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Trunk active response and spinal forces in sudden forward loading – analysis of the role of perturbation load and pre-perturbation conditions by a kinematics-driven model

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ABSTRACT

Understanding the central nervous system (CNS) response strategy to trunk perturbations could help in prevention of back injuries and development of rehabilitation and treatment programs. This study aimed to investigate biomechanical response of the trunk musculoskeletal system under sudden forward loads, accounting for pre-perturbation conditions (preloading, initial posture and abdominal antagonistic coactivation) and perturbation magnitudes. Using a trunk kinematics-driven iterative finite element (FE) model, temporal profiles of measured kinematics and external load along with subjects' weights were prescribed to predict thoracolumbar muscle forces/latencies and spinal loads for twelve healthy subjects when tested in six conditions during pre- and post-perturbation periods. Results demonstrated that preloading the trunk significantly (i.e., $p < 0.05$) increased pre-perturbation back muscle forces but significantly decreased post-perturbation peak muscle active forces and muscle latencies. Initial trunk flexion significantly increased muscle active and passive forces before the perturbation and their peak values after the perturbation, which in turn caused much larger spinal loads. Abdominal muscles antagonistic pre-activation did not alter the internal variables investigated in this study. Increase in sudden applied load increased muscle reflex activities and spinal forces; a 50 N increase in sudden load (i.e., when comparing 50 N to 100 N) increased the L5-S1 compression force by 1327 N under 5 N preload and by 1374 N under 50 N preload. Overall, forces on the spine and hence risk of failure substantially increased in sudden forward loading when the magnitude of sudden load increased and when the trunk was initially in a flexed posture. In contrast, a higher initial preload diminished reflex latencies and compression forces.

1. Introduction

Segmental compression and shear forces vary along the spinal column and depend on the external loads (gravity, load in hands and inertia), posture, passive ligamentous stiffness and activation level of trunk muscles. Due to large reflex responses and resulting spinal loads, unexpected alterations in loading and/or posture are recognized as risk factors for low back injuries (Lavender et al., 1989). Too low or too high magnitudes of and/or undue delays in reflex (feedback) activation in response to sudden perturbations are expected in patients with disordered CNS, injury or low back pain that likely exacerbate loads on spine and associated risk (Cholewicki et al., 2005; Panjabi, 1992; Reeves et al., 2008). The trunk response to sudden loads depends not only on the external perturbation itself but also on the internal pre-perturbation conditions associated with initial posture and muscle activity. Intrinsic trunk muscle and ligamentous passive stiffness values increase respectively with activation level (Bergmark, 1989; Brown and McGill, 2010; Cholewicki and McGill, 1995) and greater trunk rotations and compression (Andersen et al., 2004; Arjmand and Shirazi-Adl, 2006a; Cholewicki et al., 2000; Lee et al., 2007; Lee et al., 2006; Moorhouse and Granata, 2005; Shirazi-Adl, 2006). In accordance, higher trunk stiffness and muscle agonist/antagonist activities diminish trunk displacements under perturbations (Granata et al., 2004; Krajcarski et al., 1999; Stokes et al., 2000). Moreover, lower EMG reflex response along with smaller displacements were observed at higher pre-perturbation muscle activity (Vera-Garcia et al., 2006). Similarly, lower reflex response to perturbations were recorded at larger flexion angles (Granata and Rogers, 2007) emphasizing the marked role of the

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passive stiffness (Arjmand and Shirazi-Adl, 2006a; Granata and Wilson, 2001; McGill et al., 1994; Zeinali-Davarani et al., 2008).

