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Influence of knee angle and individual flexibility on the flexion–relaxation response of the low back musculature

Gwanseob Shin *, Yu Shu [a], Zheng Li [b], Zongliang Jiang [a], Gary Mirka [a]

[a] *Department of Industrial Engineering, North Carolina State University, Raleigh, NC 27695, USA*

[b] *Department of Electrical and Computer Engineering, North Carolina State University, Raleigh, NC 27695, USA*

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Abstract

In many occupational settings (e.g. agriculture and construction) workers are asked to maintain static flexed postures of the low back for extended periods of time. Recent research indicates that the resulting strain in the viscoelastic, ligamentous tissues may have a deleterious effect on the stability of the spine and the normal reflex response of spinal tissues. The purpose of this study was to evaluate the previously described flexion–relaxation response in terms of the interactive effect of trunk flexion angle (30°, 50°, 70°, 90°), knee flexion angle (0° (straight knees), 20°, 40°) and individual flexibiliteky (low, medium, and high). These conditions were tested under two levels of loading: no load (just supporting the weight of the torso) and trunk extension moment equal to 50% of the subject's posture-specific maximum voluntary trunk extension capacity. Surface electromyographic (EMG) data were collected from the multifidus, the longissimus, the iliocostalis, the vastus medialis, the rectus femoris, the vastus lateralis, the biceps femoris, and the gastrocnemius-soleus group from a sample of eight male participants as they performed isometric weight holding tasks in the postures defined by the combinations of trunk angle and knee angle. The results of this study showed that knee angle did have a significant effect on the lumbar extensor muscle activity but only consistently at the 90° trunk angle. Participant flexibility showed a consistent trend of decreasing lumbar extensor muscle activity with decreased flexibility across all trunk angle values. Most interesting was the interactive response of flexibility and knee angle, wherein the flexibility of the participant influenced the trunk angles at which the knee flexion angle affected the flexion–relaxation response. Highly flexible subjects showed an effect of knee angle on the flexion–relaxation response only at the 90° trunk angle; subjects in the medium flexibility category showed a similar response in both the 70° and 90° trunk angles; subject in the low flexibility group showed no knee angle effect on the flexion–relaxation response. Overall the results confirm previous results with regard to the contribution of the passive tissues to the overall trunk extension moment but also show that the tension in the bi-articular biceps femoris, which was influenced by knee flexion angle and flexibility, affects the ratio of active extensor moment contributions of the lumbar extensor musculature to passive extensor moment contributions from the muscular and ligamentous tissues. The results of this study provide empirical data describing this complicated, interactive response.

Keywords: EMG; Flexion–relaxation; Flexibility; Knee angle

1. Introduction

Low back pain is quite common in the general population, affecting up to 85% of the population sometime during their lifetime [1,12], and the compensation costs due to the work-related low back pain are estimated to be in the tens of billions of dollars annually [29]. In their review of the economic burden associated with low back pain Maetzel and Li [14], state that "the cost of illness of the low back pain (LBP) is high and is comparable to other disorders such as headache, heart disease, depression or diabetes" (p. 28). The magnitude of the problem as evidenced by these statistics motivates much of the research into the basic etiology of work-related low back pain.

The majority of the previous research into LBP has concentrated on the risk factors associated with

* Corresponding author. Tel.: +1-919-515-7210.
E-mail address: gsshin@unity.ncsu.edu (G. Shin).

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dynamic manual lifting or heavy weight handling because of their strong link with a high incidence of occupational low back pain [13,28]. These risk factors include high force lifting, high force pushing/pulling, repetitive trunk motions, awkward lifting postures and three-dimensional trunk motion characteristics. Receiving less attention are the long duration static, full flexion postures of the trunk assumed during many work tasks such as construction or agriculture. These postures are often maintained for extended periods of time but can require little or no hand held weight so are overlooked as potential problem tasks.

A body of work by Solomonow and colleagues [11,22–25] indicates the importance of considering the relationship between the viscoelastic properties of the passive tissues of the spine and a potential mechanism for spinal injury. In their most recent work [24], this team used the supraspinous ligament of the cat to assess the potential neuromuscular disorders due to static loads to these tissues. Their procedure involved a 20 min application of a load that produced strains that were within the physiologic range for this tissue and a moderate lumbar flexion. These authors note that at the completion of the loading period and as the recovery period proceeded, there was a gradual, exponential increase in the EMG signal of the multifidus back towards its original resting level. They hypothesized that as the creep in the ligament continued its recovery, the mechanoreceptors in the tissue became more sensitive and provided the afferent information necessary for the reflex response of the multifidus muscle. The collective response of the reduced elongation of the ligament and the improved reflex response of the multifidus muscle created a more stable and stiffer intervertebral joint. Many of the results generated in this work have strong correlates with an in-vivo human study performed by these authors [23] indicating that the negative impact of the reduced stiffness and stability of the spine that were the result of these static flexion loads can have a direct link to a mechanism of back injury in humans as they perform activities in fully flexed trunk postures.

