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The effect of unstable loading versus unstable support conditions on spine rotational stiffness and spine stability during repetitive lifting

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ABSTRACT

Lumbar spine stability has been extensively researched due to its necessity to facilitate load-bearing human movements and prevent structural injury. The nature of certain human movement tasks are such that they are not equivalent in levels of task-stability (i.e. the stability of the external environment). The goal of the current study was to compare the effects of dynamic lift instability, administered through both the load and base of support, on the dynamic stability (maximal Lyapunov exponents) and stiffness (EMG-driven model) of the lumbar spine during repeated sagittal lifts. Fifteen healthy males performed 23 repetitive lifts with varying conditions of instability at the loading and support interfaces. An increase in spine rotational stiffness occurred during unstable support scenarios resulting in an observed increase in mean and maximum Euclidean norm spine rotational stiffness ($p=0.0011$). Significant stiffening effects were observed in unstable support conditions about all lumbar spine axes with the exception of lateral bend. Relative to a stable control lifting trial, the addition of both an unstable load as well as an unstable support did not result in a significant change in the local dynamic stability of the lumbar spine ($p=0.5592$). The results suggest that local dynamic stability of the lumbar spine represents a conserved measure actively controlled, at least in part, by trunk muscle stiffening effects. It is evident therefore that local dynamic stability of the lumbar spine can be modulated effectively within a young-healthy population; however this may not be the case in a patient population.

1. Introduction

Goal directed occupational tasks often require the movement of unstable media (e.g. a pail of water), or the movement of stable media under unstable support conditions (e.g. lifting on ice). The human body is comprised of a multitude of unstable joints that must be controlled in the presence of these unstable external conditions. The stability of the spine in particular, and its ability to resist perturbations, is provided by active and passive tissues, with active tissues under the control of the central nervous system (CNS) (Bergmark, 1989; Panjabi, 1992; Cholewicki and McGill, 1996). If stability of the spinal column is impaired, uncontrolled perturbed intervertebral motions can result in inadequate transmission of compressive and shearing forces within the spinal column resulting in tissue strain and injury (Panjabi, 1992; Cholewicki and McGill, 1996; Granata and Marras, 2000). Previously, the local dynamic stability (LDS) of the spine has been quantified during unloaded dynamic movements (e.g. Granata and England, 2006) and simulated occupational lifting movements using maximal Lyapunov exponents (λ_{max}) (e.g. Graham et al., 2012; Graham and Brown, 2012). It has been suggested that kinematic trajectory-based LDS measures be paired with those of electromyography (EMG) based stiffness measures to quantify both the control of muscular based stiffness and the resulting kinematic stability in the dynamically moving spine (Graham and Brown, 2012). To the authors' knowledge, the effects of externally unstable lifting scenarios on the stability and stiffness of the lumbar spine have yet to be assessed within the scientific literature.

During gait tasks it has been shown that the CNS prioritizes the stability of superiorly oriented segments over those oriented inferiorly (Kang and Dingwell, 2009). Similarly, during simulated repetitive occupational lifting movements it has been suggested that inferior segments (foot, shank and thigh) can more successfully accommodate small perturbations (Graham et al., 2011). However, when small perturbations are increased in size it is expected that any postural adjustments will become more complex in nature and involve increased muscle activation and agonist/antagonist co-contraction (Brown et al., 2006; van Dieën et al., 2003). Strategies, such as increased trunk muscle co-contraction (active stiffness), have been shown to increase the velocities and displacements of an otherwise stable movement

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trajectory during unstable sitting (Reeves et al., 2006), while wearing unstable footwear (Buchecker et al., 2013) and during specialized ship-simulated lifting scenarios (Duncan et al., 2007; Matthews et al., 2007). Lumbar spine co-contraction and compression force levels have been shown to increase during lifting scenarios with liquid loads (van Dieën et al., 2001; van Dieën et al., 2003). However, only two studies (Duncan et al., 2007; Matthews et al., 2007) have investigated combined loading and support instability during lifting scenarios. Such postural strategies, however, have not been assessed in their capacity to modulate or maintain the LDS of the lumbar spine.

