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

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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

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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

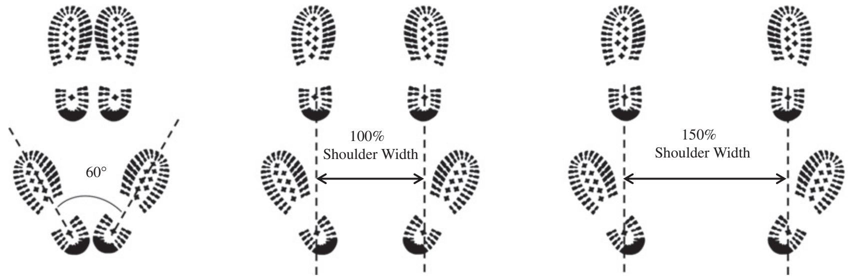

Fig. 1. Demonstration of the six different stance conditions.

trunk flexion motion. In the current study, because subjects performed sagittally symmetric trunk motions, the EMG-off angles from both sides of the same muscle were averaged.

2.5. Experimental protocol

Prior to data collection, participants' anthropometric data (including weight, height, trunk length, and trunk width) were recorded and informed consents were signed. A training session was conducted to allow participants to familiarize themselves with the experimental tasks. Surface EMG electrodes and motion sensors were then placed at the designed sites described above using double-sided tape. Participants began the experiment by performing three trunk extension maximum voluntary contraction (MVC) exertions. In each MVC trial, participants were required to stand in the back flexion–extension attachment with their pelvis and lower extremities fully secured and instructed to perform an isometric maximum trunk extension exertion in a 20° trunk forward flexion posture. Next, participants performed a total of 18 (3 WIDTH $\times$ 2 POSTURE $\times$ 3 repetitions) pace-controlled trunk flexion motions with the presentation of the experimental conditions randomly assigned. In each trial, participants were required to use seven seconds to complete a smooth trunk flexion motion (from upright standing posture to full flexion posture) while maintaining a straight knee and preassigned foot posture (Fig. 2(a)). A metronome was used to help maintain the rhythm. At least two minutes of rest was given between trials to prevent the accumulation of muscle fatigue.

2.6. Biomechanical model

A previously established EMG-assisted lumbar biomechanical model was used to calculate the internal active (muscle generated) moment (Ning et al., 2012). This model includes four trunk muscles: L3 paraspinals, L4 paraspinals, rectus abdominis, and external oblique. Normalized EMG signals from all four trunk muscles were used to estimate instantaneous muscle forces with the consideration of length-tension and force-velocity factors (Davis et al., 1998; Marras and Granata, 1997). The physiological cross-sectional areas and moment arms (to the center of the L5/S1 joint) of the muscles were estimated

