Interaction of viscoelastic tissue compliance with lumbar muscles during passive cyclic flexion–extension

Michael W. Olson a,b,d, Li Li b,d,*, Moshe Solomonow b,c,d

a Department of Kinesiology, Southern Illinois University, Carbondale, IL, USA
b Department of Kinesiology, Louisiana State University, 112 Long Field House, Baton Rouge, LA, USA
c Bioengineering Division, University of Colorado at Denver and Health Sciences Center, Denver, CO, USA
d Occupational Medicine Research Center, Louisiana State University, Health Sciences Center, New Orleans, USA

Received 7 December 2006; received in revised form 14 May 2007; accepted 16 June 2007

Abstract

Human and animal models using electromyography (EMG) based methods have hypothesized that viscoelastic tissue properties becomes compromised by prolonged repetitive cyclic trunk flexion–extension which in turn influences muscular activation including the flexion–relaxation phenomenon. Empirical evidence to support this hypothesis, especially the development of viscoelastic tension–relaxation and its associated muscular response in passive cyclic activity in humans, is incomplete. The objective of this study was to examine the response of lumbar muscles to tension–relaxation development of the viscoelastic tissue during prolonged passive cyclic trunk flexion–extension. Activity of the lumbar muscles remained low and steady during the passive exercise session. Tension supplied by the posterior viscoelastic tissues decreased over time without corresponding changes in muscular activity. Active flexion, following the passive flexion session, elicited significant increase in paraspinal muscles EMG together with increase in the median frequency. It was concluded that reduction of tension in the lumbar viscoelastic tissues of humans occurs during cyclic flexion–extension and is compensated by increased activity of the musculature in order to maintain stability. It was also concluded that the ligamento-muscular reflex is inhibited during passive activities but becomes hyperactive following active cyclic flexion, indicating that moment requirements are the controlling variable. It is conceived that prolonged routine exposure to cyclic flexion minimizes the function of the viscoelastic tissues and places increasing demands on the neuromuscular system which over time may lead to a disorder and possible exposure to injury.

© 2007 Elsevier Ltd. All rights reserved.

Keywords: Low back disorder; Electromyography; Passive flexion; Viscoelastic tissues

1. Introduction

Low back disorders are a prevalent condition in industrialized countries even with the advancement of modern technologies used in the workplace (Marras, 2000, 2005). The U.S. Department of Labor (2004) reports that injuries to the low back are the leading cause of missed work days. The treatment of these injuries costs individuals and corporations over $10 billion per year (NIOSH, 1999). The etiology of low back disorders in individuals who work in labor intensive settings is not clear. Repetitive movements performed in the labor setting may be risk factors for low back disorders.

Trunk movement can significantly alter the mechanical forces acting on the vertebrae and consequently modify the lumbar muscular activities (Dolan and Adams, 1993; Dolan et al., 1994a; Kankaanpaa et al., 1998; van Dieën and Oude Vrielink, 1998). Granata and Marras (1999) indicate that multiple factors, such as dynamic compression and shear loading, muscle fatigue, and specific requirements of the workplace affect episodes of low back
disorders. Many labor intensive tasks require either trunk flexion–extension, trunk rotation, or a combination of both, incorporating co-activation of abdominal and paraspinal muscles which increase forces acting on the lumbar spine (Marras et al., 1998; Schultz et al., 1982). Trunk flexion–extension is studied extensively in the literature as it is the primary motion performed during lifting tasks (Dolan and Adams, 1998; Dolan et al., 1994b; Esola et al., 1996; Gracovetsky et al., 1990; Ng et al., 2001; Oddsson and Thorstensson, 1990; Schultz et al., 1985). In fact, continuous trunk flexion–extension does result in modified paraspinal muscle activity suggesting this movement may be a significant factor in eliciting low back disorder (Caldwell et al., 2003; Dickey et al., 2003; Marras and Granata, 1997; Olson et al., 2004).

Altered muscle activation due to muscular fatigue is hypothesized to influence the incidence of low back injury and disorder (Bonato et al., 2003; Dolan and Adams, 1998; Elder et al., 1982; Mannion et al., 1997; Moritani et al., 1982; Petrofsky and Lind, 1982). The effects of fatigue are temporary as muscles recover from the requirements of the task over a relatively short period (Hermann and Barnes, 2000; Kuorinka, 1988; Mannion and Dolan, 1994; Potvin and Norman, 1993). Frequent breaks and extended rest periods are common strategies used to reduce the effects of fatigue in the workplace suggesting that muscle fatigue may only be a minor factor in considering low back injuries.

