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Sagittal Range of Motion of the Thoracic Spine Using Inertial Tracking Device and Effect of Measurement Errors on Model Predictions

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ABSTRACT

1 Range of motion (ROM) of the thoracic spine has implications in patient discrimination for
2 diagnostic purposes and in biomechanical models for predictions of spinal loads. Few previous
3 studies have reported quite different thoracic ROMs. Total (T1-T12), lower (T5-T12) and upper
4 (T1-T5) thoracic, lumbar (T12-S1), pelvis, and entire trunk (T1) ROMs were measured using an
5 inertial tracking device as asymptomatic subjects flexed forward from their neutral upright
6 position to full forward flexion. **Correlations between body height and the ROMs were**
7 **conducted. Effect** of measurement errors of the trunk flexion (T1) on the model-predicted spinal
8 loads **was** investigated. Mean of peak voluntary total flexion of trunk (T1) was $118.4 \pm 13.9^\circ$, of
9 which $20.5 \pm 6.5^\circ$ was generated by flexion of the T1 to T12 (thoracic ROM), and the remaining
10 by flexion of the T12 to S1 (lumbar ROM) ($50.2 \pm 7.0^\circ$) and pelvis ($47.8 \pm 6.9^\circ$). Lower thoracic
11 ROM was significantly larger than upper thoracic ROM ($14.8 \pm 5.4^\circ$ versus $5.8 \pm 3.1^\circ$). **There**
12 **were non-significant weak correlations between body height and the ROMs.** Contribution of
13 the pelvis to generate the total trunk flexion increased from ~20 to 40% and that of the lumbar
14 decreased from ~60 to 42% as subjects flexed forward from upright to maximal flexion while
15 that of the thoracic spine remained almost constant (~16 to 20%) during the entire movement.
16 Small uncertainties ($\pm 5^\circ$) in the measurement of trunk flexion angle resulted in considerable
17 errors (~27%) in the model-predicted spinal loads only in activities involving small trunk
18 flexion.

INTRODUCTION

Though less frequent than low back and neck pain, thoracic spine pain has been identified as a significant work-related disease (Briggs et al., 2009; Fouquet et al., 2014). A large population study indicates that 20% of women and 10% of men workers suffer from thoracic pain (Fouquet et al., 2015). Frequent trunk bending has been suggested to be a strong occupational risk factor for thoracic spine pain (Roquelaure et al., 2014). Consequently, surgeries on the thoracic spine have increased according to a nationwide cohort study in Japan (Nohara et al., 2004). As musculoskeletal disorders limit joint movements and thus its range of motion (ROM), motion analysis of the thoracic spine can serve as a **potential** tool for patient discrimination and subsequent diagnostic purposes. **It has been also suggested that assessment of the thoracic spine ROM can help in the diagnosis and therapy of patients with shoulder outlet impingement syndrome (Theisen et al., 2010).**

Evaluation of sagittal ROM of the thoracic spine (i.e., maximal relative flexion of T1 to T12) is also valuable in the biomechanical modeling for prediction of muscle forces and spinal loads (Arjmand and Shirazi-Adl, 2006, Arjmand et al., 2011 and 2012). Based on the general assumption of the relatively smaller sagittal ROM of the thoracic spine compared with the lumbar spine, thoracolumbar musculoskeletal models (e.g., Arjmand and Shirazi-Adl, 2006; Cholewicki and McGill, 1996; Granata and Wilson, 2001; Stokes and Gardner-Morse, 1995) as well as commercial software such as the AnyBody^(R) Modeling System (Damsgaard et al., 2006) assume the whole T1-T12 thoracic spine to move rigidly as a single body. Ligamentous passive contribution of the thoracic spine in balancing external moments is hence neglected in these models.

Generally three techniques have been used to measure sagittal ROM of the thoracic spine, including skin-surface devices such as optical marker-video camera (Tully and Stillman 1997), electronic inclinometer (Mannion et al., 2004), or electromagnetic tracking (Hsu et al., 2008; Troke et al., 1998; Willems et al., 1996), computed tomography (CT) imaging (Morita et al., 2014), and *in vitro* cadaveric investigations (White and Panjabi, 1990). These studies have reported quite different sagittal thoracic (T1-T12) ROMs varying from ~18 to 32° for forward flexion alone (Hsu et al., 2008; Mannion et al., 2004; Tully and Stillman 1997; Willems et al.,

1996) and from ~32 to 70° (Morita et al., 2014; O'Gorman and Jull, 1987; Troke et al., 1998; White and Panjabi, 1990; Willems et al., 1996) for total flexion-extension. The video camera, medical imaging, and *in vitro* techniques require equipped laboratories and thus are impractical for easy and comprehensive use in workplaces for ergonomics and in hospitals for clinical applications. The electromagnetic tracking devices are accurate, but usually confined to a limited space to allow the “sensors” to detect the magnetic field generated by a fixed “source” (Saber-Sheikh et al., 2010). Moreover, their accuracy and calibration process is adversely affected by the presence of metallic objects in the experiment environment or of implants in patients (Ng et al., 2009).

