KU LEUVEN

GROEP BIOMEDISCHE WETENSCHAPPEN

FACULTEIT BEWEGINGS- EN REVALIDATIEWETENSCHAPPEN

To what extent does 3D static alignment represent the dynamic function of the spine during gait in Adult Spinal Deformity: a pilot study

door Pauline BATAILLIE
en Ines DE CEUSTER

masterproef aangeboden, tot het behalen van de graad van Master of Science in de revalidatiewetenschappen en kinesitherapie

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Opgesteld volgens de richtlijnen van *SPINE*

WOORD VOOAF

Wij zouden graag onze promotor, prof. Jonkers, willen bedanken voor de vlotte samenwerking en de feedback op onze masterproef.

Dankzij de inbreng van en de voortreffelijke begeleiding door Thomas Overbergh, onze copromotor, hebben we dit resultaat kunnen bereiken. We waarderen zijn hulp bij de dataverwerking, zijn hulp bij het ons wegwijs maken in zijn zelfontworpen programma's en zijn constructieve feedback.

Daarnaast willen we onze copromotor Mariska Wesseling bedanken. In het eerste jaar van onze masterproef stond zij steeds paraat om ons te helpen. We appreciëren haar verduidelijkende opmerkingen bij onze tekst en vragen enorm, net als haar begeleiding tijdens dit eerste jaar.

Zonder de steun en het begrip van onze familie en vrienden was het niet mogelijk geweest om deze masterproef te realiseren. Tenslotte willen we elkaar bedanken, daar we onze masterproef met een vlotte samenwerking en in goede verstandhouding tot een goed einde hebben kunnen brengen.

Kortrijk, 15 mei 2019 P.B.

Tremelo, 15 mei 2019 I.D.C.

SITUERING

Adulte spinale deformiteit (ASD) is een heterogene aandoening die uiteenlopende afwijkingen in de 3 anatomische vlakken van de wervelkolom omvat. Enkele voorbeelden zijn scoliose, sagittale afwijkingen, kyfose, spondylolisthesis, rotatoire subluxatie en axiale vlak deformiteiten[1-4]. ASD heeft een invloed op de anatomie, symmetrie en mobiliteit van de wervelkolom en daardoor een negatieve impact op beweging en balans[2-9]. Daarnaast heeft het ook een significante impact op de levenskwaliteit[1]. Dr. L. Moke heeft recent onderzocht wat het effect is van ASD op de levenskwaliteit en concludeerde dat dynamische balanstesten een hoger potentieel hebben om de levenskwaliteit van ASD patiënten te voorspellen dan demografische en tweedimensionale (2D) radiografische spinopelvische parameters[10]. De prevalentie wordt gerapporteerd tot 68% in de oudere populatie met een tendens om toe te nemen de komende jaren door een veroudering van de bevolking[1-3].

De behandeling van ASD bestaat meestal uit een operatie. Helaas staat dit nog niet op punt: 9-36% van de ASD patiënten hebben nood aan revisie chirurgie, 30-55% hebben complicaties of ervaren functionele achteruitgang na de operatie[3,11-15]. Dit zou kunnen komen door een gebrek aan uitgebreide preoperatieve evaluatie van de wervelkolom parameters in meer alledaagse dynamische situaties. Momenteel bestaat deze preoperatieve functionele evaluatie en operatieve planning immers voornamelijk uit statische 2D radiografische metingen. Aangezien ASD een invloed heeft in alle drie de anatomische vlakken en het menselijk lichaam beweegt in drie dimensies (3D), zijn artsen er steeds meer van overtuigd dat 3D dynamische informatie over de functionaliteit van de patiënten een toegevoegde waarde zal zijn bij de evaluatie en chirurgische planning van ASD patiënten. Daarom willen wij in onze masterproef concepten voorzien die kunnen dienen als een eerste stap richting het integreren van deze dynamische evaluatie. Dit zullen we doen door gebruik te maken van 3D bewegingsanalyse. Daarnaast zullen we gebruik maken van subject-specifieke skeletale modellen van de wervelkolom[16-18]. Een subject-specifiek skeletaal model is een via de computer gegenereerd model van de wervels en het bekken, dat voor elk subject varieert naargelang de afwijking omdat het gebaseerd is op medische beeldvorming. Het voordeel van deze modellen is dat we parameters kunnen onderzoeken op het niveau van de individuele wervel terwijl we rekening houden met de afwijkende anatomie, wat een beter beeld geeft van de patiënt-specifieke kinematica en afwijking. Dit kan op zijn beurt weer leiden tot een daling in het faalpercentage van ASD operaties. Onze masterproef over subject-specifieke modellen is innovatief in de onderzoekswereld, omdat eerdere studies steeds rapporteerden over de wervelkolom als één geheel[2,4-9].

Onze masterproef kadert binnen een studie aan een multidisciplinaire onderzoeksgroep, het Institute for Orthopaedic Research and Training (IORT) Leuven en staat onder het algemeen toezicht van Dr. L.

Moke. Het IORT probeert orthopedische behandelingsmethoden te verbeteren. In onze masterproef gaan wij kijken of spinopelvische parameters op 2D radiografie representatief zijn voor het statisch alignement van de 3D wervelkolom. Daarnaast gaan wij ook onderzoeken in welke mate het 3D statisch alignement de dynamische functie van de wervelzuil tijdens de gangbeweging benadert.

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To what extent does 3D static alignment represent the dynamic function of the spine during gait in Adult Spinal Deformity: a pilot study

ABSTRACT

BACKGROUND CONTEXT: Adult spinal deformity (ASD) is a frequently present condition with significant impact on quality of life. It consists of three-dimensional (3D) deformations of the spine, which affect posture and gait. Evaluation and surgical planning have been based on two-dimensional (2D) measurements. Failure and revision rates for surgical interventions are high. 3D visualizations methods have been introduced and have a potential to provide, in combination with 3D motion capture, accurate information on the dynamic spine behavior.

PURPOSE: (1) Comparison of 2D radiographic measurements on spinal alignment with 3D measurements using a subject-specific skeletal model. (2) Comparison of 3D spinopelvic parameter values during standing and walking in a cohort of ASD patients to elucidate functional compensation mechanisms during dynamic function.

STUDY DESIGN: single-center prospective study at the University Hospitals Leuven campus Pellenberg.

PATIENT SAMPLE: 11 individuals (9 females, 2 males; mean age 59.1 years) including 10 patients with adult spinal deformity and 1 healthy control. The sagittal vertical axis ranged from -1.4 to 9.8 cm.

METHODS: Radiographic parameters were measured by an experienced spine surgeon. 3D static and dynamic parameters were collected from motion analysis. The following parameters were used for comparison: pelvic incidence (PI), pelvic tilt (PT), sacral slope (SS), lumbar lordosis (LL), thoracic kyphosis (TK), T1 spinopelvic inclination (T1SPI), T9 spinopelvic inclination (T9SPI), sagittal vertical axis (SVA) and Cobb angle. No funds were received in support of this study.

RESULTS: The 2D radiographic values deviated strongly from the 3D static values. During gait, subjects presented an increase in SVA and SS and a decrease in T1SPI and T9SPI, with varying results for the remaining spinopelvic parameters.

CONCLUSION: Radiographic measurements alone cannot predict the 3D dynamic function in ASD patients. A comparison between standing and walking parameters revealed compensation strategies used by ASD patients; increased pelvic anteversion, sacral slope and a more forward tilted trunk. 3D dynamic subject-specific information provides improved insights into patients' spinal dysfunctions and consequent compensations in comparison with static radiographic measurements.

