTRACT

Sudden loading is a major risk factor for work-related lower back injuries among occupations involving manual material handling (MMH). The current study explored the effects of external weight configuration on trunk biomechanics and trunk rotational stiffness in the sagittal plane during sudden loading. Fifteen asymptomatic volunteers experienced sudden loadings using the same magnitude of weight (9 kg) with two different configurations (medially- or laterally-distributed) at three levels of height (low, middle and high). Results of this study showed that the medially distributed weight resulted in a significantly higher peak L5/S1 joint compression force (2861 N vs. 2694 N) and trunk rotational stiffness (2413 Nm/rad vs. 1785 Nm/rad) compared to the laterally distributed weight. It was concluded that when experiencing sudden loading, a more laterally distributed weight could increase the load's resistance to physical perturbations and alleviate spinal loading during sudden loading events.

Practitioner summary: Increased trunk rotational stiffness and peak L5/S1 joint compression force were observed when undergoing a sudden load release of a medially distributed load compared to a laterally distributed load revealing a less stable hand load condition due to the reduced moment of inertia. The laterally distributed load could increase the load's resistance to physical perturbations and mitigate spinal loading during sudden loading events.

Abbreviations: EMG: electromyography; LBP: low back pain; MMH: manual material handling; MOI: moment of inertia

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Low back pain; manual material handling; sudden loading; weight configuration

1. Introduction

Low back pain (LBP) is a leading cause of disability worldwide (Hoy et al. 2014) and is one of the most common reasons of missing work days (Stewart et al. 2003; Ricci et al. 2006). Epidemiological studies have demonstrated that approximately 80% of the population will experience an episode of LBP at some point throughout their lives after reaching adulthood (Rubin 2007). In the United States, the total costs of LBP exceed $100 billion per year; two-thirds of which are indirect costs, due to days-away-from-work and declined productivity (Katz 2006).

Findings of epidemiological and biomechanical studies illustrated that work-related factors account for most of the LBP cases (McGill 1997) and occupations exposed to manual material handling (MMH) tasks lead to even a higher prevalence of LBP compared to those with no MMH involved (Andersson 1981; BLS 2016). Thus, understanding the LBP risk factors that are associated with the performance of MMH is of great importance in reducing the impact of work-related LBP. Earlier research has identified sudden loading as a major risk factor of LBP during the performance of MMH (Omino and Hayashi 1992; McCoy et al. 1997; Mannion, Adams, and Dolan 2000). Sudden loading is an event in which the human body undergoes an abrupt force exertion, and it can be triggered by any destabilising incidents (e.g. slips and falls) or impacts from external objects (Manning, Mitchell, and Blanchfield 1984). Human body responds to sudden loading with reflexive and voluntary muscle contractions to enhance both regional and whole-body stability (Lavender et al. 1989; Cholewicki, Panjabi, and Khachatryan 1997). Existing evidence exhibits that when the trunk encounters an unexpected sudden perturbation, an elevated level of both trunk flexor and extensor muscle activities is often observed

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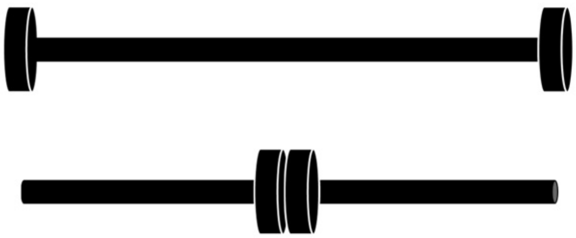

Figure 1. A demonstration of two objects with the same amount of weight but different configurations (the top panel shows a laterally distributed weight configuration and the bottom panel shows a medially distributed weight configuration).

(Thomas et al. 1998); such an increase of trunk muscle co-contraction is believed to stiffen trunk and augment spinal stability (Cholewicki, Panjabi, and Khachatryan 1997; Gardner-Morse and Stokes 1998; Granata and Marras 2000). However, an increased co-contraction of back and abdominal muscles also results in higher spinal loading, and, therefore, elevates the associated risk of LBP (Lavender et al. 1989; Lavender, Marras, and Miller 1993).

Trunk musculature, which is coordinated by the central nervous system, plays a critical role in the stabilisation of the spinal column when experiencing sudden loading (Panjabi 1992; Gardner-Morse, Stokes, and Laible 1995; Cholewicki and McGill 1996). Earlier research showed that an impaired motor control of trunk musculature (e.g. poor postural control, poor sense of lumbar position and longer trunk muscle reaction latency) could lead to spinal instability and result in lower back injuries especially during sudden loading events (Panjabi 1992; Magnusson et al. 1996; Radebold et al. 2000; McGill et al. 2003).