Experimental set ups in earlier perturbation investigations, however, varied from one study to another, which tends to complicate attempts to compare findings and draw general conclusions. Changes include sudden load duration and temporal variations, perturbation load magnitude/direction/position, loading versus unloading, subject initial posture, muscle preactivation and anticipatory conditions. In our recent in vivo study on the effect of alterations in perturbation load and pre-perturbation trunk conditions (initial preload, posture and coactivation) (Shahvarpour et al., 2014), it was found that unlike peak displacement and reflex muscle responses, peak trunk velocity and acceleration were sensitive to changes in both perturbation load and initial conditions. In corroboration with our earlier findings (Bazrgari et al., 2009), these sensitivities emphasize the potential of kinematics-driven models in decoding the complex and confounding roles of various initial conditions on the transient post-perturbation response of the human trunk. This also concurs with the fact that kinematics (velocity/acceleration profiles) and kinetics (external loads, body weight) that are used as input data in such models are recorded at greater accuracy as compared to EMG data. Effectively under sudden forward loading perturbations, back muscle reflex responses as recorded with surface EMG have shown poor to moderate reliability (Santos et al., 2011).

Following an unexpected perturbation, trunk muscles are reflexively activated to prevent lumbar instability but only after a delay period called reflex latency (Hodges and Bui, 1996; Santos et al., 2011; Staude, 2001; Vera-Garcia et al., 2006). An increase in reflex latency (in presence of prior back injuries for example) impairs adequate control and stability of the trunk (Cholewicki et al., 2005; Hodges and Richardson, 1996; Radebold et al., 2000). Any muscle activation on the other hand translates to a mechanical force only after an additional delay period called electromechanical delay (EMD). The rate of muscle force development during voluntary contractions was found to have inverse relation with EMD (Thelen et al., 1994; van Dieen et al., 1991). In addition, larger trunk flexion angles were reported to prolong EMD (Marras, 1987). In an earlier study (Bazrgari et al., 2009), the time of muscle force onset (i.e., latency including EMD) was predicted using feed-forward simulations in a kinematics-driven FE model (Bazrgari et al., 2009).

In the current study, the recently recorded trunk kinematics and applied sudden external force profiles along with body weight of 12 asymptomatic subjects under six different conditions (Shahvarpour et al., 2014) were used to drive a validated musculoskeletal nonlinear FE model of the trunk (Arjmand and Shirazi-Adl, 2005, 2006a; Bazrgari et al., 2009; Bazrgari et al., 2008). All 12 subjects at six different experimental conditions were simulated and statistical analyses performed to identify the effects of perturbation (sudden load magnitude) and initial conditions (preload magnitude, initial posture and antagonistic preactivity) on muscle reflex responses (magnitude and delay) and spinal loads (compression and shear forces at the L5-S1). It was hypothesized that the kinematics-driven FE model would (1) be sensitive to the effect of various pre-perturbation and sudden loading conditions and (2) demonstrate that the trunk muscle reflex activity and spinal loads drop in conditions associated with higher pre-perturbation intrinsic stiffness.

2. Methods

Our recent in vivo study dataset used in the current simulations are briefly described here (Shahvarpour et al., 2014). Twelve young male subjects (weight 73.0 ± 3.9 Kg and height 177.7 ± 3.0 cm) participated. Isometric maximum voluntary contraction (MVC) trials were carried out for normalization of EMG data. Superficial EMG signals of 12 muscles were recorded bilaterally at longissimus (LG, at the L1 level), iliocostalis (IC, at the L3), multifidus (MF, at the L5), rectus abdominus (RA), external oblique (EO) and internal oblique (IO). Subjects were semi-seated in a perturbation apparatus with the pelvis fixed while pre-perturbation and sudden loads were applied through a cable connected at the T8 level to a harness. A load cell along the cable measured the load applied whereas a potentiometer connected to the harness on the back measured the translation at the T8 level. Muscles EMG and trunk displacement were recorded with the sampling frequency of 1024 Hz.

