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Effects of variation in external pulling force magnitude, elevation, and orientation on trunk muscle forces, spinal loads and stability ☆

Z. El Ouaaid [a,b], A. Shirazi-Adl [a,*], A. Plamondon [b]

[a] Division of Applied Mechanics, Department of Mechanical Engineering, École Polytechnique, Montréal, Québec, Canada
[b] Institut de recherche Robert Sauvé en santé et en sécurité du travail, Montréal, Québec, Canada

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A B S T R A C T

Nowadays in various daily, occupational and training activities, there are many occasions with forces supported in hands acting at various magnitudes, elevations, and orientations with substantial horizontal components. In this work, we aim to compute trunk muscle forces, stability, and spinal loads under pulling external forces applied at 3 elevations and 13 orientations. Under an identical upright standing posture and upper body weight, the trunk active–passive response is computed using a validated iterative finite element kinematics-driven model. Pulling forces of 80, 120, and 160 N are resisted symmetrically in both hands held at 20, 40, and 60 cm elevations above the L5–S1 and oriented each in upward (−90°), inclined upward (−75°, −60°, −45°, −30°, and −15°), horizontal (0°), inclined downward (15°, 30°, 45°, 60°, and 75°) and finally downward in gravity direction (90°). In addition, in all analyses, an antagonist moment of 10 N m is applied in order to generate rather small antagonist coactivity and intra-abdominal pressures of 8–12 kPa are considered when abdominal muscles are active under upward pulling forces. Results demonstrated substantial differences in muscular response, spinal loads, and stability margin as the pulling force elevation, orientation, and magnitude altered. Compression and shear forces at lower lumbar levels peaked under forces at higher elevations acting with downward inclinations. Minimum spinal forces were computed at all elevations under pulling forces in the upward direction. Trunk stability was also maximum under these latter forces pulling upward. These findings have important consequences in rehabilitation, training, and design of safer occupational activities.

1. Introduction

Lifting and lowering of objects from a position to another have traditionally been in focus as one of the most common tasks in manual material handling environments. In these cases, the external loads are carried often in hands and are oriented primarily in the gravity direction with little or no horizontal components for example due to inertia and friction. Nowadays during regular daily, occupational, and training activities, however, there are many physical activities that involve push and pull for example of heavy objects via cable bars and handles that generate forces on spine at different elevations with substantial non-gravity horizontal components (De Looze et al., 1995; Hoozemans et al., 2007; Jager et al., 2007; Schibye et al., 2001; Theado et al., 2007). As a consequence, spinal loads, muscle activations, and trunk stability could alter potentially increasing the risk of injuries and pain (Hoozemans et al., 2004; Knapik and Marras, 2009).

The trunk response under horizontal forces in sudden loading/unloading conditions have been the subject of some earlier in vivo (Andersen et al., 2004; Cholewicki et al., 2000; Brown and McGill, 2008; Shahvarpour et al., 2014, 2015a) and computational model (Bazrgari et al., 2009a, 2009b, 2011; Shahvarpour et al., 2015a, 2015b) studies. Large compression and shear forces at the L5–S1 level accompanied with a more stable trunk were computed post-perturbation especially under higher sudden forward loads and in initially flexed postures (Shahvarpour et al., 2015a, 2015b). A deformed system or structure is stable if it does not exhibit hypermobility (large displacements) when exposed to small perturbations. Under static forward horizontal forces applied directly on spine in upright standing posture at 3 different elevations with identical moments (3 magnitudes) at the L3–L4 levels, Kingma et al. (2007) reported greater muscle EMG activity/coactivity and

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* Correspondence to: Division of Applied Mechanics, Dept. of Mechanical Engineering, École Polytechnique, P.O. Box 6079, Station 'centre-ville', Montréal, Québec, Canada H3C 3A7. Tel.: +514 340 4711x4129; fax: +514 340 4176.
E-mail address: aboulfazl.shirazi@polymtl.ca (A. Shirazi-Adl).