Previous studies have explored a number of contributing factors that could influence spinal stability and trunk biomechanics during sudden loading. Empirical measurements and theoretical analyses suggest that during sudden loading, the development of fatigue in para-spinal muscles would diminish their force-generating capacity and compromise the stability of the spine; thereby, an increased co-activation of trunk muscles would be recruited by neuromuscular control to compensate for declined spinal stability (Granata, Slota, and Wilson 2004). Studies also found that when experiencing sudden loading, grasping the load below the shoulder height resulted in considerably lower level of trunk muscle co-contraction as compared to holding the same load above the shoulder height (Ning et al. 2014), indicating a more stable spine when hand load was held low (Brown and Potvin 2005; Stokes, Gardner-Morse, and Henry 2011). Two other studies investigated the influence of foot placement and uneven ground surface on spine biomechanics during unpredicted and bimanual sudden loading; it was found that maintaining a staggered foot posture (vs. a symmetric stance with the same width) and standing on a flat ground surface (vs. a slanted ground surface) could reduce trunk muscle co-contraction and the corresponding loading to the spine due to an increased postural stability of the whole body (Zhou, Dai, and Ning 2013; Zhou et al. 2015).

A recent study has demonstrated that the weight configuration of a hand load could also change trunk muscle activations during the performance of quiet standing weight holding; holding a laterally distributed load resulted in a significantly lower activation of the trunk muscles due to an increased load moment of inertia in the frontal plane (Madinei 2017; Madinei and Ning 2017). Weight configuration is the distribution or arrangement of the parts of the weight which can alter the moment of inertia of a hand load in the frontal plane. From a mechanical standpoint, when mass is distributed farther away from its centre of rotation, it will result in a higher MOI (Figure 1). According to Newton's second law for rotation, under the same amount of torque, the load with smaller MOI will experience larger angular acceleration and consequently higher trunk stiffness is required to stabilise this load and bring it back to equilibrium (Madinei and Ning 2017). The application of this concept can be found in the Olympic-style weightlifting where weight discs are located laterally on both ends of the barbell. Likewise, acrobats often carry a long pole while performing tightrope walking to improve their postural balance and whole-body stability (McLester and Pierre 2007). However, it is still unclear how different weight configuration of the load (i.e. different moments of inertia) would influence spine biomechanics and stability during sudden loading.

The objective of the current study was to examine the effects of different weight configurations of hand load on spinal loading and trunk rotational stiffness in the sagittal plane during unexpected sudden loading events. Based on the results of previous studies, it was hypothesised that when experiencing sudden loading from the same magnitude of weight, loads with more centrally distributed weight (i.e. with smaller moment of inertia) (Figure 1) will result in an increased stiffness of the trunk musculature and consequently elevate spinal loading.

2. Method

2.1. Participants

Fifteen asymptomatic male volunteers from the student population of West Virginia University with mean

(SD) age, body weight and height of 26.3 (4.9) years, 74.6 (8.5) kg and 176.5 (5.2) cm, respectively, participated in this study. All participants were free from shoulder pain, LBP and upper/lower limb injuries. An informed consent form was obtained from each participant prior to the data collection. The experimental procedure and design for this study were approved by the Institutional Review Board of West Virginia University.

2.2. Experimental design

The weight configuration of the load (referred to as CONFIGURATION) and the load-handling height (referred to as HEIGHT) were both considered as independent variables in this experiment. The CONFIGURATION included two levels: medial (9 kg weight disc located at the midpoint of the barbell) and lateral (two 4.5 kg weight discs each located at the sides of the barbell with 100 cm distance) (Figure 1). The HEIGHT consisted of three levels: low (25 cm below the shoulder height), middle (at the shoulder height) and high (25 cm above the shoulder height). The combination of two CONFIGURATIONs and three levels of HEIGHT generated six different conditions.

The dependent variables investigated were (1) peak L5/S1 joint compression force (the maximum compressive force on L5/S1 intervertebral disc during the sudden loading event); (2) trunk stiffness (K) determined from a second-order biomechanical model described in Section 2.5.2; and (3) the maximum trunk flexion angle (the maximum increase of trunk flexion angle during the sudden loading event with respect to the initial trunk posture).