KEYWORDS: Adult spinal deformity; 3D reconstruction; subject-specific skeletal model; spinopelvic measurements; compensation strategies; gait analysis

INTRODUCTION

Adult spinal deformity (ASD) is a heterogeneous disease present in the adult population that includes various spinal malalignments in all three anatomical planes. It includes scoliosis, sagittal malalignment, kyphosis, spondylolisthesis, rotatory subluxation, and axial plane deformity[1–4]. ASD is a common medical disorder with a significant impact on the quality of life[1]. Former studies report on the relationship between radiographic parameters and functional outcome questionnaires e.g. quality of life, Oswestry Disability Index (ODI) and SF-36[5-8]. These studies found worse functional outcomes when radiographic parameters showed more deformation[5-8]. The prevalence of ASD has been reported to be up to 68% in the older population with a tendency to increase in the following years, due to an ageing population[1–3]. It is a progressive growth disease that affects anatomy, symmetry and mobility and therefore has a negative impact on motion and balance[2–4,9-13].

Previous studies, mostly based on two-dimensional (2D) motion analysis, report the change in gait pattern in individuals with ASD[2,4,9-11,13-16]. ASD patients have a more stiff gait pattern than control subjects, both due to motion restriction as a consequence of 3D structural changes and due to prolonged muscle contraction[5]. Reduced range of motion (ROM) in pelvic lift and drop, hip abduction and adduction, hip rotation and knee flexion and extension were found, with prolonged bilateral activation of several lumbar and pelvic muscles[2,4,10,13]. These could be considered as compensation mechanisms to limit the progression of imbalance[9].

ASD treatment typically consists of surgery. Unfortunately, ASD surgery is not infallible: 9-36% of ASD patients need revision surgery, 30-55% experience complications or functional deterioration[3,17-21]. This may be due to a lack of extensive evaluation of spine parameters in more dynamic conditions. Presurgical functional evaluation and operative planning is mainly based on static 2D radiographic measurements. Since, as aforementioned, ASD affects patients in all three anatomical planes, consensus is growing that three-dimensional (3D) dynamic information about the functionality of patients may provide additional insights into the evaluation and surgical planning of ASD patients.

Previous research has reported on the clinical relevance of 3D analysis in ASD patients. In 2017, Ferrero et al. found that using 3D imaging might predict pathology at earlier stages than traditional 2D radiographs[22]. The research of Kato et al. emphasises the importance of dynamic fluctuation during standing, both in healthy subjects as in subjects with a spinal deformity[23]. They concluded that the reliability of a single radiographic examination is suboptimal and even further compromised in patients with a structural instability such as spinal deformity, as they present with more postural sway[23]. Marks et al. examined the reliability of the measurement of the sagittal vertical axis (SVA) on regular standing radiographs and concluded that there was poor reproducibility of these values in

more functional positions[24]. In 2012, Somoskeöy et al. compared the accuracy of methods by conventional manual 2D and sterEOS 3D measurements[25]. They concluded there was an overall good reliability of 2D measurements, which was excelled by significantly higher reliability of 3D measurements[25].

Previous research in ASD subjects typically reports on the kinematics of the entire spine or larger parts of the spine, examining the spine as a single rigid segment[2,4,9–13]. The need is however identified to refine these analyses and examine the kinematics at the level of the vertebrae. The current state of the art in subject-specific skeletal modeling has now advanced sufficiently to allow examining in vivo intersegmental kinematic parameters throughout motion, providing a more accurate insight into the individual patients' kinematics and the contribution of specific segments[26–28]. This approach is highly innovative as current literature is largely silent on dynamic evaluation with subject-specific skeletal models.

The main purpose of this study is to investigate to what extent static alignment can represent the dynamic function of the spine in ASD patients during gait. First, we will compare 2D radiographic measurements with 3D measurements during stance in ASD patients. Second, we will evaluate the discrepancy between walking and standing posture in ASD patients by using 3D subject-specific skeletal models. In combination with 3D motion capture (MOCAP) data, we will investigate the following parameters in static and dynamic conditions: the pelvic incidence, pelvic tilt, sacral slope, lumbar lordosis, thoracic kyphosis, T1 spinopelvic inclination, T9 spinopelvic inclination, sagittal vertical axis and Cobb angle. This approach will provide a first step towards integrating dynamic evaluation using 3D motion capture with personalized musculoskeletal modelling, providing new concepts for clinical interpretation on spine function in ASD patients.

As previous studies showed discrepancy between walking and standing posture in patients with ASD, we hypothesize that the 2D radiographic parameters will not be representative for the full spinal function of ASD patients during dynamic daily activities, as this is not considered during the radiographic evaluation[2]. Large discrepancies will support the need to include dynamic evaluations on personalized deformity. By enhancing the current 2D static radiographic evaluation with 3D dynamic subject-specific information, we hope to provide improved insights into patients' spinal dysfunctions and consequent compensations, which could lead to a reduction in the failure rate of ASD surgery.

MATERIALS AND METHODS

Subjects

We performed a single-center prospective study of ten ASD subjects and one healthy control subject at the University Hospitals Leuven campus Pellenberg. The gender, age, length, mass, BMI and radiographic curvature values of each subject are reported in table 1. This study was approved by the ethical committee (S58082, Principal Investigator: Dr. Lieven Moke, MD).

Subject	Gender	Age (years)	Length (cm)	Mass (kg)	BMI (kg/m^2)	TK (°)	LL (°)	Cobb angle (°)
C	F	62	155	34	14.2	54.8	69.2	1.2
1	F	60	160.5	57.1	22.2	48.6	57.3	22.5
2	F	66	162.5	55	20.8	41.5	67.1	23.7
3	F	19	170.5	56.3	19.4	24.7	35.9	12.8
4	F	60	150	37	16.4	-13	13	33.4
5	F	75	154	48.6	20.5	35.1	55	9.3
6	M	51	169	73	25.6	40.7	56.7	24.6
7	F	47	161	67	25.8	49.3	48.4	23.7
8	M	72	172	67.8	22.9	39.9	47	15.4
9	F	69	166	61	22.1	21.8	25.5	18.6
10	F	70	159.5	62	24.4	48.1	40.8	10.8
Mean value and SD	M:F (%) 18.2 : 81.8	59.1 ± 15.1	161.8 ± 6.8	56.3 ± 11.7	21.3 ± 3.5	35.6 ± 18.2	46.9 ± 16.4	17.8 ± 8.6

Table 1: Group demographics and radiographic values, means and standard deviations. C (control subject); BMI (body mass index); TK (thoracic kyphosis); LL (lumbar lordosis)

Inclusion criteria for the patients were: age 18 years or older and the presence of an adult spinal deformity. Subjects were included if they had the ability to walk at least 50 meters independently, i.e. without a walking aid, and scored more than 25 out of 30 on the Mini Mental State Examination. Subjects with a documented neurological disease or vestibular lesion affecting balance or a current history of musculoskeletal disorders of the lower extremities affecting motor performance were excluded.

Data collection

Motion capture

Retro-reflective markers were placed by an experienced physiotherapist. The marker placement protocol consisted of a lower-limb protocol and additional markers on the head and spine to register the positioning of the head and regional spinal curvatures, aiming for detailed movement registration. A detailed figure of the placement of the retro-reflective markers can be found in figure 1.

