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Original Article

Effects of lumbar flat back posture on upper trunk acceleration during gait: A preliminary study

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Abstract

Sagittal imbalance such as flattened lumbar spine and reduction in lumbar lordosis is an essential clinical sign of back health problems. This study investigated the difference in upper body (thoracic) acceleration during gait in young individuals with lumbar flat back posture. A total of 22 younger, asymptomatic volunteers were divided into two groups: those with reduced lumbar lordosis (RLL, lumbar lordosis <20°) and those with normal lumbar lordosis (NLL, 20° ≤ lumbar lordosis < 30°). Participants walked 7 m at a self-determined speed with an accelerometer attached over the T7 spinous process. The T7-normalized medial–lateral (ML) acceleration was higher in the NLL group, whereas the T7 anterior–posterior (AP) acceleration was higher in the RLL group. These findings suggest that lumbar posture affected thoracic acceleration, and that the AP thoracic velocity changed more frequently in individuals with RLL than in subjects with normal lordosis during walking. Therefore, the RLL group required AP directional movement control during walking to improve walking efficiency.

Keywords: flat back, posture, lumbar lordosis, thoracic movement, walking

1. Introduction

A balanced sagittal profile of the spine is important to maintain ideal posture, prevent lower back pain, and promote efficient functional activities of daily living (Nairn & Drake, 2014; Vialle *et al*., 2005). The spinal curves absorb the shock of the load imposed on the spinal column, reducing and distributing the load. Cervical spine stability supports the head and thoracic curvature in determining overall spinal posture, and lumbar lordosis supports an efficient upright posture. Additionally, these curves mutually influence spinal mobility for movement patterns in other segments and regions of the spine, lower extremities, and shoulder girdle, all of which interact along a linear chain for optimal functioning of the body (Berthonnaud, Dimnet, Roussouly, & Labelle, 2005; Edmondston & Singer, 1997; Vialle *et al*., 2005).

However, sagittal imbalance, such as that occurring with a flattened lumbar spine and reduction in lumbar lordosis, is an essential clinical sign of back health problems such as lower back pain, because altered vertebral curvatures change the spatial interaction between gravity and each spinal region, causing increased stress on muscles, ligaments, bones, and discs (Neumann, 2009; Vialle *et al*., 2005). Additionally, distorted vertebral curvatures not only lead to compensatory posture and movement in adjacent segments or other vertebral curves during various activities but also affect gait by changing the relationship between center of mass (COM) displacements and the base of support (Berthonnaud, Dimnet, Roussouly, & Labelle, 2005). Previous studies have explored posture and gait in conditions involving degenerative flattened lumbar spine, such as lumbar spinal stenosis. These patients have prominent gait features including forward inclination of the trunk, reduced step length, and a wide-based gait, which increase the load to the lower back and deteriorate balance (Lee, Lee, Kim, Hong, & Yoo, 2001; Leteneur, Gillet, Sadeghi, Allard, & Barbier, 2009). Stief, Meurer, Wienand, Rauschmann, and Rickert (2015) studied 26 patients with lumbar spinal stenosis before and after mono- or bi-segmental spinal fusion surgery using biomechanical assessments of gait and trunk range of motion. They found that after surgery the patients had improved walking abilities including pain-free walking distance, walking speed, step length, as well as

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biomechanical changes such as reduced anterior pelvic and thoracic tilt, and that these did not differ from the control group. The authors also reported a decrease in maximum forward flexion after surgery and suggested that mono- and bi-level segmental changes normalized the sagittal alignment of the pelvis and thorax during walking. In other words, changes in the alignment of one or two segments of the lower lumbar region or sacrum affected the coordination of the pelvic, lumbar, and thoracic spine during walking. The coordination of these regions enables optimal posture as well as energy-efficient walking (Swinnen *et al*., 2013). Overall, many studies have focused on walking with degenerative flat back syndrome, such as cases of lumbar spinal stenosis, but few have investigated non-structural lumbar flat back posture in young subjects.