McGill and Brown (1992) indicate that prolonged static flexion of the trunk significantly alters the viscoelastic properties of the tissues leading to a possible compromise of spine stability. In healthy individuals, the flexion–relaxation phenomenon (myoelectric silent period) observed from the lumbar muscles has been related to deformation of the viscoelastic tissues at deep flexion angles (Allen, 1948; Floyd and Silver, 1951, 1955; Portnoy and Morin, 1956). Further deformation, due to prolonged repetitive loading, may signify an inability of the viscoelastic tissues in maintaining vertebral joint integrity (Dickey et al., 2003; Olson et al., 2004). Previously, we have reported the significant adaptation of the flexion–relaxation phenomenon during acute bouts of trunk flexion (Olson et al., 2006).

The direct consequences of repetitive flexion loading on the viscoelastic tissues have not been determined in humans, although indirect evidence has been presented in cadaver and animal models. Manipulation of cadaver (Dumas et al., 1987; Goel et al., 1988; Hukins et al., 1990; Pintar et al., 1992) and animal (Claude et al., 2003; Solomonow et al., 1999) ligaments through cyclic loading in creep and tension–relaxation protocols result in decreased tension within the tissues. Isolation of the passive tissues in vitro and in vivo assures that mechanical measurements are independent of muscle activity. There is no direct evidence in human models to indicate changing moments due to tension–relaxation or laxity within the posterior lumbar viscoelastic tissues after repetitive loading. Evidence of this nature would support current theories of a repetitive loading etiology responsible for low back disorders.

The purpose of this study is to examine the behavior of the lumbar viscoelastic tissues and muscles during repetitive passive cyclic trunk flexion–extension. It is hypothesized that the moments generated by the lumbar passive tissues will decrease (eg, tension–relaxation) as a result of repetitive cyclic sagittal trunk movement requiring increase in external moments. It is also hypothesized that a neuromuscular adaptation during active flexion will occur in the surrounding lumbar muscles as a result of the diminishing function of the viscoelastic tissues. The new insight gained by obtaining such information will improve and confirm our understanding of the role and importance of viscoelastic tissues in cyclic activities, the relationship with the function of the musculature and the role of the structures in development of lumbar disorders.

2. Materials and methods

2.1. Participants

Eighteen male college students volunteered to participate in this experiment. University approval and consent from each individual were obtained before data collection. The mean (SD) height, mass, and age of the subjects were 1.79 (0.08) m (range 1.65–1.96 m), 78.4 (11.2) kg (range 63.6–99.1 kg), and 21.6 (1.9) years (range 19–27 yrs), respectively. None of the participants reported any previous back or lower extremity pain or disorders.

2.2. Instrumentation

A dynamometer (830-110, Biodex Medical Systems, Shirley, New York) was used to control the passive flexion–extension movement of each person. The System3 software (Biodex Medical Systems, Shirley, New York) was used to collect applied torque, angular trunk displacement, and angular trunk velocity data. The axis of the dynamometer was aligned with the L4–L5 joint of the participant (McGill and Kippers, 1994). An attachment to the dynamometer was constructed (L-shape) to support the trunk during movement (see Fig. 1). The individual was harnessed to the horizontal arm of the attachment across the shoulders just inferior to the clavicles.

Surface electromyography (EMG) recordings were collected using an MA300 system (Motion Lab Systems, Baton Rouge, LA). EMG signals were collected from the right side of the individual using pre-gelled Ag–AgCl bipolar electrode pairs. The 1.0 cm diameter electrodes were spaced 2.5 cm apart (center to center) along the length of the underlying muscle fiber. The four placement sites were the rectus abdominis muscle, 3.0 cm lateral to the umbilicus, the external oblique muscle, 15 cm lateral to the umbilicus, and the paraspinal muscles at the L2 and L4 levels, 3.0 cm lateral to the respective spinous processes. The myoelectric signals observed from the abdominal muscles remained silent during the passive session and sporadic during the active flexion–extension sessions, and were not reported in the results. The EMG signals were amplified differentially with available gain up to 18,000 times, frequency band pass of 10–480 Hz, and common mode rejection ratio (CMRR) of 100 dB. Data were collected at 1000 Hz using a 12 bit analog to digital board and stored for
reached his maximal flexion angle based on his ability (as opposed to a common deep angle for all subjects).