Each subject performed an instrumented movement capture protocol in the motion analysis lab. Subjects were instructed to perform several movements representative of daily life, of which we will discuss gait in particular. For this study, relevant 3D marker positions were captured during a standing trial and during barefoot walking at a self-selected speed. On average, we collected 24 left and 23 right gait trials, for each subject one static standing trial was recorded (Appendix 1, table 1). 3D motion capture was recorded using a 14-camera VICON System and two embedded force plates in line.

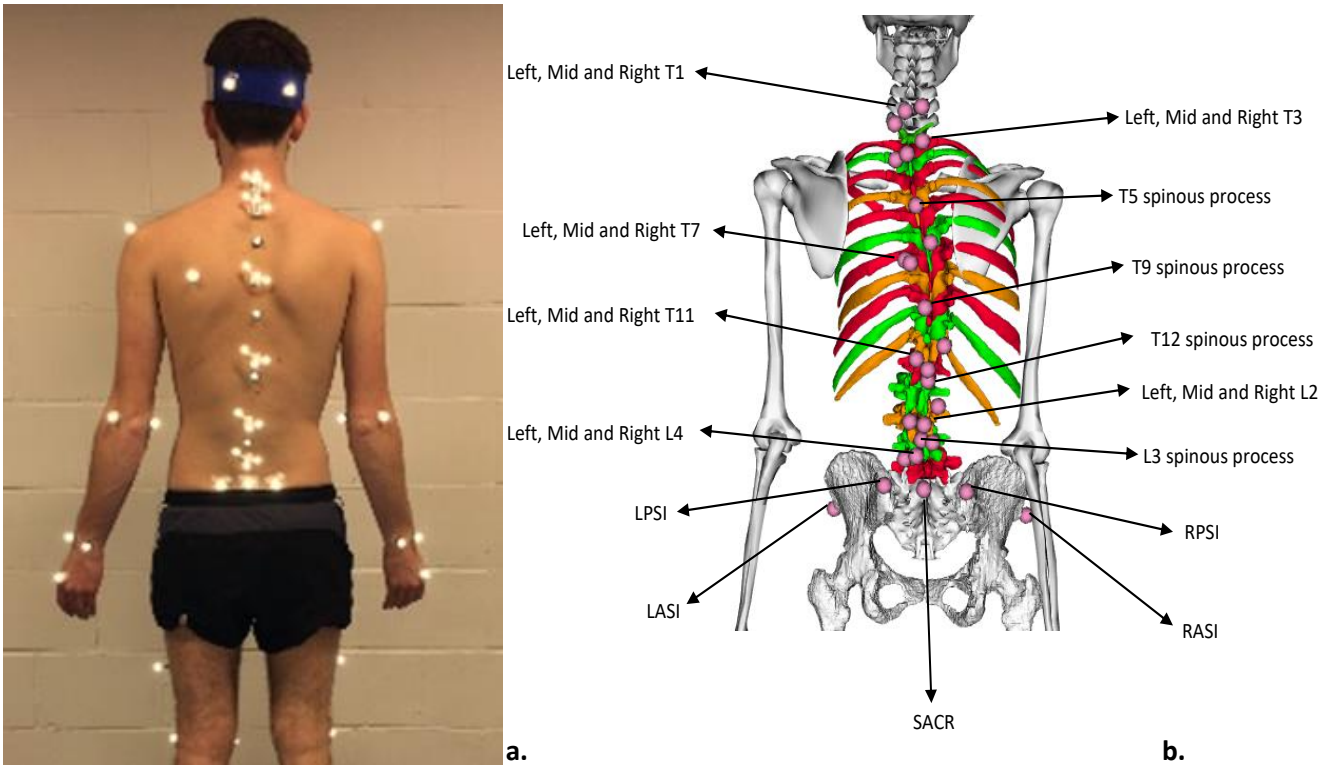


Figure 1a and b: Overview of the retro-reflective markers: clusters of 3 spine markers were placed on the spinous processes of T1, T3, T7, T11, L2 and L4, and single markers on the spinous processes of C7, T5 T9, T12 and L3. Pelvic markers were placed bilaterally on the anterior (LASI, RASI) and posterior (LPSI, RPSI) superior iliac spines. One marker was placed on the sacrum (SACR). Head markers, upper and lower limb markers were used for other purposes but excluded for further analysis. (Image 1a by T. Overbergh)

Medical Imaging

For each subject, a Computed Tomography (CT, BrightSpeed by GE Healthcare, at an inter-slice distance of 1,25 mm, pixel size 0.390625 x 0.390625 mm), Magnetic Resonance Imaging (MRI, Ingenia by Philips Healthcare, pixel spacing: 0.9309 x 0.9309 mm, slice thickness 2 mm) and EOS scan (EOS imaging, Paris, France) were taken. The EOS imaging system is a biplanair radiographic imaging modality with a limited radiographic dose in a load-bearing position[29]. The EOS images were taken simultaneously in the coronal and sagittal plane, allowing for 3D reconstructions. The retro-reflective markers were kept in place during the EOS scan.

Data processing

1. Radiographic analysis

The following clinical parameters were extracted from the radiographic images: pelvic tilt (PT), sacral slope (SS), pelvic incidence (PI), lumbar lordosis (LL), thoracic kyphosis (TK), T1 spinopelvic inclination (T1SPI), T9 spinopelvic inclination (T9SPI), sagittal vertical axis (SVA) and the Cobb angle (figure 2 and 3).

Pelvic incidence was measured as the angle between the line connecting the center of the sacral endplate to the femoral head axis and the line perpendicular to the sacral endplate[30]. ***Pelvic tilt*** was defined as the angle between a vertical reference line originating from the femoral head axis and the line connecting the femoral head axis with the center of the sacral endplate[31]. The ***sacral slope*** was determined as the angle between the tangent line to the superior sacral endplate and a horizontal reference line[32]. ***T1 spinopelvic inclination*** was measured as the angle between a vertical reference line originating from the vertebral body center of T1 and the line drawn from this vertebral body center to the midpoint of the axis through both femoral heads and ***T9 spinopelvic inclination*** was measured as the angle between the vertical line originating from the vertebral body center of T9 and the line connecting this vertebral body center and the axis of the femoral heads[33]. The sign of the T1SPI was determined by the position of the T1 vertical plumb line (figure 3). If the plumb line fell anteriorly to the T1SPI angle, the T1SPI was said to be negative. This corresponds to an anterior position of T1 in respect to the femoral heads, as visualized in figure 3b. Similarly, a positive T1SPI corresponds with a posterior position, with the plumb line posterior to the T1SPI angle. This was also applied to T9SPI. The ***sagittal vertical axis*** was defined as the C7/T1 plumb line relative to the superior sacral endplate[34]. ***Lumbar lordosis*** was determined as the angle between the superior endplate of L1 and the superior endplate of L5 and ***thoracic kyphosis*** was defined as the angle between the superior endplate of T1 and the superior endplate of T12. The ***Cobb angles*** were measured in the coronal plane by drawing tangent lines to the upper border of T1 and the lower border of T12, then erecting perpendiculars from these lines to cross[3].

The radiographic parameters were measured on the radiographic images by an experienced spine surgeon. The measured values for each subject can be found in Appendix 1, table 2.

