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DYNAMIC FE MODEL OF SITTING MAN ADJUSTABLE TO BODY HEIGHT, BODY MASS AND POSTURE USED FOR CALCULATING INTERNAL FORCES IN THE LUMBAR VERTEBRAL DISKS

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Long-term whole-body vibrations can cause degeneration of the lumbar spine. Therefore existing degeneration has to be assessed as well as industrial working places to prevent further damage. Hence, the mechanical stress in the lumbar spine—especially in the three lower vertebrae—has to be known. This stress can be expressed as internal forces. These internal forces cannot be evaluated experimentally, because force transducers cannot be implemented in the force lines because of ethical reasons. Thus it is necessary to calculate the internal forces with a dynamic mathematical model of sitting man.

A two dimensional dynamic Finite Element model of sitting man is presented which allows calculation of these unknown internal forces. The model is based on an anatomic representation of the lower lumbar spine (L3–L5). This lumbar spine model is incorporated into a dynamic model of the upper torso with neck, head and arms as well as a model of the body caudal to the lumbar spine with pelvis and legs. Additionally a simple dynamic representation of the viscera is used. All these parts are modelled as rigid bodies connected by linear stiffnesses. Energy dissipation is modelled by assigning modal damping ratio to the calculated undamped eigenvalues. Geometry and inertial properties of the model are determined according to human anatomy. Stiffnesses of the spine model are derived from static in-vitro experiments in references [1] and [2]. Remaining stiffness parameters and parameters for energy dissipation are determined by using parameter identification to fit measurements in reference [3]. The model, which is available in 3 different postures, allows one to adjust its parameters for body height and body mass to the values of the person for which internal forces have to be calculated.

1. INTRODUCTION

Long-term whole-body vibrations (WBV) include an enormous health hazard for the lumbar spine, especially for the three lower vertebrae (L3–L5). As an increasing percentage of the population is exposed to WBV because of occupational reasons, more and more people have to face the hazard of an occupational disease (e.g., in Germany: 2110 BeKV) for which compensation can be claimed.

For assessing WBV exposed industrial working places concerning this hazard and for assessing WBV protection mechanisms concerning their effectiveness, one needs to know details about the lumbar spine loads caused by typical excitation signals. These loads can be expressed as internal forces (compression force and shear force) in the vertebral disks. These forces cannot be evaluated experimentally as force transducers cannot be implemented in the force lines because of ethical reasons.

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Thus it is necessary to calculate the internal forces using a dynamic mathematical model of sitting man. This model's input data, the base excitation $q(t)$ at the man-seat-interface (see Figure 1), is easily evaluated by an experiment.

2. MODELLING

2.1. REQUIREMENTS

When modelling biomechanical phenomenons there is always the problem of individual differences even between members of a relative small group of persons for which a model may be valid. For the vast majority of all anatomical properties, there is no knowledge about their influence on the dynamical behaviour of man. However, the model presented is required to be valid for different individuals. So the modeller has to concentrate on properties that have an obvious influence on man's dynamical behaviour, like body height and body mass, that affect man's terms of inertia (mass and mass moment of inertia).

Since industrial working places are very different from each other and since experimental studies like reference [3] report that the subjective feeling of the WBV exposed person is substantially affected by the person's posture during the exposure, the model has to be adjustable to a posture that is typical for a special working place.

The objective of the model is to estimate unknown internal forces in special regions of the human body from known input signals. Therefore, a phenomenological model (this means a model that reproduces known outputs from known inputs) cannot be applied. The requirement is to build a model based on human anatomy, which is sufficiently detailed in the area of interest—the *regio lumbalis* with the lower lumbar spine itself, the lower back muscles and the viscera. Generating results with the model furthermore should not take too much computation time, so that there is a limit to the complexity (number of degrees of freedom).

2.2. DESCRIPTION OF THE MODEL

Pro/Mechanica, a FEM program family, is used for building the model. The advantage of Pro/Mechanica is that it is an "all-in-one" system with preprocessor, equation solver and postprocessor. Since the preceeding detailed analysis of the measurement data from

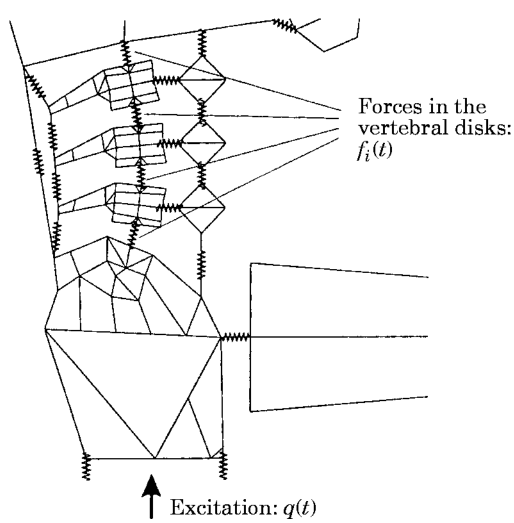

Figure 1. The model's objective.