The passive session of the experiment began as the individual was secured to the dynamometer. The angular velocity of the movement (21 ± 2 degrees s⁻¹) was calculated based on each subject’s pre-recorded range of motion, measured with the dynamometer, to assure the 5 s of flexion and extension were performed each cycle. Six cycles per minute were performed over a 10 min period. External torques applied by the dynamometer were recorded over this time period. The dynamometer was calibrated prior to and after data collection.

Each participant was instructed to remain relaxed and rest his body against the attachment arm of the dynamometer during the passive session. A computer monitor displaying the real-time EMG signals from all muscle sites was placed on the floor 30 cm in front of the individual at a 45° angle to ensure muscle silence with biofeedback. After the range of motion was determined, the relaxed participants were guided through the flexion–extension movement by the experimenter to simulate the dynamometer controlled motion. The participant was instructed to observe the EMG signals on the monitor and maintain small signal amplitude as possible in the lumbar EMG displays. This procedure was performed to ensure that the subject did not pursue any voluntary activation of the muscles. To minimize back muscle activity due to head motion, the participant was instructed to keep his chin close to the attachment arm even if the view of the monitor was obscured.

### 2.4. Analysis

Torque data were filtered with a forth-order low pass Butterworth filter at 1 Hz cut off frequency (Winter, 1979). Three dependent variables were of primary interest during the analysis of the torque-angle curve. These variables were the peak torque attained during the flexion phase (PTF), the torque applied at the deepest trunk flexion angle (TT), and the peak torque attained during the extension phase (PTE). The angle at which PTF attainment occurred was also a variable of interest since the torque supplied by the posterior viscoelastic tissues was presumed to decrease over the passive session.

Upper body (head, arms, trunk) center of mass and inertial properties were calculated based on Dempster (Winter, 1979). The position of the segments and the center of mass of each segment in its respective position during the passive session, and anthropometric data taken from each subject, allowed for the calculation of the total upper body center of mass. The moment arm was then calculated using the L4–L5 joint as the center of rotation. Torque due to the viscoelastic tissues of the back was estimated by using the following equation:

\[ T_{pt} = T_m - T_g \]

where \( T_{pt} \), \( T_m \), and \( T_g \) represent torque of the viscoelastic tissues, the measured torque, and the gravity contribution, respectively.

The calculated viscoelastic tissue torques (ligament, fascia, and passive muscle) enable us to estimate the change in the mechanical properties of the tissues over the 60 cycle passive session. This process was performed using the constant velocity portion of the flexion–extension motion.

Torque \( T_{pt} \)-angle relationship was estimated during the separate flexion (ascending) and extension (descending) phases. The differences in the area under the ascending and descending curves at constant velocity were used to estimate the energy...
dissipation of the viscoelastic tissues during each cycle (Heerkins et al., 1987). This was an important factor to consider since accelerations on either end of the movement phase could confound the data.

Raw EMG signals from the passive session and pre/post active sessions were full-wave rectified. A linear envelope was constructed with a forth-order Butterworth low pass filter with a 4 Hz cut off frequency (Bereton and McGill, 1998). The filtered EMG was separated to flexion and extension phases and compounded for future statistical analysis. A threshold level of 10% of the rectified signal was used to determine the offset of the myoelectric silent period as EMG data from this silent period were not used for analysis.

Power spectrum analysis using Fast Fourier transformation was performed on the raw EMG signal from the lumbar paraspinal (LP) muscles of the active sessions to identify possible changes in motor-units recruitment (Bernardi et al., 1996; Lindstrom et al., 1977; Moritani and Munro, 1987; Solomonow et al., 1990) due to muscular fatigue (Lindstrom et al., 1977). A window of 4 s was used for each flexion and extension phase analysis. The period of myoelectric silence was excluded from the frequency analysis and served to separate flexion and extension phases.

Integrated EMG (IEMG), based on the rectified and smoothed data, was calculated for muscle activity during the active flexion and extension sessions. IEMG activities of the lumbar paraspinal muscles during flexion and extension phases from the passive session were also calculated and compared over cycles to measure changes that may have occurred.

