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Repetitive lifting is associated with an increased risk of occupational low back disorders, yet potential adverse effects of such exposure on trunk mechanical and neuromuscular behaviours were not well described. Here, 12 participants, gender balanced, completed 40 min of repetitive lifting in all combinations of three flexion angles (33, 66, and 100% of each participant's full flexion angle) and two lift rates (2 and 4 lifts/min). Trunk behaviours were obtained pre- and post-exposure and during recovery using sudden perturbations. Intrinsic trunk stiffness and reflexive responses were compromised after lifting exposures, with larger decreases in stiffness and reflexive force caused by larger flexion angles, which also delayed reflexive responses.Consistent effects of lift rate were not found. Except for reflex delay no measures returned to pre-exposure values after 20 min of recovery. Simultaneous changes in both trunk stiffness and neuromuscular behaviours may impose an increased risk of trunk instability and low back injury.

Practitioner summary An elevated risk of low back disorders is attributed to repetitive lifting. Here, the effects of flexion angle and lift rate on trunk mechanical and neuromuscular behaviours were investigated. Increasing flexion angle had adverse effects on these outcomes, although lift rate had inconsistent effects and recovery time was more than 20 min.

Keywords: low back pain; lifting; trunk flexion; stiffness; reflex; biomechanics

1. Introduction

Low back disorders (LBDs) are among the most prevalent occupational injuries, involving annual costs in excess of $10 billion dollars for treatment in the United States alone (Martin et al. 2009). Although diverse risk factors for LBDs have been identified, performing repetitive lifting tasks involving trunk flexion is associated with a particularly high risk (Kuiper et al. 1999; Hoogendoorn et al. 2000). Epidemiological studies have also provided evidence of increased LBD risk due to repetitive lifting in several occupational sectors, such as automobile industries and parcel delivery (Punnett et al. 1991; Prado-Leon et al. 2005). Other studies have suggested that lifting conditions, specifically trunk flexion angle and lift rate, can influence the risk of LBDs (Stobbe et al. 1988; Dolan et al. 1994; Lin et al. 2002). However, the underlying mechanism (s) linking these lifting conditions and LBD development are still unclear.

Experimental studies have demonstrated changes in mechanical and neuromuscular properties of the trunk as a result of repetitive flexion (which is required typically when performing lifting tasks). The most commonly reported mechanical consequence of repetitive trunk flexion is a reduction in passive stiffness of the trunk (Parkinson et al. 2004; Olson et al. 2009, Shin and D'souza 2010), which is likely subsequent to changes in viscoelastic behaviours of spinal motion segments and passive muscle components. Cyclic flexion leads to creep and load-relaxation of human lumbar motion segments (Little and Khalsa 2005), and cyclic elongation of muscle-tendon units beyond resting length can lead to a reduction of 15% in peak tensile forces after 10 cycles (Magnusson et al. 2000). In addition to these changes in the mechanical properties of the trunk, repetitive flexion can also cause neuromuscular alterations such as muscle spasms and compromised reflex responses of paraspinal muscles (Claude et al. 2003). Such neuromuscular alterations could subsequently impair the stability of the trunk (Panjabi 2003). Moreover, recovery of trunk mechanical and neuromuscular behaviours may differ depending on the specific pattern of flexion exposures. For example, in our previous work, we found that mechanical properties recovered faster than neuromuscular behaviours after prolonged trunk flexion at a constant angle (Hendershot et al. 2011). However, opposing results (i.e. faster recovery of neuromuscular behaviours) were found after prolonged passive trunk flexion (Bazrgari et al. 2011a).

Although this existing evidence suggests mechanical and neuromuscular alterations in trunk behaviours after cyclic loading, it remains to determine whether/how such changes occur in the human trunk while performing an actual lifting task

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and the potential modifying effects of specific task demands. Accordingly, the goal of this study was to evaluate the effects of repetitive lifting on mechanical and neuromuscular behaviours of the trunk. We have reported previously that the stiffness and reflex response of the trunk are more substantially affected after an exposure to larger trunk flexion angles (Hendershot et al. 2011). In addition, changes in mechanical and neuromuscular behaviours in response to repetitive flexion exposures (in a feline model) are frequency dependent (Lu et al. 2008). As such, we hypothesised that (1) the magnitude of alterations in trunk mechanical and neuromuscular behaviours during repetitive lifting increases with trunk flexion angle and lift rate and (2) patterns of recovery are different between mechanical and neuromuscular behaviours.

2. Methods

2.1. Participants

Twelve healthy young adults with no self-reported history of low back pain or current medical conditions completed the study by providing informed consent. All experimental procedures were approved by the Virginia Tech Institutional Review Board. Participants included six males with respective mean (SD) age, stature and body mass of 22 (3) years, 182.1 (3.8) cm and 75.9 (6.3) kg; six females of 24 (3) years, 165.2 (4.4) cm and 59.1 (5.9) kg, respectively. A relatively young group of participants (from 19 to 28 years) was included to avoid potential influences related to age.