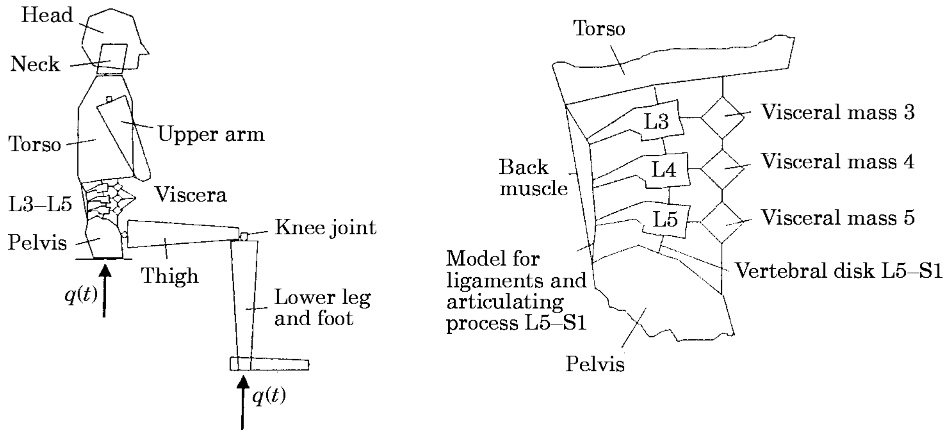

Figure 2. The complete model (left) and the *regio lumbalis* (right).

the experiment in reference [3] has not given an indication of significant non-linearities, the linear part of the Pro/Mechanica family (Pro/Mechanica *Structure*) can be used.

A view of the complete two dimensional model as well as the *regio lumbalis* is given in Figure 2. A study in reference [4] recommends modelling each vertebrae as a rigid body when elasticities of the posterior elements can be taken into account by modelling the vertebral disks. As not the whole lumbar spine (L1 to L5) but only L3–L5 is modelled here, the stiffnesses of the spinal segments T12/L1–L2/L3, which can be seen as a series connection of three springs (here: vertebral disks, because vertebral bodies are modelled as rigid bodies), are combined in the vertebral disk cranial of L3 by applying a stiffness value of a third of the stiffness of a single vertebral disk. The inertial properties of the spinal segments T12/L1–L2/L3 are included in the inertial property of the model's torso. Another part of the *regio lumbalis* is the viscera, the properties of which are represented by an elastic three-mass-chain, which has an elastic connection to the lumbar spine model. This chain does not correspond to anatomy, but it provides a good copy of the dynamically important properties of the viscera in the interesting frequency range (mass, local model shapes and local natural frequencies). The lower back muscles are combined to a linear spring. The stiffness of that spring is derived from the model in [5], which takes into account 18 different back muscles and their non-linear and frequency dependent behaviour. The distribution of muscle forces in reference [5] is obtained by applying a least squares procedure to the sum of all muscle stresses. The presented model takes the resulting stiffness of all muscles between upper torso and pelvis of the model in [5] as muscle stiffness using stiffness values at 5 Hz (main resonance of the sitting human). A constant force F (see equation (3) of section 2.3), representing the active muscle force, can be superposed to keep the torso in an erect position (see Figure 3). In addition to this constant force a constant moment M (see equation (4) of section 2.3) acting on the pelvis is necessary in order to keep the torso erect (see Figure 3). This moment represents the stabilizing influence of all muscles caudal of the pelvis. Values of force and moment are adjusted to keep the model in equilibrium when it is exposed to gravity.

The periphery of the lumbar spine model is less detailed in order to decrease computation time. Here five rigid bodies are used to build the regions cranial of the lumbar spine. These rigid bodies are upper torso, neck, head, upper arm and forearm. In Figure 2 the forearm cannot be seen as in this posture the arms are folded in front of the chest.

The caudal region of the lumbar spine also consists of rigid bodies representing pelvis, thigh, lower leg and foot. So the whole model consists of 14 rigid bodies connected by linear stiffnesses.

The model parameters (inertia, stiffness and damping) are taken from the biomechanical literature. If this literature is insufficient, parameters have to be gained by comparison of calculated model results with measurements from reference [3] (parameter identification in the frequency domain).