### 2.5. Statistical analysis

One-factor ANOVA with repeated measures (Statistix Corp.) was used to analyze selected torque and trunk angle parameters in the passive session (over time), and EMG data from active trials (pre/post active sessions). One trunk flexion–extension cycle from each minute of the passive session was used for analysis of the time component. IEMG during the passive session was analyzed with a two-way ANOVA (time × flexion–extension). The alpha level was set at 0.05.

### 3. Results

#### 3.1. Outcomes during the passive session

Fig. 2 provides typical torque data measured during the passive conditions. A significant hysteresis is apparent in every trial where the measured extension torque values were greater than the measured flexion torque values even when flexion torque increase during the repetitive cyclic movement. Peak torque applied to the dynamometer during the flexion phase occurred between erect stance (0°) and full trunk flexion (~90°, Fig. 2). Peak flexion torque (PTF) increased significantly from 45 (±4) to 60 (±4) Nm (F_{9,178} = 6.24, p < 0.0001). The PTF increased with a significant linear trend in the data (F_{1,152} = 75.91, p < 0.0001). The increasing PTF was accompanied by a significant increase in the angle at which the PTF was attained (F_{9,178} = 2.72, p < 0.006) from 32 (±4) to 36 (± 5) degrees. This observation is consistent with the behavior of tissues subjected to creep or tension–relaxation. The angle of peak flexion torque increased linearly (F_{1,152} = 19.50, p < 0.0001).

The magnitude of TT increased from 7 (±4) to 21 (±4) Nm with continuous passive cyclic movement of the trunk (F_{9,178} = 5.17, p < 0.0001). The TT was observed to increase with a linear trend (F_{1,152} = 41.16, p < 0.0001). The PTE did not increase significantly from 79 (±6) to 81 (±4) Nm, but followed a significant linear trend (F_{1,152} = 6.24, p < 0.014).

Excluding the influence of gravity and the support of the dynamometer, the behavior changes of the posterior viscoelastic tissues over time are plotted in Fig. 3 as the modeled hysteresis curve of each participant. The area within the hysteresis curve of each cycle is related to energy dissipation. Fig. 4 demonstrates that the energy dissipation decreased along a significant linear trend with the 10 min exercise (F_{1,152} = 9.57, p < 0.003).
The IEMG of the paraspinal muscles during passive flexion and extension phases at L2 and L4 were initially pooled into two groups to represent flexion IEMG and extension IEMG. However, the two pooled IEMG sets did not change during the passive session and were further pooled into one set of data ($F_{9,179} = 1.70$, $p > 0.10$, Fig. 5). This observation indicates that changes in torque applied to the dynamometer were not influenced by the activity of the paraspinal muscles.

3.2. Outcomes of the active sessions before and after the passive session

EMG frequency data during active flexion and extension were pooled between L2 and L4 levels since there was no difference between them (Fig. 6). Mean frequency increased but non-significantly from pre ($45 \pm 3$ Hz) to post ($48 \pm 3$ Hz) passive sessions ($F_{1,71} = 2.90$, $p < 0.11$), while the median frequency increased significantly from pre ($114 \pm 3$ Hz) to post ($118 \pm 5$ Hz) passive sessions ($F_{1,71} = 5.13$, $p < 0.04$). This confirmed the absence of fatigue in the lumbar muscles, which supports the contention that paraspinal muscles were active minimally during the passive session. Furthermore, it confirms that new, larger motor units were recruited post passive exercise.

IEMG activities from flexion and extension phases were compared between pre and post active sessions. Paraspinal muscle activity increased significantly from $0.031 (\pm 0.002)$ to $0.036 (\pm 0.002)$ mV s when trunk flexion was performed after the passive session ($F_{1,35} = 7.34$, $p < 0.02$, Fig. 7).

4. Discussion

The major observations of this study consist of a significant gradual decrease in torque supplied by the viscoelastic tissues during passive flexion–extension and a significant increase in LP muscular activity during active flexion–extension after the 10 min of passive trunk flexion–extension. This observation was accompanied with the absence of myoelectric activity from the abdominal musculature.
during active and passive sessions. Passive flexion did not elicit any changes in EMG activity from the posterior muscles whereas significant changes were observed during active flexion.

