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Asymmetric loading of erector spinae muscles during sagittally symmetric lifting †

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Abstract

Functional asymmetry is among the multitude of risk factors for low-back pain (LBP), the most common injury under general industrial and agricultural conditions. However, previous studies showed that normal healthy individuals exhibit some functional asymmetry, indicating that not all asymmetry causes LBP. Therefore, the threshold value that is able to discriminate between normal and pathological situations is used as critical information to predict LBP. As a preliminary study to find threshold, the purpose of this study is to quantify the magnitude of bilateral asymmetries of erector spinae muscle forces of a healthy group during sagittally symmetric lifting. Ten healthy male subjects with no history of back pathology participated in this study, which collected motion capture, force data, and electromyography signals from six infrared cameras (MCam2, Vicon), two force platforms (AMTI), and surface EMG (BME Korea). In order to quantify the magnitude of bilateral asymmetry in the trunk muscle forces, we used 3D linked segment and EMG-assisted modeling approaches, both of which were verified based on their recapitulation of previously-proposed models. The results indicated that each muscle force in the lumbar region exhibited asymmetry during the entire lifting process. In particular, the erector spinae muscle forces exhibited an approximate 24% difference between bilateral sites ($p<0.05$). The results of this study provided data from normal individuals by which to identify pathological situations and predict LBP incidence within general industrial and agricultural conditions.

Keywords: Asymmetry; EMG; LBP (Low-Back Pain); Lifting

1. Introduction

Low-back pain (LBP) is the most common injury worldwide, occurring with a lifetime prevalence of 50-90% [1, 2]. In the United Kingdom, economic losses due to LBP increased from about $27 millon in 1992 to close to $125 million in 1995 [3], comprising about 27% of overall disability payments [4]. Previous studies showed that LBP is frequently the result of lifting tasks in the workplace [5, 6], leading to proposed lifting posture guidelines designed to prevent LBP [7]. However, these recommendations have not appreciably decreased the incidence of lifting-related LBP, suggesting that LBP is not merely a result of incorrect posture, but that it is affected by a multitude of risk factors, including load, repetitions, and the patient's mental state, and others. Among these factors, bilateral asymmetry is thought to significantly increase the risk of LBP [8-11].

Asymmetry refers to an individual's random fluc-

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tuations from perfect bilateral symmetry, usually resulting from environmental factors [12]. Asymmetry leads to coordination differences that stem from one side of the body handling a greater proportion of the load [16]. For example, knee angles become asymmetrical during walking, although it is generally regarded as a symmetrical movement [13]; in addition, "symmetric lifting" contains inherent asymmetric elements, as revealed by ground reaction force, EMG, and motion analyses [5]. In sports applications, asymmetry in an athlete increased the risk of injury [10, 11] and reduced performance [14].

Despite these findings, not all asymmetries are linked to increased injuries and reduced performance. The risk tends to increase in cases where the magnitude or frequency of the asymmetries reaches a certain threshold level. Therefore, it is possible that if this threshold value were defined, it could be used to discriminate between normal and pathological situations, providing critical information for the prediction of LBP. For example, Subotnik [15] proposed such a threshold value based on bilateral differences in leg length: If the discrepancy in leg length is greater than 6.4 mm in an athlete or 19.1 mm in a non-athlete, it is pathologically important. In addition, Knapik *et al.* [8] proposed that in the knee flexors and hip extensors, strength imbalances greater than 15% were predictive of injury in female collegiate athletes.

A more recent study quantified bilateral asymmetry during symmetric lifting in a healthy group of individuals [16]. Ground reaction force studies during lifting demonstrated bilateral differences of up to 10%, and the researchers suggested that this information could be used as a standard predictor of LBP. However, this study was somewhat limited, in that the effects of asymmetries on the back determined via quantifications based on the ground reaction force do not directly translate to the actual load on the back.

Therefore, the objective of this study was to evaluate differences in asymmetry in normal subjects using erector spinae muscle forces, and to quantify the magnitude of asymmetries during sagittally symmetric lifting. The qusntified asymmetry measurements from this study will later prove useful in defining a threshold for LBP prediction.

