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Journal of Sound and Vibration 318 (2008) 1334–1347
JOURNAL OF SOUND AND VIBRATION
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Computation of trunk muscle forces, spinal loads and stability in whole-body vibration

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Received 7 August 2007; received in revised form 16 April 2008; accepted 17 April 2008
Handling Editor: S. Bolton
Available online 9 June 2008

Abstract

Whole-body vibration has been indicated as a risk factor in back disorders. Proper prevention and treatment management, however, requires a sound knowledge of associated muscle forces and loads on the spine. Previous trunk model studies have either neglected or over-simplified the trunk redundancy with time-varying unknown muscle forces. Trunk stability has neither been addressed. A novel iterative dynamic kinematics-driven approach was employed to evaluate muscle forces, spinal loads and system stability in a seated subject under a random vertical base excitation with $\sim\pm 1\,g$ peak acceleration contents. This iterative approach satisfied equations of motion in all directions/levels while accounting for the nonlinear passive resistance of the ligamentous spine. The effect of posture, co-activity in abdominal muscles and changes in buttocks stiffness were also investigated. The computed vertical accelerations were in good agreement with measurements. The input base excitation, via inertial and muscle forces, substantially influenced spinal loads and system stability. The flexed posture in sitting increased the net moment, muscle forces and passive spinal loads while improving the trunk stability. Similarly, the introduction of low to moderate antagonistic coactivity in abdominal muscles increased the passive spinal loads and improved the spinal stability. A trade-off, hence, exists between lower muscle forces and spinal loads on one hand and more stable spine on the other. Base excitations with larger peak acceleration contents substantially increase muscle forces/spinal loads and, hence, the risk of injury.

1. Introduction

Low back pain (LBP) is the leading musculoskeletal disorder in terms of cost and work-absenteeism [1]. Long-term occupational exposure to whole-body vibration (WBV) is reported to increase risk of lumbar spine disorders [2–4]. Sedentary working environment with WBV exposure and/or awkward postures increase the risk of LBP up to four-fold as compared with sitting alone [5]. Vehicle seat vibrations with high acceleration content likely cause more back injury [6] than those with low vibration levels that contribute more to the time averaged measures of exposure defined in ISO 2631-1 [7]. The direct casual association between WBV and LBP has, however, been questioned by some investigators [6,8,9] suggesting that, on the basis of existing

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0022-460X/$ - see front matter
doi:10.1016/j.jsv.2008.04.047

literature, it is not possible to confirm whether WBV exposure alone or in combination with other factors should be considered as a risk factor. Based on the premise that excessive spinal loads increase risk of back injuries, a sound risk assessment along with effective prevention, treatment, and rehabilitation programs of spinal disorders depend directly on accurate evaluation of muscle forces and spinal loads. A clearer picture of the causal role of WBV environments in back disorders can thus emerge following improved understanding on associated trunk biodynamics. Since spinal loads cannot be measured directly in vivo, biomechanical models are recognized to play indispensable role in spinal pathomechanics.

Measured driving point frequency response (impedance, apparent mass) and transmissibility functions between seat input and response at different spinal levels have led to the development of rather simple biomechanical WBV models [3,10–12]. Kitazaki and Griffin [13] developed a passive sagittaly symmetric finite element model of the upper body simulating the spine, viscera, head, pelvis and buttocks using beam, spring and mass elements. Such models can predict the overall passive response under various vibration and postural conditions and may be used to improve vehicle suspension design. The muscle forces and spinal loads cannot, however, be estimated using such deterministic models. A two-dimensional (2D) dynamic finite element model of the lower lumbar vertebrae was developed by Pankoke et al. [14]. Under applied static (including a constant extensor muscle force that depended on the posture) and dynamic loads, spinal loads at lower lumbar levels were computed. A head to sacrum finite element model including the entire spinal column and the rib cage was used by Kong and Goel [15] to study the trunk resonant frequency and transmissibility under base vertical vibration. The muscles were modeled deterministically as tension-only truss elements with a constant elastic modulus of 1.0 MPa. They reported the first vertical natural frequency in the range of 6.8–8.9 Hz depending on the muscle tension and gravity preload. Other three-dimensional (3D) dynamic models have evaluated the trunk WBV response under constant muscle forces that were estimated a priori using a static analysis with an optimization approach [16,17].

Foregoing WBV studies of the trunk have either neglected or over-simplified the redundancy in the trunk musculoskeletal system in which the spinal loads depend directly on unknown muscle forces that alter during WBV. The importance of a proper estimation of muscle forces in quantification of spinal loads in WBV has been emphasized [18,19]. The dynamic stability of the spine in WBV studies has also been overlooked. The trunk stability is maintained by activation in muscles as well as passive muscle/spinal stiffness properties and is influenced by changes in the posture, passive properties and load magnitude/height [20–24].

In continuation of our earlier isometric [20,21,25] and transient [22,26] investigations of the trunk biomechanics using the iterative kinematics-driven finite element approach, the spinal loads, trunk muscle forces and trunk stability under a random vertical base excitation is studied. The model accounts for nonlinear load- and direction-dependent properties of lumbar motion segments, complex geometry of spine, detailed muscle architecture, dynamic characteristics of the trunk, and wrapping of global extensor muscles. The input base excitation is taken from the literature [27]. It is hypothesized that high magnitude acceleration contents in vehicular vibrations increase the risk of tissue injury by generating loads in the neighborhood of safe threshold values.

