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IIE Transactions on Occupational Ergonomics and Human Factors

ISSN: 2157-7323 (Print) 2157-7331 (Online) Journal homepage: http://www.tandfonline.com/loi/uehf20

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To cite this article: Nima Toosizadeh & Maury A. Nussbaum (2014) Trunk Tissue Creep Can Increase Spine Forces During a Subsequent Lifting Task, IIE Transactions on Occupational Ergonomics and Human Factors, 2:2, 71-82, DOI: 10.1080/21577323.2014.956944

To link to this article: http://dx.doi.org/10.1080/21577323.2014.956944

Accepted author version posted online: 30 Aug 2014.
Published online: 30 Aug 2014.

IIE Transactions on Occupational Ergonomics and Human Factors, (2014), 2: 71–82

ISSN: 2157-7323 print / 2157-7331 online
DOI: 10.1080/21577323.2014.956944

ORIGINAL RESEARCH

Trunk Tissue Creep Can Increase Spine Forces During a Subsequent Lifting Task

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Received April 2014
Accepted August 2014

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OCCUPATIONAL APPLICATIONS Results from the current study show evidence of an adverse effect of prolonged trunk flexion on spine loads during consecutive lifting tasks. The time-dependent methodology introduced here can enhance task assessment based on the duration of flexion exposures. More generally, results demonstrate the importance of considering prior trunk exposures when assessing risk factors for lifting tasks. The proposed solution incorporates "time" as an independent variable, in addition to lifting weight and posture, to better assess spinal loads and maximum lifting capacity based on prior loadings. The current results also suggest that existing ergonomic guidelines or biomechanical models that do not incorporate the viscoelasticity of soft tissues or time-dependent neuromuscular alterations may underestimate spine forces and potential injury risk in some circumstances. To account for this, especially when assessing spine forces during lifting after exposure to prolonged flexion, additional safety margins should be considered.

TECHNICAL ABSTRACT ***Background:*** Prolonged trunk flexion decreases soft tissue stiffness due to viscoelastic deformations and can also lead to altered kinematics when performing a subsequent lifting task. Yet, it remains to be determined if or how these changes and alterations might increase spine forces. ***Purpose:*** A previously developed viscoelastic model was used, along with experimental data, to predict changes in peak spine forces during a lifting task performed following a prolonged flexion exposure (creep). ***Methods:*** Model inputs were obtained from an experiment using ten participants, within which lifting kinematics and muscle activity were measured both before and after creep exposure. Two sets of simulations were performed; one in which kinematics were assumed to be unchanged by creep exposure and the other incorporating measured changes in kinematics following exposure. ***Results:*** Post-exposure changes in lifting kinematics involved a reduction in the peak relative sagittal-plane flexion of superior lumbar motion segments and an increase in these flexion among inferior lumbar motion segments. Creep exposure caused increases in predicted peak spine forces during lifting at all levels of the lumbar spine (65–241 N). A substantial portion (∼51%) of this increase was estimated to be the result of muscular compensations for reduced passive tissue stiffness. ***Conclusions:*** The current study demonstrates that both changes in lifting kinematics and viscoelastic deformations resulting from creep exposures can lead to increased trunk muscle forces and spine forces during a lifting task. This evidence suggests a potential mechanical basis for previous epidemiological evidence that indicates an increased risk of low back disorders for jobs involving both trunk flexion and lifting.

KEYWORDS Creep, lifting, viscoelastic, modeling, kinematics, spine, muscle activity

INTRODUCTION

Trunk flexion combined with lifting tasks have stronger associations with low back disorder (LBD) risk compared to other occupationally related physical exposures (Burdorf & Sorock, 1997; Hoogendoorn et al., 1999; Marras, 2000; Nelson & Hughes, 2009). However, it is difficult to specify the separate effects of such exposures, since many occupational tasks (e.g., in transportation, construction, plumbing, and mining) involve combinations of trunk flexion, lifting, and other potential exposures. With respect to trunk flexion and lifting, it has been suggested that performing a lifting task following a period of trunk flexion can expose the trunk to a substantially higher risk of an LBD (McGill, 2007). This suggestion was based on the reported viscoelastic behavior of trunk soft tissues and muscle spasms following prolonged trunk flexion. To improve our understanding of LBD risk, it is important to investigate potential interactions between risk factors. Of specific interest here is the effect of creep exposure on trunk muscle activity and trunk mechanical behaviors when performing a subsequent lifting task.

Previous work has found altered trunk tissues behaviors (mechanical and neuromuscular) following creep exposures. These alterations include reduced passive stiffness of the trunk (McGill & Brown, 1992; Shin & Mirka, 2007; Bazrgari et al., 2011; Toosizadeh et al., 2012, 2013), changes in the relative sagittal flexion of the lumbar spine and hip (Marras & Granata, 1997), and increases in trunk muscle activity (Shin & Mirka, 2007; Shin et al., 2009; Bazrgari et al., 2011). Such changes may, in turn, result in additional loads on spinal motions segments and other soft tissues and a consequent increase in the risk of LBDs. Since assessing spine forces is challenging, especially using direct methods, computational biomechanical modeling is a common alternative (Stokes & Gardner-Morse, 1995; Arjmand & Shirazi-Adl, 2006).

