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Age-related differences in trunk intrinsic stiffness

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ABSTRACT

Age-related differences in trunk intrinsic stiffness, as an important potential contributor to spinal stability, were investigated here because of: (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-related difference in trunk intrinsic stiffness, the current design assessed the effects of gender, 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 were observed. Since the passive contribution of trunk tissues in the upright standing posture is minimal, our values of estimated 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), diminish their contributions to the spinal stability.

1. 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). Biomechanically, 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 complex, requiring coordination of three musculoskeletal subsystems: (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, 1992a,b). 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, 2007; Stokes et al., 2006; Hodges et al., 2009; Shahvarpour et al., 2014). A limitation of these earlier studies, however, was the lack of control over the trunk displacement and response time. As such, measures of trunk stiffness represented the combined effects of active, passive, and neural control stabilizing subsystems (Panjabi, 1992a,b), without information about relative contributions of each subsystem. This limitation was later overcome by use of displacement-controlled perturbation paradigms

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that allowed separate evaluation of each of the three stabilizing subsystems (Moorhouse and Granata, 2007). This paradigm minimizes alterations in the contribution of the passive subsystem, by minimizing the post-perturbation trunk displacement. It also enables distinguishing volitional from reflexive responses, by limiting response time to be shorter than minimum reflex delay (Moorhouse and Granata, 2007; Bazrgari et al., 2011; Hendershot et al., 2011; Miller et al., 2013; Muslim et al., 2013).

Recently, we used this method to study age-related differences in the stabilizing capabilities of the lower back musculoskeletal system, and here 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 (Iida et al., 2002; Yassierli et al., 2007; Galbusera et al., 2014), suggesting an age-related alteration in the lumbar spine stability and the associated risk of LBP. Therefore, our primary objective was to investigate age-related differences in trunk intrinsic stiffness. With aging, the ligaments become slack in the neutral posture (Iida et al., 2002), intervertebral discs become stiffer (Galbusera et al., 2014), and the volitional capability of lower back extensors declines (Yassierli et al., 2007). With this diverging evidence, it was difficult to form 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), reduced spinal stability with aging, and hence in the trunk intrinsic stiffness, was hypothesized.

2. Methods

2.1. 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 such as manual materials handling, frequent bending and twisting, non-neutral working postures, repetitive movements, or whole-body vibration (Burdorf and Sorock, 1997). Individuals with 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 22–30 range were also excluded. Consented volunteers were screened by a nurse to ensure their medical and physical eligibility for participation before experimental data collection. Our rationale for such exclusion criteria as well as decade-based grouping was to generate data that will serve as a base line, characterizing the effects of natural aging, for future cross-sectional research that will investigate the accumulation of work-related disturbances in lower back biomechanics. 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.

2.2. 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 and Burton, 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, 2012; Hendershot et al., 2011, 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 (Fig. 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 s. Isometric trunk exertion was measured (sampling rate: 3000 Hz) 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 (sampling rate: 3000 Hz) 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 in. and 1 in. medio-laterally on both sides of the midline of the participant's back), rectus abdominis (located 1 in. medio-laterally and 1 in. 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).

During perturbation tests, participants were exposed to a sequence of eight anterior–posterior position perturbations (pull and push) over a ~30 s duration, with maximum horizontal displacement, velocity, and acceleration of 10 mm, 0.25 m/s and 8 m/s^2, respectively. These displacement perturbations were generated by rotary motion of the servomotor shaft, with a maximum angular velocity of 95 revolutions/min that were translated to horizontal reciprocal displacements via a

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)

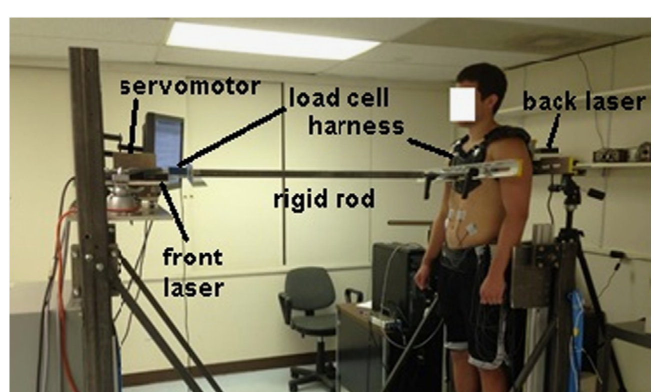

Fig. 1. The setup used for the perturbation tests, including the metal frame, pelvis restraining belt, servomotor, rigid rod, harness, laser sensors, and load cell.

crank mechanism, with a rotating radius of 0.025 m, and then transferred to the participant's trunk via the harness-rigid rod assembly. There were pseudorandom delays between perturbations to reduce the ability of participants to anticipate the exact timing of the perturbations, and thus minimize anticipatory muscle activation. 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 (20–500 Hz) and then rectified and low-pass filtered (3 Hz) to create a linear envelope.