2. Methods

2.1. Kinematics-driven approach

This approach exploits kinematics data to generate additional equations at each spinal level in order to alleviate the kinetics redundancy in the system. Initially, measured trunk kinematics (sagittal plane rotations at different vertebral levels and base vertical acceleration in this study) along with external/gravity loads are prescribed into a nonlinear finite element model of the thorocolumbar spine (Fig. 1). Implicit algorithm with unconditionally stable Hilber–Hughes–Taylor integration operator [28] is used to solve the nonlinear transient problem, resulting in the time variation of reaction moments at each vertebral level to be balanced by muscles attached to that level only. To resolve the remaining redundancy at each level, an optimization approach with the cost function of minimum sum of cubed muscle stresses is used considering only those muscles attached to the level under consideration. The inequality equations relate to unknown muscle forces remaining positive and greater than their passive force components (calculated based on instantaneous muscle length and a

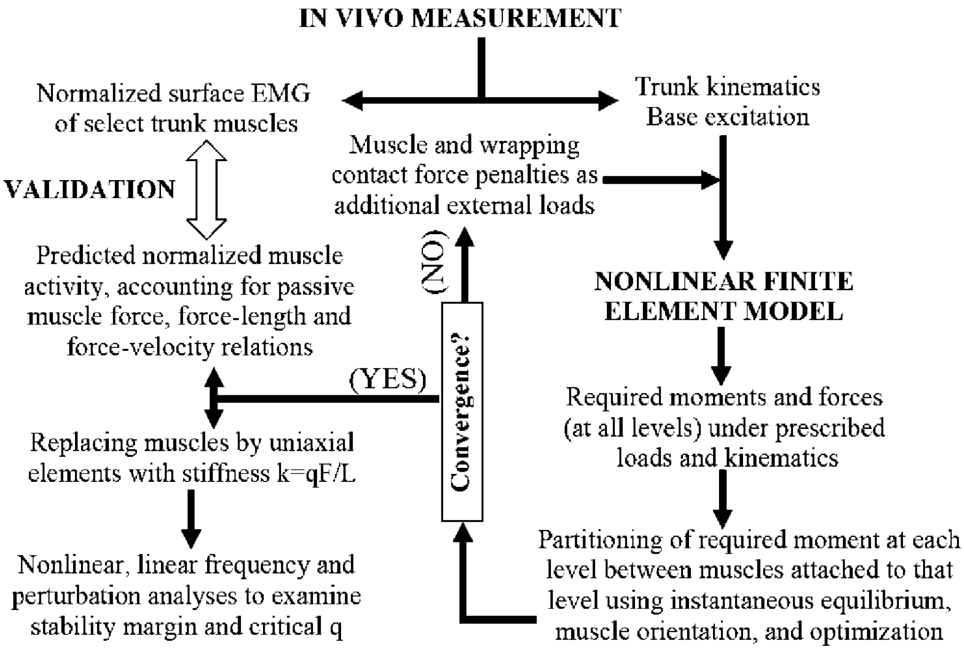

Fig. 1. Flowchart for the application of the kinematics-based approach used to determine trunk muscle forces, internal loads and spine stability under whole-body vibration base excitation. Convergence is attained if calculated muscle forces in two successive iterations remain the same. Measured muscle EMG activity, if available, is used at the post processing step in the model study for qualitative validation of the computed active forces in muscles.

tension–length relationship [29]) but smaller than the sum of their respective maximum active forces (i.e. 0.6 MPa times muscle's physiological cross-sectional area, PCSA [30]) and the passive force components is also considered. The cost function of sum of cubed muscle stresses has been demonstrated to yield plausible results comparable with reported measured EMG activities and disk pressure values [31]. For this purpose, Lagrange multiplier method is employed to analytically solve the problem which guarantees the convergence of results to a global minimum. At the end of each iteration, the penalty of muscle forces in shear and axial directions is applied along with the external loads to the spine and the procedure is repeated until the convergence is achieved (i.e. calculated muscle forces in two successive iterations remain almost the same).

Once the muscle forces are calculated throughout the vibration duration, the system stability is investigated by replacing muscles with uniaxial elements. The stiffness of each uniaxial element, k, is assigned using linear stiffness–force relation (i.e. $k = qF/L$) in which the muscle stiffness is proportional to the instantaneous muscle force, F, and inversely proportional to its current length, L, with q as a dimensionless muscle stiffness coefficient that is taken to be the same for all muscles [32]. At each instance of time, the stability margin under different q values is investigated at the loaded deformed configurations by natural frequency and linear perturbation analyses. In general, to assess the stability of a nonlinear system using the static stability criterion (i.e. divergence type), one can use linear buckling, perturbation or free vibration analyses at a deformed stressed configuration evaluated based on a prior nonlinear analysis [33]. In perturbation analysis, the translation of $T1$ vertebra under application of a unit load is used for different q values to identify the minimum (critical) q. On the other hand, smallest natural frequency of structure, determined using free vibration analysis, is also used as an indication of structural stability (i.e. system becomes unstable as the smallest natural frequency approaches zero). Finite element program ABAQUS (ABAQUS Inc., Version 6.5) is used to carryout nonlinear and linear stability analyses while the optimization procedure is analytically solved using an in-house program based on Lagrange Multiplier Method.

