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Analysis of internal torso loading in asymmetric and dynamic lifting tasks

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Abstract.
BACKGROUND: Asymmetric and dynamic lifting is known to be one of the leading causes of occupational lower back disorders (LBDs). Biomechanical modeling has been utilized to investigate lifting task characteristics so that the task demands can be kept within a limit, and internal muscles and joints are not injured.
OBJECTIVE: This study implemented AnyBody^{TM} to analyze internal torso loading in asymmetric and dynamic lifting tasks.
METHODS: A six-camera motion capture (mocap) system collected dynamic motion data of lifting 30 lb (13.6 kg) weight at 0°, 30° and 60° asymmetry. The mocap data drove the AnyBody^{TM} model, and the study investigated the effect of the asymmetry.
RESULTS: Erector spinae was the most activated muscle for both symmetric and asymmetric lifting. When lifting origin became more asymmetric toward right, erector spinae activity was reduced, but oblique muscles increased their share of activity to counter the external moment. Most muscle tensions peaked at the lift initiation phase except left external oblique and right internal oblique. Left external oblique played a minor role in the right asymmetric lifting task, and the difference of activation for right internal oblique may be due to variance of the motion. Surprisingly the lift asymmetry decreased both compression and shear forces at the L5/S1 joint.
CONCLUSIONS: This finding contradicted the results obtained from other research studies. The reduction in spine forces is postulated to have resulted from the increased oblique muscles' share in the production of back extensor moment. Since these muscles have longer moment arms, they generated lesser spine force to counteract the external moment. The subject also tended to squat as lifting origin became asymmetric, which effectively reduced the load moment on the spine.

Keywords: Biomechanical evaluation, asymmetric lifting, AnyBody^{TM} modeling, spinal forces

1. Introduction

Asymmetric lifting occurs in a great variety of workstations, and it is known to be one of the leading causes of occupational lower back disorders (LBDs). Occupational LBD is a manifestation from overloading of back extensor muscles and spinal tissues during lifting. Although many basic properties of human musculoskeletal system are measurable, internal forces of living tissue can rarely be measured directly during lifting task performance. Biomechanical modeling has been utilized to investigate lifting task characteristics so that the task demands can be kept within a limit, and internal muscles and joints are not injured.

To determine internal tissue forces more accurately, progressively more detailed anatomical models of the lower back have been introduced in modeling. However, due to presence of redundant muscle groups

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that may be active during such activities, the simple biomechanical model that can be solved from static equilibrium of forces, essentially becomes a statically indeterminate problem. Statically indeterminate problems are over defined, and as a result, Newtonian mechanics alone cannot predict the muscle forces. Two types of models have been developed to solve such statically indeterminate problems.

EMG assisted biomechanical models [1,2] were developed under the assumption that the muscle tension and electrode potential are correlated. Essentially these models partitioned the total back extensor moment during lifting into different muscle groups based on EMG signals collected from surface electromyography. However, the assumption of the correlation of muscle tension and electrode potential becomes uncertain under dynamic condition. Another category of model, optimization criterion based, assumes that muscles are recruited in such a way that a criterion function is minimized to reduce a biological cost, such as joint compression force [3,4] and muscle fatigue functions [5–7].

A detailed anatomical model of the lower back is beneficial to both categories of models. Current anatomical models of the lower back can not only consider all major muscle groups relevant in lifting activity, but also the muscle model can differentiate individual muscle fascicles of the individual muscle group [8,9] with consideration of muscle wrapping against bony structures [1,8–10].

The AnyBody™ Modeling System is commercially available, optimization criterion based modeling software. It provides by far the most detailed human torso musculoskeletal model. The torso model of AnyBody™ has been utilized effectively to validate internal muscle and joint forces [11–14], but none of the studies investigated the effect of asymmetric and dynamic aspects of lifting. This study implemented AnyBody™ to analyze internal torso loading in asymmetric and dynamic lifting tasks.

2. Method

2.1. Participant and task

This study was approved by the institutional review board. One healthy college student (1.73 cm, 75 kg) without any history of LBD during the past six months performed asymmetric lifting tasks of 0°, 30° and 60° with 30 lb (13.6 kg) dumbbell weights, placed evenly in a plastic tray, in OptiTrack™ mocap Laboratory (Fig. 1).