2. Method

2.1 Subjects & Apparatus

Ten healthy male subjects, none of whom evidenced any musculoskeletal disorders, participated in this study. The experimental instruments included a 6-MCam2 camera, 2-force platforms (AMTI), and a 12-channel surface EMG (BME Korea). Data was collected at a frequency of 120Hz for the cameras and force platforms and 1080Hz for the EMG, and each apparatus was synchronized with a Vicon 460 system. The optical sensor trajectories and ground reaction data acquired during experiments were filtered with a fourth-order Butterworth, zero-lag, low-pass filter with a 7Hz cut-off frequency. This cut-off frequency was determined by residual analysis, as previously described [17]. The optical sensors and electrodes were attached to whole body anatomical landmarks and the trunk (rectus abdominis, external oblique, internal oblique, lumbar erector spinae, latissimus dorsi, and thoracic erector spinae), respectively, in order to measure the erector spinae muscle forces. Fig. 1 shows the locations of the attached optical sensors and EMG electrodes.

Table 1. Subject Character, mean values (±S.D.).

Gender	Age (years)	Height (cm)	Weight (kg)
male	25.3 (1.6)	174.3 (5.0)	67.0 (1.7)

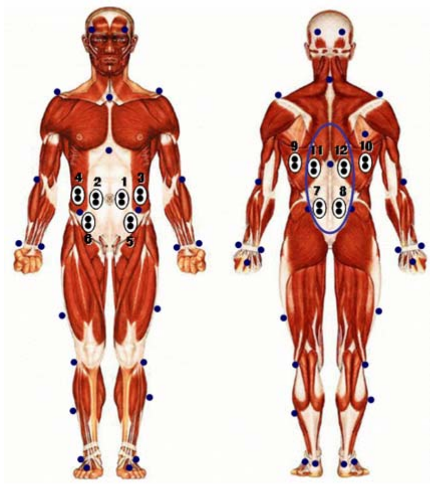


Fig. 1. Locations of the attached optical sensors and EMG electrodes. (The reference of the anatomy of the whole body, 1-2: rectus abdominis, 3-4: external oblique, 5-6: internal oblique, 7-8: lumbar erector spinae, 9-10: latissimus dorsi, 11-12: thoracic erector spinae muscles).

2.2 Research procedure

Fig. 2 is a diagram of the research flow used herein.

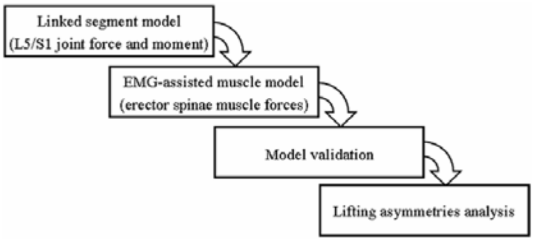


Fig. 2. The overall research process.

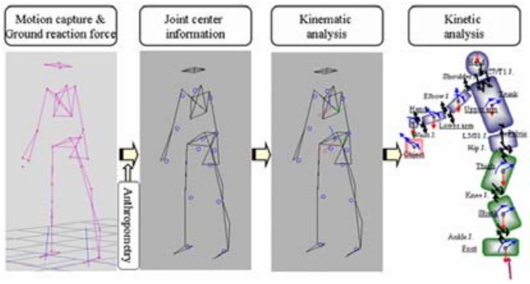


Fig. 3. LSM configuration.

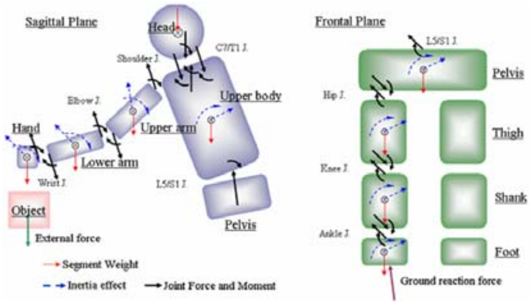


Fig. 4. The overall configuration of the 3D LSM.

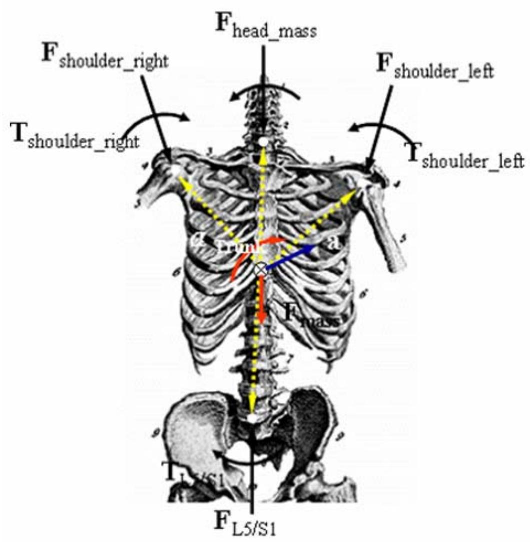


Fig. 5. Free body diagram of the trunk.