2.2. Finite element model

A sagittaly 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 is used in the model (Fig. 2) [19–21,25]. 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. The nonlinear load–displacement response in different directions along with flexion versus extension differences are represented based on numerical and measured results of previous single- and multimotion segment studies [34–37]. The flexural rigidity of the model depends also on the axial compression as reported recently [38]. The nonlinear response in flexion moment and axial compression used for different beam elements at various segmental levels is given in Fig. 3, while the shear rigidity (kGA) is assumed constant varying from 12378 N at the T12-L1 to 15157 N at the L5-S1 level. Trunk/head/arms/pelvis mass and mass moments of inertia are assigned at different levels along the spine at their respective gravity centers based on published data [39–41] (Table 1). Connector elements parallel to deformable beams are added to account for the inter-segmental damping (Fig. 2) using measured values [42,43] where translational damping = 1200 N s/m and angular damping = 1.2 N m s/rad. Due to the lack of data, damping values taken for discs are assumed to remain constant throughout vibration in the current study.

Buttocks at the base are modeled by a connector element (compression only) with nonlinear stiffness defined based on reported data in the literature (Table 2) [13,44] and damping similar to that of lumbar segments.

2.3. Muscle architecture

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles is used (Fig. 2). To simulate wrapping of global muscles (i.e. longissimus thoracis pars thoracic and iliocostalis lumborum pars thoracic), they are constrained to

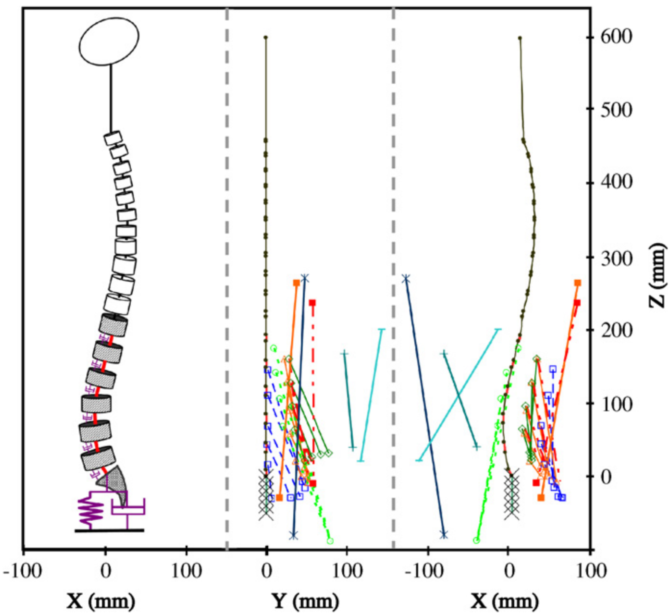

Fig. 2. Representation of the trunk including global and local musculatures in the sagittal (on the right), frontal (in the middle, fascicles on one side are shown) planes, and vertebral column (on the left) with nonlinear beam elements (representing T12-S1 disks), rigid elements (representing vertebrae, sacrum, pelvis and head), connector elements (representing the buttocks and disc dampings). 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, and RA: rectus abdominus.

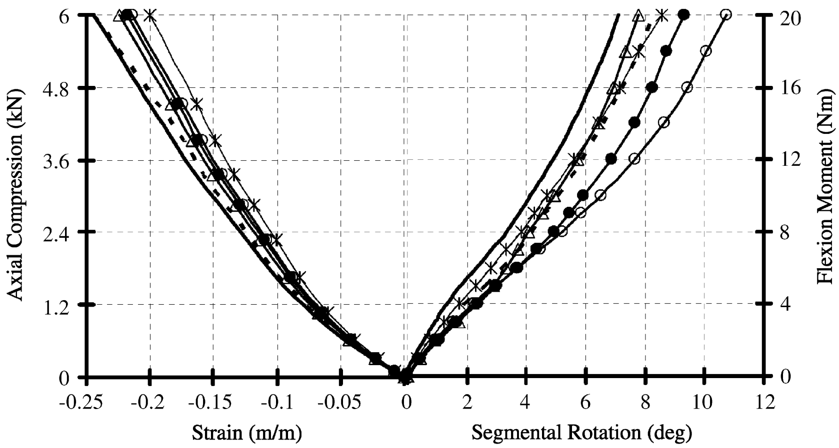

Fig. 3. Nonlinear stiffness properties considered for beam elements representing various motion segments at different levels; axial compression force-axial strain on the left and flexion moment-sagittal rotation (given for large compression preloads of ~1800 N) on the right. (—— T12-L1, ▪▪▪▪▪▪ L1-L2, —△— L2-L3, —●— L3-L4, —○— L4-L5, —✳— L5-S1).

