FINAL PROGRESS REPORT

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Aging, Neuromuscular Behavior, and risk of Occupational Low back Pain

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TABLE OF **CONTENTS**

ABSTRACT

Low back disorders, and low back pain (LBP) in particular, continue to be the leading cause of worker disability in the United States, accounting for ~ 40% of all reported occupational musculoskeletal disorders annually. Reducing the incidence and severity of LBP is also a strategic goal within most sectors in the recent NIOSH National Occupational Research Agenda. Epidemiological studies have identified a number of physical, psychosocial, and personal factors that increase the risk of developing LBP. Among the latter, age is of particular significance given the increasing numbers and proportions of older workers and the higher prevalence of LBP among them. Understanding the underlying mechanisms, specifically identifying the links between aging and a higher incidence of LBP, is an essential prerequisite for developing better control and management of such disorders in the aging workforce.

Over the project period, we investigated the role of several biomechanical mechanisms potentially linking natural aging with a higher incidence of LBP. We were interested in knowing whether age-related changes in lower back tissues would adversely affects lower back biomechanics, specifically the forces and deformations experienced within the lower back under various occupational activities. To achieve our objective, we completed two specific aims: 1) quantify alterations in trunk mechanical behavior with age, and 2) quantify the association between alterations in trunk mechanical behavior and spine biomechanics. These aims were completed by studying low back biomechanics among 60 workers (30 males and 30 females), including 12 workers in each of the following age groups: 20s, 30s, 40s, 50s, and 60s. Workers were recruited to minimize the effects of factors other than natural aging on our measures of lower back biomechanics.

Trunk resistance to passive deformation and the subsequent viscoelastic relaxation were found to increase with aging while trunk intrinsic stiffness did not change. On a more functional level, the contribution of the lumbar spine to total trunk forward bending under simple flexion-extension activity as well as during symmetric manual material handing tasks was found to decrease with aging. Such functional changes with aging are likely due to underlying age-related changes in trunk mechanical behaviors and were found to adversely affect the mechanical loading on the lower back. Specifically, age-related alterations in the manner in which workers performed symmetric manual material handling were found to be associated with higher shearing forces on the worker's lower back.

Investigating the role of biomechanical factors in the occurrence of occupational low back pain is a complex activity because of the influence of many occupational and non-occupational factors on lower back biomechanics. Given the cumulative nature of many work-related musculoskeletal disorders, including low back pain, results of this project have established a baseline for our future research wherein the influence of long-term exposure to important occupational factors (e.g., repetitive task) can be effectively distinguished from the effects of natural aging.

SIGNIFICANT (KEY) FINDINGS

Lower back viscoelastic behavior
Initial resistance of the lower back to passive deformation, as well as the subsequent relaxation over a prolonged period of sustained deformed posture (i.e., flexed posture), were larger in older versus younger workers. Resistance of the lower back to passive deformation, after the viscoelastic relaxation period, was also larger among older workers, despite significantly larger dissipation of energy during the relaxation period in older workers. Most of the significant changes observed here were found to occur after 50 years of age.

Trunk intrinsic stiffness
There was no difference in measured trunk intrinsic stiffness among workers in different age groups. In general, an increase in the level of trunk muscular activity (as a percent of maximum activity) was found to increase trunk intrinsic stiffness; an increase that was not found to be different between older and younger workers. It is noteworthy that the levels of trunk muscular efforts considered in this project were associated with significantly smaller exertion levels (i.e., output force) in older workers. Trunk intrinsic stiffness was larger in male versus female workers, a difference that was not found to be affected by aging.

Lumbopelvic ratio
For a task involving sagittaly symmetric flexion-extension, no differences was found in the maximum thoracic rotation between age groups, though older workers completed the task with smaller lumbar flexion and larger pelvic rotation. Therefore, older workers demonstrated a smaller lumbopelvic ratio as compared to younger workers. The difference in lumbopelvic rhythm between male and female workers was similar to the difference between young and older workers. Compared to a normal lifting pace, workers decreased their lumbar flexion and increased their pelvic rotation when performing the task at a faster pace. Most of significant age-related changes observed here were also found to occur after 50 years of age.

Mechanical demand on the lower back
When asked to lower a 4.5 kg load, which was held in front of body, to knee height and then return to the upright standing posture, older workers completed the task by adopting a larger pelvic rotation and smaller lumbar flexion as compared to younger workers. Such age-related differences in work methods were found to result in larger shearing forces (~ 100N) at the lower back of older versus younger workers. Compared to male workers, female workers experienced a larger shearing force (~ 30 N) in their lower back when completing the same task.

TRANSLATION OF FINDINGS

Aging and lower back biomechanics
While age-related changes in the structure and function of lower back tissues have been documented in the literature, results of this project highlighted the impact of such tissue-level changes on more functional performance (during a simulated occupational task) and the subsequent effects on lower back biomechanics. The increase in shearing force with aging suggests that a given physical task (e.g., designed to be safe using NIOSH lifting guidelines) will be more demanding on the lower back tissues that resist shearing force (e.g., lumbar spine facet joints) in older versus younger workers. Task design at the workplace, therefore, should

account for the effects of such natural age-related changes in work methods on mechanical loading of the lower back.

Enabling future research

Controlling cumulative low back pain at the workplace, via reduction of the adverse effects of occupational exposure on lower back biomechanics, requires knowledge of the effects of occupational factors on lower back biomechanics over time. Given the enormous resources required to conduct longitudinal research, cross-sectional study designs will continue to be a common alternative approach wherein workers with different exposure duration are examined to investigate cumulative low back pain at the workplace from a biomechanical perspective. Findings of this project have set the stage for enhancing such cross-sectional study efforts, by providing baseline information regarding typical age-related changes in lower back mechanics. Such changes should be accounted for when investigating the cumulative effects of occupational factors on lower back biomechanics. Without the baseline information generated in this project, these cross-sectional study efforts would be limited by confounding of natural aging (as a personal factor) and occupational factors on biomechanics of the lower back. By offering important information related to the accumulation of biomechanical disturbance over the course of worker's life, such future studies will enable effective administrative and engineering interventions aimed at preventing cumulative biomechanical disturbances.

OUTCOME/IMPACT

Potential outcomes

- With aging, workers rely more on their pelvis to perform tasks that involve trunk flexion, thus putting more shearing force on their lower back when compared to younger workers. Therefore, a reduction in the amount of flexion required to complete such tasks (e.g., elevating the lifting surface) is recommended for workers over 50 years old.

- Female workers may also benefit from interventions that reduce the amount of trunk flexion, because female workers, similar to older workers, rely more on their pelvis to complete these tasks.

SCIENTIFIC REPORT

1. Background

Low back pain (LBP) is a multifactorial disorder, with diverse occupational and non-occupational risk factors(Pope *et al.* 1991, Garg and Moore 1992, NIOSH 1997, Pope *et al.* 2002). Among identified risk factors for LBP, age is becoming increasingly important due largely to the increasing population of older workers in the US. Nearly a fifth of Americans over 55 (total = ~70 million) were in the workforce in 2008, with projections of 25% (total = ~92 million) by 2018(Toosi 2009). LBP prevalence, however, increases substantially with age(Cypress 1983, Hart *et al.* 1995, Bressler *et al.* 1999, Hoy *et al.* 2010, Johannes *et al.* 2010). Understanding the underlying mechanism(s) linking aging with a higher incidence of LBP is essential for better control and management of this disorder in the aging workforce.

Abnormal spine mechanics and LBP

Though most LBP cases remain idiopathic(White and Panjabi 1990, Bogduk *et al.* 1992), it is widely accepted that abnormal mechanics of the spinal column often results in LBP(White and Panjabi 1990, Bogduk and Twomey 1991, Kumar 2001, Adams *et al.* 2006, Panjabi 2006, McGill 2007) through stimulation of trunk nociceptors. In this context, and for a given occupational task/event (e.g., lifting/slip), abnormal mechanics means ***higher*** spinal loads and ***lower*** spinal stability; both of the latter can cause stress/strain within trunk tissues that exceeds the thresholds of trunk nociceptors(White and Panjabi 1990). Spine mechanics depend on the physical demands of the task/event, passive mechanical trunk properties, and more importantly on active mechanical trunk responses (i.e., via muscle forces) to equilibrium and stability requirements. By active mechanical responses we mean both feedforward (i.e., voluntary and reflexive responses) and feedback (i.e., proprioception and kinesthesia) mechanical responses of the neuromuscular system. While both types of active responses are important, these four components of mechanical trunk behaviors are not equally involved in different tasks/events. For example, voluntarily muscle activation (force) is critical during lifting, vs. proprioception (sensory) and reflexive muscle responses (forces) in a balance perturbation such as a slip or trip.

Research needs

Validity of causal biomechanical theory (§1.2) has not yet been consistently established. With respect to aging, the biomechanical theory suggests (Fig. 1) that age-related changes in active and passive trunk mechanical behaviors **(TMB)**[1] adversely affect spine biomechanics.

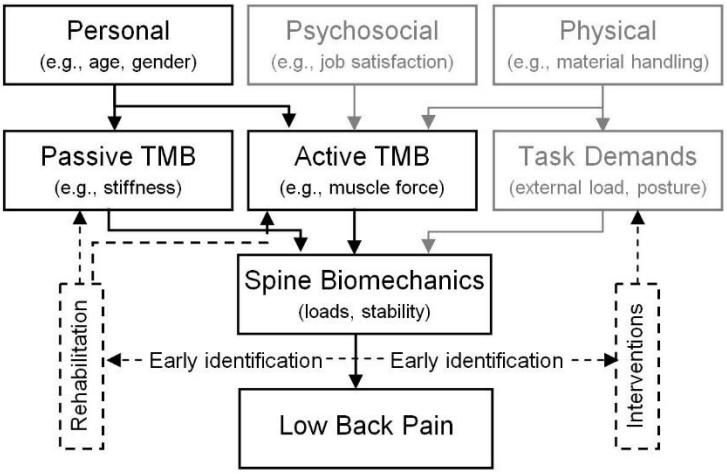

Figure 1. Causal biomechanical pathways (solid black lines) through which compromised trunk mechanical behaviors (TMB) links aging with LBP. Potential implications (gray dashed lines) are also shown.

[1] Throughout this report, by **trunk mechanical behaviors (TMB)** we mean both passive mechanical properties and active mechanical responses at the whole trunk level and not at the cell, tissue, or individual muscle levels

But this is a complex situation, in that both active and passive TMB are influenced by the ***collective*** properties and behaviors of trunk tissues (e.g., muscles, discs, ligaments, and vertebrae). Aging changes the composition, mechanical properties, and behavior of these tissues(Campbell *et al.* 1973, Lexell *et al.* 1983, Edstrom and Larsson 1987, Lexell *et al.* 1988, Becker *et al.* 1994, Evans 1995, Lexell 1995, Adams *et al.* 1996, Umehara *et al.* 1996, Hunter *et al.* 2000, Kent-Braun *et al.* 2000, Klein *et al.* 2003, Adams *et al.* 2006, Chaffin *et al.* 2006, Faulkner *et al.* 2007), though largely unknown are the resultant effects on TMB (e.g., passive stiffness and damping, intrinsic and reflexive mechanical behaviors). Earlier work has reported age-related alterations in trunk behaviors such as strength, fatigability, and range of motion(Voorbij and Steenbekkers 2001, Yassierli *et al.* 2007, Intolo *et al.* 2009). These studies, though, were mostly focused on measures of physical capability in extreme static conditions (maximum isometric contraction, endurance, etc.) and on comparisons between extreme age groups. Whether such alterations exist in measures of TMB and in more relevant occupational activities/events (e.g., lifting or balance perturbations) remain to be investigated. Evidence on age-related changes in trunk strength(Voorbij and Steenbekkers 2001) and range of motion(Intolo *et al.* 2009) suggests a nonlinear relationship between aging and alterations in TMB, though the actual progression of TMB with age is also unclear at present. Finally, the effects of age-related alterations in TMB on spine biomechanics are unknown as well.

Addressing these research gaps, our main objective was to investigate age-related changes in TMB and spine biomechanics. Since other LBP risk factors(Pope *et al.* 1991, Garg and Moore 1992, NIOSH 1997, Pope *et al.* 2002) (i.e., physical, psychosocial, and other personal factors) can also be related to LBP through their influences on TMB, task demands, and spine biomechanics (Fig. 1), this investigation focused on individuals with minimal exposure to LBP risk factors *other than age*. Age-related degradations in the load tolerance of trunk tissues may not necessarily impose a risk of biomechanical LBP, particularly if, for a given task/event and due to alterations in TMB, the resultant spinal load decreases and stability increases. According to the causal biomechanical theory, if disturbed TMB are (at least in part) responsible for occupational LBP, one would deduce **(Hypothesis 1)** that disturbances in TMB increase with age, potentially nonlinearly, and **(Hypothesis 2)** that such disturbances in TMB adversely affect spine biomechanics. Disturbances in TMB that could adversely affect spine biomechanics may include a stiffer trunk, reduced damping, increased reflex latency, and decreased reflex response (force).

Implications

Consistent with major NIOSH research objectives, this application sought to identify and investigate the relationships between a risk factor and associated occupational morbidity, using a new set of comprehensive assessment tools that have potential for diverse future applications. The growing population of older workers and the higher incidence of LBP with age highlight the importance of age in the development of occupational LBP. Clearly, a better understanding of potential causal mechanisms is not only critical to prevention (whether proactive or reactive), but also important for efficient rehabilitation and safe return-to-work of workers with occupational LBP. By quantifying age-related alterations in TMB and the resultant effects on spine biomechanics, we expected to establish a foundation from which to develop and implement age-appropriate controls for LBP. Achieving this required systematic evaluation of relationships between risk factors and LBP via available causal theories, and unraveling the relative magnitudes and mechanisms by which each risk factor contributes to occupational LBP. As such, completing the current work should open a new line of research to promote/advocate changes in work design and the work environment, and can thereby play a critical role in managing and controlling work-related LBP in our aging worker population. Important implications of our current and future work are that (cf. Fig. 1): 1) workplace interventions could

focus on controlling aspects of task demands to accommodate changes in TMB among older workers; 2) preventive or rehabilitation strategies might benefit from targeting both active and passive aspects of TMB; and 3) at-risk, but otherwise asymptomatic workers, may be identified by comprehensive evaluation of TMB and spine biomechanics and can be assisted, through changes in their working environment and behaviors, in avoiding LBP initiation.

Furthermore, the emphasis here was on aspects of the causal biomechanical theory from risk factor to abnormal mechanics, specifically influences of TMB via spine biomechanics. These aspects have not been explored previously, either in occupational or in clinical investigations. Hence, outcomes of this work can also promote future basic and clinical research on other dimensions of the causal biomechanical theory including paths relating anatomical/physiological changes to TMB alterations and abnormal mechanics to pain sensation (e.g., inflammation, biochemical and nutritional changes, immunological factors, and changes in the structure and material of endplates, discs, and neural structures).

INNOVATION

We have developed a new set of powerful tools for the comprehensive assessment of TMB and resultant spine biomechanics. These enabled us to examine, for the first time, the potential role of altered TMB in the biomechanical pathway linking aging with occupational LBP. Passive trunk stiffness and damping will be quantified using a new passive stress-relaxation test coupled with a viscoelastic model of the trunk(Toosizadeh *et al.* 2010). Measures of trunk mechanical impedance (dynamic properties) were quantified using our newly-developed system; this includes a displacement-controlled sudden perturbation device(Bazrgari *et al.* 2011a, Bazrgari *et al.* 2011b, Hendershot *et al.* 2011a) coupled with an electromyography (EMG)-based(Granata *et al.* 2005, Moorhouse 2005, Rogers and Granata 2006, Granata and Rogers 2007, Moorhouse and Granata 2007) and our new mechanical-based(Hendershot *et al.* 2011b) system identification procedures. The unique advantage of our displacement-controlled device, over the traditional force-controlled approach, is that it allowed us to quantify the relative contributions of intrinsic and reflexive aspects of TMB in trunk mechanical impedance(Bazrgari *et al.* 2011a, Hendershot *et al.* 2011a). In contrast to traditional EMG-based system identification methods, our new mechanical-based method(Hendershot *et al.* 2011b) does not rely only on data from accessible trunk muscles (using EMG) to quantify reflexive TMB. Rather, by including kinematics and kinetics it accounted for any contributions of deeper trunk muscles (i.e., not accessible to EMG) to reflexive TMB. Further, muscular responses to task/event demands, resultant spinal load (i.e., compression and shear forces at all lumbar levels), and spinal stability are all quantified with a validated finite element model of the human trunk. As such, this study was innovative in implementing an effective suite of assessment tools, and takes some important steps forward by: **1)** assessing whether age-related changes in trunk tissues can have accumulative or chronic effects on TMB; and **2)** providing new evidence regarding whether such altered TMB adversely affect spine biomechanics.

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2. Specific Aims

Linking aging with LBP via causal biomechanical theory suggests that age-related changes in trunk mechanical behavior adversely affect spine biomechanics. The aim of this application was to explore such a relationship between aging and spine biomechanics, based on hypotheses stemming from the biomechanical theory and using our new trunk mechanical behavior and spine biomechanics assessment tools. Accordingly, we proposed two specific aims to evaluate trunk mechanical behavior and spine biomechanics. It was hypothesized that: **(1)** aging is associated with increased disturbances in trunk mechanical behavior, potentially nonlinearly; and **(2)** such age-related disturbances lead to adverse effects on spine biomechanics.

