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Xiaopeng Ning
Iowa State University, xpning@iastate.edu

Sangeun Jin
Iowa State University, greenday2324@gmail.com

Gary A. Mirka
Iowa State University, mirka@iastate.edu

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Xiaopeng Ning, Sangeun Jin and Gary A. Mirka*

The Ergonomics Laboratory

Department of Industrial and Manufacturing Systems Engineering

Iowa State University

Ames, IA 50011-2164, USA

Author: Xiaopeng Ning
Mailing address: Industrial and Management Systems Engineering P.O. Box 6070 West Virginia University, Morgantown, WV, 26505, USA
Phone: 304-293-9474
Fax: 304-293-4970
e-mail: xiaopeng.ning@mail.wvu.edu

Author: Sangeun Jin
Mailing address: The Ergonomics Lab, Dept IMSE, Iowa State University, Ames, IA, 50011, USA
Phone: 515-294-1682
Fax: 515-294-3524
e-mail: sjin@iastate.edu

* Corresponding author: Gary A. Mirka
Mailing address: The Ergonomics Lab, Dept IMSE, Iowa State University, Ames, IA, 50011, USA
Phone: 515-294-8661
Fax: 515-294-9273
e-mail: mirka@iastate.edu

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Abstract

Background: Electromyography-assisted (EMG-assisted) biomechanical models are used to characterize the muscle and joint reaction forces in the lumbar region. However, during a full-range trunk flexion, there is a transition of extension moment from the trunk extensor muscles to the passive tissues of the low back, indicating that the empirical EMG data used to drive these EMG-assisted models becomes less correlated with the extensor moment. The objectives of this study were to establish the trunk flexion angles at which the passive tissues generate substantial trunk extension moment and to document how these angles change with asymmetry.

Methods: Participants performed controlled trunk flexion-extension motions in three asymmetric postures. The trunk kinematics data and the electromyographic activity from L3- and L4-level paraspinals and rectus abdominis were captured. The time-dependent net internal active moment (from an EMG-assisted model) and the net external moment were calculated. The trunk and lumbar angles at which the net internal active moment was less than 70% of the external moment were found.

Findings: The trunk flexion angle at which the net internal moment reaches the stated criteria varied as a function of asymmetry of trunk flexion motion with the sagittally symmetric case providing the deepest flexion angle of 38 degrees (asymmetry 15 degrees: 33 degrees; asymmetry 30 degrees: 26 degrees).

Interpretation: These results indicate that EMG-assisted biomechanical models need to consider the role of passive tissues at trunk flexion angles significantly less than previously thought and these flexion angles vary as a function of the asymmetry and direction of motion.

Key words: Electromyography-assisted model, Passive moment, Asymmetric bending

1. Introduction

Electromyography (EMG) -assisted biomechanical models of the lumbar region have been developed and refined over the last three decades (e.g. McGill and Norman, 1986; Cholewicki et al., 1995; Granata and Marras, 1993, 1995; Nussbaum and Chaffin, 1998). As with earlier models, this technique recognizes that the external moments encountered during lifting tasks must be countered by a net internal moment generated by the active (muscles) and passive (intervertebral discs, ligaments, and fascia) tissues of the torso. The EMG-assisted approach to modeling the muscle forces and joint reaction forces of the lumbar region marked a significant improvement over earlier models in that empirical EMG data of muscles from both the low back and abdominal regions were considered, thereby allowing the models to estimate both agonist and antagonist muscle forces. In doing so, these models were better able to estimate the internal stresses under more realistic lifting conditions (dynamic lifting, asymmetric lifting postures, and high force exertions) where antagonist muscle forces are commonly observed.

One limitation of the EMG-assisted modeling technique is that the forces generated by the passive tissues of the region are not able to be quantified empirically. For many postures through the normal range of trunk motion this is not a major concern, but in extreme flexion postures (such as lifting from ground level or performing stooped labor tasks) these passive tissues bear much of the extensor moment necessary to hold the posture (e.g. Dolan et al., 1994). If one relies solely on the moment estimates of the EMG-derived muscle forces to provide the necessary extension moment equilibrium between internal and external moments, muscle forces can be grossly under-predicted and spine reaction forces are inaccurate. Therefore, it is important to establish a trunk flexion “boundary” within which the EMG-assisted modeling technique is valid. Previous studies have considered these passive tissue contributions through anatomical modeling (e.g. McGill and Norman, 1986) and finite element modeling (e.g. Arjmand and Shirazi-Adl, 2006; Gagnon et al., 2011). Others have roughly estimated the boundary of the EMG-assisted model to be approximately from upright posture to 45^{o} of trunk flexion (e.g. Marras and Granata, 1997) but empirical validation of this range has not yet been achieved. Further, previous literature has demonstrated that trunk axial rotation (i.e. twisting) can influence the interplay between the passive and active tissues of the lumbar spine resulting in changes in the magnitude of the full flexion lumbar angle and peak trunk inclination angle (Ning et al., 2011). This complex relationship between trunk flexion angle, trunk axial rotation, and this validity boundary is best established through empirical methods. The objectives of the current paper were 1) to establish an empirical active region boundary beyond which the passive tissue moment begins to play a significant role in generating the necessary trunk extension moment and the validity of EMG-assisted model will no longer hold and 2) understand the effects of trunk asymmetry on this response. It is hypothesized that the lumbar flexion angles at which passive tissues contribute significantly to the net extensor moment will decrease in asymmetric postures.