Use of biomechanical models becomes challenging, however, when predicting time-dependent changes in spine forces, since such predictions require that the viscoelastic properties of soft tissues be accurately characterized. Several studies have modeled soft tissues viscoelastic behavior to explore time-dependent kinematics/kinetics, specifically for spinal motion segments (Li et al., 1995; Holmes & Hukins, 1996; Silva et al., 2005; Wang et al., 2005; Groth & Granata, 2008; Schmidt et al., 2010) and passive muscle components (Abbott & Lowy, 1957; Glantz, 1974; Greven & Hohorst, 1975; Sanjeevi, 1982; Taylor et al., 1990: Hedenstierna et al., 2008). Among these, Kelvin-solid models are preferable for predicting viscoelastic behaviors of trunk soft tissues and time-dependent changes in spine forces due to the convenient definition of material properties for these models and their relatively low computational costs (e.g., versus poroelastic models). Yet, it remains for these approaches to be incorporated into a biomechanical model of the trunk to facilitate assessment of time-dependent changes in spine forces.

The current study assessed the effect of prolonged trunk flexion at a constant external moment (creep) on spine forces. Here, for the first time, data were obtained from participants during actual lifting and combined with a novel viscoelastic model to assess any alterations in lifting kinematics and trunk muscle activity and subsequent alterations in spine forces before versus after creep exposure. Based on previous evidence (Shin & Mirka, 2007; Shin et al., 2009; Bazrgari et al., 2011), an increase in active trunk muscle forces and spine forces was hypothesized during a lifting task subsequent to creep exposures as a result of viscoelastic deformation. Also explored was whether changes in lifting kinematics occur after trunk creep exposure in terms of the peak relative flexion of the lumbar spine and hip and/or the peak relative flexion of lumbar motion segments. Overall, the current study aimed to provide evidence for a potential low back injury mechanism involved during occupational tasks that require both trunk flexion and lifting.

METHODS

A scalable, multi-segment model of the upper body (developed previously) was modified to account for creep responses, which was then used to estimate muscle forces and spine forces during lifting tasks before and after creep exposure. A kinematics-driven approach was used to estimate muscle and spine forces during the lifting tasks. For this, kinematics were obtained experimentally and used as model input. Below, the model development and evaluation procedure is explained briefly; for more details, readers are referred to Toosizadeh and Nussbaum (2014).

Modeling Approach

A sagittally symmetric scalable model was used, containing six sagittally deformable lumbar motion segments (T12-L1 through L5-S1) and passive posterior muscle components (18 "local" and two "global" muscles; global muscles–ICPT: iliocostalis lumborum pars thoracic, LGPT: longissimus thoracis pars thoracic; local muscles–ICPL: iliocostalis lumborum pars lumborum [four fascicles from L1 to L5], LGPL: longissimus thoracis pars lumborum [five fascicles from L1 to S1], MF: multifidus [five fascicles from L1 to S1], and QL: quadratus lumborum [four fascicles from L1 to L5]). Only passive components of posterior muscles were modeled, since passive contributions of abdominal muscles were assumed negligible for tasks involving trunk flexion. However, the active effects of abdominal muscles (i.e., active co-contraction forces) were considered in estimating spine forces by assuming a constant level of co-contraction that was set at 1.7% of the maximum contractile force for each abdominal muscle (Arjmand & Shirazi-Adl, 2006). A wrapping mechanism was used to represent changes in global muscle paths with trunk flexion as in previous work (Arjmand et al., 2006).

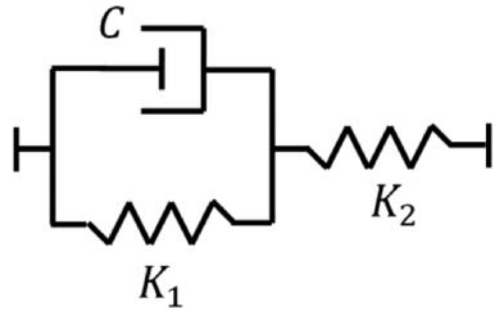

FIGURE 1 SNS model representation of intervertebral discs and passive muscles. Here K_1 and C are the respective stiffness and damping of a torsional/linear spring and damper components in parallel (Kelvin component), and K_2 is the stiffness of an in-series torsional/linear spring (Roylance, 2001). K_1 and C represent viscous responses to deformation, $K_1 + K_2$ is the steady-state stiffness once the material is totally relaxed, and K_2 is the instantaneous stiffness. The creep angle-time equation for the SNS model at a constant external moment of M_0 is

$$\theta(t) = \frac{M_0}{K_1}\left(\frac{K_1 + K_2}{K_2} - e^{-\left(t \times \frac{K_1}{C}\right)}\right).$$

Standard nonlinear solid (SNS, Fig. 1) components were used to model axial and rotational stiffnesses of each lumbar motion segment (using a "lumped" model) and muscle stiffnesses along the lines-of-action. Nonlinear elastic properties (K_2 as in Fig. 1) were defined using existing data (McCully & Faulkner, 1983; Panjabi et al., 1994; Guan et al., 2007; Bazrgari et al., 2008) for each SNS component. Viscous properties were defined using previous data for lumbar motion segments (Twomey & Taylor, 1982; Oliver & Twomey, 1995; Little & Khalsa, 2005) and trunk muscles (Glantz, 1974; Sanjeevi, 1982; Ryan et al., 2010; Ryan et al., 2011). These data were used to relate viscous parameters (K_1 and C) of each SLS component with the elastic parameter (K_2) within the current model (Table 1). Of note, this approach was used to

