ORIGINAL

Effect of leg length discrepancy on spinopelvic alignment and mobility in healthy volunteers using a shoe lift

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Abstract : Background : Leg length discrepancy (LLD) is a musculoskeletal condition in which leg length is asymmetric. LLD can adversely affect adjacent joints. A typical example is hip–spine syndrome. We aimed to investigate how LLD affects spinopelvic alignment and mobility, under conditions simulating LLD. Methods : Fifty healthy adults (31 women, 19 men ; mean age 37.1 ± 8.9 years) participated in this study. A shoe lift was applied under the left foot to simulate LLD of 1, 2, 3, and 4 cm. Spinopelvic alignment while standing upright and mobility in flexion, extension, and lateral bending were measured under conditions simulating LLD using a Spinal Mouse device. Results : Lumbar spine curvature in the frontal plane was significantly increased to the left as LLD increased. Sacral inclination angle in the frontal plane was significantly increased to the right by adding a lift. There was no significant difference in curvature of the thoracic spine in any parameters. Mobility of the lumbar spine during left lateral bending decreased as LLD increased. Inclination between T1 and S1 in the frontal plane when bending to the left decreased with increasing LLD. Conclusion : Simulating LLD affects spinopelvic alignment and mobility in the frontal plane, but not significantly in sagittal plane. J. Med. Invest. 72 : 26-33, February, 2025

Keywords **:** *Leg length discrepancy, Spinopelvic alignment, Spinopelvic mobility, Hip-spine syndrome*

INTRODUCTION

Leg length discrepancy (LLD) is a musculoskeletal condition in which leg length is asymmetric. LLD is classified as either structural or functional (1). A structural LLD occurs when there is an actual difference in length of the bony components in the lower limb, and a functional LLD occurs secondary to a rotated pelvis and axial malalignment (2). LLD may be associated with stress fracture, unilateral osteoarthritis, low back pain, and other clinical symptoms (3). Mild LLD (<3 cm) may cause clinical symptoms ; however, the association remains controversial (4, 5). It is widely accepted that adjusting leg length is not necessary as long as the difference is less than 2 cm (6). However, even less than 2 cm of LLD may influence adjacent joints. For example, a minor degenerative change at the femoral head resulting from osteoarthritis of the hip may cause mild LLD that in turn leads to compensatory changes in the lumbar spine in the frontal plane over the long term (3). This may change the lumbopelvic alignment and affect the normal function of the spine and pelvis, especially spinopelvic mobility. Restricted mobility can lead to deterioration not only at the affected joint but throughout the entire kinematic chain as well (7). This could be one of the mechanisms leading to hip–spine syndrome (8).

Many studies have investigated the relationship between LLD and static or dynamic alignment (2, 4, 6, 9-17). However, how LLD affects spinopelvic mobility is still not well understood. The purpose of this study was to examine spinopelvic mobility with simulated LLD to estimate how LLD might affect spinopelvic function.

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MATERIALS AND METHODS

Study participants

Fifty healthy adults (31 women, 19 men ; mean ± standard deviation, age 37.1 ± 8.9 years ; height 163.7 ± 8.5 cm ; and weight 60.0 ± 12.5 kg) participated in this study. The exclusion criteria were existing LLD or serious musculoskeletal disorder such as inability to apply adequate weight to the lower extremities. The above exclusion criteria were established, but no one fell into this category. Participants have no history of spinal disease and were confirmed visually and on palpation to be free of scoliosis. The study was approved by the Ethics Committee of Tokushima University Hospital. All participants provided written informed consent.

Simulated LLD

Each participant was fitted with a shoe lift under the left foot to simulate LLD of 1, 2, 3, and 4 cm (Fig. 1A–E). The material of the shoe lift was high-performance urethane foam. To standardize the simulations, each participant was asked to stand still with each foot placed on a separate body weight scale such that equal weight values were displayed on both sides. The participant was then asked to maintain a position that would give, to the extent possible, the same readings on both scales while the measurements were being taken. The participant was asked to stand upright and bend forward and backward and to the left and right with and without a simulated LLD of 1, 2, 3, or 4 cm. In this study, there were no restrictions on knee flexion of the longer leg, one of the compensations that occur as lifting increases.

