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A systems-level perspective of the biomechanics of the trunk flexion-extension movement: Part I – Normal low back condition

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ABSTRACT

Most of the previous studies of the lumbar region have not considered the influence of pelvic and lower extremity characteristics on the performance of the lumbar region. The goal of the current study was to explore these more systems-level effects by assessing the effects of a pelvic/lower extremity constraint on the biomechanical response of the lumbar spine in an *in-vivo* experiment. Twelve participants performed full range of motion, sagittal-plane trunk flexion-extension movements under two conditions: unconstrained stoop movement and pelvic/lower extremity constrained stoop movement (six repetitions in each condition over three days). Kinematics and muscle activities of the trunk and lower extremity muscles were monitored. Results showed a significant effect of pelvic/lower-extremity constraint on a number of lumbar performance measures. Trunk flexion angle was, as expected, significantly reduced with the lower extremity constraints (81 degrees (free stoop) vs. 56 degrees (lower extremity constrained)). At a more local level, there was a 6.4% greater peak lumbar flexion angle and a 9.1% increase in the lumbar angle at which the trunk extensor musculature demonstrated flexion-relaxation in the constrained stooping condition as compared to the unconstrained stooping condition. Also, the EMG of the L3/L4 paraspinals was greater in the restricted stooping as compared to the free stooping (16.3% MVC vs. 15.1% MVC).

***Relevance to Industry*:** Low back injuries are a significant challenge to many industries and developing accurate models of spinal stress at full stooping postures can help in the development of appropriate interventions to reduce prevalence.

Highlights

- We examined the role of pelvis and lower extremity during trunk flexion-extension.
- Greater lumbar flexion was observed in the lower extremity constrained condition.
- Later flexion-relaxation was observed in the lower extremity constrained condition.
- Greater EMG in low back was observed in the lower extremity constraints condition.
- A complementary interaction between trunk and lower extremity exists.

Key words: Lumbar spine; Lower extremity; Low back stability; Flexion relaxation phenomenon

1. INTRODUCTION

Standard anatomic classifications of body regions can be misleading regarding the functional biomechanical interactions between adjacent regions of the body. The existing spine biomechanics literature, for example, has provided an excellent understanding of the function of the spine as an independent unit, but a more systems-level characterization (e.g. consideration given to lower extremity influences) may provide deeper insights into its function in more realistic whole body activities. In many models and experimental studies the pelvis is regarded as a rigid, stable body on which the lumbar spine functions (e.g. Bergmark, 1989; Cholewicki and McGill; 1996, Granata and Rogers, 2007; Mirka and Marras, 1993). It is widely recognized that in real world lifting scenarios, the pelvis is not rigid or fixed but is influenced by the lower extremities and therefore documenting and quantifying these effects are important next steps in both modeling and experimental studies.

The potential influence of lower extremity structures (bones, muscles, passive tissues) on lumbar mechanics is considerable. A number of lumbar and lower extremity muscles are indirectly connected through their common insertions in the pelvis. As activation levels increase, the resulting motion of the pelvis can impact the length-tension relationship of other muscles in other regions. Many lumbar muscles originate on the ilium or sacrum (iliocostalis lumborum, quadratus lumborum, multifidus) and a number of lower extremity muscles originate on various locations on the ilium and ischium (gluteus maximus, biceps femoris, semitendinosus and semimembranosus). These posterior compartment thigh muscles span both the hip and the knee and are known to influence lumbar-pelvis interaction (i.e., lumbopelvic rhythm) and pelvis-femur interaction (i.e., pelvifemoral rhythm) (Sihvonen, 1997). The activation of the lower extremity muscles, therefore, can influence pelvic posture and thereby impact length of the low back muscles – affecting both their active tension capability as well as their passive tension. These effects have implications for spine loading and spinal stability.