2.2. Experimental design and procedures

A repeated-measures design was used, in which several measures of trunk mechanical and neuromuscular behaviours were obtained before, during and after repetitive dynamic lifting. There were six different lifting conditions, involving all combinations of three flexion angles (33, 66 and 100% of each participant's full flexion angle) and two lift rates (2 and 4 lifts/min). These lifting conditions were intended to cover, respectively, a wide range of potential exposures involving passive tissue strain and both low- and high-risk lifting rates (Marras et al. 1993). Sessions were conducted at a similar time on separate days with at least 72 hours between consecutive sessions. The presentation order of conditions was counterbalanced using two 6×6 Latin squares (one for each gender) to reduce potential order-related confounding effects.

Electromyography (EMG) of the erector spinae (at the L1 and L3 levels), rectus abdominus and external oblique muscles was collected bilaterally using bipolar Ag/AgCl surface electrodes, with electrode placements as previously reported (McGill 1991; Bazrgari et al. 2011a; Hendershot et al. 2011). Raw EMG signals were pre-amplified ($\times 100$) near the collection site, bandpass filtered (10–500 Hz), amplified ($\times 100$) and converted to root mean square (RMS; time constant $= 110$ ms) in hardware (Measurement Systems Inc., Ann Arbor, MI, USA). To measure the trunk flexion angle, a triaxial inertial measurement unit (IMU) with six degrees of freedom (Xsens Technologies XM-B-XB3, Enschede, The Netherlands) was placed over the T12 vertebral spinous process using medical-grade, double-sided tape. EMG signals were sampled at 1000 Hz and the IMU at 100 Hz.

After instrumentation and at the beginning of the first session, three trials were performed to record the lumbar flexion angle at full trunk flexion. Participants stood in a rigid metal frame and straps were used to restrain the pelvis and lower limbs (Figure 1). Subsequently, they slowly flexed forward from upright standing to full flexion (passive hanging position), with minimal muscle activity and their head facing down and their arms relaxed and hanging vertically. Participants remained in the flexed posture for 5 seconds, during which lumbar flexion angle was measured. As the pelvis was restrained, the angle measured from the IMU at T12 represented the lumbar flexion angle. The mean lumbar flexion angle across the three trials was obtained as the full lumbar flexion angle (FLFA) for each participant. Next, three trials of maximum voluntary contraction (MVC) were performed in neutral standing posture to assess trunk extension strength. During these, a rigid rod and chest harness assembly (Figure 1) were used, and participants were pulled back maximally on the rod for 5 seconds. Muscle activity (EMG) was collected and processed as described earlier, and force was measured (1000 Hz) using a load cell (Interface SM2000, Scottsdale, AZ, USA) on the harness-rod assembly. The maximum force and peak EMG values for each muscle were identified across the three trials and were used subsequently for normalisation (see Figure 1).

Repetitive lifting was performed for 40 min at a specific rate (i.e. either 2 or 4 lifts/min) that was timed by a digital metronome. The 40-min duration was based on the results of Parkinson et al. (2004), who observed a significant spinal creep after 30 min of repetitive flexion. For each lift, participants started from an upright standing posture, bent forward to grasp a 2.96 kg box located on a platform in front of their legs, lifted to an upright posture, placed the box back on the platform and then returned to an upright posture between lifts. Here, the box weight corresponds to values handled in low-risk occupational manual material handling tasks (Marras et al. 1993), represented roughly the 15th percentile of typical handled loads (Ciriello and Snook 1999), and was kept relatively low to minimise the effect of extensor muscle fatigue.

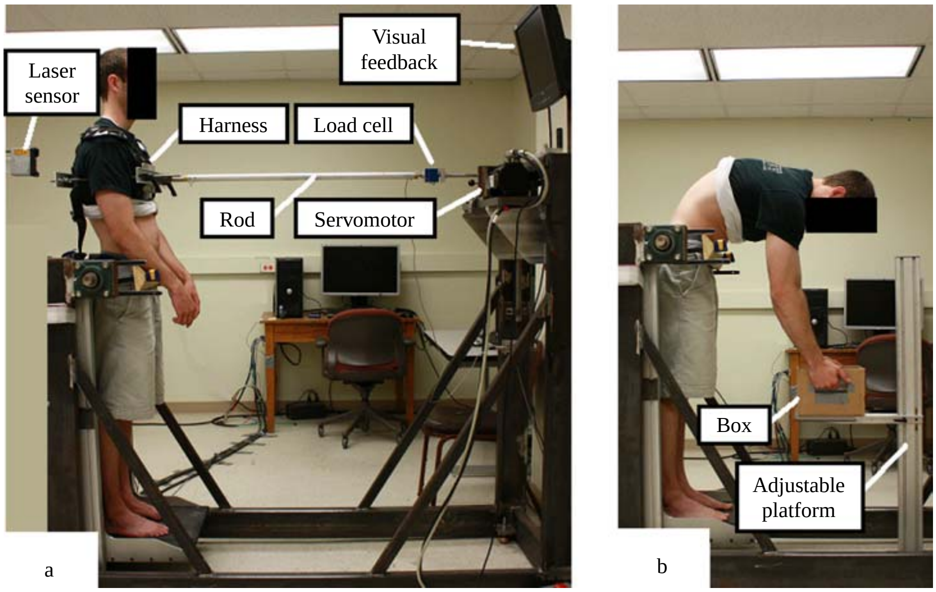

Figure 1. Experimental set-up for demonstrating a participant during (a) sudden perturbation procedure (the same set-up was used to perform MVC and fatigue tests) and (b) the start of a lifting effort. MVC = maximum voluntary contraction.