The entire process proceeded as follows. First, we developed a quantitative human model system centered on acquiring erector spinae muscle forces. This human model was formulated based on previous studies [18-22], and the present study was divided into a 3D linked segment model (LSM) and EMG-assisted (EMGA) muscle model. We validated the models following their development; we then used the models to analyze erector spinae force asymmetry during an experimental lifting exercise.

2.2.1 3D linked segment model (LSM)

Fig. 3 shows the LSM calculation process. The LSM used an inverse dynamic method that extracted information from each joint by using motion trajectories as inputs. The inputs were as follows: the anthropometric information of the subject, the three-dimensional trajectory information of the sensor attached to the whole body, and the reaction force. Initially, we calculated mass, center of mass, and inertia tensor of each segment using the anthropometry data and a regression model. We then calculated these segment parameters (mass, center of mass, joint center, inertia tensor) in a global coordinate system through the subject calibration process. The transformation matrix between this global reference system and the anatomical reference system was also derived in the subject calibration process. We then calculated the segment parameters during dynamic motion, using the transformation matrix calculated during subject calibration to convert the reference system. Finally, we calculated the joint reaction force and moment by adding the ground reaction information.

Additionally, we divided the whole body into 15 segments and 14 joints, and we used the Newton-Euler equation to develop a model of 41 degree-of-freedom, which was classified into upper body and lower body models. For the lower body models, we calculated ankle joint force and moment from ground reaction information and foot segment parameters; we calculated knee, hip, and L5/S1 joint information in the same way. Similarly, for the upper body models, we calculated wrist joint information from object external force and hand segment parameters; we calculated elbow, shoulder, neck, and L5/S1 joint information in the same way. Fig. 5 shows an example of the free body diagram of the trunk.

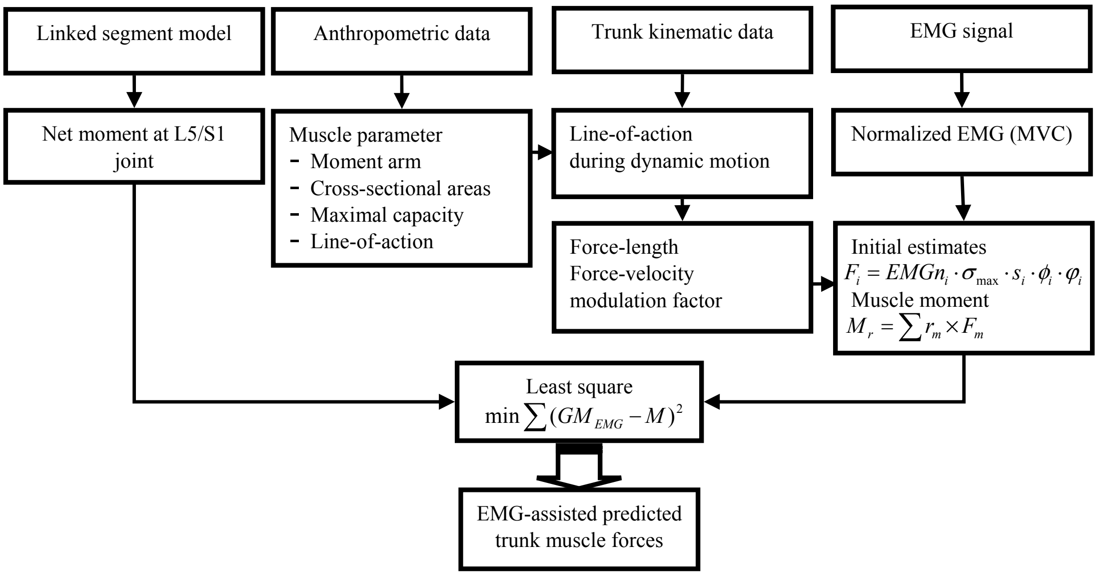


Fig. 6. Flow chart of the EMGA modeling process.

