The effects of upper limb loading on spinal shrinkage during treadmill walking

H. Watson · A. Simpson · P. E. Riches

Received: 11 December 2011 / Revised: 30 April 2012 / Accepted: 10 May 2012 / Published online: 29 May 2012
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Abstract

Introduction Everyday activities such as walking may elicit spinal shrinkage in an order of magnitude that has been related to lower back pain. The present study aims to compare the effects of unloaded treadmill walking with walking carrying loads representing everyday shopping tasks.

Materials and methods Walking tasks were performed on seven healthy males and motion analysis was used to track four reflective markers at 100 Hz, dividing the spine into three segments. Static data was collected in 5-min intervals over a 30-min period.

Results Total spinal length and lumbar segment decreased with respect to time ($p < 0.001$). Load affected the percentage length change at each spinal segment ($p < 0.005$), with the lumbar segment showing greatest height loss at the highest load. The upper and lower thoracic segments showed greater anterior lean with the heavier loads ($p = 0.000$) and the lumbar segment showed the opposite trend ($p = 0.000$).

Conclusion Results suggest that the body adopts less anterior lean with an immediate load-bearing demand, to decrease the necessary extension moment generated by the spinal extensors for spinal stability. Further postural alteration in the same direction is observed with prolonged loading. In combination with lumbar spinal shrinkage, such

postural changes are likely to increase the loading on the facet joints and subsequently unload the discs which may be beneficial for those with low back pain.

Keywords Spine · Biomechanics · Compression · Walking · Low back pain

Introduction

Low back pain (LBP) is a cause of major discomfort for large population numbers in Western society, with point prevalence rates in the range of 12–35 % and lifetime prevalence rates of 49–80 % [1]. LBP can lead to disability [2], unemployment and economic loss [1], and may also impact on social and lifestyle activities, affecting the life quality of the sufferers [3].

The mechanical risk factors relating to LBP are poorly understood; however, there are indications that bending and twisting of the spine, heavy load carriage, sudden forceful movements and whole body vibrations have strong relationships with LBP [4–8]. However, there is a lack of literature relating these daily functional activities with the mechanisms underlying LBP.

Spinal shrinkage during walking activities has been investigated in relation to LBP [9–13]. In these studies, between 3 and 12 mm of spinal shrinkage has been obtained using stadiometry, depending on the walk time, participant group, those with and without low back pain and the nature of the loading conditions. Even gentle land walking alone has caused significant shrinkage in young healthy males [14].

The shrinkage that occurs during walking and load carriage is part of the normal diurnal height change where approximately 1 % of total stature loss occurs throughout the day [15, 16]. These losses are predominantly a result of

H. Watson (✉) · A. Simpson
Health and Exercise Sciences Research Unit,
University of the West of Scotland, Hamilton ML3 0JB, UK
e-mail: hugh.watson@uws.ac.uk

A. Simpson
e-mail: alan.simpson@uws.ac.uk

P. E. Riches
Bioengineering Unit, University of Strathclyde,
Glasgow G4 0NW, UK
e-mail: philip.riches@strath.ac.uk

height reductions in the intervertebral discs. This occurs through lateral bulging of the annulus fibrosus, contributing to the initial rapid height losses [12] and subsequent fluid loss from the nucleus pulposus [17, 18], contributing to the later, slower height reductions [12].

Reducing the height of the intervertebral discs increases or causes abnormal loading on the zygapophysial joints and spinal ligaments, [19, 20] which has been associated with LBP [21]. Therefore, continuous spinal loading, which may occur during everyday activities such as shopping, maybe a contributory factor to the development of LBP [11, 22].

Consequently, this study investigates whether spinal shrinkage in the sagittal plane in normal healthy males is greater when walking carrying everyday loads equivalent to 7.5 and 15 % body weight than unloaded walking. The study will also investigate the contribution of the elastic components of the motion segments by assessing spinal length when loaded and unloaded at specific time intervals. It is hypothesized that walking carrying greater loads will produce greater spinal shrinkage than walking unloaded. To test the hypothesis, motion analysis of the spine in the sagittal plane, as described by Pollock et al. [23], is used to assess spinal shrinkage.