Recognizing that virtually any work activity (static or dynamic) that requires non-neutral trunk postures will involve bi-articular and multi-articular muscles of the low back and the thighs, it is believed that understanding the interactive response of the muscular and ligamentous systems to changes in position of any of the participating joints can lead to a deeper understanding of the etiology of low back pain and injury. Major lumbar extensors, like the longissimus, the iliocostalis, and the multifidus originate at the sacrum and pelvis, and therefore the instantaneous position of the pelvis, can influence both the passive and active forces produced by these muscles. Further, the stresses and strains in the complex ligamentous system that connects the pelvis to the spine are also directly influenced by the degree of rotation of the pelvis in the sagittal plane. The degree of rotation of the pelvis, in turn, is governed somewhat by the 'pelvifemoral rhythm' [7,16] which describes the normal interactive motions of the pelvis and the femur. Since several of the muscles of the anterior and posterior compartments of the thigh span both the knee and hip joints, the knee joint flexion angle and muscle activations may affect this pelvifemoral rhythm. In the posterior compartment changes in knee angle can have a profound effect on the hamstring length and/or tension and thereby influence the position of the pelvis. Dewberry et al. [7] illustrated this point when they controlled hamstring length and knee position as independent variables while they examined the pelvifemoral ratio. Both hamstring length and knee position (flexed vs. extended) variables had significant effects on the pelvic contribution to hip flexion. These results demonstrate that knee flexion and thereby hamstring length/tension does influence the motion of the pelvis.

Much of the research considering the implications of the interactions between knee angle and low back stress has considered it from a dynamic lifting perspective through a comparison of the stoop lift (knees straight, trunk flexed) and the squat lift (knees flexed, trunk straighter) [2,10,18,27]. Since one of the main differences between the two techniques is the knee joint angle, many studies have focused on the inter-joint coordination between knee joint and hip or L5/S1 joint during dynamic lifting tasks [5,6,26]. Burgess-Limerick et al. [5] examined the posture and muscle activities while subjects were performing lifting with self-selected technique. From the observed coactivation of the mono-articular knee extensors and bi-articular hamstrings early in the lifting movement, they suggested that the knee extensors contribute to hip extension through a tendinous action of the hamstring. Although the inter-joint coordination and the effects of leg muscles activities on dynamic lifting have been observed from many experimental studies, the quantitative relationships between leg muscle activities and lumbar extensor activities during static weight holding have not been investigated.

In tasks involving significant trunk flexion posture held statically, another important biomechanical characteristic, known as 'flexion–relaxation' phenomenon, needs to be considered [19]. In positions of full trunk flexion, the myoelectric activity of trunk extensors is diminished to low levels and it is believed that passive tissues of the back support the load through a viscoelastic stretch mechanism. Sihvonen [20] investigated the simultaneous activity of back muscles and hamstring muscles during sagittal trunk flexion and extension using surface EMG. The flexion–relaxation of back muscles occurred at a mean of 79° trunk flexion angle and hamstring muscles activity (EMG) ceased at nearly full trunk flexion (97%) angle. Despite myoelectric silence during flexion–relaxation it should be emphasized that the back extensor musculature can still generate substantial forces through passive mechanisms [15]. The biomechanical (moment arm, line of action) and physiological (stress–strain relationships, energy utilization profiles, injury mechanism) differences between active and passive trunk extensor mechanisms indicate that a detailed understanding of the interactions and tradeoffs between these two systems should be empirically derived. One important aspect of this empirical characterization is to establish the relationships between trunk angle, knee angle, and individual flexibility in the activation levels of the lumbar extensor musculature. The specific aims of the current work were to quantify the interactive effect of trunk angle and knee angle on the lumbar extensor musculature and to quantify the effect of individual flexibility in this response.

2. Methodology

2.1. Subjects

Eight male subjects with a mean (and standard deviation) age of 27 years (2.3), height 1.74 m (0.03), body mass 68.5 kg (4.9) were recruited from the university community and participated voluntarily. All subjects were free from chronic and current back problems and gave written informed consent after being introduced to the nature of the study. After a period of brief warm up and light stretching, we assessed participant flexibility level by having the subject flex his trunk forward and reach towards the ground with knees straight. Subjects were categorized as low-flexible (finger tip reach greater than +5 cm from floor, two subjects) (group 3), mid-flexible (finger tip reach between – 5 cm and + 5 cm, three subjects) (group 2), and high-flexible (finger tip reach less than – 5 cm from floor, three subjects) (group 1).

