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Capability and recruitment patterns of trunk during isometric uniaxial and biaxial upright exertion

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Abstract

Background. Work-related risk factors of low back disorders have been identified to be external moments, awkward postures, and asymmetrical dynamic lifting amongst others. The distinct role of asymmetry of load versus posture is hard to discern from the literature. Hence, the aim of this study is to measure isometric trunk exertions at upright standing posture at different exertion level and degree of asymmetry to further delineate the effects of exertion level and asymmetry on neuromuscular capability response.

Methods. Fifteen healthy volunteers randomly performed trunk exertions at three levels (30%, 60%, and 100% of maximum voluntary exertion and five different angles (0°, 45°, 90°, 135°, and 180°) of normalized resultant moments. During each trial, the normalized EMG activity of 10 selected trunk muscles was quantified.

Findings. The EMG activity of the 10 trunk muscles was significantly ($P < 0.001$) affected by the level of exertion and angle of normalized resultant moment, and their interactions. The controllability of the torque generation was reduced in biaxial exertions. The capability to generate and control the required trunk moments is significantly lowered during biaxial trunk exertions, while all muscles present higher EMG activity. These results suggest that the trunk muscles will be taxed higher while performing biaxial exertion tasks, increasing muscle fatigue possibly leading to a higher probability of low back injury.

Interpretation. The prediction of biaxial trunk performance based on uniaxial data will result in an overestimation of capability and controllability of the trunk during physically demanding tasks. This study provides a better understanding of the potential mechanisms of injury during asymmetrical and biaxial trunk exertion during work-related tasks.

Keywords: Isometric trunk exertion; Biomechanics; Ergonomics; EMG; Abdominal muscles; Lumbar spine; Low back

1. Introduction

Among various risk factors associated with work-related low back pain, the maximum physical capacity is often used to determine a safe lifting limit. It is generally believed that the closer the work-related load is to the strength capability (maximum trunk torque output), the greater the risk of injury (Davis and Marras, 2000; Lavender et al., 1993; Marras and Granata, 1995).

Davis and Marras (2000) measured dynamic strength capability and EMG activity of the trunk muscles during uniaxial or biaxial trunk exertions while manipulating resistance, velocity, or acceleration. These studies reported a significant co-activation among the trunk muscles, varying depending on whether trunk velocity or acceleration was manipulated (Davis and Marras, 2000; Kumar et al., 2003; Marras and Mirka, 1990). While these studies provide a great understanding of human performance during complex tasks, a more basic research is needed to provide a better understanding of the human motor behavior as related to various work-related symmetrical and asymmetrical trunk exertions. The variation of the measured

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strength, as reported by these studies, may be due to many factors such as changes in muscle lever arm or length–tension curve due to postural changes, as well as changes in motor behavior. These studies are limited in isolating the effect of the trunk twisting and bending torque modulation from the changes in the geometry of the muscle and the length of the soft tissues. Tsuang et al. (1993) reported that twisting and bending changes the trunk muscles' lever arm, length–tension ratio, and centroid during trunk twisting movement.

In contrast, static testing conditions, although representing less realistic conditions, may offer a better opportunity to study the contribution and interaction of individual risk factors. The aim of this study is to understand muscle recruitment patterns and the capability and controllability of trunk isometric uniaxial and biaxial exertion within upright posture, while all confounding variables such as muscle lever arm and muscle length–tension are controlled.

The hypothesis of this study is that the mean of the normalized root mean square of electromyographical signal (NRMS-EMG) of the 10 selected trunk muscles will change significantly with the magnitude (30%, 60%, and 100% of maximum voluntary exertion (MVE)) and angles of normalized resultant moments (ANRM) of the trunk (0°, 45°, 90°, 135°, and 180°). The orientation of the net resultant moment represents isometric uniaxial flexion, extension, and right transverse exertions, and two isometric biaxial exertions involving combined sagittal and transverse.

2. Methods

2.1. Subjects

Thirteen males and five females with no history of back pain for the past six months volunteered for this study. The subjects were recruited based on advertising in the New York University. The study was approved by the Internal Review Board.

The mean (±SD) age was 29.8 (4) years for males, 27.2 (8) years for females, and 28.5 (6.4) years for all subjects.