Terms of inertia and relating geometry parameters are defined in the model's standard posture (sitting man with erect torso and folded arms). Therefore, numerous literature is used [6–8]. The terms of inertia of a specific body part of the model are accepted as correct, if the part's mass and the position of its center of gravity in the sagittal plane are in good agreement to the literature. The mass moment of inertia of a single body part is not so important to be represented correctly [9, 10]. Inertia properties (masses, mass moments of inertia and location of the center of gravity of the different body parts) are listed in Table 1. The location of the center of gravity depends on the model's posture.

Stiffness properties are determined both from literature and by parameter identification, which in this case means editing parameters so that model results fit measurements in reference [3]. Since cervical and thoracic regions of the spine are not modelled here, their stiffnesses have to be included by a method recommended in reference [8]: the springs between head and neck must be adjusted to the elasticity of the cervical spine, and the spring stiffness between neck and torso has to agree with the elasticity of the thoracic spine. The dynamical behaviour of the shoulder–arm area (local mode shape, natural frequency) is expected to be the same as in the more detailed model in reference [5], so the spring between arm and torso must be adjusted like that. The human elbow joint, the knee joint and the hip joint act like pin joints, so they can be modelled by using very high translational stiffnesses in both x and z directions of the sagittal plane.

Elasticities in the lumbar spine are modelled using several springs per spinal segment: One spring, representing both vertebral disk and longitudinal ligaments, connects the midpoints of the endplates of the adjacent vertebrae. It contains a stiffness in direction of

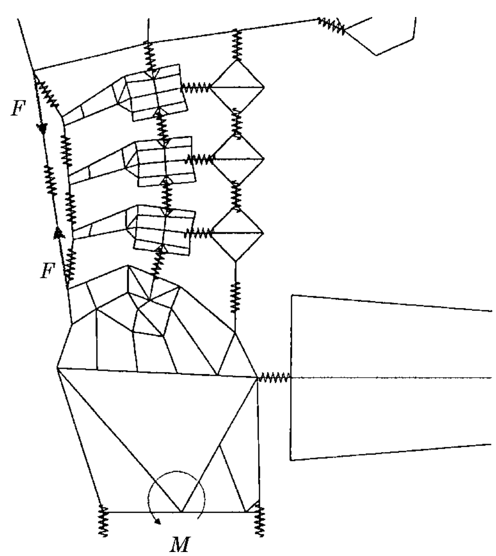

Figure 3. Force F and moment M to keep the torso erect.

TABLE 1

Inertia properties in different postures

Body part	Mass kg	Mass mom. of inerta, yy kgm^2	Standard CoG x mm	Standard CoG z mm	Bent forward CoG x mm	Bent forward CoG z mm	Relaxed CoG x mm	Relaxed CoG z mm
Head	4·42	2·96E-02	24·2	632·2	236·6	583·9	33·9	627·3
Neck	0·88	4·29E-03	18·3	585·3	215·2	541·7	24·0	581·6
Torso	18·00	2·72E-01	4·6	331·0	116·5	306·9	−11·2	329·3
Upper arm	6·67	4·95E-02	74·4	307·6	151·9	262·0	104·2	328·5
Forearm	2·74	3·05E-02	142·3	147·4	236·7	−53·0	376·2	293·8
Thigh	15·07	1·74E-01	246·1	−62·0	225·8	−89·6	249·5	−55·4
Lower leg	6·65	1.16E-01	477·8	−271·3	459·8	−301·6	480·8	−270·3
Foot	2·14	1·33E-02	561·0	−534·4	535·8	−566·9	565·4	−533·0
Pelvis	10·24	2·63E-02	−11·5	−20·0	−16·0	−30·4	−11·1	−18·8
Viscera (total)	8·00	–	–	–	–	–	–	–
Lumbar spine (total)	0·17	–	–	–	–	–	–	–
Sum	74·97	–	–	–	–	–	–	–

the spine and another stiffness perpendicular to the first. Another spring unites the properties of the posterior ligaments and the articulating processes: the ligaments can only transmit tensile forces and the articulating processes only can transmit compressive forces. Thus, both effects can be combined to a single linear spring acting between the endpoints of two adjacent spinous processes. Although stiffnesses of ligaments and articulating processes are not the same and both even non-linear, using a linear spring with a mean stiffness (dependent on estimated magnitude of relative motion; method of harmonic balance). Parameters are obtained by comparing model results with known *in vitro* measurements with cadaveric spinal segments [1, 2]. Since both springs do not act in the same line, they create the resulting bending stiffness of the spinal segment and ensure that the lumbar spine deforms like a beam as presented in references [11, 12].