2. Methods

2.1 Participants

Eleven male student volunteers from Iowa State University with average age 26.1 (SD 2.8) years, height 178.6 (SD 5.1) cm and total body mass 71.1 (SD 5.4) kg participated in this study. Subjects with a history of chronic low back pain, chronic lower extremity injury or currently experiencing discomfort in these regions were excluded from participation. All participants provided written informed consent prior to participation and all procedures were reviewed and approved by the Institutional Review Board for the Use of Human Subjects in Research.

2.2 Instrumentation

EMG signals were collected bilaterally from rectus abdominis, L3 and L4 vertebrae levels of the paraspinals at a sampling rate of 1024Hz using surface electromyography (Model: Bagnoli, Delsys Inc, Boston, MA, USA) employing 6 bipolar surface electrodes. The sampling locations for these six muscles were as follows: bilateral rectus abdominis - 5 cm above the umbilicus and 3 cm lateral to midline; bilateral L3 paraspinals – 4 cm from the L3 spinous process; bilateral L4 paraspinals - 3 cm from L4 spinous process. A magnetic field-based motion tracking system (Model: Motion Star, Ascension Technology Corporation, Burlington, VT, USA) with three sensors on the skin over the C7, T12 and S1 vertebrae were used to collect three-dimensional trunk movement at a sampling rate of 102.4 Hz. All the EMG and trunk kinematics data were synchronized and saved by using the "MotionMonitor" data acquisition software (MotionMonitor, Innovative Sports Training, Chicago, IL, USA).

An asymmetric reference frame and a lumbar dynamometer (Model: Kin/Com, Chattanooga Group, Inc., Hixson, TN, USA) were used to provide a stable system for securing the pelvis during all trials and to provide static resistance and control of trunk flexion angle during the trunk muscle maximum voluntary contraction (MVC) exertions (Mirka and Marras, 1993). During the experimental trials, the subjects utilized a series of vertical cues (vertically standing 2x4 boards) to maintain the required asymmetric postures through the full range of trunk flexion - a method established and employed in a previous paper (Ning et al., 2011).

2.3 Experimental Design

The independent variable in this study was trunk asymmetric angle (ASYM) and was considered at three levels: 0° (sagittally symmetric), 15° and 30° (both leftward twisting in the transverse plane from the mid-sagittal plane). This asymmetric angle was defined as the angle between the mid-sagittal plane and the plane containing a vertical line running through the midpoint between the ankles and the midpoint between the hands in the full flexion posture (Waters et al., 1993).

Four dependent variables were considered in this study. These four dependent variables involved two angles: lumbar flexion angle and trunk inclination angle. Lumbar flexion angle focuses on the sagittal plane angle (pitch) of the sensor on the T12 vertebra relative to the same angle of the S1 sensor. Trunk inclination angle is a more global measure of the angle between vertical and the line connecting the midpoint of the S1 sensor and the C7 sensor (Figure 1) (Ning et al., 2011). These two measures were captured at two important points in the flexion-extension motion profile. First, these two angles were found at the point in the eccentric (i.e. trunk flexion) motion where the internal moments (from EMG-assisted model) and the external moments (from simple free body diagram model) diverge (see Figure 2). Divergence of these two moments is defined as the point where the internal calculated moment is less than 70% of the external moment. This process is defined in greater detail in the *Data Processing* section. Next, these two angles were found at the point in the concentric (i.e. trunk extension) motion where the internal and external moments converge (Figure 2). It must be emphasized that the lumbar angles presented in this study are absolute not relative, that is, they were not normalized to the lumbar angles observed in the upright standing posture. The average value of lumbar angle in the upright relaxed posture in the current study was -21 degrees – reflecting the natural lordosis of the spine. Adding 21 degrees to the lumbar flexion values shown in these results provides an estimate of the <u>relative deviation</u> in lumbar angle for the lumbar angle variables.

