Dynamic Alignment Changes of the Spine, Pelvis, and Lower Limbs during Gait Analyzed Using Inertial Motion Capture in Patients with Adult Spinal Deformity

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Abstract:

Introduction: Patients with adult spinal deformity (ASD) lean forward with their trunks when walking, even if they can remain upright during static standing. However, it remains unclear which part of the spinal column is involved in forward trunk tilt and the details of the relationships between sagittal alignment during static standing and changes in dynamic parameters during walking. Therefore, this study aimed to clarify the above by analyzing the walking motion of ASD patients using inertial measurement units (IMUs).

Methods: Preoperative ASD patients were included in this study. Dynamic parameters during gait were measured by IMUs attached on the skin at the T1, T12, and S1 spinous processes, thigh, and lower leg. Walking data were divided into three phases of 10 s each (initial, middle, and final), and the average dynamic parameters at each phase were statistically compared. The relationships between the standing radiographic and dynamic parameters in the final phase were evaluated by linear regression analyses.

Results: A total of 34 patients were included in this study. Their mean age was 72 years. The inclination of IMUs on the T1, T12, and S1 and the flexion angle of T12-S1 IMUs significantly increased over time. Pelvic tilt (PT) of standing radiography was positively correlated with the inclination angles of T12 (r^2=0.22, p=0.0048) and S1 (r^2=0.16, p=0.0178) and the flexion angle of T12-S1 IMUs (r^2=0.29, p=0.0011).

Conclusions: This study showed that anteversion of the trunk in patients with ASD is due to an increase in lumbar forward bending and anterior tilt of the pelvis. Lumbar forward bending was significantly correlated with PT on standing radiography. It is important to consider the presence of poorer posture during gait than during standing when we evaluate patients with high PT.

Keywords:
adult spinal deformity, gait analysis, motion analysis, compensatory mechanism, sagittal alignment, inertial measurement units, motion capture

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Introduction

ASD is a condition caused by various etiologies, such as degenerative scoliosis, kyphosis, and kyphoscoliosis. The features of spinal alignment in ASD patients are loss of lordosis and increased kyphosis, which results in a forward leaning posture. It has been shown that sagittal imbalance can correlate with poor health-related quality of life (HRQOL), especially with upright standing and walking[1-3)]. To evaluate the individual spinal deformity in the sagittal plane, understanding the presence of compensatory mechanisms is indispensable. The basic compensatory mechanisms

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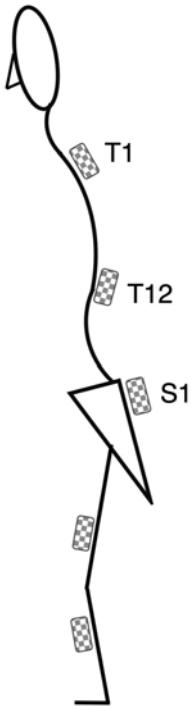

Figure 1. An illustration of IMUs attached to a subject for the analysis. IMUs are shown as plaid rectangles and are attached to the skin of the T1, T12, and S1 spinal spinous processes and the anterior thigh and lower leg.

of sagittal plane deformity are extension of the adjacent flexible spinal segments and backward rotation of the pelvis to avoid the anterior translation of the gravity line and maintain the erect position[4-7]. However, several gait analysis studies have shown that a spinal column that has regained static sagittal balance through compensatory mechanisms can lose dynamic sagittal balance during walking[8-10]. These gait analyses agreed that anterior tilt of the trunk occurs or increases as soon as a step is made. Although the retroverted pelvis in the standing position becoming anteverted during gait is considered one of the mechanisms of anterior tilt of the trunk, little is known about the change of spinal alignment from static standing to gait.

Although marker-based optical motion capture (OMC) is the gold standard for human gait analysis, inertial motion capture systems are increasingly used for gait analysis. Inertial motion capture offers several pragmatic benefits than OMC. Since inertial motion capture systems are more compact, affordable, portable, and user-friendly, they are ideal for use in clinical environments outside of the gait laboratory. Inertial motion capture is based on small yet powerful integrated circuits (IMUs), typically comprising on-board triaxial gyroscopes, magnetometers, and accelerometers[11-14]. However, to our knowledge, no study has used IMUs for gait analysis of ASD patients.

Previous motion analyses using OMC have shown that, in patients with ASD, the trunk leans forward when walking, even if the patient remains upright during static standing[8-10]. However, it remain unclear which part of the spinal column is involved in the forward trunk tilt and the details of the relationships between spinopelvic sagittal alignment during static standing and changes in dynamic parameters during walking. Therefore, this study aimed to clarify the above by analyzing the walking motion of ASD patients using 3D IMUs.