$$\sum \vec{F} = m\vec{a} \tag{1}$$
$$= \vec{F}_{L5/S1} + \vec{F}_{mass\_trunk} + \vec{F}_{C7/T1} + \vec{F}_{shoulder\_left} + \vec{F}_{shoulder\_right}$$
$$= mass_{trunk} \times \vec{a}_{trunk\_C.G}$$

$$\sum \vec{M} = I\vec{\alpha} \tag{2}$$
$$= \vec{T}_{shoulder\_left} + \vec{T}_{shoulder\_right} + \vec{T}_{L5S1} + \vec{T}_{C7T1} + \vec{r}_{shoulder\_left} \times \vec{F}_{shoulder\_left}$$
$$+ \vec{r}_{shoulder\_right} \times \vec{F}_{shoulder\_right} + \vec{r}_{head} \times \vec{F}_{head} + \vec{r}_{L5/S1} \times \vec{F}_{L5/S1}$$
$$= \frac{d(I_{trunk}\vec{\omega}_{trunk\_C.G.})}{dt} = \frac{d(I_{trunk})}{dt}\vec{\omega}_{trunk\_C.G.} + I_{trunk}\vec{\alpha}_{trunk\_C.G.}$$

2.2.2 EMGA (EMG-assisted) muscle model

The flow chart in Fig. 6 demonstrates the the EMGA muscle modeling process. We began developing the EMGA muscle model by regarding the trunk as an indeterminate body and assuming that the muscle forces around the L5/S1 joint generated the moment of the joint. We used the EMG signals from the main trunk muscles, the cross-section of the muscles, and the kinematic data from the trunk as input information. Muscle parameters were organized by trunk anthropometry data (trunk depth, trunk width), a method verified by the results of a previous study [24]. Using a regression model, we calculated the origin, maximal capacity, and the cross-sectional area of each muscle, and we extrapolated each insertion using the directional vector of the muscle. We also tracked the muscle origin and insertion by applying the transformation matrix during movement, and used this dynamic origin and insertion information to calculate force-length and force-velocity modulation factors. We normalized EMG data to the MVC value (the myoelectric maximum collected during maximum voluntary contraction). Fig. 7 shows the imaginary planes cutting the L5/S1 joint.

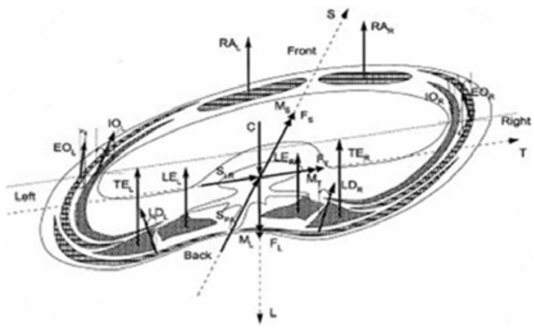


Fig. 7. The imaginary planes cutting the L5/S1 joint (scanned picture from Gagnon *et al*., 2002).

$$\vec{F}_i = EMGn_i \cdot \sigma_{\max} \cdot s_i \cdot \phi_i \cdot \varphi_i \tag{3}$$

$$EMGn_i = \frac{EMG_m}{EMG_{mvc}} \tag{4}$$

$$\sigma_{\max} = 35N/cm^2 \tag{5}$$

$$\phi_i = -3.2 + 10.2l_i - 10.4l_i^2 + 4.6l_i^3 \tag{6}$$

$$\varphi_i = 1.2 + 0.99v_i + 0.72v_i^2 \tag{7}$$

The force generated by each muscle, *i*, was represented by the product of normalized EMG, maximum muscle force generated per unit of cross-sectional area ($\sigma_{\max}$ was initially set at $35Ncm^{-2}$), muscle cross-sectional area (s_i) and modulation factor describing a relative length parameter (ϕ_i) and velocity of contraction parameter (φ_i). Also, variables l_i and v_i represent the instantaneous relative length and velocity of muscle i, respectively. Using the above equations, we calculated initial muscle force, from which we determined forces for each of the muscles by multiplying muscle forces from the early stage by the gains in values. We calculated the gain that was common to all muscles (G) on a per subject basis, using least mean square regression over the duration of each trial to obtain the best fit between the predicted moments from the EMG data and the measured moment vectors from LSM. The position vector r_i represents the lever arm of muscle i.

$$M_{EMG} = G\sum r_i \times F_i \quad (8)$$

$$\sum_{Frame}(GM_{EMG} - M)^2 = \min \quad (9)$$

2.2.3 Model validation

The LSM and EMGA muscle models were validated as follows, to boost the reliability of the information they generated. First, we validated the LSM by comparing the measured ground reaction forces from the force platforms with the estimated ground reaction forces; over all of the body segments, we summed the segment masses times the segment acceleration vector minus the gravity vector. In each case, the measured ground reaction force was the sum of the ground reaction forces from each of the force platforms. We further validated this model by comparing kinetic information for the L5/S1 joints from the upper body and lower body models. The quantitative validation results were the coefficient of correlation, RMS value, and maximum difference values. We used the coefficient of correlation to estimate how closely the upper and lower body kinetic information correlated with each other, and the RMS value represented the overall average difference between the two.