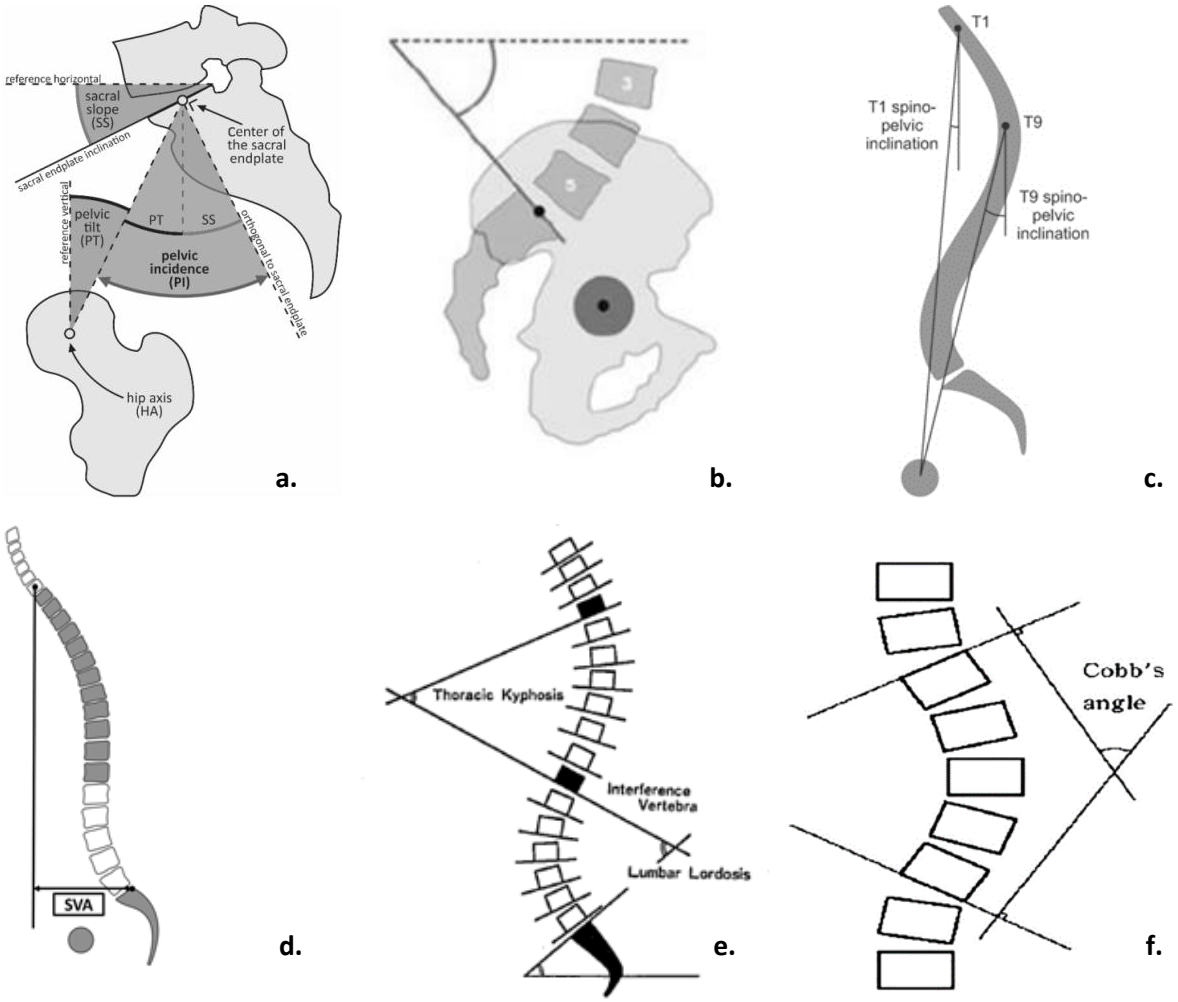


*Figure 2: An overview of the parameters used in this study **a** Pelvic incidence (PI) and Pelvic tilt (PT) (figure from Vrtovec et al.)*[35] ***b*** *Sacral slope (SS) (figure from Geiger et al.)*[32] *There is a geometrical relationship between the pelvic incidence (PI), pelvic tilt (PT) and sacral slope according to the equation: PI = PT + SS*[36]*. **c** T1 spinopelvic inclination (T1SP) and T9 spinopelvic inclination (T9SPI) (figure from Lee et al.)*[37] ***d*** *Sagittal vertical axis (SVA) (figure from Diebo et al.)*[38] ***e*** *Lumbar lordosis (LL) and Thoracic kyphosis (TK) (figure from Itoi et al.)*[39] ***f*** *Cobb angle (figure from Cheon et al.)*[40]

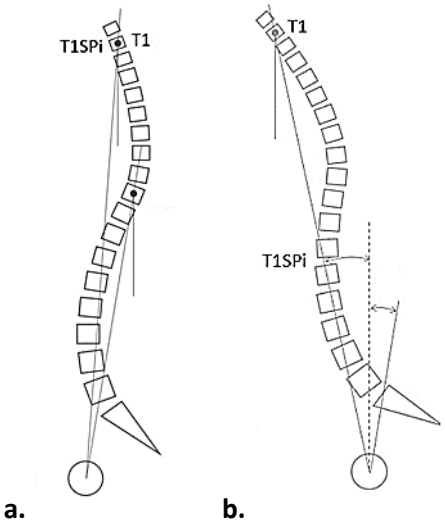


Figure 3: (a) positive T1SPI (b) negative T1SPI (figures from Park et al.)[41]

2. *Dynamic analysis*

The first purpose of this study is to investigate whether 2D radiographic measurements (radiographic values), the current method for evaluation and surgical planning, are representative for 3D measurements during a standing trial (static values). The differences between static and radiographic values (Δstat-rad) were calculated for each subject and each parameter.

The second purpose is to compare 3D parameter values during standing (static values) and walking (dynamic values). To compare whether a difference between the static and dynamic value (Δstat-dyn) was relevant, we used a cut-off score based on the standard deviation of the control subject. Differences higher than the standard deviation were considered as relevant. Positive values represent an increase in parameter value from static to dynamic conditions, negative values represent a decrease.

2.1 Modelling

CT and MRI images were imported into Mimics 19.0 (Materialise NV, Leuven, Belgium) for intensity-based segmentation. We segmented the CT and MRI images into individual vertebrae and pelvis for each patient (sacrum to thoracic vertebra 1) (Appendix 1, figure 1). These individual vertebrae and pelvis were manually registered on the EOS images, thereby creating a subject-specific skeletal model. After personalization, the intervertebral joints were redefined using a mesh-based centralization method, which places the joints in the center of the intervertebral discs. Each vertebral joint was reduced to a three-degrees-of-freedom rotational joint, i.e. a pin joint inhibiting translational movements. The skin markers of the MOCAP were reconstructed from the images and thereafter added to the subject-specific skeletal model for verification of their position in relation to the anatomical landmarks through EOS radiographic imaging. This resulted in a 3D OpenSim model, made subject-specific at the level of the bones and markers[42].

2.2 Motion capture data

The retroreflective markers from the movement capture protocol were labeled in Vicon Nexus 2.6 and preprocessed, which resulted in motion capture data (.*trc*-files).