4.1. Viscoelastic tissue properties

While moving at constant angular velocity during passive low back stretch, there are three torques developed at the lower back: extension torque produced by the viscoelastic tissue ($T_{\text{pt}}$), flexion torque due to gravitational load ($T_{g}$), and the extension torque the Biodynamics applied to the trunk ($T_{m}$). The magnitude of $T_{m}$ is a function of $T_{pt}$ and $T_{g}$ during equilibrium (Eq. (1)). Flexion of the trunk requires an elongation of the lumbar tissues resulting in increasing tension within the viscoelastic structures. The elevated tension allows the viscoelastic tissues to support the load of the trunk through greater torque production ($T_{pt}$) at the joint when deeper trunk flexion angles are attained (Floyd and Silver, 1955; Kippers and Parker, 1984). Decrease of the $T_{pt}$ over time denotes a deficient function of the ligaments, discs and fascia and a detriment to low back stability during repetitive movements.

The results of this study highlight the mechanical property changes that occur within the viscoelastic tissues during repetitive flexion. As the measurements from the dynamometer illustrate ($T_{m}$), the deeper the trunk flexes the greater the resistance supplied by the low back viscoelastic tissues ($T_{pt}$). Greater force application to the dynamometer indicates a compromise, either creep or tension–relaxation, in the viscoelastic support tissues. Implications of this event suggest the viscoelastic tissues developed a decreased capacity to maintain support of the system as continuous cyclic loading is applied. The ramifications of this compromise would bring about instability of the vertebral joints while escalating the chance of injury, or the need for an increased duration of paraspinal muscle activity to preserve joint stability. Further examination of the specific lordotic changes in the lumbar spine during continuous flexion activities is warranted based upon these observations.

The differences between the flexion and extension torque patterns (Fig. 2) are interesting to note. With prolonged cyclic passive motion, the increased $T_{m}$ during flexion was not followed by a corresponding change in $T_{m}$ during extension. In early extension there should be residual strain remaining within the viscoelastic tissues even when unloaded (Jackson et al., 2001). The linear trend of PTE increase indicates tissues were affected by the passive movement during extension.

It is possible that differential recruitment of fibers (collagen) within the tissues contributed to the pattern of observed $T_{pt}$ changes. Recruitment of the collagen fibers is dependent upon their alignment with the line of tensile force applied to the tissues (Woo et al., 1991), and influences the non-linear viscoelastic response of the tissues to loading (Stromberg and Wiederhielm, 1969; Viidik, 1972). Greater elongation of the tissues requires further participation of more fibers, since the collagen fibers are not of a uniform length along the tissue’s axis, to support and sustain the integrity of the biomaterial as external forces are applied (Chalaz et al., 1985).

Hubbard and Chun (1988) observed decreased viscoelastic strength while cyclically loading canine tendons. We did not measure viscoelastic strength, but our results showed a linear reduction in torque within the posterior tissues. Hubbard and Chan also reported most of the strength reduction occurred at the initial phase of the process. It is possible this difference in our observations is due to practice trials before the testing began to familiarize the subjects to the protocol and to the rest provided between these practice trials and the actual testing sessions. We also assumed that limited voluntary contraction of the LP muscles was present initially in the passive session. Any activation of the lumbar muscles would hinder the $T_{pt}$ pattern observed over time. EMG evidence provides support for our contention that muscle activity did not influence the calculated $T_{pt}$. This is an important factor for consideration since the most significant changes in the hysteresis loop occur within the first 10 loading cycles (Magnusson et al., 1998; Taylor et al., 1990; Yahia et al., 1991).

Energy is absorbed by the lumbar viscoelastic tissue during the lengthening period. Compliance of the viscoelastic tissues over time effectively reduces the energy (strain, heat) stored within the tissues while also diminishing the load capacity at specific physiologic strains. Greater amounts of energy stored within the viscoelastic tissues allow for considerable utilization of this energy for sustaining the position of the trunk in space. Differences between $T_{pt}$ in flexion and extension phases were apparent in the hysteresis loop as $T_{pt}$ measurements were not the same at specific angles between flexion and extension phases (Fig. 3). Viscoelastic tissue preconditioning results in little energy dissipation difference between flexion and extension phases. In the in vivo state the tissues circulate fluids and nutrients in order to maintain homeostasis (Butler et al., 1978; Toyoda et al., 1992), and water content significantly affects the compliance of biological tissues to loading (Chimich et al., 1992; Hannafin and Arnowczyk, 1994). However, a preconditioned state is observed in in vivo muscle and tendons during repetitive loading performed over longer durations of time (Kubo et al., 2002, 2005).