TABLE 1 Estimated relationships between viscous (K_1 and C) and elastic (K_2) components of spinal motion segments and muscles used in the biomechanical model

Spinal level	Sagittal flexion, K_1/K_2	Sagittal flexion, C/K_2 (1/sec)	Axial deformation, K_1/K_2	Axial deformation, C/K_2 (1/sec)
T12-L1	2.8	303.0	4.8	3703.7
L1-L2	2.3	370.4	4.1	6250.0
L2-L3	2.0	434.8	4.8	5882.4
L3-L4	1.8	454.6	3.2	1960.8
L4-L5	1.7	454.6	6.0	3846.2
L5-S1	1.6	476.2	6.0	3846.2
Muscles	—	—	3.8	31.5

approximate the viscoelastic properties of SLS components, since available data were not sufficient to define creep behaviors of all spinal motion segments and muscle groups at different loading magnitudes. The developed viscoelastic model was then used in combination with a kinematics-driven approach and an optimization algorithm to estimate muscle and spine forces. The objective of the optimization algorithm was to minimize the sum of cubed muscles stresses ($\min\{\sum_{n}^{i=1}\sigma_i\}$), with constraints of $0 \leq \sigma_i \leq \sigma\ \max_i$, where n is the number of muscles at each level of the spine, σ_i is the stress in the ith muscle, and $\sigma\max_i$ is the maximum tensile capacity of the ith muscle. A similar posterior muscle architecture was used as in the passive model but here with the inclusion of abdominal muscles (i.e., rectus abdominus, internal oblique, and external oblique). Input data for the model were the kinematics and kinetics of lifting, including lumbar motion segment rotations and the position and weight of the load (Fig. 2), all of which were measured during a lifting experiment (see below). Using an iterative procedure, muscle forces were estimated that satisfied moment equilibrium at each lumbar level, with iterations continued at each time step until convergence of estimated forces. Of note, spine forces predicted from the current model (before flexion exposure) were found comparable with *in vivo* measurements for a range of flexion angles and lifting conditions, and specific comparisons suggested <2% error in estimating spine forces at the L5/S1 level (readers are referred to Toosizadeh & Nussbaum (2014) for more details regarding model evaluation).

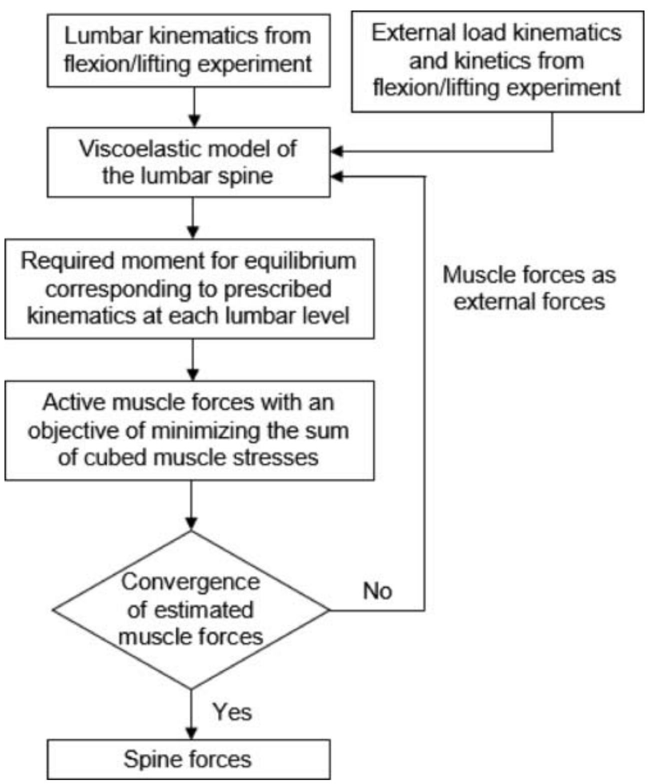

FIGURE 2 Modeling procedures for spine force estimation.

Model Evaluation

Experimental creep results (Toosizadeh & Nussbaum, 2013) were used to adjust/evaluate model-predicted viscoelastic behaviors. Briefly, a 6-minute period of prolonged trunk flexion with extra loads in the hand (overall weight of 84 N for males, and 54 N for females) was simulated in the model to elicit relatively large magnitudes of creep deformation. Creep angle time curves of the trunk from the model and the experiment were compared using fitted SNS models and material properties of spinal motion segments in sagittal flexion, and passive muscle components were modified within an iterative procedure. Coefficients of determination (R^2) and root-mean-square errors (RMSE) were obtained for the 6-minute creep exposure from linear regression of creep angle time curves from the model versus the experiment.