Measurement device

A Spinal Mouse® (SM ; Idiag AG, Fehraltorf, Switzerland) was used to measure spinopelvic mobility (Fig. 2A). The SM is a hand-held instrument that measures spinopelvic alignment by means of a three-dimensional accelerometer. The automatically obtained values are transferred to a computer to reproduce a two-dimensional graph of the spine. It is non-radiographic and non-invasive and is thus considered safe and convenient. The reliability and validity of the SM measurements have been verified in many studies, and their strong correlation with radiologic measurements confirms that the SM yields reliable measurements of spinal curvature (18-20).

Procedures

Each participant's exposed back was marked on the skin surface from C7 to S3 spinous processes for SM tracking. First, measurements were obtained without the LLD when the participant subject was standing upright and during flexion, extension, and left and right lateral bending with the participant standing barefoot with each foot placed on separate body weighing scales such that, to the extent possible, equal weight values were recorded on both scales. An examiner tracked the SM on a mark from C7 to S3. Lifts measuring 1, 2, 3, and 4 cm were then placed under the participant's left foot and the measurements were repeated.

Spinopelvic alignment and mobility were measured in the sagittal and frontal planes while performing the following movements : upright standing (sagittal plane), upright standing (frontal plane), flexion (sagittal plane), extension (sagittal plane), left lateral bending (frontal plane), and right lateral bending (frontal plane).

All bending movements, including flexion and extension, were performed in the maximum range of motion that can be moved while keeping the load as equal as possible on the left and right sides, and bending mobility from the upright position was calculated. Measurements were performed for each condition in the presence (1, 2, 3, and 4 cm) and absence (0 cm) of simulated LLD. Measurements were performed three times by one designated physiotherapist to ensure reliability, and mean values were used for the analysis.

Measurement parameters

The following parameters (Fig. 2B, C) were measured : curvature of the thoracic spine (TS : between T1 and T12) ; curvature of the lumbar spine (LS : between L1 and L5) ; sacral inclination angle (S angle : the sacral slope, defined as the angle between the horizontal plate and the sacral plate (21)) ; and inclination of the whole spine (WS : between T1 and S1).

Statistical analysis

The intra-rater reliability was confirmed using the intraclass correlation coefficient (ICC) (1, 3). ICC (1, 3) was calculated for all LS measurement conditions to confirm reliability. The data were tested for normality using the Shapiro-Wilk test. If the data were confirmed to be normally distributed, Levene's test was used to test for equal variance. When equal variance was confirmed, one-way analysis of variance was applied, followed by Tukey's test. When the data were not normally distributed, a non-parametric analysis was performed using the Steel-Dwass test. Statistical analyses were performed using R Commander 2.8.1 (The R Foundation for Statistical Computing, Vienna, Austria). An α-value of 0.05 was used to evaluate the test results.

RESULTS

The results for ICC (1, 3) are shown in Table 1. LS figures in all measurement conditions were subjected to represent a reliability of the measurement.

The results under all measurement conditions are shown in Figures 3, 4 and 5. Most of the data were not distributed normally and were therefore analyzed using non-parametric methods. However, a normal distribution was confirmed for some parameters, including LS and WS during left lateral bending, and the results for these parameters were analyzed using a parametric method.

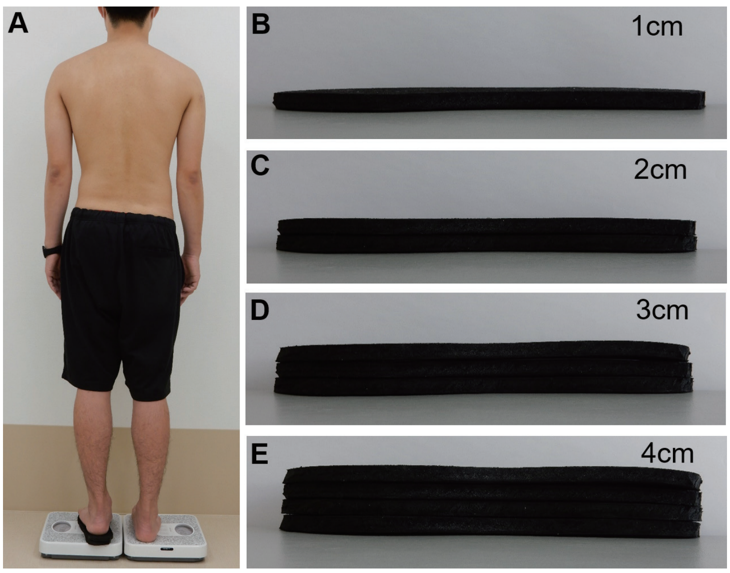

Fig. 1. Each participant was fitted with a shoe lift under the left foot to create a leg length discrepancy (A). Lateral view of shoe lift by 1 cm (B), 2 cm (C), 3 cm (D), and 4 cm (E).