2.3. 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 (Fig. 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) the trunk damping element was not included in the model for enhanced accuracy of the system identification procedure. Since tissues in the trunk are not rigid or rigidly connected, the assumption of unified motion under the perturbations applied in this study was not valid. For this reason, the mass element used in the model for the trunk was left unknown, and determined as “apparent mass” by the system identification procedure. As result of this simplification, apparent mass as estimated here will be smaller than the total trunk mass.

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).

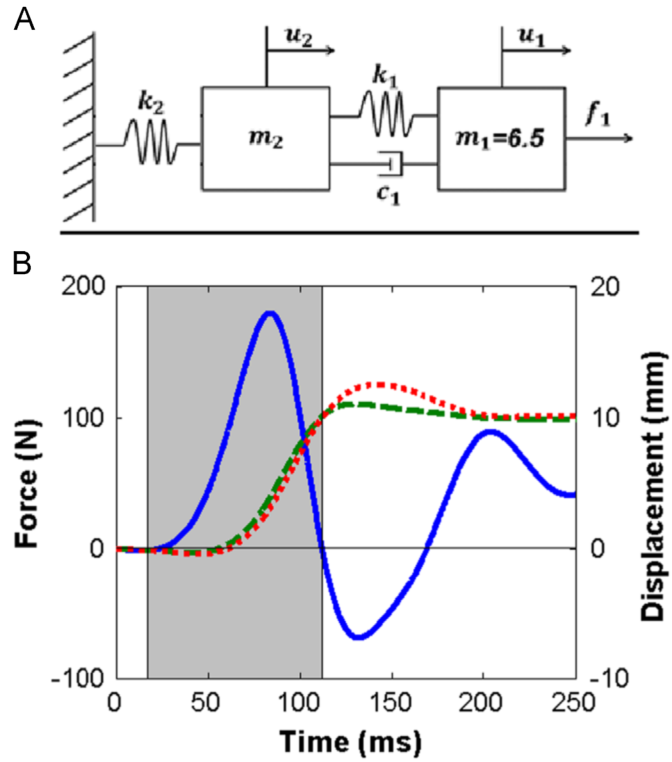

Fig. 2. (A) The spring–mass–damper model used to model perturbations. Parameters m_1 (m_2), k_1 (k_2) and c_1 respectively denote apparent mass, stiffness, damping of the harness-rigid rod assembly (the participant's trunk), and u_1 and u_2 respectively denote the displacement of the inline load cell and the trunk, and f_1 denotes the force applied externally to the system (via the load cell). (B) Samples of measured kinematics and kinetics of trunk during one single perturbation, the time window (shaded in gray) during which the connecting rod was under tension was used to estimate trunk intrinsic stiffness.

Although a time window of 1–3 s 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).

2.4. Statistical analysis

The dependent variables analyzed were: (1) the maximum recorded values of MVE efforts by the load-cell (MVEmax (N)),

(2) the difference in exerted extension forces (N) between 20% and 30% MVEmax effort conditions (when guided by force feedback) (3) the estimated trunk intrinsic stiffness (Kint (kN/m)), 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.

3. Results

3.1. Maximal and sub-maximal extension efforts

Mean values of MVE_{max} were significantly lower among older versus younger individuals (F=3.914, p=0.007) (Fig. 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, $p < 0.001$) (Fig. 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.

3.2. Trunk intrinsic stiffness – K_{int}

Summary of the ANOVA results for trunk intrinsic stiffness and apparent mass are reported in Table 2. As indicated, there was no significant difference in estimated values of K_{int} between age groups. However, both gender and the level of effort 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 (Fig. 4A). Neither age nor gender had significant interactions with the type of feedback or the level of preload.

3.3. Normalized apparent mass – M_{app}

While there was no difference in estimated values of M_{app} between age groups (Table 2), there were significant differences between genders, with males having a larger M_{app}. M_{app} was also larger with the 30% effort level for both feedback types (Fig. 4B).