Specific Aim 1: Quantify alterations in trunk mechanical behavior with age. Multiple measures of trunk mechanical behavior were proposed to be obtained from five worker groups differing in age (n_{total}=60). As supported by our preliminary data, disturbances in trunk mechanical behavior were expected to increase with age. Measures of disturbed trunk mechanical behavior were expected to be less effective active trunk mechanical behavior (e.g., reduced reflex responses) and disturbed passive trunk mechanical behavior (e.g., stiffer trunk).

Specific Aim 2: Quantify the association between disturbed trunk mechanical behavior and spine biomechanics. Spinal load and stability during sudden loading and manual material handling tasks were proposed to be quantified using our trunk model. Using data from Aim 1, model-based estimates of spine loading and stability were suggested to be then correlated with measures of trunk mechanical behavior. With disturbed trunk mechanical behavior (e.g., stiffer trunk and weaker reflex force) we expected a corresponding “poorer” spine biomechanics (higher spinal loads and lower stability). Comparisons were proposed to be made using all participants as well as by age and gender groups; the latter may indicate age- and gender-related differences in the level of association.

3. Specific Aim 1: Quantify alterations in trunk mechanical behavior with age

Status: Data collection has been completed but data analyses and publication are still ongoing. This section include our findings related to Aim 1.

Milad Vazirian, Iman Shojaei, Rebecca L. Tromp, Maury A. Nussbaum, Babak Bazrgari: Age and gender differences in trunk intrinsic stiffness. *First revision under review by Journal of Biomechanics*.

Abstract
Age-related differences in trunk intrinsic stiffness, as an important potential contributor to spinal stability, were investigated here given 1) the role of spinal instability in low back pain (LBP) development; 2) the increasing prevalence of LBP with age, and 3) the increasing population of older people in the workforce. Sixty individuals aged 20 - 70 years, in five equal-size age groups, completed a series of displacement-controlled perturbation tests in an upright standing posture while holding four different levels of trunk extension efforts. In addition to examining any age and gender-related difference in trunk intrinsic stiffness, the current design assessed effects of the level of effort and any differences in lower back neuromuscular patterns on trunk intrinsic stiffness. No significant differences in trunk intrinsic stiffness were found between the age groups. However, stiffness was significantly larger among males and increased with the level of extension effort. No influences of differences in neuromuscular pattern was observed. Since the passive contribution of trunk tissues in the upright standing posture is minimal, our estimated values of trunk intrinsic stiffness primarily represent the volitional contribution of the lower back musculoskeletal system to spinal stability. Therefore, it seems unlikely that the alterations in volitional behavior of the lower back musculature caused by aging (e.g., as reflected in reduced strength) deteriorates the spinal stability.

Introduction
Low back pain (LBP) remains a major health problem, with a high prevalence and burden throughout the world (Deyo et al., 2006). The prevalence of LBP is particularly higher among the elderly (Johannes et al., 2010). Given that the relative population of older individuals is increasing in industrialized societies (Toossi, 2012), and considering the significance of healthcare costs associated with LBP (Katz, 2006), it becomes increasingly important to investigate the reason(s) underlying the rise in the prevalence of LBP with aging.

While the specific causes of most LBP cases cannot typically be identified, spinal instability has been considered as an important casual mechanism (Panjabi, 2003). Spinal instability is clinically diagnosed by observation of relatively large intervertebral angular/translational motion while performing normal range-of-motion activities (Dupuis et al., 1985, Biely, 2006, Izzo et al., 2013, Izzo et al., 2013). Biomechanically, though, spinal instability can be defined as the failure of the lower back musculoskeletal system to maintain the spine around its equilibrium condition (static or dynamic) following a perturbation. Excessive post-perturbation deviation from the equilibrium condition (i.e., spinal instability) may expose the spine and its surrounding tissues to forces and deformations that can cause injury and/or irritate (directly or indirectly) nociceptors within the lower back tissues, thereby leading to LBP.

Stabilizing the spine is a complex task, and is implemented by coordination of three subsystems of the lower back musculoskeletal system: 1) the passive subsystem, including the spine and passive muscle forces; 2) the active subsystem (i.e., volitional and reflexive muscle forces); and

3) the neural control subsystem (Panjabi, 1992). The ability of the lower back musculoskeletal system to maintain spinal stability has been investigated using diverse sudden-loading or perturbation testing methods (Krajcarski et al., 1999, Cholewicki et al., 2000, Stokes et al., 2000, Andersen et al., 2004). In these, stabilization capability is often assessed by derivation of a measure of trunk stiffness using simple lumped-parameter models (Cholewicki et al., 2000, Lee et al., 2006). The effects of LBP (Stokes et al., 2006, Hodges et al., 2009), co-activation of antagonistic muscles (Lee et al., 2006, Shahvarpour et al., 2014), and pre-perturbation muscular efforts (Lee et al., 2007, Shahvarpour et al., 2014) on such measures of trunk stiffness have been reported. A limitation of these earlier studies, though, was the lack of control over the trunk displacement and response time. As such, measures of trunk stiffness represented the combined effects of all subsystems, without information about relative contributions of each subsystem. This limitation was later overcome by use of displacement-controlled perturbation paradigms that allowed separate evaluation of each of the three stabilizing subsystems (Moorhouse and Granata, 2007). In particular, the paradigm of displacement-controlled perturbations minimizes alterations in the contribution of the passive subsystem, by minimizing the post-perturbation trunk displacement to a relatively small range. It thus enables distinguishing volitional from reflexive responses, by controlling the response time such that it is shorter than the minimum reflex delay (Moorhouse and Granata, 2007).

Displacement-controlled perturbation tests have been utilized in several recent studies to evaluate the effects of prolonged trunk flexion (Bazrgari et al., 2011, Hendershot et al., 2011), repeated static trunk flexion and repeated lifting (Muslim et al., 2013), and exercise-induced LBP (Miller et al., 2013) on the spine stabilizing subsystems. Recently, we have used the same displacement-controlled perturbation method to study age-related differences in the stabilizing capabilities of the lower back musculoskeletal system. In this paper, we present results related to age-related differences in the trunk intrinsic stiffness. These were calculated by relating measured trunk kinematics and kinetics during the latency period of lower back muscles responses to perturbations. This approach thus represents the combined contribution of passive and volitional active subsystems to spine stability. However, estimates of trunk intrinsic stiffness in an upright standing posture, as done here, primarily represent the active volitional contribution of the neuromuscular system, given the minimal passive contributions in such a posture.

The mechanical behaviors of the passive and active tissues alter with aging, suggesting an age-related alteration in the lumbar spine stability and the associated risk of LBP. With aging, the ligaments become slack in the neutral posture (Iida et al., 2002), intervertebral discs become stiffer due to disc degeneration (Galbusera et al., 2014), and the capability of lower back extensors to generate active volitional force reduces (Yassierli et al., 2007). With this diverging evidence, it is difficult to make an explicit hypothesis regarding expected age-related changes in the trunk intrinsic stiffness. However, given the higher incidence of LBP among older versus younger individuals (Johannes et al., 2010), a reduction in the level of spinal stability with aging, hence in the trunk intrinsic stiffness, was hypothesized.

Methods

Participants

Sixty asymptomatic individuals, in five equal size and gender-balanced age groups (22-28, 32-38, 42-48, 52-58 and 62-68 years old), participated in this study after completing a consenting procedure approved by the Institutional Review Board of the University of Kentucky. Exclusion criteria were any history of work in an occupation with substantial exposure to physical risk factors for LBP, a recent (i.e., past 12 months) history of LBP that resulted in missing a workday or visiting a doctor, any neurological and musculoskeletal disorders related to the lower back,

and a body mass index outside of the 22 to 30 range. Consented volunteers were screened by a nurse to ensure their medical and physical eligibility for participation before experimental data collection. A summary of the sample is provided in Table 1. Univariate analysis of variance (ANOVA) indicated no significant differences in stature (p = 0.851) or body mass (p = 0.116) between the five age groups.

Table 1: Numbers of participants and associate anthropometry in each of five age groups. Summary values are means (SDs).

	Age Groups (years)				
	22-28	32-38	42-48	52-58	62-68
Number	6 M, 6 F	6 M, 6 F	6 M, 6 F	6 M, 5 F	6 M, 6 F
Stature (cm)	171 (8.6)	170 (6.6)	173 (8.7)	172 (11.8)	172 (10.2)
Body mass (kg)	70.0 (10.4)	73.0 (13.2)	79.3 (14.6)	78.1 (12.2)	73.7 (15.5)

Experimental Procedure

To enhance the reliability of our results, each participant completed two identical data collection sessions, which were conducted in the morning to minimize the influence of diurnal changes in lower back biomechanics (Adams M. A., 2006). During each session, each participant completed two maximum isometric voluntary exertions (MVEs) of the trunk extension and four perturbation tests. Similar to the method used previously (Bazrgari et al., 2011, Hendershot et al., 2011, Bazrgari et al., 2012, Hendershot et al., 2013, Muslim et al., 2013), participants stood in a custom fixture, were restrained at the pelvis using straps, and were attached to a servomotor (Kollmorgen, Radford, VA) via a harness-rigid rod assembly (Figure 1). MVE trials were done by locking the servomotor and instructing the participants to pull back against the harness as hard as they could for ~ 5 seconds. Isometric trunk exertion was measured using an in-line load cell (Interface SM2000, Scottsdale, AZ) that was mounted on the rigid rod. Electromyographic (EMG) activity of several bilateral muscles was collected using adhesive electrodes (DE-2.1 Differential EMG sensor, Delsys, Natick, MA). These muscle included the bilateral erector spinae at the L3 and the L5 levels (respectively located 2 inches and 1 inch medio-laterally on both sides of the midline of the participant's back), rectus abdominis (located 1 inch medio-laterally and 1 inch caudo-cephalic on both sides of the participant's naval), and external oblique (located on intersections of a horizontal line passing the navel and a vertical line passing the left/ right anterior superior iliac spines of the participant).

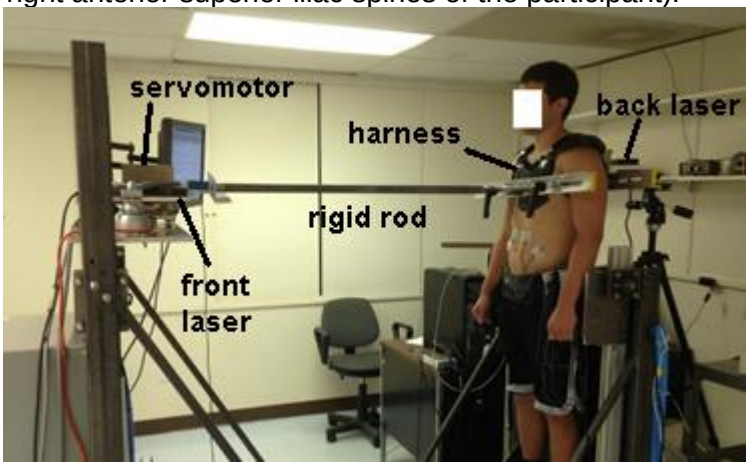

Figure 1: The setup used for the perturbation tests, including the metal frame, pelvis restraining belt, servomotor, rigid rod, harness, laser sensors, and load cell.

During perturbation tests, participants were exposed to a pseudo-random sequence of anterior-posterior position perturbations (±5mm) that were generated by the servomotor, and transferred to the participant's trunk via the harness-rigid rod assembly. In separate perturbation tests, participants were instructed to maintain an extension effort of both 20% and 30% of the mean maximum values obtained during MVEs. Two effort levels were included to investigate how different levels of volitional extension influence the trunk intrinsic stiffness among age groups. These levels of volitional extension effort were maintained using real-time visual feedback provided both from back muscle EMG activity (i.e., mean across the four erector spinae channels) and from the in-line load cell. Any age-related difference in the estimated trunk intrinsic stiffness between conditions with EMG versus force feedback was expected to be due to potential age related differences in neuromuscular pattern adopted for trunk extension (e.g., differences in abdominal co-activation). For instance, cases with versus without presence of abdominal co-activation during trunk extension would require higher activity in extensor muscles to generate the same level of effort. The use of two levels of volitional effort and types of visual feedback led to four testing conditions, and were completed in a random order. In the first session, participants received a series of perturbations to get accustomed to the sudden loading and learn how to control their exertion levels using visual feedbacks.

Trunk displacements during perturbation tests were measured using two laser displacement sensors (Optex FA, Kyoto, Japan), one targeting the back of harness at the T8 level and the other the in-line load cell. Force and displacement measurements were filtered using a 4th order, bi-directional, Butterworth filter with a cutoff frequency of 10 Hz. Previous experience had shown the appropriateness of such a filter (Bazrgari et al., 2012). Raw EMG data were bandpass filtered (20Hz-500Hz) and then rectified and lowpass filtered (3Hz) to create a linear envelope.

Data Analysis

A system identification approach was used to estimate trunk intrinsic stiffness and apparent mass, involving a two degree-of-freedom spring-mass-damper system to model the trunk and harness-rigid rod assembly (Figure 2) (Bazrgari et al., 2012). The mass of the harness-rigid rod assembly was 6.5 kg, and based on the results of a previous study (Hendershot et al., 2011), trunk damping was set to zero for enhanced accuracy of the system identification procedure.

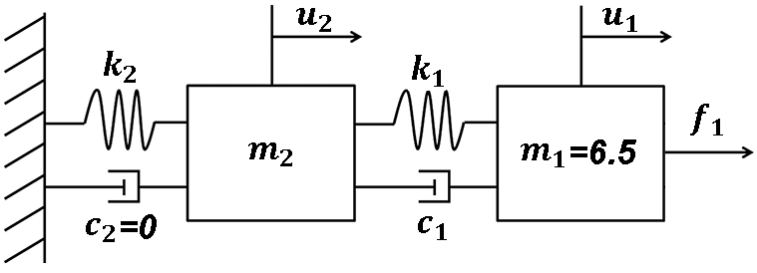

Figure 2: The spring-mass-damper model used to model perturbations. Parameters $m_1(m_2)$, k_1 (k_2) and c_1 (c_2) respectively denote apparent mass, stiffness, damping of the harness-rigid rod assembly (the participant's trunk), $u_1$and $u_2$respectively denote the displacement of the in-line load cell and the trunk, and $f_1$denotes the force applied externally to the system (via the load cell).

Due to the design of our experiment (i.e. requiring participants to hold a level of extension exertion during the perturbation test), the system identification was conducted only for perturbations causing trunk flexion (i.e. when the servomotor pulled the participant anteriorly). Backward perturbations were confounded by a change in the contact area between the harness and the trunk (i.e. from the back of the trunk due to participant's extension effort to the front of the trunk due to posterior push from the servomotor) during the perturbation. For each anterior perturbation in the sequence of perturbations (total of eight), the unknown parameters of the model were estimated using the measured trunk kinematic and kinetics collected during that specific perturbation (Bazrgari et al., 2012).

Though a time window of 1 - 3s was allocated between consecutive perturbations, some participants had difficulty adjusting their level of effort to the targeted level during small time windows. Further, feedback was given based on the mean of the two maximum values (MVE_{max}) of recorded EMG/force during MVE trails of each session. These two values were found to be significantly (p<0.001) different in the first session, and the difference approaches significance in the second session (p=0.076); in both sessions the MVE_{max} of second trial was larger than the first one. However, there was no differences in the second obtained MVE_{max} between sessions (p=0.505). Thus, only the second MVE trial of each session was considered for remaining analyses, including EMG normalization and determination of the level of effort. Specifically for the level of effort, and to assure a correspondence of the level of effort with the predicted mechanical properties, predictions from all perturbations, conditions, and sessions of each participant (i.e. 64 discrete perturbations) were pooled together, disregarding the initial categorizations for level of effort, and then were grouped based on the measured level of exertion during a time window of ~60 ms prior to each perturbation. Four new groups for effort level (two EMG-based and two force-based) were formed, each including perturbations for which the mean level of exertion effort during that time window was within 20% ±4% or 30% ±4% of the second MVEmax of each session (based on the load cell or mean EMG).

Statistical Analysis

The dependent variables analyzed were: 1) the maximum recorded values of MVE efforts by the load-cell (MVE_{max}), 2) the difference in exerted extension forces between 20% and 30% MVE_{max} effort conditions (when guided by force feedback) 3) the estimated trunk intrinsic stiffness (K_{int}), and 4) the estimated trunk apparent mass normalized to whole-body mass (M_{app}). Analysis of variance (ANOVA) was used to investigate the effects of age and gender on MVE_{max} as well as on difference in extension force between 20% and 30% MVE_{max} effort conditions. Two separate, mixed-factors ANOVAs were also used to assess the effects of age and gender, as between-subject variables, and feedback type (i.e. EMG or force) and level of effort (i.e. 20% or 30%), as the within-subject variables, on K_{int} and M_{app}. Post hoc comparisons were done, as relevant, using Tukey's HSD. All statistical procedures were conducted in SPSS (IBM SPSS Statistics 22, Armonk, NY, USA), and in all cases a p value ≤ 0.05 was considered as statistically significant.

Results

Maximal and sub-maximal extension efforts

Mean values of MVE_{max} were significantly lower among older versus young individuals (F= 3.914, p=0.007) (Figure 3A). They were also significantly lower among female versus male participants (F= 20.483, p<0.001), with respective means (SD) of 435 (171) N and 667 (257) N. The difference in exerted extension forces between 20% and 30% MVE_{max} effort conditions (when guided by force feedback) were significantly lower among older individuals (F= 6.082, p<0.001,) (Figure 3B). This difference was also lower among female versus male participants (F= 21.907, p<0.001), with respective means (SD) of 42 (19) N and 66 (26) N.