Upright standing (sagittal and frontal planes)

Figure 3 shows the results when standing upright. There was a significant increase in LS to the left in the frontal plane as the LLD increased (Fig. 3F). S angle in the frontal plane was significantly increased to the right by adding a lift (Fig. 3G). WS in the frontal plane was significantly different between 4 cm and 0 or 1 cm and that to the right was the greatest at 4 cm (Fig. 3H). There was no significant difference in TS in the frontal plane or in any parameter in the sagittal plane for any LDD (Fig. 3A–E).

Flexion (sagittal plane)

There were no statistically significant changes in any parameter during flexion in the sagittal plane for any LDD (Fig. 5A–D).

Extension (sagittal plane)

No statistically significant difference was noted in any parameter during extension in the sagittal plane for any LDD (Fig. 5E–H).

Left lateral bending (frontal plane)

Figure 4A–D show the results for left lateral bending. Changes were calculated during lateral bending and upright standing at each LLD (0, 1, 2, 3 and 4 cm). LS during left lateral bending from an upright standing position decreased as LLD increased ; the differences between 0 cm and 3 or 4 cm and between 1 cm and 3 cm were statistically significant (Fig. 4B). WS in the frontal plane when bending to the left from an upright standing position decreased with greater LLD ; the differences between 0 cm and 2, 3, and 4 cm were statistically significant (Fig. 4D). There was no significant change in TS or S angle according to LLD in the frontal plane (Fig. 4A, C).

Right lateral bending (frontal plane)

There was no statistically significant difference in any parameter during right lateral bending in the frontal plane for any LDD (Fig. 4E–H).

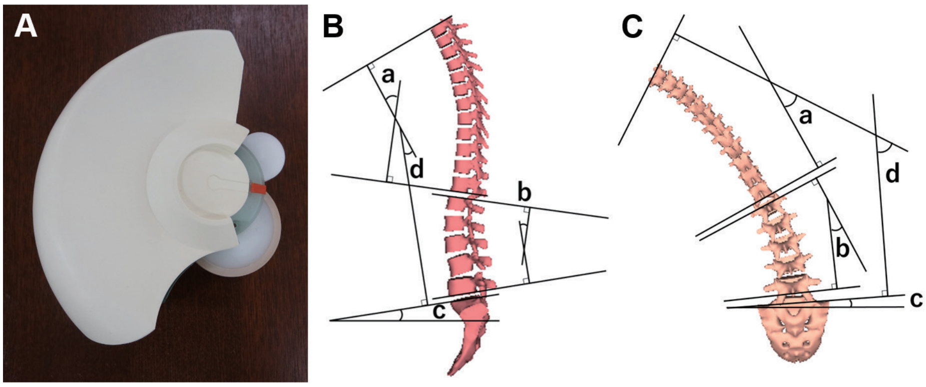

Fig. 2. (A) Spinal Mouse device. Measurement parameters : sagittal plane (B) and frontal plane (C). a : Curvature of the thoracic spine (TS : between T1 and T12), b : Curvature of the lumbar spine (LS : between L1 and L5). c : Sacral inclination angle (S angle : sacral slope, defined as the angle between the horizontal plate and the sacral plate). d : Inclination of the whole spine (WS : between T1 and S1).