2.3 Dynamic data processing

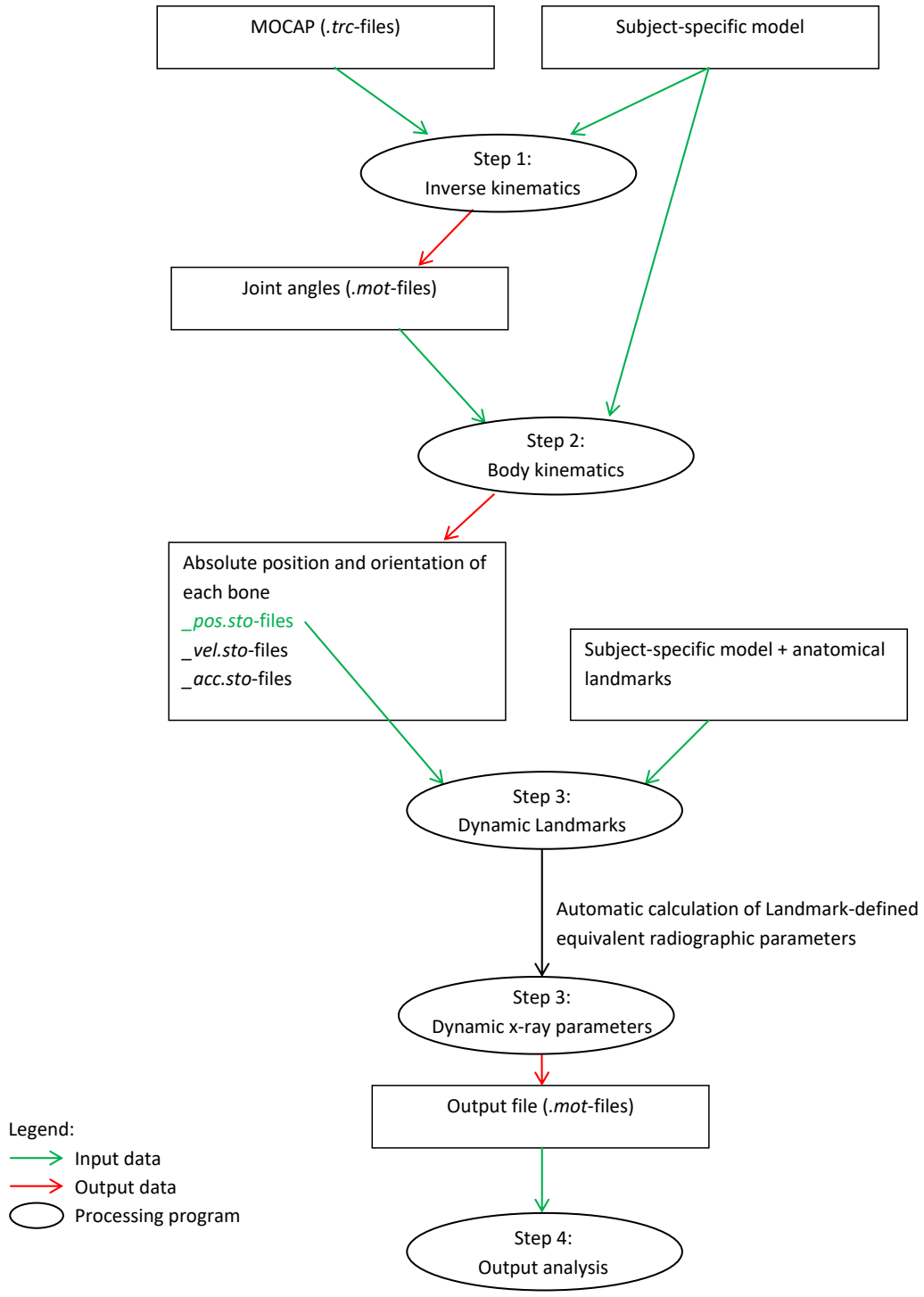


Figure 4: Flowchart of the dynamic data processing

The data for dynamic evaluation was acquired in several steps.

Processing step 1: Inverse Kinematics

First of all, the models and MOCAP data were loaded in the Inverse Kinematics algorithm of OpenSim 3.3[42]. This program provided an estimate of the relative joint angles during the gait cycle (GC). Thereafter, the individual gait cycles were time-normalized, i.e. ranging from 0 to 100%.

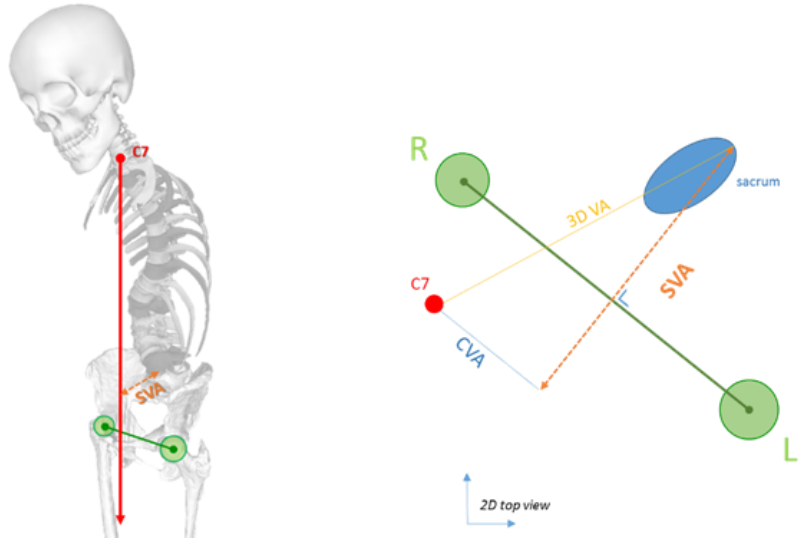
Processing step 2: Body Kinematics

The joint angles, together with the model, were batch processed using the Analyze tool of OpenSim to obtain the Body Kinematics, providing the absolute position and orientation (*_pos.sto*-files), the velocity (*_vel.sto*-files) and the acceleration (*_acc.sto*-files) of each bone expressed in the ground reference frame.

Processing step 3: Dynamic Landmarks

In the next step, custom-made software (Dynamic Landmarks) was used to obtain the absolute trajectory of indicated anatomical landmarks on the model throughout the motion (Appendix 1, figure 2). An overview of the anatomical landmarks can be found in Appendix 1, figure 3. The subject-specific skeletal model with the anatomical landmarks and the Body kinematics files (*_pos.sto*-files) was imported into Dynamic Landmarks, which was coupled with a program to automatically calculate the dynamic radiographic parameters (Appendix 1, figure 2). It took a specific combination of inserted dynamic landmarks for each desired dynamic parameter. For example: to determine the pelvic tilt, both femoral head landmarks and the superior anterior and posterior sacral endplate were required (Appendix 2).

This procedure was repeated for the left and right gait trials and a static stance trial. The program was then able to calculate the requested angles and distances of the equivalent radiographic parameters. These were achieved by projecting the actual measured anatomical landmarks onto the sagittal plane, so that calculations were expressed in a 2D plane, equivalent to the way they were measured by the spine surgeon (figure 5). Each processed gait trial resulted in an output file containing the time-expressed parameters. (*.mot*-files)



a. **b.**

Figure 5: (a) representation of the SVA in the 3D plane (b) visualization of the projection of the actual measured anatomical landmarks of SVA onto the sagittal plane in the program, Dynamic Landmarks. (Images by T. Overbergh)

Processing step 4: Output Analysis

After this process, the data was processed using a custom program called Output Analysis. Uploading the output file here resulted in visualization of a graph showing the changes of the parameter over time for each trial. The data quality was visually assessed. Discontinuous trials, originating from incomplete marker data, were removed. Next, the program calculated the average, minimum, maximum and average ROM for each parameter of all subjects (figure 6). Average values from left gait cycles and static trial data were used for further processing.

