Gait & Posture 73 (2019) 93–100

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

# Influencing lumbar posture through real-time biofeedback and its effects on the kinematics and kinetics of a repetitive lifting task

Mark Boocock[a,*], Yanto Naudé[a], Steve Taylor[b], Jeff Kilby[c], Grant Mawston[a]

[a] Health and Rehabilitation Research Institute, Auckland University of Technology, Auckland, New Zealand
[b] Department of Biostatistics and Epidemiology, Auckland University of Technology, Auckland, New Zealand
[c] School of Engineering, Computing and Mathematical Sciences, Auckland University of Technology, Auckland, New Zealand

ARTICLE INFO

*Keywords:*
biofeedback
low back/lumbar spine
manual handling
biomechanics/spine
repetitive lifting

ABSTRACT

*Background:* Repetitive, flexed lumbar postures are a risk factor associated with low back injuries. Young, novice workers involved in manual handling also appear at increased risk of injury. The evidence for the effectiveness of postural biofeedback as an intervention approach is lacking, particularly for repetitive, fatiguing tasks.
*Research question:* How does real-time lumbosacral (LS) postural biofeedback modify the kinematics and kinetics of repetitive lifting and the risk of low back injury?
*Methods:* Thirty-four participants were randomly allocated to two groups: biofeedback (BF) and non-biofeedback (NBF). Participants repetitively lifted a 13 kg box at 10 lifts per minute for up to 20 min. Real-time biofeedback of LS posture occurred when flexion exceeded 80% maximum. Three-dimensional motion analysis and ground reaction forces enabled estimates of joint kinematics and kinetics. Rating of perceived exertion (RPE) was measured throughout.
*Results:* The BF group adopted significantly less peak lumbosacral flexion (LSF) over the 20 min when compared to the NBF group, which resulted in a significant reduction in LS passive resistance forces. This was accompanied by increased peak hip and knee joint angular velocities in the BF group. Lower limb moments did not significantly differ between groups. Feedback provided to participants diminished beyond 10 min and subjective perceptions of physical exertion were lower in the BF group.
*Significance:* Biofeedback of lumbosacral posture enabled participants to make changes in LSF that appear beneficial in reducing the risk of low back injury during repetitive lifting. Accompanying behavioural adaptations did not negatively impact on physical exertion or lower limb joint moments. Biofeedback of LS posture offers a potential preventative and treatment adjunct to educate handlers about their lifting posture. This could be particularly important for young, inexperienced workers employed in repetitive manual handling who appear at increased risk of back injury

## 1. Introduction

Low back pain (LBP) is one of the most common, costly and disabling musculoskeletal conditions treated by health professionals [1]. Once affected, the recurrence of LBP is high, with 24%–80% of LBP sufferers experiencing further episodes annually [2]. Global estimates of LBP suggest a point prevalence of 12% and a one month prevalence of 23% [3]. In the United States, the estimated annual cost of back pain is in excess of $100 billion [4].

Occupations involving heavy and repeated lifting in flexed lumbar postures are considered high risk for low back injury [5–7]. Risk factors associated with LBP include the magnitude and repetition of trunk flexion, and time spent in flexed trunk postures [8–11]. Fatigue failure of the spine during repetitive loading occurs more rapidly in flexion, particularly when approaching end range of motion (ROM) [12]. Young, inexperienced workers who are in the first year of employment appear at increased risk of LBP [13].

Health professionals often prescribe postural training as a preventative measure to reduce forces on the lumbar spine when lifting [14]. Whilst this may lead to short-term changes in behaviour, the ability to maintain effects is uncertain, particularly when individuals become fatigued [15]. This may be due to impairment of the body's intrinsic feedback system [16] and/or reduced neural control of the spinal muscles when fatigued [17]. When intrinsic feedback is impaired, external feedback may be a useful approach to reduce hazardous spinal postures.

* Corresponding author at: Health and Rehabilitation Research Institute, Auckland University of Technology, Private Bag 92006, Auckland 1142, New Zealand.
*E-mail address:* mark.boocock@aut.ac.nz (M. Boocock).

https://doi.org/10.1016/j.gaitpost.2019.07.127
Received 27 March 2018; Received in revised form 19 May 2019; Accepted 2 July 2019

**Table 1**
Mean (standard deviation) demographics for the biofeedback (BF) and non-biofeedback (NBF) groups.

| | BF (n = 18) | NBF (n = 16) |
|---|---|---|
| Age (years) | 25.7 (4.6) | 25.6 (5.1) |
| Height (m) | 1.80 (0.08) | 1.84 (0.08) |
| Body weight (kg) | 79.8 (11.2) | 85.5 (13.8) |
| Body Mass Index (BMI) | 24.7 (3.10) | 25.6 (2.8) |
| Maximum range of lumbar flexion (°)[a] | 51.3 (8.1) | 45.6 (10.8) |
| Maximum range of trunk flexion (°)[a] | 53.5 (16.5) | 55.6 (8.9) |

[a] Measured from upright standing and prior to the lifting task.

The aim of this study was to determine the ability of the handler to modify lumbosacral (LS) posture in response to real-time external biofeedback (BF) during a repetitive lifting task. It was hypothesised that those provided with BF would maintain LS posture below the prescribed threshold, but there would be increased reliance on feedback with fatigue. A secondary aim was to determine the behavioural adaptations adopted to comply with feedback and the potential consequences for the risk of injury.

## 2. Methods

### 2.1. Participants

Thirty-six healthy adults were recruited from a university student population via advertising (notice board) and randomly allocated to either: 1) a BF group that received feedback on lumbar posture (n = 18); and 2) a non-biofeedback (NBF) group (n = 18). Block (BF or NBF) randomisation ensured similar sample sizes. Data from two participants in the NBF group were excluded due to data recording issues. Demographic characteristics of the participants are in Table 1.

Participants were excluded from the study if they had: a back injury or complaint in the last six months; undergone spinal surgery; any cardiovascular or neurological condition; and a musculoskeletal injury at the time of the study. None of the participants were experienced in manual handling or performed regular handling in their work. Sample size estimates (16 per group) were based on a previous study [18] using an effect size of 0.9 at an alpha level of 0.05 and power of 0.8. The study was approved by the university ethics committee and all participants gave informed consent.