2.2. Apparatus

Eight pairs of surface Ag-AgCl electrodes (Model E22x, In-Vivo Metric) were used to collect the electromyographic (EMG) muscle activity of the sampled muscles. These data were pre-amplified (1000×) and then carried via shielded cable to the main amplifiers that filtered (60 Hz, and low pass 1000 Hz) and further amplified (50×) the EMG data. An isokinetic lumbar dynamometer was used to provide the necessary static resistance for the collection of the angle-specific maximum voluntary contraction (MVC) EMG data from the lumbar extensors. This dynamometer system was also able to provide a measure of the angle-specific peak moment generated by the subject, necessary for the calculation of the hand-held loads as discussed later. A stationary chair system was developed to capture the angle specific MVC EMG data for the knee flexors and extensors.

2.3. Independent variables

Independent variables of this study included sagitally symmetric trunk flexion angles (30°, 50°, 70°, and 90°), knee flexion angles (0° (knees straight), 20°, and 40°), static moment on L5/S1 joint (no external load, and 50% of the subject- and angle specific trunk extension moment), and subject flexibility level (high, medium, low). The angle between a vertical reference line and the line joining the acromion process and greater trochanter was defined as trunk flexion angle. The angle between the line joining the greater trochanter and center of rotation of the knee and the line joining the center of rotation of the knee and the lateral malleolus of the ankle was defined as knee flexion angle.

2.4. Dependent variables

Dependent variables were the normalized EMG from lumbar extensors (multifidus, iliocostalis, and longissimus), knee extensors (rectus femoris, vastus lateralis, and vastus medialis) and knee flexors (gastrocnemius-soleus and biceps femoris). All data were collected on the right side only.

2.5. Experimental procedures

Surface electrodes were placed over eight muscles on the right side of the body of the subjects. The muscles sampled (and sampling location) were: 1) multifidus (1.5 cm to the right of the vertebral midline at L4 level), 2) longissimus (3.5 cm to the right of the vertebral midline at L2 level), 3) iliocostalis (4.5 cm to the right of the vertebra midline at L2 level), 4) vastus medialis (8–9 cm above the knee joint cleft), 5) rectus femoris (9–10 cm above the knee joint cleft), 6) vastus lateralis (8–9 cm above the knee joint cleft), 7) biceps femoris (10–12 cm above the knee joint cleft), and 8) gastrocnemius-soleus group (medial side, 10–12 cm below the knee joint cleft) with a fixed inter-electrode center to center distance of 2.5 cm. While the signal collected from the three trunk extensors almost certainly contained cross-talk from adjacent trunk extensors muscles, the electrode placement locations were chosen to maximize the contribution of the named muscle based on the relative cross-sectional areas of the muscles in the region (L2–L4). Prior to placement, the electrode placement area was shaved, abraded and cleansed with isopropyl alcohol absorbed cotton ball to lower the electrical impedance. All EMG data were collected at 1024 Hz.

After the experimental setup, the maximum voluntary trunk extension moment and lumbar extensor MVC EMG data were collected using isokinetic dynamometer as the subject performed the isometric trunk extension exertions at each of the four different trunk flexion angles (30°, 50°, 70°, and 90°). The maximum EMG of knee extensors and flexors were then collected at 0° (knees straight), 20°, and 40° knee flexion angles as the subject sat on the stationary chair system and they pulled against a secure harness. The MVC exertion for the gasctrocnemius-soleus group was accomplished by having the subject exert a maximum plantar flexion force against a stationary surface while seated in the chair. Each exertion was maintained for three seconds. All MVC trials were collected only after subject completely understood and performed consistent pre-test trials. Between consecutive MVCs a one minute rest break was given.

In the subsequent trials subjects experienced loading conditions of no load (simply holding the weight of the torso) and a condition that required the subject to produce an extension moment (about L5/S1) equal to 50% of their posture specific capacity. This 50% condition was accomplished by having the subject hold a barbell loaded with the appropriate amount of weight. The procedure used to calculate the required hand-held weight required the development of a simple biomechanical model to derive the static moment of the upper body mass as well as the posture-specific trunk extension maximum that occurred in the isokinetic dynamometer. Using these two inputs the posture-specific moment was calculated and the appropriate hand held weights were calculated for each trunk flexion condition for that subject.

Subjects performed a total of 48 trials (two repetitions of all combinations of four trunk flexion angles, three knee flexion angles, and two load conditions). At each trial, subjects bent their torso forward and flexed (or straightened in the case of 0° condition) both knees and held the barbell using both hands (Fig. 1). The experimenters used goniometers to establish when the subject had achieved the appropriate knee and trunk angles. The subject was then asked to hold that position and keep his heels in contact with floor. As soon as the posture was stable, the muscle activities were collected for 3 s. There was a 20 s rest break between consecutive trials during which the subject was allowed stand up in a relaxed posture. Task order was fully randomized.

2.6. Data processing

The raw data were filtered in software using a 10–500 Hz pass filter as well as a notch filter that eliminated 60 Hz and its aliases. Once filtered these signals were rectified (full-wave) and averaged across the three second data collection period. This processing occurred in both the data collected during the experimental trials as well as the maximum voluntary contractions. The EMG data collected during the maximum exertions were then partitioned into 1/8 s windows and the maximum of the 24 windows for each muscle in each posture were identified and were used as the denominator in the process of normalizing the experimental data.