Insert Figures 1 and 2 About Here

2.4 Protocol

Upon arrival the experimental procedures were described to the participant and written informed consent was obtained. Basic anthropometric data were gathered and a five minute, lower extremity/torso warm-up routine was followed. Surface electrodes and motion sensors were then secured in the appropriate locations.

The experiment started with two repetitions of maximum voluntary contractions (MVCs) where subjects were required to stand in the asymmetric reference frame apparatus with pelvis secured and perform isometric MVC trunk flexion and extension exertions in a 20 degree forward trunk inclination posture. This posture was selected so that low back muscles would contract near their resting length. Upon completion of two repetitions of each flexion and extension MVC exertion, a five-minute resting period was provided. The participants then performed a randomized sequence of the 15 trials (five repetitions for each of the three ASYM levels). In each trial, participants were asked to perform a slow, controlled trunk flexion-extension exertion from the upright standing posture down to a full, relaxed trunk flexion posture in approximately 20 seconds (a metronome provided an auditory cue) while maintaining the designated trunk asymmetry angle with the help of the vertical cue (Ning et al., 2011). In asymmetric trials, participants began each trial by axially rotating their trunk to the designated asymmetric angle (15^{o} or 30^{o}) and then perform the flexion-extension motion. They were also required to come back to the same trunk rotation posture at the end of the trunk extension motion then rotate back to the sagittally symmetric posture. Prior to data collection, a training session was provided to all participants in order to ensure task performance consistency. During the data collection the experimenter provided verbal feedback on technique.

2.5 EMG-assisted model

In the current study, a simplified EMG-assisted biomechanical model was developed to calculate the muscle-generated internal moment about the L5/S1 joint. Normalized (to maximum) EMG were input into the model to calculate the muscle forces. In this model the moment arms of the six muscles and physiological cross-sectional areas were estimated using established regression equations from previous studies (Jorgensen et al 2001; Marras et al., 2001) and were adjusted based on the asymmetry of the direction of motion. The appropriate length-tension and force-velocity (including appropriate modifications for eccentric exertions) modulation factors for the EMG-assisted modeling techniques were taken from the literature (Davis et al., 1998; Marras and Granata, 1997). The maximum muscle stress value (i.e. gain value of EMG-assisted model) was obtained by matching the internal and external moment profile in the latter half of the extension phase of the trunk motion. A best-fit gain was determined and applied across all muscles and all trials for each subject.

2.6 Data Processing

EMG data were filtered using a low-pass filter of 500 Hz, a high-pass filter of 10 Hz and the signal was notch filtered at 60 Hz (ambient electrical noise) and 102.4 Hz (motion tracking system) and their aliases up to 500 Hz. EMG signals were then rectified and smoothed with dual fourth-order Butterworth low-pass filter with cutoff frequency of 2Hz. The EMG data during the flexion-extension motions were normalized to MVC EMG for each muscle. These normalized EMG were then utilized in the EMG-assisted model described above to calculate the time-dependent, net internal moment. The time-dependent external moment was calculated using a simple physical model that considered the time-dependent trunk angle and upper body anthropometric variables. The time-dependent internal and external moments were then compared. The first point in time during the eccentric (flexion) phase of the trunk motion when the internal moment was less than 70% of external moment was determined as the "divergence point" and the lumbar flexion divergence angle and trunk inclination divergence angle were established at that point in time (Figure 2). Similarly, the "convergence point" was found by starting at the end of the data set and moving backwards through the data until the first time that the internal moment was less than 70% of the external moment and the lumbar flexion convergence angle and trunk inclination convergence angle were the corresponding angles. This 70% criterion value was found during pilot studies to provide a robust and consistent point of variance. Higher criteria values (e.g. 80 or 90% of external moment) proved to be too sensitive to the variable internal moment estimations, creating "false positives" for divergence.

2.7 Statistical Analysis

Prior to any statistical analysis, the assumptions of the ANOVA procedures (homogeneity of variances of the residuals, residuals normally distributed, and independence of observations) were evaluated using the graphical techniques advocated by Montgomery (2005). Once these assumptions were verified, MANOVA was performed to evaluate the effects of ASYM on the four dependent variables collectively, thereby controlling for the experiment-wise error rate. Univariate ANOVA procedures were then conducted and a Tukey-Kramer post-hoc test was performed to further explore the significant effects. The criteria p-value of $p<0.05$ was used for all statistical tests.

3. Results

Results from the MANOVA revealed a significant ($p<0.001$) effect of ASYM and subsequent univariate ANOVA results demonstrated significant asymmetry effects on several of the angles considered (Figures 3 and 4). The results showed a significant ($p<0.001$) decrease in the value of trunk inclination divergence angle with increasing levels of asymmetry of the lifting posture (from 37.7° (ASYM=0) to 26.3° (ASYM=30) (Figure 3) but showed that ASYM did not have a significant effect on the value of lumbar flexion divergence angle. Collectively, these results indicate that, during the flexion motion, passive tissues begin to contribute significantly at trunk angles less than 40 degrees, considerably less than what we have previously thought.