These observations are important to the study of the etiology of low back pain and disorder. Mechanically, repetitive trunk movement is linked to the causation of low back disorders of the spine (Dolan and Adams, 1998; Granata et al., 1999; Marras and Granata, 1997). The results reported in this study provide evidence that significant compliance within the lumbar viscoelastic tissues does occur and could be the preface to low back disorder. Further study with diagnostic imaging may help to investigate the exact detailed structural changes caused by such exercise.
4.2. EMG evidence

The EMG data supported our assumption that there was no muscle activity during the passive session. A relatively low level of EMG activity was observed and remained steady and consistent over the 10 min of cyclic flexion–extension representing normal tonic posture, while excluding the flexion–relaxation phenomenon from this analysis. The observed minimal myoelectric activity also supports our contention that the lumbar paraspinal muscles did not contribute to the measured Tm during the passive session. Andersson et al. (1996) and Stokes et al. (2003) contend that superficial muscle activity does not correlate strongly with the activity of deep muscles, particularly the multifidus. Although not very likely, fatigue due to continuous activation of the deep lumbar muscles could possibly contribute to the increased Tm over time. Increased activation of abdominal muscles would also increase the Tm during the passive session (Essendrop et al., 2002; McGill and Kippers, 1994). However, myoelectric activity recorded from rectus abdominis and external oblique muscles were ‘silent’ during the passive session, excluding this possibility.

Differences in muscle activity before and after the passive session imply that paraspinal muscular fatigue due to the passive protocol was absent as a shift of the power density spectrum to lower frequencies would indicate fatigue. The median frequency and the mean frequency of the power density spectrum increased after the passive session indicating recruitment of additional motor units (Bernardi et al., 1996; Moritani and Munro, 1987; Solomonow et al., 1990). Furthermore, IEMG activity in flexion also increased after the passive session as a compensatory result of compliance of the low back viscoelastic tissues. An increase in the number of motor units recruited may explain this IEMG increase confirming the increased median frequency. Since the IEMG represent a net increase in the peak amplitude and frequency content of the signal, one can conclude that increased activity from the musculature compensated for the lost torque in the viscoelastic tissues.

The values of the median frequency and the mean frequency were noted to be substantially different. This resulted from the fact that paraspinal muscles are mostly postural, slow twitch fibers. Consequently, their power spectrum is skewed, with more power present in the low frequency region with a very long, low amplitude toe in the higher frequencies. Since the mean frequency represents the mean of the power in the spectrum it results in a low value. Conversely, the median frequency divides the spectrum into two regions of equal area. Since the toe region asymptomatically extends deep into the high frequencies, the median frequency is placed much higher, yielding a higher value. This feature is unique to slow twitch muscles yet still sensitive to fatigue and motor units recruitment changes, with the mean frequency more sensitive than the median.

Static and cyclic manipulation of in vivo human knee ligaments indicates significant modifications in the activation of the surrounding agonist and antagonist muscles (Chu et al., 2003; Sbriccoli et al., 2005). The results of such loading schemes preformed over extended periods of time illustrate modifications in the neuromuscular system to maintain joint stability during movement. Thus, increased myoelectric activity within the lumbar muscles would dictate that further support of the intervertebral joints is necessary as compliance of the viscoelastic tissues increases. As such, the acute loading performed in this study may provide a prelude to the compensatory adaptation of the neuromuscular system and eventually the development of a chronic low back disorder if persistent.