Materials and methods

Following approval from the University of Strathclyde Ethics Committee, seven male participants without previous spinal injury, height 175.6 ± 5.2 cm, mass 72.0 ± 5.1 kg and age 23.9 ± 5.0 years (mean $\pm$ SD) volunteered.

Participants attended the laboratory on three separate occasions at weekly intervals, at the same time of day. On day 1 they walked unloaded, acting as the control and on days 2 and 3 carrying randomly assigned loads, in a similar fashion to shopping bags, equivalent to 7.5 and 15 % body weight, respectively, equally distributed across both arms. Participants refrained from physical activity for the 48 h preceding each laboratory visit. On the first occasion, normal walking speed was calculated from the average time to walk a 6-m distance. Before each trial participants were required to lie in a supine position for 2 h to standardise the baseline condition of the spine. After 1 h 45 min, the participants changed into black Lycra™ shorts and height and weight were recorded using a standard stadiometer (Seca 220, Germany) and scales (Seca 710 1021009, Germany).

Four 14-mm reflective markers were adhered directly to the skin, in the sagittal plane of the spine, as described previously [23], overlying the spinous process of C7, the region of greatest convexity of the thoracic spine, the greatest concavity of the lumbar spine and midway between the posterior superior iliac spines. These markers were used to define three spinal segments approximately

as, upper thoracic (UT), lower thoracic (LT) and lumbar (LU). On completion of the marker attachment, the participants lay down for the remaining 15 min to complete the standardisation.

Static 3D marker coordinate data were acquired using a six-camera motion analysis system (ViconPeak, Oxford, UK). The capture volume around the treadmill (HP Cosmos Pulsar 4, Germany) was calibrated with the *x*-axis in the direction of walking and the *z* axis vertically upward, with the standard deviation of the measurement of a marker less than 1 mm.

Participants initially stood on the treadmill, where two static measurements were taken 1 min apart, one loaded, one unloaded. They subsequently walked in 5-min intervals at their predetermined walking speed. After each time interval, the two measurements were repeated 1 min apart to assess elasticity in the musculoskeletal system. All data were captured at 100 Hz at time intervals of 0, 5, 10, 15, 20, 25 and 30 min. On completion of the experiment each participant's height was rerecorded using the stadiometer.

Data analysis

Raw coordinate data were converted into ASCII file and exported into Microsoft Excel 2007, where total spine length (C7 to approximately L5) and individual segment lengths were determined using standard Pythagorean formulae. All lengths were expressed as a percentage change in length from original length ($t = 0$). Lean was defined as the angle made between the segment of interest and the vertical (Fig. 1). Total spine lean and individual segment lean were calculated and compared with the unloaded lean

Fig. 1 Segment lengths, total spine length, anterior lean and curvature

at $t = 0$, as was spinal curvature through the cervicothoracic and thoracolumbar regions, which were defined as the angle between the segments.

Repeated measures ANOVA analyses (Minitab Version 15.0) were used to assess if changes in total spine, spine segments, total spine lean, segment lean or curvature varied with respect to time, load or their interaction, with significance taken if $p < 0.05$. Post hoc analysis using Tukey's pairwise comparisons were performed to compare between time intervals within loads and between loads at the same time interval where appropriate.

Results

The percentage change in total spine length and LU segment length showed a consistent height loss ($p < 0.001$) with respect to increasing time across all loading conditions, reaching a mean $\pm$ SE maximum height loss of $0.79 \% \pm 0.13$ and $1.36 \% \pm 0.40$, respectively, at the 30-min time interval (Figs. 2, 3).

