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Center of pressure trajectories, trunk kinematics and trunk muscle activation during unstable sitting in low back pain patients

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**Abstract**

Trunk motor behavior has been reported to be altered in low-back pain. This may be associated with impaired lumbar proprioception, which could be compensated by trunk stiffening. We assessed trunk control by measuring center-of-pressure, lumbar kinematics and trunk muscle electromyography in 20 low-back pain patients and 11 healthy individuals during a seated balancing task, in conditions with and without disturbance of lumbar proprioception and occlusion of vision. We hypothesized that low-back pain patients show larger postural sway, but smaller thoraco-lumbar movements than healthy individuals. Repeated measures analyses of variance indicated that the effects of proprioceptive disturbance and vision occlusion were similar between groups. Interestingly, low-back pain patients grabbed the safety rail more often, while differences between groups in sway measures were rather subtle. This suggests that low-back pain patients were more cautious. Furthermore, low-back pain patients had an about 20 degrees less flexed lumbar posture than healthy individuals, and, in contrast to our hypothesis, made larger thoraco-lumbar movements in the sagittal plane, as indicated by higher SDs of thoraco-lumbar flexion and lower (more negative) correlations between pelvis and thorax movements. Activation of the intersegmental longissimus relative to the iliocostalis muscle, which spans all lumbar segments, was lower in low-back pain patients compared to healthy individuals. This difference in muscle activation may be causal for larger thoraco-lumbar movements, and may be causative of reduced control over segmental lumbar movement, but may also reflect the need for larger corrective movements to compensate balance impairments.

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1. Introduction

Differences in trunk motor behavior between low-back pain (LBP) patients and healthy control (HC) subjects have been reported in upright standing [1,2], walking [3,4] and sitting [5,6]. Findings from studies with lumbar muscle vibration, which is known to perturb proprioceptive feedback from muscle spindles [7], suggest that these differences in motor control could in part be explained by impaired proprioception in LBP-patients [8]. Consequently, to compensate for proprioceptive deficits, LBP-patients may use trunk muscle activation strategies aimed at trunk stiffening, in order to protect the painful area [9].

Seated balancing, i.e. balancing on a chair with the lower extremities supported and a hemisphere under the seat, allows studying trunk control in an implicit, challenging and natural way, while avoiding compensation by knee and ankle motion. Previous studies compared LBP-patients and HC-subjects during this task in terms of either center-of-pressure (CoP) trajectories [10,11] or trunk kinematics [12]. Radebold et al. found that balance performance in LBP-patients was lower than in HC-subjects, especially in more challenging conditions [10], while Van Dieën et al. found larger CoP-amplitudes in subjects with a recent history of LBP, but not in subjects with current LBP [11]. The latter group demonstrated lower CoP-frequencies which was in line with earlier suggestions that LBP-patients stiffen their lumbar spine [9,13,14]. However, while such a trunk stiffening strategy would result in smaller movements of the spine, Van Daele et al. reported larger pelvis and trunk movements in LBP-patients [12]. Regrettably the authors did not report lumbar spine (i.e. trunk relative to pelvis) motion. So, although seated balancing seems a convenient task for studying trunk control [15], and published data point at impaired seated balance in patients, the exact nature of differences between patients and controls, and more specifically the question whether or not patients employ a stiffening strategy, remain to be elucidated.

Therefore, our goal was to evaluate CoP-trajectories, trunk kinematics and trunk muscle activation during seated balancing in LBP-patients. In addition, we evaluated the effects of eliminating
visual information and disturbing proprioception on postural balance. We hypothesized that LBP-patients show larger CoP-movements, coinciding with larger trunk movements, but smaller movements of the lumbar spine. In line with this, LBP-patients were hypothesized to show larger ratios of longissimus over iliocostalis, and lumbar over thoracic longissimus muscle electromyography (EMG) amplitudes, since larger ratios would reflect lumbar spine stiffening strategies. This assumption is based on differences in anatomical characteristics between muscles, i.e. the number of spinal segments crossed is larger for the iliocostalis than for the longissimus muscle and larger for the thoracic compared to the lumbar part of the longissimus muscle [16]. Model calculations indicate that preferential recruitment of muscles with multiple intersegmental insertions, over their synergist that cross more spinal segments, leads to a higher stiffness of the lumbar area [14,17,18]. Based on indications of impaired proprioception in LBP-patients, we further hypothesized that LBP-patients show larger deterioration of balance when vision is occluded and that disturbance of proprioception, through lumbar muscle vibration, degrades balance performance more in HC-subjects.

