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The effects of stance width and foot posture on lumbar muscle flexion-relaxation phenomenon

Boyi Hu [a], Xinhai Shan [b,1], Jie Zhou [a], Xiaopeng Ning [a,*]

[a] Department of Industrial and Management Systems Engineering, West Virginia University, Morgantown, WV 26506, USA
[b] College of Physical Education, Shandong Normal University, Jinan, Shandong 250014, China

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ABSTRACT

Background: Characterizing the lumbar muscle flexion-relaxation phenomenon is a clinically relevant approach in understanding the neuromuscular alternations of low back pain patients. Previous studies have indicated that changes in stance posture could directly influence trunk kinematics and potentially change the lumbar tissue synergy. In this study, the effects of stance width and foot posture on the lumbar muscle relaxation responses during trunk flexion were investigated.

Methods: Thirteen volunteers performed trunk flexion using three different stance widths and 'toe-forward' or 'toe-out' foot postures (six conditions in total). Lumbar muscle electromyography was collected from the L3 and L4 level paraspinals; meanwhile three magnetic motion sensors were placed over the S1, T12, and C7 vertebrae to track lumbar and trunk kinematics. The lumbar angle at which muscle activity diminished to a near resting level was recorded. At the systemic level, the boundary where the internal moment started to shift from active to passive tissues was identified.

Findings: For the L3 paraspinals, the flexion relaxation lumbar angle reduced 1.3° with the increase of stance width. When changed from 'toe-forward' to 'toe-out' foot posture, the flexion relaxation lumbar angle reduced 1.4° and 1.1° for the L3 and L4 paraspinals respectively. However, the active and passive lumbar tissue load shifting boundary was not affected.

Interpretation: Findings of this study suggest that changes in stance width and foot posture altered the lumbar tissue load sharing mechanism. Therefore, in a clinical setting, it is critical to maintain consistent stance postures when examining the characteristics of lumbar tissue synergy.

1. Introduction

Low back pain (LBP) continues to be a significant occupational health problem around the world (Dagenais et al., 2008). In the United States, musculoskeletal disorders account for 33% of all workplace injuries and illnesses that require days away from work. From these musculoskeletal disorders, more than 40% are back related injuries (such as LBP) (BLS, 2011). According to a recent report from the Centers for Disease Control and Prevention (CDC), back injury is the second most common reason that causes disability in adults (CDC, 2009). Previous studies have revealed that LBP is responsible for 13% of all working population sick days (Andersson, 1999) and over 90 billion dollars in annual medical expenses (Luo et al., 2003).

Although the exact etiology of LBP is still unclear (Borenstein, 2001), previous studies have shown that LBP could be attributed to genetic (MaxGregor et al., 2004), personal (e.g. obesity, smoking habits) (Richard and Edward, 1989), psychosocial (Gatchel et al., 1995), biomechanical (Kerr et al., 2001) and other (Hoogendoorn et al., 2000) risk factors. Among these risk factors, the magnitude of mechanical loading acting on the spine is highly associated with low back injuries. Direct evidence from in-vitro studies showed that excessive mechanical loading could lead to intervertebral disc rupture (Adams et al., 2000) and vertebra fracture (Brinckmann et al., 1988). In addition, in-vivo studies discovered the existence of a strong association between spinal loading (e.g. compression and shear force) and the prevalence of LBP (Marras et al. (2001a)). Therefore, a comprehensive understanding of spine biomechanics, especially spinal tissue loading, during task performance is critical for the design of appropriate control strategies in mitigating the risk of LBP.