2.2. Experimental procedure

Before the experiment, the participant put on the OptiTrack™ medium-size mocap suit. With the help of laboratory assistant, thirty-four reflective markers were attached on the suit based on OptiTrack™ standard thirty-four-marker placement protocol [15]. After standard calibration and skeleton setting up procedure instructed by ARENA™ mocap software [15], the motion data of lifting were collected through OptiTrack™ six-camera tripod setup [15] with 100 frames/seconds. A thin metal stand supported the plastic tray with dumbbell weight to prevent marker blocking. During the experiment, the participant performed 0°, 30° and 60° lifting tasks in a randomized order. The participant stood straight with feet along with the tape of pre-defined angle, and lifted from the lift origin to upright position, at a normal pace, without moving feet.

The lift origin was fixed at knuckle height (99 cm off the ground), and at a horizontal distance of 53 cm from the center of the tray to the vertical body axis, which was dimensionally identical with Marras and Davis's study [16], so that the results could be compared. Asymmetric angles were taped on a force plate for feet positioning, including a sagittal symmetric position (0°), 30° and 60° to the right of the mid-sagittal plane. The force plate was used to collect ground reaction data during the lifting. The force plate data were not used in this study, but will be used later to check the validity of AnyBody™ model.

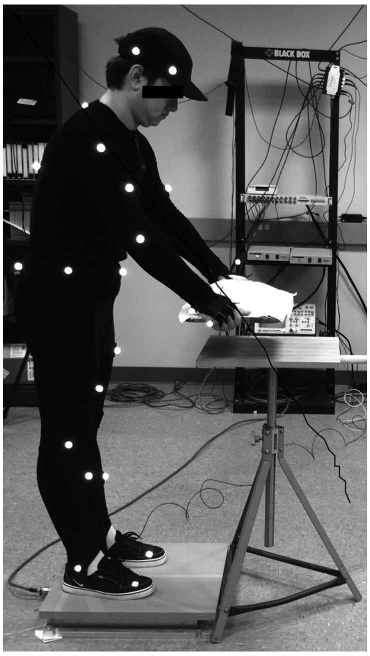

Fig. 1. Asymmetric lifting task configuration.

2.3. Data processing

ARENA™ software automatically filled missing frames less than 20, and smoothed data with cut-off frequency of 6 Hz. Gaps more than twenty frames were filled manually by visual inspection. The ".c3d" files were further trunked to capture the lifting activity only. Approximately between 160 and 220 frames were generated by ARENA™ for individual trials. Figure 2 shows the first frames of 0°, 30° and 60° asymmetric lifting simulated in inverse dynamic study by AnyBody™ model respectively.

GaitLowerExtremityProject model in AnyBody's Managed Model Repository1.31 was modified for the experimental task. Because pre-defined marker placement in AnyBody™ is different from reality, parameter and motion optimization algorithm was run before inverse dynamic calculation within AnyBody™ software. On a Sony VAIO® E series laptop computer with 2.2 GHz dual-core CPU and 3 GB RAM, inverse dynamic calculation took about 40 second/frame, but parameter and motion optimization lasted for hours depending how accurate the initial marker placement is.

3. Results

3.1. Muscle forces

AnyBody™ models muscle fibers in each muscle fascicles, for example, erector spinae (ES) is divided into a total of 29 fascicles on each side [9]. To obtain the approximate contribution by each muscle group, the fascicle forces were summed over the normalized duration of lifts (Figs 3–5).

ES was the most activated muscle for both symmetric and asymmetric lifting. Generally, RES and LES became less active as the lifting became more asymmetric. Oblique muscles became more active as