2. Methods

2.1. Participants

Twenty LBP-patients (9 female) and 11 HC-subjects (4 female) participated in the experiment. Subjects in the LBP-group had experienced LBP during the last 6 weeks or longer, and any specific diagnosis had been excluded by a general practitioner or physical therapist. Subjects were excluded when they had had previous surgery on the spine or scored >105 on a questionnaire identifying psychosocial risk factors [19,20]. Subjects participating in the HC-group did not experience LBP during the previous year. No differences between groups were found in age (HC: 32.6 ± 10.4, LBP: 33.4 ± 15.5 years, p = 0.893) and BMI (HC: 22.5 ± 2.5, LBP: 23.6 ± 3.0 kg/m², p = 0.312). LBP-patients scored 2.7 ± 1.7 on a 10 cm visual analog pain scale at the start of the measurements. The experimental protocol was approved by the local medical ethical committee and all subjects provided informed consent.

2.2. Experimental setup

An aluminum hemisphere (radius: 25 cm) was attached underneath a seat, creating instability in all directions (height of the seat relative to the lowest point of the hemisphere: 17 cm). An adjustable footplate was attached to the seat, in order to limit the influence of the lower extremities to balance control by keeping knee and ankle angles fixed at 90° (Fig. 1). Three force transducers (KAP-E, AST, Germany) recorded vertical forces with 200 samples/s. A safety rail surrounded the seat, to provide security in case of balance loss. A pulse-signal was generated when subjects touched the rail.

Trunk kinematics were measured by opto-electronic markers (Optotak 3020, Northern Digital Inc., Canada) on the spinous processes of the T1, T7, L4 and L5 vertebrae at a rate of 100 samples/s. Trunk muscle surface EMG was used to record activation of four back muscles bilaterally (Porti 17, TMS-Enschede, The Netherlands, 22-bits AD-conversion after 20X amplification, input impedance >10¹² Ω, CMRR > 90 dB). The skin was shaved and cleaned with alcohol. Based on a detailed anatomical study [16], we placed bipolar electrodes (Ag/AgCl) 4 cm lateral to T9, 6 cm lateral to T11 and L2, and 3 cm lateral to the midpoint between the spinous processes of L3 and L4, reflecting activation of the thoracic longissimus and iliocostalis muscles, and the lumbar iliocostalis and longissimus muscles, respectively. EMG-signals were recorded at a rate of 1000 samples/s and a pulse-signal synchronized the EMG-recordings to the opto-electronic and force-plate data.

To apply lumbar muscle vibration, we used a motor (Maxon Graphite Brushes S2326.946 driven by a 4-Q-DC Servo Control LSC30/2 in a velocity loop) rotating an eccentric mass. Vibration frequency was 90 Hz, and the vibration device was attached at the level of L3/4 by neoprene bands (Fig. 2).
3. Data analysis

Data were analyzed in Matlab 7.10.0 (R2010a) and the two repetitions of each condition were averaged. Details on data analysis are described below.

3.1. CoP trajectories

A running average over 2 data points down-sampled the vertical forces to 100 samples/s and forces were 8 Hz low-pass filtered (2nd order bi-directional Butterworth) before CoP-trajectories were calculated. Root-mean-square differences of the CoP with respect to its mean in the frontal (RMSx) and sagittal (RMSy) planes quantified sway amplitude. In addition, mean velocity of the resultant of the CoP-excursions in x- and y-direction (meanV) was calculated, since it is a highly reliable and sensitive parameter in this task [21]. Spectral content (Welch’s periodogram method, windowsize 10s with 9s overlap) of the CoP-trajectories was quantified via mean power frequencies in the frontal (MPFx) and sagittal (MPFy) planes. Finally, short-term diffusion coefficients (DSx and DSy) were calculated to evaluate the level of stochastic control of trunk posture [21,22].

3.2. Thoraco-lumbar kinematics

Since the vibration device covered some opto-electronic markers, thoraco-lumbar kinematics were only calculated for the REF and ECL-conditions. Opto-electronic data were 2.5 Hz low-pass filtered (2nd order bi-directional Butterworth). Lines through the T1−T7 and through the L4−L5 spinous processes reflected the high-thoracic and low-lumbar angle, respectively. Means and SDs of their difference, reflecting thoraco-lumbar flexion, as well as the correlation between high-thoracic and low-lumbar angles over time series, were calculated in the frontal and sagittal planes.