Insert Figures 3 and 4 About Here

When considering the convergence angles during the extension motion, our results indicate that both the lumbar flexion and trunk inclination angle variables are significantly ($p<0.01$ and $p<0.001$, respectively) affected by asymmetry of the trunk posture (Figure 4). However, recognizing from some of our earlier work that the full flexion angles have been shown to decrease with increasing asymmetry (Ning et al., 2011) an additional analysis was performed. In this analysis these convergence angles were subtracted from the peak angles (both for the lumbar flexion and trunk inclination angles) to establish how much extension from the full flexion posture was necessary to increase the active muscle contribution to the 70% level. When this subtraction was done the significant effect of ASYM was lost (Figure 5). This indicates that, across asymmetry levels, there is a relatively constant number of degrees of extension from the full flexion posture necessary to achieve the 70% criteria level.

Insert Figure 5 About Here

4. Discussion

The three-dimensional “active” boundary for EMG-assisted models of the lumbar spine developed in the current study identifies the range of trunk flexion motion beyond which passive tissue forces contribute significantly to the net extensor moment of the lumbar region. Previous studies have estimated this boundary to be about 45 degrees of trunk flexion for trunk motions in the mid-sagittal plane. Results of this study indicate that a more conservative estimate of this boundary is around 38 degrees in the sagittally symmetric motions. This study also considered asymmetric postures and showed that, in these asymmetric postures, the trunk inclination angle value decreases even further (33 degrees at 15 degrees asymmetry and 26 degrees at 30 degrees asymmetry). These results indicate that the pre-stressing of the passive tissue that occurs during the axial rotation of the torso, has a significant effect on the trunk inclination angles at which significant passive extensor moment is generated. Comparing these results with the peak angles shown in Ning et al. (2011), these angles represent 41%, 36%, and 29% of the full range of trunk flexion motion.

The results of the current study generally support previous studies that have used anatomical and finite element models to understand the passive tissue contribution to the net extensor moment during sagittally symmetric trunk flexion. In an early version of their EMG-assisted biomechanical model of the lumbar region, McGill and Norman (1986) found that the active muscle moment accounted for 99% of the extensor moment and noted that the participants in their study employed a “flatbacked” lifting posture with most postures generating less that 34 degrees of motion. Their results are consistent with the results of the current study as our results indicate that the rapid transition from active to passive tissue during the eccentric motion reaches our 70% criteria level at about 4-5 degrees later in the flexion range of motion than their 34 degrees of flexion. The current results also support the predictions of Arjmand and Shirazi-Adl (2006) in their finite element modeling of the lumbar spine and the empirical validation of their model. These researchers found that as trunk flexion angle increased from 40 degrees to 65 degrees (thereby increasing the external moment about the lumbar spine) the EMG activity of the longissimus and multifidus did not increase. These results indicates that the passive tissues are increasing their contribution to the net extensor moment in this region to compensate for the reduced active moment – a result that is consistent with the empirical results of the sagittally symmetric postures evaluated in the current study.

In addition to the sagittally symmetric postures, empirical evaluation of the active-passive interplay in asymmetric postures was also explored in the current study. As the complexity of these passive tissue geometries increases during asymmetric lifting postures, small errors in the modeling of the morphology of these tissues can lead to important alterations in the relationship between the flexion angle and the strain experienced in these passive tissues. The empirical approach to identifying these active boundary points overcomes these inaccuracies by simply describing the net effect of these passive tissues. The results in the asymmetric postures partially supported our hypothesis. These results are consistent with expectations because of the pre-stressing of the passive tissues that occurs during the axial rotation of the torso to achieve these asymmetric postures. It is interesting to note, however, that while the trunk inclination angle at which these moments estimates diverge is affected by asymmetric posture, the lumbar flexion angle at which they diverge does not change (8 degrees, 6 degrees, 6 degrees at 0, 15, and 30 degrees of asymmetry, respectively – differences not statistically significant). This indicates that much of the change in the pre-stressing of the passive tissues due to the asymmetric postures occurs above the lumbar level. Our previous study (Ning et al., 2011) reported that for contralateral lumbar paraspinal muscles, both trunk inclination angle and lumbar flexion angle at the point of flexion-relaxation were significantly decreased with increases in trunk asymmetry. In contrast, results from current study showed that only trunk inclination angle was significantly decreased with the increases in trunk asymmetry – not lumbar flexion angle. This contrast between the FRP angles and the boundary angles of the current study would indicate important changes in internal spinal kinematics as the torso continues to flex to this full flexion posture.