Fig. 1. The static weight holding task.

2.7. Data analysis

The data set was partitioned into two subsets by load condition and then each set was analyzed using ANOVA to examine the effects of trunk flexion angle, knee flexion angle and subject flexibility and their interactions. This preliminary ANOVA was performed to get a sense of the global effects of these variables. To test the specific hypotheses of the current study further partitioning of the data was necessary. In an effort to be more refined in the analysis of the effect of flexibility and knee angle on the flexion–relaxation response, these datasets were further partitioned by trunk angle and an ANOVA was performed on each of these trunk angle-specific datasets to identify those particular trunk angles at which the flexion–relaxation response was influenced by knee angle and individual flexibility.

3. Results

Considering the results of the initial ANOVA (Table 1) from a more global perspective yields two intuitive results and two that were not so intuitive. First, trunk angle had a strong effect on the three lumbar extensor muscles with increased trunk flexion angle leading to decreased muscle activity in the no external load condition (Fig. 2) (combination of increased reliance on the passive forces in the spine, increased moment due to forward bend, increased capacity due to the movement on the length-tension curve of the lumbar extensors, and finally severe flexion–relaxation at the 90° position) and a decreasing muscle activation profile with increasing trunk flexion in the 50% of maximum condition (Fig. 3) (a response consistent with the increased passive contribution to the extensor moment leading to the flexion–relaxation response). Second, knee angle had a strong effect on the muscles of the thigh and leg with the greater knee flexion generating greater muscle knee extensor (Fig. 4) and reduced flexor activity (Fig. 5). These results are well documented in the previous literature.

Less intuitive were the significant results related to subject flexibility. These results show that those subjects with the greatest flexibility had the highest activity levels of the lumbar extensor musculature across trunk angles (Fig. 6).

To address the fundamental questions posed by this research—What is the nature of the influence of knee angle and individual flexibility on the flexion–relaxation response?''—the dataset was partitioned by trunk angle and a series of ANOVAs assessing the impact of these variables on the muscle activities were performed for each trunk angle (Tables 2 and 3). Returning attention back to the data displayed in Figs. 2 and 3 it can be seen that the effect of the knee angle on the flexion–relaxation response is evident in the 90° trunk flexion postures in both loading conditions. These results indicate that the effect of knee angle is particularly important when the individual nears the end of their range of motion where the slight changes in pelvis rotation can have a significant impact on the passive contribution to the extensor moment. The effect of individual flexibility, on the other hand, seems to be a more consistent response across trunk angles. Fig. 6 shows that at every trunk angle the low flexibility group generated less muscle activity and relied more on the passive extensor moment.

Further exploration of this flexibility response in the extreme flexion positions (70 and 90°) showed that the knee angle response was affected by the flexibility of the subject (Fig. 7). This figure illustrates that for the high flexible subjects there was a strong effect of knee angle on the flexion–relaxation response at the 90° trunk flexion angle but there was no knee angle effect at 70° position. In the middle flexibility group it appears that this knee angle effect is also seen at the 70° position indicating that the transition point from passive to active occurs around this point for this group. In the low flexibility group there does not appear to be any knee angle effect on the flexion relaxation process indicating that, for this group, the transition point is less than 70° and regardless of knee

Table 1
ANOVA results testing the effects of trunk angle, knee angle, flexibility, and their interactions

	mult	long	ilio	vasm	recf	vasl	bicf	gast
No load								
trunk angle	***	***	***	*			***	*
knee angle				***	***	***	***	***
flexibility	**	*	*	***	***	***	***	***
trunk*knee							*	
trunk*flexibility								
knee*flexibility					**	**	*	
50% MVC								
trunk angle	***	***	***				***	
knee angle				***	***	***	***	***
flexibility	**	***	***	***	***	***	***	***
trunk*knee								
trunk*flexibility								
knee*flexibility				*		***	**	

* $p<0.05$.
** $p<0.01$.
*** $p<0.001$.

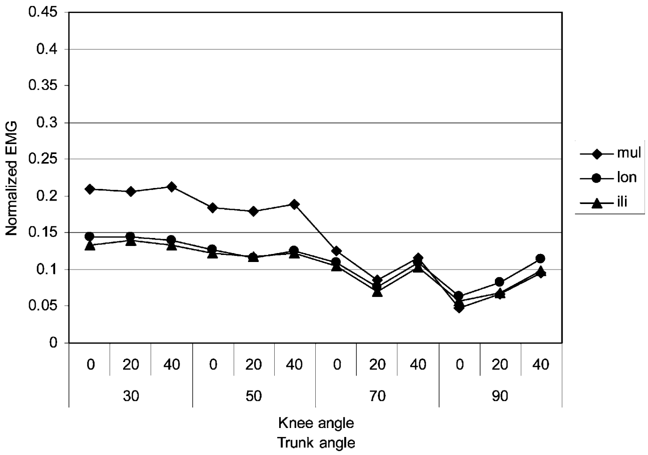

Fig. 2. Normalized EMG of lumbar extensors vs. knee angle and trunk angle (no load condition).