For each participant, platform height was adjusted so that the peak trunk flexion angle necessary to grasp the box handles was equivalent to 33, 66 or 100% of the participant's FLFA. After the period of repetitive lifting, participants maintained an upright standing posture, while constrained in the frame, for 20 min to assess immediate post-exposure effects and recovery.

Trunk behaviours were obtained using a sudden-perturbation paradigm following identical data collection and analysis procedures reported in our earlier studies (Bazrgari et al. 2011a; Hendershot et al. 2011; Miller et al. 2012). In brief, these measures were collected while participants were in an upright posture, during which a 45-second sequence of 12 small (± 5 mm), rapid (< 40 ms) anterior–posterior perturbations were imposed to the trunk via a servomotor, rigid rod and chest harness (Figure 1). Postural displacements were measured with a laser displacement sensor (Keyence LK-G 150, Osaka, Japan) and the motor encoder, whereas applied forces were measured using the in-line load cell. Prior to and during the perturbation sequences, participants maintained a constant submaximal extensor effort. The target effort was set to 10% of maximum voluntary RMS EMG in the bilateral L3 erector spinae. Measurements of mechanical and neuromuscular behaviours were recorded before, during (at 5, 10, 20, 30 and 40 min) and after (at 2.5, 5, 10 and 20 min) the repetitive lifting task (Figure 2). Short delays (~ 1 min) within the total lifting sequence were required to attach and remove the measurement equipment and to complete the perturbation sequence. Isometric reference contractions (to assess fatigue) were performed before the lifting task, after the task (after the final perturbation measurement at 40 min) and after recovery (after 20 min of standing). Reference contractions involved maintaining a submaximal extensor force of 50% MVC force for 30 seconds, which was maintained using real-time visual feedback of the force (Dolan and Adams 1998).

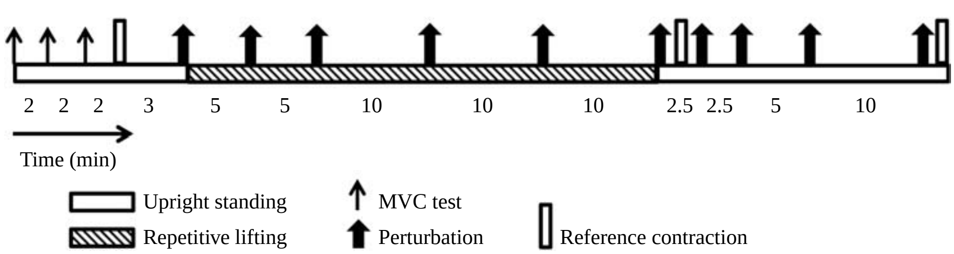

Figure 2. An overview of the experimental procedures including pre-exposure measurements (~ 10 minutes), exposure to repetitive lifting (~ 40 minutes) and a post-exposure recovery period (~ 20 minutes). Time intervals between selected measurement or tests are indicated.

2.3. Outcome measures

Similar to our previous studies (Bazrgari et al. 2011a; Hendershot et al. 2011; Miller et al. 2012), several outcome measures were obtained to characterise trunk mechanical and neuromuscular behaviour: (1) apparent mass and intrinsic trunk stiffness; (2) reflex delay; (3) maximum reflex force and (4) timing of maximum reflex force. In addition, changes in mean erector spinae EMG RMS and median power frequency (MF) were measured to assess muscle fatigue. The latency of reflexive muscle response (i.e. reflex delay) was determined as the time delay between an anteriorly directed perturbation and the onset of erector spinae muscle reflex response (Granata et al. 2005). Intrinsic trunk stiffness (because of both passive tissue and baseline muscle activation) was quantified from the trunk dynamic response to the applied perturbations in a predefined time window, which started from the onset of an anterior perturbation and ended at the reflex onset of the erector spinae musculature. Parameters of a model representing the trunk (apparent mass and intrinsic trunk stiffness) were estimated using a least-squares curve fit in MATLAB^{TM} (MathWorks, Natick, MA, USA); this was done separately for each direct-anterior perturbation by relating measured trunk kinematics to trunk kinetics (both measured in the horizontal direction at T8).