2.2. Instrumentation

A multichannel data acquisition system was developed in-house to simultaneously collect 16 channels of EMG signals and mechanical data from a B200 Isostation (Isotechnologies Inc, Hillborgh, NC, USA). EMG data were collected at a sampling rate of 1000 Hz with a system that utilizes a differential amplification method and possesses a gain of 1400, a CMRR of 130 dB, an input impedance of 10 GΩ, and a high pass filter of 25 Hz (Innovative Computer Solutions, Fresh Meadows, NY, USA). The sampling rate of 1000 Hz adequately captured the EMG signals obtained in this study (up to 450 Hz in the worst case) without appreciable distortion in recording. The Labview software (National Instrument, TX, USA) was programmed to provide real-time visual feedback of EMG and mechanical signal that was collected at 1000 Hz via A/D board (ATMIO-64F, National Instrument, TX, USA). During each trial, the program displayed a feedback screen of the subject's generated sagittal and transverse torque and 5% of the expected exertion target (see Fig. 1).

2.3. Procedure

After careful explanation of the study, and positive answer for participation each subject signed the consent form. The subject was screened by a demographic and medical questionnaire (Sheikhzadeh, 1997). The questionnaires were reviewed to ensure that the subject had no history of low back pain in the last six months, nor any neurological and physiological conditions that would make him/her unsuitable for this study.

2.4. Electrode placement

Surface electrodes and wire electrodes were used to monitor and collect the myoelectric activity of five bilateral trunk muscles. The surface electrodes were bipolar, pregelled surface electrodes (Classic Medical, Muskego, WI, USA) with an inter-electrode separation of 20 mm. Two pairs of electrodes were placed bilaterally on the medial erector spinae (ES) at level of L3–L4, latissimus dorsi (LAT) at T12–L1 level, the external oblique (EOB) muscles midway between the iliac crest and the lower border of the rib cage, and rectus abdominus (RA) muscle at level of L3–L4.

To insure the safety and accuracy of wire electrode placement, wire electrodes were placed using a needle biopsy procedure under CT-scan (CT 9800 HiLight, General Electrics) imaging by a board certified radiologist. At the L3/L4 level, the insertion of wire electrode was done

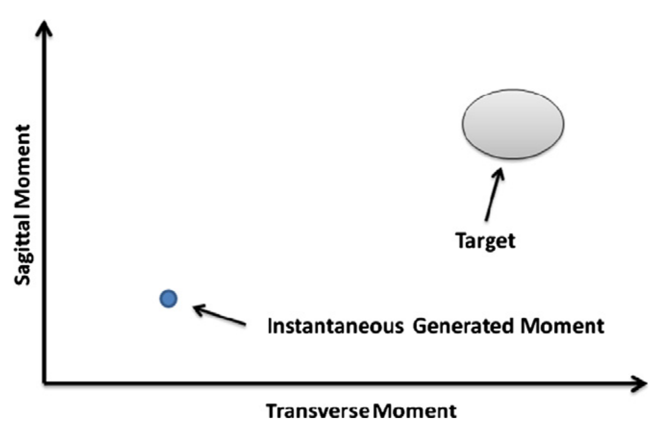

Fig. 1. The visual feedback screen provided for the subject during each trial. The horizontal axis exhibits right and left rotation transverse moment. The vertical axis exhibits flexion and extension moment in the sagittal plane. The small black circle represents the real time display of torque being exerted. The large gray circle indicates the target area and shows ±5% of target torque.

under CT-scan into the middle third region of the right and left IOB muscle.

2.5. Maximum voluntary exertion (MVE)

The subject was strapped to the B-200 Isostation in upright standing so his L3–L4 interspace was aligned with the flexion–extension axis of the machine. Each subject was restrained by the chest pad, thoracic strap, pelvic pads, and thigh straps of the B200 Isostation according to the manufacturer's recommended procedures. The trunk sagittal and transverse torqueses were measured by the B-200 Isostation. The subject was informed that the lateral axis was not locked. Only the lateral plane range of motion was monitored and the subject received audio feedback if torque produced in lateral plane caused more than 5° lateral flexion of the trunk.

Each subject performed two trials of trunk MVE randomly in four directions: flexion, extension, and right and left axial rotation. The subject was asked to perform at a maximum effort for 3 s with no jerky exertion. Each trial was followed by 2 min of rest to avoid fatigue. If the MVE of the two trials varied more than 10%, the subject was tested for a third MVE and the two trials with consistent values were selected. The average of the two trials of MVE was used to calculate percentage of MVC within the sagittal and transverse planes.