Table 1. ICC (1, 3) LS

	LLD				
	0 cm	1 cm	2 cm	3 cm	4 cm
Sagittal plane					
Upright standing	0.88 (0.81, 0.93)	0.94 (0.91, 0.96)	0.93 (0.89, 0.96)	0.92 (0.87, 0.95)	0.89 (0.83, 0.93)
Flexion	0.85 (0.77, 0.91)	0.87 (0.80, 0.92)	0.91 (0.86, 0.95)	0.91 (0.85, 0.94)	0.71 (0.54, 0.83)
Extension	0.50 (0.20, 0.70)	0.67 (0.48, 0.80)	0.70 (0.52, 0.82)	0.64 (0.43, 0.78)	0.75 (0.60, 0.85)
Frontal plane					
Upright standing	0.81 (0.70, 0.89)	0.82 (0.72, 0.89)	0.64 (0.43, 0.79)	0.81 (0.69, 0.88)	0.88 (0.81, 0.93)
Left lateral bending	0.59 (0.35, 0.75)	0.68 (0.49, 0.81)	0.80 (0.69, 0.88)	0.75 (0.60, 0.85)	0.78 (0.64, 0.86)
Right lateral bending	0.82 (0.71, 0.89)	0.80 (0.68, 0.88)	0.80 (0.69, 0.88)	0.82 (0.71, 0.89)	0.89 (0.82, 0.93)

ICC, Intraclass coefficient (95% confidence interval)
LLD, leg length discrepancy ; LS, curvature of the lumbar spine