Flexion/Lifting Kinematics

Kinematics and external kinetics of the pelvis, trunk, lumbar motion segments, and upper extremities were obtained from participants when performing dynamic stoop lifting tasks, both before and after controlled creep exposures. Ten healthy young adults with no self-reported history of low-back pain or any current medical conditions completed the study after providing informed consent. All experimental procedures were approved by the Virginia Tech Institutional Review Board. Participants included six males with respective mean (*SD*) age, stature, and body mass of 24 (3) years, 183.7 (6.1) cm, and 81.2 (6.7) kg, respectively; corresponding values for the four females were 25 (3) years, 166.9 (5.6) cm, and 66.1 (7.4) kg, respectively. Prolonged full flexion (i.e., creep exposure) was induced by participant's flexing their trunk slowly to full passive trunk flexion, remaining in this flexed posture for 6 minutes, then slowly returning to the upright standing posture. During this, two additional loads with total weight of 84 N for males and 54 N for females were attached to wrists. These weights were selected based on relative lifting capacity between genders in high-risk manual material handling (Marras et al., 1993; Mital, 1984; Snook & Ciriello, 1991) and also to be able to evaluate the current model using previous experimental results (Toosizadeh & Nussbaum, 2013). Flexion tasks were performed while participants stood in a rigid metal frame, and straps were used to restrain the pelvis. Flexion relaxation of the extensor muscles during flexion exposures were confirmed using Electromyography (EMG) data (see below), which also ensured minimal effects of muscle fatigue.

During the lifting tasks, participants were asked to lift a box, with mass set to 25% of individual body mass, from an adjustable platform at the knee height. Five lifts were done before creep exposure, with a 2-minute rest between lifts and one immediately after creep exposure. The first four pre-exposure lifts were used to provide a warm up and familiarization to the task; only the final (fifth) pre-exposure lifts were used as described later. During the lifting tasks, kinematics were tracked via reflective markers (120 Hz) using a seven-camera motion capture system (Vicon MX, Vicon Motion Systems Inc., Denver, CO, USA). Markers were placed in the mid-sagittal plane over all vertebrae processes from T12 to S1 and bilaterally over the ASIS, acromial processes, lateral humeral epicondyles, radialis, styloid processes, and second metacarpal heads. Markers were also placed on the box to track the mass center (Fig. 3). An existing transformation model was then used to estimate lumbar curvature using skin markers as follows (Lee et al., 1995):

$$v = a_1 s^3 + a_2 s^2 + a_3 s + a_4 f + a_5 d + a_0, \tag{1}$$

where v is the coordinate of the vertebral body in the x, y, or z directions; s is the coordinate of the marker in the x, y, or z directions; f is the L4 skin-fold thickness; d is the amount of L1-S1 skin distraction in the upright and flexed postures (to account for skin movement under the markers relative to bony landmarks); and a_i denotes constants (see Lee et al. [1995] for more details). By connecting the centers of adjacent motion segments, initial angles for adjacent vertebrae were estimated; changes in these angles were determined over time.

EMG of the paraspinal muscles (trunk extensors) and trunk flexors on the right side were measured using bipolar Ag/AgCl surface electrodes to explore

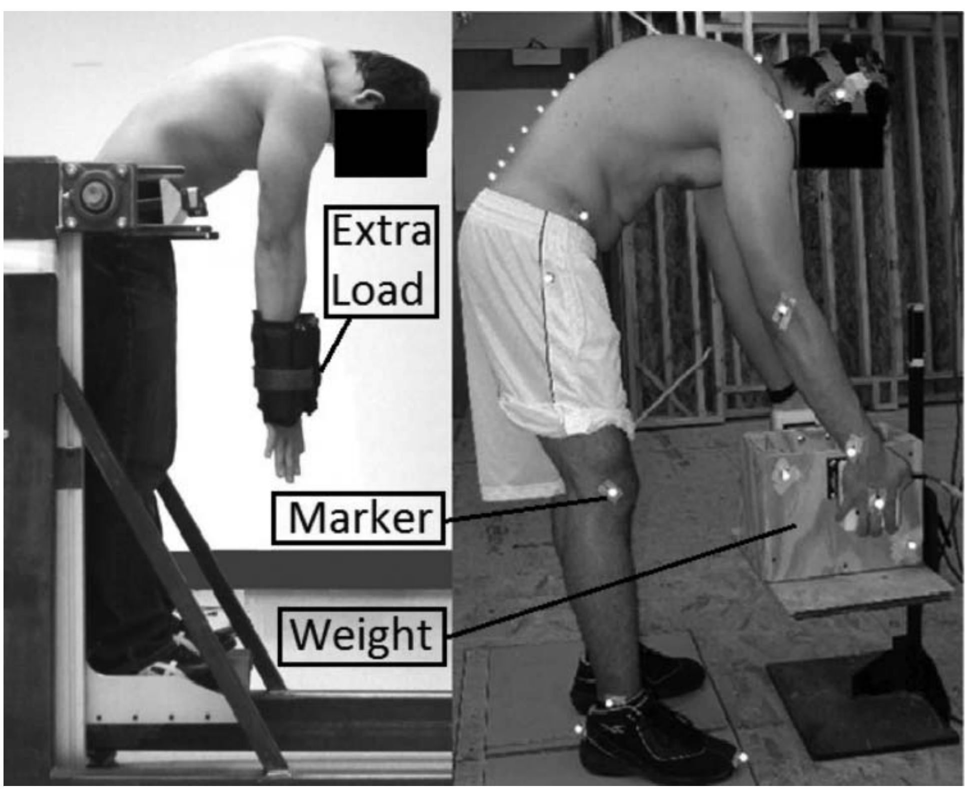

FIGURE 3 Experimental setup for creep exposure (left) and lifting (right).