4. 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

Table 2
Summary of main and interactive effects of age, gender, level of effort, and type of feedback on the trunk intrinsic stiffness and normalized apparent mass. Significant effects (p values) are highlighted in bold text, and effect sizes are indicated using eta-squared (η^2).

Factor	Intrinsic stiffness			Apparent mass		
	F	p	η^2	F	p	η^2
Age (A)	0.4	.776	.042	2.2	.086	.177
Gender (G)	30.3	**.000**	.425	7.9	**.008**	.161
Feedback type (FT)	2.3	.139	.053	0.1	.816	.001
Level of effort (LE)	24.3	**.000**	.372	64.8	**.000**	.612
A*G	0.4	.824	.035	1.9	.130	.156
A*FT	1.2	.326	.105	1.1	.385	.094
A*LE	1.3	.277	.114	1.6	.204	.132
G*FT	1.7	.199	.040	0.4	.515	.010
G*LE	0.0	.995	.000	1.6	.208	.038
FT*LE	1.4	.240	.034	0.7	.398	.017
A*G*FT	0.7	.602	.063	1.5	.216	.129
A*G*LE	0.5	.732	.047	1.5	.227	.126
A*FT*LE	0.5	.746	.045	1.3	.300	.110
G*FT*LE	0.0	.922	.000	0.1	.824	.001
A*G*FT*LE	1.1	.363	.098	0.5	.744	.045

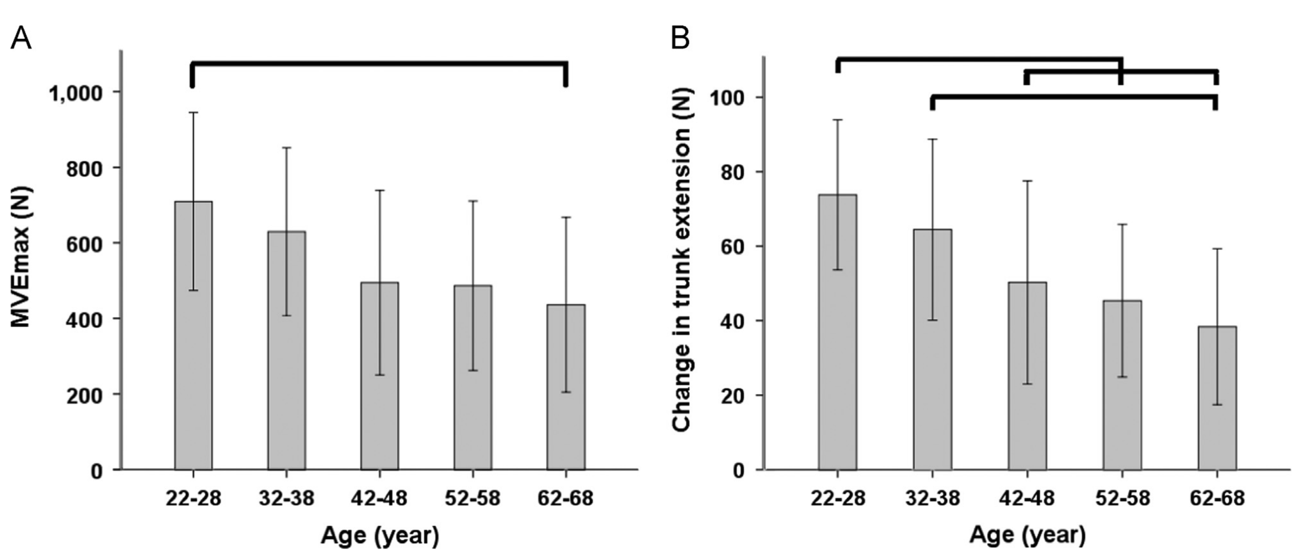

Fig. 3. (A) Mean maximum recorded forces during the MVE trials (MVE_{max}), and (B) the increase in extension forces from conditions with 20% to 30% of MVE_{max} efforts with force feedback. Brackets indicate significant paired differences.