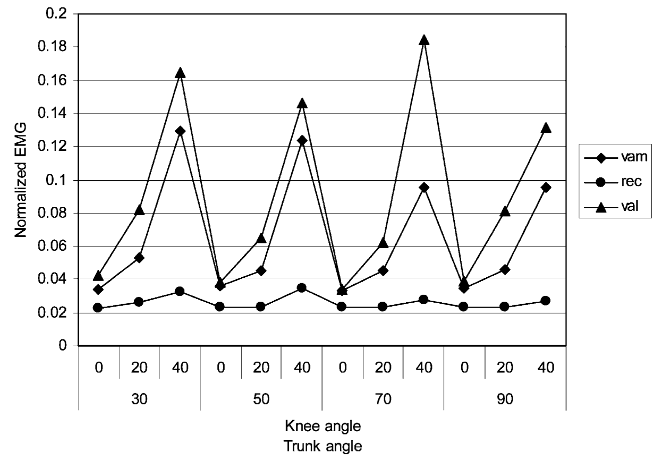

Fig. 4. Normalized EMG of knee extensors vs. knee angle and trunk angle (50% MVC condition).

angle the passive contribution is still high. Further post-hoc statistical analysis showed that only the EMG at the 90° position in the high flexibility group increased significantly ($\alpha = 0.05$) .

4. Discussion

In this study, two different load conditions were tested—no-load and 50% MVC load. From a simple biomechanical perspective, in the no-load condition the load on lumbar extensors is expected to increase as trunk angle increases because of the increase in the moment arm of the center of mass of the torso. This should continue until the passive forces in the spine contribute to the trunk extensor moment during the flexion–relaxation phenomenon. In the 50% MVC load condition considered in this study, however, the external load on lumbar extensors is expected to be relatively constant through the design of the experimental procedure. This renders two perspectives on the investigation of the flexion–relaxation response—one from a basic science perspective (the 50% MVC condition because of the tight control on the moment variable) and one from an applied perspective (the no-load condition because of its similarity to many occupational tasks such as harvesting ground crops, construction work, etc.). The results of this study have shown that in both load conditions, the normalized EMG of lumbar extensors were significantly affected by trunk angle and decreased as trunk angle approached full flexion (consistent with the flexion–relaxation response). The primary focus of the current research was to develop an empirical characterization of this response as a function of knee angle and individual flexibility.

Much of the basic physiology of the flexion–relaxation response can be seen in Panjabi et al. [17]. From the load-displacement curves, three-dimensional morphometric measurements, and a mathematical model they developed, the strains of spinal ligaments were com-

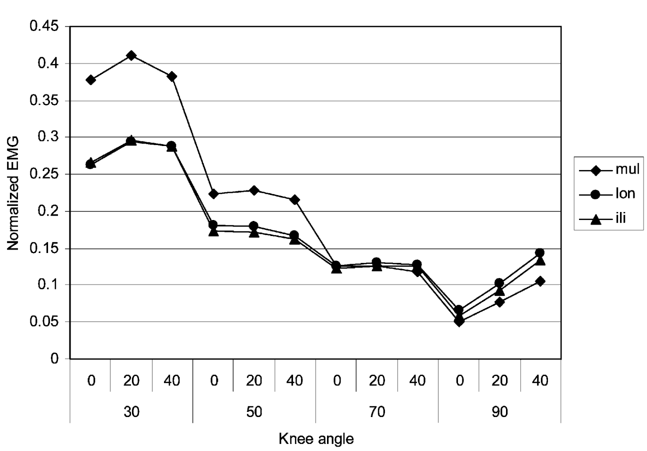

Fig. 3. Normalized EMG of lumbar extensors vs. knee angle and trunk angle (50% MVC condition).

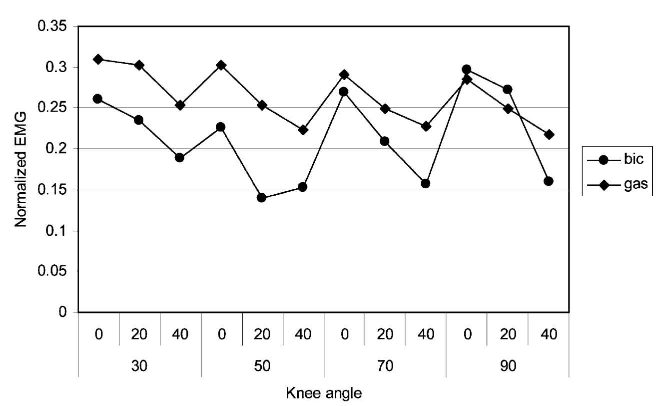

Fig. 5. Normalized EMG of knee flexors vs. knee angle and trunk angle (50% MVC condition).