Secondly, the EMGA muscle model was validated via the comparison of the predicted moment of the L5/S1 joint from prediction via the muscle forces after sampling the main muscle force of the trunk with that from the skeletal model. Fig. 8 shows the vector representation of the trunk muscles.

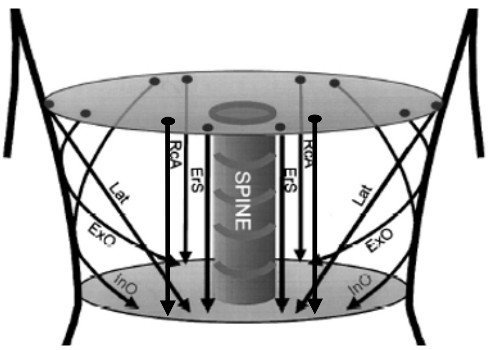


Fig. 8. Vector representation of the trunk muscles (scanned picture from William S. Marras *et al*., 1997).

2.2.4 Lifting experiments

Fig. 9 depicts the overall experimental system. We placed a 17-kg box 25 cm above the ground, between the two force platforms, and positioned the subject's feet as depicted. The box weight was selected based on a previously proposed study that a 17-kg loaded box is allowable for 90% of the population to safety perform the task [16]. Each of the ten subjects performed five lifts, with a three-minute rest between repetitions to alleviate fatigue. We did not designate a lifting posture; subjects were free to use the most comfortable posture for their ankle, knee, and hip joints.

2.3 Data analysis

We processed the erector spinae information obtained during experiments by dividing it into UW (unweighting), W (weighting), and E (entire) phases. The UW phase comprised the period from the initial posture to the raising of the box, the W phase was from raising the box to terminal posture, and the E phase was the sum of UW and W. In addition, in order to measure asymmetry between the two sides, we normalized the information from each side by dividing it by information from the total, as follows:

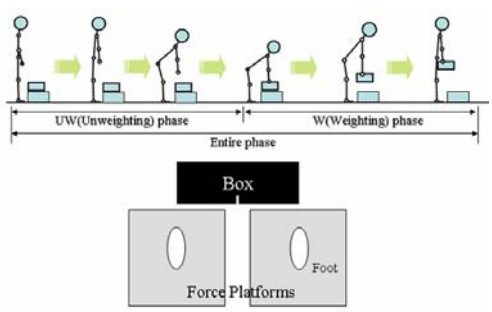


Fig. 9. Overall experimental system.

Table 2. Correlation and RMS differences between the measured and estimated ground reaction force (±S.D.).

	Correlation	RMS diff.
Forward – backward	0.58 (0.27)	14.7 (3.4)
Left – right	0.65 (0.19)	10.0 (5.1)
Upward – downward	0.93 (0.10)	8.1 (3.5)

Table 3. Correlation coefficients, RMS differences, and max. differences measured in the 40 trials (±S.D.).

	L5/S1 joint reaction force			L5/S1 joint moment		
	Correlation	RMS diff.	Max diff.	Correlation	RMS diff.	Max diff.
Lateral flexion	0.98 (0.1)	12.1 (1.2)	34.0 (16)	0.98 (0.1)	9.3 (5.0)	10.0 (1.9)
Flexion – extension	0.84 (0.2)	8.4 (3.0)	20.6 (12)	0.99 (0.1)	6.3 (0.8)	13.1 (4.7)
Twisting	0.97 (0.1)	11.1 (9.0)	20.0 (8.6)	0.90 (0.2)	4.8 (3.7)	5.7 (0.7)