To characterise trunk neuromuscular behaviours, reflex forces were first estimated by subtracting the model-estimated intrinsic force contribution from the total measured trunk response (i.e. trunk reaction force measured by the inline load cell). Magnitude and timing (with respect to perturbation onset) of the maximum reflex force were quantified to represent the overall trunk reflexive behaviour (Bazrgari et al. 2011b). For each anteriorly directed perturbation, the analysis was limited to a time window of 150 ms after reflex onset to avoid voluntary muscle responses (Bazrgari et al. 2011a; Hendershot et al. 2011). From the three reference contractions, mean values of EMG RMS and MF of the extensor muscles (erector spinae muscle at the L1 and L3 levels) were derived after normalisation following similar procedures described earlier (Dolan and Adams 1998).

2.4. Analysis

Pre-exposure differences between genders in apparent mass, intrinsic trunk stiffness, reflex delay, and the magnitude and timing of the maximum reflex force were evaluated using separate single-factor analyses of variance (ANOVA). Paired *t*-tests were used to assess overall (across all conditions) changes in these measures immediately after the exposure period. Subsequently, all post-exposure measures were normalised to pre-exposure values ([post-pre]/pre), and the acute effects of flexion angle, lift rate and gender were assessed using mixed-factor analyses of variance (ANOVAs). A similar approach, using ANOVAs, was used to assess the effect of flexion angle and lift rate at different time periods during the lifting task.

Additional ANOVAs were used to assess changes in several measures during the exposure (or recovery) periods: (1) changes in the total trunk extensor muscle activation (maximum EMG RMS) during the last three versus first three lifts; (2) changes in the total submaximal extensor efforts (mean EMG RMS) and total flexor muscle antagonistic co-contraction (mean EMG RMS) generated in the last versus first perturbation sequences and (3) fatigue, as determined by changes in mean EMG RMS and mean MF of the trunk extensor muscles during the reference contractions. To achieve this, 'time' (i.e. pre-exposure, post-exposure, and recovery for reference contractions) was defined as an additional independent variable. As relevant, *post hoc* pairwise comparisons were performed using Tukey's honestly significant difference tests. All analyses were done using JMP^{TM} (Version 8, SAS Institute Inc., Cary, NC, USA), and statistical significance was concluded when $p < 0.05$. Summary results are presented as means (SDs). Reflex delay measures from one condition of one participant (66% FLFA and 4 lifts/min) were excluded because of an inability to capture EMG-based reflex responses consistently.

3. Results

3.1. Pre-exposure

Pre-exposure apparent mass and intrinsic trunk stiffness were significantly larger among males than females (Table 1). All measures of trunk reflexive behaviour (reflex delay and the magnitude and timing of maximum reflex force) were comparable between genders.

Table 1. Pre-exposure measures of trunk behaviours.

Measure	Male	Female	ANOVA	p Value
Apparent mass (kg)	20.3 (3.4)	16.7 (3.0)	$F_{(1,10)} = 5.6$	0.040*
Intrinsic trunk stiffness (N/m)	8968 (1039)	6139 (852)	$F_{(1,10)} = 42.2$	<0.0001*
Reflex delay (ms)	62.4 (4.5)	60.7 (5.5)	$F_{(1,9)} = 0.5$	0.48
Maximum reflex force (N)	194 (36)	193 (46)	$F_{(1,10)} = 0.004$	0.95
Timing of maximum reflex force (ms)	160.4 (8.2)	157.1 (8.3)	$F_{(1,10)} = 0.8$	0.38

Note: Means (SDs) are provided. *Significant differences between genders.

Table 2. Summary of statistical effects of repetitive lifting on trunk behaviours.

Measure	Overall difference	Main effect: Flexion angle	Main effect: Lift rate	Main effect: Gender
Intrinsic trunk stiffness	$t_{(71)} = -8.3$, $p < 0.0001$*	$F_{(2,45)} = 5.9$, $p = 0.0051$*	$F_{(1,45)} = 0.1$, $p = 0.75$	$F_{(1,10)} = 0.5$, $p = 0.49$
Reflex delay	$t_{(70)} = 2.7$, $p = 0.010$*	$F_{(2,44)} = 4.0$, $p = 0.025$*	$F_{(1,44)} = 2.8$, $p = 0.10$	$F_{(1,10)} = 0.8$, $p = 0.38$
Maximum reflex force	$t_{(71)} = -6.6$, $p < 0.0001$*	$F_{(2,45)} = 3.6$, $p = 0.034$*	$F_{(1,45)} = 4.0$, $p = 0.051$	$F_{(1,10)} = 2.9$, $p = 0.12$
Timing of maximum reflex force	$t_{(71)} = 2.5$, $p = 0.016$*	$F_{(2,45)} = 11.3$, $p = 0.0002$*	$F_{(1,45)} = 0.1$, $p = 0.74$	$F_{(1,10)} = 1.9$, $p = 0.20$

Note: The 'overall difference' column indicates whether there were significant post- versus pre-exposure changes across all conditions. Main effects of experimental conditions are also indicated. *Significant effects.