2.3. Apparatus and equipment

A custom-made wood structure was built and served as a guide for participants to maintain the horizontal distance of the load and adjust their performances according to the specified heights. Weight discs were tightly secured to two identical wooden bars and constituted the two configuration conditions with equal weights of 9 kg (i.e. medial and lateral). Eight bipolar surface electromyography (EMG) electrodes (Bagnoli, Delsys, Inc., Boston, MA) were utilised to record the myoelectric activity from eight trunk muscles at sampling frequency of 1024 Hz. These electrodes were placed over the skin of bilateral erector spinae (4 cm lateral to the L3 spinous process) (Marras and Davis 2001), bilateral multifidus (2 cm lateral to the L5 spinous process) (Ning et al. 2011; Hu, Ning, and Nimbarte 2013), bilateral external oblique (10 cm lateral to the umbilicus and 4 cm above the ilium with the angle of 45°) and bilateral rectus abdominus (2 cm above and 3 cm lateral to the umbilicus) (Marras and Davis 2001). The locations of electrodes are presented in Figure 2(a,b). Trunk and upper extremity kinematics were captured using an eight-camera (MX-13 series) 3D optical motion tracking system (Vicon, Nexus, Oxford, UK) at a sampling frequency of 100 Hz. Eleven reflective markers were placed over the spinous processes of C7, T12 and L5 vertebrae, the most dorsal point on acromioclavicular joint of the left and right shoulders, the most caudal point of the lateral epicondyle of the left and right elbows, the ulnar side of both wrists and the end sides of weight bars (Ning et al. 2014). The EMG and kinematics data were recorded using Nexus 10.7 software (Vicon, Nexus, Oxford, UK). A lumbar dynamometer (Humac Norm, CSMi, MA) with a back flexion module was used to secure participants' pelvis and lower extremities and also to provide a static resistance during the trunk maximum voluntary contraction (MVC) trials.

2.4. Procedure

A complete description of the experimental protocol was provided for participants upon their arrival, and informed consent forms were then signed. After that, participants' anthropometric data, including body height, weight, trunk length (the vertical distance from L5/S1 joint to the top of the head), trunk width (at iliac and xiphoid process levels) and trunk depth (at iliac and xiphoid process levels) were collected. A brief warm-up session was provided to allow participants to stretch their trunk and shoulder muscles and become familiar with the experiment protocol. Eight bipolar surface EMG electrodes were then attached to the skin of the above-mentioned muscles using double-sided tape. At the beginning of the experiment, participants were secured in a 20° trunk forward flexion posture and performed two repetitions of isometric maximum trunk flexion/extension exertions against a static resistance provided by the dynamometer (Marras and Mirka 1993). Each MVC trial lasted for 6 s and 2 min of rest was provided between the exertions to avoid muscle fatigue (Caldwell et al. 1974; Marras and Davis 2001).

When finished with MVC trials, 11 reflective markers were attached to the above-described locations, and participants were required to perform a total of 24 trials (six conditions with four repetitions for each condition) in a completely randomised order. Each trial started with participants closing their eyes, standing with feet shoulder-width apart and gripping the barbell without supporting its weight; the load was then