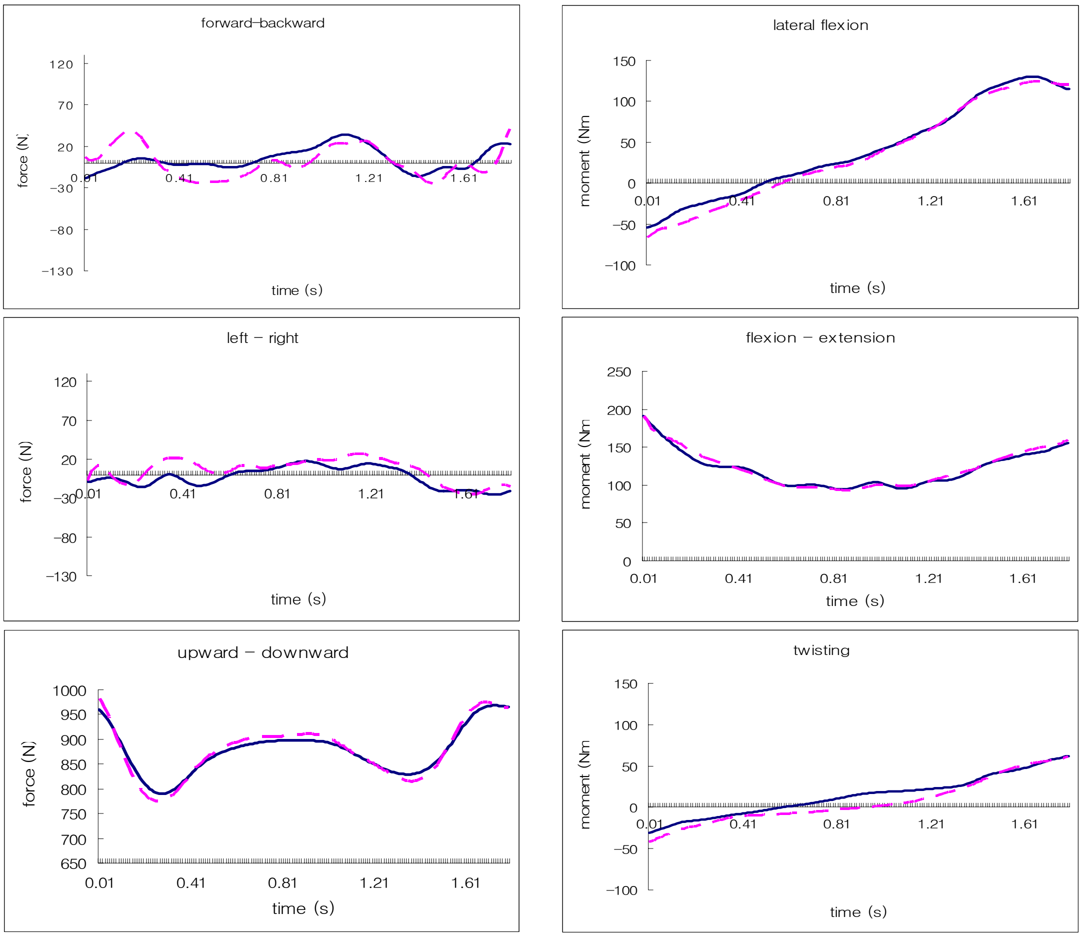
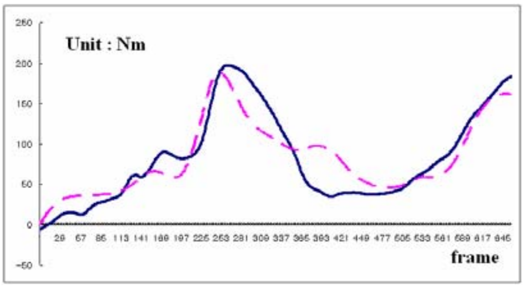
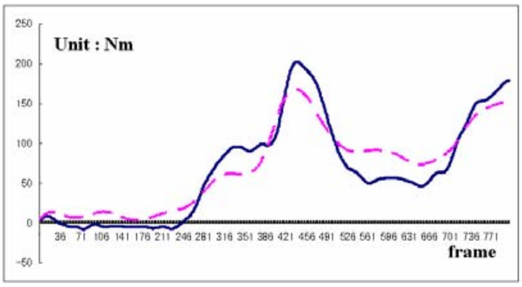


Fig. 10. Left panels: an example of the measured (solid line) and estimated (dashed line) ground reaction force during a twisting motion. Right panels: an example of the L5/S1 joint reaction moments of the lower body models (solid line) and upper body models (dashed line) during a twisting motion.