The spring representing the lower back muscles has the resulting stiffness of all back muscles of the model in reference [5]. The viscera model is requested to have the same eigenvalues and eigenvectors as the model in reference [5], where the viscera is divided into several visceral layers. Each layers is represented by a single mass and is connected to the adjacent vertebral body and the two adjacent visceral layers by several springs.

The spring between pelvis and ground (=seat) is used for modelling the elasticities of the pelvis and the tissue between pelvis and ground. Its stiffness is obtained by parameter identification (criterion: fitting natural frequency of human main resonance, measured in reference [3]). All stiffnesses used in the presented model are listed in Table 2.

Energy dissipation can be modelled in Pro/Mechanica only by assigning a modal damping ratio to the calculated undamped eigenvalues. These model damping ratios (see Table 3) are a result of parameter identification (criterion: fitting magnitudes of impedance and transfer functions from seat to head in x and z, measured in reference [3]).

2.3. ADAPTATION OF BODY HEIGHT, BODY MASS AND POSTURE

An adaptation to body height, body mass and posture seems to be necessary, because measurements in reference [3] report a non-neglectable influence of those values on the human's dynamical behaviour: for example, an increasing body mass (and body height)

TABLE 2

Stiffnesses of all springs in the model (at the connections head-neck, neck-torso and pelvis-ground, stiffnesses should be understood as the sum of both springs applied there)

Connection	Spring stiffness (N/m) Axial	Spring stiffness (N/m) Perpendicular	Res. torsional stiffness (Nm/rad)
head–neck	120 000	∞	1200
neck–torso	120 000	∞	1200
torso–upper arm	66 650	75 740	–
upper arm–forearm	∞	∞	0
torso–forearm (contact)	100 000	0	–
torso–L3, disk	266 000	66 000	–
torso–L3, posterior	12 800	0	–
lumbar spine, disk	800 000	200 000	–
lumbar spine, posterior	38 000	0	–
back muscle	10 000	0	–
viscera–torso	32 000	22 000	–
viscera, internal	32 000	22 000	–
viscera–lumbar spine	22 000	1000	–
viscera–pelvis	32 000	22 000	–
pelvis–thigh	∞	∞	0
thigh–lower leg	∞	∞	0
pelvis–seat	64 000	132 000	147·4

results in a decreasing frequency of the impedance maximum, while the modulus of the impedance maximum is increasing [3].

Measurements in reference [3] have been done in three different postures: an erect posture with folded arms (called standard posture), a relaxed posture typical for truck drivers (called steering wheel posture) and a bent forward posture typical for crane operators. Both frequency of the impedance maximum and its modulus are higher in the bent forward posture than in standard posture. The modulus of the impedance maximum in steering wheel posture is lower than in standard posture [3].

The model can be adapted to represent the specific person for which the internal forces in the lumbar spine shall be calculated by transferring the parameters *BH* (for body height) and *BM* (for body mass) to Pro/Mechanica. These parameters (which do not have any physical meanings) have to be calculated from the personal data of the specific person

TABLE 3

Mode shapes 1–8, person 176 (82·0 kg) in standard posture, with damping ratio

Mode	Natural frequency (Hz)	Modal damping ratio (%)	Description
1	0·594	10	1. spinal bending mode
2	2·747	35	2. spinal bending mode
3	4·681	26	1. vertical mode
4	7·775	20	3. spinal bending mode
5	11·420	5	Local mode, shoulder
6	14·341	10	Local mode, viscera, vertical
7	15·392	10	Local mode, viscera, horizontal
8	18·376	15	2. vertical mode

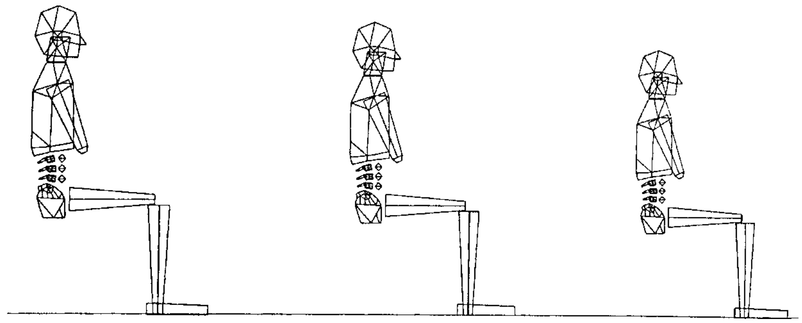

Figure 4. Models in standard posture; different values for BH (and BM).