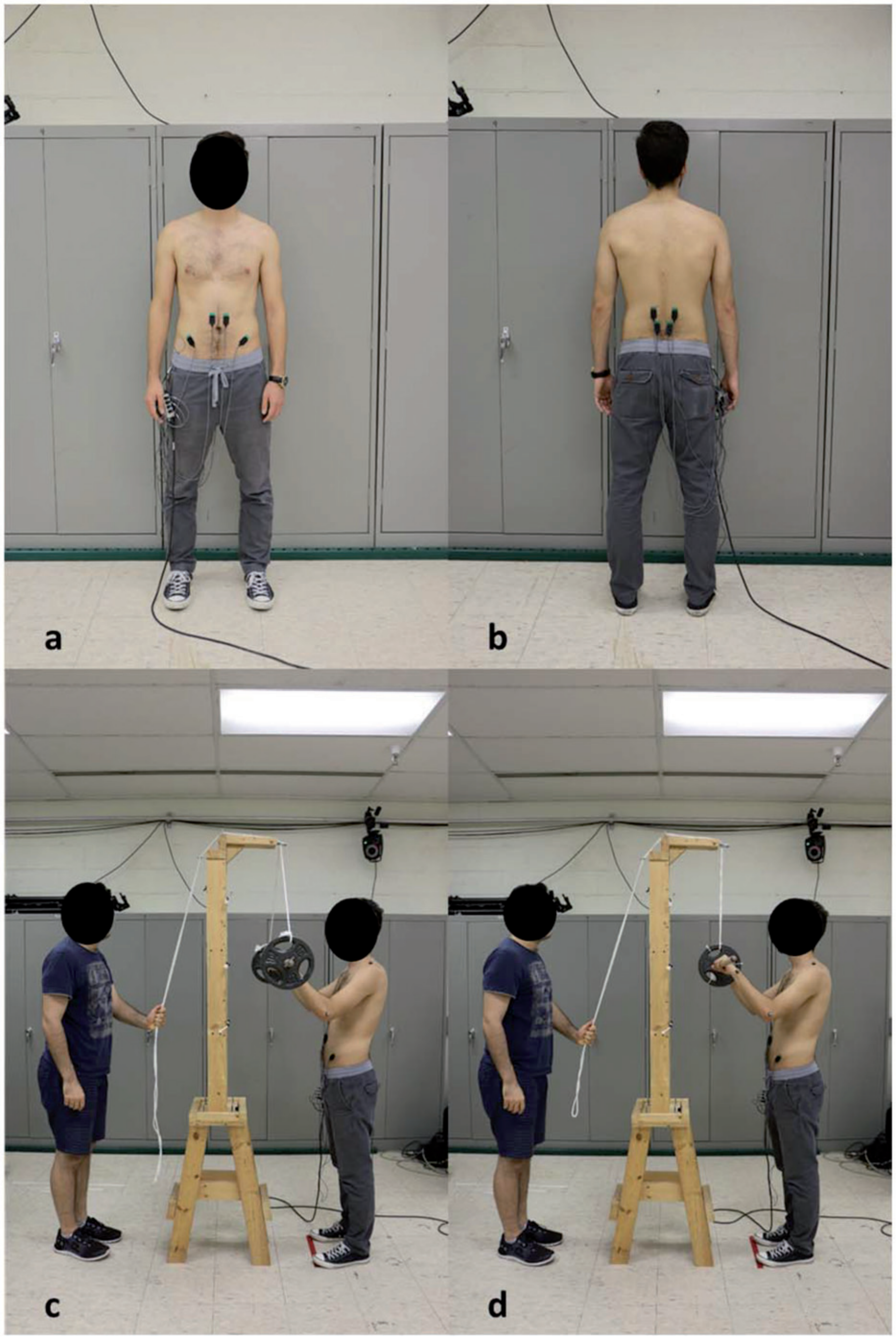

Figure 2. A demonstration of the location of EMG sensors (a, b), side view of the data collection apparatus and the postures participants used when undergoing the sudden load release of a laterally distributed weight at middle height (c) and a medially distributed weight at middle height (d).

suddenly released by the experimenter with no preceding notice (Figure 2(c,d)). Immediately after the load was released but before it landed, participants were instructed to open their eyes, grasp the falling load, bring the load back to its initial position and hold it steady for three seconds. An experimenter then took the load and lowered it to the ground. At least 1 min of rest was provided between the trials to prevent shoulder and back muscle fatigue. To maintain a constant external flexor moment to the L5/S1 joint, participants were instructed to stand straight in front of the wooden structure such that the horizontal distance between the midpoint of their ankles and the projected location of the load's centre of mass remained at 45 cm throughout the whole tasks (Zhou, Dai, and Ning 2013). Throughout the experiment, the locations of hand grip remained consistent such that there was a 20 cm distance between the third metacarpals of both hands, and arms remained in prone condition.

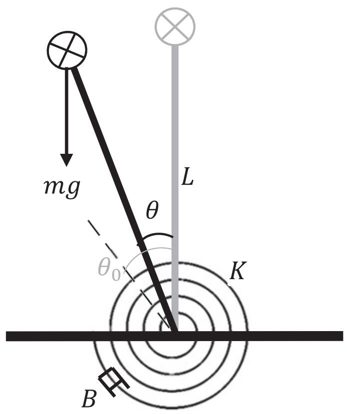

Figure 3. A free body diagram of a second-order trunk model oscillating after the sudden load release. *mg* is the trunk and external weight, *L* is the vertical distance from the L4/L5 joint to the centre of trunk mass supposed to be at the T9 level, *K* is the rotational stiffness of the modeled trunk, *B* is the damping coefficient of the modeled trunk and θ_0 is a hypothetical resting angle of torsion spring.