Materials and Methods

Preoperative ASD patients were included in this study. The inclusion criteria were age >60 years, and categories based on the sagittal modifiers of the SRS-Schwab classification[15] were marked, or there was moderate deformity in at least one sagittal modifier. Patients who had musculoskeletal diseases such as hip and knee osteoarthritis and who had neurological disorders (e.g., spinal canal stenosis, Parkinson's disease) were excluded. Moreover, we excluded cases of iatrogenic lumbar kyphosis due to inadequate spinal fusion surgery and fixed kyphotic healing after lumbar vertebral fractures, as pathologies resulting in loss of spinal mobility.

The radiographic sagittal parameters were measured on full-length standing sagittal radiographs with fingers on clavicles and shoulders in 45° of forward elevation. Radiographic parameters were T5-T12 thoracic kyphosis (TK), T 12-S1 lumbar lordosis (LL), PT, pelvic incidence (PI), sagittal vertical axis (SVA), and PI minus LL (PI-LL).

Motion analysis

Motion analysis was performed on a circuit of a square walkway comprising two parallel 35-m and 10-m straight paths. All patients walked at their own speeds until they were fatigued. Dynamic parameters measured in three phases were used for the analysis: the initial 10 s (initial phase), the middle 10 s (middle phase), and the final 10 s of walking (final phase). In each measurement phase, the average of the dynamic parameters at each step was calculated.

An IMU (Myomotion Research Package MR3 version 3.86, Noraxon USA, Inc., Scottsdale, AZ, USA) that was equipped with gyroscopes, accelerometers, and magnetometers and could capture the motion at the attached site precisely, with accuracy of ±0.4° and ±1.2° under static and dynamic conditions, respectively, was used[11,12]. The sensors were attached to the skin on the spinous processes of T1, T 12, and S1 and the anterior part of the thighs and lower legs with elastic wrap to minimize extraneous movement (Fig. 1). The upper end of the thoracic spine measurement in the radiograph was set at T5, but in motion analysis, the upper thoracic sensor was placed at T1 to evaluate the movement of the entire thoracic spine. The placement position of the IMU on the spine was determined by palpating the spinous processes on the skin. The inclination signal of each sensor was calibrated to zero in the standing posture just before the start of walking, and based on this, changes in the horizontal and vertical directions were defined as positive and negative, respectively; thus, we were able to accurately record

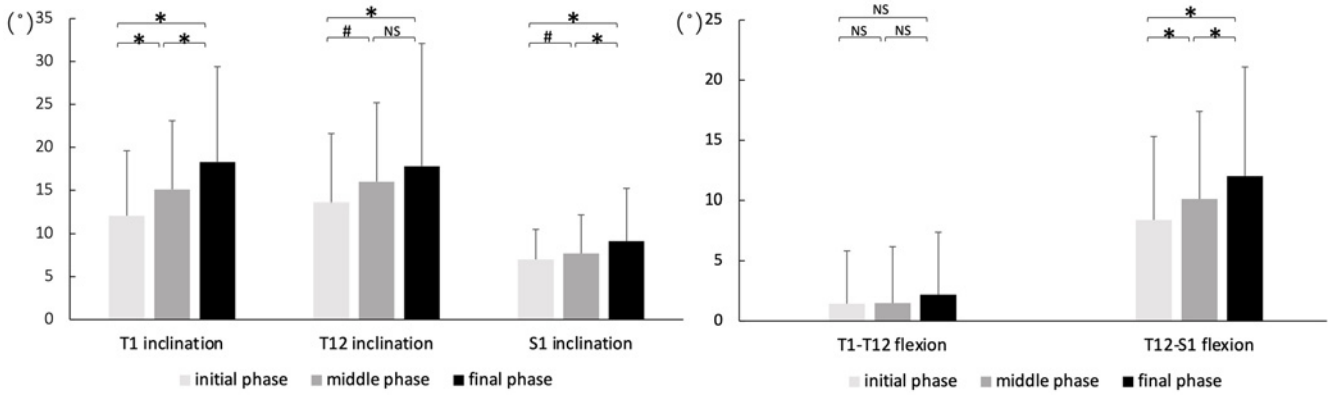

Figure 2. Spinal and pelvic dynamic parameters in the three phases. The inclination angle of each sensor in the sagittal plane is shown on the left graph, and the flexion angle of the spine is shown on the right graph. * and # denote significant differences between these groups (*, p<0.001; #, p<0.05); NS, not significant

the direction of the inclination.