Fig. 2 Pooled mean percentage change ($\pm$ SE) in spinal length with time

Fig. 3 Pooled mean percentage change ($\pm$ SE) in LU segment length with time

However, the UT segment height losses were inconsistent across time ($p < 0.01$) and the LT segment failed to show any height loss ($p = 0.641$).

Load affected the percentage length change at each spinal segment ($p < 0.005$); however, the LU segment showed the greatest height loss at the highest load, $1.43 \% \pm 0.23$ compared with the lesser loads ($p < 0.001$), but the UT and LT segments showed the opposite trend ($p = 0.001$ and 0.004 , respectively) (Fig. 4). The inter-segmental variation in length change resulted in the non-significant effect of load on the percentage length change of the total spine ($p = 0.263$).

Load had a significant effect on anterior lean of the total spine ($p = 0.037$), with the two heavier loads causing greater anterior lean, $4.19^\circ \pm 0.25$ and $4.33^\circ \pm 0.21$, respectively, compared with $3.88^\circ \pm 0.24$ in the unloaded condition. Spinal shape analysis also reveals that during the loading and unloading conditions at individual time intervals the anterior lean is less during loading and with respect to time (Fig. 5).

Fig. 4 Percentage change ($\pm$ SE) in segment length with respect to load

Fig. 5 Average spinal shape with and without 15 % body weight load (relative to L5 marker)

Load also affected UT and LT segmental lean ($p = 0.000$) with the heavier loads causing greater anterior lean (or less posterior lean where the spine is in lordosis, i.e., the LT segment) than the unloaded condition, $18.92^\circ \pm 0.52$ and $17.67^\circ \pm 0.34$, respectively, compared with $16.48^\circ \pm 0.39$ in the unloaded condition for the upper thoracic segment and $-9.52^\circ \pm 0.28$ and $-10.01^\circ \pm 0.32$, respectively, compared with $-10.61^\circ \pm 0.39$ in the unloaded condition for the lower thoracic segment. In the UT segment the 7.5 % load caused greater anterior lean than the 15 % load ($p < 0.001$). In the LT segment there was no difference in anterior lean between the two heavier loads ($p = 1.000$).

Load also affected lumbar anterior lean ($p = 0.001$), with heavier loads tending to show less anterior lean than the control, $2.82^\circ \pm 0.66$ and $3.40^\circ \pm 0.81$, respectively, compared with $4.73^\circ \pm 0.66$ in the unloaded condition. However, there was no difference in anterior lean between the two heavier loads ($p > 0.5$).

Load had a main effect on thoracolumbar curvature ($p = 0.000$), with 7.5 and 15 % loads both causing less curvature $167.65^\circ \pm 0.83$ and $166.59^\circ \pm 0.98$ than the unloaded condition $164.66^\circ \pm 0.97$ ($p = 0.000$ and 0.002 , respectively). There was no difference in curvature between the two heavier loads ($p = 0.052$).

Load also had a main effect on cervicothoracic curvature ($p = 0.000$), with 7.5 % load causing more curvature than the unloaded condition ($p = 0.000$) $151.56^\circ \pm 0.64$ and $152.91^\circ \pm 0.63$, respectively; however, the 15 % load did not reach significance ($p = 0.056$) $152.32^\circ \pm 0.48$. The 15 % load produced less curvature than 7.5 % load ($p = 0.012$).

Discussion

The load carriage mechanisms of the spine are complicated, with spinal shrinkage and postural changes evident in this study, but dependent on spinal level. The percentage change in total spine length and the LU segment showed a consistent height loss with respect to time. The UT segment also displayed height losses but these were not consistent with time. However, the observed height losses are in agreement with previous walking studies [9–13, 15, 16]. The variations in segment and total spine lengths may be partly explained by differences in anterior lean and curvature to the various loading conditions.