We have recently used an inertial tracking device (Xsens MTx, Xsens Technologies, Enschede, Netherlands) for measurement of lumbopelvic rhythm during trunk flexion-extension (Tafazzol et al., 2014). **Compared to other skin-surface methods**, inertial tracking devices are source-less (no cameras), low-cost, light, portable and, hence, indispensable easy-to-use tools for the quick recording of the spinal ROMs. This study, hence, has **two** aims:

- (1) Measure total (T1-T12), lower (T5-T12) and upper (T1-T5) thoracic, lumbar (T12-S1), pelvis (P), and total trunk (T1) ROMs and their movement rhythms in the sagittal plane as asymptomatic subjects flex forward from their neutral upright posture to full forward flexion in unconstrained upright standing using an inertial tracking device. **Correlations between body height and the foregoing ROMs were also conducted.**
- (2) Quantify the effect of inherent errors (uncertainties) in the *in vivo* measurements of upper trunk flexion angle (T1) on the responses (i.e., predicted spinal loads) of a previously developed musculoskeletal finite element model of the thoracolumbar spine (Arjmand and Shirazi-Adl, 2005 and 2006; Arjmand et al., 2009, 2010, 2011 and 2012). **In other words, this modeling study aims to investigate the sensitivity of biomechanical models to the measured trunk flexion angle that is required as input into these models.** As magnitude of trunk flexion angle affects the trunk posture, position of its center of mass, and thus trunk external moments, it is hypothesized that errors in the measurements of trunk flexion angles will have an impact on the model predictions for spinal loads.

METHOD

Inertial tracking device: Four small (38 mm × 53 mm × 21 mm) and light (~30 g) inertial and magnetic sensors (Xsens MTx, Xsens Technologies, Enschede, Netherlands) are used to capture the 3D rotations of the pelvis, lumbar and thoracic spine. Each sensor has triaxial accelerometers, gyroscopes, and magnetometers whose signals are fused using a Kalman filter to estimate drift-free 3D orientations of the sensor coordinate system with respect to the reference earth-fixed coordinate system which, in turn, is created using information from the inertial and magnetic sensors (Tafazzol et al., 2014). Sensor data are transferred to a PC through Bluetooth technology at 100 Hz sampling rate. Before starting the trials, accuracy of the inertial sensors was assessed by measuring the angles between two (fixed and movable) arms of a goniometer as described elsewhere (Tafazzol et al., 2014). The error (difference between the known angle of the goniometer arms and the measured angle by the inertial sensors) was found less than 0.6° indicating that the device can be used for accurate measurement of the spine kinematics.

Subjects and protocol: Forty young healthy male students with no history of back surgery or recent back, hip or knee complications volunteered for the study (Table 1). They signed an informed consent form after being familiarized with the test. Inertial sensors were securely attached to the overlying skin of the T1, T5, T12, and S1 spinous processes using double-sided tapes to minimize movement of the sensors with respect to the underlying skin (Figure 1). After warming up by performing angular movements in all three anatomical planes including full flexion and extension for 2-3 minutes, participants were requested to flex forward (to their maximum voluntary flexion) from a neutral upright position with their feet shoulder width apart and their knees extended (three trials). Subjects were instructed to move slowly at their own pace (4-5 seconds to reach full trunk flexion) and remain in the sagittal plane.