The human lumbar spine is a structure that has a high degree of complexity. In general, lumbar tissues can be divided into two main types: active and passive. Active lumbar tissues refer to the contractive component of muscles. Passive lumbar tissues on the other hand include ligaments, fascia, vertebrae, discs, and all other tissues that do not voluntarily generate force. During trunk motion, active and passive lumbar tissues act in concert to initiate, maintain, or stop trunk motions. Early studies have found that during trunk bending, the lumbar

* Corresponding author at: Industrial and Management Systems Engineering, P.O. Box 6070, West Virginia University, Morgantown, WV 26506, USA.
E-mail addresses: bohu@mix.wvu.edu (B. Hu), xhshan_jn@aliyun.com (X. Shan), jizhou@mix.wvu.edu (J. Zhou), xiaopeng.ning@mail.wvu.edu (X. Ning).
[1] Mailing address: The Biomechanics Laboratory, College of Physical Education, Shandong Normal University, 88 Wenhua East Road, Jinan, Shandong 250014, China.

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extensor muscles will suddenly cease action when reaching close to full trunk bending posture (Floyd and Silver, 1955). This phenomenon illustrated the close interaction between active and passive lumbar tissues and was later referred as the flexion-relaxation phenomenon (FRP). Contemporary literature has recently studied the FRP to enhance our understanding of lumbar tissue neuromuscular behaviors (Olson et al., 2004) and the load sharing mechanism between the active and passive lumbar tissues (Solomonow et al., 2003).

Previous studies have reported that lumbar extensor muscle FRP (and the underlining lumbar tissue synergy) could be affected by a number of factors including the speed and direction of trunk motion (Ning et al., 2011; Sarti et al., 2001), lumbar muscle fatigue (Descarreaux et al., 2008), and ligament creep (Shin et al., 2009). It was also discovered that the increase of knee flexion could reduce tension on the lumbar posterior tissues and result in a delay of lumbar extensor muscle FRP (Shin et al., 2004). Further, existing literature has demonstrated that adopting different lower extremity postures could significantly change the lumbar biomechanical responses during lifting tasks. More specifically, one study observed smaller lumbar external loading when lifting with increased stance width (Cholewicki et al., 1991). A more recent investigation found that the increase of stance width significantly reduced trunk range of motion and sagittal acceleration during lifting (Sorensen et al., 2011); such changes could consequently reduce lumbar external loading (Marras and Granata, 1997) which have been confirmed with previous findings. To reach equilibrium, the reduced lumbar external loading may further lead to smaller lumbar tissue loading and changes to the associated lumbar tissue synergy. Studies have also discovered that maintaining an outwardly rotated foot posture could change lower extremity muscle activation patterns during deadlifting (Escamilla et al., 2000, 2001, 2002) and result in a smaller knee joint internal rotational moment during squat exercises (Almosnino et al., 2013). Although the existing evidence has demonstrated that the changes in posture of the lower extremities could alter trunk and lower extremity biomechanics during task performance, it is still unclear how changes in stance posture affect the lumbar muscle FRP.

FRP illustrates the electromyographic (EMG) silence of a local muscle during trunk flexion motion. However, it is insufficient in describing the systematic behavior of the lumbar tissue synergy. To achieve a more comprehensive understanding of the lumbar tissue load sharing mechanism, a recently defined global (systemic) variable: active region boundary (ARB), warrants further investigation. ARB describes the systematic shift of internal loading from the active, contractive component of the lumbar muscles to the passive, elastic lumbar tissues (Ning et al., 2012). To identify the ARB, the L5/S1 joint external loading will be compared to the internal active moment generated by muscle contraction; the point at which the active moment starts to drastically decrease will be identified as the ARB (described in detail in 'Methods'). The estimation of the internal active moment utilizes an existing lumbar biomechanical modeling approach which uses anthropometric measurements and muscle EMG as model inputs (Ning et al., 2012). Such modeling approaches have demonstrated relatively high accuracy and reliability in estimating trunk muscle forces and the corresponding internal active moments (Granata and Marras, 1993, 1995).