the force distribution among superficial trunk muscles. Electrode placements followed earlier protocols (El-Rich et al., 2004; McGill, 2005), targeting the iliocostalis lumborum pars thoracic, longissimus thoracis pars thoracic, iliocostalis lumborum pars lumborum, longissimus thoracis pars lumborum, multifidus, rectus abdominus, internal oblique, and external oblique. Prior to applying electrodes, the skin was prepared using abrasion and cleaned with alcohol; raw EMG signals were pre-amplified ($\times$100) near the collection site, then bandpass filtered (10–500 Hz), amplified, and converted to root-mean-square (RMS) in hardware (Measurement Systems Inc., Ann Arbor, MI, USA) to minimize motion artifacts and other sources of noise. RMS values were then normalized (nEMG) to peak EMG RMS obtained from maximum voluntary contractions (MVCs) performed by each participant. Six trials of MVCs (three trials each in extension and flexion) were performed in standing posture. During MVCs, a rigid frame was used to constrain the lower limbs, pelvis, and trunk. Participants pulled back/pushed forward maximally for 5 seconds. Paired *t*-tests were used to assess changes (before versus after creep exposure) in peak values of relative lumbar motion segment flexion (sagittal plane, as percent of total lumbar flexion), total lumbar flexion, hip flexion, and trunk muscle nEMG. Statistical analyses were done using JMP™ (Version 8, SAS Institute Inc., Cary, NC, USA), and significance was concluded when $p < 0.05$.

Flexion/Lifting Simulations

Both creep exposure and lifting tasks were simulated in the model. Two sets of lifting simulations were performed using a quasi-static analysis, since inertial loads were relatively small (mean lifting duration was ~3 seconds). The first assumed similar kinematics for the lifting tasks done prior to and after creep exposure (identical kinematics) by using the pre-exposure kinematics as input. Specifically, mean lifting kinematics across participants from the last pre-exposure lifting tasks were used as input to the model, and the model was scaled based on individual anthropometry. The second considered (accounted for) changes in kinematics due to creep exposure (modified kinematics). In this, kinematics from the final pre-exposure lifting task were used along with kinematics from the single post-exposure lift. These two sets of simulations using the two datasets were performed to explore the separate effects of creep deformation and altered kinematics on any changes in spine forces following creep exposure. An additional set of simulations was done using the modified kinematics and with only elastic (rather than viscoelastic) properties incorporated in the model. These latter simulations were used to isolate the effects of creep deformation on spine forces. From these three sets of simulations, the separate contributions of kinematic changes and creep deformation, and the joint contribution of both factors, were identified. The reason for performing separate simulations with different modeling and kinematic input conditions was to assess the extent of error that might be introduced in estimating spine forces if one of the factors is neglected (e.g., using a simplified biomechanical model or with limited kinematics input). Outcome measures obtained from the model for a given lifting task were peak spine forces at each level of the lumbar spine and peak muscle forces.

RESULTS

Experiment Results: Kinematics and nEMG

Peak relative flexion of the lumbar motion segments changed between lifting tasks performed before versus after creep exposure. Post-exposure, the flexion of superior (i.e., T12-L1 and L1-L2) and inferior motion segments (i.e., L4-L5 and L5-S1) of the lumber spine were estimated to significantly decrease ($p < 0.025$) and increase ($p < 0.027$), respectively (Fig. 4). No significant changes, however, were observed for the L2-L3 and L3-L4 motion segments ($p > 0.540$). Overall results suggest that lumbar flexion (T12-S1) decreased post-exposure (from 64.1° to 61.1°, $p = 0.029$), whereas hip flexion increased (from 37.4° to 39.8°, $p = 0.003$). Peak nEMG increased in all muscle groups following creep exposures (Fig. 5), though only some of these changes were statistically significant.

Model-Based Results: Creep, Spine Forces, and Muscle Forces

Comparison between the creep angle time curves from the model and experimental results during creep

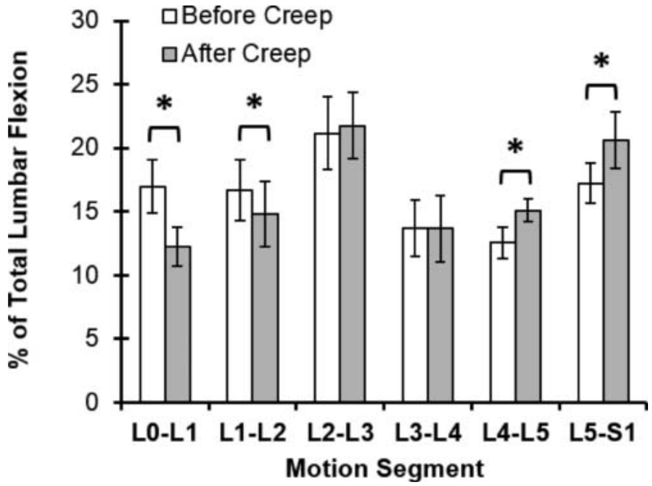

FIGURE 4 Changes in peak relative flexion of lumbar motion segments during lifting tasks performed before and after creep exposure. Mean and standard error values are presented, and the asterisk indicates a significant post-exposure change.

