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Seated whole body vibrations with high-magnitude accelerations—relative roles of inertia and muscle forces

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ABSTRACT

Reliable computation of spinal loads and trunk stability under whole body vibrations with high acceleration contents requires accurate estimation of trunk muscle activities that are often overlooked in existing biodynamic models. A finite element model of the spine that accounts for nonlinear load- and direction-dependent properties of lumbar segments, complex geometry and musculature of the spine, and dynamic characteristics of the trunk was used in our iterative kinematics-driven approach to predict trunk biodynamics in measured vehicle's seat vibrations with shock contents of about 4 g (g: gravity acceleration of 9.8 m/s^2) at frequencies of about 4 and 20 Hz. Muscle forces, spinal loads and trunk stability were evaluated for two lumbar postures (erect and flexed) with and without coactivity in abdominal muscles. Estimated peak spinal loads were substantially larger under 4 Hz excitation frequency as compared to 20 Hz with the contribution of muscle forces exceeding that of inertial forces. Flattening of the lumbar lordosis from an erect to a flexed posture and antagonistic coactivity in abdominal muscles, both noticeably increased forces on the spine while substantially improving trunk stability. Our predictions clearly demonstrated the significant role of muscles in trunk biodynamics and associated risk of back injuries. High-magnitude accelerations in seat vibration, especially at near-resonant frequency, expose the vertebral column to large forces and high risk of injury by significantly increasing muscle activities in response to equilibrium and stability demands.

1. Introduction

The cost associated with the low back pain (LBP) has steadily risen over the years exceeding $100 billion in US alone during 1998 (Luo et al., 2004). Extensive literature reviews have demonstrated a strong association between the whole body vibration (WBV) and LBP (Hulshof and van Zanten, 1987; Bovenzi and Hulshof, 1999), thus encouraging efforts to reduce risks involved (Lings and Leboeuf-Yde, 2000). Those exposed to WBV have been reported to be 1.4 (Boshuizen et al., 1992) to 9.5 (Bongers et al., 1990) times more prone to back pain. While some investigators have indicated a stronger association with LBP for the duration of exposure as compared to the vibration magnitude (Lis et al., 2007), others have suggested that vehicle seat vibrations with high acceleration content (Stayner, 2001) cause more back injury than those with low vibration levels that contribute more to the time averaged measures of exposure defined in ISO 2631-1 (1997).

Trunk response to WBV can be affected by many factors such as trunk posture, muscle activation, backrest, footrest, and magnitude/frequency of excitation inputs (Pope et al., 1987; Paddan and Griffin, 1988; Fairley and Griffin, 1989; Broman et al., 1991; El-Khatib et al., 1998; Hinz et al., 2002). Erect posture, in general as compared to a relaxed (flexed or slouched) posture, increases trunk resonant frequency and reduces the corresponding transmissibility gain (or modulus of apparent mass) (Fairley and Griffin, 1989; Broman et al., 1991). In lower excitation frequencies (i.e. 2–4 Hz), gain of transmissibility has conversely been reported to be larger in a relaxed posture (Pope et al., 1987). Muscle activity tends to increase both trunk resonant frequency and transmissibility gain (Fairley and Griffin, 1989; Broman et al., 1991), while a shift toward lower frequencies may occur due to muscle fatigue (Goel et al., 2001). Driving on a seat without backrest increases the peak of transmissibility with the maximum occurring at a bent forward posture (Hinz et al., 2002).

Since pathological studies relate a diverse spectrum of lumbar lesions to mechanical loads (Bogduk and Twomey, 1991), accurate determination of load distribution among passive and active components of the human trunk is essential in effective prevention, evaluation, and treatment of spinal disorders. Infeasibility of direct measurements along with

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practical difficulties and validity concerns associated with indirect measurements of muscle forces and spinal loads have presented biomechanical models as indispensable tools for estimation of spinal loads during various occupational and recreational activities. Existing WBV biodynamic models do not, however, properly simulate the redundancy in the trunk musculoskeletal system in which the spinal loads and stability depend directly on unknown muscle forces that alter during the WBV period (Bazrgari, 2008; Bazrgari et al., 2008a). Muscle forces not only balance the moment of external, gravity and inertial loads, but also maintain the spinal stability. Due to much smaller lever arms, trunk muscles are at a mechanical disadvantage compared to external, gravity, and inertial loads; an effect that increases spinal loads substantially beyond applied forces (Arjmand, 2007; Bazrgari, 2008). The importance of the proper estimation of muscle forces in evaluation of spinal loads in WBV conditions has been recognized (Bluthner et al., 2002; Seidel, 2005).