Table 1
Trunk segments mass, mass moment of inertia and mass center locations

Level	% TM[a]	% BM	*Ixx*	*Iyy*	*Izz*	CG-*z*	CG-*x*
Head–neck	–	6.94	27.18	29.34	20.13	597.60	−10.00
Upper arms	–	2*2.8	12.63	11.30	3.80	447.38	30.00
Lower arms	–	2*1.6	6.45	5.99	1.20	426.85	30.00
Hands	–	2*0.6	1.31	0.88	0.50	405.81	30.00
T1	3.59	1.28	6.70	2.00	8.70	467.60	−8.00
T2	3.88	1.38	3.40	2.40	9.10	447.38	−12.00
T3	4.15	1.47	8.40	3.20	11.50	426.85	−20.00
T4	4.46	1.58	8.30	3.40	11.70	405.81	−28.00
T5	4.72	1.68	8.00	3.50	11.50	384.14	−33.00
T6	5.03	1.78	7.80	3.90	11.60	361.70	−39.00
T7	5.29	1.88	7.40	4.10	11.50	338.40	−43.00
T8	5.60	1.99	7.20	4.40	11.60	314.12	−45.00
T9	5.91	2.10	7.20	4.70	11.80	288.94	−48.00
T10	6.17	2.19	8.90	6.20	15.00	262.94	−48.00
T11	6.47	2.30	9.00	6.20	15.20	235.30	−46.00
T12	6.74	2.39	11.00	7.20	18.10	204.56	−44.00
L1	7.04	2.50	11.10	6.50	17.50	171.07	−37.01
L2	7.30	2.59	10.90	6.00	16.80	135.03	−29.00
L3	7.61	2.70	10.70	5.50	16.10	97.55	−17.00
L4	7.87	2.79	11.20	5.30	16.40	58.90	−10.00
L5	8.19	2.91	12.20	5.60	17.70	20.57	−6.00
S1	0.00	0.00	0.00	0.00	0.00	0.00	0.00
Pelvis	–	11.00	75.00	30.00	80.00	−89.00	0.00

[a]TM: trunk mass, BM: body mass, *Ixx*, *Iyy*, *Izz*: mass moments of inertia, respectively, in anterior–posterior, transverse and longitudinal directions ($kg\,m^2 \times 10^{-3}$), CG-*z*: height of the centers of mass with respect to the S1 (mm), CG-*x*: anterior–posterior distance from corresponding vertebral centers with negative indicating anterior position (mm). Upper arms, lower arms and hands centers of mass are considered posteriorly at T2, T3 and T4 vertebral levels, respectively.

follow curved paths whenever their distances from T12 to L5 vertebral centers decrease below 90% of their respective values at undeformed configuration (i.e. to reach the limit values of 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12-L5 vertebrae, respectively). This wrapping mechanism, similar to that formulated in our earlier simulations [38,45,46], is considered in order not to allow the line of action of these muscles approach unrealistically close to the

Table 2
Buttocks' nonlinear stiffness properties

Force (N)	Deflection (m)	Tangent stiffness (kN/m)
0.0	1.0	
0.0	0.0	0.0
−12.5	−0.001	12.5
−62.5	−0.002	50
−1232.5	−0.02	65
−7232.5	−0.05	200

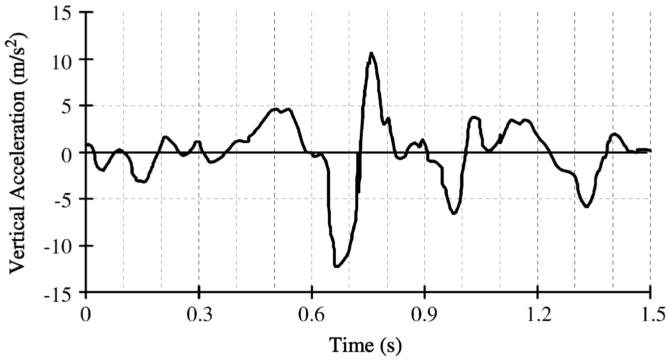

Fig. 4. The input random seat vertical acceleration at the base considered in the model based on the measured un-weighted acceleration at the seat–driver interface of a hydraulic excavator [27].

vertebrae while at the same time simulating a maximum of 10% reduction in their lever arms as observed during forward flexion tasks [47,48].

2.4. Input excitations and parametric studies

The input seat vertical acceleration at the base in the model is taken based on measured unweighted acceleration at the seat–driver interface of a hydraulic excavator [27] (Fig. 4). Seidel et al. [27] have reported trunk response and seat–subject interaction for a subject, freely sitting on a rigid seat, under a base excitation at three different postures (i.e. driving, bent forward and erect). In accordance with these measurements, two different lumbar postures are considered in the current model study that remain unchanged during the entire vibration duration (Table 3); an erect posture as the reference case [25] and a flexed posture in which the lumbar lordosis is flattened at all levels by a total of 10° [49,50] to simulate a relax sitting posture (i.e. slouch posture). To evaluate the effect of antagonistic co-activity in abdominal muscles on load distribution between active and passive systems and trunk stability, a-priori low to moderate abdominal coactivity levels of 2%, 1% and 0.5% [51] are considered in the internal oblique (IO), external oblique (EO), and rectus abdominus (RA) and analyses are repeated for the case with erect posture. Finally, effect of changes in the axial stiffness of the spring at the base simulating buttocks (i.e. softening as much as to yield deflections of ~15 mm and stiffening as much to simulate a rigid base) on trunk response to base excitation as well as system natural frequency is also investigated.