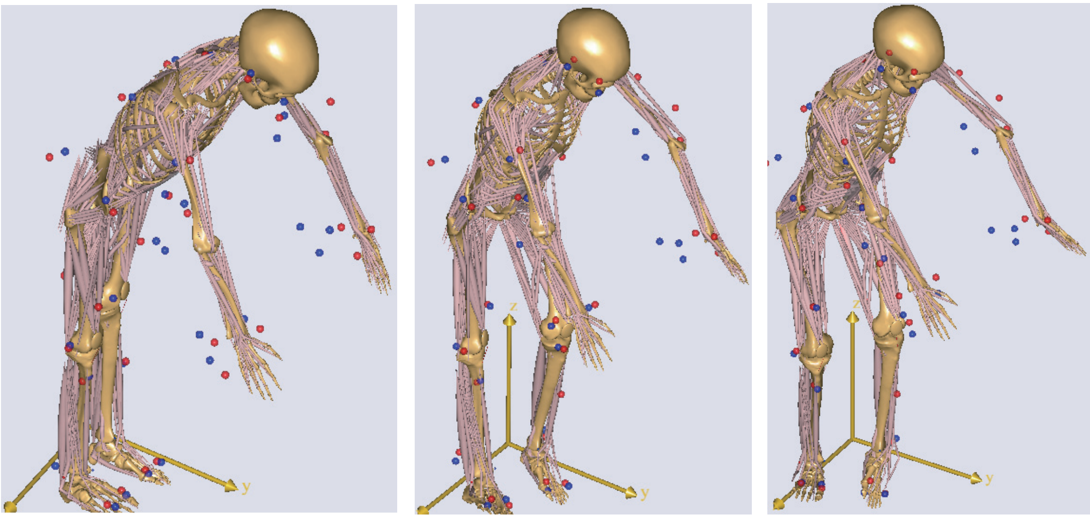

Fig. 2. First frames of 0°, 30° and 60° asymmetric lifting initialized in inverse dynamic study by AnyBody™ model.

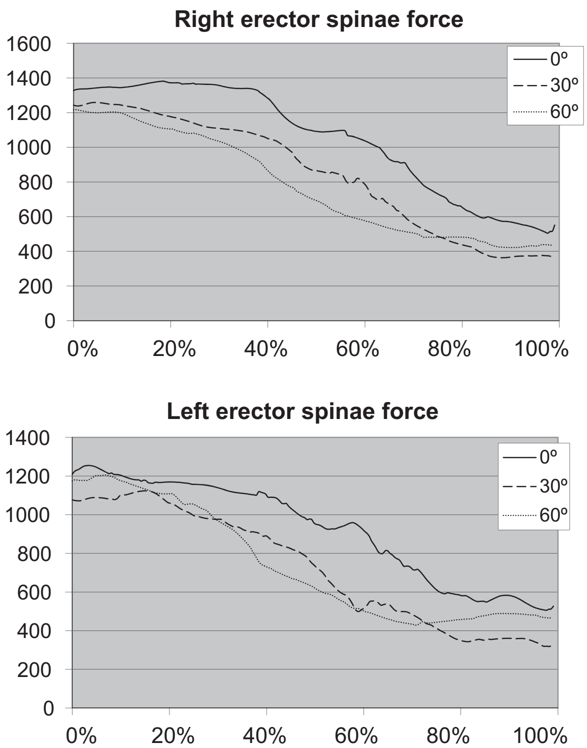

Fig. 3. Right erector spinae (RES) force (N), left erector spinae (LES) force (N) development during the lifting.

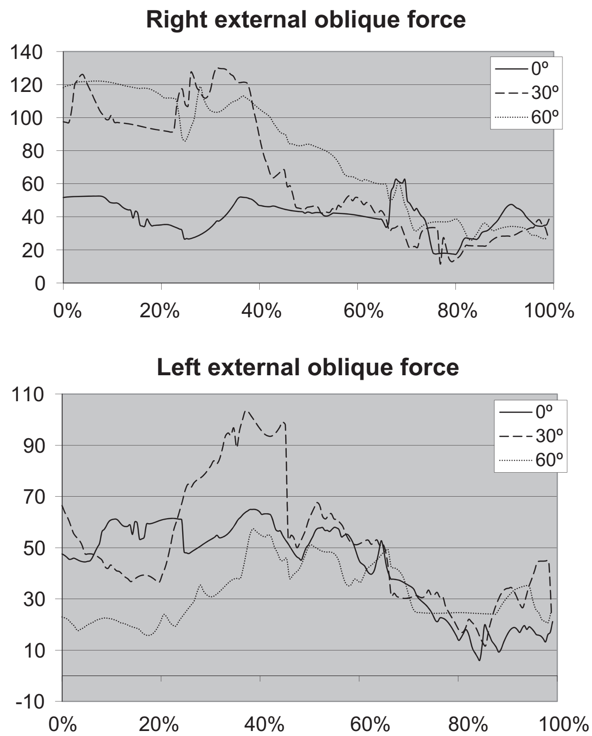

Fig. 4. Right external oblique (REO) force (N), left external oblique (LEO) force (N) development during the lifting.

the lifting became more asymmetric. Majority of the muscles were most active during the lift initiation phase, with exceptions for LEO and RIO. Since at the lift origin the load is farthest from the spine, as well as the upper body is maximally bent, the stronger muscle activity is expected. LEO played a minor