Six initial conditions were considered (Table 1). In conditions 1 to 4 (C1-C4) the effects of changes in preload (5 and 50 N) and sudden load (50 and 100 N) were investigated. In C5, subjects flexed forward (10 cm anterior translation at the T8 level causing $\sim 20^\circ$ of trunk flexion) before perturbation. Finally in C6, subjects preactivated abdominal muscles, attempting to maintain the activity level of EO at 10% of MVC using a visual biofeedback while in the upright posture similar to C1-C4. Recorded EMG showed significantly greater pre-perturbation activity in abdominals in C6 when compared to C2 and C4 (Shahvarpour et al., 2014). Five trials were performed for each condition. The perturbation force was applied suddenly and randomly during 5 s.

2.1. FE model studies

For the sake of simulations, one trial was chosen randomly for each subject and condition as statistical analysis rejected the effect of learning between trials. Simulation durations covered periods starting 256 ms pre-perturbations and 1 s after. Recorded trunk displacement in this period was resampled to 50 Hz. With the pelvis fixed, sagittal rotations at the T12 and lumbar levels for each subject at 6 conditions were estimated based on the measured T8 translations (Shahvarpour et al., 2014) and intersegmental rotation ratios (Bazrgari et al., 2009). Velocity and acceleration profiles were calculated from displacements. Due partly to the stiffening effect of the ribcage and in accordance with earlier studies (Belytschko et al., 1973; Nussbaum and Chaffin, 1996; Schultz et al., 1973), the T1-S1 motions are limited in the model to those at the T12-S1 levels thereby neglecting relative rotations at the T1-T12 levels. Some thoracic rotations at T1-T12, albeit much smaller than lumbar rotations, have nevertheless been reported (Gercek et al., 2008; Morita et al., 2014).

Iteratively and driven by angular velocity profiles at different levels as well as external load and subject-specific gravity forces distributed along the spinal height, the trunk FE model was iteratively analyzed to estimate muscle recruitment patterns and spinal loads during pre- and post-perturbation periods at all 6 conditions for all 12 subjects. The FE model (Bazrgari et al., 2008, 2009) consisted of 7 rigid bodies representing sacrum, L5 to L1 vertebrae and thorax-head-hands segments (Fig. 1). Six nonlinear shear-deformable beam elements, with mechanical properties based on previous studies (Oxland et al., 1992; Shirazi-Adl, 2006; Yamamoto et al., 1989), accounted for passive stiffness of motion segments. Seven connector elements parallel to beams accounted for intersegmental damping (Kasra et al., 1992; Markolf, 1970). Distributed mass and inertial properties at each vertebral level were based on the literature (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983).

Trunk musculature was represented by 46 local lumbar and 10 global thoracic muscles (Fig. 1). For global extensor muscles, nonlinear trajectories (wrapping of muscles plus contact forces) were taken into account (Arjmand et al., 2006); global muscles were constrained not to approach the T12 to L5 vertebral centers more than 90% of their respective initial distances at the undeformed configuration. In case of wrappings, muscle forces remained identical in various segments assuming frictionless contact at wrapping points (Shirazi-Adl, 1989, 2006; Shirazi-Adl and Parnianpour, 2000).Wrapping contact forces were applied as external forces in subsequent iteration. The trunk FE model of each of 12 subjects was driven by its angular velocity profiles at different levels (based on the measured translation profile at the T8) as well as external load profile and gravity forces distributed along the spinal height. It was iteratively analyzed to estimate muscle recruitment patterns and spinal loads at all times during pre- and post-perturbation periods under 6 conditions. The objective function of the minimum sum of the cubed muscle stresses at each vertebral level was considered (Arjmand and Shirazi-Adl, 2006b). All unknown muscle forces were bound to be greater than the muscle

Table 1
Independent parameters in the six experimental conditions considered in this work.

Condition	Pre Load (N)	Sudden Load (N)	Initial posture	EO preactivation
C1	5	50	Upright	–
C2	5	100	Upright	–
C3	50	50	Upright	–
C4	50	100	Upright	–
C5	5	50	10 cm Anterior Translation	–
C6	5	100	Upright	10%