5. Conclusions

Calculations based on the observed data indicated reduced tension in the posterior lumbar viscoelastic tissues as a result of continuous passive cyclic trunk flexion–extension exercise without concurrent active muscle contraction. A consequence of the decreased tension within the viscoelastic tissues was increased muscle activity during active flexion performed after the passive motion session. Changes of the hysteresis loop observed from the measured and modeled torque were in agreement. Those changes indicate that prolonged cyclic trunk flexion motion greatly modifies mechanical and neural functions in the low back without the concomitance of muscle fatigue. This research provides information suggesting that repetitive stretches of low back viscoelastic tissue are significant factors in modifying the mechanical and neuromuscular functions of the lumbar structures which may, over time, lead to low back disorder. The results of this study also filled the gap in the exploration of neuromuscular disorders in humans through the use of animal models (Claude et al., 2003; Jackson et al., 2001; Solomonow et al., 1999; Sbriccoli et al., 2005). These observations are pertinent to the understanding of everyday movements that may lead to low back disorder.

Acknowledgement

This work was supported in part by the Occupational Medicine Research Center grant from the LSU Board of Regents and in part by grant RO1-OH-07622 from NIOSH.

References


United State Department of Labor 04-460. Lost work-time injuries and illnesses: characteristics and resulting days away from work, 2002; 2004.

van Dieën JH, Oude Vrielink HHE. Evaluation of work-rest schedules with respect to the effects of postural workload in standing work. Ergonomics 1998;41:1832–44.


Michael W. Olson is currently an assistant professor in the department of kinesiology at Southern Illinois University Carbondale where he teaches and serves as director of the Biomechanics and Integrative Movement Laboratory. He received his B.S. in biological sciences from the University of Wisconsin-Whitewater in 1997 and his M.S. in exercise science from University of Wisconsin-La Crosse in 1999. In 2006 he earned his Ph.D. in kinesiology from Louisiana State University. His present research is aimed at studying the biomechanics of the spine and its interrelationship with the control parameters in locomotion.

Dr. Li is an associate professor and director of the Biomechanics Laboratory in the Department of Kinesiology at the Louisiana State University.

He received the BS in Physics from Peking university, MS in Biomechanics from National Institute of Sports Science of China, and PhD. in Biomechanics from University of Massachusetts.

Dr. Li’s research focus on the neuromuscular control of human movements. His publications cover topics from gait transition, dynamical systems theory, stability of gait and posture, neural control of cycling, to low back disorder.

Dr. Li is a Fellow of American College of Sports Medicine.

Dr. Moshe Solomonow is a Professor and director of the Bioengineering Division and the Musculoskeletal Disorders research Laboratory in the Department of Orthopedics at the University of Colorado Health Sciences Center in Denver. He was a Professor and Director of Bioengineering and of The Occupational Medicine Research Center at Louisiana State University Health Sciences Center in New Orleans, Louisiana from 1983 to 2005.

He received the B.S., and M.Sc. in Electrical Engineering from California State University and the Ph.D. in Engineering Systems and Neuroscience from the University of California, Los Angeles.

Under his leadership, technology was developed for several translational projects related to: Myoelectric control of upper limb prosthetics for amputees; Electronic walking orthosis for paraplegics; Smart orthosis for Anterior Cruciate Ligament deficient patients; and smart braces for individuals with low back pain.

He is the Founding Editor of The Journal of Electromyography and Kinesiology, and serves on the Editorial Board of several bioengineering and medical journals. Dr. Solomonow is/was a consultant to the National Science Foundation, National Institutes of Health, Centers for Disease Control, The Veterans Administration and scientific agencies of several European and Asian governments and Canada. He was a council member of the International Society of Electrophysiological Kinesiology, the International Society of Functional Electrical Stimulation, and the IEEE-Biomedical Engineering Society. He published over 130 refereed journal papers on musculoskeletal disorders including: motor control, Electromyography, muscle, tendon, ligament and joint Biomechanics, electrical muscle stimulation, prosthetics and orthotic systems for paraplegic locomotion, and supervised more than 150 engineering, physical therapy, physiology, neuroscience, medical students and orthopaedic residents, as well as postgraduate students and fellows from several countries.

Dr. Solomonow organized the EMG Tutorial Workshop in the ISB Congress, the Canadian Society of Biomechanics, The Human Factors and Ergonomics Society, and The Society for Clinical Movement Analysis, was on the organizing committee of numerous conferences and gave keynote and symposia lectures in many others. He received the Crump Award for Excellence in Bioengineering Research (UCLA), the Distinctive Contribution Award from Delta 7 Society (France), The Doctor Medicine Honoris Causa (Vrije Universiteit, Brussels), The I. Cahen Professorship (LSUHSC) and the 1999 Volvo Award for Low Back Pain Research.