2.5. Data processing and analysis

EMG signals were filtered using a band pass filter between 10 and 500 Hz and a notch filter at 60 Hz and its aliases. The filtered signals were then full-wave rectified and smoothed using a moving average technique with a 200-data point sliding window. Muscular EMG from all experimental trials wes subsequently normalised to the corresponding maximal EMG values obtained during MVC trials. Following that, trunk and upper extremity kinematics were calculated using extracted 3D coordination data from the 11 reflective markers. Trunk flexion angle was calculated as the angle between trunk segment (the line connecting C7 and L5 reflective markers in the sagittal plane) and the coronal plane (Ning et al. 2011).

2.5.1. EMG-assisted biomechanical model

To estimate the external moment exerted on the L5/S1 intervertebral joint of the spine, a multi-segment dynamic model, consisting of seven body segments (trunk, left and right upper arms, forearms and hands), and the external load, was developed. The estimations of masses and body segments centres of mass were made according to previous work (Pheasant 1986). Muscle forces and the corresponding internal moment and spinal compression force at the L5/S1 joint were calculated using an established EMG-assisted modelling approach (Marras and Granata 1997; Ning, Jin, and Mirka 2012). This model estimated the internal moment at L5/S1 joint using Eq. 1, where 'Gain_i' denotes muscle gain value, 'NEMG_i' is the normalised EMG with respect to the MVC and 'A_i' represents the cross-sectional area of muscle *i*; '$f(l_i)$' and '$f(v_i)$' are the muscle force-length and force-velocity modulation factors of muscle *i* respectively and 'r_i' represents the moment arm vector of muscle *i* (Marras and Granata 1997; Davis, Marras, and Waters 1998). Anthropometric data (e.g. body mass, height, trunk length, width, and depth) obtained from each participant were used to estimate the subject-specific moment arms and cross-sectional areas of the trunk muscles (Jorgensen et al. 2001; Marras et al. 2001). The gain value (maximum muscle stress) was then derived from matching the internal and external moments at the static period of load holding (last 3 s) of all trials.

$$\vec{M} = \sum_{i=1}^{8} \vec{r}_i \times \text{Gain}_i \times \text{NEMG}_i \times A_i \times f(l_i) \times f(v_i) \quad (1)$$

2.5.2. Trunk stiffness model

Trunk rotational stiffness can be estimated based on trunk kinematics in response to a sudden loading event (Hunter and Kearney 1982; Lacquaniti, Licata, and Soechting 1982; Hogan 1990; Tsuji et al. 1995 Winters, Stark, and Seif-Naraghi 1988). In a model developed by Cholewicki, Simons, and Radebold (2000), the trunk was represented as a second-order system with viscoelastic properties, oscillating freely around L4/L5 joint after a sudden load release (Figure 3). The frequency and amplitude of such oscillations recorded promptly after the load release but before the occurrence of voluntary muscle contraction, are determined by the trunk inertia (*I*), damping coefficient (*B*) and stiffness coefficient (*K*) which were established prior to the load release. For small trunk angles (θ), the governing equation of motion for the trunk angular oscillations can be expressed by:

$$I\ddot{\theta} + B\dot{\theta} + K(\theta - \theta_0) = mgL\sin\theta, \quad (2)$$

where *mg* is trunk and external load, *L* is the vertical distance from the centre of trunk mass supposed to be at T9 level to the L4/L5 joint and θ_0 is a hypothetical resting angle of the rotational spring. Subject-specific trunk mass (including head and arms) and moment of inertia were calculated from the anthropometric data (Winter 2005). A curve-fitting algorithm was implemented to acquire the best match between the measured and modelled trunk deflection trajectories and, consequently, determine coefficients *B*, *K* and a constant *C* (encompassing θ_0 and integration constants). The curve-fitting algorithm was merely dependent upon the trunk flexion angle which implies no interference of arm rotation with calculations.

Table 1. The results of MANOVA and univariate ANOVA.

Independent variables	MANOVA	ANOVA		
		Compression force	Trunk stiffness	Max trunk flexion
Configuration	**<0.001**	**<0.001**	**<0.001**	0.191
Height	**<0.001**	**<0.001**	**<0.001**	**<0.001**
Configuration × height	0.479	N/A	N/A	N/A

Bold values indicate significant results.