### 2.2. Lifting task

Both groups performed the same repetitive lifting task, previously shown to induce back muscle fatigue [18]. Participants lifted and lowered a box weighing 13 kg at a frequency of 10 lifts/minute (Fig. 1A). The box was 30 × 25.5 × 25 cm (length/width/height) and had two cylindrical handles (2.8 cm in diameter) extending 6 cm either side of the box, 17 cm above its base. An electronic metronome provided an audible cue to commence each lift and lower. Participants initially lifted the box to an upright standing position with their arms extended and the box resting against their thighs. It was then lowered onto a platform, 15 cm above the floor. Further details of the lifting task can be found elsewhere [18].

Participants were not informed of task duration, but were verbally encouraged to continue for as long as possible. They could stop at any time if they felt excessive discomfort or were unable to continue. All participants were stopped after 20 min. Every minute, participants rated perception of physical exertion using Borg's 15 point Rating of Perceived Exertion (RPE) scale [19].

### 2.3. Biofeedback of lumbosacral posture

The BF group received real-time feedback on lumbar posture using two wireless Shimmer inertial sensors (Shimmer Sensing, Ireland) fixed to the skin superficial to the first lumbar (L1) spinous process and sacral body (S1) (Fig. 1B). Purpose-designed software provided audible feedback (high pitched tone) when 80% of maximum LS ROM was exceeded, a posture beyond which passive loading on the lumbar spine significantly increases [20]. Maximum LS ROM for each participant was determined at the beginning of each session using the method described in Section 2.4, from which 80% LS ROM was calculated. Prior to the lifting task, participants in the BF group were familiarised with the auditory feedback by adjusting LS posture to turn on and off the sound while performing practice lifts. Participants were instructed to adjust their LS posture to prevent the audible tone. The NBF group performed a similar period of familiarisation, but without feedback. Participants were instructed to maintain a fixed, symmetrical foot position, and maintain a hold on the box handles. No other instructions about lifting technique were provided. LS angular displacement and velocity were recorded continuously, along with the number and times feedback was provided to participants.

### 2.4. Kinematic measures

A nine camera motion analysis system (Qualysis AB, Sweden) sampling at 120 Hz recorded 3-dimensional (3D) kinematics. Seventy-seven lightweight, retro-reflective markers were attached to participants' skin to track the position and movement of body segments (Fig. 1A). Markers defined the dimension and axis of body segments, with cluster markers used to track segments and box [18]. Initially, participants adopted a standing position as a reference posture ('static' trial) for biomechanical modelling. Medial markers on the ankle and knees were removed after the static trial.

### 2.5. Lumbar range of motion

Two pairs of reflective markers on small rods fixed to the inertial sensors provided a measurement of LS angle (Fig. 1B) [21]. Prior to lifting, participants maximally flexed their spine while standing [22], with their knees slightly flexed. LS angle was expressed as a percentage of the ROM from upright standing to full flexion:

$$\%LSF = (\Theta t - \Theta s)/(\Theta m - \Theta s) \tag{1}$$

where:
$\%LSF$ = percentage lumbosacral flexion
$\Theta t$ = peak flexion angle during the lifting task
$\Theta s$ = standing angle during the 'static trial'
$\Theta m$ = maximum lumbosacral flexion angle

Percentage trunk flexion (%TF) was expressed similarly.

### 2.6. Force plate data

Participants stood with each foot on an AMTI (Advanced Mechanical Technology Inc., USA) force platform. 3D Ground reaction forces and moments were sampled at 1200 Hz and synchronised with kinematics. Kinematic and kinetic data for two complete lifting cycles were recorded each minute, where a cycle was one complete lift and lower of the box.

### 2.7. Biomechanical model

A 15 segment, rigid-link dynamic biomechanical model of the lower body and upper limbs, pelvis, trunk and head was constructed in Visual 3D (C-Motion Inc., USA) (Fig. 1C). Body segments were represented as geometric objects [23] and scaled to each person's anthropometrics. The mass, centre of mass and inertial characteristics of each segment were estimated using Winter's [24] regression equations. Kinematic and kinetic data were smoothed using a recursive Butterworth lowpass filter