Data analysis: The device measures real-time 3D orientation of the sensors coordinate systems with respect to a global fixed coordinate system (S_R) (Tafazzol et al., 2014). As the orientation of the sensor unit coordinate system (S) is not necessarily aligned with that assumed for the body

segment (B), a body segment to sensor orientation matrix (${}^B_S R$) needs to be defined (Figure 1). Orientation of a body segment frame (B) (i.e., T1, T5, T12, and S1) with respect to the global earth-fixed frame (G) is thus computed as ${}^B_G R = {}^S_G R {}^B_S R$. Relative rotation of each body segment to the initial upright posture ($t = 0$) (${}^B_G R|_r$) can be computed as ${}^B_G R|_r = {}^B_G R {}^B_G R^{-1}_{t=0}$ where ${}^B_G R_{t=0}$ is the unknown orientation of the body segment with respect to the global coordinate system. Subsequently, and to eliminate this unknown matrix one can write ${}^B_G R|_r = ({}^S_G R {}^B_S R)({}^B_S R^{-1}_{t=0} {}^S_G R^{-1}_{t=0})$. It is assumed that no movement occurs between the sensors and underlying skin (sensors are attached firmly to the skin), i.e., ${}^B_S R {}^B_S R^{-1}_{t=0} = I$ where I is the unit (identity) matrix. Relative rotation of each body segment to the initial upright posture can thus be measured as ${}^B_G R|_r = {}^S_G R {}^S_G R^{-1}_{t=0}$. The anatomical angles were computed based on the direction cosine matrices (Grood and Suntay, 1983; Tafazzol et al., 2014). Changes in orientation of the T1 with respect to the global coordinate system, S1 with respect to the global coordinate system, T12 with respect to the S1, T1 with respect to the T5, T5 with respect to the T12, and finally T1 with respect to the T12 as subject flexes forward from upright posture to full forward flexion are considered, respectively, as total trunk (T1), total pelvis (P), total lumbar (L), upper thorax (T1-T5), lower thorax (T5-T12), and total thorax (T1-T12) rotations (ROMs) in the sagittal plane. **It is to be noted that for sagittally symmetric (two-dimensional) movements one may assume that total trunk (T1) flexion is equal to algebraic summation of pelvis (P), total lumbar (L), and total thorax (T1-T12) flexion (Tafazzol et al., 2014). Pearson's Correlation analyses between the measured ROMs and body height were conducted to measure their degree of linear relationship.** Moreover, paired t-tests were conducted to compare T1-T5 and T5-T12 ROMs as well as total lumbar (L) and total thoracic (T1-T12) ROMs.

Biomechanical modeling: To investigate the effect of inherent errors in the *in vivo* measurements of trunk flexion angle on the responses of biomechanical models (i.e., predicted spinal loads) a previously developed kinematics-driven musculoskeletal finite element model of the spine was used (Arjmand and Shirazi-Adl, 2005 and 2006; Arjmand et al., 2006, 2009, 2010, and 2011). The model uses an optimization algorithm to estimate trunk muscle forces and spinal loads while satisfying equilibrium requirements at all spinal joints. Inputs of the model are

posture-dependent variables including trunk sagittal flexion (T1) and lumbopelvic ratio (L/P, the ratio of lumbar sagittal rotation over pelvic sagittal rotation for a given trunk flexion angle) as well as load-dependent variables including the magnitude of handled load (M) and its sagittal position with respect to the shoulder joint (D) (Arjmand et al., 2011). In the present study, the model was run to predict compressive and shear loads at the L5-S1 disc during several static symmetric activities in the sagittal plane with and without 20 kg load in the hands ($M = 0$ or 20 kg) and arms in the gravity direction ($D = 0$) at different trunk flexion angles (20, 40, 60, 80, and 100°) with two levels of trunk flexion uncertainty ($\pm 5^\circ$) at each trunk flexion angle. Magnitude of the lumbopelvic ratio for each trunk flexion angle and its two levels of uncertainties was considered based on the measured mean values of all subjects in this study.

RESULTS

In vivo study: Mean of peak voluntary total flexion of trunk (T1) was 118.4° (SD 13.9°), of which 20.5° (SD 6.5°) was provided by the relative angular movement of the T1 to T12 (i.e., T1-T12 ROM), and the remaining by relative rotation of the T12 to S1 (i.e., lumbar spine ROM) (50.2°, SD 7.0°) as well as pelvis (P) (47.8°, SD 6.9°). Lower thoracic spine (T5-T12) contributed more to the total thoracic (T1-T12) ROM as compared with the upper thoracic spine (T1-T5) (14.8°, SD 5.4° versus 5.8°, SD 3.1°) ($p < 0.0001$). Thoracic spine ROM (T1-T12) was significantly smaller than lumbar spine ROM ($p < 0.0001$). **Non-significant weak correlations were found between upper trunk ($r = 0.25$, $p = 0.124$), T1-T12 thoracic ($r = 0.21$, $p = 0.190$), lumbar ($r = 0.006$, $p = 0.973$), and pelvis ($r = 0.19$, $p = 0.305$) ROMs and body height.** Mean lumbopelvic ratio (L/P, calculated as the total lumbar rotation to the total pelvis rotation) for 10 intervals of trunk flexion (i.e., 10%, 20%,..., and 100% of full trunk flexion) decreased from 2.96 to 1.07 as trunk flexed forward (Figure 2). While the lumbar spine and pelvis contributed more to the trunk flexion during, respectively, early and final stages of forward flexion, both upper (T1-T5) and lower (T5-T12) thoracic spine had a simultaneous movement with an almost constant trend during the entire movement (Figure 2). Contribution of the pelvis to generate the total trunk flexion increased from ~20 to 40% and that of lumbar decreased from ~60 to 42% as subjects flexed forward from upright to maximal flexion while that of the thoracic spine (T1-