2.6. Uniaxial and biaxial exertions

The computer guided uniaxial and biaxial exertion conditions by generating real time visual display of generated torque and a target area as shown in Fig. 1. Each subject was asked to produce isometric torque along each axis until they reached to the target area. During biaxial MVE, subject was instructed to reach the MVE target in each axis simultaneously. The target area was set to be equal to ±5% of 0%, 30%, 60%, and 100% of the isometric trunk MVE for attempted sagittal flexion and extension, and axial trunk rotation.

To reduce the number of testing conditions, the risk of fatigue, and subject discomfort, only axial exertion to the right side was tested. The condition of relaxed upright standing in the dynamometer provided the baseline EMG for the normalization of the data. Each subject performed a total of 32 trials (2 repetitions × 16 conditions) of uniaxial and biaxial exertions as shown in Fig. 2. Level of exertion was randomly tested. Between each testing condition, the subject rested for 2 min. The trials were randomized and subject blinded to level of exertion.

Before each test, the feedback screen was displayed on the computer terminal and subject was asked to practice reaching the target area and testing the lateral audio feedback signal. Based on this performance, verbal feedback was provided. The practice was continued until the subject was comfortable with the task and indicated that he/she was ready to start the test.

2.7. Data reduction

For the purpose of statistical analysis the time window representing the most stable 3 s were used for computation. Occasionally the best selected value for biaxial trails was out of the designated target value. The capability index at each instant of time was computed by Eq. (1), which quantifies the distance between continuous exerted moments and expected target torque value. This index serves as a normalized deviation of performed exertion from desired target. A capability lower index value indicated a greater ability to accurately match the desired (target) torque levels:

$$\text{Capability index} = \frac{\sqrt{(\tau_S - \tau_{ST})^2 + (\tau_T - \tau_{TT})^2}}{\sqrt{(\tau_{ST})^2 + (\tau_{TT})^2}} \quad (1)$$

where τ_S and τ_T are the performed sagittal and transverse torque, respectively; τ_{ST} and τ_{TT} are the sagittal and transverse target torque respectively.

For the selected three seconds of each trial, the variability of the generated torque was quantified by the coefficient of variation (CV) computed based on the ratio of the standard deviation over the mean of the most stable 3 s of torque generated in each axis. The higher value of CV shows a larger variability of performance that indicates a lower controllability over the generated torque by the trunk muscles.

For all 10 trunk muscles, the mean of the normalized root mean square of the EMG signal (NRMS-EMG) was calculated for the selected 3 s window. Based on previous recommendations, the time constant of 50 ms was used for the RMS calculation. As suggested by Lavender et al. (1992) and Seroussi and Pope (1987), the EMG signal of each muscle was normalized to the maximum and rest value of that muscle. In this study, the highest mean RMS-EMG value among all the trials, the MVE and test conditions, was selected for the normalization procedure.

2.8. Statistical analysis

Descriptive statistics (mean and standard deviation) were computed for demographic data, maximum isometric torque, and EMG activity of the 10 trunk muscles, indices of capability and controllability of performance. Multivariate analysis of variance, MANOVA, with a repeated measures design (SAS Institute Inc., 1985) was used to determine the main effects of ANRM (θ) and the level of exertion on the NRMS-EMG of selected trunk muscles (Fig. 2). If the overall MANOVA was significant at $P < 0.05$, a univariate analysis of variance (ANOVA) was performed. The Tukey test was performed on the levels of ANRM and level of trunk exertion, if the effect of the univariate test was statistically significant. In addition, repeated measures ANOVA was used to determine the effect of ANRM on the capability index.