Other studies have demonstrated these inter-region biomechanical effects through interactions of active and passive tissues. Several studies have revealed that the sacrotuberous ligaments can stabilize the sacroiliac (SI) joint during nutation of the sacrum via the activation of the biceps femoris and gluteus maximus muscles (Vleeming, et al., 1989a; Vleeming, et al., 1989b; van Wingerden, et al., 1993). In contrast, the sacroiliac ligaments can stabilize the SI joint during counter-nutation of the sacrum via activation of the erector spinae muscles, and the tension of the ligament decreases during activation of the gluteus maximus and traction of lumbodorsal fascia (Vleeming et al., 1996). The results suggest that there is a complementary interaction between trunk and lower extremity to achieve the stable foundation of the sacrum-ilium system. Recently, van Wingerden et al. (2004) demonstrated that SI joint stability increases with even slight activation of the erector spinae, the gluteus maximus and the biceps femoris muscles. In addition, Vleeming et al. (1995) showed the functional role of the lumbodorsal fascia in load transfer between spine, pelvis, and lower extremity by dissection in ten embalmed human cadavers and traction to various muscles such as gluteus maximus, external oblique, latissimus dorsi and biceps femoris. Through the lumbodorsal fascia these muscles may play an important role in stabilization of the trunk motion system during trunk flexion, trunk extension and trunk rotation. Pool-Goudzwaard et al. (1998) demonstrated through a biomechanical model that the lumbodorsal fascia can transmit force from the lower extremity to the trunk. In summary, this fascia creates a strong link between the trunk (i.e., spinal column) and lower extremity (i.e., pelvis) by bracing the lumbar spine and SI joints, and enhances the trunk-system level stability achieved by both pelvic stabilization and spinal stabilization.

The goal of the current study was to investigate the biomechanical interactions between the lumbar region of the spine and the pelvis/lower extremities during full range of motion, sagittal plane trunk flexion-extension movements. These are explored by documenting the impact of pelvic/lower extremity constraints on lumbar and lower extremity muscle activation profiles and lumbar and trunk kinematics. It is hypothesized that constraining the thighs and pelvis will significantly affect lumbar kinematics and muscle activations through changes in the passive tissue contributions to stability and total trunk extension moment.

2. METHODS

2.1. Participants

Twelve male participants were recruited from the Iowa State University with average age 28.3 (SD 4.7) years, height 175.9 (SD 2.7) cm, and weight 73.5 (SD 6.6) kg. Participants were screened by questionnaire for chronic problems or current pain in the low back or lower extremities before experiment. Each participant provided written informed consent prior to participation, using a form approved by the institutional review board (IRB) at Iowa State University.

2.2. Apparatus

A lumbar dynamometer (Marras and Mirka, 1989) was used to provide the static resistance necessary to perform maximum voluntary contractions (MVCs) (both trunk flexion and extension). Surface electromyography was used to capture the activities of the twelve sampled muscles including right and left pairs of: L4 paraspinals (2 cm lateral from L4 spinous process), L3 paraspinals (4 cm lateral from L3 spinous process), rectus abdominis, external oblique, gluteus maximus and biceps femoris (Model DE-2.1, Bagnoli™, Delsys, Boston, MA) (data collected at 1024 Hz). A magnetic field-based motion analysis system was used to capture the instantaneous trunk motions (Ascension Technology Corporation, Shelburne, VT; The MotionMonitor™, Innovative Sports Training, Chicago, IL) (data collected at 102.4Hz). Four magnetic motion sensors were placed over the S1, T12, C7 vertebrae as well as one over the xiphoid process. The pitch angle of each of these sensors captured the angle in the sagittal plane. An electrical metronome was used to maintain a constant pace for trunk flexion and extension.

The platform on which the participants stood during the experimental trials could be set up for the two different experimental conditions. In the free stooping condition the participants were free standing on the platform during the trunk flexion-extension motions – knees were locked straight, but there were not any external restrictions on the pelvis or the thighs. In the restricted stooped condition the participants' legs and pelvis were secured to a stable structure (the same vertical structure that was used to secure the pelvis during the MVC exertions) thereby maintaining verticality of the lower extremity (Figure 1). The straps used to secure the thighs were cinched tightly across the midthigh level. The strap at the waist level was likewise cinched tightly, but was not a "clamp" that eliminated any pelvic rotation (Figure 2).