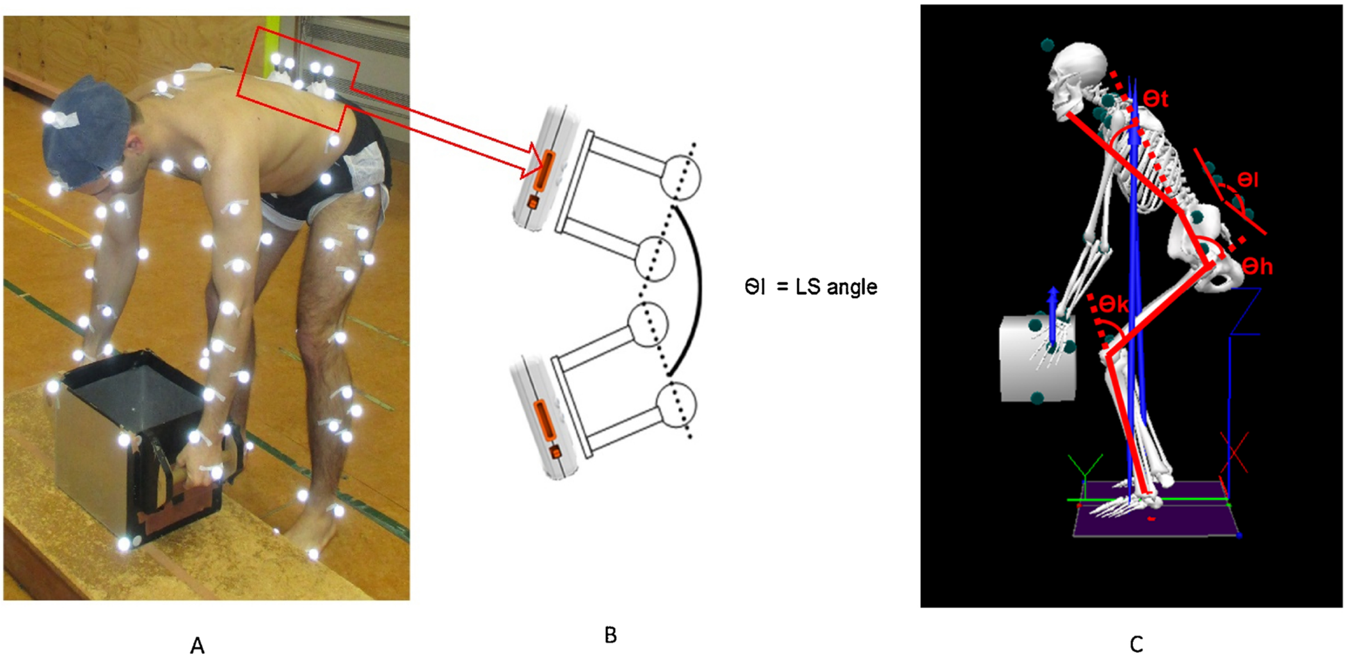


**Fig. 1.** A participant performing the lifting task (A). Real-time biofeedback on lumbar posture was provided by two inertial sensors attached to the lumbar spine (B). Two pairs of reflective markers attached to the sensors provided additional kinematic measures of lumbosacral (LS) angle (B). Relative joint angles (ϴl = lumbosacral angle; ϴt = trunk angle; ϴh = hip angle; ϴk = knee angle) and joint moments (knee, hip and back) were calculated using a fifteen segment, rigid-link, dynamic biomechanical model and external ground reaction forces (C).

with a cut-off frequency of 6 Hz.

### 2.8. Bending moment

Bending moments resisted by passive lumbar spinal structures ($I$) were estimated as a function of percentage lumbosacral flexion (%LSF) using the equation from Dolan et al. [20]:

$$I = 7.97 \times 10^{-5} \times \%LSF^{3} + 12.9 \quad (2)$$

### 2.9. Data analysis

Relative joint angles and angular velocities were determined at the pelvis, hip, and knee joints (Fig. 1C). Inverse dynamics were used to estimate joint reaction forces and net moments about the trunk, hips, and knees. Joint moments were normalised according to body weight. Dependent measures included mean peak angular displacements (flexion), angular velocities (extension) and joint moments (extension) of the two cycles (when the box left the platform to when it came to rest in upright standing) measured each minute. Independent measures were group allocation: BF or NBF.

### 2.10. Statistical analysis

Each outcome measure was analysed using mixed models to account for correlation of repeated measures within participants. Based on a previous study [18], linear mixed models with quadratic terms were applied that included two random effects: an intercept and a slope per participant. Fixed-effect terms were used to estimate differences between groups for intercept, slope and curvature, by introducing each additional term individually. Pairs of sequential models (one with and one without the additional fixed-effect term) were compared using the likelihood-ratio Chi-squared test. Differences in intercepts provided an indication of group differences at the start of the task, while the slopes or curvature provided evidence of differences over time. The models were fitted using R version 3.3.3 [25] and 'lme4' software [26]. Independent t-tests (IBM SPSS Statistics v24, USA) determined differences between groups at 20 min. A statistical significance of 0.05 was applied throughout.

## 3. Results

There was no difference between groups for demographic variables (Table 1). Sixteen participants in the NBF group (100%) and 15 (83%) in the BF group completed 20 min, with median lifting times of 20 and 18 min, respectively. The three participants who failed to complete 20 min cited lower back discomfort as the primary reason for discontinuing.

At the start of the task, the NBF group took significantly longer to perform a lift than the BF group (Table 2). At 20 min, the NBF group had reduced lifting times, but this remained significantly longer than the BF group.

### 3.1. Lumbosacral posture feedback

Two participants in the BF group received no feedback throughout the task. The number of participants who received feedback increased until the 7$^{th}$ minute, with nine receiving feedback between 6th and 7th minutes. Thereafter, the number receiving feedback reduced, with two participants receiving feedback after 13 min. Most feedback occurred between the 10th and 11th minute (mean = 3.16; 95% confidence interval (CI) = 1.37–4.95). This declined steadily after the 14th minute, reaching a mean of 1.16 (95%CI = 0.08–2.25) in the 19th–20th minute.

### 3.2. Rating of perceived exertion

There was no between group difference in RPE initially (mean = 6 ("No exertion at all") at 1 min for both groups), but significant differences in the slope of the fitted models (quadratic). At 20 min, mean RPE was significantly higher (mean = 17.1 ("Very hard"); 95%CI = 16.0–18.2) in the NBF group compared to the BF group (mean = 15.8 ("Hard (heavy) – Very hard"); 95%CI = 14.9–16.7) (P = 0.006).

**Table 2**
Mean and 95% confidence intervals (CI) for the kinematic and kinetic variables of the biofeedback (BF) and non-biofeedback (NBF) groups at the start (1st minute) and end (20th minute) of the lifting task.