One caveat that must be noted in the interpretation of the trunk inclination angles is that the experimental apparatus employed in this study did restrict pelvic rotation in the sagittal plane. In true free-dynamic lifting tasks this pelvic rotation will allow for a much greater range of sagittal plane trunk motion (i.e. trunk inclination angle in the current study) before the passive tissues are placed in significant tension. On the other hand, the lumbar flexion angles as defined in the current study account for sacral motion and therefore can be interpreted similarly in both the current experimental apparatus and in the free-dynamic lifting scenario. While the results of the current study are not able to provide an estimate of the passive force developed by an individual ligament or disc, these results can provide a standard for evaluating the predictions of anatomical and finite element models.

A comparison of the divergence angles (eccentric motion) and convergence angles (concentric motion) provides some interesting insights into the role of the passive tissues. During the sagittally symmetric flexion motion (eccentric contraction of the low back musculature) the passive tissues lumbar flexion divergence angle was 8 degrees (29 degrees when the initial 21 degrees of lordosis are added) while the lumbar flexion convergence angle in the sagittally symmetric posture was 22 degrees (43 degrees when the upright 21 degrees of lordosis are added). This significant differential indicates an important distinction in the function of the passive tissues that should be considered as these EMG-assisted models seek to describe spinal loading during both flexion and extension motions. At the full flexion posture (just before the beginning of the trunk extension motion) the full weight of the torso is supported by the passive forces (the flexion-relaxation response) and creep deformation occurs in the viscoelastic tissues (Solomonow et al., 2003; Shin et al., 2010). As the individual commences the extension motion the passive tissues go through a relatively rapid shortening/de-stressing resulting in a need for significant increases in the active muscle component early in the extension motion. At about 8 degrees (Figure 5) from the full flexion posture the active moment exceeds the 70% criteria. In contrast, as the individual flexes forward from the upright relaxed posture, the 70% criteria level is met at a lumbar flexion angle corresponding with a value of 24 degrees from the full flexion posture. Similar back extensor muscle activation pattern has also been observed from previous studies (Solomonow et al., 2003; Olson et al., 2006). The difference between 8 degrees from full flexion in trunk extension and 24 degrees from full flexion in trunk flexion is likely the result in differences in the force producing capabilities of these viscoelastic, passive tissues resulting from different levels of strain experienced immediately prior to the movement. This perspective is supported by the work of Ambrosetti-Giudici et al. (2009) which showed that significant stress-relaxation occurs during the first few seconds of strain on the tissue, particularly at higher strain levels. In the former case, the passive tissues had been strained while in the full flexion posture and therefore have lost their force producing capability at a given level of lumbar flexion, while in the latter case this pre-strain is not present and therefore the passive tissue force is generated with a lower level of required lumbar flexion. Regardless of the underlying cause, this difference should be recognized within the EMG-assisted models to ensure that the passive tissue forces are correctly considered.

It is recognized that the choice of threshold level has a profound effect on the resulting boundary angles. The 70% value used in the current study was established through pilot studies and supported in post-hoc analysis of the full data set. In preliminary pilot work the research team reviewed the traces of the internal and external moments and then evaluated these plots and the predictions of the angles where these two plots diverged as a function of different threshold values. The ideal situation would be when the external moment value abruptly and permanently departs from the internal moment, and we would like to have that threshold value as close to 100% as possible (generating a very sensitive model). From these preliminary analyses we established that the 70% criteria provided stable (from trial to trial) estimates of the trunk inclination and lumbar flexion angles. Once the full dataset was collected we applied 70%, 80%, and 90% criteria values and found that the 70% value was indeed the most consistent across multiple repetitions. The 90% and 80% values generated a high number of false positives simply because of the variability of the predicted internal moment. The internal moment was not quite as smooth as the external moment and this variability, inevitably, resulted in many instances wherein the internal moment was less than 90% (or 80%) of the external moment early within the flexion movement, but would recover, a few degrees later, to regain a match of the external moment. This resulted in the 90% and 80% criteria having higher variability (standard deviation 6.6 and 6.5 degrees, respectively) while the value for 70% was significantly reduced to 4.7 degrees.