3. Results

The input vertical acceleration at the base varied primarily in the range of $\pm 5\,m/s^2$ with rms value of $1.4\,m/s^2$ and two relatively sharp peaks in input signals (i.e. -12 and $+10\,m/s^2$) (Fig. 4). The computed vertical accelerations at the L3 level and head as well as base reaction force were in general agreement with measured values (Fig. 5) [27]. The effect of posture on computed accelerations was found to be negligible. The net

Table 3
Prescribed total sagittal rotations (degree) at different levels in two modeled seated postures (negative: flexion)

Level	Erect	Flexed
Head-T12	−9.9	−15.9
L1	−6.2	−11.5
L2	−2.5	−6.5
L3	0.3	−2.1
L4	1.8	1.8
L5	3.5	6.0
S1	5.0	9.0

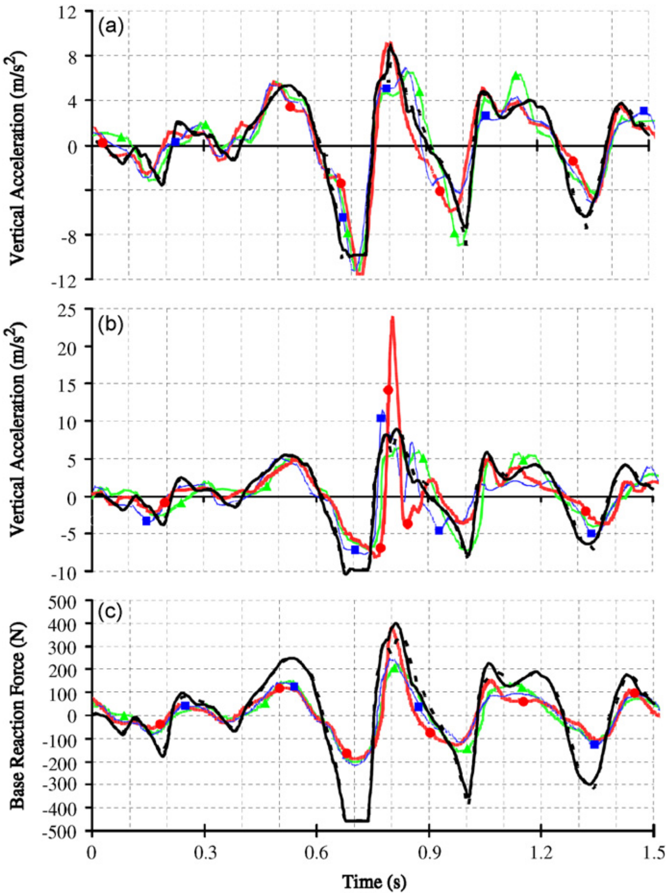

Fig. 5. Measured and predicted response at the L3 (a) and head (b) along with reaction force at seat–subject interface (c). Three postures considered in experimental studies (i.e. driving, bent, and upright) [27] and two in the current model studies (i.e. erect and flexed) are presented. The base reaction force is given relative to the upper body weight.

moment at the S1 base remained almost positive during the entire period and was greater in the flexed posture than in the erect one (Fig. 6). This net moment is generated by trunk gravity and inertia forces above the S1 level and is balanced by passive ligamentous structure and muscle forces. The contribution of passive ligamentous spine in balancing the net external moment was initially at the onset of vibration at 48% and 87% that dropped at the peak positive acceleration to minimum values of 13% and 22% in erect and flexed

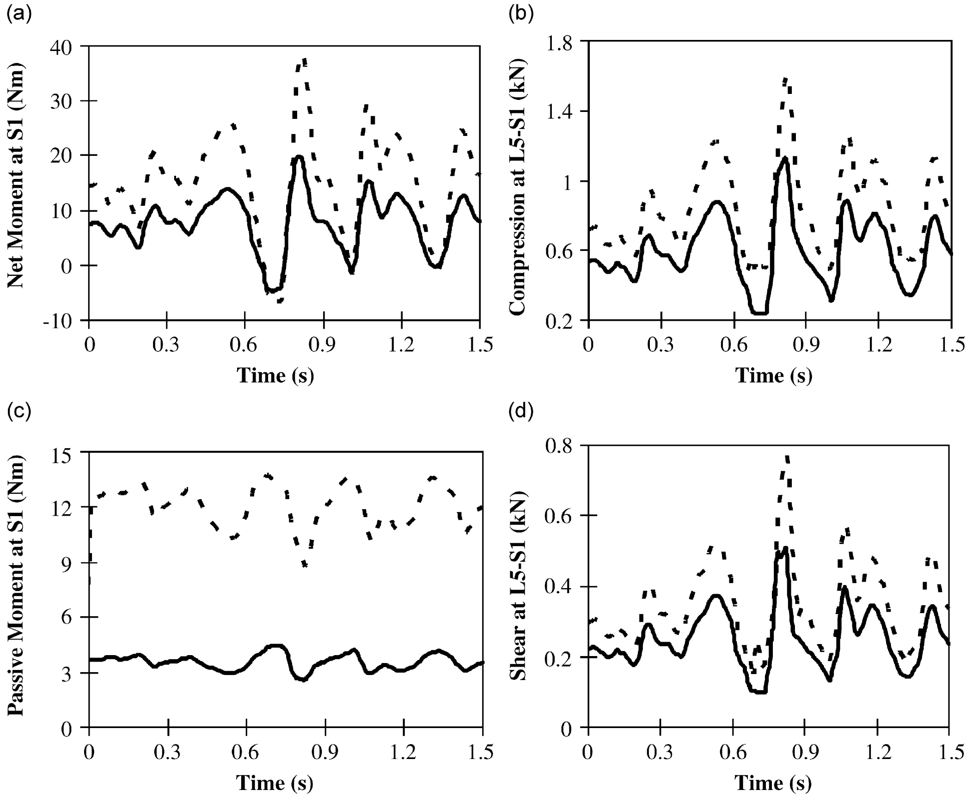

Fig. 6. Predicted temporal variations of internal spinal forces at the L5-S1 disc mid-height in local directions (b: compression force, d: shear force), the net moment at the S1 (to be resisted by all muscles and passive ligamentous spine, a) and the passive ligamentous moment at the S1 (c) for erect and flexed postures. ——— Erect and ▪▪▪▪▪▪ flexed.