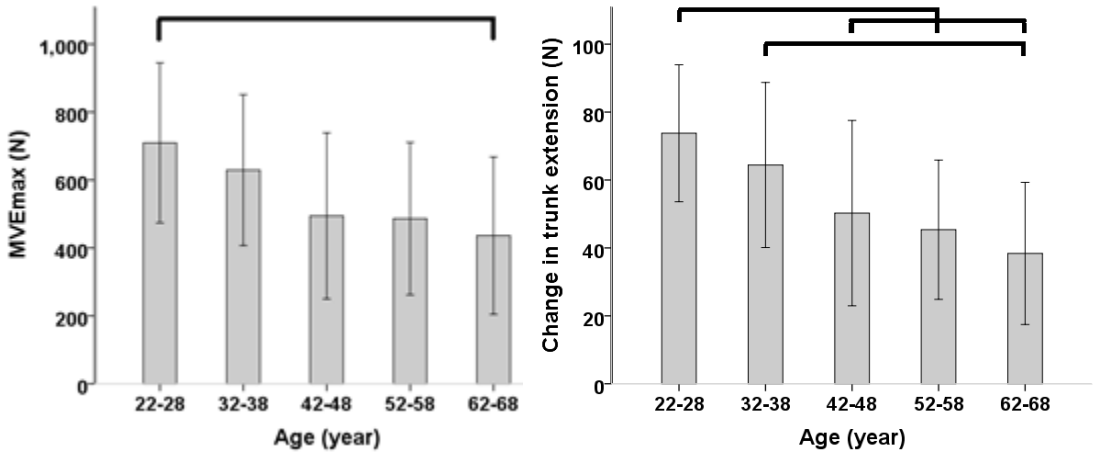

Figure 3: A) Mean maximum recorded forces during the MVE trials (MVE_{max}), and B) the increase in extension forces from conditions with 20% to 30% of MVE_{max} efforts with force feedback

Trunk intrinsic stiffness - K_{int}

There was no significant difference in estimated values of K_{int} between age groups, (F=0.47, p=0.755). However, both gender (F=31.698, p<0.001) and the level of effort (F=26.57, p<0.001) had significant influences on K_{int}, Males had a larger K_{int} in all four feedback conditions, and K_{int} was larger with the 30% effort level for both feedback types (Figure 4A). Neither age nor gender had significant interactions with the type of feedback (F=1.23, p=0.314) or the level of preload (F=1.46, p=0.231).

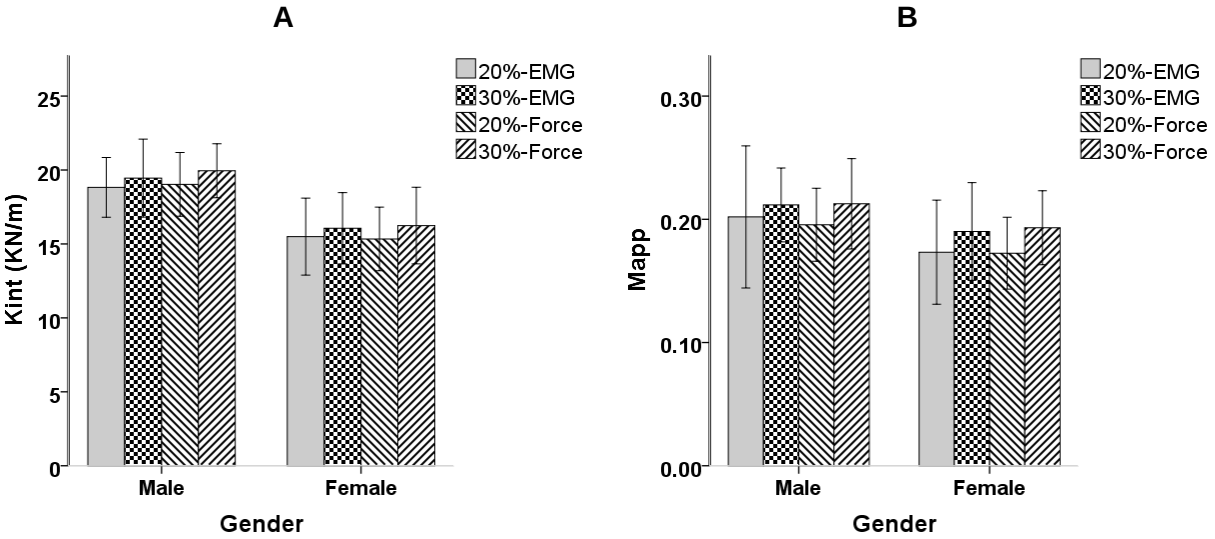

Figure 4: Mean (SD) A) estimated trunk intrinsic stiffness (K_{int}) and B) normalized apparent mass, across genders and the level of extension efforts. The level of extension efforts were 20% and 30% of MVE_{max} values and were guided either by EMG (i.e., 20%-EMG, 30%-EMG) or force (20%-force, 30%-force) feedback.

Normalized apparent Mass - M_{app}

While there was no difference in estimated values of M_{app} between age groups (F=1.96, p=0.117), there were significant differences between genders (F=6.84, p= 0.012), with males having a larger M_{app} (Figure 4B). M_{app} was also larger with the 30% effort level for both feedback types (Figure 4B).

Discussion

The purpose of this study was to assess whether there are age-related differences in trunk intrinsic stiffness, given the role of such stiffness as a contributor to spinal stability (Bergmark, 1989). Considering the reported causal role of spinal instability for LBP, it was hypothesized that trunk intrinsic stiffness would be lower in older versus younger individuals. However, our hypothesis was not supported because no differences were found in estimated trunk intrinsic stiffness between groups of participants ranging from ~20 to ~70 years old. Trunk intrinsic stiffness was larger among males versus females and increased with 10% increase in extension effort (regardless of type of feedback). Lack of interaction between age/gender and level of effort on trunk intrinsic stiffness suggests that such significant increase in trunk intrinsic stiffness with 10% increase in extension effort to be comparable across all ages/genders. However, the associated increase in output extension force with a 10% increase in the effort was found to be significantly different between age groups and genders. In other words, the increase in level of effort from 20% to 30% of an individual's maximum effort, in spite of being associated with significantly less addition to the exerted trunk extension forces in older versus younger participants, and among females versus males, provided the same level of increase in trunk intrinsic stiffness across all age and gender levels. Further, a lack of interaction between age/gender and type of feedback on trunk intrinsic stiffness suggests no age/gender-related differences in the adopted neuromuscular pattern to generate the extension effort. Therefore, our results as a whole suggest that an age-related deterioration of the spinal stability due to a decrease in volitional contribution of neuromuscular system to the trunk intrinsic stiffness is not likely.

The prevalence of LBP has been shown to be higher among the elderly (Johannes et al., 2010), a cohort whose participation at work place is increasing (Toossi, 2012). Aging changes the composition, mechanical properties, and behavior of tissues within the trunk (Lexell et al., 1983, Edstrom and Larsson, 1987, Hunter et al., 2000, Kent-Braun et al., 2000). Age-related changes in the neuromuscular system include reduced numbers of active motor units, decreased motor unit firing rates, fewer muscle fibers, and smaller fiber sizes (Campbell et al., 1973, Lexell et al., 1988, Evans, 1995). At the functional level, aging is associated with reduced strength and fatigability (Yassierli et al., 2007) of trunk muscles. The observed lower values of MVE_{max} among older individuals in our study is consistent with earlier reports of reduced strength (Yassierli et al., 2007) that were also evaluated using maximum voluntary exertion tests. The relative masses of bones and muscles, compared to other tissues, has been reported to decrease with age (Adams M. A., 2006, Faulkner et al., 2007). Such age-related differences in the relative composition of lower back tissues, however, was not associated with any significant changes in the inertial response of the trunk to displacement perturbations, as was reflected in the current measure of apparent mass; suggesting similar inertial response from tissues (e.g., fat) replacing muscle and bone with aging.

Trunk intrinsic stiffness, estimated here using perturbations in an upright standing posture, primarily represents the active volitional contribution of the lower back musculoskeletal system to spinal stability. Another factor that affects spinal stability, but has not been considered equally in the literature, is the trunk apparent mass. A higher trunk apparent mass is associated with a higher inertial force, and exacerbates the destabilizing effect of any displacement perturbation. Our results indicated a larger normalized trunk apparent mass among males versus females, as

well as in conditions with 30% versus 20% of MVE_{max} effort. These results highlight a potential destabilizing role of trunk inertia for individuals/conditions that are associated with larger trunk intrinsic stiffness.

While generalizability was somewhat reduced by our rather restrictive inclusion/exclusion criteria, such control was needed to enhance our ability to evaluate the hypotheses and draw conclusions about the mechanics of back pain and potential and preventable etiologies of LBP. Future studies can apply the current methods to a broader range of occupational risk factors and types of LBP. Moreover, there are age-related changes in other elements within the lower back musculoskeletal system that also contribute to spinal stability, but these contributions were not included in our measure of trunk intrinsic stiffness. With aging, the central part of the intervertebral disc becomes dry, fibrous, and stiff (Umehara et al., 1996), while tendon and ligaments become weaker (Becker et al., 1994). These changes suggest an alteration in the role of the passive subsystem in spinal stability. Similarly, the contribution of the neural control subsystem to spinal stability is likely to change because of age-related changes in ligament behaviors. Alterations in the mechanical behavior of spinal ligaments can result in sensory-motor disturbances (Solomonow, 2006). For instance, creep deformation of spinal ligaments is associated with delayed and reduced stretch-reflex responses of trunk muscles, which is an important contributor to spinal stability (Moorhouse and Granata, 2007). Finally, modeling of the lower back as a single degree-of-freedom system is a modeling limitation imposed by experimental limitations, related to the measurement of trunk kinematics and kinetics during sudden perturbation experiments. Integrating advances in imaging technology and finite element modeling of the lower back (Shojaei et al., 2015), future research may be able to offer more accurate estimates of the contributions of lower back neuromuscular systems to spinal stability.

In conclusion, active voluntary contributions of the lower back neuromuscular system and apparent mass (i.e., the two contributing elements to spinal stability) were found to be consistent across a range of participant ages. Therefore, spinal instability may be less likely to be responsible for a higher prevalence of LBP in older people. The role of other contributing elements to spinal stability (i.e., passive tissue and active reflexive) as well as equilibrium-based parameters (e.g., compression and shear forces under various activities) should be investigated in future work.

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4. Specific Aim 2: Quantify the association between disturbed trunk mechanical behavior and spine biomechanics

Status: Data collection has been completed but data analyses and publications are still ongoing. This section includes our findings related to Aim 2.

Iman Shojaei, Milad Vazirian, Emily Croft, Maury A. Nussbaum, Babak Bazrgari. Age related differences in mechanical demands imposed on the lower back by manual material handling tasks. *First revision under review by Journal of Biomechanics*.

Abstract
The prevalence of low back pain (LBP) increases with age, yet the underlying mechanism(s) responsible for this remains unclear. To explore the role of biomechanical factors, we investigated age-related differences in lower-back biomechanics during sagittally-symmetric simulated manual material handling tasks. For each task, trunk kinematics and mechanical demand on the lower back were examined, from among 60 participants within five equal-sized and gender-balanced age groups spanning from 20 to 70 years old. The tasks involved lowering a 4.5 kg load from an upright standing posture to both knee height and a fixed height and then lifting the load back to the initial upright posture. During these tasks, segmental body kinematics and ground reaction forces were collected using wireless inertial measurement units and a force platform, respectively. Overall, older participants completed the tasks with larger pelvic rotation and smaller lumbar flexion. Such adopted trunk kinematics resulted in larger peak shearing demand at the lower back in older vs. younger participants. These results suggest that older individuals may be at a higher risk for developing lower back pain when completing similar manual material handling tasks, consistent with epidemiological evidence for higher risks of occupational low back pain among this cohort.

Introduction
Low back pain (LBP) is the leading cause of disability globally; ahead of 290 other health-related conditions (Buchbinder et al., 2013). Occupationally, 28% of workers develop disabling LBP at some time, with 8% being disabled each year (Manchikanti, 2000), and disabling work-related LBP accounts for ~ 40% of all lost work days in US industries (Manchikanti, 2000). Healthcare costs related to LBP are also substantial, with annual estimates of ~ $100 billion in the U.S. (Katz, 2006). Accordingly, investigating the underlying mechanism(s) responsible for LBP at workplace is essential.

Diverse occupational and non-occupational risk factors have been recognized for LBP (Manchikanti, 2000; Marras, 2000). Among these risk factors, manual material handling (MMH) appears strongly associated with the occurrence of occupational LBP (Garg and Moore, 1991; Hoy et al., 2010; Marras, 2000). Given the physically demanding nature of many MMH tasks, biomechanical loading of the lower back during MMH has been extensively investigated to understand how different MMH parameters may contribute to development of occupational LBP, such as lifting method (Bazrgari et al., 2007), symmetry vs. asymmetry (Arjmand et al., 2010; DeVita et al., 1991; Marras and Davis, 1998), and task dynamics (Buseck et al., 1988; Marras et al., 1993). Convergent findings from these studies and others suggest that excessive mechanical demands on lower back tissues during MMH tasks are likely involved in the causal pathway leading to occupational LBP. However, most reported studies involved younger participants and/or workers, yielding results that may not be generalizable to an older population.

The population of most countries is aging, however, an increasing number of older individuals is remaining in the workforce. In the U.S., for example, nearly a fifth of individuals over 55 (total = ~70 million) were in the workforce in 2008, with projections of 25% (total = ~92 million) by 2018 (Toossi, 2009). Of concern is that the prevalence of LBP increases with age, with reported annual rates of 38% and 12% for older and younger populations, respectively (Bressler et al., 1999; Johannes et al., 2010; Manchikanti, 2000; Peek-Asa et al., 2004). Age-related physiological changes are widely reported, such as reduced muscle strength and joint flexibility (Brown et al., 1994; Hyatt et al., 1990). Such age-related alterations may affect the way an older worker performs MMH tasks and/or the mechanical demand resulting from these tasks. Therefore, the relationship between occupational MMH performance and LBP risk, via biomechanical demands on the lower back, needs to be evaluated for older individuals.

There are a few reports of age-related differences in MMH biomechanics in the literature (Boocock et al., 2015; Shin et al., 2006; Song and Qu, 2014a, b). All of these earlier studies involved two groups, of young and older individuals, and have generally reported decreased lumbar flexion and increased pelvic rotation among older individuals during lifting. The only aspect of mechanical demand during MMH reported in these studies has been the net moment at the lower back, which was not found to be different between the two age groups (Song and Qu, 2014b). Evidence on age-related changes in trunk strength (Voorbij and Steenbekkers, 2001) and range of motion (Intolo et al., 2009) suggests a nonlinear relationship between aging and MMH biomechanics may be present. Therefore, the main objective of the present study was to more precisely assess age-related differences in work methods and the resultant mechanical demand on the lower back during symmetric MMH tasks. Work methods during simulated MMH tasks were quantified using measures of trunk kinematics (i.e. pelvic and thoracic rotations, lumbar posture and lumbopelvic ratio) and mechanical demand on the lower back was quantified using estimates of both net moment and forces at the lower back. On the basis of earlier evidence regarding significant functional change in the trunk after the fourth decade of life (Intolo et al., 2009), it was hypothesized that individuals over 50 would adopt trunk kinematics that would lead to larger mechanical demand at the lower back when completing the same MMH task as younger individuals.

Method

Participants

Sixty healthy individuals completed the study, and were recruited to form five equal-sized and gender-balanced age groups: 22-28, 32-38, 42-48, 52-58, and 62-68 years old (Table 1). These age groups were intended to represent individuals in the 1st through 5th decades of working life. Univariate analysis of variance (ANOVA) indicated no significant differences in stature (p = 0.851) or body mass (p = 0.127) between the five age groups (Table 1). Participants completed an informed consent procedure approved by the University of Kentucky Institutional Review Board prior to any data collection. All participants reported engaging in regular, moderate levels of physical activity, had a BMI between 22 and 30, and were free from any self-reported musculoskeletal disorders or other medical conditions that might have substantially influenced the experimental results. Individuals with a history of working in physically demanding occupations were excluded, to minimize potential confounding related to prior exposure to LBP risk factors. To avoid potential confounding from LBP-induced changes in trunk neuromuscular behaviors (Ahern et al., 1986; Arendt-Nielsen et al., 1996), individuals with a recent (1 year) history of LBP were also excluded. To recruit the sixty eligible subjects, we screened > 200 individuals.

Table 1: Mean (SD) participant characteristics. Each age group included six male and six female participants.