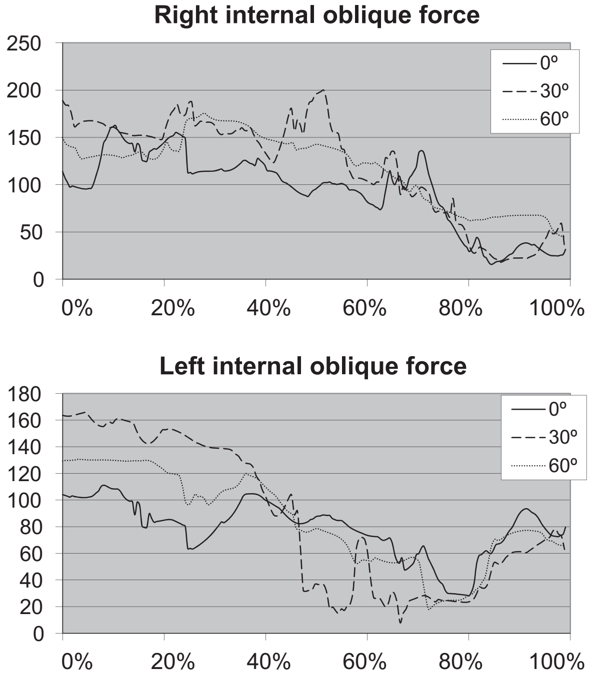

Fig. 5. Right internal oblique (RIO) force (N), left internal oblique (LIO) force (N) development during the lifting.

role in right asymmetric lifting task, and the difference of activation for RIO may be due to variance of the motion.

However, some observations cannot be properly explained. The zig-zag pattern of oblique activation may be due to the dynamic effect of lifts, resolution of mocap, or error tolerance of AnyBody^{TM} calculation. More data from different subjects need be collected for conclusive results. The more oblique forces for 0° or 30° than 60° at certain instances were also not explainable.

3.2. L5/S1 joint forces

L5/S1 joint compression, anterior-posterior (A-P) shear and lateral shear forces over the normalized duration of lifts are presented in Fig. 6. Compression and A-P shear forces followed the similar pattern, which was identical with ES muscle forces. At the beginning and the end of lifting, the joint loads were steadier than between, probably due to the requirement of movement control. Comparing with 0° lifts, L5/L1 maximum compression force reduced from 3156 N to 2963 N by 6.1% and 2888 N by 8.5% for 30° and 60° respectively; maximum A-P shear force increased 2.3% to 568 N for 30°, but reduced 6.5% to 519 N for 60° respectively comparing with 0° from 555 N; absolute lateral shear force reduced from 52.6 N for 0° to 44.6 N by 15.2% and to 23.8 N by 54.8% for 30° and 60° respectively. In general, joint forces reduced as lifting origin became more asymmetric.

4. Discussion

ES is the main extensor of trunk. When the ES fascicles of one side act together, they produce combined lateral flexion and rotation to the same side [17]. During asymmetric lifts, the support of the