The effect of load on the individual spinal segments was inconsistent with the LU segment showing greatest height loss at the heaviest load and the UT and LT segments showing the opposite trend. This inconsistency maybe partially explained by the effect of heavier loads causing greater anterior lean than the unloaded condition in the UT

and LT segments. Anterior lean causes the vertebral spinous processes and hence the overlying skin markers to move apart, negating to an extent, the skin markers moving together as a result of stature loss. In the LU segment, the heavier loads caused less anterior lean and therefore would have less effect on the measured shrinkage.

These differences in anterior lean and height losses of the segments resulted in the non-significant effect of load on the percentage change in length of the total spine. This may have been confounded further by changes in spinal curvature, when carrying the heavier loads. Load carriage not only causes height losses through compression of the intervertebral discs, but also through increases in spinal curvature [10, 24]. This may have relevance to LBP, evidenced by three-dimensional modelling, which has suggested that greatest pressure occurs on the fifth lumbar vertebra during forward flexion [25].

In the current study, both of the heavier loads caused less curvature than the control in the thoracolumbar region and suggest height losses, in particular in the LU segment, are due to losses from the intervertebral discs. Considering that anterior lean decreased with heavier loads in the LU segment, this would strengthen the authors' suggestion that height losses were due to changes in the discs in the LU region. The reduction of anterior lean in this segment maybe a result of purposeful spine straightening to lessen muscular activity in the lumbar spine, to better balance the load through the intervertebral discs.

In the cervicothoracic curvature, the 7.5 % load caused greater curvature than the control and the 15 % load did not. Considering the significant effect of anterior lean on the UT and LT segments has led to inconsistent height losses with respect to loading conditions, we propose that any height reductions occurring in these segments are a result of both curvature and losses from the intervertebral discs. However, considering the fact that significant anterior lean occurred with heavier loads in the UT and LT segments, any shrinkage that may have occurred as a result of either effect is likely to have been masked by the phenomena of the skin markers moving apart.

The spinal shape analysis revealed less anterior lean during loading at time $t = 0$, in the 15 % body weight loading condition. This maybe a subconscious strategy to move the weight vector of the load more over the lumbar spine, reducing the flexion moment of the load and thus the required extension moment of the spinae erector muscles. This spinal shape change between loaded and unloaded conditions at individual time intervals will also reduce the moment arm of the load about the most anterior point of the lumbar lordosis and reduce the moment arm about the most posterior point of the thoracic kyphosis.

The spinal shape analysis also shows a tendency for less anterior lean with respect to time. As fluid is lost from the

discs, this analysis suggests that there is a spinal shape compensation mechanism, actively but subconsciously controlled by muscular activity. This is supported by the curvature data, which suggest an active straightening of the thoracolumbar spine under load. This would reduce the flexion moment of the load about the lumbar spine as previously described. A reduction of the flexion moment would reduce the loading on the disc and any subsequent future fluid loss. However, this may increase loading on the neural arch, in particular the facet joint [19, 20]. Since fluid loss from the disc also increases facet joint loading, the effects would be combined. In combination with lumbar spinal shrinkage, such postural changes are likely to increase the loading on the facet joints and subsequently unload the discs which may be beneficial for those with low back pain.

Conclusion

Total spine length and LU segment length decreased with respect to time in a 30-min walking task. The height losses observed in the LU segment were greater with heavier loads and appear to be a result of fluid losses from the intervertebral discs. The UT and LT segments exhibited greater height losses in the unloaded condition but maybe a result of greater anterior lean during load carriage with the heavier loads.

The reduction in both anterior lean during instantaneous loading and with time and the straightening of thoracolumbar curvature with load are both suggestive of a spinal shape compensation mechanism to reduce the flexion moment of the load about the lumbar spine. This would reduce loading on the disc but may increase loading on the neural arch and in combination with increased facet joint loading as a result of the discal fluid losses may exacerbate pain in the pathologically abnormal spine. However, walking maybe beneficial for back pain sufferers insofar as it may encourage facet loading and hence unload the discs.

Acknowledgments We would like to thank SIPBS at the University of Strathclyde for their assistance in, and the use of their laboratory.

Conflict of interest None.

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