(body height h_{body} in m, body mass m_{body} in kg). If both parameters are set to 50%, the model behaves like a 50% male. The parameters are calculated by equations (1) and (2).

$$BH = (h_{body}/1{\cdot}742)50, \qquad BM = (m_{body}/74{\cdot}97)\,(50^3/BH^2). \tag{1, 2}$$

The parameters BH and BM affect only geometry and inertial properties of the model. They do not affect stiffnesses or energy dissipation (modal damping ratio).

Models with different body height (and body mass) are shown in Figure 4.

Three different postures have been realized according to the experimental set-up of the study in reference [3] (see Figure 5). Any other posture can be created, if translation and rotations of pelvis and torso relative to the standard posture are known. The static forces (muscle force F and moment M on pelvis), which are necessary to keep the torso in equilibrium have to be changed when the model's posture changes. So they depend on the posture as well as on the parameters BH and BM and can be calculated by using equations

$$F = (BH/50)^2(BM/50)F_0, \qquad M = (BH/50)^3(BM/50)M_0. \tag{3, 4}$$

In equations (3) and (4), F_0 and M_0 are the static force and moment, that have to be applied when the model is set to represent the 50% male, i.e., $BH = BM = 50\%$. They depend on the posture and can be taken from Table 4.

3. CALCULATION OF DYNAMIC FORCES IN THE VERTEBRAL DISKS

The forces in the vertebral disks S1L5, L4L5 and L3L4 according to a base motion $q(t)$ at seat and feet (see Figure 2) can be calculated in two successive steps because of the

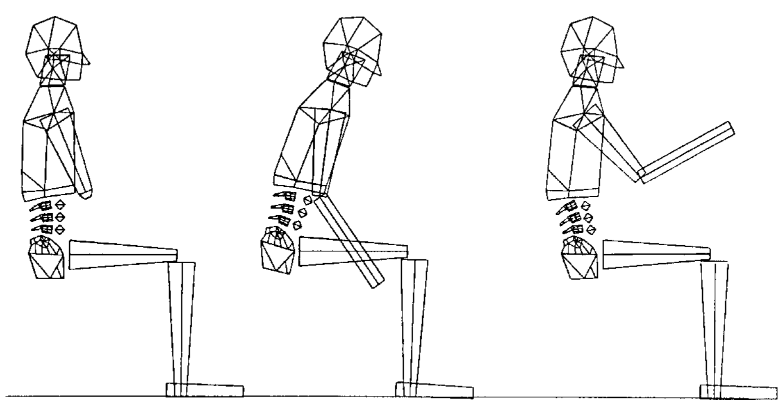

Figure 5. Realized postures: standard; bent forward; relaxed.

TABLE 4

Static forces for 50% male

Posture	F_0 (N)	M_0 (Nm)
standard	218	9
bent forward	605	45
relaxed	317	10

linearity of the model: (i) calculation of static forces according to gravity and time-invariant muscle forces F, M; (ii) calculation of dynamic forces according to a base motion $q(t)$ with mean value $\bar{q}(t) = 0$. The resulting forces are calculated by a superposition of the static and the dynamic forces (linear model). These force-time-functions can be processed by evaluating the crossing of class boundaries [13], so they are ready to use for assessing the fatigue strength of the lumar vertebral disks under the used excitation function $q(t)$, if the fatigue limit for the disk material is known.

The crossing of class boundaries procedure is a procedure for classification of time functions. In this case, 10 classes are spaced equally between maximum and minimum of the resulting force–time functions. These classes are limited by the dashed lines shown in Figure 6 (upper diagram). A cross is counted, when the signal (time function) crosses one of these dashed lines—the class boundaries—in a rising part of the curve. So the criterion of a countable cross of the class boundary b_j is given by, where f^i and f^{i+1} are force values at two succeeding time steps t^i and t^{i+1},

$$f^i < b_j < f^{i+1}. \tag{6}$$

A typical force-time function with classification is shown in Figure 6.

4. VERIFICATION OF THE MODEL

4.1. STATIC VERIFICATION OF SPINAL SEGMENTS

The most important part of the model is the lumbar spine, consisting of the spinal segments. These spinal segments have been verified using results from *in vitro* measurements reported in references [1] and [2] with different static load cases: compression, anterior shearing and flexion have been simulated with the model (according to the experimental set-up in references [1, 2]), and the calculated results have been compared with the results of the measurements. This comparison is shown in Table 5. There is a very good agreement in the main movements (printed *italic*) and a good agreement in the coupled movements.

4.2. VERIFICATION OF THE COMPLETE MODEL'S DYNAMIC BEHAVIOUR—COMPARISON TO MEASUREMENTS IN REFERENCE [3]

Verification of the complete model is done in the frequency domain with the transfer functions from seat to head in the x and in z directions and the mechanical input impedance. The calculated curves are compared with measurements in reference [3]. In the time domain, the calculated force between seat and pelvis, due to an excitation $q(t)$ used in the experiments in reference [3], is compared with the measured force between seat and pelvis.