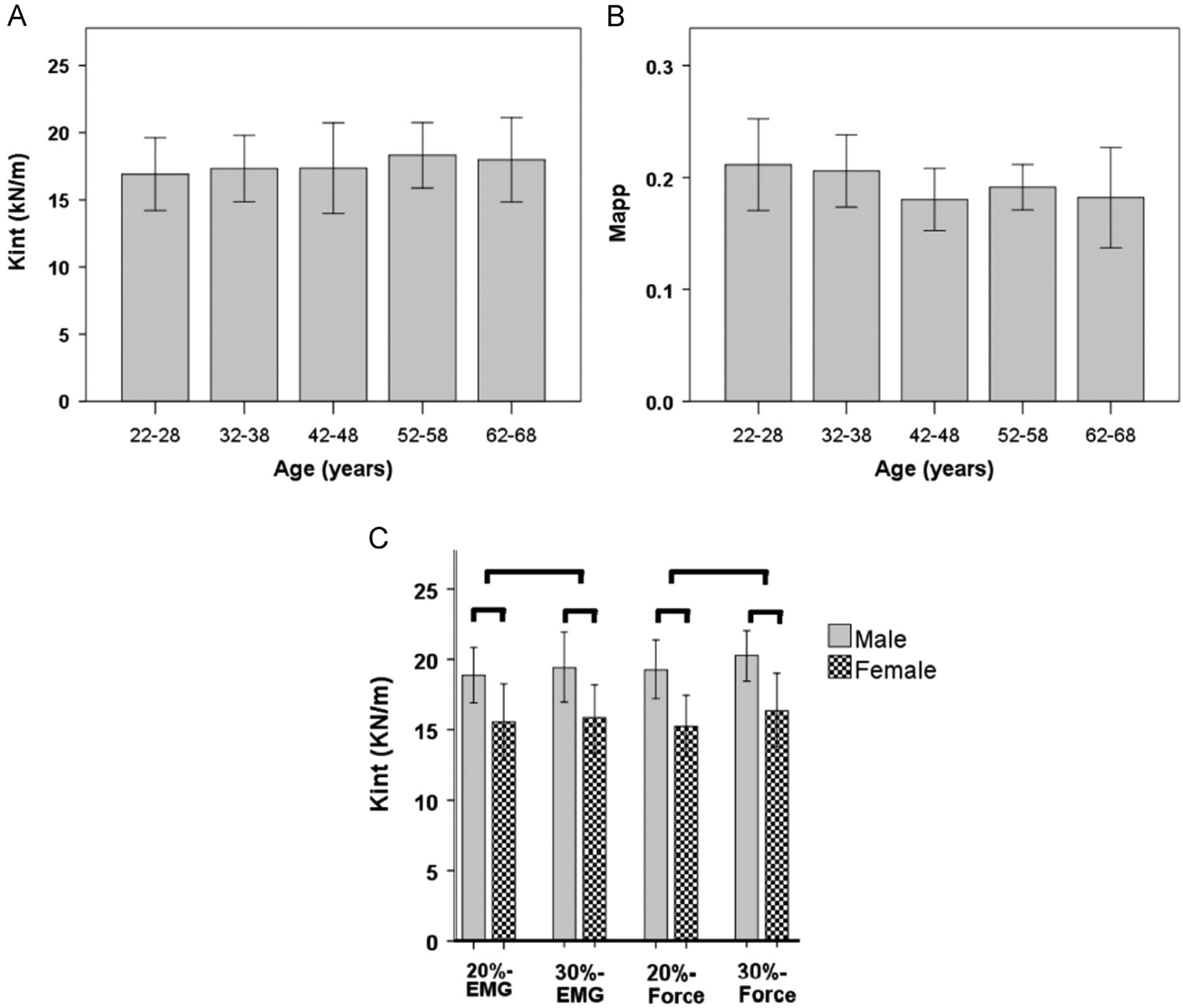

Fig. 4. (A) Estimated trunk intrinsic stiffness (K_{int}) in each age group. (B) Normalized apparent mass (M_{app}) in each age group. C) Trunk intrinsic stiffness (K_{int}) for genders and the level of extension efforts. The level of extension efforts were 20% and 30% of MVE_{max} values and were guided either by EMG (i.e., 20%-EMG, 30%-EMG) or force (20%-force, 30%-force) feedback. All values are means (SD), and significant paired differences between genders and levels of extension effort are highlighted in C).

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 and Burton, 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., in press), 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 to spinal stability (i.e., quantified via changes in trunk intrinsic stiffness and apparent mass) was found to be consistent across a range of participant ages. Therefore, deterioration of spinal stability due to age-related changes in volitional trunk neuromuscular behavior is not 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.

Conflict of interest statement

We declare that all authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented therein.