3.2. Post-exposure

Intrinsic trunk stiffness significantly decreased by 527 (541) N/m across all exposure conditions (Table 2), and the effect of exposure was significantly larger with increased flexion angle (Figure 3). There was a significant increase in reflex delay post-exposure. Although the mean difference was small (1.2 [3.9] ms), it was also significantly larger with increased flexion angle (Figure 3). Both the magnitude and timing of maximum reflex force significantly changed post-exposure, respectively, decreasing and increasing; pre- versus post-exposure values of maximum reflex force were 193 (41) versus 167 (44) N and 159 (8) versus 161 (12) ms for the timing of maximum reflex force. These changes were also significantly affected by the flexion angle, with larger changes after exposure to increased angle (Figure 3). However, flexion angle had no interactive effect on any of the outcome measures ($p > 0.11$). There were no main effects of lift rate, although the effect on changes in maximum reflex force approached significance ($p = 0.051$), with respective decreases of 20 (28) and 33 (28) N in the 2 and 4 lifts/min conditions. There were also no main effects of gender on any of the outcome measures. No interactive effects involving lift rate or gender were significant ($p > 0.11$), except for a

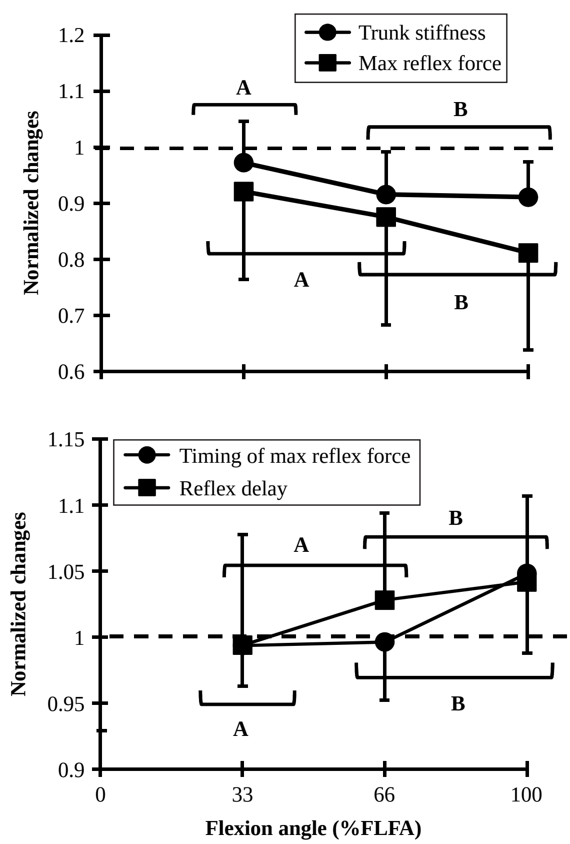

Figure 3. Effects of flexion angle on intrinsic trunk stiffness, the magnitude and timing of maximum reflex force, and reflex delay. Normalised changes are indicated (e.g. a value of 1.05 indicates a 5% increase from pre-exposure levels). Error bars indicate SDs, and *post hoc* groupings are indicated by brackets and letters. FLFA, full lumbar flexion angle.

gender × lift rate effect on maximum reflex force ($p = 0.030$). Regarding the latter, females had a more substantial reduction in maximum reflex force in response to the higher lift rate, whereas the reduction among males was comparable between the two lift rates.

Total trunk extensor muscle activity (EMG RMS) increased by ~5% during lifting at the end of the exposure period (i.e. last vs. first three lifts); however, the main effect of time was not significant ($p = 0.73$). EMG RMS values were comparable between the last and first perturbation trials during the lifting exposures ($p = 0.16$ for total extensor muscle EMG RMS and $p = 0.99$ for total flexor muscle EMG RMS). There were no significant changes in EMG RMS or MF values during the reference contractions performed pre- and post-exposure or after recovery ($p > 0.54$). Moreover, there was no interactive effect of time with flexion angle, lift rate and gender for any of these outcome measures ($p > 0.19$).

3.3. Recovery

Post-exposure reductions in intrinsic trunk stiffness recovered by ~43% after 20 min of upright standing (Figure 4), whereas changes in reflex delay were recovered by ~70% after the same recovery period. In contrast, recovery of maximum reflex force was slower with an overall increase of ~14% as compared with post-exposure values. No recovery was evident for the timing of maximum reflex force, which instead continued to increase during the recovery period. Comparisons between initial (pre-exposure) and final (post-recovery) values showed that only reflex delay was fully