Age group	Age (years)		Stature (m)		Body Mass (kg)	
	M	F	M	F	M	F
22-28	25.6(1.0)	23.5(2.3)	177.8(6.8)	164.9(3.7)	78.5(4.7)	61.4(6.4)
32-38	33.5(2.2)	34.0(1.2)	173.0(5.1)	167.4(7.1)	81.3(10.3)	64.5(10.2)
42-48	44.5(1.8)	45.1(1.4)	179.9(4.8)	166.2(5.4)	88.0(12.0)	70.1(12.1)
52-58	54.3(1.7)	56.0(2.3)	180.5(10.4)	163.4(6.0)	85.4(11.3)	72.0(8.7)
62-68	65.6(1.6)	65.0(2.7)	179.7(6.2)	163.5(5.7)	86.3(11.1)	61.0(4.1)

Experimental procedures

To enhance the reliability of our results, each participant completed two identical experimental sessions, which were conducted in the morning to minimize the influence of diurnal changes in lower back biomechanics (Adams et al., 2007). In each session, participants completed two sagittally-symmetric MMH tasks while standing on the center of a force platform (AMTI, Watertown, MA). The first task (Task-1) involved lowering a 4.5 kg load from an upright standing posture to knee height and then lifting the load back to the initial upright posture (Fig. 1). This task was designed to simulate a MMH tasks with a subject-specific, anthropometrically-designed target height. The second task (Task-2) involved lowering the same load from the upright standing posture and placing it on a horizontal surface (i.e., fixed height for all subjects) located in front of the participant (Fig. 1). In Task-2, to assure the consistency in the amount of load supported by the surface versus the participant's hand, subjects were instructed to put the load completely on the surface for roughly 2-sec. while holding their final lowering posture. This surface was located 55 cm anteriorly and 40 cm superiorly to the center/top of the force platform (Fig. 1). Task-2 was designed to simulate an MMH task with a fixed target height (e.g. placing a load on a horizontal surface with a non-adjustable height). Both tasks were completed using a freestyle method and self-selected pace. All participants completed Task-1 prior to Task-2 and with no practice before the actual testing; however, testing was repeated in cases where the participants violated the task instruction regarding target height. During MMH tasks, trunk and lower-body kinematics were tracked and sampled (50 Hz) using wireless Inertial Measurement Units (IMUs; Xsens Technologies, Enschede, Netherlands), and ground reaction forces were sampled (1000 Hz) from the force platform. Accelerometers were attached to the T10 vertebral process, sacrum (S1), right thigh (superior to the knee joint), and right shank (superior to the ankle joint) using straps. Raw kinematics and kinetics data were low-pass filtered using a fourth-order, bidirectional, Butterworth filter, with respective cutoff frequencies of 6 Hz and 50 Hz (Kristianslund et al., 2012; Hendershot and Wolf, 2014).

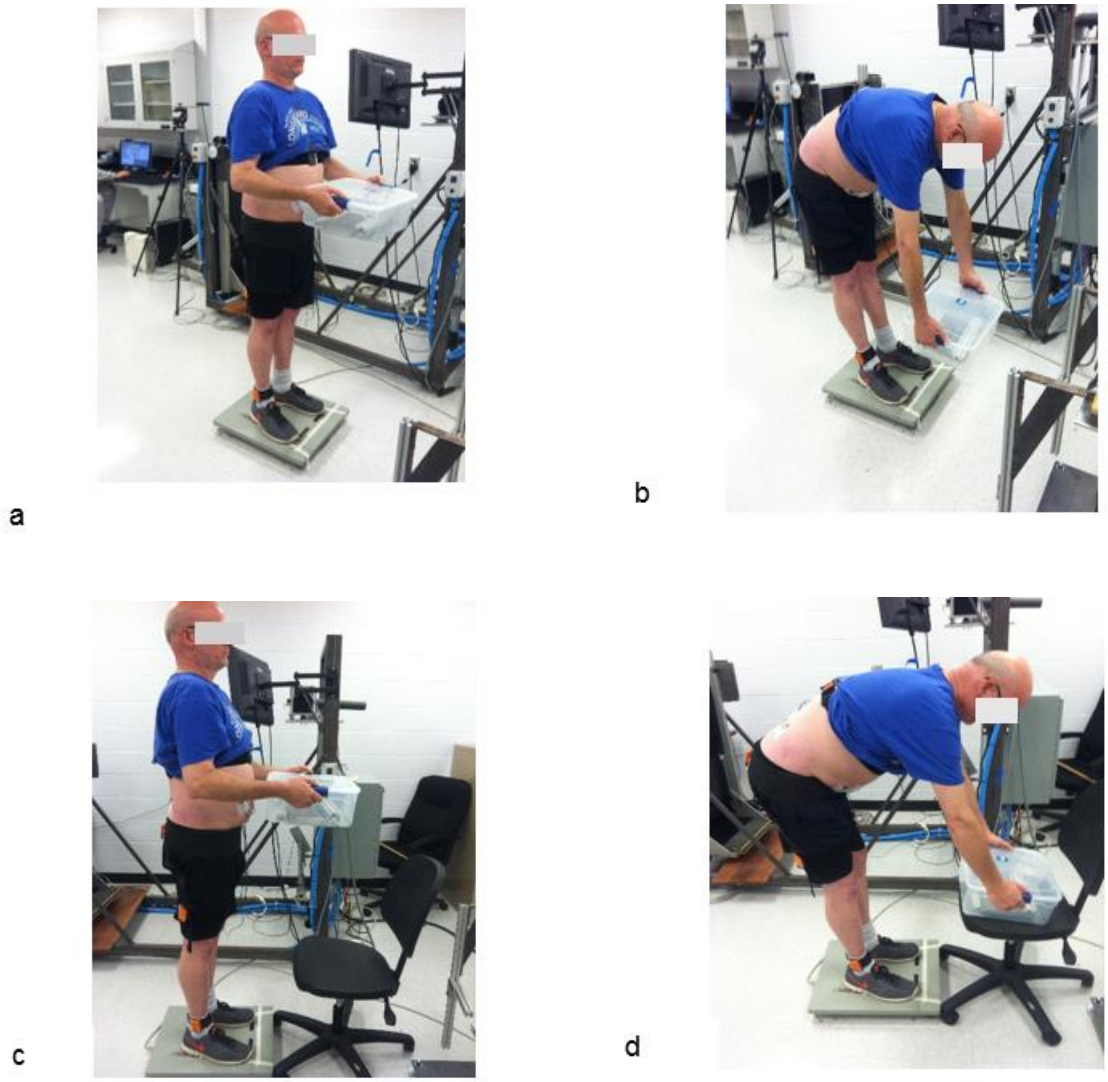

Figure 1: MMH task procedures: a) standing posture for Task-1; b) flexed posture for Task-1; c) standing posture for Task-2; and d) flexed posture for Task-2. Note that sensor locations are only for illustration, whereas in the actual experiment they were attached directly to the skin surface.

Modeling procedure

A three-dimensional, rigid body, linked-segment model (Freivalds et al., 1984; Kingma et al., 1996) of the lower extremities and pelvis was developed in MATLAB (The MathWorks Inc., Natick, MA, USA, version 7.13) to estimate the net reaction forces and moments at the lower back. The model included seven segments (pelvis and bilateral feet, shanks, and thighs) that were connected using frictionless point-contact joints (Fig. 2). . Anthropometric and inertial properties of each segment were estimated from individual stature and mass using existing regression equations (Winter, 2009). Rotation matrices output from IMUs were used to calculate angular velocity and acceleration for each segment using successive numerical differentiation procedure, and linear velocity and acceleration were found using the relationship between linear and rotational velocity (Meriam and Kraige, 2012) and the assumption that the ankle remains in

a fixed planar position, which provided a constant reference for the model (Freivalds et al., 1984). Due to task symmetry, similar kinematics was assumed for right and left extremity limbs.

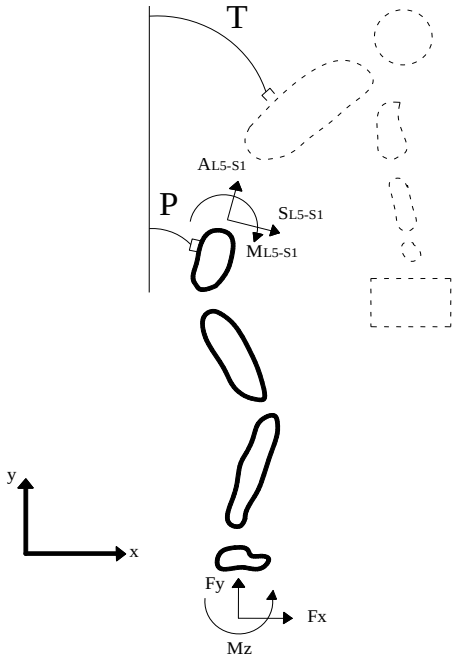

Figure 2: Lateral view of the linked-segment model. Segments with solid lines were used for "bottom-up" inverse dynamics modeling. $A_{L5\text{-}S1}$, $S_{L5\text{-}S1}$ and $M_{L5\text{-}S1}$ represent axial, shearing and moment demand, respectively, at the lower back. F_x, F_y and M_z denote ground reaction forces. Pelvic (P) and thoracic (T) rotations are shown in the figure.

A bottom-up inverse dynamic approach was used to estimate reaction forces and moments at the lower back (via stepwise estimates at the ankle, knee, and hip joints). Estimated reaction forces and moment at the upper side of the pelvis segment were assumed to represent mechanical demands imposed at the lower back (Freivalds et al., 1984; Song and Qu, 2014).The validity and reliability of top-down versus bottom-up calculation of loads in the lower back have been reported in earlier studies (Freivalds et al., 1984; Kingma et al., 1996; Hendershot and Wolf, 2014; Song and Qu, 2014). Projections of the lower back reaction forces perpendicular (i.e., axial) and parallel (i.e., shearing) to the L5-S1 intervertebral disc were calculated to demonstrate the contribution of task demand (i.e., trunk weight and inertia, and external load) to total axial and shearing forces (i.e., sum of external demand and internal tissues response) experienced at the lower back. Following results of Schwab et al. (2006), the orientation of the L5-S1 intervertebral disc in the standing posture was considered to be 51º (with respect to the gravity direction) for the 22-28 and 32-38 age groups, 50° for the 42-48 and 52-58 age groups, and 54° for the 62-68 age group. Since a similar sacral slope in standing posture has been reported between genders (Mac-Thiong et al., 2011), the same values of sacral orientation were used here for males and females. Estimated forces and moments were normalized to body mass and stature * body mass, respectively (Hendershot and Wolf, 2014). However, to present such kinetic measures in a more clinically-meaningful sense (but without affecting the results of statistical analyses), normalized values were multiplied by constants equal to mean body mass and mean stature * body mass across participants. Finally, all kinetic data were time-normalized to a 100% of a lifting task cycle.

Dependent Measures and Statistical analysis

The dependent measures included several measures of trunk kinematics and task demands, as well as the duration of self-selected paces. Kinematic outcome measures for each lifting task were the measured pelvic and thoracic rotations along with corresponding values of lumbar flexion and lumbopelvic ratio, all at the time of maximum thoracic rotation. At this time, lumbar flexion was considered to be the difference between the thoracic and pelvic rotations while lumbopelvic ratio was calculated as the ratio of lumbar flexion to pelvic rotation (Granata and Sanford, 2000). Measures of task demands were the maximum values and corresponding timings for axial, shearing and moment demand of the MMH task at the lower back. Preliminary statistical analyses showed no significant difference (p = 0.31) between the maximum values of task demands during the lowering and lifting phases. Thus, for maximum values with two possible timings, the first half of the task cycle (i.e. lowering phase) was considered for statistical analysis of timings. For each participant and dependent measure, the mean value across two trials (i.e. two sessions) was used for statistical analyses.

Due to the large number of dependent measures, a two-way multivariate analysis of variance (MANOVA) was used to control the experiment-wise error rate (Swanson and Holton, 2005). For any significant effect of the independent variables (i.e. age group and gender) and their interaction (age*gender) identified by MANOVA, a follow-up univariate two-way ANOVA was performed. Significant univariate ANOVAs were followed by post hoc analyses using Tukey's procedure. Since the two lifting tasks were independently designed, separate statistical analyses were performed for each task. All statistical analyses were performed using SAS (version 9.4, Dell Inc), and summary values are reported as means (SD). In all cases, a p value ≤ 0.05 was considered as statistically significant. However, for cases with violation of parametric model assumptions, the p value 0.05 was reduced to 0.01.

Results

From MANOVA, gender (p<0.001) and age (p<0.001) had significant effects on the set of dependent measures for both tasks, and a non-significant interaction effect (p=0.143). During Task-1, the contribution of peak pelvic to peak thoracic rotation was larger (F = 7.12, p < 0.001) among older participants, while the contribution of peak lumbar to peak thoracic rotation was smaller (F = 4.53, p = 0.003) (Figure 3). Age -related differences in the relative contributions of pelvic and lumbar to thoracic rotation were also evident in the corresponding lumbopelvic ratios (F = 4.41, p = 0.004) during Task-1 (Figure 3). Similar age-related differences in trunk kinematics (pelvic rotation: F = 2.91, p = 0.031; lumbar rotation: F = 6.37, p = 0.003; lumbopelvic ratios: F = 4.9, p = 0.002) were observed during Task-2 (Figure 4). For both MMH tasks, thoracic rotations were not different across age groups or genders (Task-1: 87(12) º, Task-2: 81(11) º). In addition to age-related differences, the contribution of peak pelvic to peak thoracic rotation was significantly larger (Task-1: F = 10.97, p = 0.002; Task-2: F = 4.63, p = 0.036) among female versus male participants (Task-1: 32 (12) º vs. 24 (10) º; Task-2: 28 (9) º vs. 23 (9) º). Further, the contributions of peak lumbar to peak thoracic rotations were significantly larger among male versus female participants, at 64(9) º vs. 53(12) º during Task-1 (F = 17.20, p <0.001), and 60(8) º vs. 49(12) º during Task-2 (F = 19.36, p < 0.001). Lumbopelvic ratios were also larger among male versus female participants (3.3(1.9) vs. 2.0(1.5)) during Task-1 (F = 10.15, p =0.003), and 3.1 (1.7) vs. 2.0 (1.1) during Task-2 (F = 9.19, p = 0.004).

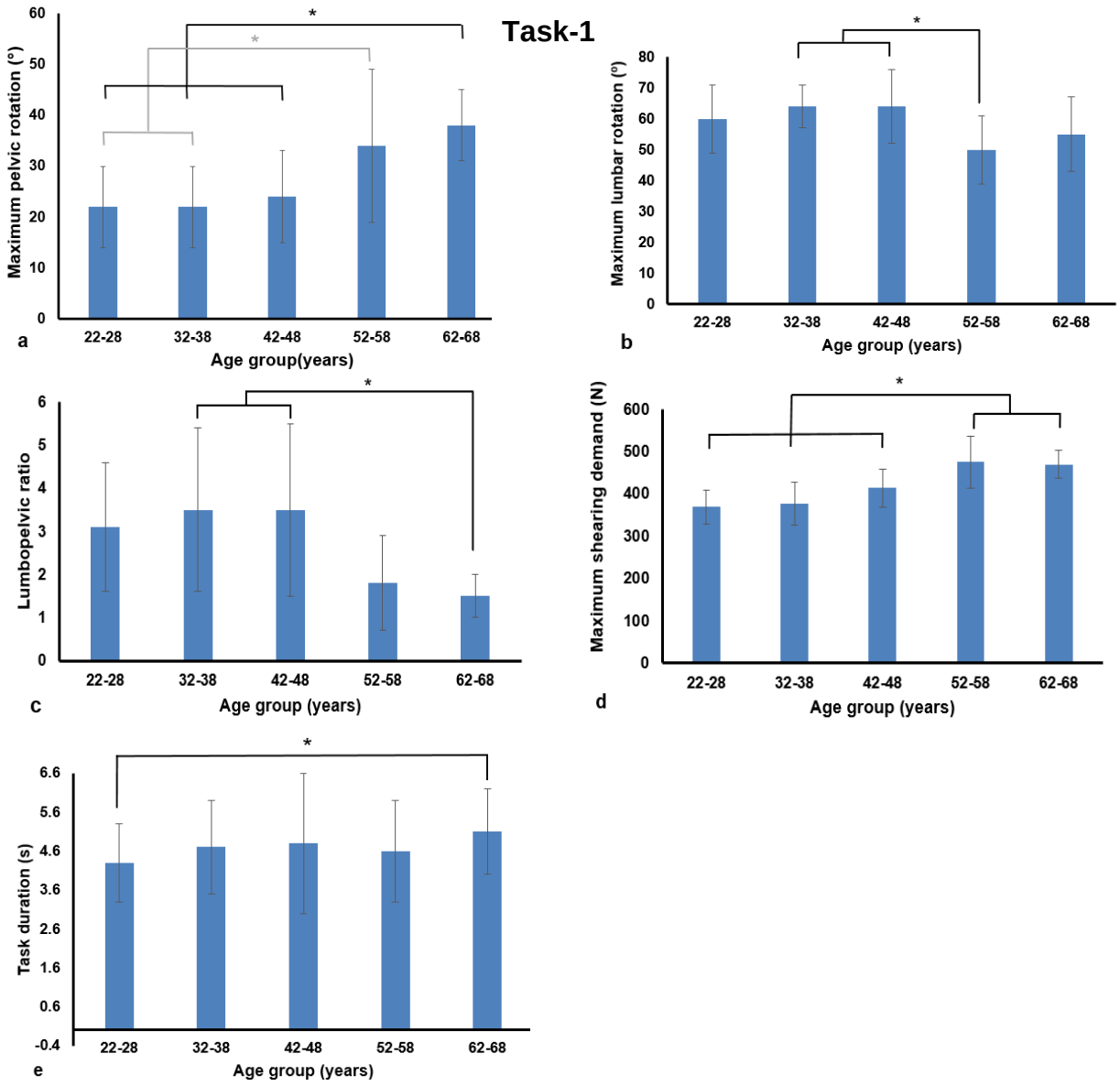

Figure 3: Age-related differences in: a) maximum pelvic rotation; b) maximum lumbar rotation; c) lumbopelvic ratio at maximum thoracic rotation; d) maximum shearing demand; and e) task duration for Task-1. Significant paired differences between age groups are indicated with brackets.