In clinical settings, the accuracy and effectiveness of LBP diagnosis and evaluation is the key in treating this common condition (Marras et al., 1993). Recently, a number of studies have used the absence or alteration of lumbar muscle FRP to differentiate between asymptomatic and LBP patients (Neblett et al., 2003, 2010; Watson et al., 1997). FRP has also been recognized as an indicator of lumbar neuromuscular alterations caused by LBP (Alschuler et al., 2009; Shirado et al., 1995). The ARB can also be used as a complement to the lumbar muscle FRP in the diagnosis and assessment of LBP.

The objective of the current study was to investigate the changes in lumbar active and passive tissue synergy during trunk bending when maintaining different stance postures. More specifically, this study investigated the effect of stance width and outward foot rotation on lumbar extensor muscle FRP and the global lumbar boundary condition ARB during trunk bending motion. Based on the results of previous studies (Cholewicki et al., 1991; Escamilla et al., 2000; Sorensen et al., 2011), it was hypothesized that the increase of stance width and outward foot rotation would reduce lumbosacral joint external loading and cause FRP to occur earlier on lumbar extensor muscles. The global condition ARB is also expected to occur earlier with the increase of stance width and outward foot rotation.

2. Methods

2.1. Participants

Thirteen male volunteers from the student population and nearby residents of West Virginia University participated in this study with informed consent. Their average age, body weight and height were 25.5 years (SD 2.7), 172.8 cm (SD 5.0) and 73.8 kg (SD 6.9), respectively. Participants with chronic or current back, upper/lower extremity disorders or pain were excluded. The research protocol was approved by the Institutional Review Board of West Virginia University.

2.2. Instrumentation

Muscle activities were collected using bipolar surface EMG electrodes (Bagnoli, Delsys, Boston, MA, USA). Eight bipolar electrodes were placed over the skin of the left and right L3 paraspinals (4 cm lateral from the L3 spinous process), L4 paraspinals (2 cm lateral from the L4 spinous process), rectus abdominus (1 cm above and 2 cm lateral from the umbilicus) and external oblique (15 cm lateral from the umbilicus) (Ning et al., 2011). Lumbar and trunk kinematics were collected using a magnetic field-based motion tracking system (Motion Star, Ascension, Burlington, VT, USA); three motion sensors were secured to the skin over the spinous processes of the C7, T12, and S1 vertebrae. The EMG and kinematic data were synchronized using the MotionMonitor software (MotionMonitor, Innovative Sports Training, Chicago, IL, USA) with a sampling frequency of 1024 Hz. A dynamometer and a trunk flexion-extension attachment (HUMAC Norm, Computer Medicine, Stoughton, MA, USA) were used to provide static trunk resistance and lower extremity restriction during the maximum voluntary contraction (MVC) trials (described in detail in the 'Experimental protocol' section).

2.3. Independent variables

Two independent variables were involved in the current study: stance width (WIDTH) and foot posture (POSTURE). Based on the previous literature (Sorensen et al., 2011) three WIDTH levels were selected: narrow (feet together), moderate (shoulder width), and wide (150% shoulder width). POSTURE had two levels: toe-forward (0° between feet) and toe-out (60° between feet). The combination of the two independent variables generated six different conditions (Fig. 1): narrow toe-forward (NF), narrow toe-out (NO), moderate toe-forward (MF), moderate toe-out (MO), wide toe-forward (WF), and wide toe-out (WO).

2.4. Dependent variables

Four dependent variables were investigated. 1.) Maximum lumbar flexion angle (Max-L). The lumbar flexion angle was defined as the angular difference between the T12 and S1 motion sensors in the sagittal plane. 2.) Lumbar flexion angle at ARB (LARB). The ARB was identified using criteria complying with existing literature (Ning et al., 2012). 3.) L3 paraspinals EMG-off lumbar angle (L3L) and 4.) L4 paraspinals EMG-off lumbar angle (L4L). L3L and L4L were defined as the corresponding lumbar flexion angles when FRP occurred (i.e. onset of EMG silence) on L3 paraspinals and L4 paraspinals respectively during