Some previous studies have investigated non-structural flat back posture. Claus, Hides, Moseley, and Hodges (2009) quantified the range of surface spinal curves in flat back posture during sitting under imitated and facilitated conditions and reported that flat back posture was associated with a lower degree of kyphosis (imitated: 4.6°; facilitated: 3.4°), and lumbar lordosis was close to zero (imitated: −1.5°; facilitated: 0.1°). Shin and Yoo (2019) classified the lumbar angles of young healthy subjects with flat back posture into global and regional angles and investigated their characteristics. Some previous studies also have investigated the relationship between non-structural posture and lower back pain. Smith, O'Sullivan, and Straker (2008) classified 766 adolescents into sagittal standing posture subgroups using a set of bony markers and determined whether the subgroups varied in terms of spinal pain. They found that 172 (22.5%) of the 766 participants had flat back posture, and that compared with those classified as having a neutral posture, adolescents with a flat back posture scored 1.78 times higher on measures of back pain. Harrison, Colloca, Harrison, Janik, Haas, and Keller (2005) reported that anterior trunk translation in standing subjects increased extensor muscle activity and load and stresses acting on the intervertebral discs in the lower thoracic and lumbar regions. To date, most previous studies related to non-structural flat back posture have classified participants by posture and quantified the relationship of posture to lower back pain under a range of static postures; very few studies have examined movement and gait patterns. Furthermore, with regard to biomechanics research focused on the lumbar spine, few studies have considered the interaction between lumbar posture and thoracic movement despite the close anatomical and biomechanical relationship between them.

Various quantitative methods have been used for gait analysis, such as the GAITRite Portable Walkway system and three-dimensional motion-capture systems. The GAITRite system is a reliable tool for measuring spatial and temporal parameters of the footstep pattern. However, it is limited to measuring direct trunk movements. Three-dimensional motion-capture systems are accurate but are costly and require a controlled environment and are thus not available in many clinical settings (Chung & Ng, 2012; Van der Linden, Kerr, Hazlewood, Hillman, & Robb, 2002; Webster, Wittwer, & Feller, 2005). Among the alternatives are tri-axial accelerometers, which are widely used to monitor activities of daily living. These have been used in gait analyses (Asai, Doi, Hirata, & Ando, 2013; Shin, An, & Yoo, 2016), evaluations of upper body sway (Asai, Doi, Hirata, & Ando, 2013), and measurement of energy expenditure (Shin, An, & Yoo, 2016). These instruments are useful in that they are accurate, highly wearable, low in cost, capable of long-term monitoring, and can be used alone or as supplements to other equipment (Shin, An, & Yoo, 2016; Simon, Ilharreborde, Souchet, & Kaufman, 2015). Therefore, we used these to investigate upper trunk (thoracic) acceleration during gait in young individuals with lumbar flat back posture (i.e., reduced lumbar lordosis).

2. Materials and Methods

2.1 Subjects

The sample size in this study was determined from a pilot study with 6 subjects. G-power 3.1.2 software (Franz Faul, University of Kiel, Germany) calculated a required sample size of 12 subjects (group 1 = 6 subjects, group 2 = 6 subjects) with a significance level of .05, power of .95, and effect size of 2.56 (calculated using mean and standard deviation from the pilot study). In total, 22 young male subjects were recruited for this study. None had a history of disease or any problems with walking, or any history of musculoskeletal pathology during the prior 12 months, surgery, or traumatic injury. No subjects had any functional restrictions, respiratory or neurological disorders, or reported any pain elsewhere in the spine or lower limbs.

Participants were divided into two groups: those with reduced lumbar lordosis (RLL, global lumbar lordosis angle (T10–S2): <20°) and those with normal lumbar lordosis (NLL, 20° ≤global lumbar lordosis angle <30°). The RLL group consisted of 11 subjects with an average age of 23.4 ± 2.8 (standard deviation) years and average height and weight of 173.26 ± 5.8 cm and 67.9 ± 6.2 kg, respectively. The NLL group consisted of 11 subjects with an average age of 21.9 ± 2.5 years and average height and weight of 173.36 ± 3.6 cm and 66.9 ± 4.0 kg, respectively. Table 1 lists the characteristics of all subjects. Ethical approval was obtained from the Inje University Ethics Committee for Human Investigations, and written informed consent was obtained from all participants (IRB number: INJE 2018-06-007).