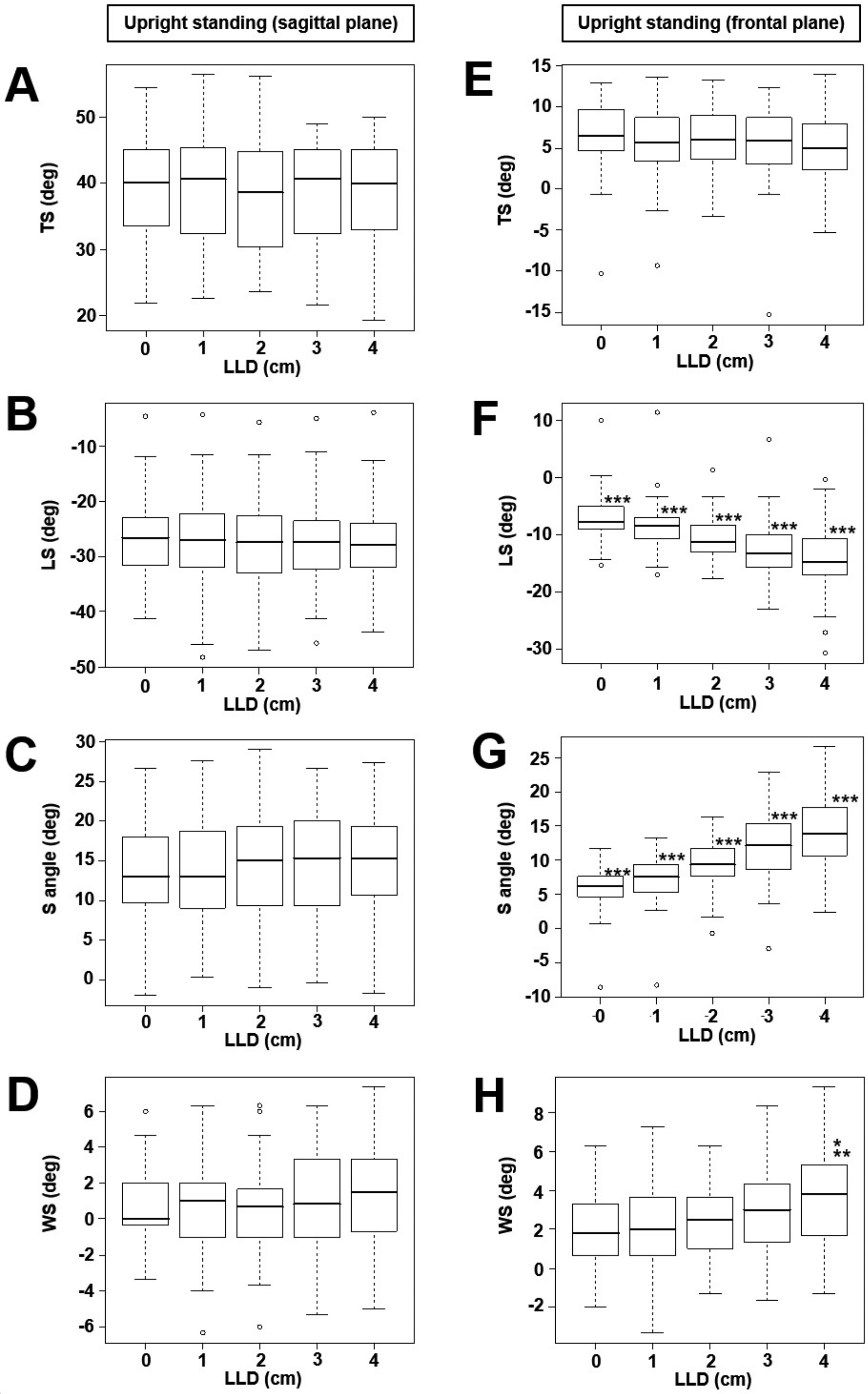

Fig. 3. Measurement parameters (TS, LS, S angle, and WS) during upright standing in the sagittal plane (A–D) and frontal plane (E–H). (A, E) TS, (B, F) LS, (C, G) S angle, and (D, H) WS. *Significant difference ($p<0.05$) from LLD of 0 cm. **Significant difference ($p<0.05$) from LLD of 1 cm. ***Significant difference ($p<0.05$) from any LLD. LLD, leg length discrepancy ; TS, curvature of the thoracic spine ; LS, curvature of the lumbar spine ; S angle, sacral inclination angle ; WS, inclination of the whole spine