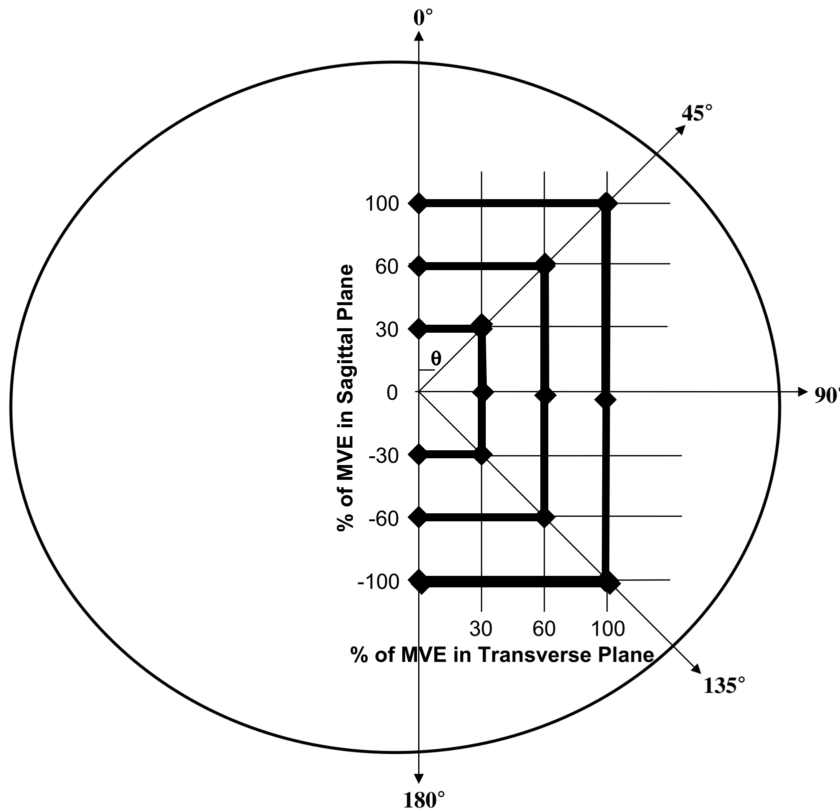

Fig. 2. Schematic diagram of test conditions used in this experiment. The test conditions (♦) are shown for three levels of exertion (shown by thick black line) and five angles of net resultant moment (θ). The 0° indicates the trunk isometric exertion in sagittal flexion exertion, 180° indicates trunk extension, 90° indicated right axial exertion and 45° and 135° indicate biaxial exertions: right axial exertion and flexion, and right axial exertion and extension, respectively.

3. Results

3.1. EMG activity of trunk muscles

The result of the repeated MANOVA indicated significant effect of ANRM and level of exertion and interaction effect on the mean of NRMS-EMG of the 10 trunk muscles (Table 1).

Table 1
The outcome, F and (P value), of multivariate and univariate analysis of variance of mean NRMS-EMG of 10 trunk muscles for the main effects of angle (five levels), exertion (three levels), and their interaction effect

NRMS-EMG	Independent variables		
	Exertion	Angle	Exertion * angle
MANOVA			
All muscles	26 (0.001)	28 (0.001)	3 (0.001)
ANOVA			
Right EOB	171 (0.0001)	79 (0.0001)	4 (0.0002)
Left EOB	209 (0.0001)	101 (0.0001)	5 (0.0001)
Right IOB	146 (0.0001)	48 (0.0001)	2 (0.003)
Left IOB	110 (0.0001)	23 (0.0001)	3 (0.012)
Right RA	125 (0.0001)	156 (0.0001)	13 (0.0001)
Left RA	246 (0.0001)	325 (0.0001)	21 (0.0001)
Right ES	79 (0.0001)	139 (0.0001)	3 (0.011)
Left ES	148 (0.0001)	274 (0.0001)	5 (0.001)
Right LAT	181 (0.0001)	60 (0.0001)	4 (0.004)
Left LAT	67 (0.0001)	23 (0.0001)	3 (0.001)

The results of ANOVA demonstrate that the trunk muscles are significantly affected by the magnitude of exertion ($P < 0.001$), angle of exertion ($P < 0.001$), and their interaction ($P < 0.01$). Mean trunk muscle activities are depicted in Fig. 3 for the different exertion levels and angle of normalized resultant moment.

All muscles showed a higher level of activity during 100% MVE followed by 60% and 30% MVE. The result of the post-hoc Tukey test indicated all three levels of exertion are significantly ($P < 0.001$) different from each other. Fig. 3 shows that all the abdominal, muscles demonstrated significantly higher activity during uniaxial and biaxial flexion (0°, 45°), compared to uniaxial and biaxial extension (135°, 180°).

The left EOB and right IOB had similar patterns of activity. Both muscles had their highest activity during biaxial flexion and right transverse exertion (45°). This activity was significantly ($P < 0.03$) different than uniaxial exertion (0°, 90°) (see Fig. 3, Supplementary material, Table S1). The right EOB and left IOB had a similar pattern of activation.