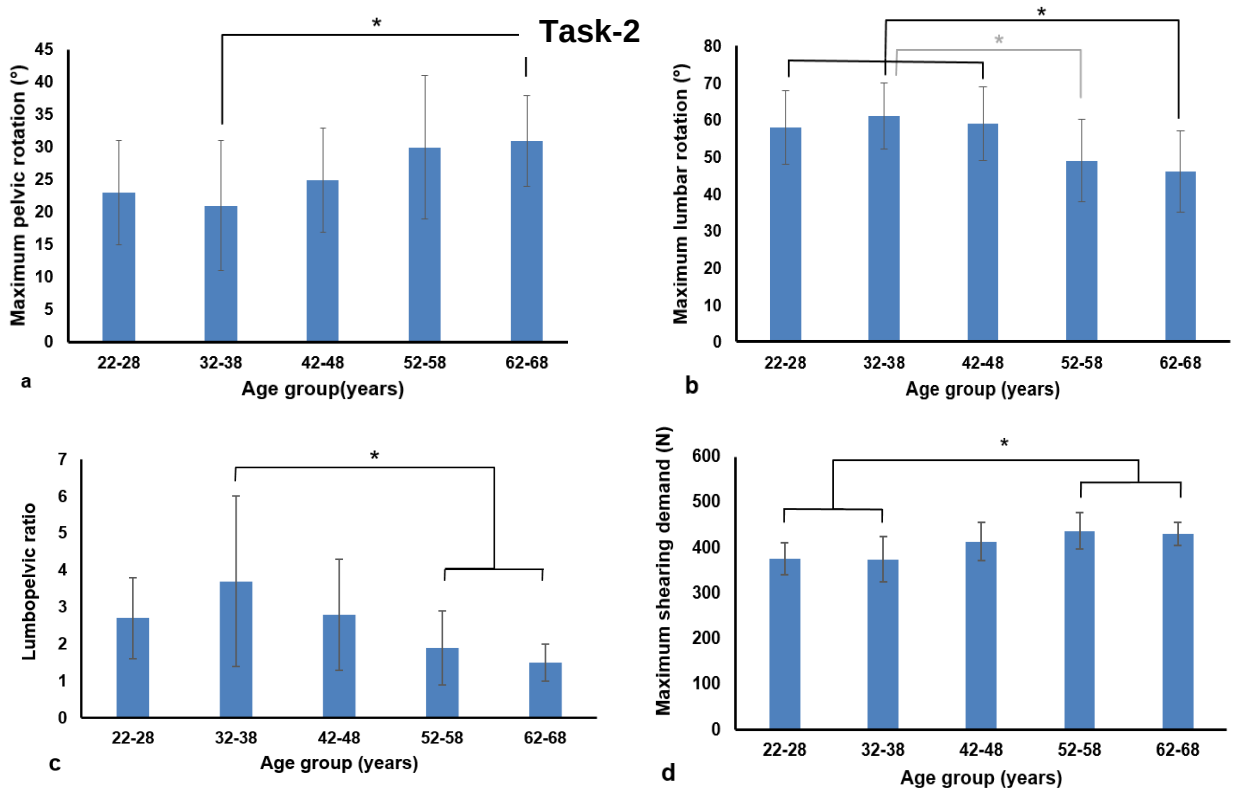

Figure 4: Age-related differences in: a) maximum pelvic rotation; b) maximum lumbar rotation; c) lumbopelvic ratio at maximum thoracic rotation; and d) maximum shearing demand for Task-2. Significant paired differences between age groups are indicated with brackets.

Temporal patterns of mean lower back loads for each age group during Task-1 and Task-2 are depicted in Figures 5 and 6, respectively. For both tasks, there were no significant differences between age groups or genders in the maximum values of net moment (Task-1: 92(23) Nm, Task-2: 112(24) Nm), axial demands (Task-1 and Task-2: 364(16) N), or the timing of maximum net moment (Task-1: 49(7) %, Task-2: 31(7) %), axial (Task-1 and Task-2: 4.8(2.5) %) and shearing demands (Task-1: 51(5) %, Task-2: 32(6) %). Shearing demand of the task was, however, larger (Task-1: $F = 16.11$, $p < 0.001$; Task-2: $F = 7.15$, $p = 0.001$) among older versus younger participants in both tasks (Figures 3 and 4) and among female versus male participants (Task-1: 435(69) N vs. 406(56) N; Task-2: 419(42) N vs. 393(48) N) during both tasks (Task-1: $F = 6.73$, $p = 0.012$; Task-2: $F = 6.65$, $p = 0.013$). Finally, it took longer ($F = 3.57$, $p = 0.035$) for older participants to complete Task-1 (Figure 3), while there were no significant differences in task duration between genders in Task-1 (4.9 (1.3) s) or between age groups or genders in Task-2 (7.7(1.7) s).

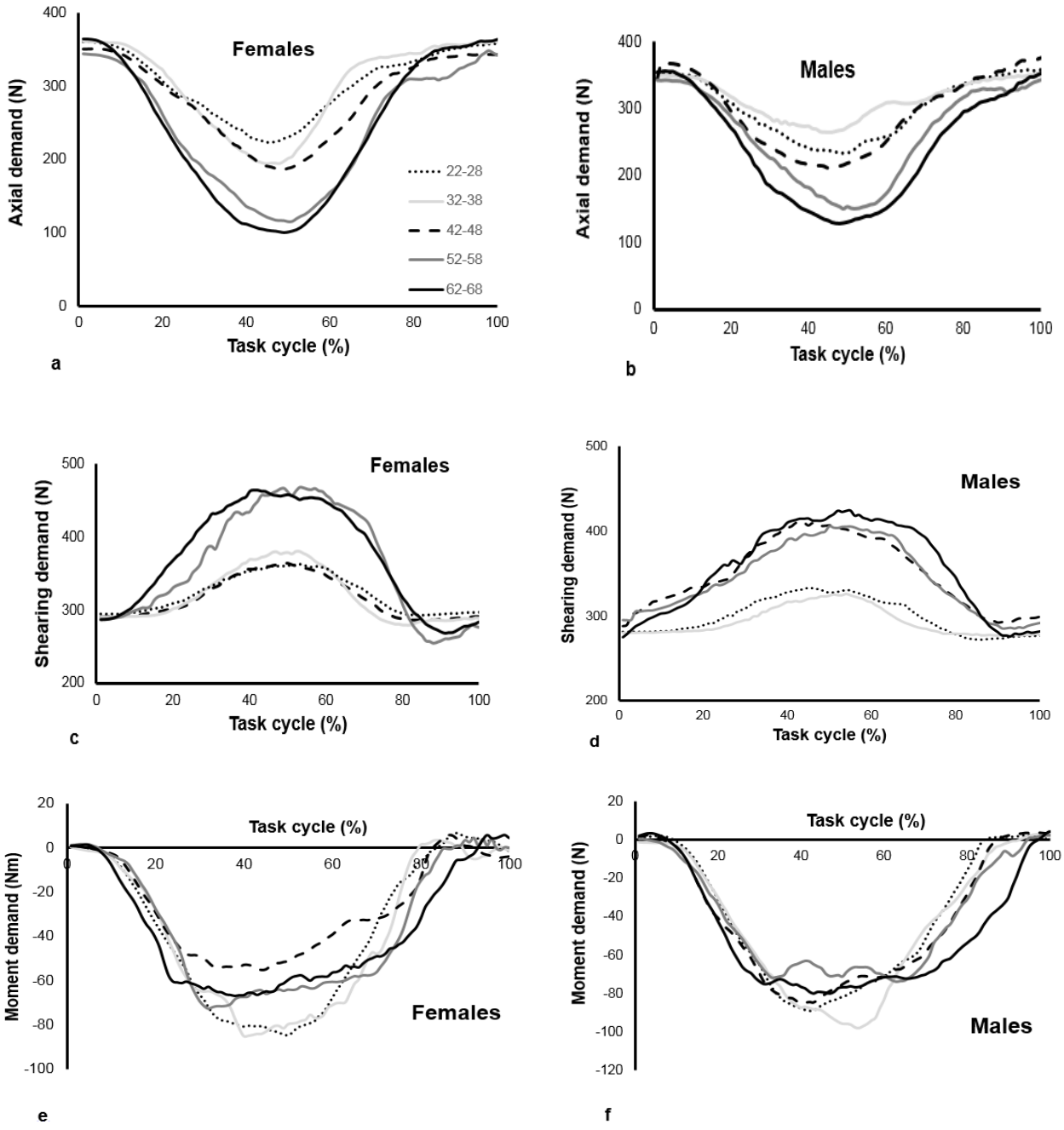

Figure 5: Mean values of axial, shearing, and moment demands of Task-1 at the lower back for different age and gender groups.

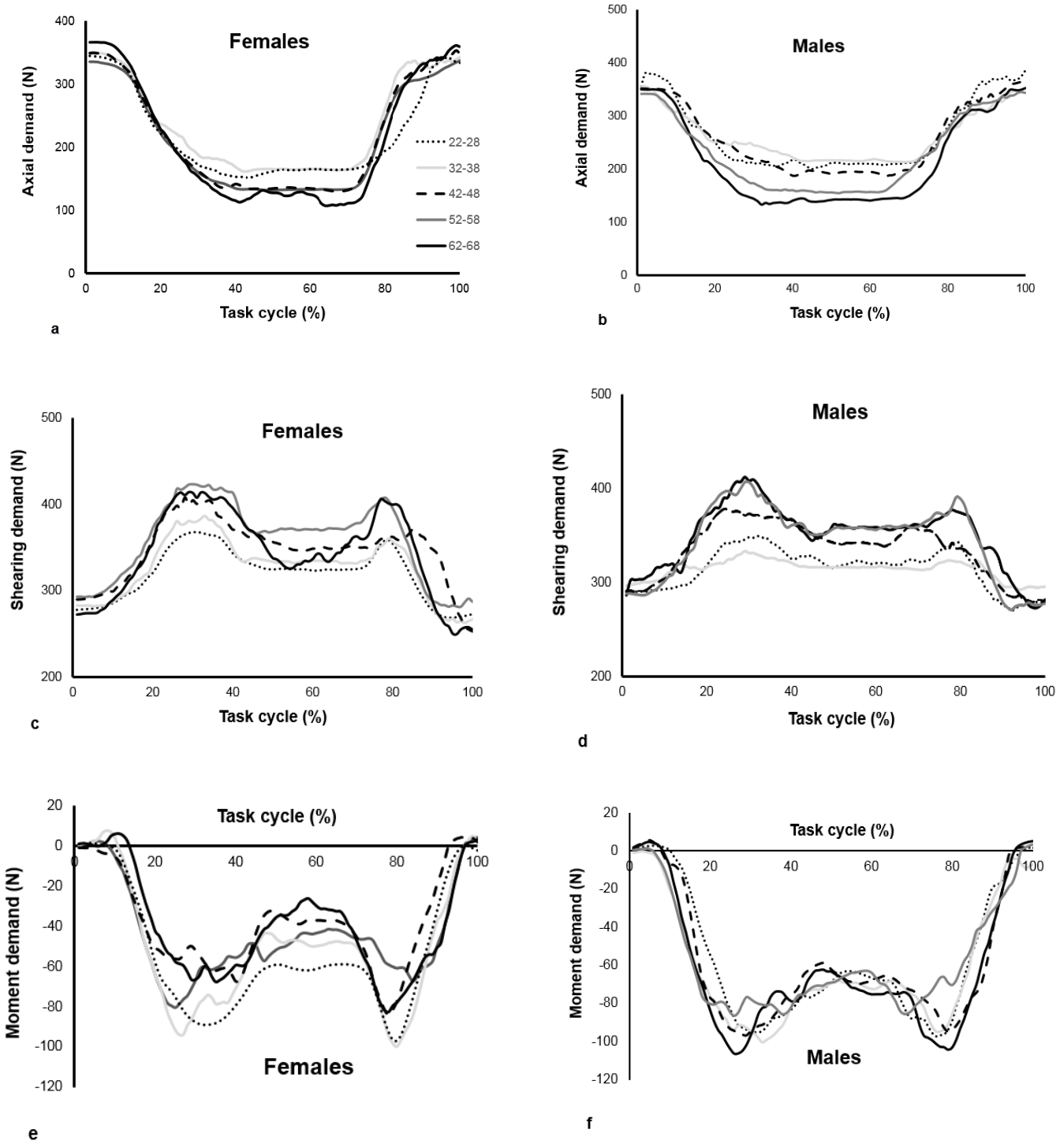

Figure 6: Mean values of axial, shearing and moment demands of Task-2 at the lower back for different age and gender groups.

Discussion

The main purpose of this study was to assess potential age-related differences in trunk kinematics (i.e., measures of work methods) and the resultant mechanical demands on the lower back during two sagittally-symmetric MMH tasks. Five gender-balanced age groups were formed to enable more specific evaluation of age-related differences in MMH biomechanics. Similar levels of maximum thoracic rotation were adopted by participants in both tasks, regardless of age and gender. However, the contribution of pelvic rotation relative to lumbar flexion to such level of thoracic rotation was larger among older participants. Such adopted kinematics resulted in bearing a higher shearing load on the lower back in older participants (i.e., confirming our hypothesis). For both measures of work methods and the resultant mechanical demand, significant differences were observed between individuals older versus younger than 50 years.

Studying groups of young and older people, Song and Qu (2014b) similarly reported larger pelvic rotation, smaller lumbar flexion, and similar thoracic rotation among older individuals when performing various symmetric MMH tasks. Shin et al (2006) also reported smaller lumbar flexion among older individuals during sagittally symmetric MMH tasks; a difference that was not statistically significant, though, likely due to the small sample size. These studies suggested that larger pelvic rotation and smaller lumbar rotation among older individuals might be a protecting strategy for reducing the moment demand of the task on the lower back tissues. This suggestion, though, seems unlikely given their reported results and our current ones indicating similar or larger moments at the lower back of older individuals. Alternatively, such age-related differences in the work methods could be a natural response of the neuromuscular system to changes in the musculoskeletal system with aging, such as a stiffer lumbar spine and weaker back muscles (Adams et al., 2007), that make it more demanding to flex the lumbar spine versus rotating the pelvis. This reasoning may also help explain the observed age-related differences in the duration of Task-1.

The lumbopelvic ratio found in our youngest age group, 2.9 (1.3), was consistent with the reported value of 2.5 (2.4) by Granata and Sanford (2000) that was obtained from relatively young participants (mean age (SD): 23.8 (3.1)) while lifting a 10 kg load. Estimation of lumbopelvic ratio from data reported by Song and Qu (2014b) resulted in values of 3.2 and 1 in the younger and older groups, respectively, comparable with ratios of 2.9(1.3) and 1.5 (0.5) here. Contrary to the earlier suggestion (Song and Qu, 2014b), rather than being a protecting strategy, age-related differences in work methods may predispose older individuals to a higher risk for LBP. A changed (larger or smaller) lumbopelvic ratio relative to the “normal” condition (healthy individuals), as seen in our study among older (higher reports of LBP) versus younger and female (higher reports of LBP) versus male participants (Bressler et al., 1999; Johannes et al., 2010), has been suggested as a clinical indicator of higher LBP risk (Esola et al., 1996; Kim et al., 2013). This suggestion is consistent with higher reports of LBP among older individuals and females (Johannes et al., 2010). From a biomechanical point of view, our results suggest that such age-related differences in work methods impose a higher risk of injury not due to moment demands imposed by the task (to be balanced by lower back tissues) but due to the task-related shearing demand (to be balanced by facet joints and intervertebral discs).

Though we have not accounted for the effects of internal tissue response to task demand on the lower back loading, it should be mentioned that larger pelvic rotation at similar thoracic rotation among older individuals causes the shearing projection of internal tissues response to moment demand of the task to act in the same direction (posterior-anterior) of shearing demand of the task (Arjmand and Shirazi-Adl, 2005), therefore substantially increasing the shearing force

acting on the lumbar spine. Furthermore, using a finite element simulation, Tafazzol et al. (2014) showed that a decrease in lumbopelvic ratio resulted in higher shearing and compression forces at the L5-S1 disc. Considering the smaller lumbopelvic ratios in older versus younger individuals in our study, the former likely experience even higher total spinal loads after accounting for the effects of internal tissue responses to the task demand (Tafazzol et al., 2014). Given the principal role of facet joints in resisting shearing forces (Adams et al., 2007), our findings are consistent with a higher prevalence of LBP due to facet joint pain in older individuals.

There are several limitations associated with our study that should be kept in mind when interpreting the reported results. First, the MMH tasks were performed with a self-selected pace in the sagittal plane. Both velocity and asymmetry of trunk motion have been suggested to affect lower back mechanics, but were not investigated here. Second, symmetry of lower extremity kinematics was assumed, with only data from the right side used in the linked-segment model. Given the sagittal symmetry of the current MMH tasks, though, this assumption was not expected to have introduced substantial error. Third, we did not exclude individuals who had a history of LBP more than one year prior to the study. Any persistent LBP-related alterations in trunk neuromuscular behavior could thus have impacted our results. Finally, the reported loading at the lower back only represents the task demand; more detailed models should be used for characterizing the internal tissue responses and estimating spinal loads using the resultant muscle forces.

In summary, our outcomes for the work methods and resultant demands were in general different between the three younger groups and the two older groups. The mechanical behavior of lower back tissues and the prevalence of LBP have been reported to significantly alter after the fourth decade of the life (Manchikanti, 2000). Our results suggest that such age-related differences in mechanical behavior of lower back tissue result in adaptation of work methods that can predispose individuals to a higher risk of spinal injury and LBP.

Acknowledgment
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5. Supplemental studies

In addition to the studies summarized above, which were included in the formal proposal and conducted to address the specific aims, one more study was completed by the PI and colleagues on a related topic. Given that the student researcher who completed this study was supported by grant funding, this study is also included in the report.

Iman Shojaei, Navid Arjmand, Babak Bazrgari. An optimization-based method for prediction of lumbar spine segmental kinematics from the measurements of thorax and pelvic kinematics. *International Journal of Numerical Methods in Biomedical Engineering (In Press)*.