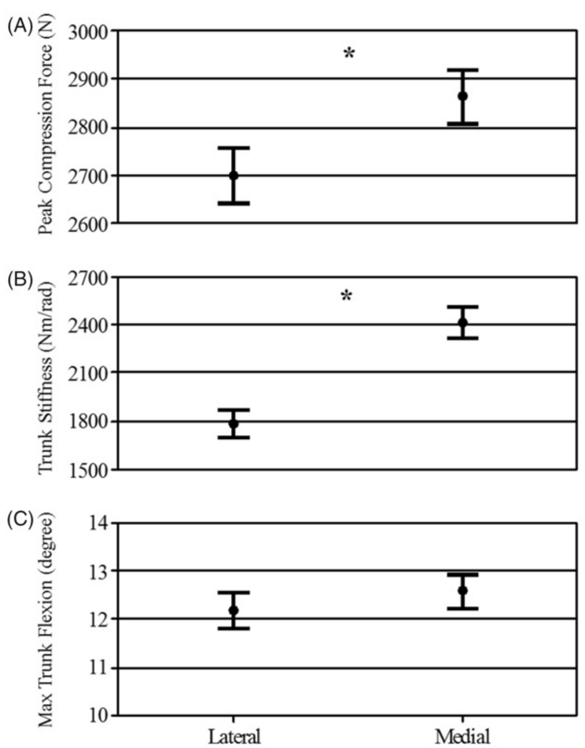

Figure 4. Peak L5/S1 compression force (A), trunk stiffness (B), and maximum trunk flexion angle (C) caused by sudden loading under the two different load configurations (three HEIGHT levels combined). Asterisks denote statistical difference. Bars indicate the corresponding standard error.

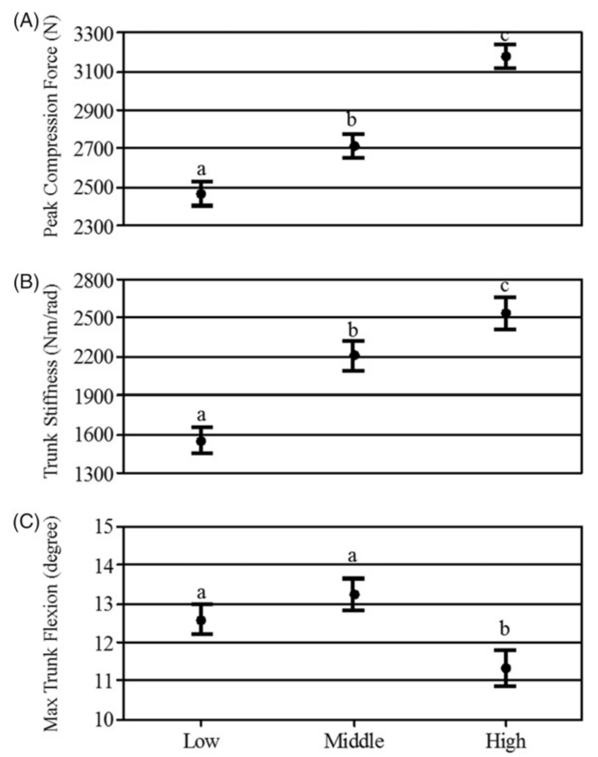

Figure 5. Peak L5/S1 compression force (A), trunk stiffness (B) and maximum trunk flexion angle (C) caused by sudden loading under the three different load handling heights (two CONFIGURATION levels combined). Different lowercases denote conditions that are statistically different from one another. Bars indicate the corresponding standard error.

Further, the point-mass representation of the trunk and upper limbs was supported by the regression analysis which showed no correlation between arms and trunk rotations ($R^2 < 0.1$).

2.6. Statistical analysis

The ANOVA assumptions were first validated among all dependent measures (e.g. normality of residuals and equality of variances), variables that violated the assumptions were transformed using deterministic mathematical functions until all assumptions were satisfied (Montgomery 2012). Multivariate analysis of variance (MANOVA) was subsequently conducted to reveal the statistical significance of the main effects, CONFIGURATION and HEIGHT, as well as their interaction on all dependent variables, collectively. Significant effects were further analysed using repeated measures univariate ANOVA. Tukey–Kramer post-hoc tests were conducted on dependent variables that were significantly affected by HEIGHT to further investigate the differences between each two HEIGHT levels. The level of significance, $\alpha = 0.05$, was chosen for all statistical analyses. All statistical analyses of the current study were performed by Minitab 17 software (Minitab Inc., PA).