Both right and left ES showed the higher EMG activity during biaxial extension and right rotation (135°), followed

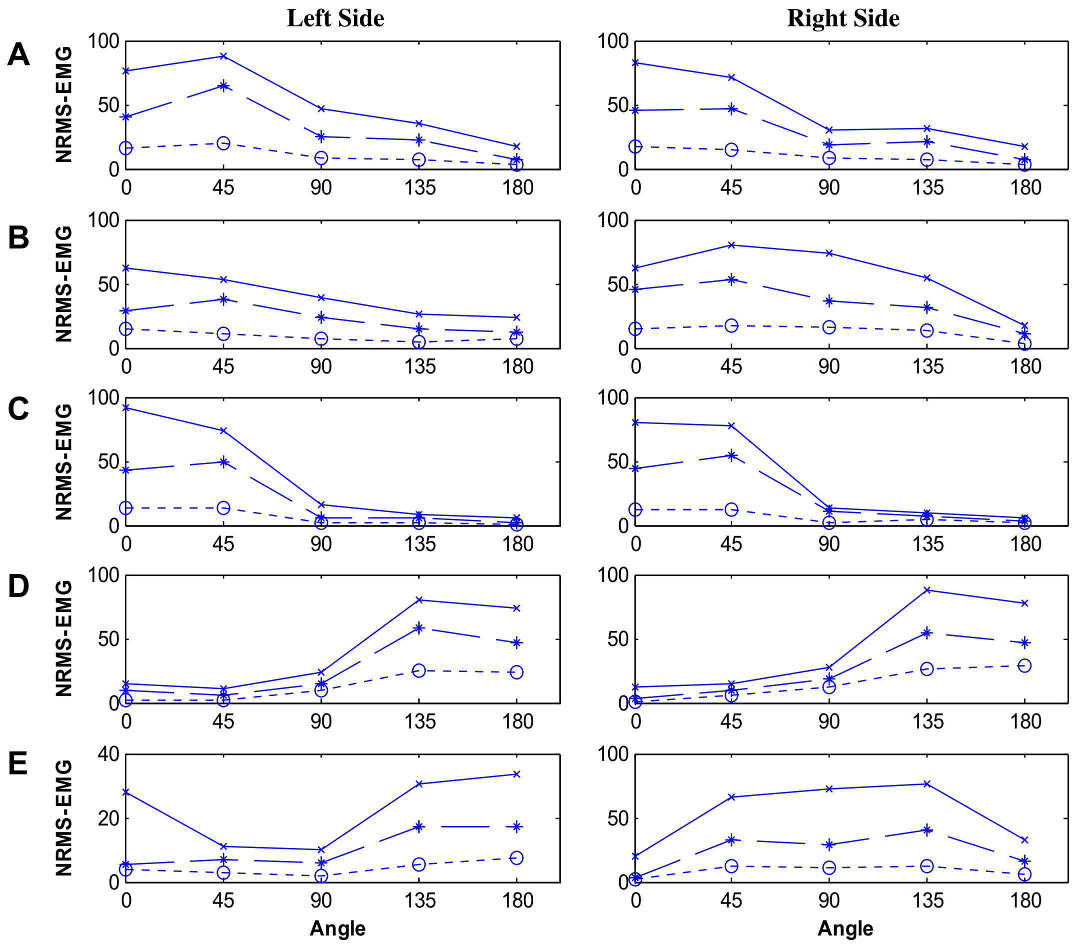

Fig. 3. Mean NRMS-EMG (Y-axis) of left and right external oblique (A), internal oblique (B), rectus abdominis (C), erector spinae (D), latissmus dorsi (E), muscle for different angle of net moments (X-axis) and three level of exertion, 30% (○), 60% (∗), 100% (X) ($n = 15$).

by uniaxial extension (180°), and rotation (90°). The results of the Tukey test suggest that the differences between uniaxial extension and biaxial extension and right rotation were not statistically significant.

The right LAT muscle had the highest activity during biaxial right rotation and extension (135°), followed by uniaxial right rotation (90°), and biaxial right rotation and flexion (45°). The EMG activity of right LAT muscle for three angles of exertion (45°, 90°, 135°) was not significantly different from each other, however they were all significantly different than the uniaxial extension (0°, 180°), (see Supplementary material, Table S1).

3.2. Uniaxial and biaxial exertions

The mean and standard deviation ($n = 15$) of the trunk muscle torques produced with respect to each plane of exertion are presented in Table 2. The mean (SD) of muscle torques produced during the 100% exertion was 120.0 (37) N m for uniaxial flexion, 121.6 (43) N m for uniaxial extension, and 53.7 (19) N m for uniaxial right axial exertion.