Table 1. General characteristics of the participants (N = 22)

Variables	Group 1 (n = 11)	Group2 (n = 11)	p
Age (years)	23.36 ± 2.84	21.91 ± 2.47	.215
Height (cm)	173.55 ± 5.77	173.36 ± 3.59	.930
Weight (kg)	67.91 ± 6.17	66.93 ± 4.03	.664
BMI (kg/m^2)	22.63 ± 2.69	21.89 ± 1.29	.418
Global (º)	-14.03 ± 4.88	-23.18 ± 1.68	<.001*

All values are mean ± standard deviation, Abbreviation: BMI, body mass index, Group 1, Global lumbar lordosis (T10 relative S2 angle) < 20º; Group 2, 20º≤ and <30º, *p< 0.05

2.2 Examiner and instruments

Before task performance, participants were asked to stand in a comfortable posture with feet slightly apart, looking ahead. The examiner marked the participant's T10–S2 spinous processes (Claus, Hides, Moseley, & Hodges, 2009). The examiner identified T10 by counting down vertebrae starting at T7. The S2 spinous process identified the posterior superior iliac spine line. Marker placement has been shown to be a potential source of error in measurement of spinal movement with skin markers. To minimize this error, one investigator identified the relevant spinous processes and marked them accordingly on all participants. The examiner measured the global lumbar lordosis angle (T10–S2) using a dual inclinometer (Acumar, Lafayette Instrument Co., Lafayette, IN, USA) (Figure 1). We measured three times and used the mean value (ICCs = 0.984).

Gait time, normalized upper trunk (T7) accelerations in each of three directions, and energy expenditure during gait were measured using a tri-axial accelerometer (Fit Dot Life, Suwon, Korea). Participants were asked to stand in a comfortable posture with feet slightly apart, looking ahead. The examiner identified the most prominent cervical process as the participant's C7 spinous process, and the vertical line drawn 7 levels down was set as the T7 spinous process. A sensor was fixed with double-sided adhesive tape over the T7 spinous process (Figure 2). Raw data were measured on the x, y, and z acceleration axes; data were transferred automatically to a computer using a USB cable. A sensor range of –2 G to +2 G was used. Another accelerometer with start and stop buttons operated by a hand switch was used by the investigator. Before the test, we synchronized the two accelerometers with acquisition software (Fitmeter Manager 2; ver. 1.2.0.14, Fit Life, Inc., Korea). Data were collected at a sampling rate of 32 Hz. Gait time was calculated by reference to the accelerometer data. The root mean square (RMS) of trunk acceleration is often used as a measure of gait variability (Iosa, Fusco, Morone, & Paolucci, 2014). Optimal dynamic balance results in smooth trunk acceleration during walking, so a low RMS is considered evidence of a healthy gait. However, because acceleration RMS is highly correlated with walking speed, the RMS ratio was employed for normalization. Normalized anterior–posterior (AP), medial–lateral (ML), and vertical (VT) accelerations were calculated using the following formulas (Shin, An, & Yoo, 2016):

$$\mathrm{AP} = \left(\sum_{i=1}^{n}\sqrt{z_i^{\,2}}/\sum_{i=1}^{n}\sqrt{x_i^{\,2}+y_i^{\,2}+z_i^{\,2}}\right)\times 100,$$
$$\mathrm{ML} = \left(\sum_{i=1}^{n}\sqrt{x_i^{\,2}}/\sum_{i=1}^{n}\sqrt{x_i^{\,2}+y_i^{\,2}+z_i^{\,2}}\right)\times 100,$$
$$\mathrm{VT} = \left(\sum_{i=1}^{n}\sqrt{y_i^{\,2}}/\sum_{i=1}^{n}\sqrt{x_i^{\,2}+y_i^{\,2}+z_i^{\,2}}\right)\times 100$$

2.3 Procedures

The test was explained to participants in advance, after which the accelerometer was attached with double-sided tape over the T7 spinous process. Participants were asked to walk 7 m on a pathway at a self-determined speed. Participants started walking 2 m before measurement began and continued walking 2 m after measurement ended to avoid effects due to acceleration and deceleration; 3 m of walking were retained for analysis. After two practice trials, participants performed three measurement trials. Participants rested for 1 minute between trials.