As an example, figure 6 shows the lumbar lordosis of subject 3 during the average gait cycle. These graphs were created for each subject and for each parameter. We will apply the terms *static* and *dynamic* values to describe the results. *Static* and *dynamic* values were calculated as the averages of values obtained throughout the static trial and the gait trials, respectively. Figure 6 shows an explanation of how we obtained the *dynamic* values.

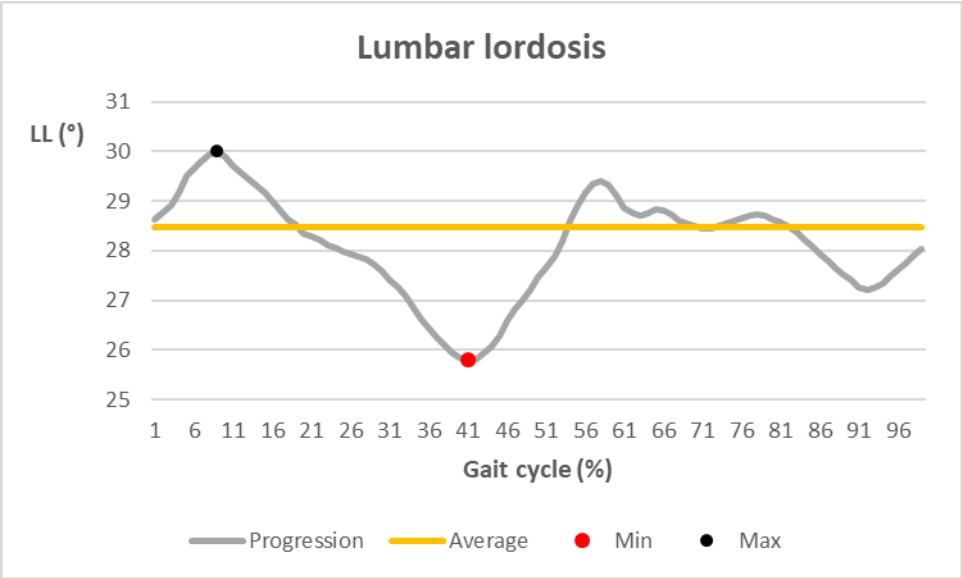


Figure 6: Example of gait cycle and average values for lumbar lordosis (LL). The grey progression curve was calculated as the average of all left gait trials of this subject. The yellow line represents the average value of the progression graph, which we will call the dynamic value. The black and red dots represent the respective maximum and minimum value of the progression curve.

RESULTS

General findings on the relationship between static, dynamic and radiographic parameters

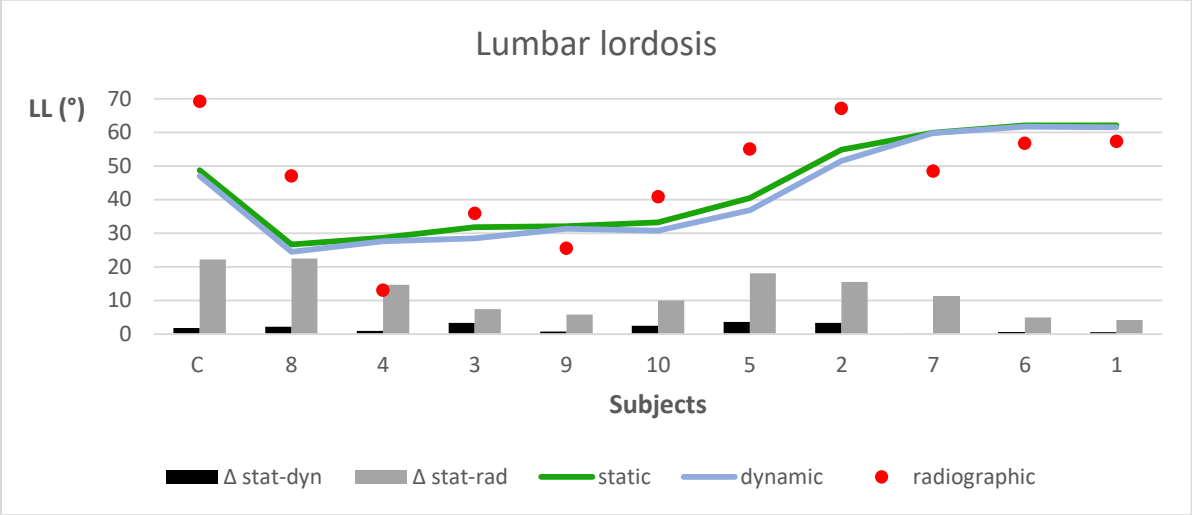


Figure 7: Comparison of static, dynamic and radiographic values for lumbar lordosis (LL). The red dots represent the values measured on the radiographic images. The static values are visualized by a green line, the dynamic values by a blue line. The grey bars show the size of the differences between static and radiographic values. The discrepancy between standing (static) and walking (dynamic) values is shown as black bars.

Figure 7 shows a comparison of the radiographic, static and dynamic values for the lumbar lordosis of each subject. Subjects were placed according to the size of the average static value, with the control subject on the left. In 6 of the subjects, including the control subject, the radiographic value was higher than the static value, indicating an overestimation of the 3D values by the radiographic value. In the remaining 5, the static value was higher, indicating an underestimation of the dynamic function by the radiographic value. The difference between static and radiographic values ranged from 4.2 to 22.5 degrees.

The difference between the 3D static and dynamic values ranged from 0.1 to 3.6 degrees. The static values were higher than the dynamic lumbar lordosis for each of the subjects, indicating a loss of lumbar lordosis from static to dynamic conditions.

Changes in parameter values from static to dynamic conditions

Figure 8 shows a more detailed overview of the previously mentioned differences between static and dynamic values. Subjects were sorted as function of increased ΔSVA.

Sagittal vertical axis (SVA)

All subjects presented a larger SVA during walking than in standing posture. The difference ranged from 1.5 to 4.5 cm. In the control subject, a difference of 2.2 ± 0.6 cm was seen.

Thoracic kyphosis (TK)

The thoracic kyphosis curve did not show a clear trend. The ΔTK ranged from -1.5 to 14.1 degrees. 6 out of the 10 ASD patients did not present a manifest change in thoracic kyphosis when the static trial was compared to the dynamic ones. All four of the other ASD patients and the control subject demonstrated a larger TK during walking than during the static trial. The control subject showed a ΔTK of 8.1 ± 2.8 degrees.

T1 spinopelvic inclination (T1SPI)

Figure 3 documents the spinopelvic parameter T1SPI in detail. All differences for T1SPI are negative, indicating a more forward position of T1. There was a positive relationship between ΔT1SPI and ΔSVA. The ΔT1SPI varied from -1.5 to -6.8 degrees. In the control subject, the ΔT1SPI was -2.9 ± 0.9 degrees.

T9 spinopelvic inclination (T9SPI)

No clear relation with ΔSVA was observed. For 10 of the subjects, including the control subject, there was a relevant decrease in T9SPI, indicating a more forward position of vertebra T9 during gait. The ΔT9SPI ranged from -0.9 to -6.9 degrees. For the control subject, the T9SPI decreased by 2.5 ± 1.0 degrees from static to dynamic conditions.