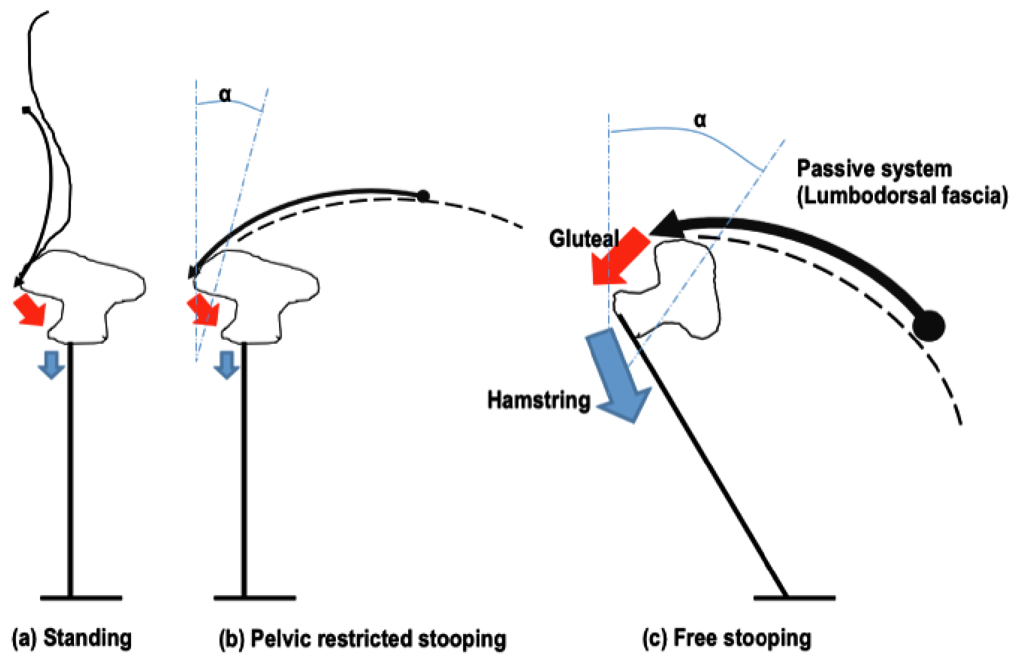

Figure 1. Representation of the difference in postures assumed during the restricted and free stooping conditions.