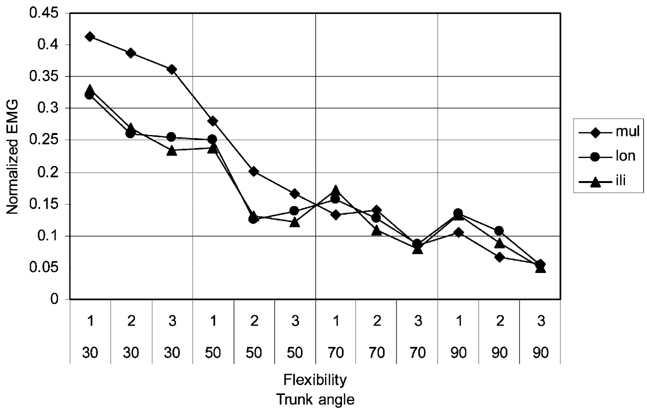

Fig. 6. Normalized EMG of lumbar extensors vs. flexibility and trunk angle (50% MVC condition) (group 1: high-flexible, group 2: mid-flexible, group 3: low-flexible).

puted as functions of various lumbar movements (flexion, extension, lateral bending, and rotation). In flexion movements, ligaments were stretched linearly as flexion angle increased and the interspinous and supraspinous ligaments had the highest strains and the capsular ligaments and the ligamentum flavum were strained less, relatively. One of the positive aspects of this response is a reduction in the required active muscle contribution to the extensor moment, often resulting in reduced muscular energy consumption and fatigue. The negative aspects of this response, however, are a concern. First, increasing the ligamentous component to the extensor moment will increase spinal loading due to the shorter moment arm of the ligaments relative to the muscular tissue. Second, the ligamentous tissues display viscoelasticity indicating that the extensor moment provided by these tissues will vary over time due to creep. Further, when the spine is returned to the non-flexed, upright postures the strain in these tissues does not immediately disappear, leaving a spine without some of the passive (through mechanical support) and active (through loss of reflex muscle control) sources of stability. Solomonow and colleagues have developed an

Table 2
ANOVA results testing the simple effect of knee angle at each trunk angle (data set was partitioned by trunk angles and the effect of knee angle was tested)

Load	Trunk angle	multifidus	longissimus	iliocostalis
No load	30			
	50			
	70	**	***	***
	90	*	**	**
50% max.	30	*	**	
	50			
	70			
	90	*	**	**

$^{***}p< 0.001$; $^{**}p< 0.01$; $^{*}p< 0.05$.

Table 3
ANOVA results testing the simple effect of flexibility at each trunk angle (data set was partitioned by trunk angles and the effect of flexibility was tested)

Load	Trunk angle	multifidus	longissimus	iliocostalis
No load	30	**		*
	50			
	70	*		
	90			
50% MVC	30		*	***
	50	*	***	***
	70			
	90			

$^{***}p< 0.001$; $^{**}p< 0.01$; $^{*}p< 0.05$.

understanding of this response in a series of previously published papers [11,22–25]. While the current study did not employ conditions that were of sufficient duration to activate the viscoelastic response, it does provide information regarding postural (knee angle, trunk angle) and individual factors (flexibility) that are sufficient to place the biomechanical system in a position where stress is placed on these tissues that can lead to these viscoelastic changes.

The contribution of passive tissues in generating extensor moments during lifting was examined by Dolan et al. [8]. In their study, participants performed dynamic lifting from a floor level while EMG was collected from the erector spinae (L3 and T10 level). From an equilibrium equation of moments acting on the lumbar spine, they defined the amount of difference between extensor moment and moment by muscle activity as the passive extensor moments. That is, the extensor moment is equal to the sum of the moment by trunk muscles and the moment by passive tissues. In their results, the relationship between the extensor moments from passive tissues (including intra-abdominal pressure) and trunk flexion was approximately bi-linear. The passive extensor moment increased slightly from the lordotic standing position up to 80% flexion, but it increased considerably after 80% flexion, in both women and men subject groups. While the activity performed in this study was dynamic, and the task performed in the current study was static, the similarities between these previous results and those shown in Figs. 2 and 3 are strong in that it was at the extreme 90° flexion posture that the effect of knee angle on the flexion–relaxation response was most obvious across participants.