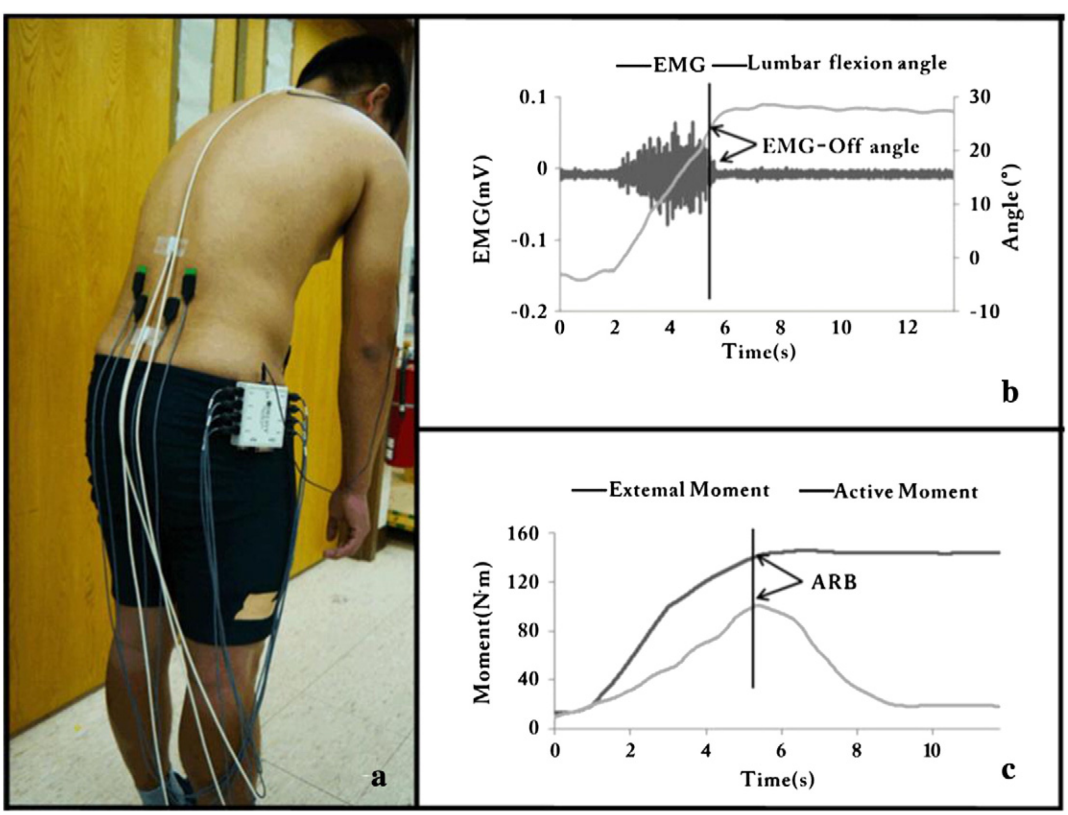

Fig. 2. (a) A snapshot during trunk flexion motion; (b) Demonstration of the "EMG-Off" point of a back muscle and the corresponding lumbar flexion angle; (c) ARB is identified as the location beyond which internal active moment quickly diminishes.

using previously reported regression equations (Jorgensen et al., 2001; Marras et al., 2001b). External moments were estimated using a simple physical model which considers trunk flexion angle (calculated using the position data of C7 and S1 motion sensors) (Ning et al., 2012), instantaneous acceleration (both linear and angular), upper body mass, and center of mass, which were obtained from previous literature (Mirka et al., 1998).

2.7. Data processing

A customized program (Matlab 2010, MathWorks, Natick, MA, USA) was developed to process EMG and kinematics data. Raw EMG data were first transferred into frequency domain and then passed through a 500 Hz low-pass filter, a 10 Hz high-pass filter, and a notch filter at 60 Hz and its aliases. The EMG data were then transferred back to time domain and fully rectified; a moving window of 512 points (0.5 s duration) was used to generate a standard deviation (SD) profile for each low back extensor muscle. The SD profile was compared with the full flexion SD and the last point during the flexion motion segment at which the SD was 3 times larger than that of full the flexion SD was selected as the EMG-Off point (Jin et al., 2012; Ning et al., 2011). The corresponding lumbar flexion angle at EMG-Off point was then identified (Fig. 2(b)). The ARB was identified by comparing the external moment and the internal active moment profiles, and was defined as the first point in time where the internal moments reduced to less than 70% of the external moments during trunk flexion (Ning et al., 2012) (Fig. 2(c)).

2.8. Statistical analysis

Multivariate analysis of variance (MANOVA) was first performed to investigate the effects of the independent variables and their interactions on the set of dependent variables. The independent variables that demonstrated significant effects were further analyzed using univariate ANOVA (Montgomery, 2005). Tukey–Kramer *post hoc* analyses were then performed on the significant effects with more than two levels to further explore the differences between the two conditions. A criteria *P*-value of 0.05 was used for all statistical analyses.

3. Results

MANOVA results revealed significant effects of both WIDTH and POSTURE. However, their interaction effect was found not significant and therefore was not further investigated (Table 1). The analysis of muscle EMG data revealed that FRP consistently occurred for all back extensor muscles during trunk flexion motion regardless of what stance postures were used. The increase of WIDTH caused FRP to occur significantly earlier on the L3 paraspinals (i.e. reduced L3L angle). While a similar trend was also observed for the L4 paraspinals, however, such an effect was not statistically significant (Fig. 3). POSTURE significantly affected the onset of FRP for both lumbar extensor muscles; more specifically, FRP occurred earlier on both L3 paraspinals and L4 paraspinals (i.e. reduced L3L and L4L angles) when participants maintained a 'toe-out' foot posture (Table 2). Kinematics data showed that despite the small changes in magnitude, the maximum lumbar flexion angle

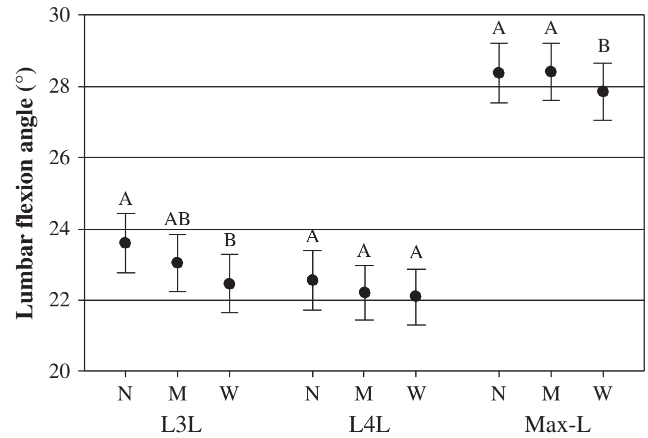

Fig. 3. EMG-Off lumbar angles of L3 and L4 paraspinals and the maximum lumbar flexion angle under different stance width (N: Narrow; M: Moderate; W: Wide). Different letters denote angles that are statistically different from one another. Bars indicate the corresponding standard error.

significantly decreased with the increase of WIDTH and change of POSTURE from 'toe-forward' to 'toe-out' (Fig. 3, Table 2). Finally, the LARB was not significantly affected by any main or interactive effects (Table 1).