The purpose of this study was to assess the effects of instability applied through loading vs. support (hands vs. feet) interfaces on the LDS and rotational stiffness of the lumbar spine during a repetitive simulated lifting movement. It was hypothesized that both loading and support instability would result in an increased muscular-driven spine rotational stiffness, as well as a decreased lumbar spine LDS. This is contrary to the positive correlation between stiffness and LDS generated through manipulation of lifting mass reported in previous work (Graham and Brown, 2012); we suggest that an increased prevalence of neuromuscular control errors will accompany the unstable external conditions tested in the current study resulting in a decreased LDS. It was also hypothesized that when both instability scenarios were combined that this would result in the largest observed increase in spine rotational stiffness and decrease in local dynamic stability.

2. Materials and methods

2.1. Participants

Fifteen males (mean age 22 ± 2.6 years; height 1.8 ± 0.06 m; and mass 77.4 ± 10.6 kg) participated in the study. Exclusion criteria included persistent pain within the past year (causing absence from school, work or regular activity), or treatment for pain or injury in any region of the body. All participants were rested, and had not completed any intense physical activity 24 h prior to testing. The protocol was approved by the University Research Ethics Board.

2.2. Materials

As a means to elicit instability at both the loading and support interfaces two separate methods were employed. A BOSU® ball was used to perturb the lifter's stability from the support interface (Fig. 1c and d), whereas a liquid loading box was used to perturb lifters through the loading at the hands. The liquid loading box had a total volume of 11.5 L which was filled with 3.25 L of water to elicit an unstable load. Both de-stabilizing methods were chosen for their capacity to provoke a continuous challenge for the CNS to adapt at the support and loading interface respectively.

Surface EMG was collected bilaterally from the latissimus dorsi (LD), thoracic erector spinae (TES), lumbar erector spinae (LES), external oblique (EO), internal oblique (IO) and rectus abdominus (RA) muscles. Prior to placement of the surface electrodes (Blue Sensor, Medicotest Inc., Ølstykke, Denmark), skin sites were shaved and prepped with rubbing alcohol. Raw EMG signals were band-pass filtered from 10–1000 Hz, amplified (AMT-16, Bortec Calgary, AB, Canada) and captured digitally at 2048 Hz. Optoelectric kinematic data (Optotrak 3D Investigator, Northern Digital, Waterloo ON, Canada) were obtained from rigid bodies of three non-collinear markers placed on the thorax (T12) and sacrum (S1), and were sampled at 32 Hz. A simple mechanical push-button switch was installed into the base of the loading box to distinguish between repeated lifts for the rotational stiffness analysis.

2.3. Procedure

Prior to testing all participants completed an orientation session to allow for acclimation to lifting under unstable conditions, as well as lifting to a beat of a metronome. Upon completion of this session, EMG electrodes were placed on the bellies of the muscles of interest. Maximal voluntary isometric contractions (MVICs) were done for each muscle following protocols previously used throughout the scientific literature (e.g. Vera-Garcia et al., 2010). Each MVIC was completed three consecutive times, separated by 2 min of rest between contractions. The maximum rectified, filtered voltage (Section 2.4) from each set of MVIC trials was used to normalize EMG data for each respective muscle.

Following the MVIC trials, each participant was outfitted with the thoracic (T12) and sacral (S1) kinematic rigid bodies. Next, three separate calibration trials were performed including a (1) quiet standing bias trial, (2) spine flexion range of motion (ROM) trial and (3) static 45° loaded (10.2 kg) spine flexion trial. Data from these trials were used to (1) normalize kinematic data relative to a neutral lordotic spine curve, (2) tune the EMG-driven muscle model to maximum spine flexion ROM and (3) calibrate the EMG-driven model moment output to a known external moment.

Each lifting scenario was then randomly administered to each participant. The lifting scenarios consisted of a stable surface-stable load (SSSL) control trial (Fig. 1a and b), a stable surface-unstable load (SSUL) trial, an unstable surface-stable load (USSL) trial and an unstable surface-unstable load (USUL) trial (Fig. 1c and d). Each lifting trial was completed with a maintained participant-specific base of support, as well as a maintained lifting rate of 10 lifts/min (controlled via a metronome). The lifted load for each scenario was held at a constant of 8 kg with 40% of the load (3.25 kg) being replaced with water to elicit an unstable load. Each lifting scenario consisted of a total of 23 consecutive lifts (raising and lowering of the load). Shelf heights were set based on anatomical landmarks (lower shelf at the level of the tibial tuberosity and upper shelf at the level of the anterior superior iliac spine) for each participant in standing, and adjusted to compensate for added participant height when standing on the BOSU® (Fig. 1). Participants were instructed to lift the load as naturally as possible using a freestyle-type of lift. Participants were given 10 min of rest between lifting trials to avoid fatigue influences on outcome measures.