3. Results

The results of MANOVA revealed significant main effects of both CONFIGURATION and HEIGHT, although, their interaction effect was not significant and, thus, was not further analysed (Table 1). Results of univariate ANOVA demonstrated that a sudden loading from a medially distributed weight resulted in

Table 2. The mean (SD) values of dependent variables at different HEIGHT and CONFIGURATION conditions.

	Low		Middle		High	
Dependent variables	Lateral	Medial	Lateral	Medial	Lateral	Medial
Peak L5/S1 compression force (*N*)	2348 (596)	2559 (598)	2640 (585)	2767 (593)	3095 (698)	3256 (613)
Trunk stiffness (Nm/rad)	1305 (900)	1799 (1073)	1918 (1089)	2496 (1069)	2133 (1055)	2943 (1232)
Maximum trunk flexion angle (degree)	12.3 (4.3)	13.0 (3.6)	13.1 (4.0)	13.5 (4.4)	11.3 (4.4)	11.4 (4.6)

a significantly higher peak L5/S1 compression force compared to a laterally distributed weight (on average, 2861 N vs. 2694 N) (Figure 4(A)). Significantly higher trunk rotational stiffness in the sagittal plane was observed in the medially distributed weight condition in comparison to the laterally distributed weight condition (on average, 2413 Nm/rad vs. 1785 Nm/rad) (Figure 4(B)). The maximum trunk flexion angle was not significantly affected by CONFIGURATION (Figure 4(C)). In addition, HEIGHT significantly influenced all three dependent variables. A significant rise of peak L5/S1 joint compression force was observed with an increase of load handling height (on average from 2453 N to 3175 N) (Figure 5(A)). Trunk rotational stiffness was also considerably elevated with an increase of load handling height (on average from 1552 Nm/rad to 2538 Nm/rad) (Figure 5(B)). Finally, a significantly smaller maximum trunk flexion angle (11.3°) was observed when the load handling position was set at 'high', while the difference between 'Low' and 'Middle' conditions was not statistically significant (Figure 5(C)). Table 2 provides mean (SD) values of all dependent variables under different CONFIGURATION and HEIGHT conditions.

4. Discussion

The objective of this study was to investigate the influence of different weight configurations of hand load on spinal loading and trunk rotational stiffness during sudden loading events. As described earlier, different weight configurations of a hand load create different rotational moments of inertia. Such differences could result in changes in neuromuscular control and muscle activation patterns to cope with external perturbations (e.g. sudden loading).

We hypothesised that when experiencing a sudden loading from the same magnitude of weight, loads with more centrally distributed weight (i.e. with smaller moment of inertia) will result in elevated trunk muscle co-contraction and spinal loading. Confirming our initial hypotheses, findings of the present study demonstrated an increased spinal compression force and trunk rotational stiffness in the sagittal plane when handling medially distributed weight. When weight is medially distributed, the moment of inertia of the load decreases in the frontal plane, as a result of this reduction, when experiencing equal torques in the frontal plane, it would undergo larger angular acceleration compared to the laterally distributed weight (Madinei and Ning 2017). Thus, the trunk musculature is stiffened through muscle co-contraction to stabilize the load with larger displacements and bring it back to its equilibrium (Granata and Marras 2000). It is to be noted since the loads were harmonically wobbling around the centre of barbell the net moment in the frontal plane was negligible. As a result, no lateral bending was observed during the experiment. Further, the NEMG outcomes demonstrated no significant differences between peak NEMG of left and right trunk muscles which further confirms that the net external moment in the frontal plane was negligible. However, due to the larger displacements of the medially distributed load (i.e. the load with smaller moment of inertia), the proprioceptive system was likely sensing a more unstable load condition; thus, the CNS system increased the activity of trunk muscles to enhance the trunk rotational stiffness to be able to stabilize the load with larger displacements and bring it back to the equilibrium.