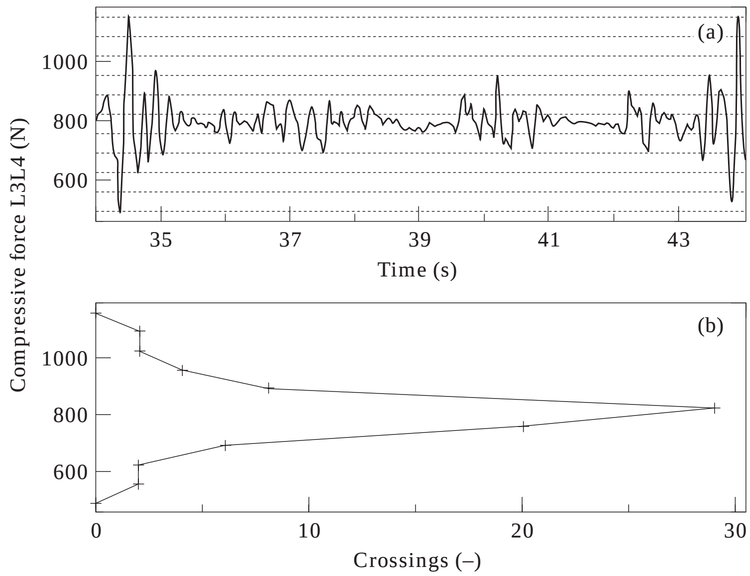

Figure 6. Typical results, for person 176, standard posture. (a) Force-time function; (b) crossing of class boundaries.

Figures 7 and 8 show calculated and measured mechanical impedances for an average person (Figure 7, body mass 82·0 kg) and a very light person (Figure 8, body mass 47·2 kg). The results show a good agreement with the measurements in the area of the human's main resonance in the z direction, while greater differences occur at higher frequencies.

Figure 9 compares the calculated transfer function from seat (z excitation) to head in x direction with the measured one. Again, there is a satisfactory agreement in the area of the dominating mode (here: first spinal bending mode at frequency $f < 1$ Hz). At higher frequencies ($f < 5$ Hz), results are less good.

A simulation result is shown in Figure 10. It shows the influence of body mass on the impedance: with increasing body mass, the maximum moves to lower frequencies and reaches higher magnitudes. The derivative of the impedance $I(\Omega)$ at $\Omega = 0$, which is the vibrating mass m^{vib} of the person, increases with the body mass (Ω = angular frequency):

$$m^{vib} = (\delta/\delta\Omega)I(\Omega)|_{\Omega=0}. \tag{6}$$

TABLE 5

Verification of the spinal segments

Static load case	Literature	Movement	Measurement	Model result
Compression	Berkson *et al*., 1979	*displacement in z*	−0·5 *mm*	−0·5 *mm*
		displacement in x	0 mm	0·05 mm
		Pitch (rotation in y)	0°	0·25°
Anterior shearing	Berkson *et al*., 1979	displacement in z	0·05 mm	−0·01 mm
		displacement in x	0·6 *mm*	0·6 *mm*
		Pitch (rotation in y)	1°	0·67°
Flexion	Schultz *et al*., 1979	displacement in z	−0·02 mm	−0·12 mm
		displacement in x	1·7 mm	1·42 mm
		Pitch (*rotation y*)	5·5°	5·65°

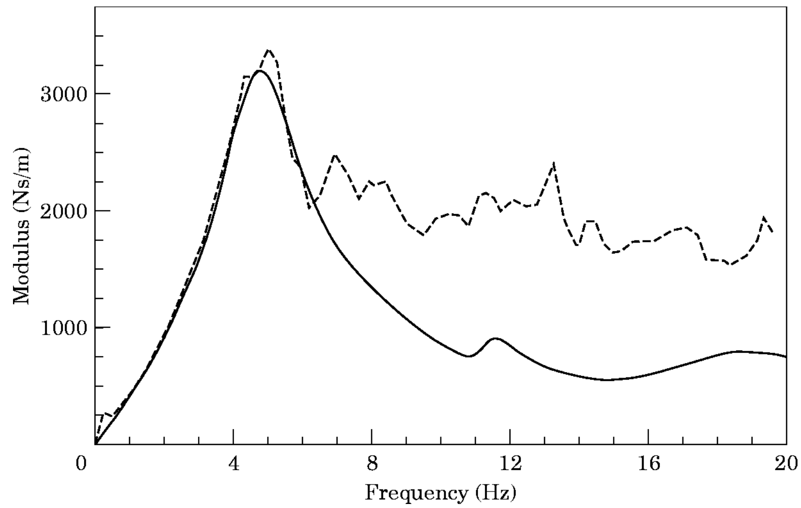

Figure 7. Comparison of measured [3] (----) and calculated (——) input impedance modulus. Person 176 (82·0 kg body mass); standard posture.