| | Start (1st minute) | | | | | End (20th minute) | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | BF | | NBF | | P Value* | BF | | NBF | | P Value† |
| | Mean | 95% CI | Mean | 95% CI | | Mean | 95% CI | Mean | 95% CI | |
| Time to perform lift (s) | 1.08 | 1.01–1.14 | 1.56 | 1.37–1.76 | < 0.001 | 1.07 | 0.99–1.14 | 1.31 | 1.17–1.45 | = 0.01 |
| *Kinematic variables – peak values* | | | | | | | | | | |
| LSF angle (°) | 176.3 | 173.4–179.1 | 188.2 | 182.7–193.8 | = 0.001 | 186.5 | 182.6–190.4 | 201.7 | 197.0–206.3 | < 0.001 |
| %LSF | 45.6 | 40.1–51.1 | 71.7 | 59.7–83.7 | = 0.001 | 64.3 | 57.3–71.3 | 98.4 | 91.0–105.8 | < 0.001 |
| LSF angular velocity (°/s)[b] | 35.4 | 30.7–40.1 | 30.5 | 24.3–36.7 | NS (0.217) | 49.3 | 41.2–57.5 | 57.1 | 47.2–66.9 | NS (0.272) |
| TF angle (°) | 24.3 | 20.6–27.9 | 35.4 | 28.5–42.4 | = 0.007 | 27.4 | 23.7–31.1 | 48.3 | 43.5–53.2 | < 0.001 |
| %TF | 42.1 | 35.9–48.1 | 63.9 | 52.9–74.8 | = 0.002 | 47.3 | 41.2–53.4 | 87.7 | 80.4–94.9 | < 0.001 |
| TF angular velocity (°/s)[b] | 29.4 | 25.2–33.5 | 37.5 | 35.6–39.5 | NS (0.145) | 47.5 | 29.4–65.7 | 69.4 | 66.1–72.6 | NS (0.079) |
| Hip angle (°)[a] | 87.5 | 87.1–87.9 | 93.0 | 87.1–98.9 | NS (0.159) | 86.3 | 85.8–86.8 | 87.5 | 81.8–93.1 | NS (0.775) |
| Hip angular velocity (°/s)[b] | 137.4 | 134.7–140.0 | 80.9 | 67.5–94.3 | < 0.001 | 141.9 | 139.7–144.0 | 89.6 | 78.9–100.4 | < 0.001 |
| Knee angle (°)[a] | 61.1 | 51.5–70.7 | 61.3 | 51.3–71.3 | NS (0.978) | 59.3 | 53.7–64.9 | 53.2 | 44.1–62.4 | NS (0.472) |
| Knee angular velocity (°/s)[b] | 102.5 | 99.3–105.6 | 73.9 | 58.9–88.8 | = 0.015 | 113.1 | 110.0–116.3 | 82.3 | 65.2–99.4 | = 0.015 |
| *Kinetic variables – peak values* | | | | | | | | | | |
| L5/S1 moment (Nm/kg) | 2.40 | 2.36–2.45 | 2.16 | 2.04–2.27 | NS (0.177) | 2.38 | 2.35–2.42 | 2.17 | 2.02–2.32 | NS (0.953) |
| Passive bending moment (Nm/kg) | 0.3 | 0.25–0.34 | 0.54 | 0.4–0.69 | = 0.004 | 0.44 | 0.34–0.54 | 1.12 | 0.93–1.3 | < 0.001 |
| Hip moment (Nm/kg) | 1.45 | 1.33–1.56 | 1.48 | 1.36–1.59 | NS (0.544) | 1.42 | 1.26–1.57 | 1.41 | 1.32–1.51 | NS (0.682) |
| Knee moment (Nm/kg) | 0.46 | 0.44–0.49 | 0.40 | 0.33–0.47 | NS (0.089) | 0.40 | 0.37–0.43 | 0.37 | 0.27–0.46 | NS (0.372) |

Abbreviations: LSF Lumbosacral flexion; TF Trunk flexion; NS none significant.
* P value denotes the intercept difference of fitted linear models.
† P value denotes the difference between groups based on independent *t*-tests at 20 min.
[a] Hip and knee values are for the left limb.
[b] Angular velocities refer to peak values of extension during the lift.

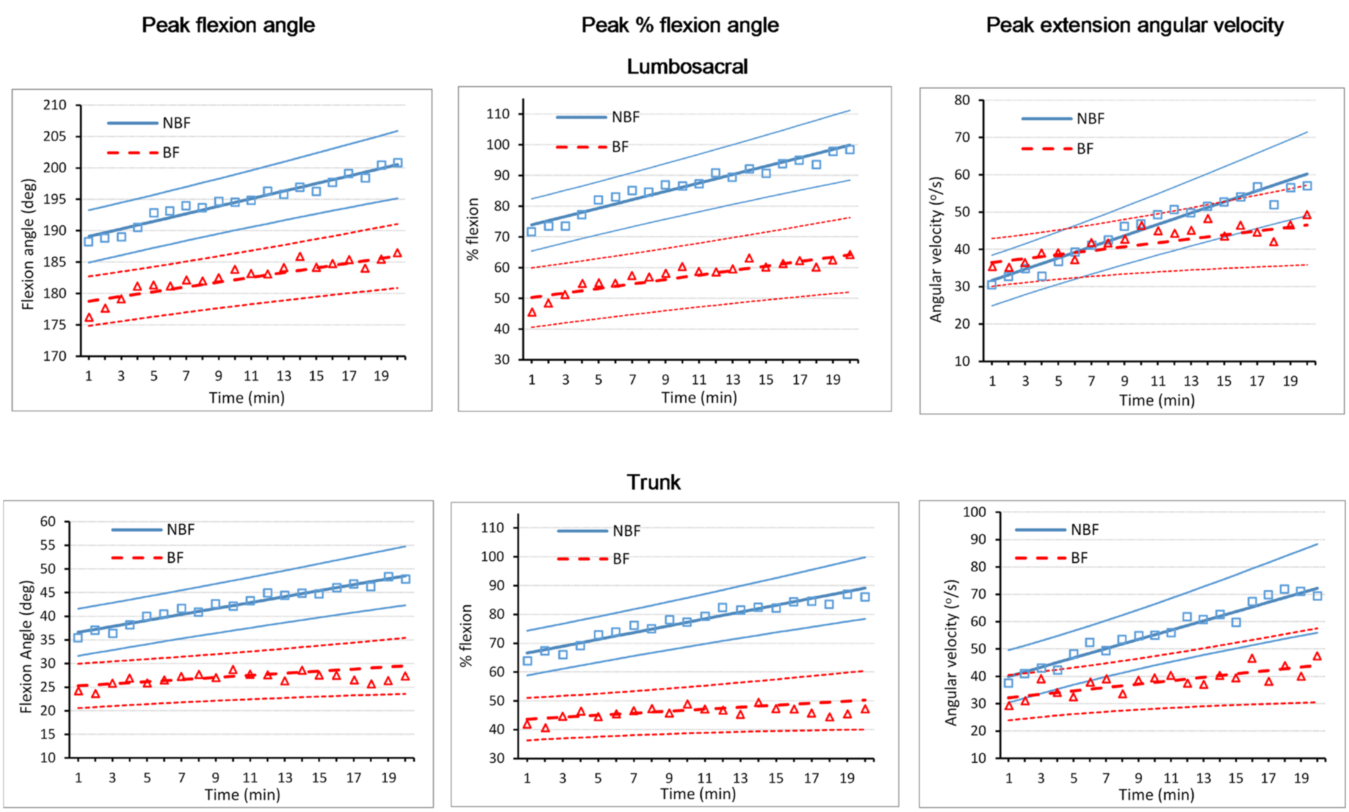


**Fig. 2.** Mean peak lumbosacral and trunk flexion angles, percentage flexion and extension angular velocities for the biofeedback (BF) and non-biofeedback (NBF) groups during each minute of the 20 min lifting. Lines indicate best-fit linear models and corresponding 95% confidence intervals.