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spinal loads under higher external forces at lower elevations. Similar results were reported by El Ouaaid et al. (2014a, 2014b) under forward (pulling) horizontal forces resisted in hands in front at two elevations yielding identical moments (3 magnitudes) at the L5–S1. These latter works, however, also investigated the effect of changes in force orientation (5 levels) at a fixed elevation and identical moments (3 magnitudes) at the L5–S1 on muscle EMG activity and spinal loads. Despite identical external moments at the L5–S1 and similar trunk postures, substantially different trunk muscular responses with moderate alterations (up to 24%) in spinal loads were predicted as the pulling force orientation varied. Compression and shear forces at the L5–S1 as well as forces in extensor thoracic muscles progressively decreased as the orientation of external forces varied from downward gravity (90°) to inclined upward direction (−25°). In contrast, forces in local lumbar muscles followed reverse trends.

Using EMG-driven models, Hoozemans et al. (2004) and Knapik and Marras (2007) estimated loads on spine in different push–pull conditions. In the former study, one- and two-handed push and pull of carts at 3 weights and 2 handle elevations (shoulder and hip) were considered. Low cart weights and push–pull at shoulder level were recommended to reduce net moment and spinal compression at the L5–S1. Larger compression forces under pulling whereas larger shear forces under pushing were reported in the latter study under forces oriented horizontally at 3 magnitudes (resistance levels) and 3 elevations (handle heights). The effects of changes in the external force orientation and elevation on estimation of spinal loads are generally overlooked in many existing lifting tools and regression equations (Arjmand et al., 2015).

In continuation of our earlier studies where forces at 5 orientations at a fixed height and 2 elevations under a horizontal force (for a total of 18 loading cases) were varied to maintain identical flexion moments at the L5–S1(El Ouaaid et al., 2014a, 2014b), in this study and in an upright standing posture, pulling forces are applied through both hands at 3 fixed magnitudes, 3 elevations, and 13 orientations (from down ward gravity to upward for a total of 117 loading cases). We aim here to compute trunk muscular coordination, muscle forces, spinal compression/shear forces, and stability using the iterative finite element kinematics-driven model of the trunk. It is hypothesized that, in standing posture and under identical applied pulling forces and posture, muscle forces, spinal loads and trunk stability substantially alter as orientation of pull changes.

2. Methods

2.1. Kinematics-driven model

One of 12 subjects in our earlier in vivo studies with the body weight 68.3 kg and height 181.5 cm is modeled here (El Ouaaid et al., 2013b, 2014a). Three pulling forces of 80, 120, and 160 N are considered symmetrically at both hands. Each force is carried at 20, 40, and 60 cm heights above the L5–S1 in 13 different orientations; upward (−90°), inclined upward (−75°, −60°, −45°, −30°, and −15°), horizontal (0°), inclined downward (15°, 30°, 45°, 60°, and 75°) and finally downward in gravity direction (90°) (Fig. 1). In all analyses and in accordance with our earlier EMG measurements, a constant antagonist moment of 10 N m is considered (see Eq. (3) below, El Ouaaid et al., 2013a, 2014b) in order to generate some (<4% activity in abdominal muscles and <9% activity in global extensor muscles) antagonistic coactivity; in abdominal muscles under downward forces with net flexion moment at the T12 or in global extensor muscles under upward forces with net extension moment at the T12. Abdominal, local lumbar, and global extensor muscle forces as well as spinal compression/shear forces at the L5–S1 and L4–L5 levels are estimated. Finally, the trunk stability margin (i.e., additional pull on top of the existing loads that the system can support in the deformed configuration without exhibiting hypermobility) is evaluated.

A thoracolumbar T1–S1 nonlinear kinematics-driven finite element model (Arjmand and Shirazi-Adl, 2006a, 2006b, 2006c; Bazrgari et al., 2009a, 2009b; El Ouaaid et al., 2014b) (Fig. 1) along with a coupled objective function, Eq. (1) below, (El Ouaaid et al., 2013a) is employed to estimate trunk muscle forces, compression/shear forces at all T12–S1 levels, and trunk stability. The multi-segment T1–S1 model consists of six shear deformable beams with nonlinear proprieties (see Figs. 1 and 2) to represent the overall stiffness of T12–S1 motion segments (i.e. disc, facets and ligaments) at different directions and rigid elements to represent thoracic spine T1–T12 (as a single body) and lumbosacral vertebrae (L1–S1). A sagittaly symmetric muscle architecture consisting of 46 local lumbar (inserted into lumbar vertebrae) and 10 global (inserted into the thoracic spine) muscle fascicles (Fig. 1) are considered. The subject trunk weight is distributed eccentrically at different spinal levels (Pearsall, 1994) while the weight of the upper arms, forearms, hands, and head/neck, estimated based on anthropometric data (de Leva, 1996) and subject body weight, are applied at their mass centers measured in the upright posture while resisting loads in hands.