This simulation has been done to ensure that the model results make sense.

A time domain comparison of measured and calculated force (z direction, dynamic part only) between seat and pelvis is shown in Figure 11 (compressive force: negative sign). Differences occur only at peaks (high frequency events). This results agrees with the comparison of measured and calculated impedance (see Figure 7): there are only small differences at low frequencies but greater differences at higher frequencies.

The differences in the impedance at higher frequencies are, as known from the more detailed model in reference [5] using a more flexible software, a result of the way of modelling energy dissipation (Pro/Mechanica: only proportional damping; in general: discrete damper elements). The viscoelasticity of the vertebral disks, which has an increasing effect on the impedance at higher frequencies (sensitivity study in reference [5]),

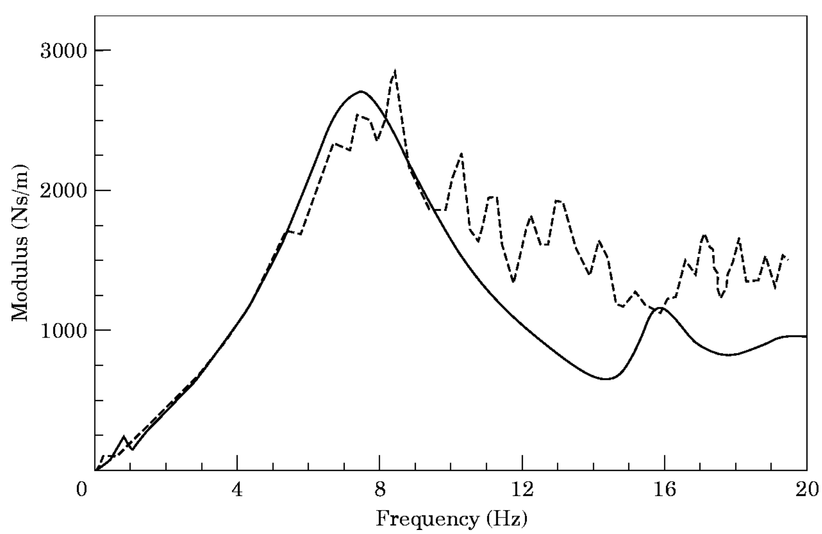

Figure 8. As Figure 7 but person 403 (47·2 kg body mass); bent forward posture.

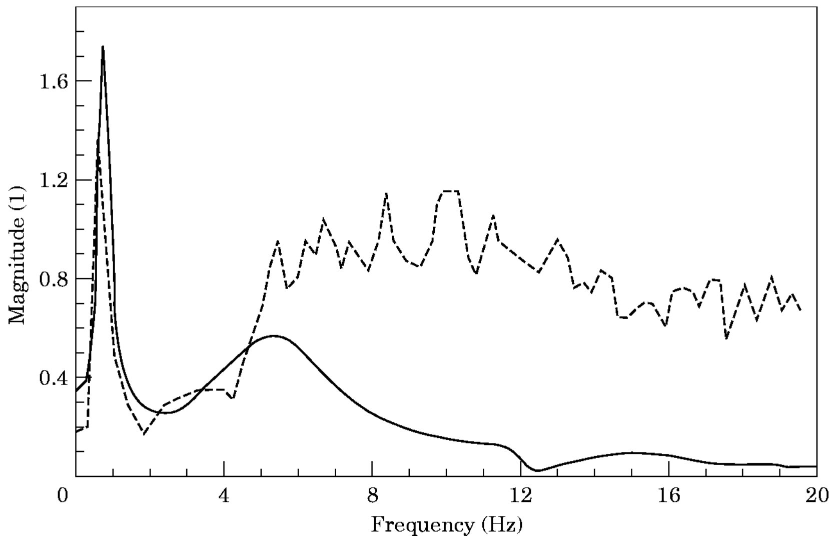

Figure 9. Comparison of measured [3] (----) and calculated (——) transfer function magnitude from seat z to head x. Person 176 (82·0 kg body mass), bent forward posture.

cannot be modeled without using discrete damper elements. So the possibility of modelling viscoelasticity would result in an increasing impedance in the frequency range where it is too low at the moment.