In both load conditions, subjects' flexibility had significant effects on EMG of all muscles (Table 1). The trends in lumbar extensor muscle activity showed a consistent reduction across trunk angles with reduced flexibility. In this study a modified sit-and- reach test was used that had the subjects assume the same basic posture but in a standing position to better to gain a

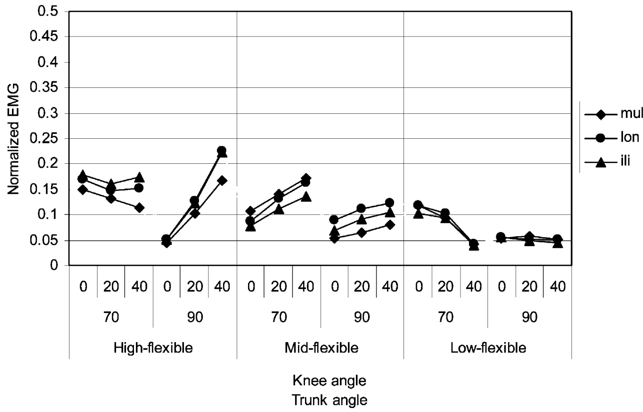

Fig. 7. Normalized EMG of lumbar extensors vs. knee angle, trunk angle, and flexibility (50% MVC condition).

better simulation of the experimental task to be performed. The sit-and-reach test results have been found to be exclusively determined by hamstring flexibility [21]. Since the length and tension on hamstring have effects on the pelvic contribution to pelvis rotation (and thereby the passive spine extensor mechanism), relatively larger active extensor moments would be expected with high-flexible subjects than low or mid-flexible subjects and these were the findings of the current research. Further exploration of this flexibility response in the extreme flexion positions (70 and 90°) showed that the knee angle response was significantly affected by the flexibility of the subject (Fig. 7), leading to the formulation of a simple multi-joint articular model that considers these interactive effects.

While standing upright without any trunk or knee flexion, the hamstring muscles and spine ligaments are in moderate length with little passive tension (Fig. 8a). Inclination of the torso from this position involves flexion of the vertebral column (including sacroiliac flexion) and hip flexion. The lumbar spine contributes more to early forward bending (0–30°) than the hip, and the hip contributes more to late forward bending (60–90°) than the lumbar spine [9]. The pelvifemoral rhythm that describes these relative motions is expressed as the ratio of angle change between the pelvis and vertical reference to angle change between the femur and vertical reference. The contribution of pelvis rotation to hip flexion has been shown to vary as a function of knee angle. The mean ratio of pelvis rotation to hip flexion has been reported to be 26% when knees are flexed [3], and 39% when knees are straight, with subjects in the supine position [4]. In the standing position, the pelvis rotation could contribute 18.1% to hip flexion (leg raise with knees flexed) [16].

As the person bends forward, the pelvis freely rotates forward (α) until the passive tension in the hamstrings begins to influence pelvic rotation. As the person continues to flex forward (Fig. 8b) and the forward rotation of the pelvis continues to be restrained, the passive extensor mechanism of the spine is activated. As this process continues with greater forward trunk flexion, the ratio of active to passive extensor moment continues to decrease to the point of full flexion–relaxation. When knee flexion occurs at the trunk flexion posture where the passive mechanism was just activated (Fig. 8c), the distance between the origin and insertion of the biceps femoris is reduced and thereby releases the tension on the pelvis allowing it to rotate forward and deactivating the passive extensor mechanism requiring greater active moment from the lum-

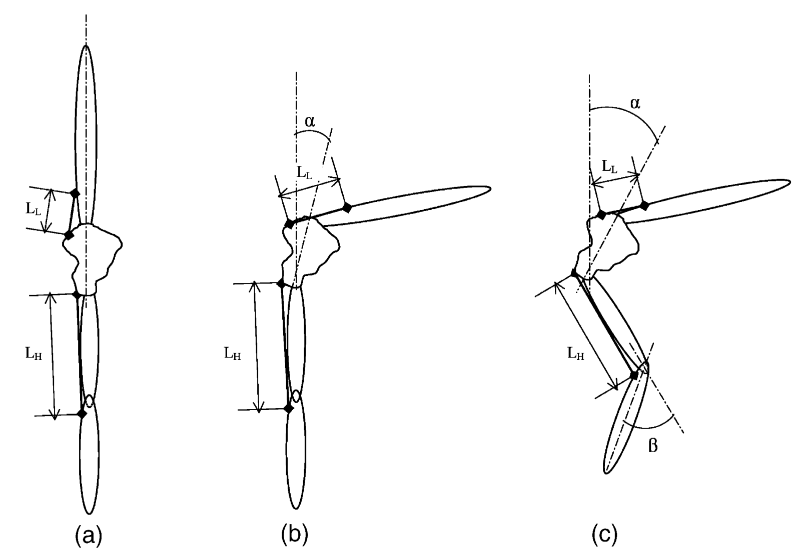

Fig. 8. Simple biomechanical model showing the interaction between trunk flexion angle, knee flexion angle and individual flexibility. L_H = distance between origin and insertion of hamstring; L_L = length of a spine ligament (e.g. supraspinous ligament); α = pelvis angle relative to vertical reference; β = knee flexion angle.

bar extensors. Individual flexibility plays a role in this process in that the amount of knee flexion required to release hamstring tension is going to be dependent on this characteristic. In fact, in some combinations of trunk angle and flexibility, no amount of knee flexion is going to be adequate to release the tension in the hamstrings and allow the pelvis to rotate forward (as seen in the low flexible group data in Fig. 7).