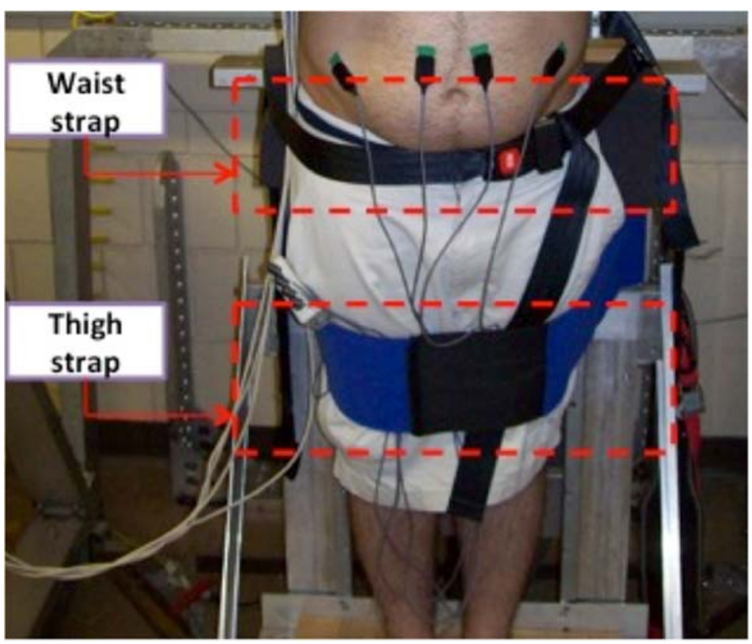

Figure 2. Lower extremity constraint in the lumbar dynamometer

2.3. Experimental design

There was one independent variable, POSTURE, with two levels: free stooping and restricted stooping. There were six kinematic dependent variables in this study: 1) peak hip flexion angle (pitch angle from the S1 sensor), 2) peak trunk flexion angle (pitch angle from the xipoid process sensor), 3) peak lumbar flexion angle (difference between the pitch angles from the T12 and S1 sensors), 4) peak lumbothoracic flexion angle (difference between the pitch angles from the xiphoid process and S1 sensors), 5) the EMG-Off lumbar angle for the L3 paraspinals, and 6) the EMG-Off lumbar angle for the L4 paraspinals. The peak values listed above simply refer to the maximum of that measure seen during a given trial. The EMG-Off lumbar flexion angles are the lumbar flexion angles at which the muscle activity level was indistinguishable from that seen in the full trunk flexion posture (i.e. point of the beginning of flexion-relaxation response). This EMG-Off angle was identified as the first point during the trunk flexion motion at which the normalized EMG (NEMG) signal was reduced to value that was less than three times the full flexion NEMG (flexion-relaxation) value (Jin, Ning, and Mirka, 2012). As these were sagitally-symmetric tasks, these angles from the right and left pairs of each muscle were averaged resulting in an EMG-Off L3 and an EMG-Off L4 value for each trial.

There were three EMG-related dependent variables that are here labeled "Agonist", "Antagonist" and "Lower Extremity". The value for "Agonist" was calculated for each trial as the average of the normalized EMG (NEMG) of four muscles (right and left pairs of the L4 and L3 paraspinals) over a specified "80-20" range of trunk flexion angles (80-20 range defined below). The value of "Antagonist" was calculated for each trial as the average of the NEMG of four muscles (right and left pairs of the rectus abdominis and external obliques) in the same 80-20 range of trunk flexion angles. The value for "Lower Extremity" was calculated for each trial as the average NEMG of four muscles (right and left pairs of the gluteus maximus and biceps femoris) in the same 80-20 range of trunk flexion angles. In the experimental trials the specified range of lumbar flexion angles in which the rectified signals were averaged (herein called the 80-20 range) began during the flexion motion as the participant reached 80% of the full lumbar flexion angle and continued through the full flexion posture and then ended as the participant passed through that same angle (20% of extension motion) during the returning extension motion.

2.4. Task and procedure

Upon arrival the experimental procedures were described to the participant and written informed consent was obtained. Participant was fitted with motion and EMG sensors and then performed the MVC exertions. Participants performed two repetitions of the isometric trunk flexion and extension MVC exertions in the lumbar dynamometer while assuming a 20 degree trunk flexion posture. MVC exertions for the gluteus maximus and biceps femoris were performed against manual resistance provided by the experimenter while the participant assumed an upright standing posture (two repetitions for each).

Prior to performing the experimental trials, the participants stood in an open space (no restrictions on pelvis or thighs) in an upright comfortable posture and then bent forward to a full trunk flexion posture. These baseline data defined full range of trunk flexion. In subsequent experimental trials participants were asked to do slow, controlled flexion and extension trunk motions consisting of two free stooping trials and two restricted stooping trials. Each of these trials consisted of a 5 second flexion motion (to full flexion), 4 seconds of holding at full flexion and then 5 seconds to extend back to upright posture in time with a metronome sound (one beat per second). The order of the free stooping vs. restricted stooping sequences was randomized across participants. This procedure was repeated on three separate days as part of the preliminary data collection of the companion paper (Jin and Mirka, 2015). Data collected over the three days resulted in six repetitions per participant per condition.

2.5. Data processing

Kinematic Variables. The "peak" sagittal plane angles were simply the greatest values seen during the full flexion-extension motion for each trial. The EMG-Off variables, on the other hand, were expressed as a percent of the lumbar full range of motion. The percentage of range of flexion was calculated using two calibration data points: the lumbar angle (T12-S1) in upright standing and the lumbar angle in the full flexion postures in the pre-experimental trials (Equation 1) (Dolan et al., 1994).

$$\text{Percentage Flexion (\%)} = \frac{[LF - LF_{standing}]}{[LF_{fullflexion} - LF_{standing}]} \times 100 \qquad (1)$$

EMG Variables. The unprocessed EMG data were filtered (high-pass 10 Hz, low-pass 500 Hz and notch filtered at 60 Hz and 102.4 Hz and their aliases) and full wave rectified. For the MVC exertions, the data were averaged over 1/8 second windows. The maximum 1/8 second window was identified for each muscle group and was used as the denominator in order to normalize the EMG data during lifting tasks. All rectified EMG data were averaged in this window and then normalized with the muscle-specific maximum value. It is important to note that these muscle specific maximum values were collected in the 20 degree trunk flexion posture while the task EMG data were collected in a more fully flexed trunk posture. While it would have been ideal to have collected the MVC EMG data in this full flexion posture, participant safety concerns led to this modified normalization approach. Finally, the average normalized EMG values across the four muscles in each group (Agonist, Antagonist, Lower Extremity) were calculated.