2.4 Statistical analysis

All data were analyzed using the SPSS statistical package (version 18.0 for Windows; SPSS, Chicago, IL, USA). The independent samples t-test was used to analyze between-group differences in demographic, T7 acceleration, and walking time. Statistical significance was determined at a level of $p < 0.05$.

Figure 1. Measurements of lumbar posture by dual inclinometer

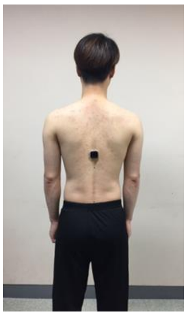

Figure 2. Accelerometer attachment (Experimental condition)

3. Results

The average global lumbar lordosis angle of the RLL group (–14.03° ± 4.88°) was significantly lower than that of the NLL (–23.18° ± 1.68°, P<0.001). The normalized AP acceleration of T7 in the RLL group (39.02 ± 5.26%) was significantly higher than that in the NLL group (32.55 ± 3.47%, P=0.003), whereas the normalized ML acceleration in the RLL group (44.70 ± 5.36%) was lower than that in the NLL group (51.98 ± 7.89%, P=0.021). The normalized VT acceleration, average gait time and energy expenditure during gait were not different between groups (P>0.05).

Table 2. Comparison of T7 acceleration during walking between groups (N = 22)

Variables	Group 1 (n = 11)	Group 2 (n = 11)	95% CI	p
AP (%)	39.02 ± 5.26	32.55 ± 3.47	2.46 ~ 10.47	.003*
ML (%)	44.70 ± 5.36	51.98 ± 7.89	-13.27 ~ -1.27	.021*
VT (%)	62.99 ± 7.16	59.07 ± 9.04	-3.36 ~ -11.2	.274
Gait time (sec)	2.36 ± 0.39	2.58 ± 0.45	-0.02 ~ 2.08	.256
Mean activity (m/s^2)	6.31 ± 1.27	5.28 ± 0.45	-0.59 ~ 0.19	.054

All values are given as mean ± standard deviation. Abbreviations: AP, anterior–posterior; ML, medial–lateral; VT, vertical; CI, confidence interval, Group 1, global lumbar lordosis angle < 20°; Group 2, 20°≤ and <30°, Global, T10 relative S2 angle, p<0.05.

4. Discussion

We examined differences in normalized directional upper trunk acceleration, gait time, and energy expenditure between the RLL and NLL groups during walking. Our primary findings indicated that in the RLL group, AP velocity changed frequently, whereas ML movement was reduced, during walking.

In this study, the normalized AP acceleration of T7 was significantly higher in the RLL group than in the NLL group. Directional acceleration is defined as the temporal rate of change in directional motion velocity. Increased acceleration means that directional velocity changes frequently during walking (Chen & Chou, 2010; Studenski *et al*., 2011; Van Kan *et al*., 2009). Generally, the acceleration is calculated by root mean square (RMS; i.e., the sum of the three directional vectors), and acceleration in each direction is expressed by the directional root mean square. However, low acceleration values do not always mean stability because the acceleration increases with velocity over time (Stokes, Andersson, & Forssberg, 1989; Swinnen *et al*., 2013). Thus, directional acceleration values were normalized to the directional mean RMS/ mean RMS and are expressed as % RMS (Shin, An, & Yoo, 2016).