Another cause for differences in impedance at higher frequencies is the way of modelling muscles. In this model, muscle stiffnesses are linear and do not depend on frequency and muscle action potential. A study in reference [5] shows, that impedance increases at higher frequencies, when frequency dependance and dependance on muscle action potential is included in the muscle stiffness model.

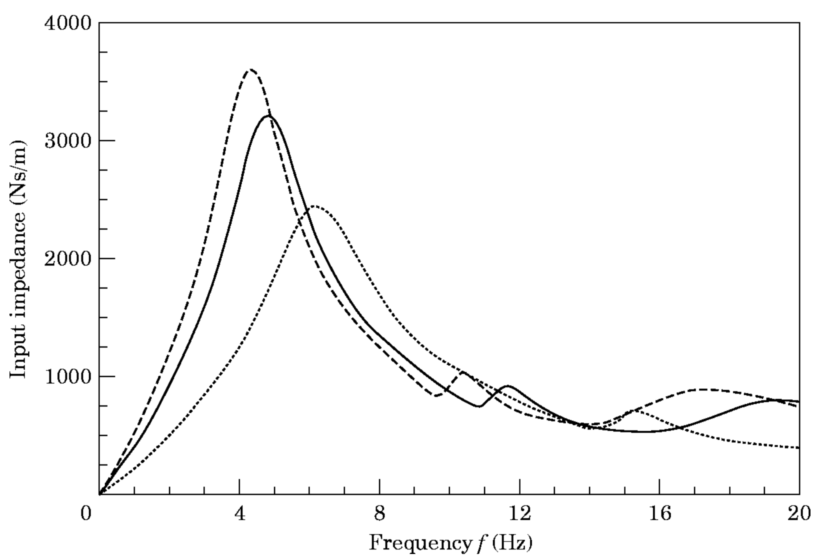

Figure 10. Model results showing changes in impedance magnitude with body height and body mass. Person number, body mass (kg): ——, 176, 82·0; ····, 403, 47·2; ----, 402, 103·0.

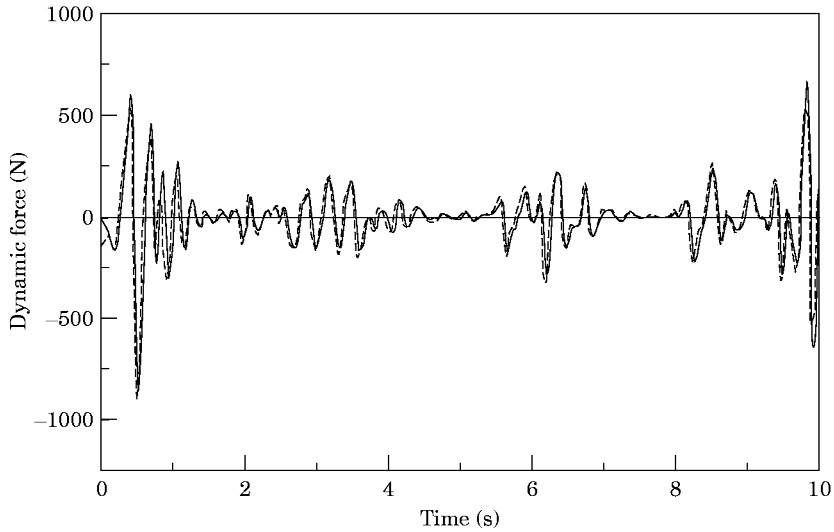

Figure 11. Comparison of measured [3] (----) and calculated (——) force between seat and pelvis (dynamic part).

However, the differences in the impedance do not imply any remarkable effects on the seat–pelvis force in the time domain as can be seen in Figure 11: there is no greater difference between measured [3] and calculated force, because the biggest amount of energy of the input signal (excitation $q(t)$ at the seat) occurs at lower frequencies (2–4 Hz), which are reproduced correctly by the model. If the excitation had dominant spectral density in the higher frequency range ($f >$ 7–8 Hz), the model would probably underestimate the internal forces in the lumbar spine and therefore should be used with care then.

5. CONCLUSIONS

The model can be used as a tool for estimating compressive forces and shear forces in the lumbar vertebral disks from an arbitrary base-excitation function $q(t)$. The computation time is short enough to be acceptable (approx. 8 min for 10 s excitation time on a 166 MHz PC). The spectral density of the excitation signal $q(t)$ should be biggest in the lower frequency range up to 7 Hz.

The model can be developed using Pro/Mechanica in one or more of the following ways: more detailed representation of muscular system and ligaments in the *regio lumbalis*; expansion to a three-dimensional model; different excitation functions at different body parts (pelvis, feet, arms, lower back); addition of the spinal segments T12L1–L2L3.

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