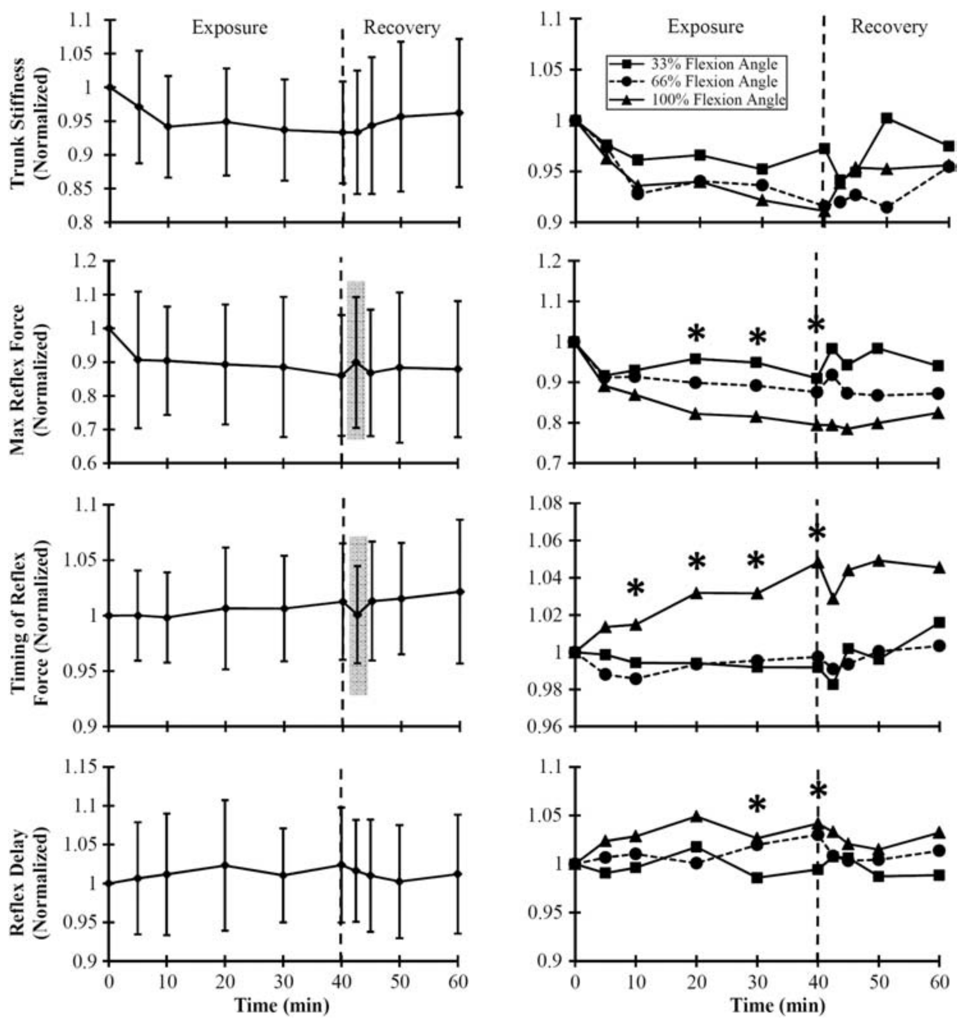

Figure 4. Normalised changes in mean trunk behaviours across all loading conditions (intrinsic trunk stiffness, magnitude and timing of maximum reflex force, and reflex delay) during the exposure and recovery periods (left), and the effect of flexion angle on mean values of the same parameters across both lift rates (right). Error bars indicate SDs and asterisks indicate significant effects of flexion angle at a given time. An apparent hyperexcitability during recovery is highlighted with grey bars.

recovered ($t_{(68)} = 1.01$, $p = 0.32$), whereas the other outcome measures were still significantly different from pre-exposure values ($p < 0.012$). Qualitatively, no consistent differences in recovery behaviours were evident between flexion angle exposures for any of the outcome measures (Figure 4).

4. Discussion

4.1. Intrinsic trunk behaviour

Intrinsic trunk stiffness, which here represents the overall contributions of passive trunk tissues and baseline muscle activation around the neutral standing posture, decreased by $\sim 7\%$ across all lifting conditions after 40 min of repetitive lifting. Muscular fatigue is a potential confounding influence, because at a similar level of EMG RMS fatigued muscles can produce smaller forces as compared with an unfatigued state; yet, there was no evidence of fatigue after the lifting task (i.e. based on EMG RMS and MF). Furthermore, baseline muscle activation, another potential confounding effect, was not significantly changed between initial and final perturbations. As such, the observed decrease in stiffness here likely resulted from reductions in passive contributions to intrinsic trunk stiffness. Previous *in vivo* studies have reported similar reductions in passive stiffness of the human trunk after repetitive flexion/lifting exposures (Parkinson et al. 2004; Olson et al. 2009; Shin and D'souza, 2010; Muslim et al. 2013). Noticeable creep deformation was also observed in studies of repetitive flexion exposure using feline spines (Solomonow, 2011), which can be interpreted as a reduction in rotational stiffness. Although repetitive lifting is different from repetitive flexion, similar viscoelastic deformation and consequently reduction in passive trunk stiffness are expected due to stretching/deformation of soft tissues.