postures, respectively (Fig. 6). Local compression and shear forces reached their maximum values at the L5-S1 level and were larger in flexed posture with maximum values of 1608 and 771 N, respectively, as compared with 1131 and 508 N for the erect posture. The associated computed forces in global extensor (LG and IC) and flexor (RA, IO and EO) muscles showed no coactivity and followed the temporal trends predicted for net moments and spinal loads (Fig. 7). Due to larger net moments, muscle forces were greater in the flexed posture than in the erect posture. It is important to note that the subject momentarily separated from the seat when the seat downward acceleration exceeded that of gravity resulting in a substantial decrease in base reaction force and compression on the spine. Nevertheless, the duration of this period and the exceeding acceleration magnitude were both small.

The critical (minimum) muscle stiffness coefficient to maintain trunk stability was substantially influenced by muscle activity and posture (Fig. 8). The trunk was much more stable (i.e. smaller critical q values) during periods with activity in abdominal muscles and on the contrary less so at periods with peak activity in extensor muscles (Fig. 8). The flexed posture resulted in an overall improvement in trunk stability due primarily to greater resistance provided by passive tissues.

Introduction of 2%, 1% and 0.5% as a-priori antagonistic coactivity in the abdominal IO, EO and RA muscles, respectively, substantially improved the trunk stability in the erect posture (i.e. requiring much smaller critical q values) during periods when no abdominal activity was otherwise present (Fig. 9). The foregoing prescribed antagonistic abdominal coactivities diminished the critical muscle stiffness coefficient from 89 to 21 at the onset of vibration and from 131 to 39 at the peak upward acceleration. The spinal loads, on the contrary, increased in the same periods by as much as 196 N in compression force and 138 N in shear force.

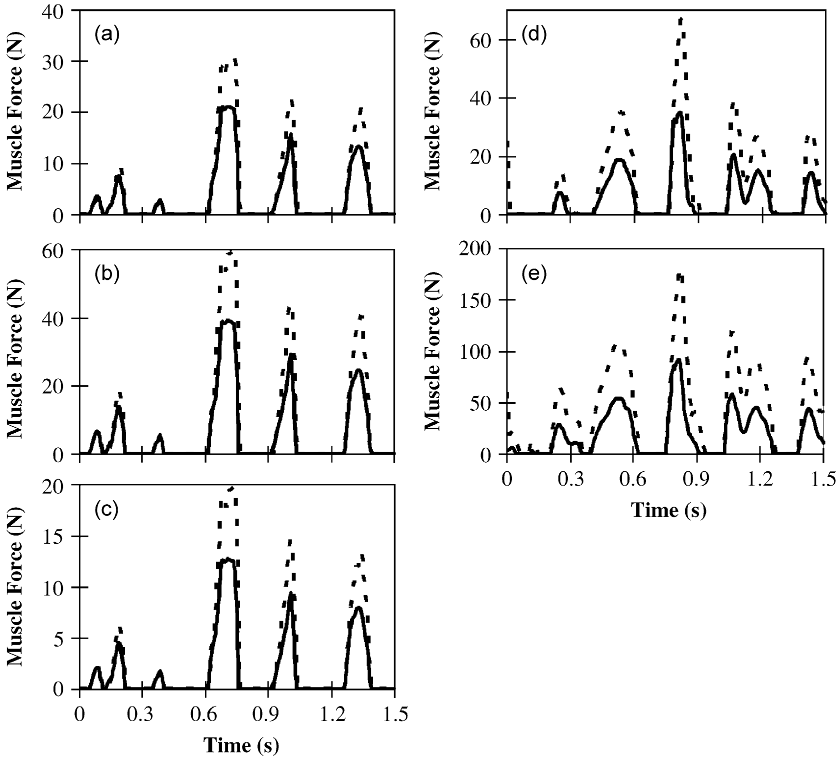

Fig. 7. Predicted temporal variations of active forces in global muscles (one side only): (a) external oblique, (b) internal oblique, (c) rectus abdominus, (d) iliocostalis and (e) longissimus. ——— Erect and ▪▪▪▪▪▪ flexed.

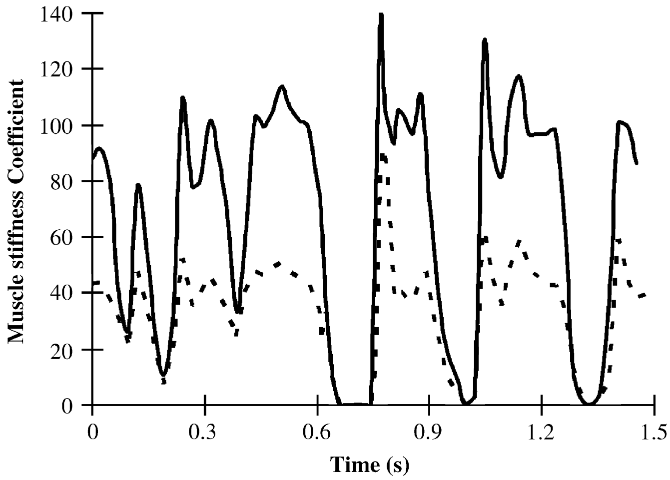

Fig. 8. Predicted temporal variations of minimum (critical) muscle stiffness coefficient, q, for two different lumbar postures. A lower q value indicates a higher trunk stability. ——— Erect and ▪▪▪▪▪▪ flexed.