exposure yielded respective R^2 and RMSE values of 0.99° and 0.35°. Peak spine forces, specifically compression and antero-posterior shear forces, were predicted to increase following creep exposure; such increases were found using both "identical" and "modified" lifting kinematics (Fig. 6). Across all lumbar levels, respective increases in peak spine forces were 3.4% and 17.0% larger in compression and shear when using the modified versus identical kinematics. The largest increase in compression force occurred at the L5-S1 motion segment, which increased by 233 N following creep exposure; antero-posterior shear force had the largest

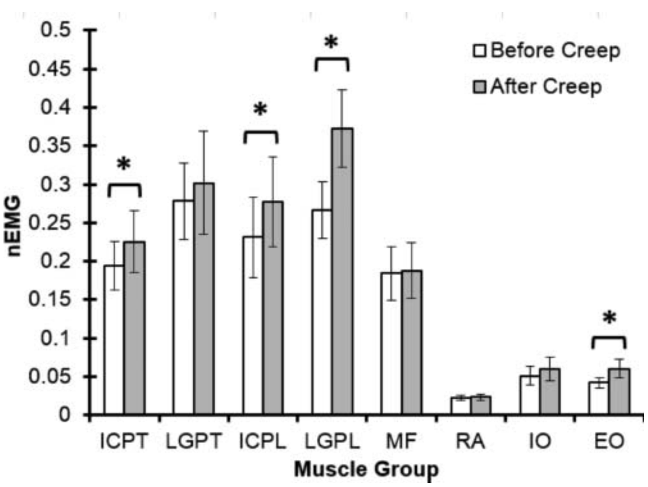

FIGURE 5 Increase in muscle activity (nEMG) during lifting tasks performed before and after creep exposure. Mean and standard error values are presented, and the asterisk indicates a significant post-exposure change.

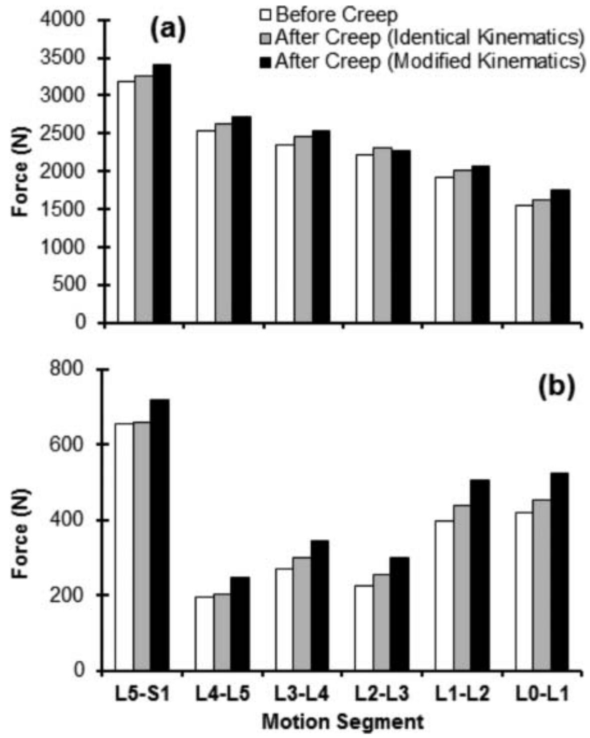

FIGURE 6 Increases during lifting tasks performed before and after creep exposures: (a) in predicted peak compression force and (b) in antero-posterior shear force; the latter predictions were done using identical and modified kinematics.

increase (106 N) at the L1-L2 motion segment. Peak muscle forces (i.e., summation of passive and active forces) increased when performing the lifting task following creep exposures (Table 2). Post-exposure increases in peak muscles forces were larger when the modified (9.3%) versus identical (3.3%) kinematics were used, and this increase mainly resulted from additional activity predicted in the global muscles. When using the model with elastic properties and modified kinematics, a 4% (133 N) increase in compression and 7% (~49 N) increase in antero-posterior shear forces were predicted at L5-S1 following creep exposure.

DISCUSSION

Viscoelastic deformation predicted by the model had a high correspondence with experimental results. The difference between model-predicted and experimental values of total creep angle was ~0.5°, or 8% of the total creep angle. As such, it is expected that estimated decreases in stiffness within trunk passive components was reasonably well predicted using the model.

In support of the study hypotheses, alterations in kinematics and spine forces during lifting were

TABLE 2 **Predicted changes in peak passive and active muscle forces (all values in Newton) during lifting tasks performed prior to and immediately following a simulated 6-minute creep exposure; muscles are listed at the level of origin**

		Before creep			After creep (identical kinematics)			After creep (modified kinematics)		
Spinal level	Muscle group	Active	Passive	Total	Active	Passive	Total	Active	Passive	Total
T12-L1	ICPT	666	70	736	696	70	766	764	68	832
	LGPT	1156	60	1216	1209	60	1269	1329	59	1388
L1-L2	ICPL	53	20	73	57	20	77	42	20	62
	LGPL	38	12	50	41	12	53	30	12	42
	MF	63	30	93	72	30	102	58	29	87
	QL	42	17	59	45	17	62	34	16	50
L2-L3	ICPL	42	20	62	52	20	72	19	21	40
	LGPL	24	12	36	26	12	38	7	13	20
	MF	49	28	77	49	27	76	10	29	39
	QL	21	17	38	21	16	37	4	16	20
L3-L4	ICPL	0	37	37	0	36	36	36	36	72
	LGPL	0	17	17	0	17	17	20	17	37
	MF	0	44	44	0	44	44	54	43	97
	QL	0	15	15	0	15	15	14	15	29
L4-L5	ICPL	7	39	46	0	39	39	0	42	42
	LGPL	4	19	23	0	19	19	0	21	21
	MF	8	39	47	1	39	40	0	43	43
	QL	2	14	16	0	14	14	0	15	15
L5-S1	LGPL	0	20	20	0	20	20	0	22	22
	MF	0	29	29	0	29	29	0	31	31