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References

Adams, M.A., Burton, K., 2006. The Biomechanics of Back Pain. Churchill Livingstone, Edinburgh.
Andersen, T.B., Essendrop, M., Schibye, B., 2004. Movement of the upper body and muscle activity patterns following a rapidly applied load: the influence of pre-load alterations. Eur. J. Appl. Physiol. 91 (4), 488–492.
Bazrgari, B., Hendershot, B., Muslim, K., Toosizadeh, N., Nussbaum, M.A., Madigan, M.L., 2011. Disturbance and recovery of trunk mechanical and neuromuscular behaviours following prolonged trunk flexion: influences of duration and external load on creep-induced effects. Ergonomics 54 (11), 1043–1052.
Bazrgari, B., Nussbaum, M.A., Madigan, M.L., 2012. Estimation of trunk mechanical properties using system identification: effects of experimental setup and modelling assumptions. Comput. Methods Biomech. Biomed. Eng. 15 (9), 1001–1009.
Becker, C.K., Savelberg, H.H., Barneveld, A., 1994. In vitro mechanical properties of the accessory ligament of the deep digital flexor tendon in horses in relation to age. Equine Vet. J. 26 (6), 454–459.
Bergmark, A., 1989. Stability of the lumbar spine. A study in mechanical engineering. Acta Orthop. Scand. Suppl. 230, 1–54.
Burdorf, A., Sorock, G., 1997. Positive and negative evidence of risk factors for back disorders. Scand J Work Environ Health 23 (4), 243–256.
Campbell, M.J., McComas, A.J., Petito, F., 1973. Physiological changes in ageing muscles. J. Neurol. Neurosurg. Psychiatry 36 (2), 174–182.
Cholewicki, J., Simons, A.P., Radebold, A., 2000. Effects of external trunk loads on lumbar spine stability. J. Biomech. 33 (11), 1377–1385.
Deyo, R.A., Mirza, S.K., Martin, B.I., 2006. Back pain prevalence and visit rates: estimates from U.S. national surveys, 2002. Spine (Phila Pa 1976) 31 (23), 2724–2727.
Edstrom, L., Larsson, L., 1987. Effects of age on contractile and enzyme-histochemical properties of fast- and slow-twitch single motor units in the rat. J. Physiol. 392, 129–145.
Evans, W.J., 1995. What is sarcopenia? J. Gerontol. A Biol. Sci. Med. Sci. 50 (spec No.: 5–8).
Faulkner, J.A., Larkin, L.M., Claflin, D.R., Brooks, S.V., 2007. Age-related changes in the structure and function of skeletal muscles. Clin. Exp. Pharmacol. Physiol. 34 (11), 1091–1096.
Galbusera, F., van Rijsbergen, M., Ito, K., Huyghe, J.M., Brayda-Bruno, M., Wilke, H.J., 2014. Ageing and degenerative changes of the intervertebral disc and their impact on spinal flexibility. Eur. Spine J. 23 (Suppl. 3), S324–S332.
Hendershot, B., Bazrgari, B., Muslim, K., Toosizadeh, N., Nussbaum, M.A., Madigan, M.L., 2011. Disturbance and recovery of trunk stiffness and reflexive muscle responses following prolonged trunk flexion: influences of flexion angle and duration. Clin. Biomech. 26 (3), 250–256.
Hendershot, B.D., Bazrgari, B., Nussbaum, M.A., 2013. Persons with unilateral lower-limb amputation have altered and asymmetric trunk mechanical and neuromuscular behaviors estimated using multidirectional trunk perturbations. J. Biomech. 46 (11), 1907–1912.
Hodges, P., van den Hoorn, W., Dawson, A., Cholewicki, J., 2009. Changes in the mechanical properties of the trunk in low back pain may be associated with recurrence. J. Biomech. 42 (1), 61–66.
Hunter, S.K., Thompson, M.W., Adams, R.D., 2000. Relationships among age-associated strength changes and physical activity level, limb dominance, and muscle group in women. J. Gerontol. A: Biol. Sci. Med. Sci. 55 (6), B264–B273.
Iida, T., Abumi, K., Kotani, Y., Kaneda, K., 2002. Effects of aging and spinal degeneration on mechanical properties of lumbar supraspinous and interspinous ligaments. Spine J. 2 (2), 95–100.
Johannes, C.B., Le, T.K., Zhou, X., Johnston, J.A., Dworkin, R.H., 2010. The prevalence of chronic pain in United States adults: results of an internet-based survey. J. Pain 11 (11), 1230–1239.
Katz, J.N., 2006. Lumbar disc disorders and low-back pain: socioeconomic factors and consequences. J. Bone Joint Surg. Am. 88 (Suppl. 2), S21–S24.
Kent-Braun, J.A., Ng, A.V., Young, K., 2000. Skeletal muscle contractile and non-contractile components in young and older women and men. J. Appl. Physiol. (1985) 88 (2), 662–668.
Krajcarski, S.R., Potvin, J.R., Chiang, J., 1999. The in vivo dynamic response of the spine to perturbations causing rapid flexion: effects of pre-load and step input magnitude. Clin. Biomech. 14 (1), 54–62.
Lee, P.J., Granata, K.P., Moorhouse, K.M., 2007. Active trunk stiffness during voluntary isometric flexion and extension exertions. Hum. Factors 49 (1), 100–109.