Softening of the buttocks did not affect the response as much. On the contrary, stiffening of buttocks increased the magnitude of accelerations; for example the peak acceleration at the head reached $\pm 17\,\mathrm{m/s^2}$ for the case with rigid buttocks. Furthermore, the first natural frequency of the system under gravity load and prescribed postures was ~5.5 Hz that altered to ~12.4 or 3.9 Hz as buttocks became completely rigid or softer, respectively.

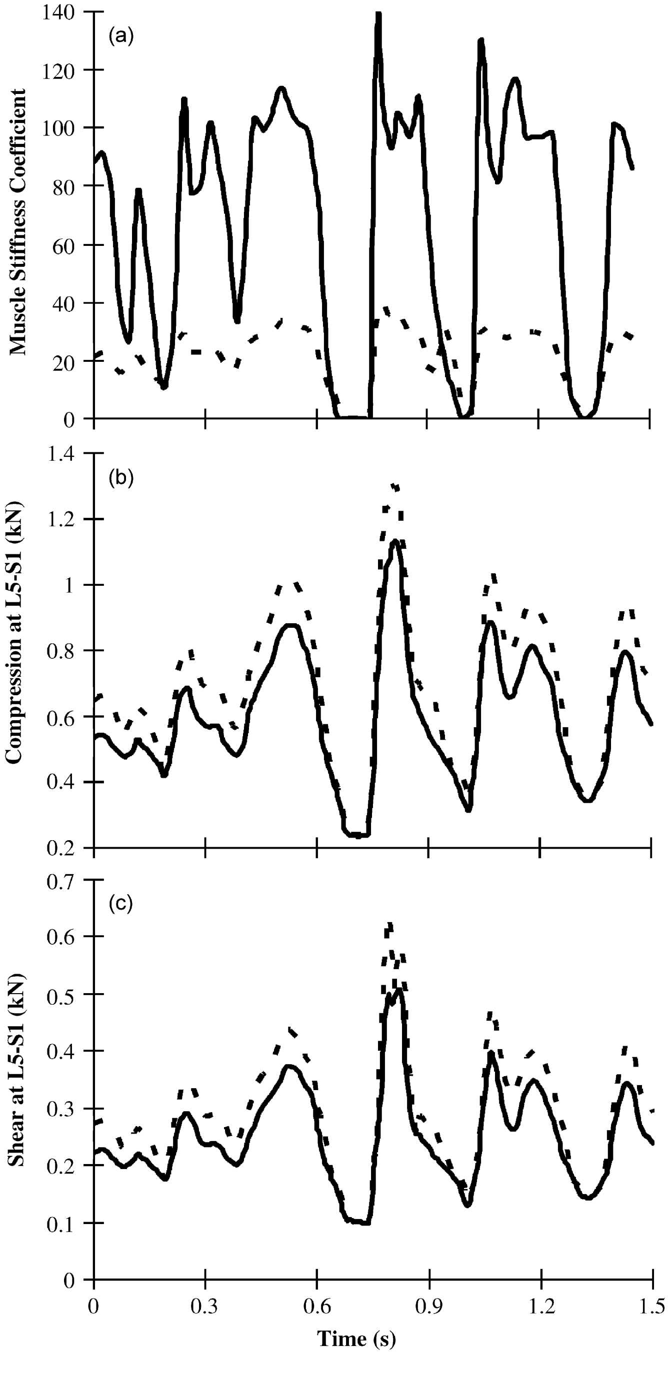

Fig. 9. Predicted temporal variations of minimum (critical) muscle stiffness coefficient, q (a) and compression (b) shear forces (c) at the lowermost L5-S1 levels in the erect posture for cases with and without prescribed coactivity in abdominal muscles (see the text for coactivity levels). ——— No coactivity and ▪▪▪▪▪▪ with coactivity.

4. Discussion

The iterative kinematics-driven finite element approach was used to solve the redundant passive–active trunk system at the seated position subject to an input random WBV. The time variations of trunk muscle forces, spinal loads and trunk stability were evaluated. This is a novel investigation performed in response to the recognized need for development of more anatomically detailed biomechanical models of trunk in WBV biodynamics [19,52].

There was a good agreement between predicted and measured [27] accelerations at the L3 and head levels (Fig. 5). Noticeable differences in the force magnitudes at the seat–subject interface at the times of peak acceleration (Fig. 5) could partly be due to the dynamic contributions of thighs and legs that were absent in the model but present in measurements. The predicted dynamic component of base reaction force reached the maximum of 400 N at peak positive acceleration (i.e. $10\,m/s^2$) whereas the minimum of -460 N (i.e. equal to the trunk weight) at the peak negative acceleration (i.e. $-12.2\,m/s^2$).

A single L2-L3 lumbar motion segment has been reported to yield an axial natural frequency of ~32 Hz [42]. This natural frequency dropped to ~18 Hz when considering a finite element model of two lumbar motion segments, L4-S1 [53] and furthermore to ~11 Hz as the entire lumbar spine is considered [15]. The incorporation of buttocks in the current model with proper stiffness and damping values [13,44] diminished the first vertical natural frequency of the seated trunk under gravity from ~12 to ~5.5 Hz, in agreement with earlier measurement studies [15,54]. With no constraint on sagittal displacements, vibration analysis at the loaded configuration yielded the lowest frequency of ~1 Hz that is also in agreement with the reported values in the literature [55].

