Journal of Electromyography and Kinesiology 24 (2014) 387–393

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Effect of changes in orientation and position of external loads on trunk muscle activity and kinematics in upright standing

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ARTICLE INFO

Article history:
Received 6 August 2013
Received in revised form 13 February 2014
Accepted 15 February 2014

Keywords:
Electromyography
Trunk muscles
Force orientation
Kinematics
L5-S1 moment
Force position

ABSTRACT

Forces at different heights and orientations are often carried by hands while performing occupational tasks. Trunk muscle activity and spinal loads are likely dependent on not only moments but also the orientation and height of these forces. Here, we measured trunk kinematics and select superficial muscle activity of 12 asymptomatic subjects while supporting forces in hands in upright standing. Magnitude of forces in 5 orientations (−25°, 0°, 25°, 50° and 90°) and 2 heights (20 cm and 40 cm) were adjusted to generate flexion moments of 15, 30 and 45 N m at the L5-S1 disc centre. External forces were of much greater magnitude when applied at lower elevation or oriented upward at 25°. Spinal kinematics remained nearly unchanged in various tasks.

Changes in orientation and elevation of external forces substantially influenced the recorded EMG, despite similar trunk posture and identical moments at the L5-S1. Greater EMG activity was overall recorded under larger forces albeit constant moment. Increases in the external moment at the L5-S1 substantially increased EMG in extensor muscles ($p < 0.001$) but had little effect on abdominals; e.g., mean longissimus EMG for all orientations increased by 38% and 75% as the moment level altered from 15 N m to 30 N m and to 45 N m while that in the rectus abdominus increased only by 2% and 4%, respectively. Under 45 N m moment and as the load orientation altered from 90° to 50°, 25°, 0° and −25°, mean EMG dropped by 3%, 12%, 12% and 1% in back muscles and by 17%, 17%, 19% and 13% in abdominals, respectively. As the load elevation increased from 20 cm to 40 cm, mean EMG under maximum moment decreased by 21% in back muscles and by 17% in abdominals.

Due to the lack of EMG recording of deep lumbar muscles, changes in relative shear/compression components and different net moments at cranial discs despite identical moments at the caudal L5-S1 disc, complementary model studies are essential for a better comprehension of neuromuscular strategies in response to alterations in load height and orientation.

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1. Introduction

Low back pain (LBP) is a major health problem in the industrialized world and is by far the most prevalent and costly musculoskeletal disorder among US industries (NIOSH, 1997). Spine disorders are the most prevalent cause of chronic disability in persons less than 45 years (Ashton-Miller and Schultz, 1997). Although the cause of most low-back disorders remains unknown, biomechanical factors need to be taken into account when searching for an adequate understanding of the mechanisms involved (Marras et al., 2001).

The difficulty to determine with acceptable accuracy the loads on the trunk passive ligamentous and active musculature is a critical hindrance towards development of ergonomics guidelines for safer occupational environments. Realistic estimation of spinal loads in various daily recreational and occupational activities is also essential for the effective rehabilitation, surgical/conservative treatment, implant design, and performance enhancement programs. Towards this goal, a number of predictive equations have been proposed using biomechanical models of various complexities (Arjmand et al., 2011; El Ouaaid et al., 2009, 2013; Fathallah et al., 1999; Hoozemans et al., 2008; McGill et al., 1996; Merryweather et al., 2009). These works have to different degrees of accuracy considered the effect of posture and load magnitude/lever arm on spinal loads. They however have not accounted for the effect of changes in load height and orientation that exists for example while pulling and pushing objects. Alterations in the

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http://dx.doi.org/10.1016/j.jelekin.2014.02.005
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height at which weights are held in hands (Granata and Orishimo, 2001) or horizontal anterior shear forces are applied on the trunk (Kingma et al., 2007) have, despite identical net moments (at the L5-S1 and L3-L4 lumbar discs, respectively), been found to markedly influence muscle electromyography (EMG) activities. Changes in the load orientation held in hands at constant spinal moments have not yet been investigated. The resultant external force magnitude as well as its relative anterior shear and axial force components substantially alter when orientations vary.

Trunk muscle activity and loads at a spinal level are likely dependent not only on the moments of external loads at that level but also on the orientation and height of the external loads. Despite similar posture and identical external net moment (at the L5-S1 disc center), changes in magnitude, elevation and direction of applied force vectors would alter the absolute and relative magnitudes of compression, shear and moment loads at different spinal levels with likely consequences on the neuromuscular response. The current work aims to measure trunk kinematics and surface EMG activities of select muscles in asymptomatic volunteers in upright standing carrying loads via cables at different orientations and locations. Magnitude of forces is chosen so as to maintain the net flexion moments at the L5-S1 disc center constant. For this purpose, two heights, 5 orientations and 3 moment levels were considered. It is hypothesized that, despite identical posture and moments at the L5-S1 disc center, the orientation and height of the externally applied load vector markedly influence muscle EMG activity levels. In this event, such parameters should explicitly be included in predictive equations if and when accurate spinal loads are expected.