3.3. Trunk muscle activation

We applied eight repetitive band-stop filters to remove 50 Hz interference and its harmonics up to 400 Hz (4th order bi-directional Butterworth; target frequency ± 0.5 Hz). Subsequently, EMG-signals were 30 Hz high-pass filtered (2nd order bi-directional Butterworth), to reduce contamination by the electrocardiogram, and absolute Hilbert amplitudes were calculated. VIB-conditions were discarded from this analysis since vibration affected the EMG-signals. In four LBP-patients, left lumbar iliocostalis signals were discarded from analysis due to technical malfunctioning. After averaging over left and right muscles, ratios of longissimus over iliocostalis (long/ilioc) and of lumbar over thoracic (lumb/thor) longissimus muscle EMG-amplitudes were calculated, with larger ratios reflecting activation aimed at spinal stiffening [14,17,18].

4. Results

4.1. CoP-trajectories

Strikingly, six LBP-patients grabbed the safety rail 2.2 ± 1.6 (range 1–5) times, all during the ECL-condition, while only one HC-subject grabbed the safety rail once, during the REF-condition. Because subjects continued the task immediately after any balance loss, and to prevent introducing a bias between groups by excluding trials with large CoP-excursions, both trials with and without balance loss were included in the analysis.

RMS, MPF and DS are shown in Fig. 3. No effect of group was found on RMS and MPF (all $p \geq 0.312$). Larger diffusion coefficients, indicating more ‘random’ control, were found in LBP-patients, with this difference between groups reaching significance in the frontal ($p = 0.044$), but not in the sagittal plane ($p = 0.068$). MeanV tended to be about 20% larger in LBP-patients (1.15 ± 0.34 cm/s vs. 0.94 ± 0.17 cm/s in HC-subjects), but this effect did not reach significance ($p = 0.069$). Significant main effects of condition were found on meanV and on RMS and DS in both directions, but not on MPF. Pairwise comparisons revealed a 12% increase in meanV due to lumbar muscle vibration (REF: 0.78 ± 0.24 cm/s vs. VIB: 0.87 ± 0.27 cm/s, $p = 0.003$). Significant effects of vibration were also found on DS (both $p < 0.026$), but not on RMS (both $p > 0.376$). In the ECL-condition, meanV approximately doubled to 1.57 ± 0.50 cm/s and RMS and DS were significantly larger in both directions (all $p < 0.001$). No significant interactions between group and condition were found on any COP-measure (all $p > 0.086$).

4.2. Thoraco-lumbar kinematics

Kinematics data from two LBP-patients were not available, so these results are based on 18 LBP-patients. Since no significant differences between groups were found in the frontal plane (all $p > 0.351$), Fig. 4 shows thoraco-lumbar kinematics in the sagittal plane only. Trunk posture was about 20° more upright in LBP-patients than in HC-subjects ($p = 0.007$). In addition, and in contrast with our hypotheses, LBP-patients demonstrated larger SDs of thoraco-lumbar flexion ($p = 0.005$), coinciding with a lower (more negative) correlation between high-thoracic and low-lumbar segment angles ($p = 0.018$) compared to HC-subjects, implying that thoraco-lumbar movements were larger in LBP-patients.
Vision occlusion resulted in an increase of thoraco-lumbar movements \((p < 0.001)\), while the average thoraco-lumbar flexion and the correlation between low-lumbar and high-thoracic angles were unaffected \((all \ p \geq 0.215)\). No significant interactions between group and vision condition were found on thoraco-lumbar kinematics \((all \ p \geq 0.093)\).

### 4.3. EMG ratios

In contrast with our hypothesis, the long/iliac ratio (Fig. 5, left panel) was significantly lower in LBP-patients compared to HC-subjects \((p = 0.004)\). The lumb/thor ratio (Fig. 5, right panel) appeared somewhat higher in LBP-patients, but this difference did not reach significance \((p = 0.177)\). No significant main effects of vision occlusion and no significant interactions between group and vision occlusion were found on EMG-ratios \((all \ p \geq 0.101)\).

### 5. Discussion

We compared CoP-trajectories, thoraco-lumbar kinematics and EMG-ratios between LBP-patients and HC-subjects during a seated balancing task. The differences between groups were most

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**Fig. 3.** Seated balancing COP-amplitude (upper panel), frequency (middle panel) and level of stochastic control (lower panel) in both planes of motion in the reference (REF), vibration (VIB) and eyes closed (ECL) conditions for healthy control (HC) and low back pain (LBP) groups. Error bars represent SDs over subjects within groups.