T12) remained almost constant (~16 to 20%) during the entire forward flexion movement (Figure 3).

Modeling study: While a 5° error in the measurement of total trunk flexion (T1) resulted in considerable errors in the predicted spinal loads at small trunk flexion angles (e.g., up to 27% error for L5-S1 compression at 20° trunk flexion without hand load), it generated negligible errors in the predicted spinal loads in tasks involving larger trunk flexion angles (e.g., less than 3% error for L5-S1 compression at 100° with or without hand load) (Figure 4).

DISCUSSION

This study primarily aimed to measure normal sagittal thoracic (T1-T12) spine ROM, whose magnitude has been somewhat controversial, in healthy population using, for the first time, inertial tracking sensors. Mean ROM of the thoracic spine in full forward flexion was found 20.5° (SD 6.5°) most of which was generated by the relative flexion of the T5 to T12 (lower thoracic). Contribution of the thoracic spine to generate the total forward trunk rotation remained almost constant (~16 to 20%) during forward flexion. **Body height and the measured ROMs had non-significant weak correlations.** Small uncertainties ($\pm 5^\circ$) in the measurement of total trunk flexion angle could result in considerable errors in the model prediction for spinal loads (hypothesis was confirmed).

Limitations: The *in vivo* study suffers from the problem common to all skin-mounted devices, i.e., the unavoidable inter skin-vertebra and sensor-skin movements. To minimize the skin-sensor movements the sensors were securely attached to skin overlying vertebrae by double-sided tapes and were continuously monitored during the tests. **As inertial sensors are relatively large compared to the spinous processes, their angular movement may have been partly recorded from the adjacent (upper or lower) vertebrae. The sensor on the T1, however, was carefully fixed below the C7 vertebra so to exclude movements from the cervical spine.** Coupled rotations in the coronal and transverse planes though, in accordance with the instruction to subjects to remain in the sagittal plane, were small ($<3^\circ$) but remained unavoidable during the tests. **The sensor at mid-thoracic spine was placed on the T5 spinous process rather than**

the T6 that lies in the middle of the twelve thoracic vertebrae. This was because in video-camera techniques upper trunk flexion angle is sometimes evaluated via a marker located at the T5 spinous process (Arjmand and Shirazi-Adl, 2005 and 2006; Bazrgari et al., 2008).

It was found that evaluation of the total trunk flexion angle using markers on the T5 (rather than T1) may underestimate trunk angle by up to 5.8° in full flexion.

The model predicted spinal loads based on an optimization algorithm that does not assign any force to antagonistic (abdominal) muscles. This shortcoming does not affect our conclusions as antagonistic co-activities do not vary considerably for a given trunk flexion and its two small levels of uncertainties ($\pm 5^\circ$). The model predictions for back muscle forces and intervertebral disc compression loads had been, however, in agreement with *in vivo* data for EMG activities of back muscles and intradiscal pressure values (Arjmand et al., 2009 and 2011).