There are several limitations to the generalizability of the results of this study that must be noted. First, as noted previously, these exertions were performed in an exercise apparatus that restricted the motion of the pelvis. Recognizing the important interplay between the low back and the lower extremities (lumbopelvic and pelvifemoral rhythms) may, for some, limit the interpretation of the current results to only the lumbar flexion angles, seeing the trunk inclination angles as fundamentally influenced by the restrictions of the pelvic motion. The utility of the trunk inclination angle results may depend on the type of apparatus being employed in a given study (free-dynamic lifting vs. controlled lifting). Second, as with all EMG-assisted models, the empirical data collected in this experiment represents the activity of only a subset of the muscle fascicles that may play a role in the generating moments about the spine. Those chosen for capture in this experiment represent the majority of the muscle cross-sectional area, however not all muscles/fascicles that could play a role were considered in the EMG-assisted model used in this study. In addition, previous studies have reported a uniform force-length-velocity modulator (Vance et al., 1994; Baratta et al., 2000) however a two step length-tension modulator and force-velocity modulator was used in the current study. The uniform force-length-velocity modulator may improve the force prediction accuracy however only exiguous differences should exist when finding the active region boundary because the shift of external moment between active and passive tissue at boundary line is rather clear and obvious. These simplifications and our choice of the 70% threshold were appropriate for the current purposes, but other choices could impact the boundaries predicted by the model.

5. Conclusions

Results demonstrated that passive tissues generate significant loads at trunk angles less than previously thought and the effect of asymmetric lifting postures was to further reduce the trunk flexion angles necessary for their activation. The results also note an important difference in the trunk angles necessary to initiate these passive tissue forces when comparing the concentric and eccentric motions of the trunk, indicating a need for considering direction of motion when performing EMG-assisted modeling.

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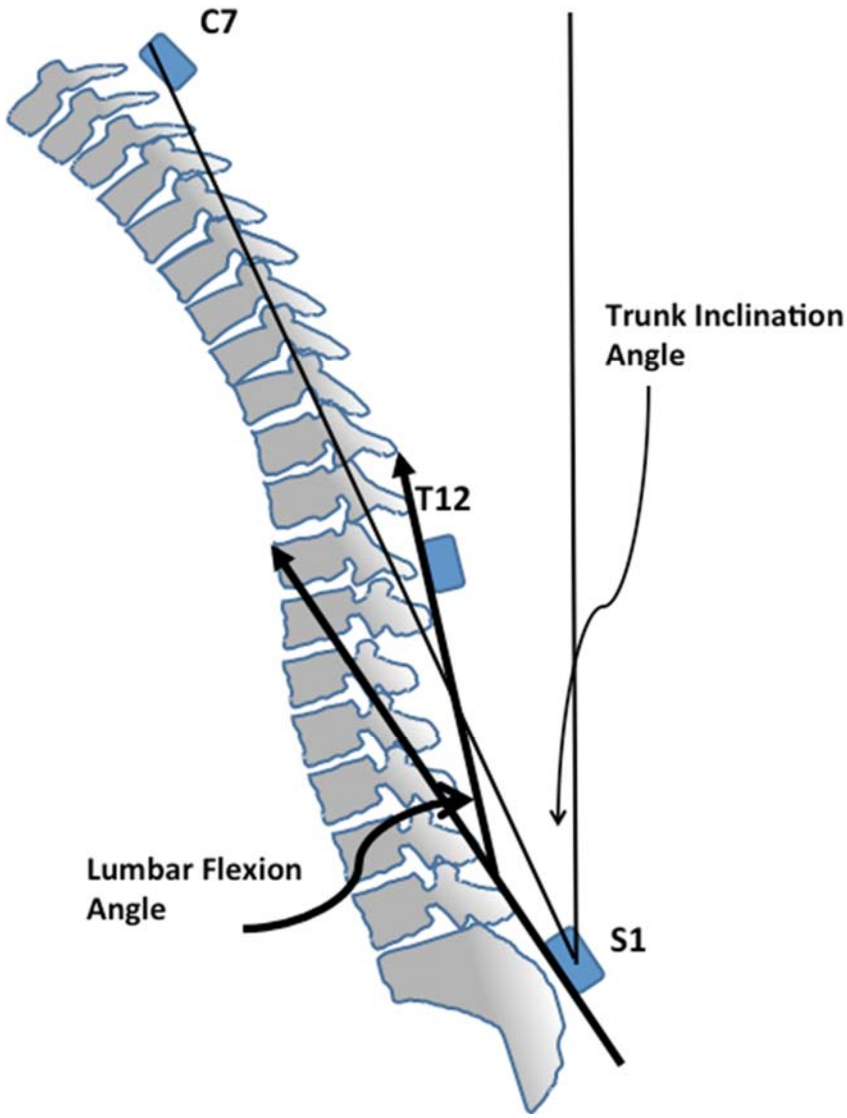

Figure 1: Definition of lumbar flexion and trunk inclination angles (adapted from Ning et al., 2011). (In the posture shown in this figure, lumbar flexion angle would be negative due to the natural lordosis of the lumbar spine.)