Pelvic tilt (PT)

The pelvic tilt decreased during walking in 8 subjects, indicating an increase in pelvic anteversion during gait. The ΔPT ranged from -6.1 to 0.9 degrees. In the control subject, a difference of -1.3 ± 1.1 degrees was seen. There were no relevant changes in the three remaining subjects.

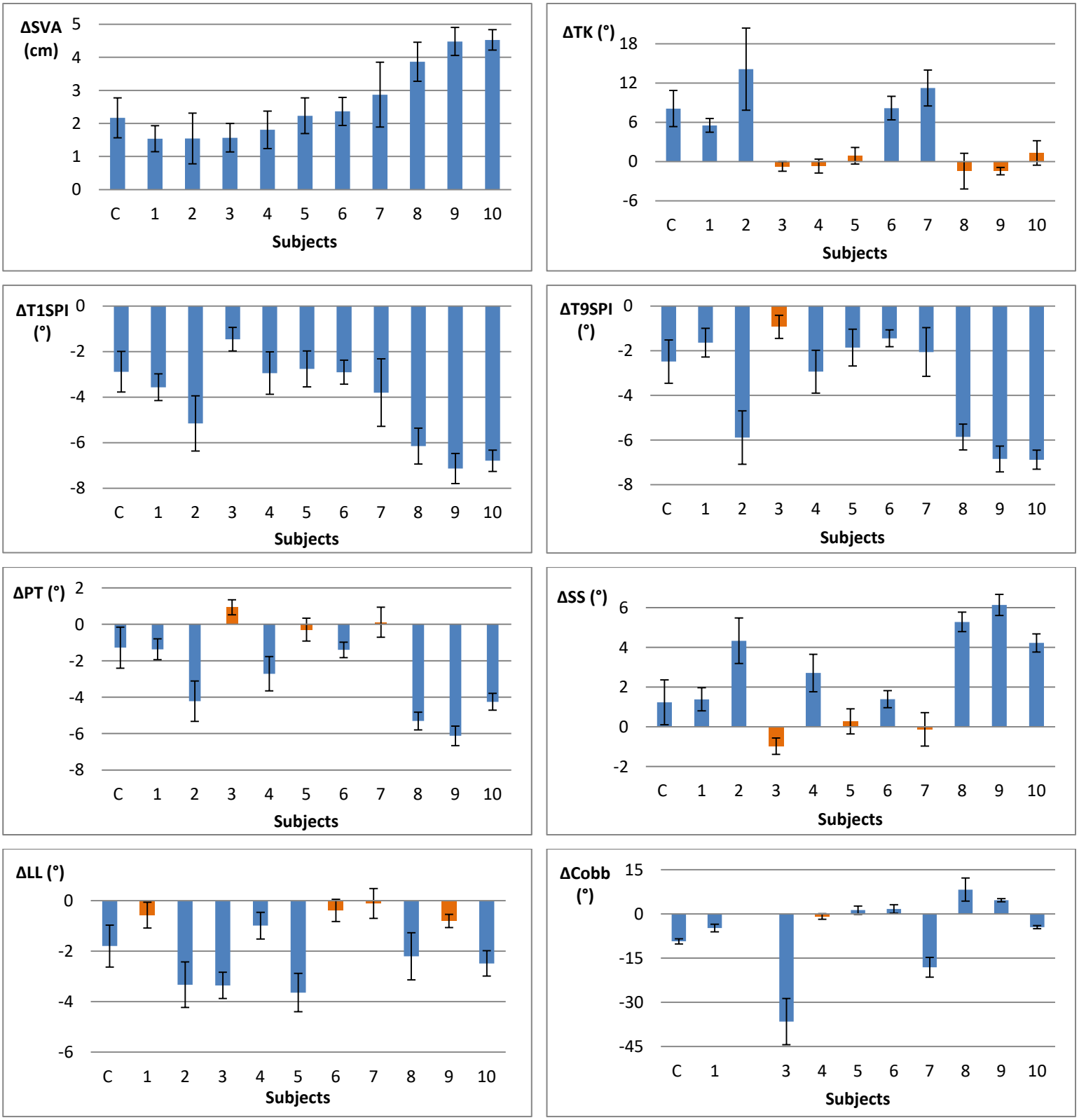


Figure 8: Changes and standard deviations in spinopelvic parameter values from static (standing) to dynamic (walking) conditions. Clinical relevant changes are indicated in blue, irrelevant changes are indicated in orange. Positive values represent an increase in parameter value from static to dynamic conditions. Negative values represent a decrease. The control subject is always on the left. Control subject (C); difference (Δ); sagittal vertical axis (SVA); T1 spinopelvic inclination (T1SPI); thoracic kyphosis (TK); T9 spinopelvic inclination (T9SPI); sacral slope (SS); lumbar lordosis (LL); pelvic tilt (PT).

Sacral slope (SS)

With increasing SVA differences, there was an increase in the ΔSS, indicating an increase in pelvic anteversion during gait. The ΔSS ranged from -1.0 to 6.1 degrees. Three of the subjects showed no relevant difference between static and dynamic values. The sacral slope increased from standing to walking in the eight remaining subjects. The control subject showed a ΔSS of 1.2 ± 1.1 degrees.

Lumbar lordosis (LL)

The ΔLL ranged from -0.1 to -3.6 degrees and did not follow a clear pattern when compared to the ΔSVA. In four subjects, there was nearly no difference. In those subjects where we did find a change, the lumbar lordosis was always higher in the static trial than during walking, indicating a loss of lumbar lordosis during walking. For the control subject, ΔLL was -1.8 ± 0.8 degrees.

Cobb angle

The differences of the Cobb angles did not show a clear relationship with the SVA scores. The ΔCobb ranged from -36.6 to 8.2 degrees. For subject 2, the Cobb angle could not be calculated. One subject showed no relevant changes. Five of the other subjects, including the control subject, showed a decreased Cobb angle during walking. For the control subject, this was a decrease of 9.2 ± 0.9 degrees. The 4 remaining subjects showed an increase in the Cobb angle during walking.

DISCUSSION

In the present study, we investigated to what extent static alignment can represent the dynamic function of the spine in ASD patients during gait. We will discuss this research question in two different sections.

To what extent can 2D radiographic measurements predict 3D static alignment of the spine in ASD patients?

Our results indicated that radiographic measurements alone are not representative for the dynamic spine function in ASD patients, which already partially confirms our hypothesis.

As visualized in figure 7 for the lumbar lordosis, the static spinal alignment calculated based on the radiographs (2D) and the static motion capture trial in combination with the subject-specific skeletal model were considerably different, which was the case for all parameters. This indicated that the static 2D representation observed in a radiograph was not as informative as the 3D spinal alignment measured from the static motion capture combined with a subject-specific musculoskeletal model. This emphasizes the need for additional analyses (i.e. EOS acquisition + motion capture).

The difference between both values benchmarks the true difference between the radiographic parameters measured on 2D images, and the static trial evaluated based on 3D images. However, a high inter- and intra-rater variability during the measurements of the radiographic parameters, as previously concluded in a study of Khalsa et.al, who reported high variability in the assessment of spinopelvic parameters in patients with lumbosacral transitional vertebrae using radiographs, may have been a contributing factor in this analysis[43]. Indeed, the measurement of radiographic parameters is strongly susceptible to subjectivity during the placement of the lines by which a parameter is determined, with even the slightest change having a large impact on the result (Appendix 3, figure 1).