(a) Example 1



(b) Example 2

Fig. 11. Comparison of measured (solid line) and predicted (dashed line) moments at the L5/S1 joint during a lifting task.

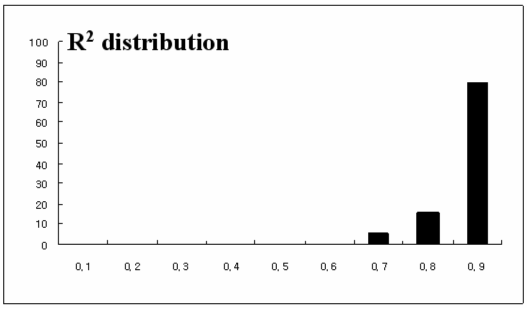


Fig. 12. The distribution of R^2.

$$Right(\%) = \frac{Right\ muscle\ force}{Right\ muscle\ force + Left\ muscle\ force} \times 100 \quad (10)$$

$$Left(\%) = \frac{Left\ muscle\ force}{Right\ muscle\ force + Left\ muscle\ force} \times 100 \quad (11)$$

3. Results

3.1 Model validation results

3.1.1 3D LSM

Fig. 10 represents the results of LSM validation; the left panel is the comparison of the measured ground reaction force (solid line) with the estimated ground reaction force (dashed line), and the right panel is the comparison of the measured L5/S1 joint moments from the lower body model (solid line) and the upper body model (dashed line). All moments are presented in the pelvic anatomical reference system. The ground reaction force example demonstrated that the measured and estimated information agreed closely, and the moment example revealed that the lower body and upper body models also agreed closely.

The coefficient of correlation and RMS difference quantitatively represent the comparison results between measured and estimated ground reaction forces (Table 2). We tested a total of 40 trial experimental results, comprised of lifting, twisting, and optional motions, and both the correlations and the RMS differences were the most similar in the upward-downward direction. This observation is in agreement with a previous study [18], in which correlation in the forward-backward direction represented the minimum value in the range of 0.079 to 0.781; our results showed a minimum value of 0.58 in the forward-backward direction as well. The previously reported RMS difference represented a range of median values from 16.9 to 25.3, and our RMS differences were similar.

Table 3 lists the correlation coefficients, RMS differences, and maximum quantitative differences measured in the 40 trials. These validation results are similar to those reported in a previous study [19].

3.1.2 EMGA muscle model

The EMGA muscle model was validated via the comparison of the predicted moment of the L5/S1 joint from prediction via the muscle forces after sampling the main muscle force of the trunk with that from the LSM. The comparison yielded an R^2 value of more than 0.7, and nearly 80% of the R^2 distribution was above 0.9 (Fig. 12). The distribution of R^2 in this study was similar to the distribution of the correlation coefficients in a previous study [24].

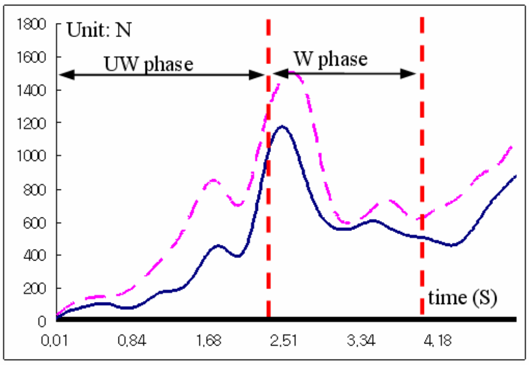
3.2 Asymmetry

Fig. 13 shows the bilateral asymmetry of erector spinae muscle forces, calculated by using the LSM and EMGA muscle models during a symmetric lifting task. The upper panel represents the lumbar erector

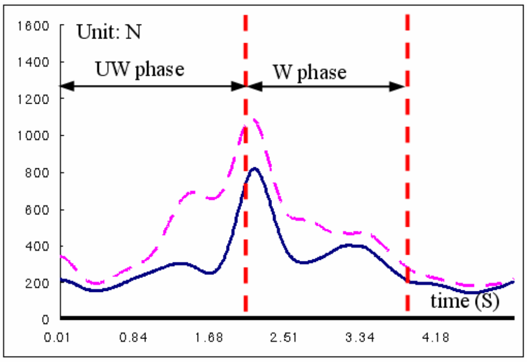
Table 4. Asymmetry of lumbar & thoracic erector spinae muscles (±S.D.).