The capability of the torque generation was reduced during the 100% MVE biaxial exertion (Table 2). The higher capability index represents a larger discrepancy between the actual exerted torque and the expected target torque. The result of the statistical analysis reveals that the value of the capability index was significantly ($P < 0.0001$) affected by the level of exertion (Table 3). The result of the Tukey test suggests that the value of this index during 100% exertion was significantly different than for 30% and 60% exertion. The angle of net resultant moment and its interaction with level of exertion did not affect the capability index significantly.

Poor controllability as demonstrated by high CV was observed during the biaxial 100% MVE testing condition and followed by uniaxial 100% MVE conditions (see Supplementary material, Table S2).

4. Discussion

4.1. Moment generated during biaxial exertion

Previous studies demonstrated a decrease in the maximum trunk strength (Kumar and Narayan, 2006; Marras

Table 2
The means and standard deviations of measured sagittal and transverse trunk moments (N m) for three levels of exertion (30%, 60%, and 100% MVE) and different angles of net resultant moments (ANRM) during testing conditions, ($n = 15$)

ANRM (°)	30% MVE		60% MVE		100% MVE	
	Sagittal	Transverse	Sagittal	Transverse	Sagittal	Transverse
0	37.9 (12)	0.7 (1)	76.4 (24)	0.7 (1)	120.0 (37)	0.3 (3)
45	37.8 (12)	18.0 (7)	0.1 (1)	33.1 (12)	105.4 (33)	42.5 (15)
90	−0.1 (1)	17.9 (7)	−0.7 (2)	35.3 (13)	0.5 (2)	53.7(19)
135	−37.2 (13)	17.8 (7)	−73.6 (26)	34.2 (15)	−110.0 (38)	45.9 (16)
180	−39.0 (13)	0.6 (1)	−73.8 (25)	1.7 (1)	−121.6 (43)	2.6 (4)

Table 3
The means and standard deviations of the capability index (Eq. (1)) for three levels of exertion (30%, 60%, and 100% MVE) and five angles (0°, 45°, 90°, 135°, and 180°) of net resultant moments (ANRM) during testing, ($n = 15$)

ANRM (°)	30% MVE	60% MVE	100% MVE
0	1.25 (1.16)	1.67 (9.12)	8.75 (5.60)
45	0.98 (0.54)	4.10 (2.55)	28.53 (17.69)
90	0.68 (0.41)	1.69 (1.21)	7.41 (7.94)
135	1.13 (0.56)	3.09 (1.48)	22.33 (23.31)
180	1.14 (0.80)	2.65 (2.42)	5.62 (4.39)

and Granata, 1995) and maximum acceptable weight during asymmetrical lifting (Vink et al., 1992). Garg and Badger (1986) reported a 7–22% decrease in the maximum acceptable weight as the angle of asymmetry increased during a lifting task. (Vink et al., 1992) reported a 30% reduction of the trunk extension MVE during an upright twisting posture compared to symmetrical exertion. Axial torque production in the direction of twisting posture declined up to 25%, where as exertion in opposite direction may increased up to 30% (Kumar et al., 2002a).

Other studies demonstrated the trunk limitations in producing torsional MVE (Marras and Granata, 1995; McGill and Hoodless, 1990; Pope et al., 1987) or extension MVE (Kumar and Garand, 1992; Vink et al., 1992) as a function of trunk upright twisting posture. These studies suggested that the reduction of the MVE during asymmetrical exertion is due mainly to the changes in muscle parameters, i.e., changes in the lever arm and centroid of the muscle.

Theoretical simulation of biaxial and triaxial strength models of trunk also predicted that based on uniaxial exertions the strength capacity during triaxial and biaxial tasks would be overestimated (Parnianpour et al., 1997).

The isometric testing protocol of our study demonstrates that other factors beside changes in muscle parameters are limiting the trunk performance during asymmetrical exertions. Compared to uniaxial sagittal exertion, the maximum sagittal torque during biaxial exertions was reduced by 12% and 10% in flexion and extension, respectively. Compared to uniaxial transverse exertion, the biaxial torsional torque was reduced by 15% and 20% when occurring with flexion torque and extension torque, respectively (Table 2).

The capability index was higher during biaxial exertions indicating the discrepancy between the maximum biaxial capability and expected target torque, based on uniaxial strength. The controllability, as presented by coefficient of variation, was high during both biaxial conditions as well as uniaxial rotation.