Lee, P.J., Rogers, E.L., Granata, K.P., 2006. Active trunk stiffness increases with co-contraction. J. Electromyogr. Kinesiol. 16 (1), 51–57.
Lexell, J., Henriksson-Larsen, K., Winblad, B., Sjostrom, M., 1983. Distribution of different fiber types in human skeletal muscles: effects of aging studied in whole muscle cross sections. Muscle Nerve 6 (8), 588–595.
Lexell, J., Taylor, C.C., Sjostrom, M., 1988. What is the cause of the ageing atrophy? Total number, size and proportion of different fiber types studied in whole vastus lateralis muscle from 15- to 83-year-old men. J. Neurol. Sci. 84 (2–3), 275–294.
Miller, E.M., Bazrgari, B., Nussbaum, M.A., Madigan, M.L., 2013. Effects of exercise-induced low back pain on intrinsic trunk stiffness and paraspinal muscle reflexes. J. Biomech. 46 (4), 801–805.
Moorhouse, K.M., Granata, K.P., 2007. Role of reflex dynamics in spinal stability: intrinsic muscle stiffness alone is insufficient for stability. J. Biomech. 40 (5), 1058–1065.
Muslim, K., Bazrgari, B., Hendershot, B., Toosizadeh, N., Nussbaum, M.A., Madigan, M.L., 2013. Disturbance and recovery of trunk mechanical and neuromuscular behaviors following repeated static trunk flexion: influences of duration and duty cycle on creep-induced effects. Appl. Ergon. 44 (4), 643–651.
Panjabi, M.M., 1992a. The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement. J. Spinal Disord. 5 (4), 383–389 (discussion 397).
Panjabi, M.M., 1992b. The stabilizing system of the spine. Part II. Neutral zone and instability hypothesis. J. Spinal Disord. 5 (4), 390–396 (discussion 397).
Panjabi, M.M., 2003. Clinical spinal instability and low back pain. J. Electromyogr. Kinesiol. 13 (4), 371–379.
Shahvarpour, A., Shirazi-Adl, A., Mecheri, H., Lariviere, C., 2014. Trunk response to sudden forward perturbations-effects of preload and sudden load magnitudes, posture and abdominal antagonistic activation. J. Electromyogr. Kinesiol. 24 (3), 394–403.
Shojaei I., Arjmand N., Bazrgari B., An optimization-based method for prediction of lumbar spine segmental kinematics from the measurements of thorax and pelvic kinematics. Int. J. Numer. Method Biomed. Eng. http://dx.doi.org/10.1002/cnm.2729, in press.
Solomonow, M., 2006. Sensory-motor control of ligaments and associated neuromuscular disorders. J. Electromyogr. Kinesiol. 16 (6), 549–567.
Stokes, I.A., Fox, J.R., Henry, S.M., 2006. Trunk muscular activation patterns and responses to transient force perturbation in persons with self-reported low back pain. Eur. Spine J. 15 (5), 658–667.
Stokes, I.A., Gardner-Morse, M., Henry, S.M., Badger, G.J., 2000. Decrease in trunk muscular response to perturbation with preactivation of lumbar spinal musculature. Spine (Phila Pa 1976) 25 (15), 1957–1964.
Toossi, M., 2012. Labor force projections to 2020: a more slowly growing workforce. Montly Labor Review, pp. 43–64.
Umehara, S., Tadano, S., Abumi, K., Katagiri, K., Kaneda, K., Ukai, T., 1996. Effects of degeneration on the elastic modulus distribution in the lumbar intervertebral disc. Spine (Phila Pa 1976) 21 (7), 811–819 (discussion 820).
Yassierli, M.A., Nussbaum, H., Iridiastadi, Wojcik, L.A., 2007. The influence of age on isometric endurance and fatigue is muscle dependent: a study of shoulder abduction and torso extension. Ergonomics 50 (1), 26–45.