		Bilateral diff.	Most asym.	Most sym.
Thoracic	UW	17.7 (5.7)	23.0 (5.8)**	11.6 (9.7)
	W	9.6 (5.7)	15.8 (2.8)*	3.2 (0.9)
	Entire	9.8 (7.7)	18.7 (3.5)*	1.9 (1.0)
Lumbar	UW	17.1 (6.9)	24.1 (6.4)*	12.0 (6.6)
	W	9.0 (4.5)	12.3 (3.4)*	4.4 (4.3)
	Entire	10.4 (4.5)	13.8 (2.9)*	7.4 (4.0)

(*p<0.05, **p<0.01 between right and left side
Bilateral difference = absolute average % difference across all subjects
Most asym. = the average of 5 lifts with the greatest difference from 50%.
Most sym. = the average of 5 lifts with the least distance from 50%.



(a) Lumbar erector spinae muscle



(b) Thoracic erector spinae muscle

Fig. 13. Bilateral force results of lumbar and thoracic erector spinae muscles from a 618N male subject. (Dashed purple line: right side of the trunk; solid blue line: left side of the trunk)

spinae muscle forces, and the lower panel represents the thoracic erector spinae muscle forces. The dashed purple line represents the right side of the trunk and solid blue line represents the left side of the trunk.

Both the thoracic and lumbar erector spinae muscles exhibited an average asymmetry of 200 N. In particular, the thoracic erector spinae muscle yielded a difference of more than 400 N during the UW phase; the maximum difference in the lumbar erector spinae was approximately 500 N.

Table 4 shows the normalized results for the left and right sides, from the information obtained from all ten subjects. We noted bilateral differences of up to 24.1% at the lumbar erector spinae and 23.0% at the thoracic erector spinae in normal people from the results, whereas we noted almost symmetric lifting due to the 4.4% difference at the lumbar erector spinae and a 1.9% difference at the thoracic erector spinae. Also, the bilateral difference of the UW phase was bigger than the W phase.

4. Discussion

The principal objective of this study was to assess bilateral asymmetries during symmetric lifting, using the load on the erector spinae, and to quantify the magnitude of bilateral asymmetries in a healthy group of individuals. The human model was developed in order to sample the load on the erector spinae, the utility was verified, and the asymmetries of the erector spinae muscle were analyzed by experiments conducted on a healthy group. This study involved the use of the load on the erector spinae, which is directly associated with LBP, as compared with using the ground reaction force in the previous study. In the future, the data from this study will aid in proposing a reliable threshold value for distinguishing between normal and pathological lower back situations.

We validated our LSM and EMGA muscle models using methods similar to those in previous studies. We validated the LSM by comparing both ground reaction and the L5/S1 joint information; other reports have already proven that validation through ground reaction information is a far more rigorous method than comparing L5/S1 joint information conducted on the upper and lower body models [18]. Validation results from both the ground reaction and L5/S1 joint information in this study were similar to those reported in previous studies [18, 19]. The ground reaction information in the upward-downward direction showed the highest correlation between predicted and actual results, perfectly matching results of a previous study [18]. Validation of the EMGA muscle model, not unlike the LSM, replicated the findings of previous reports; moreover, both of the studies yielded coefficients of correlation that were greater than 0.8

The results of this study show that the erector spinae muscles exhibit asymmetry as high as 24%, a much larger figure than the maximum of 10% (based on upward-downward ground reaction information) reported by Maines [16]. One explanation for the discrepancy is that the results of the present study come from a more detailed analysis than that of ground reaction information, *i.e.*, we evaluated the erector spinae muscles, which have a significant influence over lifting tasks. When considering muscle forces on the trunk as a whole, the bilateral difference of erector spinae muscles will be compensated, resulting in lower asymmetry maxima. Future steps will include this compensatory mechanism of trunk muscle forces.

The results of this study are summarized as follows:

(1) In order to quantify the bilateral asymmetry in the erector spinae muscles during a lifting task, we developed LSM and EMGA muscle models and validated them based on previous studies; this validation reinforced the reliability and validity of the results.
(2) Bilateral differences between the erector spinae muscles during a symmetric lifting task ranged from 3.2% to as high as 24% in a healthy male group.

As a preliminary step toward identifying an asymmetry threshold value that is predictive of LBP, we quantified bilateral asymmetry of the erector spinae muscles in a normal healthy group. However, this study was limited in that the experiments did not include a pathological group. Future experiments conducted in parallel on pathological and healthy groups will subsequently fine-tune the threshold value for LBP prediction.

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