2.6. Statistical analysis

All statistical analyses in this study were conducted using SAS® and Minitab®. Prior to model analysis, diagnostic tests were performed on the data, including, test for homoscedasticity (Bartlett's Test and Levene's Test) and normality (Anderson-Darling Normality Test). Dependent variables that violated one or more assumption were transformed so that the ANOVA assumptions were fully satisfied (Montgomery, 2001). ANOVA employing a randomized complete block design (blocking on participant) was used to test the effects of POSTURE on the dependent measures. A *p*-value less than 0.05 was the standard level for significance.

3. RESULTS

3.1. Trunk kinematics

The ANOVA revealed statistically significant effects of POSTURE on all six kinematic variables considered (Table 1). The results showed a significantly higher peak trunk flexion angle and peak hip flexion angle in free stooping than in the restricted stooping condition. In contrast, the other four dependent measures showed a significantly higher peak value in the restricted stooping condition as compared to the free stooping condition (Figure 3). The difference in the response of these two groups of dependent variables reflects the difference between an absolute measure (peak trunk flexion angle (xiphoid process sensor pitch angle) and peak hip flexion angle (S1 pitch angle)) and a relative measure (final four are all sensors that are evaluated relative to the S1 sensor).

Table 1. Results of the ANOVA for the kinematic and EMG measures. (* indicates significance at the $p<0.05$ level)

	Mean Difference (= free – restricted)
Trunk flexion angle	25.2° *
Hip flexion angle	30.0° *
Lumbar flexion angle	-2.3° *
Thoracic flexion angle	-2.7° *
EMG-off angle (L4)	-3.2° *
EMG-off angle (L3)	-2.5° *
NEMG Agonist	-1.2% *
NEMG Antagonist	1.2%
NEMG Lower Extremity	2.3% *

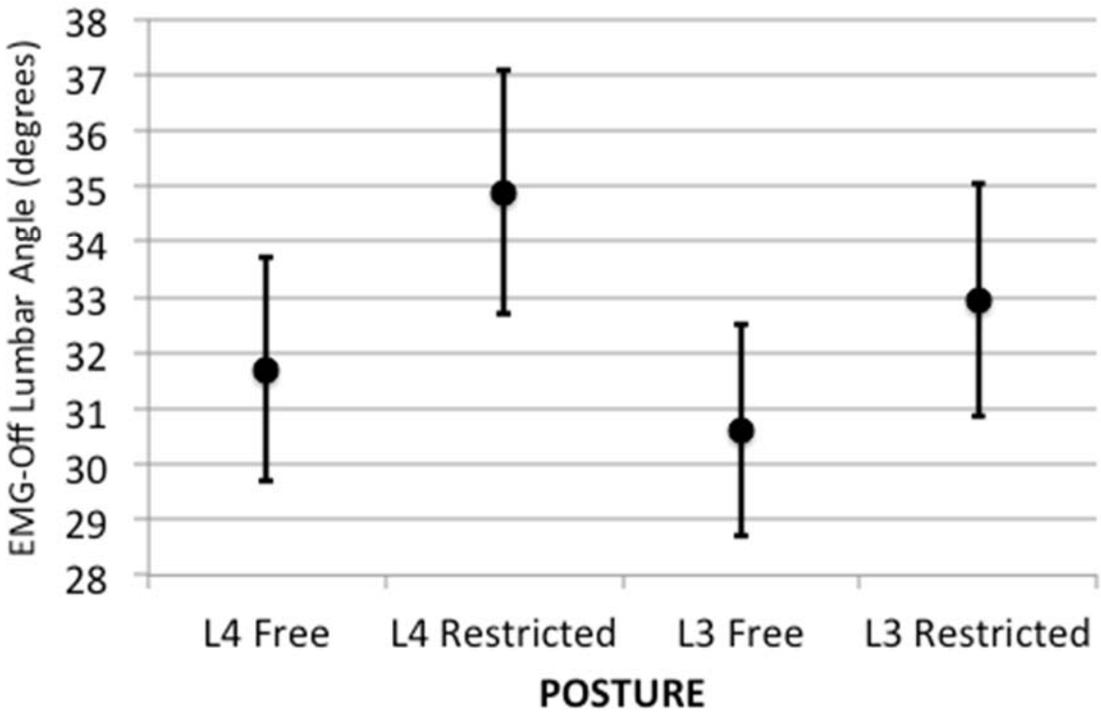

Figure 3. Comparison between two stooping postures in EMG-off lumbar angles (Error bars show 95% confidence interval)