Local inclination during walking was defined as the inclination angle of each sensor in the sagittal plane (i.e., T1, T 12, and S1 inclination is the sagittal inclination of the sensor at the level of the T1, T12, and S1 spinous processes, respectively). For the dynamic alignment of the spine, the T1-T12 and T12-S1 flexion angles were defined as the difference between T1 and T12 inclinations and between T12 and S1 inclinations, respectively. Similarly, the hip extension and hip flexion angles were defined as the differences between S 1 and thigh inclination, and the knee extension and knee flexion angles were defined as the differences between ipsilateral thigh and lower leg inclination. Sagittal plane deviation from the standing posture just before the start of walking was defined such that forward lean of the trunk is represented by a positive value, where a larger value indicates a more horizontal orientation. For the hip and knee joints, change in the extension and flexion directions of each joint was recorded as negative and positive, respectively.

Statistical analysis

Data were statistically analyzed using JMP 9.0 (SAS, Cary, NC). Spinal, hip, and knee dynamic parameters were compared among three phases (initial, middle, final) using the Wilcoxon signed-rank test. The relationships between the radiographic sagittal parameters and lumbar dynamic parameters in the final phase were evaluated by simple linear regression analyses. The level of significance was set at 0.05.

Results

Patients' data

A total of 34 patients, consisting of 27 women and 7 men, were included. Their mean age was 72 years (SD=8.1 years; range=60-82 years). Curve types of the SRS-Schwab classification were thoracolumbar/lumbar (TL/L) in 7 patients, double curve (D) in 5, and no major coronal deformity (N) in 22. The mean walking time and speed on the walkway were 3.53±1.64 min and 48.8±13.5 m/min, respectively. The mean values for each radiographic sagittal parameter were as follows: TK, 30.4°±23.6°; LL, 16.7°±22.9°; PT, 36.6°±12.6°; PI, 51.5°±15.5°; PI-LL, 35.2°±22.6°; and SVA, 114.6±82.2 mm.

Motion analysis

Fig. 2 shows the spinal and pelvic dynamic parameters of three selected sections. The inclination of the IMUs on the T1, T12, and S1 spinous processes significantly increased by time period (T1 inclination (initial phase, 12.1°±7.5°; middle phase, 15.1°±8°; final phase, 18.3°±11.1°), T12 inclination (initial phase, 13.6°±8°; middle phase, 16°±9.2°; final phase, 17.8°±14.3°), S1 inclination (initial phase, 7.01°±3.5°; middle phase, 7.7°±4.5°; final phase, 9.12°±6.1°)). The flexion angle of T1-T12 IMUs showed no significant change. However, the flexion angle of T12-S1 IMU significantly increased over the time period (initial phase, 8.41°±6.9°; middle phase, 10.1°±7.3°; final phase, 12°±9.1°).

Table 1 shows the hip and knee joint dynamic extension/flexion angles. The hip extension angle decreased over the time periods, although there was no significant change. There were no trends and no significant changes in knee joint dynamic parameters.

Regression analyses of the radiographic parameters vs. lumbar dynamic parameters

PT was significantly correlated with T12 inclination (r^2= 0.22, p=0.0048), S1 inclination (r^2=0.16, p=0.0178), and T 12-S1 flexion (r^2=0.29, p=0.0011) (Fig. 3). PI-LL was significantly correlated with S1 inclination (r^2=0.12, p=0.0415), but not with T12 inclination (p=0.0618) and T12-S1 flexion (p=0.0824). SVA was not significantly correlated with T12 inclination (p=0.3386), S1 inclination (p=0.0628), and T12-S1 flexion (p=0.6937).

Table 1. Hip and Knee Joint Dynamic Parameters in the Three Phases.

	Initial phase	Middle phase	Final phase	p
Rt hip extension (˚)	−7.4±5.2	−6.9±6.5	−5.1±8.2	NS
Rt hip flexion (˚)	27.2±9.2	28±9.5	27.8±11.7	NS
Lt hip extension (˚)	−8±6.7	−6.9±8.4	−5±9.2	NS
Lt hip flexion (˚)	31.7±9.53	32.7±9.38	32.8±11	NS
Rt knee extension (˚)	−4.03±7.81	−4.83±9.66	−4.35±9.39	NS
Rt knee flexion (˚)	42.3±13.9	42.5±16.8	41.2±16.6	NS
Lt knee extension (˚)	−5.1±10	−6.21±11.4	−4.50±7.94	NS
Lt knee flexion (˚)	41.4±15.4	42.5±17.1	43.1±18.3	NS