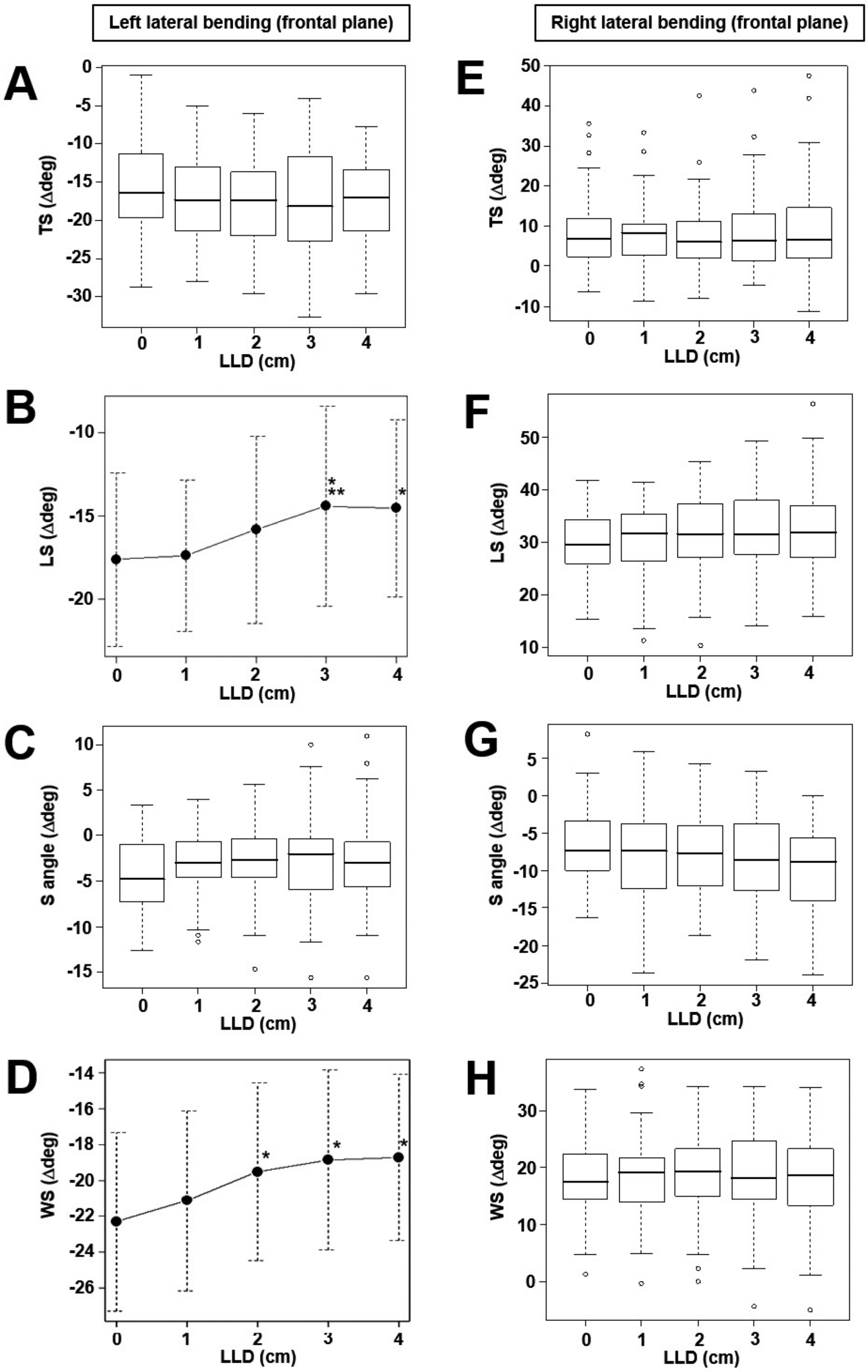

Fig. 4. Variation in measurements of TS, LS, S angle, and WS during left lateral bending (frontal plane, A–D) and right lateral bending (frontal plane, E–H). (A, E) Variations in TS, (B, F), LS, (C, G), S angle, and (D, H) WS. *Significant difference ($p<0.05$) from LLD of 0 cm. **Significant difference ($p<0.05$) from LLD of 1 cm. LLD, leg length discrepancy ; TS, curvature of the thoracic spine ; LS, curvature of the lumbar spine ; S angle, sacral inclination angle ; WS, inclination of the whole spine