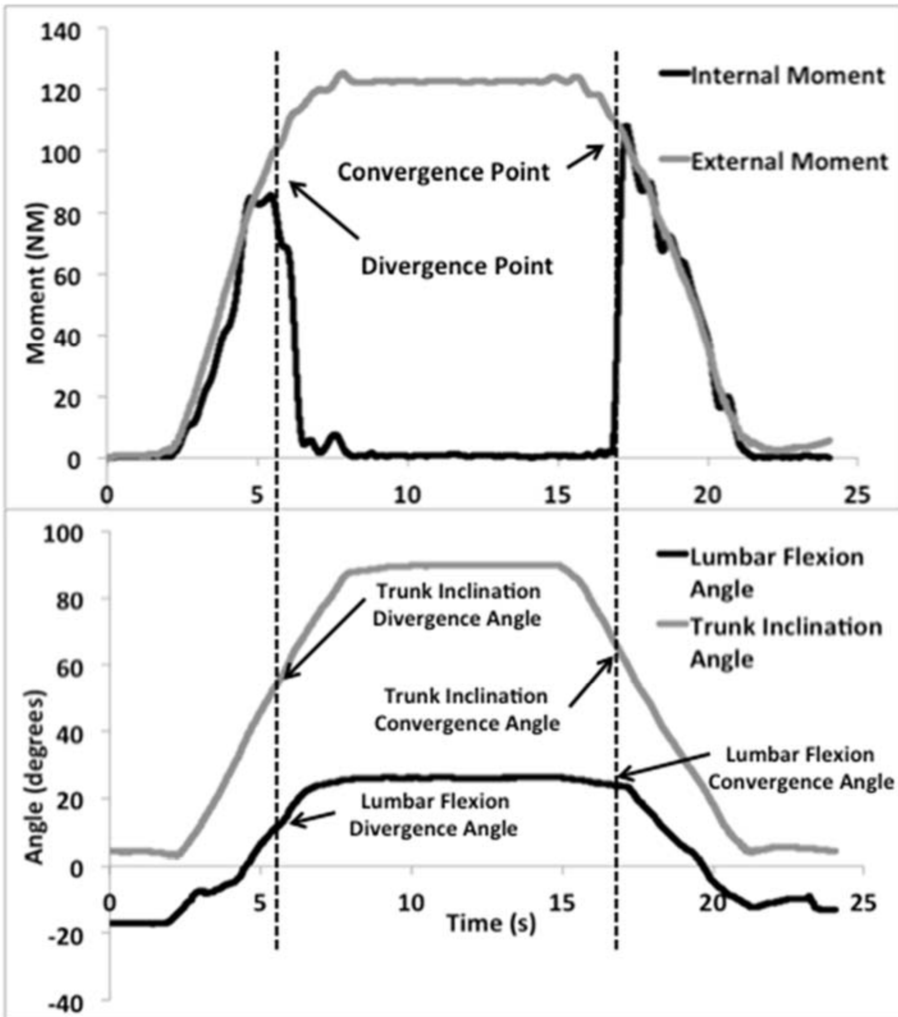

Figure 2: Graphic showing the process of establishing divergence and convergences angles. Top panel shows the process of establishing the 70% threshold, while the bottom panel identifies the corresponding lumbar and trunk angles.

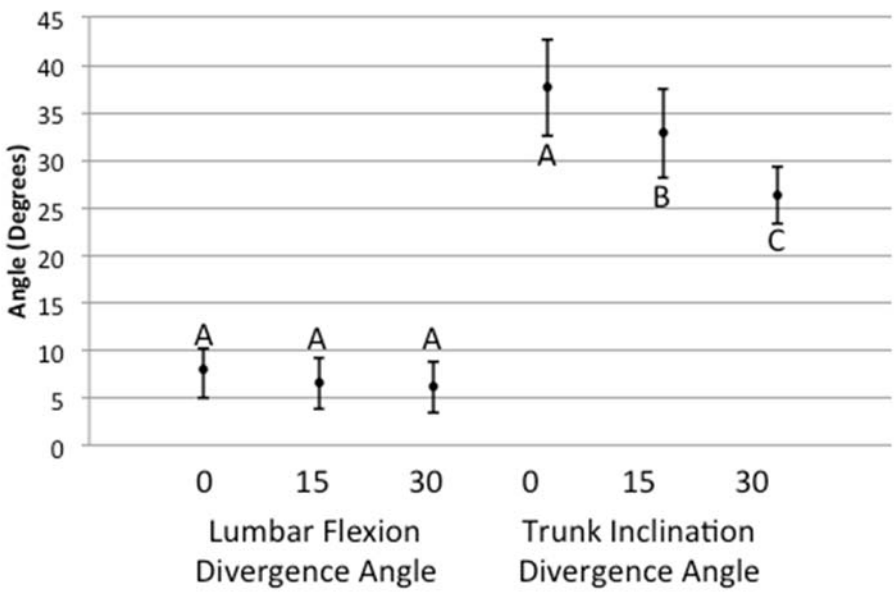

Figure 3: Effect of asymmetry on the lumbar flexion divergence angle and the trunk inclination divergence angle. Values with the same letter are not statistically significantly different. 95% confidence intervals are shown.