To interpret these findings, we conducted further data analyses and discovered that the increase in foot stance width often resulted in smaller external moment during trunk flexion motion (Fig. 4). To support this assertion, we compared the external moment at 85%, 90%, and 95% of maximum lumbar flexion (averaged across all trials) for each subject. These three levels were chosen because our current results and earlier studies showed that lumbar muscle FRP normally occur when the lumbar angle reaches between 85% and 95% of its maximum (Hu et al., 2013; Ning et al., 2011; Olson et al., 2004; Solomonow et al., 2003). Results of this analysis showed that the increase of stance width significantly and consistently reduced the external moment at all three lumbar angle levels (Fig. 5). A similar mechanism was also observed among different foot posture conditions; namely the 'toe-out' foot posture resulted in smaller external moment which caused FRP to occur earlier among the lumbar extensor muscles.

4. Discussion

The objective of the current study was to evaluate the influence of stance width and foot posture on the biomechanical responses of the lumbar spine during trunk bending motion. Based on the results of previous literature, it was hypothesized that the increase of stance width and outward foot rotation would cause lumbar muscle FRP and global condition ARB both to occur earlier. Results of the current study partially supported our initial hypothesis. Specifically, the increase of stance width engendered smaller lumbar EMG-Off angle; namely, the FRP occurred earlier on lumbar extensor muscles during trunk flexion (Fig. 3).

FRP is determined by the interplay of active and passive lumbar tissues in response to the external loading. During the trunk bending motion, the internal moment (about the L5/S1 joint) produced by trunk muscle contraction and passive tissue deformation counterbalances the external moment. For each individual, the passive lumbar tissue

Table 1
Results of MANOVA and ANOVA on independent major and interactive effects. Bolded numbers represent results that are statistically significantly different between different conditions.

Independent variables	MANOVA	Dependent variables			
		L3L	L4L	Max-L	LARB
WIDTH	$\mathbf{P < 0.001}$	$\mathbf{P < 0.001}$	$P = 0.30$	$\mathbf{P < 0.001}$	$P = 0.14$
POSTURE	$\mathbf{P < 0.001}$	$\mathbf{P < 0.001}$	$\mathbf{P < 0.001}$	$\mathbf{P = 0.03}$	$P = 0.20$
WIDTH ∗ POSTURE	$P = 0.36$	$P = 0.09$	$P = 0.78$	$P = 0.11$	$P = 0.74$

Table 2
Mean values of dependent variables for different foot postures. Bolded numbers represent results that are statistically significantly different between different conditions.

POSTURE	Dependent variables			
	L3L	L4L	Max-L	LARB
Toe-forward	**23.7**	**22.8**	**28.4**	11.9
Toe-out	**22.3**	**21.7**	**28.1**	11.1

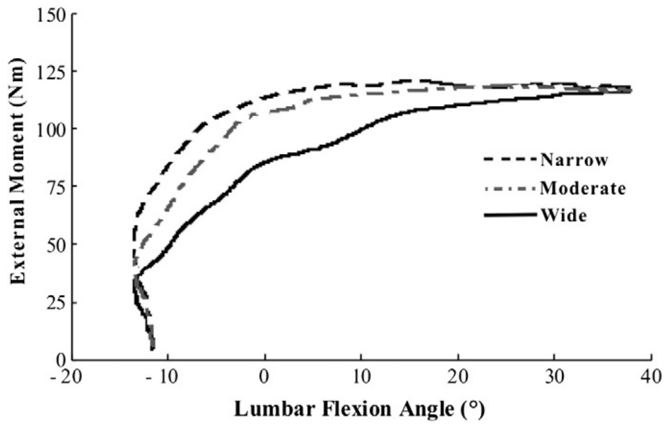

Fig. 4. The L5/S1 joint external moment profiles of a typical subject when performing trunk bending motion with different stance width.

deformation and the associated passive tissue forces are largely determined by the magnitude of lumbar flexion (Jia et al., 2011; McGill and Norman, 1986). As demonstrated in Figs. 4 and 5, a reduction in the total external loading would in turn reduce internal active moment and cause the lumbar muscles to cease activity earlier (i.e. reduced lumbar EMG-Off angles).