Although walking appears to take place at a stable forward propulsion, in fact, the body speeds up and slows down slightly with each stage of gait. When the supporting limb is in front of the body's COM, the body slows down, whereas when the supporting limb is behind the body's COM, body speed increases. These gait mechanics generate trunk acceleration. Additionally, COM displacement is reduced by thoracic and pelvic motion, weight transfer generates ML displacement of the COM, and arm swing influences trunk movement during walking to conserve energy (Crosbie, Vachalathiti, & Smith, 1997; Swinnen *et al*., 2013). Thus, trunk ML movement contributes to efficient walking. We found that the RLL group frequently changed in terms of upper trunk oscillation velocity in the AP direction, whereas trunk oscillation velocity was lower in the ML direction during walking. The range of thoracic spine flexion and extension movement was larger than that of transverse or coronal angles because the apophyseal joints were slightly more vertically orientated. Individuals in the RLL group exhibited decreased lordosis (more flexion of the lumbar spine) and kyphosis (more extension of the thoracic spine), allowing easy posterior (extension) movement of the thorax and thereby increasing the moment arm of the extensor muscles and putting stress on posterior structures such as the facet joints (Neumann, 2009). Nairn and Drake (2014) reported that increased lumbar flexion tended to reduce axial rotation in the mid-thoracic region, and changes in thoracic movement were altered by lumbar spine posture. To our knowledge, few studies have examined walking in individuals with a non-structure problem or young, healthy flat back posture. Although we cannot directly compare the results of this study with those of previous studies, we have interpreted our results in the context of related studies. Ohtaki and Mamizuka (2014) evaluated the gait characteristics of patients with lumbar spinal stenosis before and after surgery and reported significant changes in postoperative maximum walking distance and walking time. Postoperative subjects also had a reduced sway range, chiefly in the AP direction, maintaining upper body steadiness and walking smoothness; thus, sway in the AP direction decreased after recovery from normal lordosis, and sway in the ML direction was not statistically significant. It is important to consider a multi-segmented approach when assessing trunk movement. Our findings are partially supported by the findings of Boulet, Boudot, and Houel (2016), who demonstrated the relationship of each spinal curve and the center of pressure position with velocity in healthy subjects. They reported that the lumbar lordosis angle increased with increases in the thoracic kyphosis angle. Furthermore, decreasing the thoracic curve is consistent with a forward shift of the center of pressure with high velocity and improves the lumbar lordosis angle, which increases ML velocity. Together, these findings demonstrate the relationship of biomechanics to lumbar and thoracic movement in a static posture, and may explain why the RLL group exhibited greater AP directional acceleration than did the NLL group.

We found that gait time and energy expenditure did not differ between groups. Although the RLL group exhibited reduced ML acceleration compared to the NLL group, this is an energy-inefficient walking pattern (Crosbie, Vachalathiti, & Smith, 1997; Studenski *et al*., 2011). This result is supported by the findings of Leteneur, Gillet, Sadeghi, Allard, & Barbier (2009), who demonstrated that thoraco-lumbar extension moments were higher for forward leaners and flexion moments were higher for backward leaners, in a study of 25 young men divided according to natural backward or forward trunk inclination during level walking. That study also reported no difference in walking velocity between groups, despite a strong correlation between trunk inclination and lumbar lordosis, implying inter-joint synergy. Auvinet *et al*. (2002) reported that among active, healthy adults and elderly subjects walking at a comfortable speed for 40 m, walking variables (i.e., walking speed, stride length) were higher in men than in women, and began to decrease during the sixth and seventh decades in men and women, respectively. Although that study did not provide raw data on posture, the problem of unstructured posture did not appear to affect walking velocity. Therefore, we conclude that since our participants were asymptomatic young men who walked only 3 m, there were likely no differences in walking speed or energy consumption between groups, despite differences in upper trunk acceleration in each direction that made it difficult to accurately evaluate energy consumption. This study was also limited by a small sample size, so caution should be exercised in generalizing the results. Moreover, we measured only thoracic acceleration and did not measure lumbo–pelvic acceleration, so future research should include longer gait times and investigate lumbo–pelvic movement.

5. Conclusions

In this study, we compared normalized directional upper trunk acceleration, gait time, and energy expenditure between RLL and NLL group gaits during walking. We found that during walking, lumbar posture affected thoracic acceleration, and that the AP thoracic velocity changed more frequently in individuals with RLL than in subjects with normal lordosis. Therefore, the RLL group required AP directional movement control during walking to improve walking efficiency.

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