3.2. Muscle activity

The ANOVA of the muscle activity data revealed significantly higher Agonist in the restricted stooping than in free stooping (16.3% MVC vs. 15.1% MVC), and the NEMG of Lower Extremity showed significantly higher NEMG in free stooping as compared to the restricted stooping (10.5% MVC vs. 8.2% MVC). There was no significant difference in the value of Antagonist between the two stooping conditions.

4. DISCUSSION

In 1989 Bergmark developed a model of the lumbar spine that included both a local and a global system (Bergmark, 1989). In this conceptual model the local system included muscles that had their origin or insertion on the vertebrae while the global system included muscles that connected the pelvis to the thoracic cage. The muscles of the local system were said to control the curvature of the spine and provide stiffness to the spine while the muscles of the global system functioned to generate the internal moments necessary to counter the net external moment. In the current paper we have taken this modeling concept one step further by considering a "super-global" system that considers those muscles that influence the biomechanics of the lumbar spine indirectly through their action on the pelvis. The empirical work in this study focused on gathering data from the super-global biomechanical systems to reveal the functional role of these components during a simple trunk flexion-extension movement.

At the highest level, the results of the kinematic analysis of the current study show the significant role of pelvis mobility in trunk flexion-extension with greater trunk flexion and hip flexion in the free stooping technique - an intuitive result which supports previous studies that demonstrated significant pelvic rotation in trunk the free flexion-extension motion (Sihvonen, 1997; Paquet et al., 1994; Sarti et al., 2001). Peak lumbar flexion angle and peak thoracic flexion angle, on the other hand, were significantly greater in the restricted stoop as compared to the free stoop condition. At first glance this may appear counterintuitive, but as one considers the effects of the biceps femoris and gluteus maximus at the full flexion positions during the free stooping technique, one notes that these muscles will be exerting a force on the pelvis, inducing a posterior rotation of the top of the ilium. This in turn induces a passive tissue force in the ligamentous tissues of the lumber region and thereby a restriction on the range of motion on the lumbar and thoracic spine. This interpretation is supported by the work of Olson et al. (2006), which showed that the hamstring muscles are fully stretched around the full flexion and suggested passive pulling tension generated by the muscles on the pelvis.

This view is also supported by the EMG-off angles in L3 and L4 paraspinals that revealed a significantly later (i.e. deeper) initiation lumbar flexion angle of flexion-relaxation in restricted stooping condition as compared to the free stooping condition. In terms of muscle activation levels during these movements, the agonist group showed a greater muscle activation level in the restricted stooping as compared to the free stooping. This result, again, would be consistent with earlier transition to the passive mechanism. By contrast the muscle activity of the lower extremity group was greater in the free stooping condition thereby providing an active tension on the pelvis in these near full flexion postures.

The results of this study suggest a significant role of the super global system as an active stabilizer of the pelvis and a passive stabilizer of low back system. Reeves et al. (2007) used the interesting analogy of a ball on a curved surface to describe the stability of a biomechanical system as well as the robustness of the system to perturbations. They developed this as an analogy of the spine by describing the steepness of the walls of a concave surface as representational of the overall trunk stiffness. To extend this nice analogy to the super global system advocated in the current study, the ball and bowl model must be controlled relative to a larger system (Figure 4). In this new model the bowl in which the ball is resting could describe the pelvis; in the Reeves et al. (2007) model the bottom of the bowl was always flat and stable representing a fixed rigid foundation while the current model allows for instability of this bowl analogy by having a curved underside. In addition there are structures (the cables in the figures) that connect the pelvis to the stable ground and these represent pelvic stabilizers such as ligamentous tissues and muscle of the thigh. When these stabilized are shown as “taut” (Figure 4 (A) & (C)) they act to stabilize the pelvis; however, when they are “slack” (Figure 4 (B) & (D)) the pelvis can move thereby increasing system instability. This new model has the capability of considering the stability provided by the lower extremities, especially pelvic stabilization for stable foundation and SI joint stability. The steepness of bottom of the bowls in the figure could represent contribution of the active tissues such as the erector spinae, the gluteus maximus and the biceps femoris muscles on pelvic stabilization. Deconditioned or fatigued muscles or viscoelastic elongation of ligamentous tissues due to stress-relaxation may influence the baseline stabilization of these tissues and may require more active stabilization from other muscles. This topic is the focus of the Part II companion article.