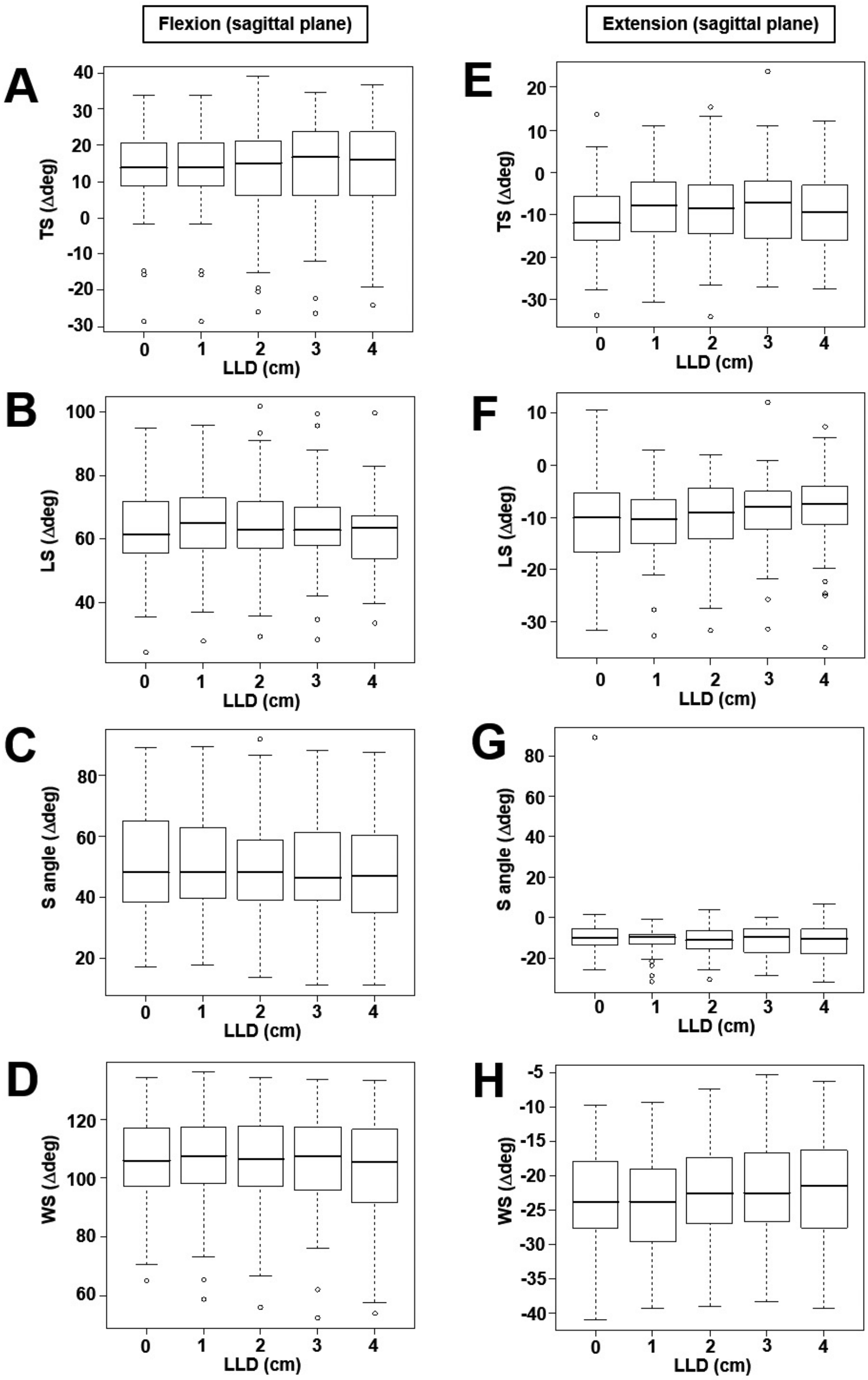

Fig. 5. Variation in measurement parameters (TS, LS, S angle, and WS) during flexion (sagittal plane, A–D) and extension (sagittal plane, E–H). (A, E) Variations in TS, (B, F), LS, (C, G) S angle, and (D, H) WS. LLD, leg length discrepancy ; TS, curvature of the thoracic spine ; LS, curvature of the lumbar spine ; S angle, sacral inclination angle ; WS, inclination of the whole spine

DISCUSSION

This study revealed that even 1 cm of LLD may induce a lumbar frontal curve and sacral tilt in the frontal plane when standing upright. LS in the frontal plane was increased on the elongated left side causing a convex lumbar frontal curve on the right side. The elongated leg raised the ipsilateral pelvis and the pelvis tilted toward the contralateral side. The direction of the curvature and inclination did not vary according to the level of lift in this simulated LLD study. LS and WS on bending toward the lift from an upright standing position decreased in the frontal plane with greater LLD. On the other hand, LLD did not affect spinopelvic parameters in the sagittal plane related to alignment when standing upright or mobility when bending. LLD did not affect parameters related to TS in the sagittal or frontal plane.

The important finding of the present study was that even LLD of 1 cm may affect lumbar curvature and pelvic obliquity in the upright standing position in the frontal plane. Although a previous study reported no significant difference in spinal posture resulting from LLD (9), there was a significant difference in alignment of the lumbar spine in the present study. We found that LS in the frontal plane was increased toward the elongated left side, thereby creating a convex lumbar frontal curve on the right. This finding is consistent with the general concept that LLD causes a lumbar frontal curve with convexity toward the shorter leg (10). Morimoto *et al*. reported that the direction of convexity of LLD-related lumbar frontal curve is uncertain when the LLD is <3 cm ; however, it was markedly directed toward the shorter leg when the LLD was >3 cm (11). However, in the present study, all degrees of LLD resulted in convexity toward the shorter side. The difference in these results depends on whether assessments were made in individuals who have a temporary LLD created using a shoe lift and not actual LLD as in our study or in individuals who have acquired LLD due to long-term unilateral osteoarthritis and are adapting by modifying their movement patterns as in the study by Morimoto *et al* (11). Assogba *et al*. reported that patients with actual LLD showed significantly decreases in mechanical work and energy cost compared with healthy participants with artificial LLD induced by soles (22). Cooperstein and Lew suggested that the direction of pelvic torsion might vary depending on whether LLD is of structural or functional origin (12). In the present study, S angle in the frontal plane was increased towards the right, meaning that the elongated leg raised the ipsilateral pelvis and the pelvis tilted toward the contralateral side. Walsh *et al*. noted that pelvic obliquity was the most common mechanism used to compensate for LLD of up to 2.2 cm and that flexion of the knee of the longer leg developed for larger degrees of discrepancy (13). In our study, the compensation strategies were not specified, when the weights displayed on the two scales were equal. Our results suggest that pelvic obliquity could still be one of the compensatory mechanisms used even when LLD exceeds the 2.2 cm shown in the previous study. In the upright standing position, we found significant increases in compensatory height and contralateral tilt in the 4 cm condition (Fig. 3H). This result may also be clinically meaningful as a compensatory strategy for increased LLD.