Abstract
Given measurement difficulties, earlier modeling studies have often used some constant ratios to predict lumbar segmental kinematics from measurements of total lumbar kinematics. Recent imaging studies suggested distribution of lumbar kinematics across its vertebrae changes with trunk rotation, lumbar posture and presence of load. An optimization-based method is presented and validated in this study to predict segmental kinematics from measured total lumbar kinematics. Specifically, a kinematics driven biomechanical model of the spine is used in a heuristic optimization procedure to obtain a set of segmental kinematics that, when prescribed to the model, were associated with the minimum value for the sum of squared predicted muscle stresses across all the lower back muscles. Furthermore, spinal loads estimated using the predicted kinematics by the present method were compared with those estimated using constant ratios. Predicted segmental kinematics were in good agreement with those obtained by imaging with an average error of ~ 10%. Compared to those obtained using constant ratios, predicted spinal loads using segmental kinematics obtained here were in general smaller. In conclusion, the proposed method offers an alternative tool for improving model-based estimates of spinal loads where image-based measurement of lumbar kinematics is not feasible.

Introduction
Limited applicability of measurement tools has prompted application of biomechanical models to gain improved insights into the lower back mechanics (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2007, Bergmark 1989, Calisse et al. 1999, Marras and Sommerich 1991). Many of the existing biomechanical models of the lower back are kinematic-driven wherein measured kinematics is input into the model. The outputs of these models are the resultant forces and moments at one or multiple levels of the lumbar spine that should balance the external demand of the task (Reeves and Cholewicki 2003). The output forces and moments from these kinematic-driven models are then used to estimate muscle forces required to satisfy equilibrium and/or stability of the lumbar spine (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2007, Arjmand and Shirazi-Adl 2005). Estimation of muscle forces using the outputs from the kinematic-driven models is, however, an indeterminate problem since the number of unknown muscle forces is often more than the number of output forces and moments (i.e., equations). The indeterminate problem can be resolved by increasing the number of equations in the kinematic-driven model and/or by selecting a set of muscle forces from all admissible combinations. The latter case could be based on optimizing some neuromuscular or biomechanical aspects of lower back and/or achieving an overall agreement with indirect measures of muscle forces (i.e., electromyography) (Arjmand and Shirazi-Adl 2006, Bean et al. 1988, Larivière et al. 2000, Marras and Granata 1997, McGill 1991, Stokes and Gardner-Morse 2001).

Increasing the number of equations, on the other hand, can be achieved by introducing additional degrees of freedom into the model. However, the main challenge of such an approach is the

measurement of spinal motion for the added degrees of freedom. To overcome this challenge in the earlier works wherein each lumbar motion segment was represented by a degree of freedom, thorax and pelvic kinematics were obtained from measurements but intervening vertebral kinematics in the lumbar region was calculated by distributing the lumbar kinematics (i.e., thorax minus pelvic kinematics) across its vertebrae using a set of constant ratios (Arjmand and Shirazi-Adl 2006, Cholewicki and McGill 1996, McGill and Norman 1986). These ratios represented the percent contribution of each lumbar motion segment to the total lumbar motion and were obtained from earlier imaging studies on the lumbar spine under rather simple tasks of voluntary maximum rotations with constraint pelvis (Dvorak et al. 1991, Frobin et al. 1996, Pearcy et al. 1984, Potvin et al. 1991, Yamamoto et al. 1989). Recent imaging studies using Magnetic Resonant Imaging and stereo radiograph techniques suggest that relative contribution of segments' motion to total lumbar spine motion changes with trunk flexion angle, loading and state of the pelvis (i.e. constraint or free) (Pearcy et al. 1984, Yamamoto et al. 1989, Gatton and Pearcy 1999, Aiyangar et al. 2014, Chen and Lee 1997, Harada et al. 2000). As such, the assumption regarding the distribution of lumbar kinematics across its vertebrae using constant ratios is required to be revised. This is particularly important because ethical consideration, technical limitations and associated cost still hinder application of imaging methods for measurement of lumbar spine segmental kinematics.

The primary objective of this study was to present development and validation of a method for prediction of lumbar segmental kinematics from its measured total kinematics. In this method, a kinematics-driven model of the spine (Reeves and Cholewicki 2003) is used within a heuristic optimization procedure to obtain a distribution of lumbar kinematics between its vertebrae that is associated with the minimum sum of quadratic muscle stresses across all the lower back muscles. A secondary objective of this study was to determine differences in spinal loads between those predicted using kinematics obtained by the new method proposed here and those predicted using kinematics data obtained using constant ratios. Since spine motion is the resultant of internal neuromuscular response to external task demand, it was hypothesized that predictions of lumbar segmental kinematics using a neuromuscular cost function, as partly done in the present study, would be in agreement with those obtained by the image-based measurements. The neuromuscular cost function that was used to obtain lumbar segmental kinematics was the sum of squared muscle stress across all lower back muscles. Hence, it was further hypothesized that the resultant lumbar segmental kinematics to be associated with smaller spinal loads when compared to the traditional approach using the constant ratios.

Method

For a given total lumbar kinematics, the set of segmental kinematics associated with the minimum sum of squared muscle stresses across the entire lower back muscles was obtained using a kinematic-driven model of the spine (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2009) within an optimization procedure (i.e., henceforth called the global optimization). Because of the dependency of the objective function of global optimization procedure on outputs of the kinematics driven model, we start the method section with a brief description of the kinematic-driven model.

Kinematics driven model

The model includes a nonlinear finite element model of the spine wherein equilibrium equations are satisfied across the entire lumbar spine. This model was developed using ABAQUS finite element software (Version 6.13, Dassault Systemes Simulia, Providence, RI) and its nonlinearity arise from 1) the nonlinear geometry of the model (i.e. force-deformation effect and large deformations), and 2) the nonlinear mechanical properties of elements used to simulate the behavior of passive spine components (i.e. ligaments and intervertebral discs). In the finite

element model, as schematically shown in Figure 1, the intervertebral discs are simulated by nonlinear flexible beam elements and vertebrae as rigid elements.

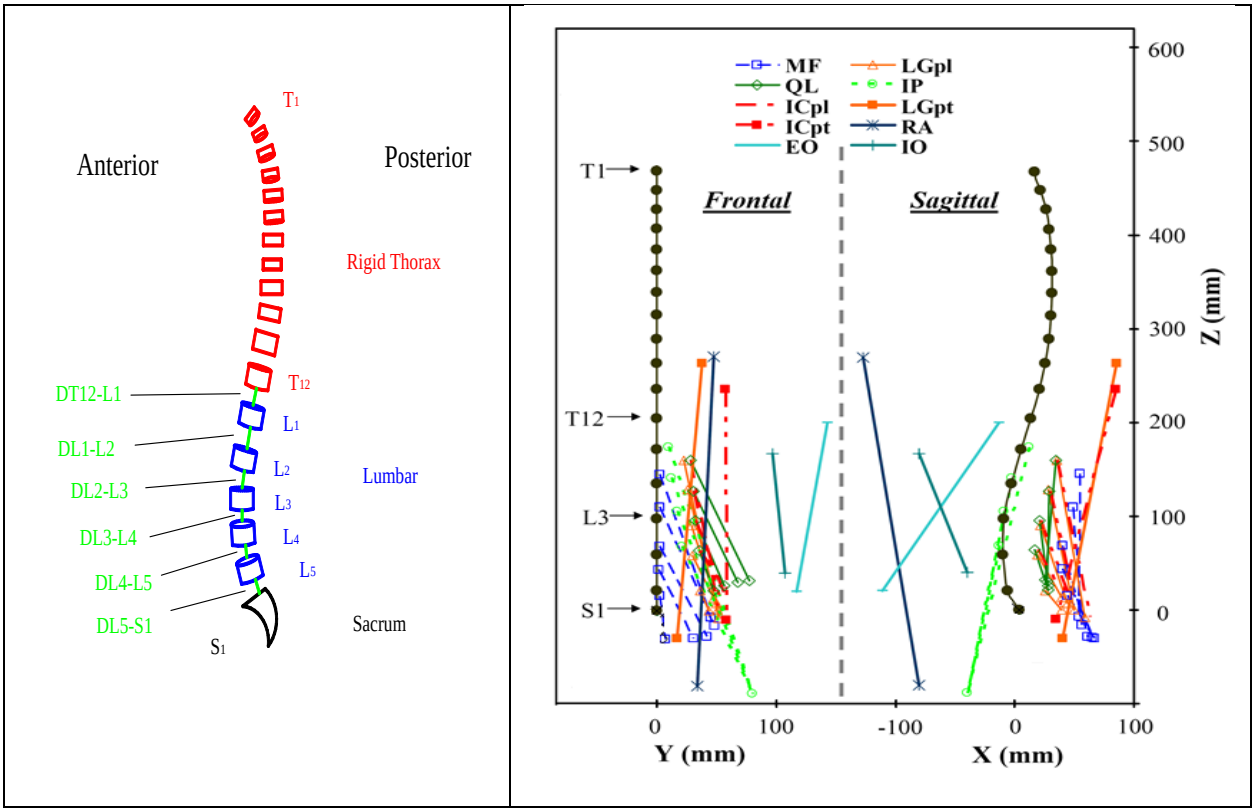

Figure 1. The spine model and the representation of the dimensions (mm) and the musculatures in the sagittal and frontal planes in upright posture. ICpl: iliocostalislumborum pars lumborum, ICpt: iliocostalislumbroum pars thoracis, IP: iliopsoas, LGpl: longissimusthoracis pars lumborum, LGpt: longissimusthoracis pars thoracis, MF: multifidus, QL: quadratuslumborum, IO: internal oblique, EO: external oblique and RA: rectus abdominus. For the sake of clarification only muscles on one side are shown. Mass and mass moment of inertia along with geometrical information are provided in Appendix (Tables A.1 and A.2).

Inputs to the finite element model includes displacement boundary conditions (i.e. sagittal plane segmental kinematics prescribed on the spine vertebrae from the T12 to the S1 level, and constrained out of plane displacements in all levels) along with the gravity (i.e. ~50% of total body weight and distributed across the entire spine (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2009) and external loads (i.e., held in hands). The main outputs of each finite element analysis are six moments; each representing the sagittal plane mechanical demand of activity on muscles attached to a spinal level from the T12 to the L5. These moments are, hence, used to predict muscle forces at each level. Specifically, forces in a muscle architecture including 56 muscles (Fig. 1) that are attached to spine from thorax to pelvis are predicted such to balance moment outputs from finite element simulations. It should be noted that estimation of muscle forces is a post processing to finite element simulations therefore muscles are not simulated in the model. Only insertions and origins of muscles (see Table A.2) are simulated in the finite element model to be able to estimate moment arm and line of action of each muscle during the post processing procedures. Since the number of attached muscles to each level (i.e., 10 muscles in each level from the T12 to the L4 and 6 muscles in the level L5) is more than the number of equilibrium equations (i.e., three at each vertebra), an optimization procedure (hereafter called local

optimization) is used to obtain the set of muscle forces that minimizes the sum of squared muscle stresses at that specific level. This cost function has been reported to result in prediction of muscle forces that are in better agreement with electromyography data (Arjmand and Shirazi-Adl 2006). The local optimization procedure is formulated as follows for each level:

$$\begin{cases} \text{Minmize} \sum_{i=1}^{m} \left(\frac{F_i}{PCSA_i}\right)^2 \\ \text{Subject to} \sum_{i=1}^{m} r_i \times F_i = M \end{cases} \quad (1)$$

Where F_i, $PCSA_i$,r_i respectively denote the force, physiological cross section area, and the moment arm of the i^{th} muscle. m is the number of muscles attached to that level and M is the output of the finite element analysis and represents the mechanical demand of activity on the muscles attached to that level (see Appendix and references (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2009, Bazrgari et al. 2008, Arjmand and Shirazi-Adl 2006, Arjmand et al. 2008) for additional details of the model).

Because of the simplicity of constraints and the cost function as well as the immediate availability of known variables in the local optimization procedures, a closed-form optimization technique (i.e., Lagrange Multiplier Method) is used for the local optimizations. Note that a local optimization problem is solved for each level, from the T12 to the L5, for a total number of six distinct local optimization problems. Finally the estimated muscles forces are applied to the FE model as external loads and the above described procedure is repeated till a convergence is achieved in estimated muscles forces (i.e., changes in all estimated muscle forces during two consecutive iterations is < 1%).

Global optimization procedure: A given value of total lumbar kinematics can mathematically be distributed across its vertebrae using numerous different distributions. From all possible mathematical distributions, a distribution of lumbar kinematics across it vertebrae that is associated with minimum sum of squared muscle stress across all the lower back muscles is obtained in the global optimization procedure. For each set of generated segmental kinematics by the global optimization procedure, muscle stresses are calculated using predicted muscle forces by the kinematic-driven model. Specifically, the set of generated kinematics is prescribed to the kinematics-driven model to predict muscle forces (see previous section). Since evaluation of cost function in the global optimization procedure depends on the outcomes of the kinematic-driven model for each set of generated segmental kinematics (i.e. a distribution), a classic closed-form optimization algorithm is not applicable. Alternatively, a heuristic algorithm is employed wherein a genetic algorithm with 100 generations and 30 individuals in each generation is utilized (Kaveh et al. 2013, Kaveh and Rahami 2009, Rahami et al. 2015). The stop criterion is considered as the tolerance of 10^{-3} for both variables and objective function. This tolerance value has been chosen by trial-and-error and the results are not influenced by this choice. In particular, a much stricter tolerance limit of 10^{-6}, while doubling the simulation time, only resulted in prediction of segmental kinematics that were <0.05 degree different than those obtained using the tolerance limit of 10^{-3}. The global optimization procedure was hence formulated as

$$\begin{cases} \boldsymbol{\theta} = [\theta_{L_1}\theta_{L_2}\theta_{L_3}\theta_{L_4}\theta_{L_5}] \\ \text{Minmize } (\sum_{i=1}^{n=56}\left(\frac{F_i}{PCSA_i}\right)^2\left(1+\alpha\sum_{i=1}^{n=56}max[0,g_m]\right)) \\ \text{Subject to} \\ 0 \le F_i \le \sigma_{max} \times PCSA_i \\ -9.6^{\circ} \le \theta_{T_{12}} - \theta_{L_1} \le 6^{\circ} \\ -9.6^{\circ} \le \theta_{L_1} - \theta_{L_2} \le 6^{\circ} \\ -12^{\circ} \le \theta_{L_2} - \theta_{L_3} \le 3.6^{\circ} \\ -14.4^{\circ} \le \theta_{L_3} - \theta_{L_4} \le 1.2^{\circ} \\ -15.6^{\circ} \le \theta_{L_4} - \theta_{L_5} \le 2.4^{\circ} \\ -10.8^{\circ} \le \theta_{L_5} - \theta_{S_1} \le 6^{\circ} \end{cases} \quad (2)$$

Where θ_{L1} to θ_{L5} are vertebral rotations of the L1 to the L5 respectively and are generated by global optimization procedure. n = 56 denotes the number of trunk muscles in the model. F_i and $PCSA_i$ respectively denote the force and the physiological cross section area of i^{th} lower back muscle, g_m is the number of predicted muscle forces that exceed the muscle force boundaries, α is a penalizing value (see next paragraph), and σ_{max} is the maximum allowable stress in the muscle (i.e., 1.0 MPa) (Winter 1991). θ_{T12} and θ_{S1} are inputs of global optimization and respectively represent the rotation of theT12 and the S1 vertebrae. The rotational inequality constraints denote modified sagittal plane range of motion of lumbar motion segments with negative sign denoting flexion. These were obtained by adding a 20% increase to the mean reported values in Adams *et al*. (2007) to account for individuals' variability.

For some instances, the generated segmental rotations (i.e. θ_{L1} to θ_{L5}) within the global optimization procedure result in prediction of moments in one or multiple lumbar levels in the finite element model that could not be balanced by muscles attached to those levels. To avoid lack of solution from the local optimization procedure in such cases, no constrain is imposed on the predicted muscle forces in the local optimization procedure. This would in turn result in prediction of muscle forces that exceeded muscle force boundaries. However, exclusion of such solutions from the global optimization procedure is assured by penalizing the associated value of objective function. In particular, constraints associated with predicted muscle forces are checked inside the global optimization procedure and cases that violate the constraints are penalized by adding a large number, α, to the cost function.