There are several limitations of the current research. First, the empirically derived transition points are not very precise because of the large range of trunk and knee angles considered. Our intention in choosing the range used was to be sure to see a transition from pure active to active-passive combination to chart this response. Future research may focus in the trunk angles of $>60^\circ$ and use steps of much smaller value to gain more precise definition of these transition points. A second limitation is related to the generalizabilty of the results. The subject population in the current study was all male of similar anthropometry. For a first attempt at empirically deriving these transition points it was reasoned that having an homogenous population would decrease the nuisance variance in the sample and allow for a clearer picture of the response. Including women and subjects of varied anthropometry would broaden the generalizabilty of these results and is another logical next step in this research effort.

Future research should focus on the verification of the implications of these results with regard to the viscoelastic effects. The duration of trunk flexion at each task in this study was just 3 s, too brief to expect significant viscoelastic characteristics in the strain response of ligaments. The results of this study have indicated the specific multi-joint postures at which the passive mechanism is activated. The viscoelastic changes in these tissue would then be expected to commence in these postures and the resulting variation of ligaments strain would be expected to affect the amount of the passive extensor moments and thereby the required compensation by the active trunk extensor mechanism. Further research on this topic will include long duration of static flexion to cause viscoelastic strain of ligaments and examine the variation of the passive/active ratio of extensor moment as a function of time.

5. Conclusions

The present study examined the effects of trunk flexion angle, knee flexion angle and individual flexibility on the myoelectric activities of lumbar extensors, knee extensors, and knee flexors with the intention of gaining empirical information about the effects of these variables on the flexion–relaxation response of the low back musculature during static trunk flexion tasks both with and without load. The results have shown an important interactive effect of knee angle and individual flexibility on this flexion–relaxation response. The most flexible subjects showed a knee angle effect only at the 90° trunk flexion position. The mid-level flexible subjects showed a knee angle response at both the 70 and 90° trunk flexion positions and the least flexible subjects showed no knee angle effect on the flexion relaxation response. This interactive effect on the flexion–relaxation response has been described in a simple multi-joint biomechanical model.

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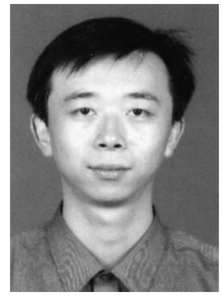
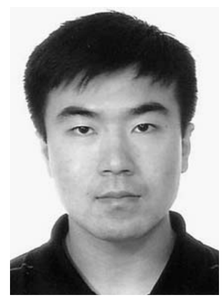
Gwanseob Shin received his BE degree in Mechanical Engineering from the Korea University (Seoul, Korea) in 1996, MS degree in Industrial Engineering from the North Carolina State University in 2002. He is currently a PhD degree candidate at the North Carolina State University. His current research interests are viscoelastic properties of spine ligaments during prolonged static lumbar flexion and its biomechanical modeling.



Yu Shu received a Bachelor of Science degree in Industrial Engineering from the Tsinghua University, Beijing, China. He joined the Ergonomics laboratory of North Carolina State University to study biomechanics, cognitive ergonomics and occupational health and safety. He is currently a research assistant in the Ergonomics laboratory and doing a research on the effects of wrist splints on activities of forearm muscles in different working conditions.



Zheng Li received his B.E. from the Department of Automation, Tsinghua University, Beijing in 2001 and M.S. in Electrical Engineering from North Carolina State University in 2003 respectively. He is currently a Ph.D. student with the Department of Electrical and Computer Engineering in North Carolina State University. His research is focused on networks based control, computational intelligence and application of neural networks on multiple areas. He is a student member of Institute of Electrical and Electronics Engineers (IEEE) from 2001.

Zongliang Jiang obtained his BS degree in Engineering Mechanics from Shanghai Jiao Tong University, China in 1999, and an MS degree in Computer Science from North Carolina State University in 2003. He is currently a PhD student in Industrial Engineering at North Carolina State University. His research interests include the study of cumulative trauma disorders, spine biomechanics, biomechanical modeling and ergonomics.

Gary Allen Mirka is an Associate Professor of Industrial Engineering at North Carolina State University where he has been on the faculty since 1992. He received his BS, MS and PhD degrees in Industrial and Systems Engineering from The Ohio State University. His current research interests involve spine biomechanics, stochastic modeling of biomechanical systems, and ergonomic intervention effectiveness research. Dr. Mirka was a recipient of the 1993 Volvo Award on Low Back Pain Research.