The present study estimated the rotational stiffness of trunk in the sagittal plane from its kinematic response to a sudden loading event. The current trunk stiffness model considers the lump-sum effect of both involuntary and voluntary muscle contractions; thus, the estimations of trunk stiffness were described as effective trunk stiffness. Based on the results, significantly higher effective trunk rotational stiffness in the sagittal plane was observed when experiencing medially distributed weight (Figure 4(B)). In the current study, the elevated stiffness of the lumbar spine was achieved by an increase of co-activation of agonistic and antagonistic trunk muscles. Such an increase of trunk muscle co-contraction could also elevate spinal stability (Bergmark 1989; Gardner-Morse, Stokes, and Laible 1995; Cholewicki and McGill 1996; Cholewicki, Panjabi, and Khachatryan 1997). However, this enhancement in trunk stiffness and stability resulted in increased spinal loading, which may elevate the risk of back injury (Lavender, Marras, and Miller 1993; Ning and Mirka 2010; Ning et al. 2014). A previous study suggested that co-contraction of trunk muscles contributes to improved spinal stability and increased spinal compression, yet the stability improvement outweighs the increase of compression loads (Granata and Marras 2000). Moreover, results of the current experiment revealed no significant effect of weight configuration on the maximum trunk flexion angle.

In agreement with previous studies (Granata and Orishimo 2001; Ning et al. 2014), higher trunk rotational stiffness in the sagittal plane and higher spinal compression force were observed when holding a load in a higher position. It has been well-documented that this increase in stiffness of torso musculature is to compensate for the reduced trunk and spinal stability (Granata and Marras 2000; Granata and Orishimo 2001). In addition, the general decreasing trend of the maximum trunk flexion angle at a higher load handling position aligns with previous research (Ning et al. 2014). Brown, Vera-Garcia, and McGill (2006) demonstrated that under the same amount of external flexor moment, peak trunk flexion angle would reduce as the stiffness of the lumbar region increases.

Sudden loading incidents due to loss of control or external impacts on the spine can lead to back injuries (Manning, Mitchell, and Blanchfield 1984; Omino and Hayashi 1992; McCoy et al. 1997). The results of this study suggest that to undergo a sudden load release of the same amount of weight, a symmetrical and more laterally distributed load would increase the load's moment of inertia which in turn elevates the resistance of the load to physical perturbations (i.e. more stable condition). Further, the load with a larger moment of inertia not only reduces spinal compression forces which could mitigate the risk of LBP (Lavender, Marras, and Miller 1993), but also constitutes a more stable load handling condition that might potentially reduce the risk of losing balance during sudden loading events (McCoy et al. 1997).

Finally, a number of limitations of this study need to be noted. First, due to the relatively small trunk flexion during sudden loading, only spinal compression force was evaluated. More comprehensive spinal loading (shear force and torsional force) should be assessed in future studies, especially when experiencing larger trunk posture deflections. Second, shoulder and elbow rotation were not considered separately when assessing the trunk stiffness. Indeed, the interaction between trunk, shoulder and elbow rotation whose mean (SD) flexion values observed in the current experiment were 12.4° (4.3°), 22.8° (13.8°) and 12.1° (8.3°), respectively, warrants further investigation. It needs to be noted that the curve-fitting algorithm employed to calculate trunk stiffness was based on the trunk flexion angle which means arm rotation did not interfere with the calculations. In other words, only trunk flexion angle data, taken from the time of load release to the point of maximum trunk deflection, were used for stiffness calculations, which reflect the independence of the calculations from arm motions. Further, in order to make sure arm rotation did not influence trunk flexion a regression analysis was conducted whose results showed no correlation between arms and trunk rotations ($R^2 < 0.1$). Thus, a point-mass representation of the trunk and upper limbs would be a plausible assumption to make in this study. Third, in order to avoid shoulder and back muscle fatigue, only one level of weight, one specific hand and trunk posture, and symmetric tasks were tested in the current study; heavier weights and more complex hand and body postures as well as asymmetric weight holding may result in larger impact to the trunk which warrant further investigations. Finally, only young, male participants from the student population were recruited in this experiment. The responses of older individuals, female participants and experienced workers need to be studied in the future.

5. Conclusion

Results of the present study provided important information regarding the impact of sudden loading on trunk biomechanical responses when handling loads with different configurations (or distributions). These findings suggest that the weight configuration of the hand load should be considered during the design and assessment of MMH tasks involving sudden loadings. According to our results, it was concluded that when confronted with a sudden loading incident, the load with a larger moment of inertia (i.e. laterally distributed load) could help mitigate the compressive forces on the spine due to an increase in the load's resistance to physical perturbations (i.e. more stable condition).

Disclosure statement

No potential conflict of interest was reported by the authors.

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