It is generally believed that complex trunk exertion (asymmetric push, pull, and lifting) is hazardous to the musculoskeletal system. The chance of overexertion is present during a complex trunk exertion even when individual components of demands are below maximal uniaxial trunk strength. In light of the results of this study, all trunk muscles were significantly affected by the level and angle of exertion. A higher level of co-activity seen during biaxial exertions leads to a higher internal loading of the spine, in addition to larger strain and stress in the annulus fibers, facet joints and soft tissues, and which subsequently may increase the risk of injury (Arjmand et al., 2006; Arjmand and Shirazi-Adl, 2006).

4.2. EMG activity of trunk muscles

The statistical analysis suggested that the means of NRMS-EMG of the 10 selected trunk muscles were significantly affected by the magnitude of uniaxial and biaxial trunk exertion. The significant interaction effect suggests that these factors have a combined effect on NRMS-EMG of trunk muscles. Results of other studies (Kumar et al., 2002b; Lavender et al., 1992; Marras and Granata, 1995) in this area support our findings, although no other study utilized identical protocols. Other isometric studies measured torsion (Marras and Granata, 1995) or lateral (Lavender et al., 1992) moments as a function of trunk angles and direction of exertion.

Lavender et al. (1992, 1993) investigated the effect of asymmetrical exertion on symmetrical upright posture. Their study was similar to the present study with respect to quantifying the activity of trunk musculature as a function of the direction and magnitude of an external net resultant moment. However, Lavender and his colleagues defined the net resultant moment as a combination of lateral and sagittal exertion; where as in our study, it is defined as a combination of sagittal and transverse exertion. Furthermore, the net resultant moment in one study was 10–50 N m with increments of 10 N m (Lavender et al., 1992), and 20 and 40 N m in another study (Lavender et al., 1993), which generally consider low level exertion compared to reported trunk MVE (Perez and Nussbaum, 2002). In the present study, the magnitude of torque is defined as combination of 30%, 60%, and 100% MVE along each axis.

In the present study, we attempted to decouple the EMG activity of the trunk muscles with respect to the activity required to generate moments along the sagittal and transverse axis. The rationale behind this approach is based on the anatomical and biomechanical observation of the trunk muscles during asymmetrical trunk exertion. The activation of the trunk muscles produces moments along all three axes. In the case of planar flexion and extension, the RA muscles and ES muscles respectively are considered as primary movers. In both cases, these muscles are primarily aligned within the plane of sagittal exertion. However, axial rotation is achieved by the net resultant of the co-activation of all trunk muscles and no single muscle is defined as the primary mover. No muscle has a primary vector direction to generate trunk axial rotation (Carlsoo, 1961). The cocontraction of trunk muscles will produce moments along three trunk axes. The limitation of number of trunk muscles to fulfill task demand during biaxial exertion raises questions concerning the allocation of muscles to exert high level force in two planes simultaneously.

Among anterior muscles, the means of NRMS-EMG of the right IOB and left EOB muscles were significantly affected by the level and orientation of the net resultant moment and their interaction (Table 1). Independent of the level of exertion, Fig. 3 and the results of the Tukey test (see Supplementary material, Table S1) show significantly higher levels of EMG activity for these muscles during biaxial exertion compared to uniaxial flexion or uniaxial right rotation. During 100% MVE, mean the NRMS-EMG activity of the right IOB muscle was 62.5 (±22)% during uniaxial flexion, 74.3 (±26)% during uniaxial rotation, and 80.6 (±19) during biaxial exertion.

The mean of NRMS-EMG of the right IOB and left EOB muscles during biaxial extension and right rotation was significantly lower than during axial rotation and significantly higher than during maximum extension. This study, as well other studies (Lavender et al., 1992; Lavender et al., 1993), assumed that the trunk performance is symmetrical for right and left rotation. Therefore, the left exertion was not examined in detail in this study and it was not included in the statistical analysis.

In summary, compared to the ES and RA muscles, muscles that are oriented in an oblique direction (i.e., IOB, EOB, LAT), are taxed more during biaxial trunk exertion. All the trunk muscles showed higher activity during biaxial exertion accompanying the plane that muscle is considered as primary movers. Although the amplitude of EMG activity was different for each muscle, the trend of muscle activity changes for the three levels of exertion was very similar as expected due to concept of spatial (directional) tuning of trunk muscle activation (Todorov, 2002; Vasavada et al., 2002).