Thoracic kyphosis, lumbar lordosis, pelvic tilt, and inclination in the sagittal plane were not influenced by LLD in the upright standing position. In the present study, the temporary simulated LLD was asymmetrical in the frontal plane, and therefore LLD may have resulted in significant changes in spinopelvic alignment or mobility in the frontal plane, but no significant changes in the sagittal plane. In addition, the flat shoe lift was used in the present study, which may have resulted in no significant changes in the sagittal plane.

In particular, the clinically meaningful results in this study were that lateral bending to the long leg side may decrease the range of motion of the lumbar spine, and forcing repetitive movements in that direction may result in overload. Spinopelvic mobility in the frontal plane was influenced by simulated LLD to some extent. Alignment in the upright standing position was already affected by LLD, and thus spinopelvic mobility was restricted. LS and WS in the frontal plane during lateral bending to the left from an upright standing position decreased with greater LLD. A previous study that examined trunk mobility under simulated LLD conditions found that lateral flexion of the trunk increased toward the side of the lift (14). Furthermore, Gibson *et al*. reported that lateral flexion toward the longer leg was greater when measured radiographically and explained this observation in terms of the pre-existing sacral tilt (15). In our study, sacral tilt and a lumbar frontal curve were already present under greater LLD in the upright standing position. The right convex lumbar frontal curve created by the lifts on the left side meant that the lumbar spine was already laterally flexed to the left. Therefore, it is possible that the left lumbar facet joints were compressed by the temporary lumbar frontal curve created by the lifts during standing in the upright standing position and that mobility during left lateral bending was restricted. Some reports have noted that LLD might increase the mechanical stress on the longer side (1, 16). It might also be predicted that mobility during left lateral bending may be restricted to avoid an increase in mechanical stress. The participants in our study were requested to distribute their weight equally between the feet while the measurements were recorded. Therefore, any lateral weight shift was minimized. This could be the reason why mobility of the lumbar spine decreased with greater LLD during bending to the left, unlike in previous studies.

There was a dimensional limitation when using the SM device in this study. Spinopelvic compensation for LLD can be achieved not only by a lumbar frontal curve and pelvic tilt but also by spinopelvic rotation (12, 17). In this study, the SM device was used to measure spinopelvic alignment and mobility in the sagittal and frontal planes. It is important to assess changes in three-dimensional spinopelvic alignment, including in the transverse plane, during compensation for LLD (17). Previous reports have confirmed that Spinal Mouse is highly reliable (18-20) : the standard error of the measurement (SEM) ranges from 0.322° to 4.965° in the sagittal plane and from 0.958° to 4.820° in the frontal plane (20). The existence of errors is a limitation of this study.

The changes in spinopelvic alignment and mobility in response to changes in LLD were temporary in this simulation study. However, if individuals with actual LLD participate regularly in tasks involving repetitive mechanical loading while in specific postures, these changes could lead to structural deformities that cause clinical symptoms in the lumbopelvic region (2).

In actual LLD, there is an inequality in the force exerted on each foot (23). In the present study, we instructed each participant to apply the same weight to both lower limbs in order for measurements to be taken under at constant conditions, unlike in measurements in the natural position. Therefore, this constitutes a limitation of this study.

One of the most common compensations is knee flexion of the longer leg as the lift increases (13). However, in this study, there were no restrictions on knee flexion. The purpose of this study was not to intentionally suppress knee flexion to create compensation at the spinal and pelvic girdle, but to let nature determine what compensatory pattern would emerge, and to set the condition for equal loading on both sides only. This condition setting is a limitation of this study.

This study was performed under conditions that simulated LLD, and individuals with structural LLD might need to develop a different spinopelvic strategy to compensate for LLD. Further investigations in patients with structural LLD are needed.

CONCLUSIONS

This study revealed that even 1 cm of LLD may induce a lumbar frontal curve and sacral tilt in the frontal plane when standing upright. Simulating LLD affects spinopelvic alignment and mobility in the frontal plane, but not significantly in sagittal plane. The clinically meaningful results in this study were that lateral bending to the long leg side may decrease the range of motion of the lumbar spine, and forcing repetitive movements in that direction may result in overload.

CONFLICTS OF INTEREST

The authors declare that no conflicts of interest exist.

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