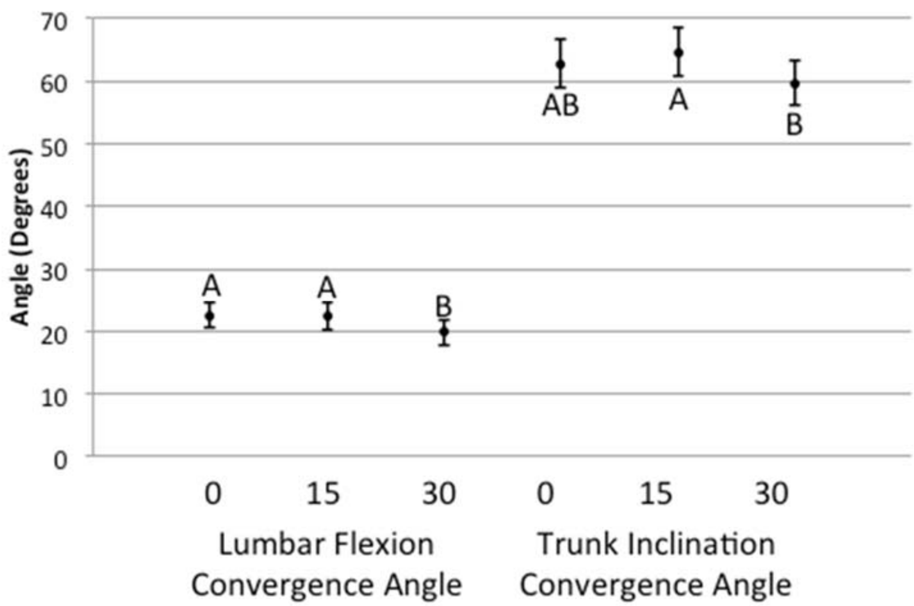

Figure 4: Effect of asymmetry on the lumbar flexion convergence angle and the trunk inclination convergence angle. Values with the same letter are not statistically significantly different. 95% confidence intervals are shown.

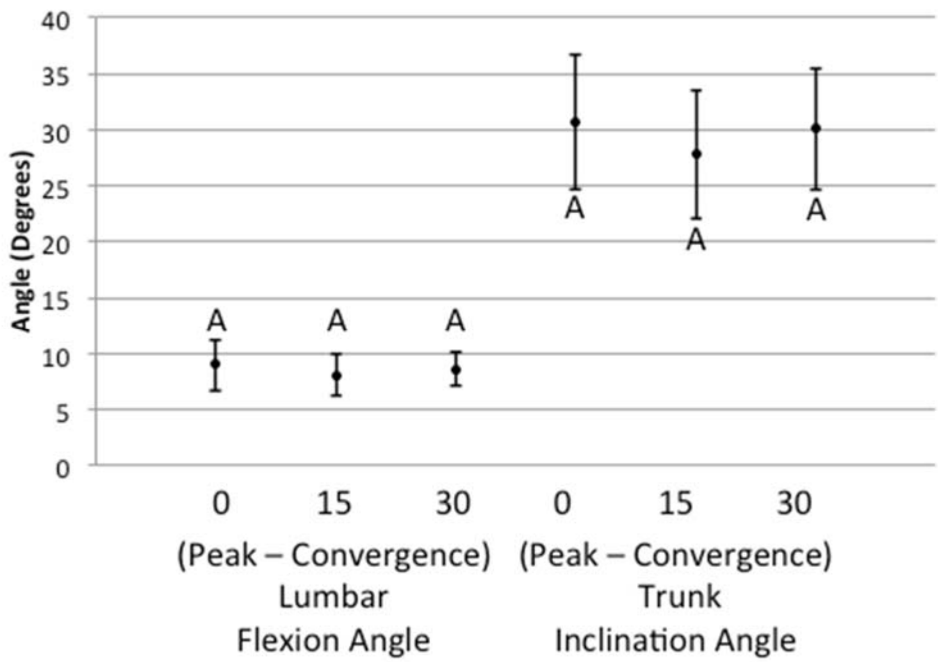

Figure 5: Effect of asymmetry on the "Peak - Convergence" lumbar flexion and trunk inclination angles. These values represent the number of extension degrees from the full flexion posture necessary to reach the convergence of the internal and external moments. Values with the same letter are not statistically significantly different. 95% confidence intervals are shown.