Analysis of Results: The measured sagittal thoracic (T1-T12) ROM (20.5° , SD 6.5°) fell within the lower range of those reported in the literature using other measurement techniques (Table 2). In agreement with the only *in vitro* cadaveric (White and Panjabi, 1990) and an electromagnetic tracking (Willems et al., 1996) approach, but in disagreement with the CT medical imaging in supine posture (Morita et al., 2014) most of this ROM was generated by movements from the lower (T5-T12) thoracic spine. Medical imaging techniques measuring ROM in the supine posture have important limitations as maximal flexion cannot be reached by subjects (Morita et al., 2014) (only 31.7° of full flexion-extension ROM compared with $\sim 58\text{--}70^\circ$ measured by other techniques in standing posture as in Table 2). *In vitro* studies that remove ribs and muscles (White and Panjabi, 1990), however, may overestimate ROM of the thoracic spine as the stabilizing role of the sternum, rib cage, and muscles are eliminated (Watkins et al., 2005). For instance, while full flexion-extension ROM of the T11-T12 was measured to be $\sim 12^\circ$ in cadaveric specimens (White and Panjabi, 1990), an *in vivo* ultrasound-guided invasive method using Kirschner wires inserted into the T11 and T12 spinous processes (Gercek et al., 2008) and an *in vivo* radiography imaging in upright posture (Dvorák et al., 1991; Leferink et al., 2002) indicated a full flexion-extension ROM of only 2.7° and 5° , respectively. These latter values, compared with the foregoing *in vitro* investigation, agree better with our findings (i.e., total T5-T12 ROM of 14.8°) especially knowing that ROM of the thoracic motion segments decreases from the lower to upper levels. Differences between the measured ROMs in the foregoing *in*

vivo studies can also be partly due to different upright and full flexion postures. For instance, while in the current study subjects were asked to bend forward to their maximum voluntary flexion from their neutral upright posture, Troke et al. (1998) asked subjects to bend forward from a standardized start position to their full flexion so that they can look back through their legs.

Lumbar ROM (50.2°, SD 7.0°) also fell within the range of those reported in the literature using other skin-mounted devices and radiography imaging (Tafazzol et al., 2013). Total sagittal trunk rotation value of 118.4° (SD 13.9°) was in agreement with other reported data (Arjmand et al., 2009; Bazrgari et al., 2008; Esola et al., 1996; Hsu et al., 2008; Tafazzol et al., 2013). In agreement with the findings of our previous study using the inertial tracking device (Tafazzol et al., 2013) as well as those of others (e.g., Esola et al., 1996; Rose et al., 1988; Mayer et al., 1984; Paquet et al., 1994) we found a simultaneous rhythm between the lumbar and pelvis movements during forward trunk flexion (Figure 2). While the lumbar spine contributed more to the trunk rotation during early stages of forward flexion, the pelvis contributed more during the final stages of forward flexion (Figures 2 and 3). On the other hand, contribution of the thoracic spine (T1-T12) to generate the total forward trunk flexion remained almost constant (~16 to 20%) during the entire movement (Figure 3). The L/P ratio (calculated by dividing the total lumbar rotation to the total pelvis rotation at each trunk angle) reduced with the trunk flexion angle (Figure 2) as also reported by others (Esola et al., 1996; Granata and Sanford, 2000; McClure et al., 1997; Porter and Wilkinson, 1997; Rose et al., 1988). It has been suggested that evaluation of the lumbopelvic rhythm can help discriminate between healthy and low back pain individuals (Esola et al., 1996; McClure et al., 1997; Porter and Wilkinson, 1997). Significant difference in the thoracic sagittal ROM between patients with a shoulder outlet impingement syndrome and a control group with no shoulder pathology has been observed (Theisen et al., 2010). Assessment of the mobility of the thoracic spine can hence be important in the diagnosis and therapy of patients with shoulder outlet impingement syndrome. Future studies should evaluate pattern of the thoracic spine movement in upper back pain patients to investigate how this measure may help discriminate between individuals with and without upper back pain.

The modeling investigation indicated that an experimental error of only 5° in the measurement of trunk flexion angle in tasks involving small trunk flexion resulted in large errors in the predicted spinal loads (Figure 4). For instance, an error of 5° in trunk flexion of 20° (i.e., 15° or 25° rather than 20°) resulted in under- or overestimation of ~27% and 20% in the predicted L5-S1 compression without and with 20 kg in hands, respectively (22% and 17% for shear loads without and with 20 kg in hands, respectively). On the other hand, such errors for tasks involving larger trunk flexion angles (e.g., 80° or 100°) considerably reduced to < 3% for compression and < 8% for shear loads (Figure 4). This was because an error in the measurement of trunk flexion angle in tasks involving small trunk flexion directly resulted in a considerable change in the moment arm of gravity loading and thus external moments, muscle forces and spinal loads while it mainly affected the vertical position of center of mass of the upper trunk in tasks involving larger trunk angles. **Moreover, effect of larger errors in the measurement of trunk flexion angle on the model predictions can also be seen in Figure 4 (compare predictions for trunk flexion angle of 25° with those of 35° for an error of 10°, for instance).**

Our *in vivo* measurements also indicated that the relative sagittal rotation of T1 to T12 remained smaller than 10° for forward flexion of up to 50% of the full ROM (Figure 2). For the sake of simplification, biomechanical models of the thoracolumbar spine may therefore consider thorax to move rigidly during these activities. However, as small errors in the measurement of trunk flexion angle in these activities result in considerable errors in the model predictions (Figure 4), accurate measurement of the trunk flexion angle becomes crucial. On the other hand, although model predictions during tasks involving larger trunk flexion are less sensitive to the errors in the measurement of trunk flexion angles, but the adverse effect of assuming the thorax to move rigidly may increase due to neglecting the passive contribution of the ligamentous spine.