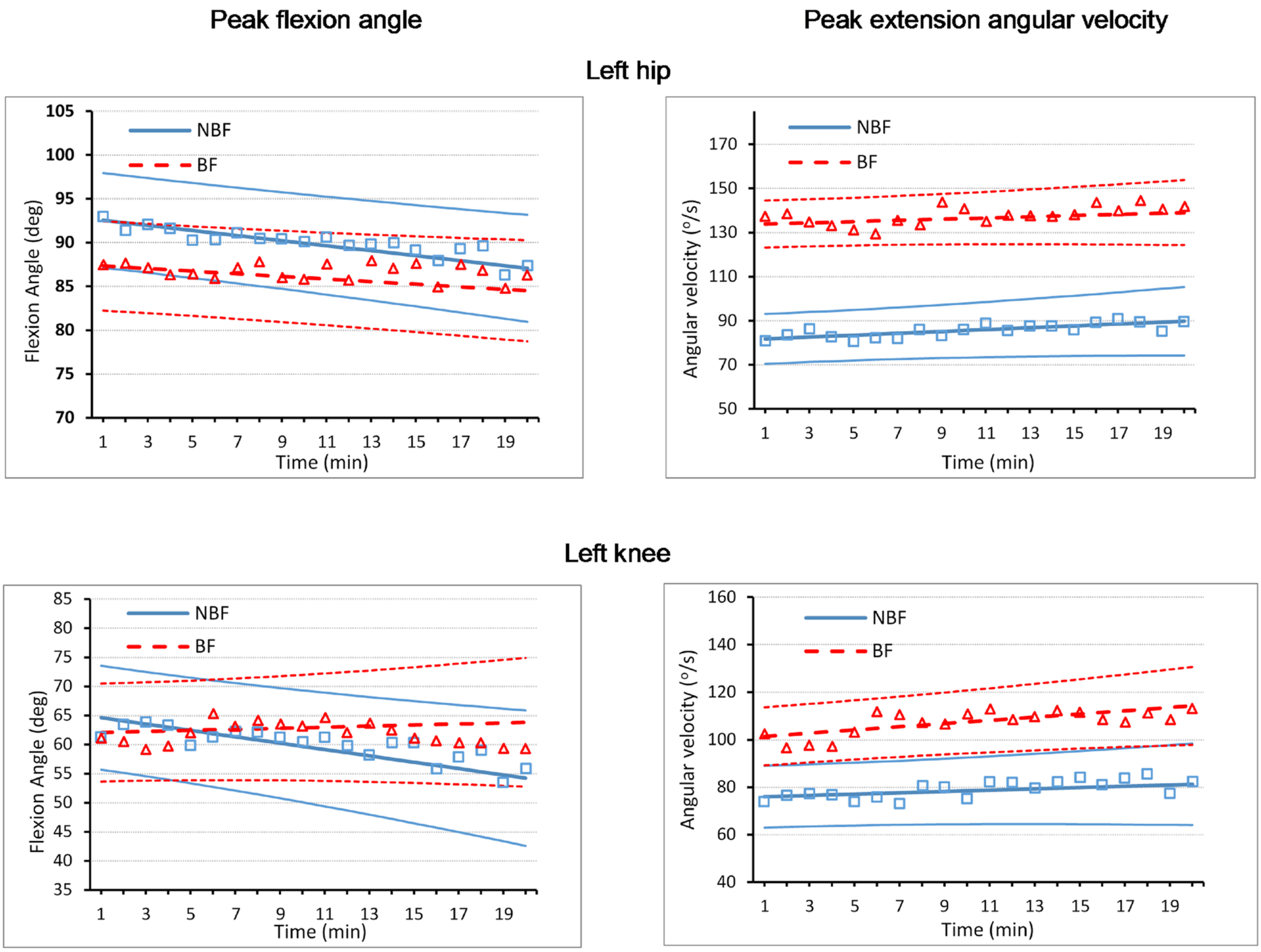


**Fig. 3.** Mean peak joint flexion angles (hip and knee) and extension angular velocities for the biofeedback (BF) and non-biofeedback (NBF) groups for each minute of the 20 min lifting. Lines indicate best-fit linear models and corresponding 95% confidence intervals.

### *3.3. Kinematics*

Linear models provided the best fit for the kinematic data.

#### *3.3.1. Lumbosacral posture*

%LSF differed significantly between groups, for both the intercepts ($P < 0.001$) and slopes ($P = 0.033$) of the fitted models (Fig. 2; Table 2). At the start of the task, the BF group flexed less than the NBF group. Both groups increased peak %LSF over time, but at a reduced rate in the BF compared to the NBF group (mean change = 18.7% and 26.7%, respectively). At 20 min, %LSF was significantly different between groups ($P < 0.001$).

For peak LS angular velocities (Table 2), there was a significant difference in slopes of the fitted models ($P = 0.005$), with the NBF group increasing angular velocities at a greater rate than the BF group (Fig. 2).Whilst at the start, LS angular velocities were higher in the BF group, intercepts of the fitted models were not significantly different ($P = 0.25$).