4.3. Implication of study

Measurement of trunk strength is frequently used in clinical evaluation of LBP patients (Schenk et al., 2006). The results of this study raise some concern with respect to the measurement of mechanical data and EMG data during the MVE. Commonly, the protocols for measuring trunk strength consist of quantification of trunk maximum voluntary exertions and planar exertion along each trunk axis. The torque generated during the MVE is thought to be an indication of the maximal capability of an individual for handling an external load (Ayoub and Mital, 1989). It is questionable how the information derived from uniaxial plane exertions can be used to predict the capabilities of an individual in situations that involve biaxial exertions. The results of the present study indicate that it is incorrect to predict an individual's ability to perform high-level biaxial exertion based on planar trunk exertion performance. An estimation of individual performance, based on a vectorial sum of planar exertion, will significantly over-estimate the capability of performing biaxial trunk exertion. Industrial jobs often require individuals to have high capability and controllability during asymmetric lifts, i.e., lifting heavy objects and placing them at special area on shelf. The results of this study suggest that both capability and controllability of the subjects should be considered for the ergonomic design of these jobs.

To enhance the biomechanical interpretation, the EMG signal of a muscle is often normalized to the range of the highest and lowest activity of the muscles observed in the experiment. Normalization has been commonly used to compare EMG activities of muscles within a subject or between subjects and tasks (Marras and Mirka, 1990). Generally, the highest activity of a muscle is taken from the isometric exertion along each axis of the trunk. The statistical analysis reveals that the NRMS-EMG activity of the LAT, IOB, and EOB muscles is significantly higher during the biaxial than the uniaxial exertion. This implies that the activity of each measured muscle during biaxial exertion is a better estimate of the maximal activity of the muscle. Therefore, biaxial exertion value should be used for the normalization of the muscle activity.

Observation of high EMG activities during a task is generally associated with either higher physical demand or high precision and control required by task. While higher EMG activity of the trunk muscles represents higher force production of the trunk muscles, the index of capability was higher during biaxial exertions. Compared to uniaxial sagittal exertions, during biaxial exertions, flexion and extension torques were reduced by 12% and 10%, respectively. Similarly, the transverse torque was reduced by 15% and 20% when accompanied by flexion and extension torques, respectively. Furthermore, biaxial exertions resulted in higher coefficient of variability which indicates poor controllability.

This study demonstrates better understanding of higher risk of low back injury that has been associated with asymmetrical exertion that requires biaxial trunk exertion.

Based on present results, it is incorrect to assume a person's ability to perform the biaxial exertion based on uni-planar performances. This study allows link the concept of relative load or utilization ratio (Khalaf et al., 1997; Parnianpour et al., 1997) to the higher risk of low back injury. At a higher utilization ratio, as defined by ratio of the task demand to the maximum muscular strength, an individual is more fatigue prone and more susceptible to the loss of control and coordination. Biaxial exertions lead to a higher utilization ratio than each component of the load would have caused individually. The lower controllability observed even during the short 3 s duration of this study is alarming since the onset of muscle fatigue would only exacerbate the loss of control.

The less available strength resources and lower controllability of trunk muscles would diminish their capability to respond appropriately to any perturbations to the trunk position or load. More research must address these novel perspectives for injury prevention strategies that have more specific motor control implications. The reduction in controllability at high biaxial loads may have implication for risk assessment methods in ergonomics and control training strategy during functional restoration.

The emerging coactivation and stability requirements should be accounted during asymmetric exertion at upright standing posture, since higher stiffness (stability) caused by coactivation has negative consequences on maximum strength and endurance. Maximum strength is provided when the least coactivity is elicited. Hence, controllability strength and endurance will both suffer during asymmetric exertions (Franklin and Granata, 2007).

5. Conclusion

Biaxial trunk exertion tasks represent unique situations, which are different from uniaxial exertion. Controllability index was lower and capability index was higher during biaxial exertion while trunk muscles were taxed more and muscles demonstrated higher EMG activity. Higher EMG activity results in higher spinal forces, higher rate of fatigue and possibly higher risk of injury.

The higher capability and controllability indices during high levels of exertion suggests that uniaxial exertion cannot be based for evaluation of biaxial exertion. Biomechanical and ergonomical job evaluation that rely on uniaxial tasks will underestimate the spinal forces and risk of injury during biaxial tasks. The clinical evaluation of low back pain patients performing biaxial exertion allows better estimation of functional capability and motor deficits. Future study should demonstrate that index of controllability can enhance discriminating power of strength models for identifying those with low back pain.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.clinbiomech.2007.11.017.

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