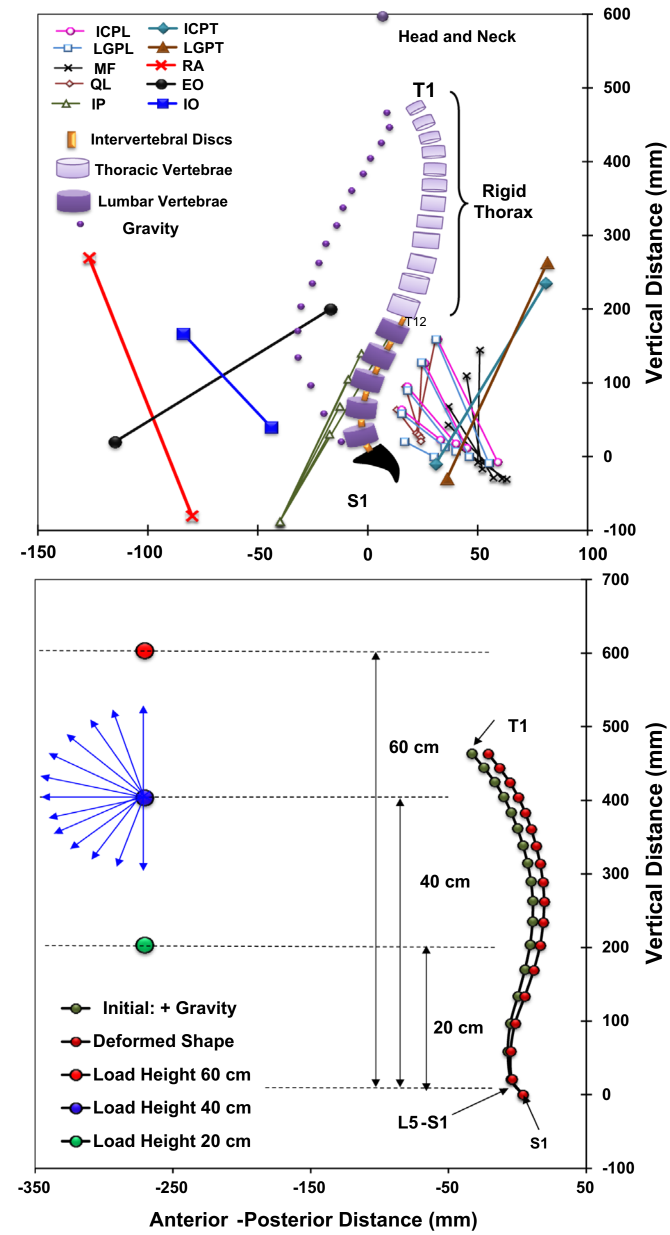

Fig. 1. The sagittaly-symmetric FE model with musculature in the upright standing posture at initial configuration before application of gravity loads (axes are not to the same scale) (top). Six inter-vertebral beam elements (shown as discs) represent the stiffness of various motion segments. Global muscles: ICPT, iliocostalis lumborum pars thoracic; LGPT, longissimus thoracis pars thoracic; IO, internal oblique; EO, external oblique and RA, rectus abdominus. Local lumbar muscles: ICPL, iliocostalis lumborum pars lumborum; LGPL, longissimus thoracis pars lumborum; MF, multifidus; QL, quadratus lumborum and IP, iliopsoas (Bogduk et al., 1992; Stokes and Gardner-Morse, 1999). Mass centers of the head and neck as well as trunk itself are also depicted. Bottom: upright standing postures under gravity load without (El-Rich et al., 2004) and with external force (El Ouaaid et al., 2014a) supported in hands at 3 magnitudes, 3 elevations and 13 orientations.