#### *3.3.2. Trunk posture*

For peak percentage trunk flexion (%TF), there was a significant difference in the intercept ($P < 0.001$) and slope ($P = 0.004$) of the fitted models, with the NBF group adopting greater flexion at the start and increased rate of change over time (Fig. 2).Whilst there were no significant differences in the intercepts of the fitted models for mean peak trunk angular velocities, slopes were significantly different ($P < 0.001$).

#### *3.3.3. Lower limb kinematics*

There was no significant difference between right and left lower leg kinematics, so only left hip and knee kinematics are presented (Fig. 3). There was no significant difference in mean peak hip flexion between groups. In contrast, intercepts for mean peak hip extension velocity was significantly different between groups ($P < 0.001$), as was hip velocity at 20 min ($P < 0.001$). However, slopes of the fitted models were not significantly different.

There was no significant difference in peak knee flexion at the start of the task. However, there was a significant differences in the slopes of the fitted models for peak knee flexion ($P = 0.03$), with the BF maintaining similar knee flexion throughout compared to a reduction in knee flexion in the NBF group. At the start, the BF group adopted significantly higher peak knee angular velocities than the NBF group ($P = 0.008$). No significant differences were found for the slopes of the fitted models.

### *3.4. Kinetics*

Quadratic models provided the best fit for the kinetic data (Fig. 4). Peak back moments were not significantly different at the start or end of the task, although there was a significant difference in the curves of the fitted models ($P = 0.007$).

As there was no significant difference between mean right and left

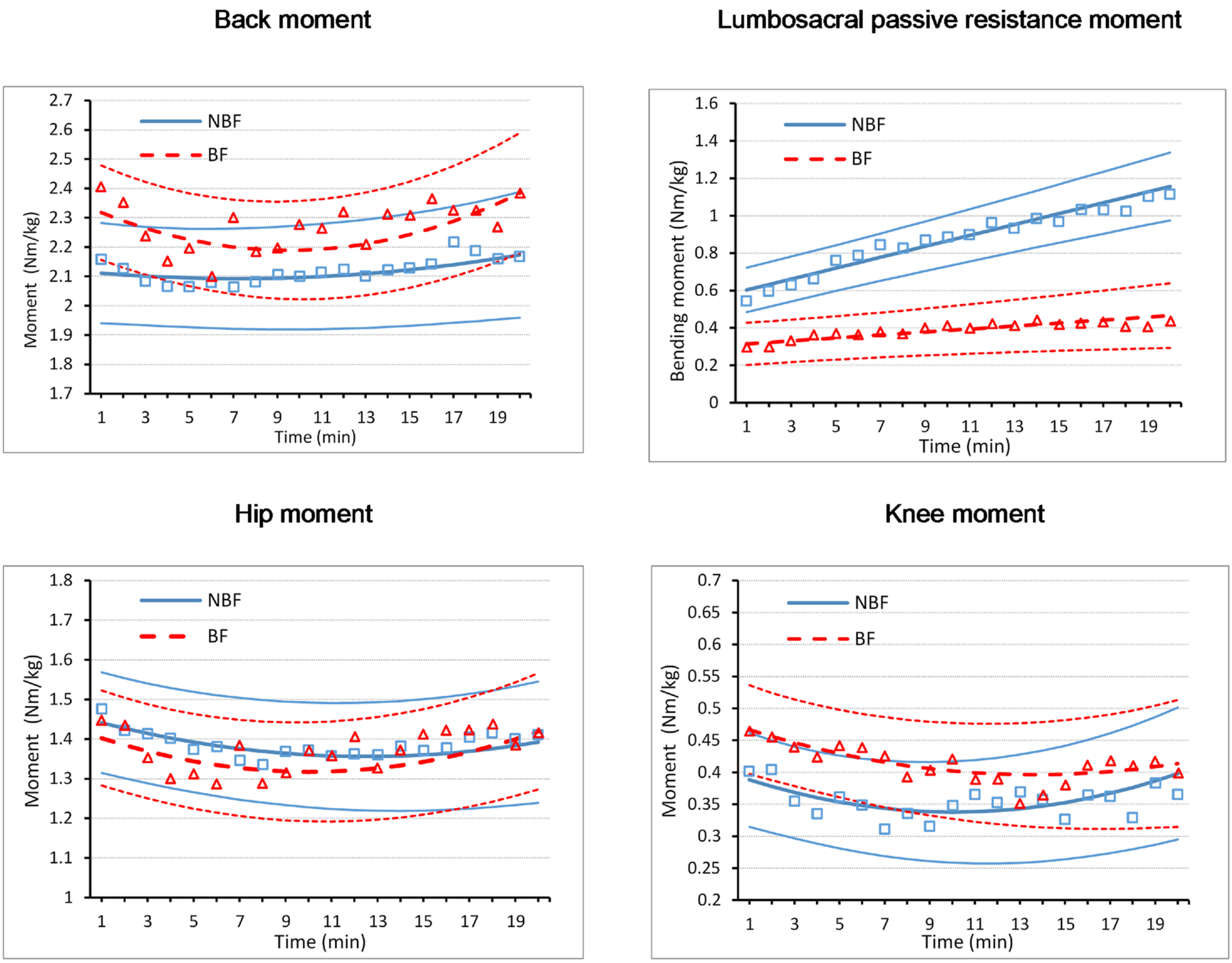


**Fig. 4.** Mean peak joint moments (back, lumbosacral passive resistance, hip and knee) for the biofeedback (BF) and non-biofeedback (NBF) groups for each minute during the 20 min lifting. Lines indicate best-fit quadratic (back, hip and knee) and linear (lumbosacral passive resistance) models, and corresponding 95% confidence intervals.

peak hip and knee moments, only left lower kinetics are reported (Fig. 4). There was no significant difference in the intercepts and curves of the fitted models for peak hip and knee moments.

### *3.4.1. Passive bending moment on the lumbar spine*

Linear models provided the best fit for the peak bending moments on passive spinal structures (Fig. 4). These were significantly higher at the start ($P < 0.001$) and increased at a greater rate (slope; $P < 0.001$) in the NBF group compared to the BF group.