In summary the global optimization procedure works as follows (Figure 2):

1. Generating a set of lumbar segmental rotation (i.e., **θ**) that satisfy the rotational constraints in Eq. (2)
2. Sending the **θ** to the kinematic-driven model
    a. **θ** and external loads are prescribed on the nonlinear finite element model in the kinematic-driven model and simulation is run within ABAQUS finite element software
    b. Moment outputs from finite element simulation are input to the local optimization procedure in Eq. (1) to calculate muscle forces
    c. Using predicted muscle forces the loading of nonlinear finite element model is updated and the procedure (from 2.a) is repeated till convergence is achieved in

calculation of muscle forces (see (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2007, Bazrgari et al. 2009) for details)

3. Calculating the cost function in Eq. (2) using the predicted muscle forces in step 2 and the penalties due to any violation of force constraints
4. Modifying the **θ** by the optimization algorithm in Eq. (2) and repeating the previous steps until finding the minimum cost function in Eq. (2) regarding the stop criterion

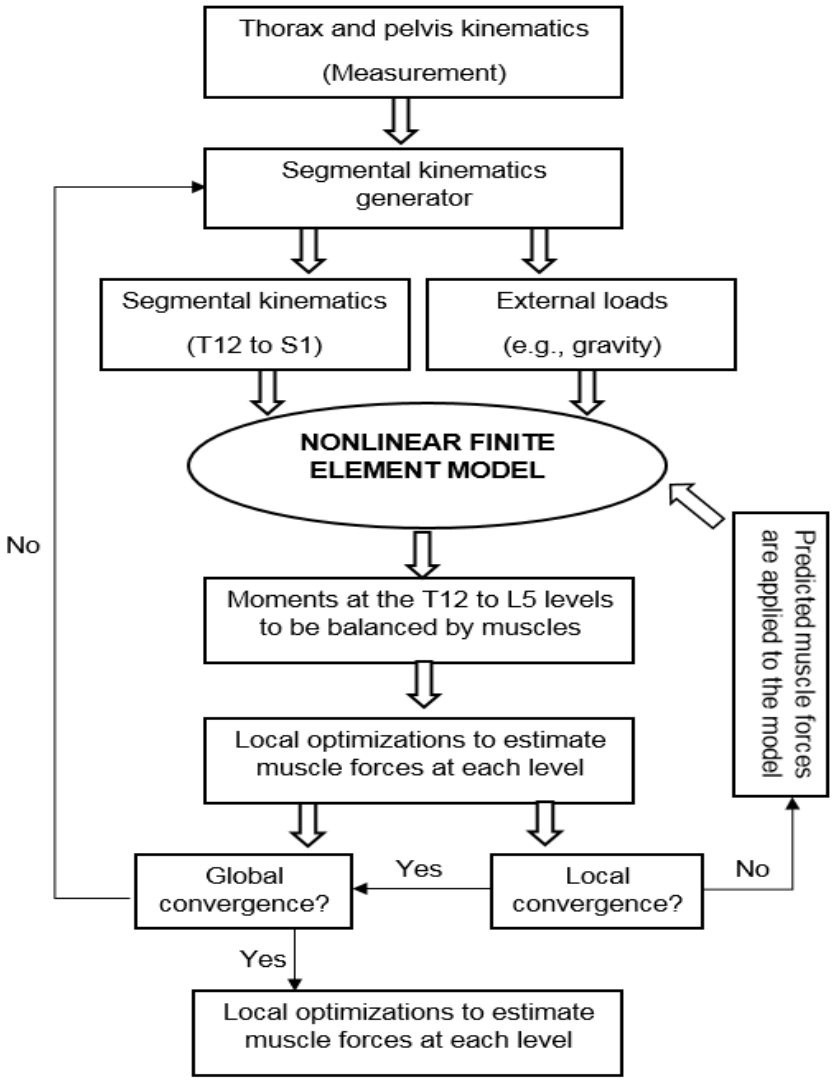

Figure 2. The algorithm for global optimization procedure

Validation

To compare segmental kinematics predicted by the present method with those obtained by imaging studies, several simulations were performed. Flexion angles, loading conditions, thorax and pelvic kinematics were selected such to allow comparison with results from earlier works (Pearcy et al. 1984, Aiyangar et al. 2014, Chen and Lee 1997, Harada et al. 2000) and are summarized here. Pearcy *et al.* (1984) used biplanar radiography method to quantify the movements of the lumbar spine during a motion from the standing erect posture to full forward flexion while the movement of pelvis was constrained. This condition was simulated by fixing the S1 at an angle similar to its angle in standing erect posture and applying 60 degrees of total lumbar flexion to the model. The second imaging study with which we compared our model

predictions was the work by Aiyangar *et al.* (2014). Using dynamic stereo-X-ray imaging, Aiyangar *et al.* (2014) measured lumbar kinematics when subjects performed flexion tasks with a weight of 4.55 kg in hand and constraint pelvis. For our simulations, the weight of 4.55 kg was applied to the T3 vertebra (i.e., the approximate level where the center of shoulder is attached to the trunk) using a rigid element, the S1 was fixed similar to the previous simulation and segmental kinematics were obtained for different flexion angles as reported in Aiyangar *et al*. (2014). The other two studies that were used for comparison included: 1) cineradiography lumbar motion analyses by Harada *et al.* (2000) and 2) radiographic measurement of lumbar spine by Chen and Lee (1997). Both studies involved sagittal plane trunk flexion with pelvis motion was constrained in the former and free in the latter (flexed ~ 10 degrees). For each of the above four comparisons, the error between our predictions and reported measurements, denoted by E, was calculated as the sum of difference between measured and predicted rotations across all lumbar levels divided by the total lumbar flexion.

Effects on spinal loads

To compare spinal loads that were estimated using the segmental kinematics obtained by the proposed method here (called hereafter Disribution-1) and those obtained by the traditional method of constant ratios (called hereafter Distribution-2), measured kinematics of thorax and pelvis were obtained from an earlier work (Arjmand and Shirazi-Adl 2006). Measured kinematics data included mean thorax and pelvic sagittal plane rotations of 15 asymptomatic participants under twelve different forward bending conditions. Each forward bending condition was one of the twelve possible combination of 1) two trunk flexion angles (i.e., 40 and 65 degrees), 2) three lumbar postures (i.e., free, lordotic, and kyphotic) and 3) two loading conditions (i.e., with and without a weight of 180 N). Distribution-2 involved allocations of 8%, 13%, 16%, 23%, 26%, and 14% of total lumbar kinematics respectively for the T12-L1, L1-L2, L2-L3, L3-L4, L4-L5, and L5-S1 motion segments. For each experimental condition and set of lumbar segmental kinematics (i.e., Distribution-1 vs. Distribution-2), the kinematics-driven model (Arjmand and Shirazi-Adl 2006, Bazrgari et al. 2007, Bazrgari et al. 2009) was used to calculate muscle forces and spinal loads.

Results

Validation:

Distribution of lumbar kinematics across its vertebrae predicted by the proposed method here and those reported in earlier studies are presented in Table-1. Kinematics data in this table are only given for the reported segments in the measurement studies. Mean error of our predictions across all validation studies was 10.5%; while, the associated value was 12% in the method with constant ratios.

Table-1: Comparison of predicted (P) segmental rotation (deg) by the current method and those obtained using constant (C) ratios with measured (M) segmental kinematics in earlier imaging studies (Pearcy et al. 1984, Chen and Lee 1997, Harada et al. 2000, Aiyangar et al. 2014).

Reference	Total		T12-L1	L1-L2	L2-L3	L3-L4	L4-L5	L5-S1	E (%)
(Chen and Lee 1997)	60	M	4.8	7.1	8.9	13	14.8	10.69	
		P	4.8	10	10.7	11.9	12.5	10	14.83
		C	4.8	7.8	9.6	13.8	15.6	8.4	8.92
(Pearcy et al. 1984)	60	M		8	10	12	13	9	
		P		8	10	12	13	9	0.00
		C		7.8	9.6	13.8	15.6	8.4	10.77
(Aiyangar et al. 2014)	30	M			6.6	6.8	7.6	5.3	
		P			6.3	5	6.6	8.4	23.57
		C			4.8	6.9	7.8	4.2	12.17
	37.5	M			8.2	8.5	9.2	6.6	
		P			7.9	7.6	8.5	9.2	13.85
		C			6	8.6	9.8	5.2	13.23
	45	M			10.3	10.3	11	7.9	
		P			9.9	10.3	10.7	8.7	3.80
		C			7.2	10.4	11.7	6.3	13.92
	52.5	M			12	12	12.9	9.2	
		P			9.4	10.4	11.5	9.4	12.58
		C			8.4	12.1	13.7	7.3	13.88
(Harada et al. 2000)	45	M				10.9	10.9	9.3	
		P				11.5	9.9	9.6	6.11
		C				10.4	11.7	6.3	13.83
	55	M				13.3	14.4	10.3	
		P				11.9	12.5	10	9.47
		C				12.6	14.3	7.7	8.95
Average		P							10.5
		C							12

Effects on spinal loads:

Lab-based measured values of the thorax (T12) and the pelvis (S1) sagittal plane kinematics for all twelve conditions are listed in the Table-2. For each experimental condition, the predicted segmental kinematics by the proposed method here, as percentages of the total lumbar kinematics, is also presented in the Table-2. Such distribution of the lumbar kinematics across its vertebrae altered with experimental condition. Rather than reporting predicted forces in all muscles, due to space limitation, we chose to report sum of muscle forces at each spinal level (Figure 3). Predicted muscle forces at the T12, L1, L4, and L5 (L2 and L3) using kinematics obtained by the method of this study (i.e., Distribution-1) were lower (higher) than those predicted using Distribution-2 (Figure 3). The values of global optimization cost function (i.e., sum of squared muscle across all the lower back muscles) calculated using muscle forces obtained by Distribution-2 were considerably more (up to twice) than the values obtained by the Distribution-1 (Figure 4). The resultant compression (shear) forces, predicted using Distribution-2, were in general higher (lower) than those obtained using Distribution-1 with maximum values occurring at the L5-S1 (L5-S1) and L4-L5 (L5-S1) disc's mid-plane for respectively Distribution- 2 and Distribuition-1 (Table-3). The maximum (minimum) decrease in the predicted total joint load (i.e., vector summation of shear and compression forces) from those obtained using Distribution-2 to those obtained using Distribution-1 was 568 N (-86 N) and occurred at the L5-S1 (T12-L1) joints during 40 degree trunk flexion with lordotic lumbar posture and 180 N load in hand (65 degree

trunk flexion with kyphotic lumbar posture and 180 N load in hand). Lower differences in spinal loads were associated with higher flexion angles, more kyphotic posture, and no load conditions.

Table-2: Lumbar segmental rotations as percentages of total lumbar flexion. These ratios were kept fixed for all experimental conditions in the Distribution-2. Lab-based measures of the mean T12 and the S1 rotations were the same in both methods and are also presented in the table. D-1 and D-2 denote Distribution-1 and Distribution-2. F, L and K respectively denote free, lordotic and kyphotic lumbar postures.

	D-2 All conditions	D-1(0N)						D-1 (180N)					
		Flexion 40˚			Flexion 65˚			Flexion 40˚			Flexion 65˚		
		F	L	K	F	L	K	F	L	K	F	L	K
T12(degree)		41	41	41	62	62	62	46	46	46	65	65	65
S1(degree)		16	25.2	10.6	27	36.50	23.4	16	23.5	11.5	27	33.6	24.8
T12-L1 (%)	8	8	8	9	11	16	8	18	21	14	9	-2	6
L1-L2 (%)	13	16	17	16	20	25	18	26	31	23	21	14	20
L2-L3 (%)	16	15	13	16	19	20	18	22	23	20	21	18	20
L3-L4 (%)	23	10	-3	13	14	3	18	5	-4	10	17	19	17
L4-L5 (%)	26	20	15	20	14	5	19	1	-8	8	9	22	15
L5-S1 (%)	14	32	51	26	22	31	21	28	36	25	24	29	23

Table-3: Predicted compression and shear forces (N) at different lumbar levels. F, L and K respectively denote free, lordotic and kyphotic lumbar postures and D-1 and D-2 denote Distribution-1 and Distribution-2 respectively.

			Load=0N						Load=180N					
			Flexion 40˚			Flexion 65˚			Flexion 40˚			Flexion 65˚		
			F	L	K	F	L	K	F	L	K	F	L	K
Compression (N)	T12-L1	D-1	1005	958	1027	1300	1178	1382	1723	1645	1828	2377	2214	2438
		D-2	1026	997	1049	1329	1242	1364	1816	1714	1883	2313	2179	2369
	L2-L1	D-1	1099	1041	1124	1434	1319	1510	1903	1788	1988	2658	2516	2705
		D-2	1179	1119	1214	1618	1505	1670	2167	2024	2262	2910	2702	2991
	L3-L2	D-1	1258	1214	1286	1688	1581	1750	2136	2041	2259	3020	2915	3120
		D-2	1300	1261	1337	1842	1712	1887	2423	2264	2530	3331	3097	3419
	L4-L3	D-1	1474	1455	1490	2042	1953	2069	2561	2443	2664	3550	3366	3680
		D-2	1409	1378	1418	1958	1884	1987	2606	2481	2674	3572	3366	3662
	L5-L4	D-1	1638	1622	1644	2309	2191	2364	2863	2792	2942	4060	3945	4151
		D-2	1653	1644	1645	2291	2233	2307	3006	2918	3046	4073	3907	4144
	L5-S1	D-1	1560	1486	1584	2210	2032	2329	2654	2569	2775	3853	3973	4023
		D-2	1874	1877	1871	2595	2519	2631	3302	3234	3379	4408	4233	4476
Shear (N)	T12-L1	D-1	155	138	179	359	278	397	305	227	369	646	477	709
		D-2	161	124	193	345	270	391	312	227	368	614	478	636
	L2-L1	D-1	157	146	171	304	228	363	187	141	253	540	540	617
		D-2	180	147	196	366	283	395	318	251	374	625	514	679
	L3-L2	D-1	156	198	135	242	317	214	280	321	233	256	353	232
		D-2	137	168	139	261	230	275	196	164	211	378	316	403
	L4-L3	D-1	255	267	229	286	251	328	171	190	180	317	635	409
		D-2	265	283	249	392	403	385	410	422	401	602	596	603
	L5-L4	D-1	306	392	242	299	414	252	358	457	305	333	530	308
		D-2	226	294	180	250	345	258	269	371	200	263	375	298
	L5-S1	D-1	709	825	636	899	1051	821	1254	1392	1148	1586	1603	1479
		D-2	683	800	607	835	983	768	1161	1314	1039	1431	1619	1363

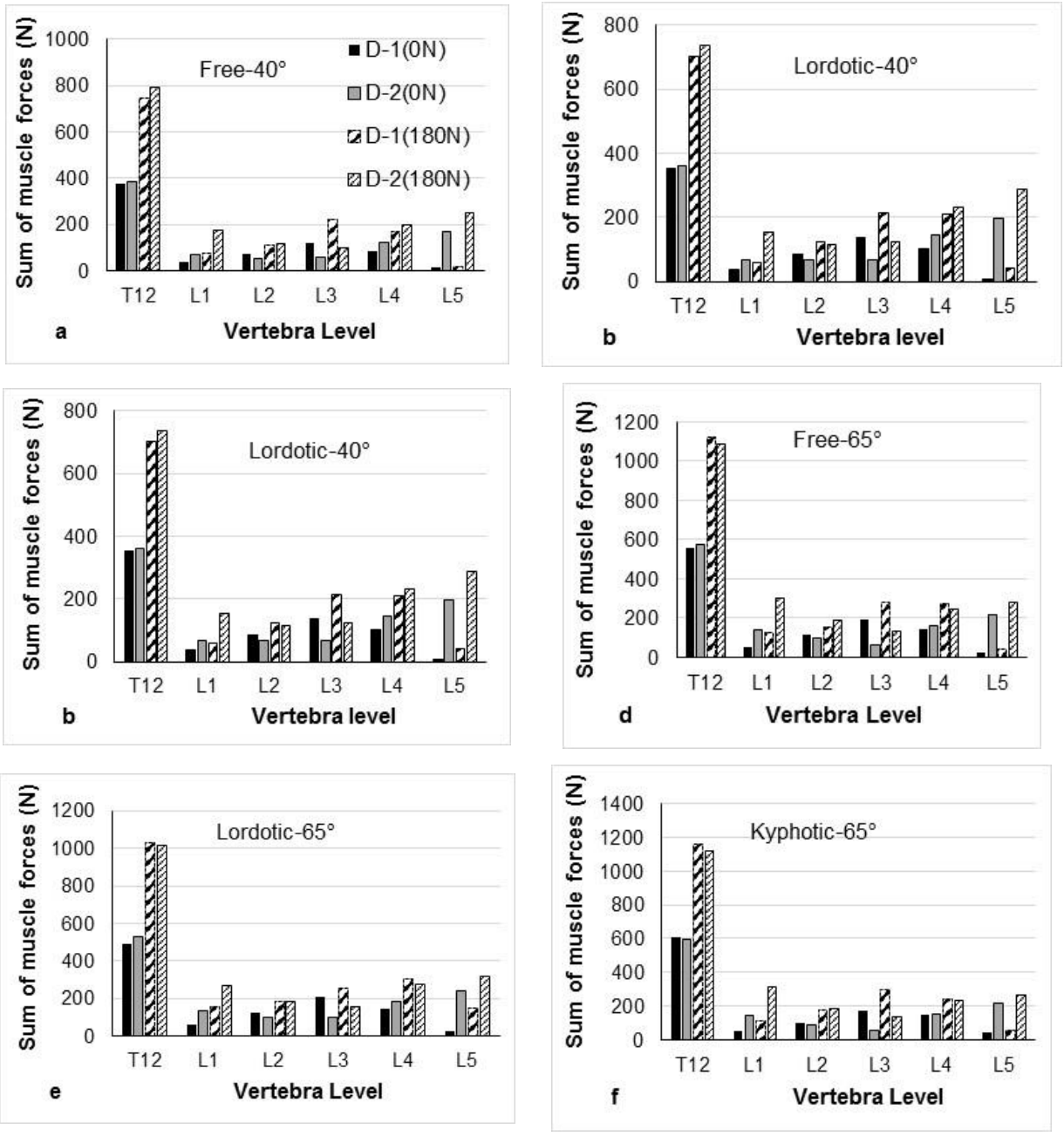

Figure 3. Summation of predicted muscle forces at each level of the lumbar spine using the Distribution-1 (D-1) and the Distribution-2 (D-2) of lumbar flexion across its vertebrae with and without an external load of 180 N. For each case, lumbar posture (i.e. free, lordotic or kyphotic) and trunk flexion angle (i.e. 40° or 65°) are given in the subplot's title.

DISCUSSION

An optimization-based method was developed and validated to predict distribution of total lumbar kinematics across its vertebrae. Since the adopted neuromuscular strategy in response to physical demands of activities, involving the lower back, determines the lumbar segmental kinematics and given the limitations of studies used to establish the constant ratios (e.g., small sample size and involving the end range of motion), the proposed method here, despite complexity, offers great potentials for future research. Predicted distributions using this method

were found to be in good agreement with imaging studies. However, the error between segmental kinematics obtained by this new method and measured kinematics were found in average to be slightly smaller than the error between segmental kinematics obtained using constant ratios and measured segmental kinematics. Distribution of measured lumbar kinematics across its vertebrae using constant ratios (i.e. Distribution-2) resulted in prediction of muscles forces that did not minimize the sum of quadratic muscle stresses among the entire lower back muscles. However, they were not always associated with prediction of higher level of spinal loads.