Results of this study also indicated that the increase of stance width generated larger changes on the L3 paraspinals compared to the L4 paraspinals. The lumbar EMG-Off angle of the L3 paraspinals decreased 1.3° from 'narrow' to 'wide' stance width, while the L4 paraspinals lumbar EMG-Off angle only reduced 0.4°. The L4 paraspinals was also less affected by foot posture; the reduction of the EMG-Off lumbar angle from 'toe-forward' to 'toe-out' foot posture was 1.4 and 1.1° for the L3 paraspinals and L4 paraspinals respectively. One potential explanation for this phenomenon may come from the different functions that these two muscles serve in the lumbar region. The sampling locations of the EMG electrodes suggest that signals of the L3 paraspinals are mainly gathered from the longissimus and signals of the L4 paraspinals are primarily generated by muscles that are close to the spinous process such as the multifidus. According to previous investigations, the longissimus muscle mainly serves as a global muscle that initiates trunk motion (Bergmark, 1989); whereas the primary function of the multifidus is to maintain lumbar stability and stiffness (Ward et al., 2009). In the current experiment, although external moment varied between conditions, the L4 paraspinals could have been activated for a longer period of time in order to maintain lumbar stability and was therefore less affected by the changes of external loading.

Although the active and passive tissue interaction was altered at the individual muscle level, at the system level, the location of the ARB was not significantly affected by any main or interactive effects, which differed from our initial hypothesis. According to its original definition, the ARB recognizes the posture beyond which the internal active moment drastically decreases and the passive lumbar tissues become the dominant load bearer (Ning et al., 2012). The current results suggest that although changes in lower extremity posture altered the activation pattern (i.e. onset of EMG silence) of individual lumbar muscles, the systematic loading sharing mechanism between active and passive lumbar tissues is much less affected.

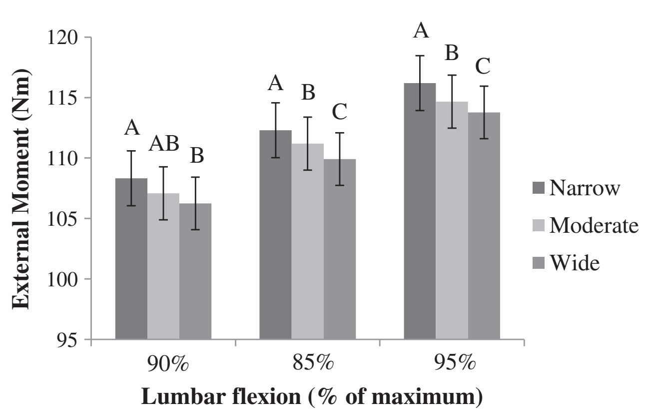

Fig. 5. External moment at 85%, 90%, and 95% of maximum lumbar flexion angle under different stance widths. Different letters denote angles that are statistically different from one another. Bars indicate the corresponding standard error.

The current research has several limitations that need to be noted. First, to eliminate the possibility of unwanted muscle fatigue, no hand load was used in the current design. Future study should investigate the potential interactive effect between hand load and stance postures on the active and passive lumbar tissue interaction during trunk bending. Second, the biomechanical differences (e.g. lumbar EMG-Off angles) revealed in the current study were often significant but relatively small in magnitude. This could be due to the relatively conservative independent variable conditions that we tested (e.g. maximum foot rotation was 60°). It is possible that more distinct foot postures or wider stance widths may generate larger differences; especially when a hand load is added. Third, to standardize the experimental protocol and to eliminate the effect of motion speed on FRP (Sarti et al., 2001), a controlled rhythm was used and the potential interaction between stance posture and motion speed was not explored.

5. Conclusion

In conclusion, during trunk bending, the increase of stance width and outward foot rotation (i.e. the change from 'toe-forward' to 'toe-out' foot posture) will cause the lumbar extensor muscles to cease activity earlier. However, the global boundary condition (i.e. ARB) of active and passive lumbar tissue interaction will remain unchanged. For the clinical setting, our results suggest that it is necessary to maintain constant lower extremity posture during the assessment of the lumbar muscles and motion segments; especially when lumbar muscle FRP is used to identify LBP symptoms, examine the progress of recovery, or differentiate LBP patients from healthy individuals. On the other hand, when examining the global condition ARB restrictions on lower extremity postures can be less rigorous.

Conflict of interest statement

The authors of this manuscript submitted for possible publication in the scientific journal *Clinical Biomechanics* do hereby state that none of us have any conflicts of interest to disclose with regard to any financial and personal relationships with other people or organizations that could inappropriately influence this work.

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