Changes in intrinsic trunk stiffness caused by repetitive lifting were more substantial with larger trunk flexion angles (Figures 3 and 4). Our earlier results for prolonged trunk flexion exposures also demonstrated a larger decrease in intrinsic trunk stiffness with increasing flexion angle (Hendershot et al. 2011; Toosizadeh et al. 2012); however, there is a subtle but notable difference in these effects of flexion angle. In this study, reductions in intrinsic trunk stiffness were almost equal in response to exposures involving 66 and 100% of full flexion (Figure 3), suggesting a non-linear effect of increasing flexion angle on decreases in stiffness. In our earlier study of prolonged trunk flexion (Hendershot et al. 2011), the results suggested increasing effects up to the full flexion exposure. One contribution to this difference is likely the way in which the maximum (100%) flexion exposure level was achieved. Here, the maximum trunk flexion angle was used, whereas in previous study the flexion-relaxation angle was used. As such, maximum trunk flexion angle was larger here, leading to a larger range of flexion angle exposures.

Qualitatively, alterations in intrinsic trunk stiffness during both the exposure and rest periods showed exponential behaviours (Figure 4) similar to those reported earlier (McGill and Brown 1992; Solomonow 2011). Approximately 87% of the decrease in intrinsic trunk stiffness occurred in the first 10 min of exposure, and 81% of the total recovery occurred within the first 10 min of the standing recovery period. Furthermore, the rate of recovery of intrinsic trunk stiffness appeared to be slower here as compared with the rate of changes during exposure, which is also consistent with previous experiments (Hoops et al. 2007; Solomonow 2011).

On the basis of earlier results, spine rotational stiffness and stability are closely related (Bazrgari and Shirazi-Adl 2007; Graham and Brown 2012) and the reductions in passive stiffness observed here could compromise trunk stability and increase the risk of LBDs (Hoogendoorn et al. 2000). To stabilise the trunk, an increase in paraspinal muscle activation may be required to compensate for reductions in passive stiffness (Marras and Granata 1997; Olson et al. 2009). We observed that the total trunk extensor muscle activation (EMG RMS) increased during lifts performed at the end versus beginning of the exposure periods. Although the increases were small ($\sim 5\%$ across all exposure conditions) and not statistically significant, this effect may still be of relevance given that the moment arms of paraspinal muscles are also relatively small. As such, even small increases in muscle forces (which is considered to be the case given a lack of evidence for muscle fatigue) could result in important increases in spine loads (e.g. compressive forces).

4.2. Trunk neuromuscular behaviour

Repetitive lifting reduced and delayed maximum reflex forces. Three main factors have been hypothesised to influence muscle reflexive behaviour: laxity in viscoelastic tissues, fatigue and habituation (Solomonow et al. 1999; Granata et al. 2005; Jackson et al. 2009; Bazrgari et al. 2011a; Hendershot et al. 2011). Here, EMG-based measures provided no evidence for muscle fatigue and the effect of habituation was likely minimised using pseudorandom timing of the perturbations. As such, and in agreement with previous work (Sánchez-Zuriaga et al. 2010), the observed changes in trunk muscle reflexive behaviours likely arose from strains in ligaments and other soft tissues rather than muscle fatigue. Reductions in reflex force as a result of strains in ligaments were observed in earlier work, which showed that reflexive activities of the feline multifidus decreased during/after repetitive loading exposures (Solomonow et al. 1999). Comparable changes in the magnitude and timing of reflex responses have also been observed in human skeletal muscles surrounding other joints after passive stretching (Rosenbaum and Hennig 1995; Avela et al. 1999).

Similar to intrinsic trunk stiffness, trunk reflexive behaviours were significantly influenced by the trunk flexion angle (Figures 3 and 4). Although the trunk flexion angle during lifting is an important factor in work design, no study has, to our knowledge, investigated the effect of this factor on trunk behaviours. Overall, the current results indicate an adverse effect of increased trunk flexion angle during repetitive lifting on reflexive and mechanical behaviours of the trunk. Regarding recovery behaviours, reflex delay and maximum reflex force tended to recover during 20 min of standing; in contrast, the timing of reflex force presented an increasing trend even during the recovery period (Figure 4). There was an apparent muscular hyperexcitability in the first measures during the recovery period (i.e. 42.5 min) (Figure 4), which are noticeable as respective increases and reductions in the magnitude and timing of maximum reflex force immediately after the exposure period. Similar behaviour was also observed using feline spines and was suggested as a protective mechanism to compensate for reductions in intrinsic trunk stiffness immediately after flexion exposures (Claude et al. 2003; Solomonow 2011).

Although not statistically significant, reductions in the maximum reflex force tended to be larger with a higher lift rate. Prior evidence suggested an increased creep and reductions in trunk reflexive behaviours with an increasing rate of repetitive flexion (Lu et al. 2008). However, the lifting task here involved active muscle activity, as opposed to repetitive passive trunk flexion as in Lu et al. (2008). Moreover, there is a higher likelihood of extensor muscle fatigue with increasing lift rate (Kim and Chung 1995). Therefore, high lift rates were avoided here to prevent extensor muscle fatigue and potential confounding effects on the outcome measures.