Global muscles: ICPT, iliocostalis lumborum pars thoracic; LGPT, longissimus thoracis pars thoracic. Local muscles: ICPL, iliocostalis lumborum pars lumborum; LGPL, longissimus thoracis pars lumborum; MF, multifidus; and QL, quadratus lumborum.

observed as a result of creep exposure. Post-exposure, the contribution of hip flexion to total trunk flexion during lifting increased, while total lumbar flexion decreased. This effect is consistent with previous work (Marras & Granata, 1997) that found qualitatively similar changes in hip and lumbar flexion when performing a lifting task following repetitive lifting. As suggested by these authors, a trade-off between hip and lumbar flexion may be used to decrease trunk external moments via reducing the moment arm between L5-S1 and the load and/or trunk masses. However, as was apparent here from the model results, these kinematic changes did not fully compensate for other mechanical alterations induced by creep exposure, and predicted spine forces were larger following creep exposure. Also, within the lumbar spine, there was a post-exposure reduction in the peak relative flexion of superior lumbar motion segments and an opposing increased flexion of inferior segments. Although between-subject variability of the magnitude of these kinematic changes was substantial, a similar qualitative pattern of kinematics changes was observed among all participants. While an increase in peak spine forces was found as a result of changes in the peak relative flexion of lumbar motion segments, the underlying mechanisms responsible for these changes, and/or any physiological benefits (or disadvantages), are not obvious.

For compression forces, predicted increases in peak values at the L5-S1 level were 2.5% (81 N) for identical kinematics and 7.3% (233 N) for modified kinematics. Of note, accounting for changes in lifting kinematics increased the levels of predicted antero-posterior shear force in superior motion segments up to ~27% (106 N). Overall, these creep-induced changes emphasize the importance of considering both viscoelastic deformations and kinematics alterations for task assessment. Accordingly, to explore the effect of viscoelasticity on model estimates of spine forces, the simulation was performed with only elastic material properties. Without viscoelasticity, a 4.3% increase in total spine forces (i.e., summation of compression and antero-posterior forces) was predicted only from changes in lifting kinematics. As such, using elastic biomechanical models may underestimate changes in spine forces following creep exposures. Such errors would be of particular importance when assessing tasks involving prolonged (or repetitive) loading.

An increase in peak predicted muscle forces was found during lifting tasks post-creep, and such increases were found for all simulation conditions. Passive muscle forces during lifting were predicted to decrease by only 0.7% (~4 N) when using identical kinematics. In support of previous investigations (Toosizadeh et al., 2012; Toosizadeh & Nussbaum, 2013), these observations suggest a predominant contribution of spinal motion segments in viscoelastic changes rather than passive muscle components. Additional trunk muscle activity was also directly observed here during lifting tasks performed after versus before creep exposure. Active muscle forces from the model and direct measurements (nEMG) indicated comparable levels of increased activity in global muscles post-exposure (respective increases of 14.8% and 12.3%). However, model predictions of local muscle activity were inconsistent with nEMG. Specifically, the model predicted a 7% decrease in active muscle forces post-creep, while nEMG increased by a mean of 21% across participants. As such, the model-based predictions of spine forces likely underestimated actual values. However, since the maximum force capacity of local muscles is relatively small (Marras et al., 2001), the magnitude of this underestimation is likely modest. Although the loading conditions were different, the overall nEMG increase of ~19% following flexion exposure is comparable with previous work that found a 16% increase in paraspinal muscle activity during lifting following cyclic flexion (Olson & Solomonow, 2009). In contrast, in another study, ~30% increases were found in erector spinae and multifidus nEMG after 6 minutes of trunk flexion exposure (Shin & Mirka, 2007). Differences in nEMG changes may have arisen from differences in lifting protocols. In the latter study, specifically, participants were asked to lift a box from their full trunk flexion angle, whereas lifting occurred from knee height in the present study.

An increased level of co-activity was observed when participants performed the lifting task following creep exposure, with a mean 22% increase in nEMG across all abdominal muscle groups. This increase, however, was statistically significant only for the external oblique, perhaps due to the large variability in nEMG. Additional co-activity can impose additional forces on lumbar motion segments and may therefore be an additional mechanism whereby creep exposure increases the risk of an LBD during a subsequent lifting task. Since a constant abdominal muscle co-activity was used in the model-based simulations, this study's approach was incapable of predicting changes in co-activity post-creep. Alternative approaches, such as EMG-assisted or optimization hybrid models, could be implemented to provide better predictions of changes in muscle activity. Since fatigue was controlled for during prolonged flexion exposures, the increased level of co-activity observed in the experiment likely resulted from neuromuscular alterations/adaptations. Previous work also suggested an increase in trunk muscle co-activity after prolonged trunk flexion (Hendershot et al., 2011), which may be a protective neuromuscular mechanism to increase trunk stiffness and stability (Lee et al., 2006).