2.4. Data processing

Raw kinematic data from each of the lifting trials were low-pass filtered (4th order dual-pass Butterworth 10 Hz cut-off). Spine flexion-extension (FE), lateral bend (LB) and axial twist (AT) angles were obtained between thoracic and sacral rigid bodies using a Cardan rotational sequence (FE, LB, and AT). These data were then expressed relative to a neutral upright standing spine posture. The Euclidean norm of these angles was taken over time and used during LDS analysis.

Raw EMG data from each trial were linear enveloped by rectifying and low-pass filtering (2nd order Butterworth 2.5 Hz cut-off). All raw EMG data were then normalized to each participant's maximum MVIC voltages which were linear enveloped in the same manner. All normalized EMG and mechanical switch data were then down-sampled to 32 Hz to be inputted into the spine rotational stiffness model.

2.5. Quantifying lumbar rotational stiffness

The muscular contribution to lumbar spine rotational stiffness was obtained using an anatomically detailed EMG-driven biomechanical model representing 58 muscle lines of action crossing the L4/L5 joint (Brown and McGill, 2010). Data inputted into this model included the time-varying linear enveloped EMG signals and lumbar spine angles. Muscle force estimates were made by

$$F_m = NEMG_m PCSA_m \sigma_m l_m \, v_m G \tag{1}$$

where F_m is the force produced by muscle m about its line of action (N), $NEMG_m$ is the normalized EMG signal for muscle m (%MVIC), $PCSA_m$ is the physiological cross-sectional area of muscle m (cm^2), σ_m is the stress generated by muscle m (set at 35 N/cm^2), l_m is the length coefficient of muscle m (unitless), v_m is the velocity coefficient of muscle m (unitless) and G is the participant specific calibration gain (unitless).

Coefficients accounting for the muscle force-length and force-velocity relationships were adapted from McGill and Norman (1986). Muscle lengths and velocities were obtained by rotating vertebral muscle attachments at each spinal level in accordance to kinematic lumbar spine angles. Muscle optimal lengths and passive force-generating characteristics were tuned to participant specific spine flexion ROM, based on previously unpublished work in which we have adjusted these factors to match internally modeled with externally determined low back moments over full lumbar spine ranges of motion.

Each participant-specific gain factor (G) was obtained by finding the best match in the static 45° loaded spine flexion trial between the experimentally determined moment (3DSSPP, Centre for Ergonomics, University of Michigan, Ann Arbor, MI) and the moment estimated by the EMG-driven model. This was done to accommodate for differences in participant muscle size.

To estimate the muscular contribution to lumbar spine rotational stiffness the following formula was employed (Potvin and Brown, 2005):

$$S_Y = \sum_{m=1}^{58} F_m \left[\frac{A_X B_X + A_Y B_Y - {r_Y}^2}{l} + \frac{q {r_Y}^2}{L} \right]_m, \tag{2}$$

where S_Y is the rotational stiffness about the y-axis (flexion/extension) of the L4/L5 joint, F_m is the muscle force (N) obtained from Eq. (1), l is the 3D length of the muscle vector that crosses the L4/L5 joint, L is the full 3D length of the muscle, r is the 3D muscle moment arm across the axis of interest, A_X and A_Y are the origin coordinates with respect to the L4/L5 joint (0,0,0), B_X and B_Y are insertion coordinates with respect to L4/L5, and q is the stiffness gain factor relating muscle force and length to stiffness (Bergmark, 1989) (set at 10 (e.g. Granata and Marras, 2000; Graham and Brown, 2012)).

S_X and S_Z were also computed for the LB and AT axes, respectively. The Euclidean norm of the three stiffness dimensions was also calculated to get an estimate of overall 3D stiffness. Maximum, minimum and mean stiffness values were computed for each