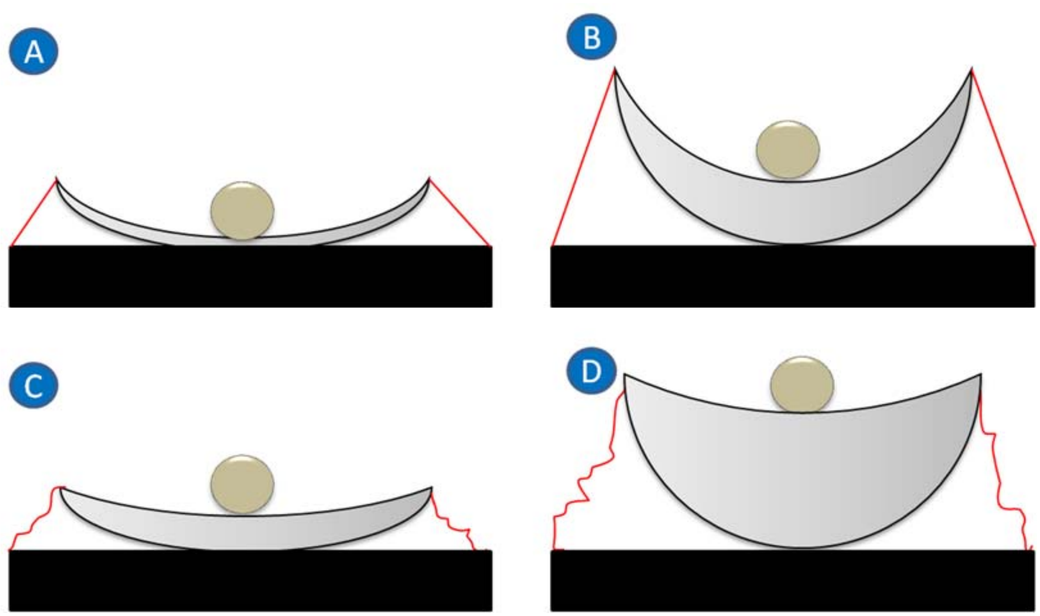

Figure 4. Extension of the Reeves et al. (2007) “ball on curved surface” model to consider a systems-level modeling. The ball on the surface represents the system stability and the cables represent the structures that can provide the stability of the pelvis.

While it is recognized that the effects of these constraints on these kinematic variables and EMG variables (increased peak lumbar flexion angle by 6.4%, increased EMG-off angles by 9.1%, increased EMG of the L3/L4 paraspinals by 7.9%) were relatively modest, these results demonstrate an important biomechanical interplay between the low back and the lower extremities consistent with the results of previous studies (Shin and Mirka, 2007; Shin and D'Souza, 2010; Solomonow et al., 2003) and provide important insights into spine loading that can be used to help design jobs requiring full stooping postures. The results of this line of research could be used to inform more theoretical models (e.g. Hou et al., 2007), EMG-assisted biomechanical models (e.g. Marras and Granata, 1997) as well as field applications, particularly for those ergonomic assessment techniques that utilize whole body approaches to risk assessment (e.g. Kee and Karwowski, 2007).

5. CONCLUSIONS

The results of this study collectively support the significance of the lower extremity influence on lumbar mechanics during deep trunk flexion motions. These effects include kinematic effects and muscle activation profile effects and can be understood through a logical evaluation of their impacts through the pelvis. Extending the modelling approach advocated by Bergmark (1989) to include these super global system effects would seem appropriate for those activities that approach the full flexion posture of the torso.

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