As noted, there was substantial between-subject variability in observed changes of muscle activity (nEMG) when performing a lifting task following flexion exposure. At the extremes, one participant had slightly reduced muscle activity (~4%) following creep exposure, while for another, muscle activity increased more substantially (~40%). Notable differences in kinematics changes were evident when comparing the lifting kinematics for these particular individuals. The former had more substantial changes in lumbar motion segment flexion compared to other participants, specifically the highest post-exposure reduction in peak relative flexion in superior levels and the largest increase in peak relative flexion of inferior levels. For the latter participant (with a 40% increase in muscle activity), an increase in peak relative flexion in superior levels and a reduction in inferior levels was observed that was in contrast to other participants. In terms of LBD pathology, both types of flexion-induced alterations (i.e., increase in muscle activity and changes in lumbar motion segment flexion) may contribute to soft-tissue injury (McGill, 2007). Thus, the risk of LBD development may involve different mechanisms that result from individual differences in mechanical and neuromuscular responses to prolonged flexion.

Overall, some limitations exist for the current modeling approach, which should be considered when interpreting the outcomes. First, viscoelastic material properties in response to prolonged loading were derived here from human hamstring, rat tail, and papillary muscles (Glantz, 1974; Sanjeevi, 1982; Ryan et al., 2010, 2011), which were the only data available. The assumption of identical viscoelastic properties for all trunk extensor muscle groups likely introduced some errors. Further, viscoelastic responses of soft tissue to prolonged loading have not been reported for different loading magnitudes, and it was assumed that identical relationships exist between elastic and viscous SNS components at different loading magnitudes (as in Table 1). These properties were, however, evaluated and calibrated based on empirical measures (responses to creep exposures) and are considered to provide the best predictions of viscoelastic behaviors given available evidence. Second, use of an SNS model with a single retardation time constant does not capture the dual-phase creep responses of spinal motion segments (Toosizdeh & Nussbaum, 2014). This limitation could be addressed using more complex models, such as generalized-Kelvin solid models (Toosizadeh & Nussbaum, 2013), although experimental data are not yet available for defining relevant material properties. Third, predicted load distribution among different components of spinal motion segments (e.g., discs and ligaments) were not evaluated here, and they would be quite difficult to achieve in a non-invasive manner. Among these components, the contributions of ligaments to changes in spine forces were not considered. However, previous investigations have reported relatively small load bearing from ligaments during lifting tasks (Potvin et al., 1991; Cholewicki & McGill, 1992), and time-dependent changes in ligaments loads are likely negligible.

Further limitations are related to measured lifting kinematics; although the implemented method for estimating lumbar motion segment angles using skin markers was previously validated and incorporates skin distraction in the calculations, there is still likely a 0.05° to 0.56° error inherent in such estimations (Lee et al., 1995). The magnitude of such an error is, however, smaller than the observed changes in lumbar flexion angle following creep exposure. Furthermore, as evaluated in recent work (Zemp et al., 2014), using a surface marker set is considered appropriate for quantifying changes in lumbar angle in quasi-static conditions. Finally, the optimization approach used here allowed estimation of changes in muscle and spine forces following viscoelastic deformation. However, the underlying mechanisms of the neuromuscular system may be more complex and variable within/between individuals. For instance, differences in local and abdominal muscle activities observed here between model predictions and nEMG data may be due to implementing a more complex neuromuscular strategy to maintain spine stability (McGill, 2007). As such, in future studies, alternative modeling approaches, such as the use of "EMG-driven" models, may be of benefit.

In summary, the current study provides new evidence that exposure to trunk flexion changes mechanical loading during a subsequent lifting task as a result of kinematic alterations and viscoelastic deformation of the lumbar spine. An adjustment between the load sharing between active and passive tissues was apparent and was likely an adaptation to viscoelastic deformation. This adjustment led to increased contributions from active muscle force and, consequently, additional forces on spinal motion segments. As such, the current study provides evidence consistent with previous epidemiological studies that a combination of risk factors (here, prolonged trunk flexion and lifting) may contribute to the risk of LBD development. Models accounting for time-dependent effects of task demands and associate tissue responses may thus be of future benefit for job evaluation and design. Results here suggest that a more comprehensive evaluation is required when a task involves both prolonged flexion and lifting. More specifically, using purely elastic modeling or neglecting kinematic alterations in lifting due to viscoelastic deformation can lead to overestimation of load lifting capacity. Moreover, within the current study, high between-subject variability in kinematics and neuromuscular behaviors in response to extreme flexion exposure were found, which indicate a need for moving toward subject-specific modeling to improve task assessments and, ultimately, facilitate occupational injury prevention.

CONFLICT OF INTEREST

The authors declare no conflict of interest. Dr. Potvin served as Editor for this paper during the review process.

ACKNOWLEDGMENT

The authors thank Dr. Babak Bazrgari for his assistance in developing the finite-element model.

FUNDING

This work was supported by an award (R01 OH008504) from the Centers for Disease Control and Prevention (CDC). The contents are solely the responsibility of the authors and do not necessarily represent the official views of the CDC.

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