We have employed the iterative kinematics-driven approach to study the trunk biomechanics in static (Kiefer, 1999; El-Rich, 2005; Arjmand, 2007) and dynamic (Bazrgari, 2008) conditions. The method accounts for nonlinear load- and direction-dependent properties of lumbar motion segments, complex geometry and musculature of the spine, dynamic characteristics of the trunk, and wrapping of global extensor muscles in forward flexion. Forces in muscles are computed under prescribed spinal kinematics separately at different levels yielding results in satisfaction of stability condition as well as equilibrium equations at all directions/levels along the spine. It has been reported (Robinson, 1999) that occupants of off-road and industrial vehicles are exposed to whole body vibration and repeated shocks with peak acceleration contents in the range of 2–6 g. In order to delineate the effects of high-magnitude shock vibration content (likely lost while averaging in ISO 2631-1 (ISO 2631-1)) on trunk biodynamics, the objective of the present novel study is set to calculate trunk muscle forces, spinal loads and stability under WBV with high ~4 g acceleration contents at two different frequencies of ~4 and 20 Hz. The effects of changes in the spinal posture and of the presence of antagonistic abdominal coactivity on the response are also determined. It is hypothesized that muscle forces generated in response to trunk equilibrium and stability demands in such WBV environments would substantially increase spinal loads and associated risk of back injuries.

2. Method

Resolving the redundancy to estimate unknown muscle forces is a primary challenge facing biomechanical models of the spine. In the kinematics-driven approach, measured kinematics data are exploited in a nonlinear finite element model of the spine to generate additional equations at each spinal level in order to alleviate the kinetics redundancy in the system at that particular level. Additional description of the method can be found elsewhere (Arjmand, 2007; Bazrgari, 2008). The method uses a sagittally symmetric head–pelvis model made of six nonlinear deformable beams representing T12-S1 spinal discs, seven rigid elements for lumbar vertebrae (L1–L5) and head-T12 (as a single body), and a connector element to simulate the buttocks (Fig. 1). The beams represent the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments) at different levels with nonlinear axial compression-strain and sagittal/lateral/axial moment-rotation relations defined based on earlier studies (Yamamoto et al., 1989; Oxland et al., 1992; Pop, 2001; Shirazi-Adl et al., 2002). Trunk/head/arms/pelvis mass and mass moments of inertia are assigned at different levels along the spine at their respective gravity centres based on published data (Zatsiorsky and Seluyanov, 1983; de Leva, 1996; Pearsall et al., 1996). Inter-segmental damping is modelled by connector elements parallel to deformable beams (Fig. 1) using measured values (Markolf, 1970; Kasra et al., 1992) with translational damping = 1200 Ns/m and angular damping = 1.2 Nms/rad. A nonlinear force-displacement (compression only) property is used to represent buttocks mechanical response (Bazrgari et al., 2008a). For muscles, a sagittally symmetric architecture consisting of 46 local (attached to lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles is used (Fig. 1).

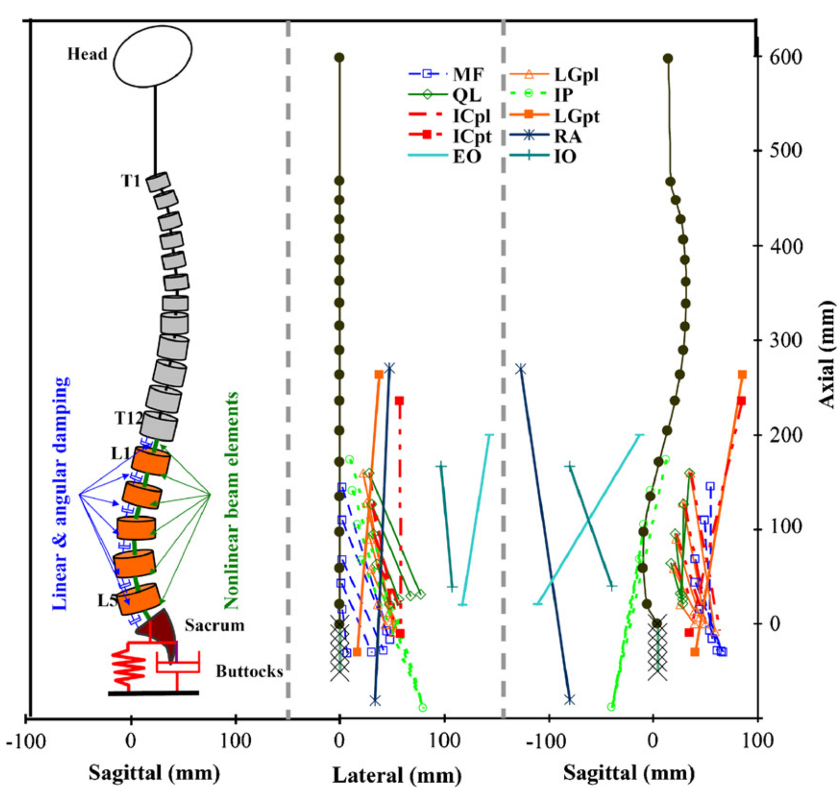

Fig. 1. Trunk model used in the study including global and local musculatures in the sagittal (on the right), frontal (in the middle, fascicles on one side are shown) planes, and vertebral column (on the left). ICpl: iliocostalis lumborum pars lumborum, ICpt: iliocostalis lumborum pars thoracic, IP: iliopsoas, LGpl: longissimus thoracis pars lumborum, LGpt: longissimus thoracis pars thoracic, MF: multifidus, QL: quadratus lumborum, IO: internal oblique, EO: external oblique, RA: rectus abdominus.