## 4. Discussion

Participants provided with biofeedback on LS posture were able to control peak flexion below a prescribed threshold when repetitively lifting. They not only adopted LS flexion well below the prescribed threshold, but changes over time were at a significantly reduced rate, reaching approximately 64% of full flexion by the end of 20 min compared to almost full flexion (98%) in the NBF group. Beyond 80% LSF contribution to the extensor moment shifts markedly from the back extensor muscles to posterior passive spinal structures (e.g. lumbodorsal fascia, supraspinous ligament) [20], increasing the risk of ligament and vertebral end plate injury [27]. Our estimates of the bending moments resisted by the passive spinal structures was approximately two and half times greater at 20 min in the NBF group, compared to the BF group.

The changes in LS posture are consistent with a previous study investigating the effect of age on lifting strategies [18]. Older participants (mean age = 47 years) showed significantly reduced LS flexion (approximately 80%) at 20 min compared to a younger group (mean age = 24 years). Experience and greater postural awareness were considered potential moderating factors in limiting LSF in the older group. Given that young, inexperienced workers in the first year of employment appear to be at increased risk of low back pain [13], targeting manual handling interventions to these workers seems prudent.

In the NBF group, behavioural adaptation in response to repetitive lifting was consistent with other studies, showing a shift from predominantly ‘squat’ (bent knees, upright trunk) to ‘stoop’ lifting (extended knees, flexed trunk) [27,28]. This is likely due to fatigue and a move to a less physiologically demanding lifting technique (lower oxygen consumption) [29]. In contrast, the BF group maintained similar hip and knee flexion throughout. However, peak hip and knee angular velocities were higher during the lift, and the duration of the lift was shorter. Higher angular joint velocities have been associated with improved joint stability to protect against the negative physiological effects of fatigue [30]. Decreasing lift duration during repetitive lifting is consistent with other studies and suggestive of a strategy to increase rest periods between lifts [28].

The benefits of faster movements in increasing joint stability may also explain the increased change in LS angular velocities observed in the NBF compared to the BF group. Other studies have shown increasing trunk velocities [28] and increased co-contraction of abdominal muscles during repetitive lifting [31]. Although increased co-activity of trunk muscles may benefit stability, it may be detrimental in increasing biomechanical loads on the spine. Furthermore, faster extension movements have been found to be significant predictors of anterior shear forces [32] and an increased risk of back injuries in repetitive handling jobs [7]. Despite changes to LS and trunk posture, there was no evidence of differences in the magnitude of knee, hip or back moments between groups.

Feedback provided to participants diminished beyond 10 min, suggesting a learning effect that was not inhibited by fatigue. There was also a significant learning effect during the familiarisation period, with the BF group initially adopting significantly less LSF (22%) than the NBF group. Mawston et al. [21] demonstrated a similar learning effect and postural adaptations immediately following a single, prior exposure to sudden spinal loading.

When compared to the NBF group, measures of perceived exertion did not suggest that the BF group found the task more physical demanding. In fact, over the duration of the task the BF group rated the task as less physically demanding. This may have been due to participants associating overall perceptions of physical exertion with localised muscle fatigue. Studies have shown that increased spinal flexion decreases muscle thickness of the erector spinae [33]. This may elevate intramuscular pressure and decrease local blood flow, a suggested cause of low back pain [34]. Some caution is needed when interpreting these findings, as three of the BF group failed to complete the 20 min due to back discomfort.

The laboratory-based nature of this study limits the extrapolation of findings to the working environment. While only male participants took part in the study, the high predominance of male employees in heavy manual jobs makes the study relevant to a working population. Recruiting participants without prior lifting experience also seems pertinent as novice workers are considered at high risk of musculoskeletal injury [13]. Although no restrictions were placed on the lifting technique adopted, aspects of lifting task were constrained to control for potential confounders. The task was restricted to symmetrical lifting and participants had to adopt a stationary foot position and maintain hold on the box throughout. Sensors attached to the back may inhibit application of the method in some work situations, such as sitting. Further work should investigate the effects of LS biofeedback during complex lifting tasks (e.g. asymmetric) in realistic work environments.

## 5. Conclusion

The literature suggests that manual handling training alone is insufficient and should incorporate some measure of proficiency [35]. Providing biofeedback on LSF during repetitive lifting enabled participants to avoid end range of LSF and reduced loading on the passive spinal structures. Biofeedback of LS posture offers a potential adjunct to educate handlers when lifting. This could be particularly important for young, inexperienced workers employed in repetitive manual handling jobs who appear at increased risk of back injury. When presented with biofeedback on lumbar posture, a strategy adopted by participants involved increased knee and hip angular velocities.

## Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors

## Declaration of Competing Interest

The authors declare that they have no conflict of interest.