2. Methods

Twelve healthy university student males with no recent back complications and back pain history participated voluntarily after signing an informed consent. Participants had 72.98 ± 3.87 kg weight, 177.67 ± 3.03 cm height and 23.25 ± 1.82 kg/m^2 body mass index (BMI). Kinematics was measured using 12 clusters: one on each foot, thigh, upper arm and forearm in addition to one on the pelvis, T12, C7 and head. Four LED markers were glued on each cluster (except for the feet with seven LEDs) for a total of 54 LEDs. Positions of the markers were measured in three dimensions at a sampling rate of 30 Hz using a Five-camera Optotrak system (Northern Digital, Waterloo ON, Canada). Clusters on each segment were related to anatomical landmarks by recording while pointing at each landmark with an Optotrak probe containing 25 markers. A total of 48 anatomical markers were probed in order to find joint centers. An additional LED was placed on the loading cable to track its spatial positions. Kinematics was low-pass filtered at 10 Hz.

A trunk dynamometer was used (Lariviere et al., 2009) to record EMG at maximal efforts. In a semi-seated position and instructed to generate maximal efforts (Baratta et al., 1998), subjects performed three trials in extension, two in flexion, two in lateral bending (on each side) and two in axial rotation (on each side). Each trial lasted 10 s while the subject exerted maximal force pushing against a harness. The dynamometer signals were collected at a sampling rate of 128 Hz. EMG signals were collected at a 1024 Hz using 12 active surface electrodes with single-differential dry surface electrodes [Model DE-2.3, DelSys Inc., Wellesley, MA; bandpass filter 20–450 Hz, preamplification gain 1000, CMRR −92 dB, input impedance > 10^{15} Ω//0.2 pF, noise 1.5 μV (RMS, R.T.I.)] composed of two parallel silver bars (10 mm long, 1 mm wide) spaced 10 mm apart. After the skin at the electrode sites was shaved and abraded with alcohol, electrodes were positioned bilaterally over trunk muscles: longissimus LG (∼3 cm lateral to the midline at the L1), iliocostalis IC (∼6 cm lateral to the midline at the L3), multifidus MF (∼2 cm lateral to the midline at the L5), rectus abdominus RA (∼3 cm lateral to the midline above the umbilicus), external oblique EO (∼10 cm lateral to the midline above umbilicus and aligned with muscle fibers) and internal oblique IO (∼2 cm below and 7 cm medial to the anterior superior iliac spine) according to previous guidelines (De Foa et al., 1989; McGill, 1991). The difficulty in capturing the MF muscle with surface electrodes (Stokes and Gardner-Morse, 2003) is acknowledged. The electrodes intended to pick the activity of the IO likely capture the activation of the transverse abdominus as well (Marshall and Murphy, 2003).

All measured EMG signals were bandpass filtered (30 and 450 Hz, 8th order zero-lag Butterworth IRR filter) to remove high frequency noise and ECG artifacts; ECG being dominant in torso EMG signals (Redfern et al., 1993). Subsequently, the root mean square (RMS) of task trials was calculated choosing a time-window of 6 s without overlap. For EMG signals recorded during the MVCs (lasting 10 s each), maximal root mean square (RMS) values were calculated over successive (50-ms overlapped) 500-ms time-windows. For normalization, the EMG data of each muscle obtained from all trials under various loads were divided to their maximal value for each subject. Recorded left and right EMG data for each muscle in maximal efforts and regular tasks were averaged due to insignificant differences ($p > 0.15$).

While in upright standing position, each subject performed 6 distinct static tasks at 2 repetitions each (Fig. 1). These were further repeated at 3 load levels generating identical flexion moments of 15 N m, 30 N m and 45 N m at the L5-S1 disc: (1) Force heights: Anteriorly-directed horizontal forces were carried in hands at 2 heights of H_1 = 20 cm and H_2 = 40 cm with respect to the L5-S1 disc center. (2) Force orientations: At the H_2 = 40 cm height, the orientation of the force vector supported at hands via a cable (Fig. 1) varied from downward vertical (90°) to downward inclined (50° and 25°), horizontal (0° which is the same as that considered above) and finally to upward inclined (−25°) directions.

The weight carried in hands via the cable (Fig. 1) was adjusted for each orientation and subject in a manner to yield the desired moments at the L5-S1 disc center. For this purpose, the sagittal coordinates of the L5 marker on the skin was taken to be 8.99 cm posterior and 0.94 cm proximal to the L5-S1 disc center (Snyder et al., 1971).

To avoid fatigue, 1 min rest was considered after each trial lasting about 10 s. To minimize the effect of inter-subject changes in gravity moment at the L5-S1 level, a near-homogeneous population was considered with nearly identical heights and weights. Subjects were instructed to look forward at all times and to keep the posture and position of hands constant as much as possible (monitored using fixed reference bars at the hand levels and behind the subjects). The data recorded at the 2nd repetition were considered for subsequent analyses.

Repeated analysis of variance (ANOVA with post hoc Tukey) was carried out on the normalized EMG of each muscle (3 extensors and 3 abdominals) and weights held in hands to analyze the effects of load orientations (5 levels) and moments (3 levels). Additional analyses were performed to study the effect of 2 heights (H_1 and H_2) at 3 moment levels. Moreover, the effects of orientation and moment on 3 dependent kinematics variables of trunk, pelvic and lumbar rotations were also analyzed at 3 instances of 2 s, 5 s and 8 s during tests.

3. Results

Analysis of trunk kinematics demonstrate that the subjects indeed held their posture almost unchanged at all times as moment,

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