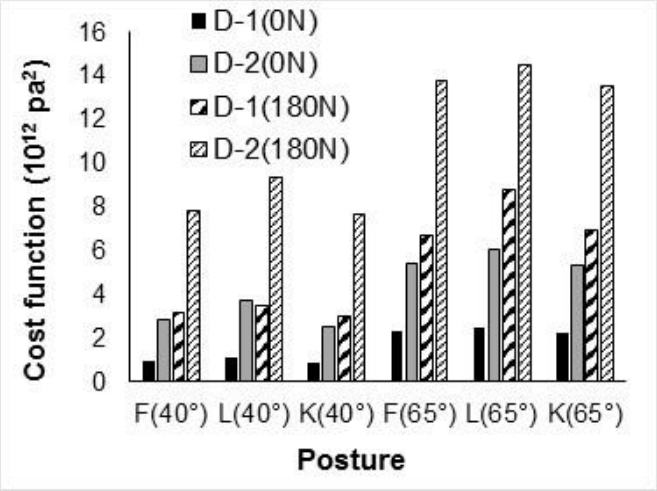

Figure 4. The value of global optimization cost function calculated using the predicted muscle forces by the Distribution-1 (D-1) and the Distribution-2 (D-2) of lumbar flexion across its vertebrae with and without an external load of 180 N. For each case, the trunk flexion angle is denoted by the number given in the parenthesis and the lumbar posture is denoted by F: free, L: lordotic, and K: kyphotic.

Although our results were in good agreement with results from earlier imaging study, it should be noted that the mathematical optimization-based model used in this study may not represent the actual underlying physical phenomenon. The cost function used in our model was merely selected on the basis of its reported capability for prediction of muscles forces that were in better agreement with electromyography measurements (Arjmand and Shirazi-Adl 2006). Furthermore, experimental studies are also prone to measurement errors that should be kept in mind when interpreting the reported results here. For instance, there are considerable differences in reports of measured segmental kinematics of lumbar spine under similar activity (Table-4). The differences between these measurements, when calculated using error variable E, ranges between 11.04% and 44.07% (Table-5) and could potentially be due to many different reasons (e.g., subject and examiner variability as well as differences in measurement tools and methods).

Table-4: Reported segmental rotations (deg) for sagittal plane full flexion

Reference	L1-L2	L2-L3	L3-L4	L4-L5	L5-S1
(Pearcy et al. 1984)	8	10	12	13	9
(Miyasaka et al. 2000)	10	13	15	17	13
(Mannion et al. 2004) (examiner 1)	9	13	17	11	9
(Mannion et al. 2004) (examiner 2)	10	14	17	9	12
(Plamondon et al. 1988)	5	9	11.5	13	
(Kanayama et al. 1996)			10	15	12.5

For conditions studied here (i.e., Table-2) and with few exceptions, greater ratios of total lumbar kinematics were allocated to the T12-L1, L1-L2, L2-3 and L5-S1 motion segments in the Distribution-1 as compared with the Distribution-2. In particular, between 21% and 52% of total lumbar flexion was allocated to the L5-S1 motion segment in the Distribution-1 as compared to 14% in the Distribution-2. Allocation of such higher level of segmental kinematics to the L5-S1 in the Distribution-1 resulted in significant drop in the predicted moment which in-turn caused significant decreases in predicted muscle forces at the L5 levels (Figure 3). Sum of physiological cross sectional areas of muscles attached to the L5 vertebra (i.e., 432 mm^2) is much smaller than those attached to the other lumbar levels (i.e., 632 mm^2, 758 mm^2, 905 mm^2, and 866 mm^2 for respectively L1, L2, L3, and L4 levels). As such any reduction of muscle forces at the L5 level could impact the cost function of global optimization with an amount that is two to four times higher than the reduction in the cost function due to similar amount of change in muscle force at other levels. Accordingly, distribution of lumbar kinematics across its vertebrae in the Distribution-1 has been such to result in the minimum sum of muscle forces at the L5 level followed the L1, L2, L4, and L3 levels (Figure 3).

Table-5: The difference between reported sagittal plane ranges of flexion in terms of error (%)

Reference	(Pearcy et al. 1984)	(Miyasaka et al. 2000)	(Mannion et al. 2004)(examiner 1)	(Mannion et al. 2004)(examiner 2)	(Plamondon et al. 1988)	(Kanayama et al. 1996)
(Pearcy et al. 1984)		26.67	19.82	31.58	11.04	20.98
(Miyasaka et al. 2000)			20.47	18.46	35.29	18.18
(Mannion et al. 2004)(examiner 1)				11.57	35.03	38.93
(Mannion et al. 2004) (examiner 2)					44.07	35.76
(Plamondon et al. 1988)						14.14
(Kanayama et al. 1996)						
Average	25.46					

While trunk was in the flexed posture for all conditions of the second part of this study, negative ratios obtained for some levels by the Distribution-1 indicated extension in these segments. Although this finding seems counter-intuitive, earlier in vivo studies of lower back kinematics have also reported extension in some lumbar segments while trunk was in a flexed posture (Gatton and Pearcy 1999, Lin et al. 1994). Although such agreement with earlier works further support the superiority of our method versus traditional use of constant ratios to obtain segmental kinematics, we couldn't use these earlier studies (Gatton and Pearcy 1999, Lin et al. 1994) for validation purpose due to lack of sufficient data. In contrast to sub-maximal lumbar flexion angles (Table-2), lumbar segmental rotations were not found to be affected by the external load at postures close to its range of motion. Further investigation of our results revealed that at submaximal lumbar rotation where the instantaneous ratio of lumbar to pelvic rotations dropped below 2 (i.e., increasing contribution from pelvis to trunk flexion), a larger portion of total lumbar

flexion was allocated to the L5-S1 L1-L2, and L2-L3 levels as compared to the L3-L4 and The L4-L5 levels.

There are some limitations regarding the finite element model used in this study, in particular modeling of muscles line of action as straight line that should be kept in mind when interpreting the validity of the proposed method (Arjmand et al. 2006, Bazrgari et al. 2007, Bazrgari et al. 2007). The main limitation of optimization-based models of lower back is their inability to capture differences in the behaviors of the central nervous system among different individuals. With advances in imaging technology (Wu et al. 2014), the proposed method here can alternatively be used to investigate such differences in the behavior of central nervous system by searching for a cost function that predict the closest kinematics to the image-based measures of lumbar spine kinematics. This may include searching cost functions that optimizes a given aspects of lower back mechanical environment as related to risk of developing low back pain; for example cost functions that optimize: 1) the level of strain in all or selects muscles, 2) the shear and/or compression forces in one or more spinal levels, 3) axial and/or rotational deformation in one or more motion segments, or 4) passive versus active tissue contributions to equilibrium. Ultimately, the availability of such tool provides a unique platform for a comprehensive study of lower back mechanical environment and may contribute to our efforts for control and management of LBP.

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Appendix

Table-A.1: Mass and mass moment of inertia along with corresponding location of center of mass (unloaded configuration) for all trunk segments

Level	% TM*	% BM	CG - z	CG - x	Ixx	Iyy	Izz	Initial Unloaded Geometry	
								X (mm)	*Z (mm)*
Head-Neck	--	6.94	597.60	-10.00	27.18	29.34	20.13	-	-
Upper Arms	--	2*2.8	447.38	30.00	12.63	11.30	3.80	-	-
Lower Arms	--	2*1.6	426.85	30.00	6.45	5.99	1.20	-	-
Hands	--	2*0.6	405.81	30.00	1.31	0.88	0.50	-	-
T1	3.59	1.28	467.60	-8.00	6.70	2.00	8.70	-12.6	467.6
T2	3.88	1.38	447.38	-12.00	3.40	2.40	9.10	-17.7	447.4
T3	4.15	1.47	426.85	-20.00	8.40	3.20	11.50	-22.1	426.9
T4	4.46	1.58	405.81	-28.00	8.30	3.40	11.70	-25.1	405.8
T5	4.72	1.68	384.14	-33.00	8.00	3.50	11.50	-26.8	384.1
T6	5.03	1.78	361.70	-39.00	7.80	3.90	11.60	-27.6	361.7
T7	5.29	1.88	338.40	-43.00	7.40	4.10	11.50	-27.6	338.4
T8	5.60	1.99	314.12	-45.00	7.20	4.40	11.60	-26.7	314.1
T9	5.91	2.10	288.94	-48.00	7.20	4.70	11.80	-24.9	288.9
T10	6.17	2.19	262.94	-48.00	8.90	6.20	15.00	-21.7	262.9
T11	6.47	2.30	235.30	-46.00	9.00	6.20	15.20	-16.7	235.3
T12	6.74	2.39	204.56	-44.00	11.00	7.20	18.10	-9.6	204.6
L1	7.04	2.50	171.07	-37.01	11.10	6.50	17.50	-1.1	171.1
L2	7.30	2.59	135.03	-29.00	10.90	6.00	16.80	6.8	135.0
L3	7.61	2.70	97.55	-17.00	10.70	5.50	16.10	13.0	97.6
L4	7.87	2.79	58.90	-10.00	11.20	5.30	16.40	14.1	58.9
L5	8.19	2.91	20.57	-6.00	12.20	5.60	17.70	10.2	20.6
S1	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.0	0.0
Pelvis	--	11.00	-89.00	0.00	75.00	30.00	80.00	-	-

* TM: Trunk mass, BM: Body mass, CG-z: height of the centers of mass with respect to the S1 (*mm*), CG-x: anterior-posterior distance from corresponding vertebral centers with negative indicating anterior position (*mm*), *Ixx*, *Iyy*, *Izz*: Mass moments of inertia respectively in anterior-posterior, transverse and longitudinal directions ($Kg.m^2 * 10^{-3}$). Upper arms, lower arms, and hands centers of mass are considered posteriorly at T2, T3, and T4 vertebral levels, respectively. The geometry of the spine in the sagittal plane (mm) (X: anterior-posterior, Z: vertical)

Table-A.2: The coordinates of origins and insertions of the trunk muscles (initial unloaded geometry) and the physiological cross sectional area for muscles on each side

Level	Muscle Name*	Origin			Insertion			PCSA (mm^2)
		X(mm)	*Y(mm)*	*Z(mm)*	*X(mm)*	*Y(mm)*	*Z(mm)*	
T12	ICpt	84.9	57	235.3	34.9	58	-10	660
	IO	-80	96	167	-40	107	40	1200
	RA	-126.7	47.3	269.8	-80	34	-80	567
	EO	-13	141.5	200	-111	116	20	1576
	LGpt	85.6	37.8	262.9	40	17	-30	1345
	MF	54.9	2.5	145	54	45.2	-7.4	96
L1	QL	34.8	28.3	159.3	26	77	32	88
	IP	12	10	174.3	-40	79	-88	252
	ICpl	35.7	28.3	159.3	63	52	-7	108
	LGpl	35	22.1	159.6	59	51.3	-8.4	79
	MF	48.8	2.5	110	56	47.9	-16.5	138
L2	QL	28.6	30.5	126.9	27	67	30	80
	ICpl	29.8	30.5	126.9	49	52	12	154
	LGpl	28.4	26	128.2	50	50.7	0.1	91
	MF	40.6	2.6	68.4	61	41.6	-28.3	211
L3	QL	21	32.3	95.3	28	56	26	75
	ICpl	21.9	32.3	95.3	44	55	18	182
	LGpl	22	28.9	90.1	44	49.6	7.3	103
	MF	40.6	1.5	43.4	65	30.4	-29.5	186
L4	QL	17.1	35.1	63.9	28	47	21	70
	ICpl	19.3	35.1	63.9	37	58	23	189
	LGpl	19.3	30	58.6	39	47.1	13.3	110
L5	MF	43.9	2.3	15.2	67	7.6	-30.4	134
	LGpl	25.7	36	20.6	39	42.9	0	116

* ICpl: Iliocostalis lumborum pars lumborum, ICpt: Iliocostalis lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus thoracis pars lumborum, LGpt: Longissimus thoracis pars thoracic, MF: Multifidus, QL: Quadratus lumborum, IO: Internal oblique, EO: External oblique, and RA: Rectus abdominus. x, y, and z represent anteroposterior, transverse (lateral), and longitudinal directions, respectively.

LIST OF PUBLICATIONS

The following lists publications that have resulted to date from work supported (directly or indirectly) by this grant, and a list of future publications that are anticipated.

Journal Articles

1. Shojaei I, Arjmand, A, Bazrgari B [2015]. The effects of lumbar spine segmental rotations on predictions of spinal loads. *International Journal of Numerical Methods in Biomedical Engineering* (In Press)

Conference Proceedings

1. Shojaei I, Vazirian M, Croft E, Nussbaum MA, Bazrgari B [2015]. Lower back biomechanics during manual material handling task; the effects of aging. *Biomedical Engineering Society's Annual Meeting,* October 7-10, 2015 Tampa, FL, USA. Online only, not paginated (1 pp.).
2. Vazirian M, Shojaei I, Tromp R, Nussbaum MA, Bazrgari B [2015]. Age-related alterations in trunk intrinsic stiffness. *Biomedical Engineering Society's Annual Meeting,* October 7-10, 2015 Tampa, FL, USA. Online only, not paginated (1 pp.).
3. Shojaei I, Vazirian M, Croft E, Bazrgari B [2015]. Age-related differences in responses of lower back tissues and the resultant spinal loads during lifting. *39th annual meeting of the American Society of Biomechanics,* August 5-8 2015 Columbus, OH, USA Online only, not paginated (2 pp.).
4. Vazirian M, Shojaei I, Agarwal A, Bazrgari B [2015]. The lumbopelvic ratio during trunk flexion: the effects of age, gender and motion pace. *39th annual meeting of the American Society of Biomechanics,* August 5-8 2015 Columbus, OH, USA Online only, not paginated (2 pp.).
5. Shojaei I, Bazrgari B [2014]. Analytical solution for obtaining the lumbar spine segmental rotations. *Biomedical Engineering Society's Annual Meeting*, October 22-25, 2014 San Antonio, TX, USA. Online only, not paginated (1 pp.).
6. Vazirian M, Agarwal A, Koch B, Tromp R, Bazrgari B [2014]. Sensitivity of lumbopelvic rhythm to risk factors of low back pain. *Biomedical Engineering Society's Annual Meeting*, October 22-25, 2014 San Antonio, TX, USA. Online only, not paginated (1 pp.).

Dissertation/Thesis:

Koch B: [2013] The study of trunk mechanical and neuromuscular behaviors. MSc Dissertation, University of Kentucky

Tromp R: [2014]. Body armor induced changes in the trunk mechanical and neuromuscular behavior. MSc Dissertation, University of Kentucky

Croft E: [TBD] The effects of spinal manipulation on the lower back biomechanics of healthy individuals. . MSc Dissertation, University of Kentucky

Publications submitted or In preparation

1. Shojaei I, Vazirian M, Croft E, Nussbaum MA, Bazrgari B [2015]. Age-related changes in mechanical demands imposed on the lower back by manual material handling tasks. *Submitted to Journal of Biomechanics*
2. Vazirian M, Shojaei I, Tromp R, Nussbaum MA, Bazrgari B [2015]. Age and gender differences in intrinsic trunk stiffness. *Submitted to Journal of Biomechanics*
3. Vazirian M, Bazrgari B [2015]. Lumbopelvic coordination under trunk movements in the sagittal plane: A review of experimental studies. *Submitted to Applied Biomechanics 2015*

4. Vazirian M, Shojaei I, Bazrgari B [2016]. Alterations in lumbopelvic coordination under trunk movements in the sagittal plane; the effects of age. *To be submitted to Gait and Posture*
5. Shojaei I, Vazirian M, Bazrgari B [2016]. Age-related alterations in active and passive mechanical responses of lower back tissues to lifting demands and the resultant spinal loads. *To be submitted to Ergonomics*
6. Shojaei I, Bazrgari B [2016]. Changes in viscoelastic response of lower back to passive flexion in the sagittal plane with age. *To be submitted to Clinical Biomechanics*

CUMULATIVE INCLUSION ENROLLMENT REPORT

Study Title: Aging, Neuromuscular Behavior, and risk of Occupational Low back Pain

Comments: 5R21OH010195

Racial Categories	Ethnic Categories									Total
	Not Hispanic or Latino			Hispanic or Latino			Unknown/Not Reported Ethnicity			
	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	
American Indian/ Alaska Native										0
Asian	2	4								6
Native Hawaiian or Other Pacific Islander										0
Black or African American	1	2								3
White	32	30								62
More Than One Race		1								1
Unknown or Not Reported										0
Total	35	37	0	0	0	0	0	0	0	72

INCLUSION OF CHILDEREN

The participant age range in this application was 22-68 in order to represent and compare workers in (roughly) their first five decades of working life (i.e., 22-28, 32-38, 42-48, 52-58, and 62-68 years old). Since it is less likely for individuals <22 years old to develop occupational back pain, 22 was the lower age limit in the present study. As such and based on the DHHS definition of children (i.e., less than 21 years), no children was included in the proposed investigation.

MATERIALS AVAILABLE FOR OTHER INVESTIGATORS

All experimental data are available. Software (MatLab™ code) is available for the biomechanical models. These may be accessed by contacting the PI, whose information is provided below. Note that the investigators wish to limit such access to individuals requesting data or software for research purposes only.

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