**Fig. 4.** Thoraco-lumbar kinematics in the sagittal plane for both groups in the reference and eyes closed conditions. Error bars represent SDs between subjects within groups.
prominent in thoraco-lumbar kinematics, with, in contrast to our hypothesis, larger thoraco-lumbar movements in LBP-patients. Moreover, LBP-patients demonstrated more upright trunk postures than HC-subjects. Surprisingly, the ratio of longissimus over iliocostalis muscle activation was lower in the LBP-group. Since the effects of lumbar muscle vibration and vision occlusion were similar in both groups, differences in task performance could not be attributed to differences in weighting of these information sources. Instead, differences in trunk muscle recruitment may explain the current findings.

The long/ilioc and lumb/thor EMG-ratios were hypothesized to be higher in LBP-patients under the assumption that these ratios would reflect trunk stiffening strategies. The finding that neither long/ilioc nor lumb/thor longissimus over iliocostalis ratio was higher with LBP, together with the findings that sway frequency was not lower and thoraco-lumbar movements were not smaller in LBP-patients compared to HC-subjects, suggests that LBP-patients did not stiffen their spine during unstable sitting. Surprisingly, the long/ilioc ratio was in fact smaller and thoraco-lumbar movements were larger in LBP-subjects. Preferential recruitment of the longissimus muscle, which has multiple segmental insertions, may allow for subtle adjustments of lumbar curvature, while increased activation of the iliocostalis muscle, which spans more lumbar segments and has a larger moment arm, would cause larger thoraco-lumbar movements. While this interpretation, implying more subtle control in HC-subjects than in LBP-subjects, could explain the current findings, future studies will be needed to elucidate the causality of these differences in trunk muscle recruitment and LBP. Specifically, we do not know whether this activation pattern caused larger movements, or was a response to the necessity to make larger movements as a consequence of other deficits, e.g. delayed responses [10].

The rather subtle differences between groups in terms of CoP-trajectories could not fully explain the finding that LBP-patients grabbed the safety rail more often that the HC-subjects. Since sway amplitude did not differ between groups, we think that LBP-patients were more cautious and grabbed the safety rail quicker to prevent expected balance loss, while the HC-subjects took more risk in maintaining balance without touching the safety rail.

We found a lower (more negative) correlation between thorax and pelvis movements in LBP-patients compared to HC-subjects, which appears to be in contrast with previous findings [12]. However, although Van Daele et al. interpreted their between-subjects correlation of trunk and pelvis angular deviations as support for the trunk stiffening hypothesis [12], the high correlation only indicates that subjects with large angular deviations of the pelvis also demonstrated large angular deviations of the thorax. We calculated correlations per subject in the time domain, which, in accordance with the larger SD of thoraco-lumbar angle, was lower in LBP-patients compared to HC-subjects, indicative of more countermovement of the thorax and pelvis segments in the patients.

In contrast to previous findings on trunk control in other experimental tasks [8,23], no reduced effect of lumbar muscle vibration was found in the LBP-group. Moreover, paraspinal muscle vibration did not affect sway amplitude or frequency. These findings, combined with the large effect of closing the eyes, suggest that certainly visual and possibly vestibular information were dominant in this seated balancing task. This supports a recent finding on standing on an unstable surface, that the relative importance of proprioception may be reduced when no direct relation exists between proprioceptive feedback and orientation of the body relative to the gravitational field [24].

We did not normalize EMG to percentages of (sub)maximal voluntary contractions, since this may introduce a bias between groups [9]. We are confident that the reported ratios of EMG-amplitudes provide robust measures of trunk muscle recruitment. Unfortunately, we cannot exclude that the difference in mean thoraco-lumbar flexion between the groups may partly explain the differences in the other outcome measures. A future study with specific instructions on trunk posture could elucidate the relation between thoraco-lumbar flexion and balancing behavior. Furthermore, no intramuscular EMG was recorded, so activation of deep muscle fibers was only partly detected [25,26]. Although we therefore cannot exclude that LBP-patients used a stiffening strategy based on recruitment of deep muscles, this would not fit with the group effects found on kinematics. Moreover, reported changes in activity of deep muscles in LBP-patients [27,28] indicate inhibition rather than increased activity of these muscles. One might speculate that such inhibition could underlie the larger amplitudes of trunk movements due to insufficient stabilization of the lumbar spine. Future studies could measure intramuscular EMG during seated balancing to evaluate the contribution of deep intersegmental muscle fibers.

In conclusion, LBP-patients demonstrated trunk control that differed from HC-subjects during a seated balancing task. While being more cautious can explain why LBP-patients grabbed the safety rail more often, LBP-patients also showed more upright trunk postures and larger thoraco-lumbar movements compared to HC-subjects. Furthermore, LBP-patients showed lower activation of the intersegmental longissimus muscle relative to the iliocostalis muscle, which spans all lumbar segments.

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Conflict of interest
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