## References

[1] R.A. Deyo, D. Cherkin, D. Conrad, E. Volinn, Cost, controversy, crisis: low back pain and the health of the public, Annu. Rev. Public Health 12 (1) (1991) 141–156.
[2] D. Hoy, P. Brooks, F. Blyth, R. Buchbinder, The epidemiology of low back pain, Best Pract. Res. Clin. Rheumatol. 24 (6) (2010) 769–781.
[3] D. Hoy, C. Bain, G. Williams, L. March, P. Brooks, F. Blyth, et al., A systematic review of the global prevalence of low back pain, Arthritis Rheum. 64 (6) (2012) 2028–2037.
[4] J.N. Katz, Lumbar disc disorders and low-back pain: socioeconomic factors and consequences, J. Bone Jt. Surg. 88 (Suppl. 2) (2006) 21–24.
[5] J.L. Kelsey, P.B. Githens, A.A. White, T.R. Holford, S.D. Walter, T. O'Connor, et al., An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc, J. Orthop. Res. 2 (1) (1984) 61–66.
[6] M. Magnusson, M. Granqvist, R. Jonson, V. Lindell, U. Lundberg, L. Wallin, et al., The loads on the lumbar spine during work at an assembly line. The risks for fatigue injuries of vertebral bodies, Spine (Phila Pa 1976) 15 (8) (1990) 774–779.
[7] W.S. Marras, S.A. Lavender, S.E. Leurgans, S.L. Rajulu, W.G. Allread, F.A. Fathallah, et al., The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders. The effects of workplace factors, trunk position, and trunk motion characteristics on risk of injury, Spine (Phila Pa 1976) 18 (5) (1993) 617–628.
[8] W.S. Marras, S.A. Lavender, S.A. Ferguson, R.E. Splittstoesser, G. Yang, Quantitative dynamic measures of physical exposure predict low back functional impairment, Spine (Phila Pa 1976) 35 (8) (2010) 914–923.
[9] W.S. Marras, S.A. Lavender, S.E. Leurgans, F.A. Fathallah, S.A. Ferguson, W.G. Allread, et al., Biomechanical risk factors for occupationally related low back disorders, Ergonomics 38 (2) (1995) 377–410.
[10] L. Punnett, L.J. Fine, W.M. Keyserling, G.D. Herrin, D.B. Chaffin, Back disorders and nonneutral trunk postures of automobile assembly workers, Scand. J. Work Environ. Health (5) (1991) 337–346.
[11] A. Ramond-Roquin, J. Bodin, C. Serazin, E. Parot-Schinkel, C. Ha, I. Richard, et al., Biomechanical constraints remain major risk factors for low back pain. Results from a prospective cohort study in French male employees, Spine J. 15 (4) (2015) 559–569.
[12] S. Gallagher, W.S. Marras, A.S. Litsky, D. Burr, Torso flexion loads and the fatigue failure of human lumbosacral motion segments, Spine (Phila Pa 1976) 30 (20) (2005) 2265–2273.
[13] A. Van Nieuwenhuyse, L. Fatkhutdinova, G. Verbeke, D. Pirenne, K. Johannik, P.R. Somville, et al., Risk factors for first-ever low back pain among workers in their first employment, Occup. Med. 54 (8) (2004) 513–519.
[14] S.M. McGill, Low Back Disorders: Evidence-Based Prevention and Rehabilitation, 3rd ed., Human Kinetics, Champaign, IL, 2015.
[15] S.H. Roy, C.J. De Luca, D.A. Casavant, Lumbar muscle fatigue and chronic lower back pain, Spine (Phila Pa 1976) 14 (9) (1989) 992–1001.
[16] O.M. Giggins, U.M. Persson, B. Caulfield, Biofeedback in rehabilitation, J. Neuroeng. Rehabil. 10 (2013) 60.
[17] M. Parnianpour, M. Nordin, N. Kahanovitz, V. Frankel, Volvo award in biomechanics. The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements on the motor output and movement patterns, Spine (Phila Pa 1976) 13 (9) (1988) 982–992.
[18] M. Boocock, G.A. Mawston, S. Taylor, Age-related differences do affect postural kinematics and joint kinetics during repetitive lifting, Clin. Biomech. (Bristol, Avon) (2015) 30136–30143.
[19] E. Borg, L. Kaijser, A comparison between three rating scales for perceived exertion and two different work tests, Scand. J. Med. Sci. Sports 16 (1) (2006) 57–69.
[20] P. Dolan, A.F. Mannion, M.A. Adams, Passive tissues help the back muscles to generate extensor moments during lifting, J. Biomech. 27 (8) (1994) 1077–1085.
[21] G.A. Mawston, P.J. McNair, M. Boocock, The effects of prior exposure, warning, and initial standing posture on muscular and kinematic responses to sudden loading of a hand-held box, Clin. Biomech. (Bristol, Avon) 22 (3) (2007) 275–281.
[22] P. Dolan, M. Earley, M. Adams, Bending and compressive stresses acting on the lumbar spine during lifting activities, J. Biomech. 27 (10) (1994) 1237–1248.
[23] J.E.P. Hanavan, A personalized mathematical model of the human body, J. Spacecr. Rockets 3 (3) (1966) 446–448.
[24] D.A. Winter, Biomechanics and Motor Control of Human Movement, Wiley, New York, 2005.
[25] R. Core Team, R: A Language and Environment for Statistical Computing, (2016) (Accessed), http://www.R-project.org/.
[26] D. Bates, M. Mächler, B. Bolker, S. Walker, Fitting linear mixed-effects models using lme4, J. Stat. Softw. 67 (1) (2015).
[27] P. Dolan, M.A. Adams, Repetitive lifting tasks fatigue the back muscles and increase the bending moment acting on the lumbar spine, J. Biomech. 31 (8) (1998) 713–721.
[28] J.P. Mehta, S.A. Lavender, R.J. Jagacinski, Physiological and biomechanical responses to a prolonged repetitive asymmetric lifting activity, Ergonomics 57 (4) (2014) 575–588.
[29] S. Kumar, The physiological cost of three different methods of lifting in sagittal and lateral planes, Ergonomics 27 (4) (1984) 425–433.
[30] P.L. Silva, S.T. Fonseca, J.M. Ocarino, G.P. Goncalves, M.C. Mancini, Contributions of cocontraction and eccentric activity to stiffness regulation, J. Mot. Behav. 41 (3) (2009) 207–218.
[31] W.S. Marras, G.A. Mirka, Electromyographic studies of the lumbar trunk musculature during the generation of low-level trunk acceleration, J. Orthop. Res. 11 (6) (1993) 811–817.

[32] J. Alderson, L. Hopper, B. Elliott, T. Ackland, Risk factors for lower back injury in male dancers performing ballet lifts, J. Dance Med. Sci. 13 (3) (2009) 83–89.
[33] K. Watanabe, K. Miyamoto, T. Masuda, K. Shimizu, Use of ultrasonography to evaluate thickness of the erector spinae muscle in maximum flexion and extension of the lumbar spine, Spine (Phila Pa 1976) 29 (13) (2004) 1472–1477.
[34] S. Konno, S. Kikuchi, Y. Nagaosa, The relationship between intramuscular pressure of the paraspinal muscles and low back pain, Spine (Phila Pa 1976) 19 (19) (1994) 2186–2189.
[35] S. Lavender, E.P. Lorenz, G.B. Andersson, Can a new behaviorally oriented training process to improve lifting technique prevent occupationally related back injuries due to lifting? Spine (Phila Pa 1976) 32 (4) (2007) 487–494.