NS, not significant

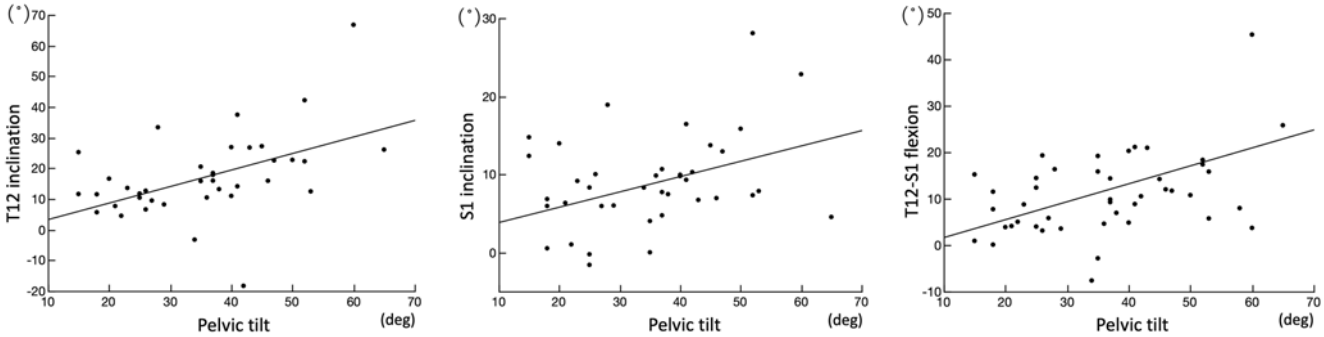

Figure 3. Regression analyses between the PT and the lumbar dynamic parameters in the final phase. PT was significantly correlated with the T12 inclination (r^2=0.22, p=0.0048), the S1 inclination (r^2=0.16, p=0.0178), and T12-S1 flexion (r^2=0.29, p=0.0011).

Discussion

Previous studies of the dynamic spinal alignment changes of ASD patients during gait used an OMC system comprising cameras and reflective markers. To our knowledge, this study was the first to use IMUs to measure dynamic spinal alignment changes of ASD patients during gait. Shiba et al. analyzed the gait of ASD patients using an OMC system and reported that anterior trunk and PT occurred during gait[8]. However, since their study evaluated trunk tilt using the axis between the line connecting the pinna and the greater trochanter, the regional spinal alignment change during gait was unknown. This study showed that the anterior trunk tilt in ASD patients is caused by anterior PT (i.e., inclination of the IMU on the S1 spinous process significantly increased by time period) and anterior lumbar forward bending (i.e., flexion angle of T12-S1 IMUs significantly increased by time period). It was found that these changes occur just after the patient starts walking and progress gradually until the patient stops walking.

To date, research has been conducted to observe changes in posture due to walking using methods other than IMUs or OMC. Bae et al. measured radiographic parameters at prewalk and post-10-min walk and found significant decreases in LL and PT at postwalk in the participants overall[16]. They showed that, in cases in which sagittal compensatory mechanisms were effective in the standing position before gait, walking reduced LL and pelvic retroversion, but thoracic spine compensation was maintained. Yin et al. measured sagittal radiographic parameters at prewalk and post-10-min walk and observed a significant decrease in LL consistently[17]. The results of this study agree with the results of the previous ones mentioned above; therefore, it appears that walking causes lumbar forward bending and anterior tilt of the pelvis.

However, the results for changes in the alignment of the thoracic spine during walking are different among the studies, and there is no consensus. Miura et al. conducted a gait analysis using an OMC system. They reported that, although the thoracic spine flexes during walking, there is no significant change in the lumbar spine[8]. However, this study found no significant changes in thoracic spine alignment during walking. In previous studies, an increase in TK after walking did not always occur, but it was observed in some cases[16,17]. Bae et al. showed that, in cases in which sagittal compensatory mechanisms were effective in the standing position before gait, thoracic spine compensation was maintained even after gait[16]. To summarize these findings, it appears that increased TK during walking is not a universal change. We believe that there are multiple factors that cause changes in the thoracic spine during walking, such as the sagittal compensatory mechanisms, which we mentioned above, and the flexibility of the thoracic spine[18]. It is not possible to make a definitive conclusion, and future research should have more participants and investigate patients with various sagittal compensatory effects.