Repetitive versus prolonged strain of trunk soft tissues appears to result in divergent influences on trunk reflex force. In contrast to repetitive lifting (this study), prolonged flexion exposures leads to an increase in post-exposure maximum reflex force (Bazrgari et al. 2011a; Hendershot et al. 2011). Although an adverse effect of trunk flexion on reflexive behaviour is apparent, the underlying mechanisms that can lead to either larger or smaller reflex forces are still unknown. Further, the timing of maximum reflex force and reflex delay increased here after repetitive lifting. In this work, the timing of maximum reflex force and reflex delay were estimated by different methods, respectively, using measured force and EMG, and consistency between these results provides some level of consensual validity. In general, reflex responses of the trunk muscles play an important role in controlling the stability of the spine, and with less energy expenditure as compared with voluntary co-contraction of trunk muscles (Franklin and Granata 2007; Moorhouse and Granata 2007). Therefore, any alteration in reflexive behaviours, as found in this study, could compromise the efficiency of the neuromuscular system in stabilising the spine. Of note, higher reflex forces following prolonged flexion exposures in our earlier work (Bazrgari et al. 2011a; Hendershot et al. 2011) were suggested as a compensatory mechanism for intrinsic trunk stiffness reduction. Yet as found in this study, concurrently reduced stiffness and reflex force could impose a higher risk of LBDs.

4.3. Implications, limitations and conclusions

In general and in support of our first hypothesis, the present results indicate adverse effects of larger trunk flexion angles during lifting on both mechanical and neuromuscular properties of the trunk. Except for maximum reflex force, which decreased by $\sim 10\%$ across all conditions after lifting task, all outcome measures showed no substantial changes for lifting from heights associated with 33% of maximum trunk flexion angle. This suggests that some potential adverse effects of repetitive lifting could be avoided by controlling lifting height (i.e. to result in trunk flexion less than about one-third of flexion range-of-motion). Limited evidence was found to suggest that a reduction in lift rate would have similar beneficial effects.

Overall, and as illustrated in Figure 4, recovery (to pre-exposure values) was faster for mechanical properties as compared with recovery of the magnitude and timing of maximum reflex force (i.e. supporting the second hypothesis). Further, although the period of recovery here was only 20 min, trends in the outcome measures (except for reflex delay, which had a more rapid recovery) suggest that full recovery requires a longer recovery time versus the exposure time. As such, relatively long rest/recovery periods may be needed to prevent the accumulation of mechanical and neuromuscular disturbances during occupational tasks requiring repetitive lifting. However, additional studies are required to determine more specifically what rest periods or duty cycles would be sufficient. For example, this study involved exposures that were limited to 40 min to minimise potential confounding effects of prolonged standing posture on trunk behaviours (e.g. axial creep and muscle fatigue). Future studies should consider longer work/rest periods, a larger range of lift rates and more realistic working conditions.

One potential limitation in our data analyses is related to the method used for estimating intrinsic trunk stiffness. As discussed earlier (Bazrgari et al. 2011a; Hendershot et al. 2011), measured reductions in intrinsic stiffness might be conservative (underestimated), since measurements were performed in a posture (neutral standing) that is associated with the lowest intrinsic trunk stiffness (Parkinson et al. 2004; Shirazi-Adl 2006). Another limitation of this study is that changes in trunk flexion range of motion as a result of viscoelastic deformations were not measured, since perturbation measurements were performed immediately after the lifting task. As such, direct measurements of viscoelastic deformation imposed by repetitive lifting were not available here, but which should be explored in future studies. Furthermore, trunk damping has recently been suggested as a means for better understanding spinal stability and neuromuscular control (Cholewicki et al. 2005); however, we forced trunk damping to zero to prevent our model predictions of the dynamic responses of the trunk to be dominated by damping. This simply transferred the predominant response from damping to stiffness, to facilitate comparisons with existing literature. However, some caution is warranted when interpreting the relative contribution of elastic and viscous components here and in earlier work, because the models used likely represent oversimplifications of the spine and predicted properties may not be the best physical representation of the system.

In summary, the present work provides experimental evidence of mechanical and neuromuscular alterations in the human trunk caused by repetitive lifting. Our earlier results indicated an increase in reflex force after prolonged trunk flexion, and this increase was suggested to compensate for decreases in passive trunk stiffness (Bazrgari et al. 2011a; Hendershot et al. 2011). Alterations in passive trunk stiffness and reflex responses here indicate a deterioration in both behaviours following a repetitive lifting task. Consequently, simultaneous changes in both passive trunk stiffness and neuromuscular behaviours can be interpreted as inducing a higher risk of LBD when performing demanding tasks (such as lifting heavy weights) following repetitive versus prolonged trunk flexion exposures. As such, and in agreement with previous suggestions (McGill 2007), it may be important in job design to avoid demanding tasks after repetitive lifting unless sufficient rest/recovery is provided.

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