The excitation at the base substantially influenced, via the inertial and muscle forces, the spinal compression and shear forces. The maximum spinal compression and shear forces predicted at the L5-S1 increased significantly from the initial (static) values of, respectively, 535 and 222 N in the erect posture and 717 and 297 N in the flexed posture to peak (dynamic) values of 1131 and 508 N in the erect posture and 1608 and 771 N in the flexed posture. Back pain prevention and rehabilitation programs are designed based on the premise that excessive loads in the ligamentous spine could cause injury. The compression strength of lumbar motion segments has been reported to be in the range of 2–10 kN [56–58]. Jager and Luttmann [59] reported values of 5.81 ± 2.58 kN for males and 3.97 ± 1.5 kN for females based on relatively large sample populations. The strength in shear force has been reported to be >1 kN [60,61]. Notwithstanding the effect of strain rate, earlier injuries/degeneration, fatigue and combined loading on these strength values [62], lower risk of injury could be associated with conditions that yield smaller loads on spine. Hence, considering the mild shock contents of $\sim\pm 1$ g in vibration input data of the current study as compared with reported larger shock values of $\sim\pm 2$–6 g in off-road and industrial vehicles [12], it is likely that spinal loads in latter vibration environments approach and even exceed strength limits causing injury.

The biomechanical advantages of preservation or flattening (i.e. flexing) of the lumbar lordosis in various activities such as sitting or lifting remain controversial. In a recent study [20], the free style posture or a posture with moderate lumbar flexion was advocated as the posture of choice in static lifting tasks. The 10° flattening in the lumbar lordosis from the erect to flexed posture in the current sitting position under WBV increased the net moment, muscle forces and passive spinal loads and hence the risk of tissue injury while on the other hand substantially improved the trunk stability. The latter effect was due to the increased stiffness of muscles and passive ligamentous spine in the flexed posture. The moment at the S1 as well as compression and shear forces at the L5-S1 increased in the flexed posture by, respectively, 6.8 Nm, 182, and 75 N at the onset of vibration and by 19 N m, 482, and 240 N at the time of peak positive acceleration response. Similarly, the introduction of antagonistic coactivity in abdominal muscles at low to moderate levels increased the passive spinal loads while improving the spinal stability. It appears hence that an increase in trunk stability can be achieved only at the cost of higher passive loads and hence greater risk of tissue injury suggesting a tradeoff in muscle activities.

Activation in muscles has opposite effects on the spinal stability; on one hand, it increases compression force on spine (i.e. destabilizing role); but on the other, it offers greater stiffness associated with larger activation (i.e. stabilizing role). Furthermore, due to larger lever arms [51], abdominal muscles even with much smaller forces are more efficient than extensor muscles in stabilizing the trunk. Results demonstrate much lower critical q values in presence of abdominal activities, being agonistic or antagonistic. The positive role of antagonistic activities in enhancement of the spine stability has been recognized in the literature [24,25,63]. It should be emphasized that the choice of the widely used linear force–stiffness relationship (i.e. intrinsic muscle stiffness) [20,32] assumed in this study rather than a nonlinear one (i.e. reflexive stiffness) [64,65] has absolutely no influence on computed muscle forces and, hence, internal spinal loads. Nevertheless, the choice of force–stiffness relationship would influence the system stability. Moreover, consideration of delays in muscle spindle reflex response [65] in vibration would have no bearing on muscle forces computed in this study as these forces are inclusive of passive, reflexive and areflexive contributions under prescribed motions. As the damping matrix of the system is symmetric and positive definite, for a given loading and stressed state, the spinal system ceases to be stable as its lowest natural frequency reaches zero [33]. Static perturbation and frequency analyses yielded identical critical muscle stiffness coefficient values when instantaneous stressed configurations were considered.

Due to changes in contact area between buttocks/thighs and the seat as the lumbar posture alters, the stiffness of the element simulating buttocks may need to be modified [55]. Although no such changes were considered in the current study, the first natural frequency of the model was found to be highly dependent on the buttocks stiffness demonstrating the likely indirect effect of posture on the natural frequency. Rigid buttocks increased the system resonant frequency from 5.5 to 12.4 Hz, increased the acceleration response especially at the time of peak acceleration, and resulted in larger net moments. Reverse trends were found as the buttock stiffness decreased. These findings likely indicate the effect of seat cushion compliance on vibration response.

In conclusion, a detailed trunk muscle architecture along with nonlinear properties of the ligamentous spine, wrapping of global extensor muscles and trunk dynamic characteristics (inertia and damping) was used in our kinematics-driven model to evaluate muscle forces, spinal loads and trunk stability under a random base WBV reported in the literature [27]. The predictions satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Large net moments, muscle forces and spinal loads along with a deteriorated stability margin were predicted at the vibration durations with peak accelerations. The flexed posture, compared with the erect one, increased muscle forces and spinal loads but improved trunk stability. Agonistic or antagonistic activities in abdominal muscles substantially improved spinal stability. Results points to a likely tradeoff between the opposing effects of higher muscle forces in improving trunk stability on one hand but increasing spinal loads and risk of injury on the other. Additional muscle coactivity, as a compensatory response to an injury in the passive spine or to insufficient stability for example, would further increase the spinal loads and the risk of fatigue and failure.

Acknowledgment

The work was supported by grants from the NSERC—Canada and Aga Khan Foundation.

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