This study showed that pelvic anteversion increased over time from the initial phase to the final phase. Previous stud-

ies of 3D gait motion analysis using OMC for ASD patients also showed increased pelvic anteversion[8,9]. In patients with lumbar kyphoscoliosis involving sagittal compensation, pelvic retroversion would work to compensate for global sagittal alignment in the static standing position, but such compensatory mechanisms by pelvic retroversion would not work during gait. Shiba et al. considered pelvic compensation, in which the hip extensor muscles work as posterior rotators of the pelvis in static standing, but during gait, the hip extensor muscles no longer work as posterior rotators of the pelvis because they are used for hip extension in walking motion[8]. Moreover, Bae reported that PT decreased after 10 min of walking compared to static standing radiography and postulated that loss of compensatory mechanisms may be due to fatigue of pelvic extensor muscles[16]. In this study, the hip extension angle tended to decrease, although not significantly. The tendency toward a decreased hip extension angle over the walking phase may be due to progressive hip extensor muscle fatigue with gait. With the additional stress of walking, hip extensor muscles were exhausted, which also resulted in pelvic anteversion.

T12 inclination, S1 inclination, and T12-S1 flexion angle in the final phase were significantly correlated with PT on static standing radiography in this study. To date, two motion analysis studies have reported the relationship between sagittal alignment changes during walking and static standing sagittal alignment parameters, but there is no consensus. Miura et al. reported that there was no significant correlation between the increase in thoracic or lumbar kyphosis and static spinopelvic parameters[9]. Arima et al. analyzed the gait posture of ASD patients and showed that PI minus LL mismatch and PT were significantly correlated with a discrepancy between standing and walking trunk forward tilt[10]. However, since their study evaluated trunk tilt using the axis between the line connecting the pinna and the greater trochanter, the relationship between the dynamic regional change of spinopelvic alignment and radiographic sagittal parameters during static standing remained unknown. In this study, IMUs classified anterior trunk tilt into anterior pelvic tilt and lumbar flexion and showed for the first time that each correlates with PT during static standing. From these results, it can be predicted that the larger the PT to maintain sagittal balance during static standing, the greater the change in lumbar flexion and pelvic anteversion during gait. Gait analysis is generally a time-consuming task. Therefore, based on the results of this study, estimating changes in dynamic spinal alignment during walking from the PT on the static standing radiograph is indispensable in clinical practice for patients with ASD.

This study had some limitations. First, muscle strength and/or activity were not measured. As mentioned above, back and hip extensor muscles play important roles in sagittal compensation mechanisms. Gait analysis such as muscle evaluation is needed in the future. Second, the accuracy of IMUs must be taken into account. However, Berner et al. validated IMU systems relative to optimal motion capture for gait kinematics and concluded that IMU-based gait analysis is reliable, with an intraclass correlation of 0.71-0.99 and mean absolute percentage errors <7.40%[14]. Regarding the accuracy of measuring instrument placement on the skin, there is a study indicating that the accuracy of marker placement is lower in the lumbar spine, where soft tissue is thicker, than in the thoracic spine, where soft tissue is thinner[19]. The fact that the IMUs were not placed in the lumbar vertebrae itself in this study was an advantage in terms of accuracy of placement. However, to accurately determine the installation location, it would be best to confirm the location by radiography. Third, parameters at standing position just before the start of walking were not presented, because the inclination signal of each sensor was calibrated to zero in the standing posture just before the start of walking. To accurately measure inclination signal at standing position in future study, we should adopt different method of calibration. Fourth, HRQOL data were excluded in this study. However, the relationships between HRQOL data and static standing radiographic parameters have been detailed in many studies[1-3]; this study mainly focused on dynamic motion assessment. Fifth, we have not examined how the spinal column and pelvis change when walking in healthy elderly people of the same age group as the control of this study.

In conclusion, by using IMUs, this study demonstrated for the first time that the anteversion of the trunk that occurs during walking in ASD patients is due to an increase in lumbar forward bending and anterior tilt of the pelvis. Lumbar forward bending and anterior tilt of the pelvis were significantly correlated with PT on static standing radiography in this study. Therefore, it is important to consider the presence of poorer posture during gait than during static standing when we evaluate patients with high PT.

Conflicts of Interest: The authors declare that there are no relevant conflicts of interest.

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Author Contributions: F.A. and H.T. designed the study; F.A., S.I., D.T., H.M., H.U., H.A., and T.I. obtained the data; F.A., D.T., and S.I. analyzed the data; H.T. supervised the study; F.A. wrote the first draft; S.I. edited the manuscript.

Ethical Approval: This study was approved by the Ethics Committee of Dokkyo Medical University (approval code: R-58-4J).

Informed Consent: Informed consent for publication was obtained from all participants in this study.

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