To what extent does 3D static alignment represent the 3D dynamic function during gait in ASD patients?

3D static alignment clearly differed from dynamic function during gait, as can be concluded based on the 3D dynamic parameter differences, visualized in figure 8. This finding is important as it confirms our hypothesis and supports the clinical need for dynamic evaluation of spinal function in patients with ASD to identify compensation strategies during dynamic function. Detailed images and descriptions of individual compensatory mechanisms are presented in Appendix 3, table 1. We will discuss below the general compensatory mechanisms observed in this cohort of ASD patients.

During gait, most subjects walked with a more forward tilted trunk as reflected by the SVA, representative of the trunk alignment with the vertical axis, T1SPI and T9SPI, both representative of the trunk alignment with the femoral heads. However, each of these parameters described a specific characteristic of trunk alignment, as will be discussed below.

The observed differences in the sagittal vertical axis (SVA) during dynamic function are in accordance with recent literature[4,43]. Based on the work of Kyrölä et al, an SVA above 4,6 cm is considered a threshold for serious deterioration[36,44]. Using this threshold for classification based on the 3D static measures, only two subjects of our cohort (subjects 9, 10) exceeded this value. However, considering SVA during gait, five subjects would be classified as having a serious deterioration (subjects 4, 7, 8, 9, 10). This supports our hypothesis that 3D dynamic information provides additional insights into the evaluation of ASD patients.

This increase in sagittal vertical axis affected the positioning of T1 and T9 with respect to the femoral heads (as reflected by T1SPI and T9SPI):

A similar magnitude of change in SVA and T1SP1 was observed. However, the variability between subjects was large: for the majority of the subjects (subjects 2, 3, 4, 6, 8, 9) the location of T1 changed from a posterior position during the standing trial to an anterior position during walking. In subjects 7 and 10, the already anteriorly located T1 during stance moved even more anteriorly during walking; in only two subjects (subjects 1, 5) a posteriorly positioned T1 during the standing trial was normalized during walking. The more anteriorly located T1 during walking might be a consequence of the forward tilting thorax, as the subjects did not have a clear sight unless they protracted their heads – resulting in the displacement of T1.

The magnitude of change in SVA and T9SP1 was more variable. In contrast to T1, the T9 vertebral body was located posteriorly to the femoral heads in both standing and walking conditions. During walking, T9 moved more anteriorly with respect to the femoral heads. The observed anterior shift of T9 can be related to two mechanisms: Firstly, a more anterior position of a rigid thoracic spine, as reflected by an equal magnitude of change in position of T1 and T9. This mechanism was present in 6 subjects (subjects 3, 4, 5, 8, 9, 10). Secondly, an increase in pelvic anteversion, as reflected by the larger sacral slope during walking. This was present in all but three subjects (subjects 3, 5, 7). With increased anteversion, the femoral heads were positioned more posteriorly. As a result, the distance between the femoral heads and the T9 vertebral body decreased. This was also confirmed by the observation that the highest changes in T9SPI matched the highest changes in the sacral slope. Indeed, previous literature found that the pelvic retroversion used by ASD patients to compensate

for their sagittal malalignment is reduced during walking[36,45]. This supports our findings that ASD patients lose the compensatory retroversion during walking – but present an increased anteversion.

The relation between the more forward tilted trunk (as reflected by SVA) and the changes in thoracic curvature (as reflected by TK) was not clear. Six of the subjects walked with a more forward tilted trunk, but presented no relevant change in the curvature of the thoracic spine, only a rigidly displaced thoracic spine. In the remaining four subjects (subjects 1, 2, 6, 7), this curvature change between standing and walking was relevant. Three subjects (subjects 1, 6, 7) walked with increased head protraction, but presented only a small change at the lower thoracic region – thus were more flexible, which increased the curvature of the spine.

Most of the subjects walked with a more flattened lumbar back (represented by the decrease in LL). There was no relationship with the pelvic anteversion, although previous literature stated that the use of anteversion increases the lumbar lordosis in asymptomatic subjects[36,46]. In fact, this relationship was reversed in our study, as most subjects walked with more anteversion combined with a more flattened lower back. The effects of the increased pelvic anteversion might be more pronounced at the level of the lower limbs, as discussed by Ferrero et al[47].

The changes in the coronal curve (represented by the Cobb angle), the only parameter measured in a different plane, seemed not to relate to the sagittal parameters. This was not unexpected. In this study, the Cobb angles were measured at a fixed level (i.e. as the angle between vertebrae T1 and T12), instead of the more common method to determine the angle based on the level of the most-tilted vertebrae. The motivation to use a fixed level was standardization across subjects. Furthermore, the results of the Cobb angles may not have been fully representative for the true deviations due of the inability of the Cobb angle measurement to account for coronal deviation around the thoracolumbar region, as present in at least one of the patients. This was a limitation to this study.

Nevertheless, the use of the Cobb angle determined at the level of the highest deviation as defined in the literature should be explored in future work. Preliminary analysis performed on one of the subjects in this study (subject 8), resulted in increased Cobb angles, but also indicated less differences between standing and walking values. This result indicates the relevance for further exploring the relationship between sagittal and coronal parameters (Appendix 4, figure 1).

As mentioned before, the present study had several limitations. In addition to the measurement of the Cobb angle discussed above, the control subject presented large sagittal curvature values and an excessively straight posture. This may raise some concern about the representativity. Nevertheless,

the ASD subjects presented with comparable compensation strategies as the control subject, independent of the questionable representativity of the values present in the control subject. We were unable to perform statistics, because of a small and heterogeneous group. In future research, we would recommend implementing a more homogenous and larger group. If statistics could be implied, we would opt for the Spearman test to measure the correlation between the radiographic and static/dynamic parameters and make a comparison between the pathological and the non-pathological group by using an independent T-test or the Mann-Whitney U-test. In subject 2, an outdated marker protocol was used, which missed a cluster marker for the T1 vertebral body – causing a large standard deviation, an incorrect 3D subject-specific skeletal model and unreliable results for the parameters based on T1. Subject 5 lacked the anatomical landmark of T1, because of unclear imaging. We used the T2 body to determine the necessary parameters for this subject, which influenced the parameters based on the T1 vertebral landmark. Figure 6 shows how the used average values were a simplification of the dynamic behavior of the spinopelvic parameters, causing an underestimation of the intra-subject variability. Nevertheless, extreme values in ROM due to outliers in motion performance were therefore excluded and consistency was provided in the comparative evaluation of subjects and parameters. As mentioned earlier, another limitation was the lack of integration of other body parts. The lower limbs specifically, are known to play an important role in the compensation strategies in adult spinal deformity[36,47].

CONCLUSION

2D radiographic measurements alone are insufficient to represent the 3D alignment of the spine. 3D static evaluation seems to be more representative for the dynamic function of the spine during gait. Nevertheless, ASD patients compensate for their sagittal malalignment during gait by increasing pelvic anteversion, sacral slope and a forward tilted trunk. Using 3D dynamic subject-specific information provides improved insights into patients' compensation strategies. Future research with larger and more homogeneous study samples will provide more in-depth insights.

CONFLICT OF INTEREST

There was no conflict of interest regarding this article

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