In summary, knowledge of the thoracic spine ROM during forward trunk flexion could have implications in patient discrimination and in biomechanical models. Inertial tracking devices allow for straightforward measurement of spinal ROMs. Thoracic sagittal ROM, mostly provided by movements from the lower (T5-T12) motion segments, was significantly smaller than that of the lumbar. **Body height and ROMs had weak correlations. The effect of assuming the thoracic spine as a single rigid body on model predictions will be the subject**

294 **of future studies.** While impractical for easy use in workplaces and clinical centers, medical
295 imaging techniques that allow measurements in the upright posture without exposure to
296 radiation, such as upright magnetic resonance imaging remain as the gold-standard approach for
297 static measurements of the spinal kinematics and ROMs **as its measurements are based on**
298 **images taken *in vivo* and from the vertebrae rather than the overlying skin.**

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Figure Captions

Figure 1: Position of the inertial sensors on the T1, T5, T12, and S1 spinous processes as well as a schematic of axis definitions for the derivation of the relative body segment rotations with respect to the earth-fixed global coordinate system.

Figure 2: The pelvis, lumbar (T12-S1), upper (T1-T5) thoracic, lower (T5-T12) thoracic, total thoracic (T1-T12) sagittal rotations (all on the right vertical axis) as well as lumbopelvic ratio (on the left vertical axis) at different forward trunk flexion angles (as percentage of full trunk flexion).

Figure 3: Contribution (as percentage) of the pelvis, lumbar spine (T12-S1), and thoracic spine (T1-T12) in generating total forward trunk flexion angle (T1) at different stages of forward flexion.

Figure 4: The L5-S1 compression (a) and posterior-anterior shear (b) loading at different trunk flexion angles (T1) with two levels of uncertainties ($\pm 5^\circ$) with or without 20 kg in hands.

Table 1: Body height, body weight, age, and body mass index (BMI) (mean (SD)) of volunteers participated in the study.

	Body height (cm)	Body weight (kg)	Age (year)	BMI (kg/m²)
All subjects	177.8 (7.3)	75.5 (10.8)	22.5 (1.8)	23.9 (3.1)
Range	163-194	54-99	20-26	18-28

Table 2: Comparison of the mean (SD) values (in degrees) of full thoracic (T1-T12) range of motion (ROM) (either from the upright standing posture to full flexion (upright-flexion) or from full extension to full flexion (flexion-extension)) measured in the current study with those reported in the literature. Equipment used, number of subjects, and their posture under which measurements were carried out are also indicated when available.

	Equipment	Number of subjects (posture/range)	T1-T12 ROM mean (SD)
Current Study	Inertial tracking device (Xsens)	40 (upright/upright-flexion)	20.5 (6.5)
Willems et al. (1996)	Electromagnetic tracking device	60 (seated/ upright-flexion)	32.0 (-)
Tully and Stillman (1997)	Skin marker and video-camera	22 (upright/ upright-flexion)	17.8 (8.6)
Mannion et al. (2004)	Electronic inclinometer (The Spinal Mouse)	20 (upright/upright-flexion)	25.0 (-)
Hsu et al. (2008)	Electromagnetic tracking device	18 (upright/upright-flexion)	26.0 (9.0)
O'Gorman and Jull (1987)	Inclinometer	120 (upright/ flexion-extension)	62.0 (-)
White and Panjabi (1990)	Cadaveric specimens	5 (-/flexion-extension)	64.0 (-)
Willems et al. (1996)	Electromagnetic tracking device	60 (seated/ flexion-extension)	58.0 (-)
Troke et al. (1998)	Electronic goniometer (CA6000)	11 (upright/flexion-extension)	70.0 (16.2)
Morita et al. (2014)	Computed tomography (CT) Imaging	50 (supine/flexion-extension)	31.7 (11.3)

Figure 1

Figure 2

Figure 3

Figure 4