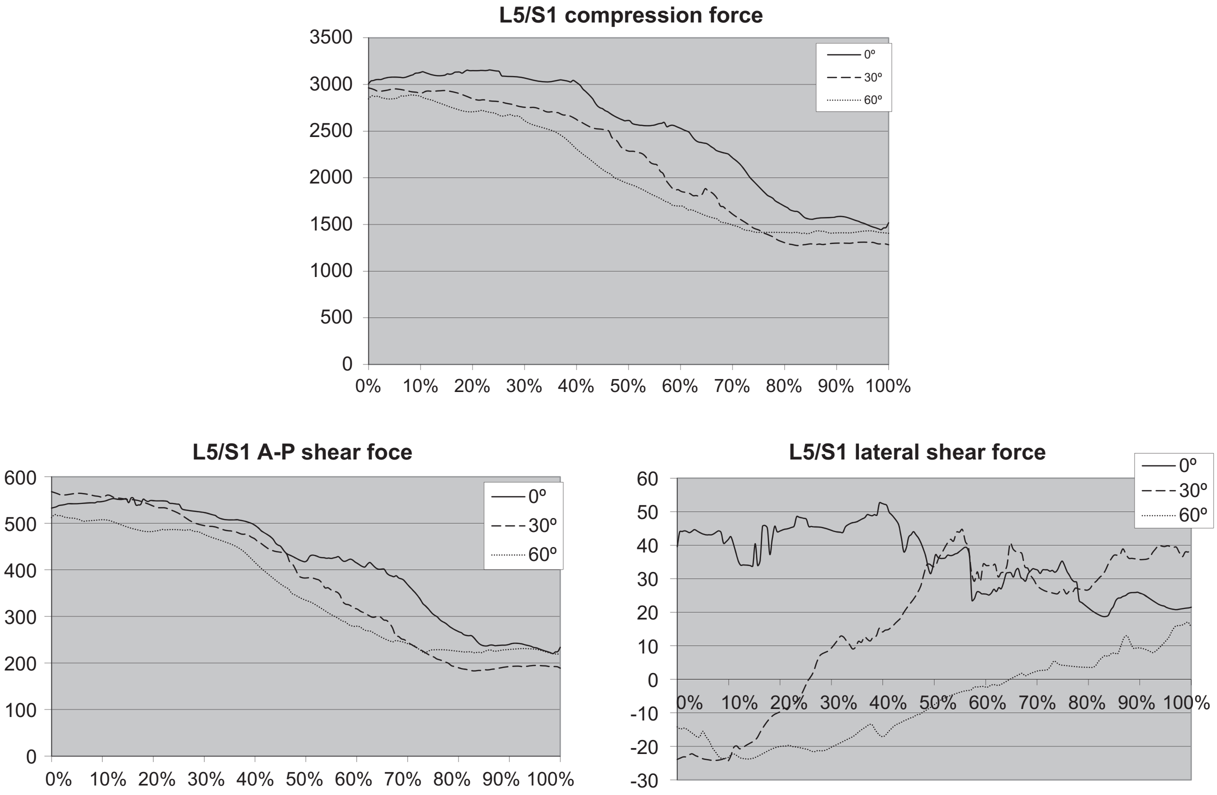

Fig. 6. L5/S1 compression, anterior-posterior (A-P) shear and lateral shear forces (N) during the lifting.

external load is shifted from the large ES muscles to smaller, less capable oblique muscles [18]. Biomechanically, ES has smaller moment arm than oblique muscles referring to lumber joint, so ES is less efficient to support external moment generated by upper body weight and hand loads. When the support of the external moment shifts from ES to oblique muscles, which also means shifting to more efficient muscles, the joint forces should reduce. However, oblique muscles are much weaker than ES, so they are less activated during symmetric lifting to minimize muscle fatigue. Furthermore, from observation (Fig. 2), the participant tended to flex and twist his knee joints more as lifting origin became more asymmetric, which may also be a strategy of our body to reduce internal joint forces.

According to NIOSH [19], the tolerance level for compression loading of the spine is expected to be around 3400 N. At this level of compression, micro fractures of the vertebral endplate begin to occur. The threshold limits for spine lateral and A-P shear are probably less than 900 N [16]. Reducing A-P shear and compressive forces should be considered a priority to prevent LBDs [16]. In this study, joint forces did not exceed the limitation. However, if certain factors such as lifting speed, lifting height and lifting weight become more demanding, joint forces may exceed the tolerance level, and long time working under those circumstances may develop LBDs.

The average maximum L5/S1 compression force derived from ten subjects by Marras et al.'s EMG assisted model [16] was 3600 N, 3900 N and 4050 N for asymmetric lifting of 0°, 30° and 60° toward right, which was presented graphically. Compression forces increased as the lifting origin became more asymmetric, which was contradicted with this study. A-P shear force was approximately 910 N, 850 N and 830 N for 0°, 30° and 60 asymmetry respectively. Comparing with this study, both A-P shear force deceased as the lifting origin became increasingly asymmetric, but the force predicted by Marras et al. was about 350 N higher than this study. Lateral shear force predicted by them ranged from 210 N to 350 N, which was far higher than the values predicted by AnyBody^{TM} in this study. Generally, they found compression and lateral shear forces increased as the lift origin became more asymmetric, whereas A-P shear force decreased. The EMG assisted model is based on the assumption that the muscle tension correlates well with the electrode potential. This assumption is valid when the muscle contraction is isometric, that is muscle fiber lengths remain unchanged during force production [20]. However during dynamic situation, when muscle fibers generates force as well as change their lengths, sliding action of muscle fibers underneath the fixed surface electrodes, also induces electrode potential [20]. Unless the dynamic part of the electrode potential is separated from the gross electrode potential, EMG may not accurately estimate force generation by the muscle fibers.

5. Conclusions

Commercially available AnyBody^{TM} biomechanical model provides by far the most detailed human anatomical model, which is driven by criterion optimization algorithm. To our knowledge, the model has been used for the first time to evaluate dynamic and asymmetric lifting. ES was the most activated muscle for both symmetric and asymmetric lifting. When lifting origin became more asymmetric toward right, ES activity was reduced, but oblique muscles increased their share of activity to counter the external moment. Most muscle tensions peaked at the lift initiation phase except left external oblique and right internal oblique. Surprisingly the lift asymmetry decreased both compressive and shear forces at the L5/S1 joint. This finding contradicted the results obtained from other research studies.

Future research

More data from different subjects should be collected for conclusive results. Force plate data should be used later to check the validity of AnyBody^{TM} model.

Conflict of interest

The authors have no conflict of interest to report.

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