ANNALS of the ORADEA UNIVERSITY.

Fascicle of Management and Technological Engineering

MODELING OF HUMAN DYNAMICS

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Abstract: Vibration is most simply defined as oscillating motion. It could be periodic or nonperiodic. Repeated loading of the lumbar spine occurs in activities of daily living like lifting and driving. The chronic exposure results in mechanical and chemical changes in the spinal components leading to spinal degeneration. In a chronic vibration environment, the prevalence of low-back problems is dependent on a host of factors including subject age, subject posture, magnitude of input vibration, and exposure time. It is imperative that efforts be made to understand the effects of wholebody vibration on the spine and how these can be prevented. This paper focuses on the contributions of the mathematical models in this area.

I. INTRODUCTION

Vibration is most simply defined as oscillating motion. It could be periodic or nonperiodic. Repeated loading of the lumbar spine occurs in activities of daily living like lifting and driving. The chronic exposure results in mechanical and chemical changes in the spinal components leading to spinal degeneration. These disorders in a person may lead to discomfort, loss in productivity, and an enormous increase in health care cost to society. In a chronic vibration environment, the prevalence of low-back problems is dependent on a host of factors including subject age, subject posture, magnitude of input vibration, and exposure time. It is imperative that efforts be made to understand the effects of wholebody vibration on the spine and how these can be prevented.

This paper focuses on the contributions of the mathematical models in this area. These models help us understand the likely basis for these effects and help us identify ways in which the effects may be prevented or reduced.

Although the human body is a unified and complex active dynamic system, lumped parameter models are often used to capture and evaluate human dynamic properties. Lumped parameter models consisting of multiple lumped masses interconnected by ideal springs and ideal dampers have proven to be effective in

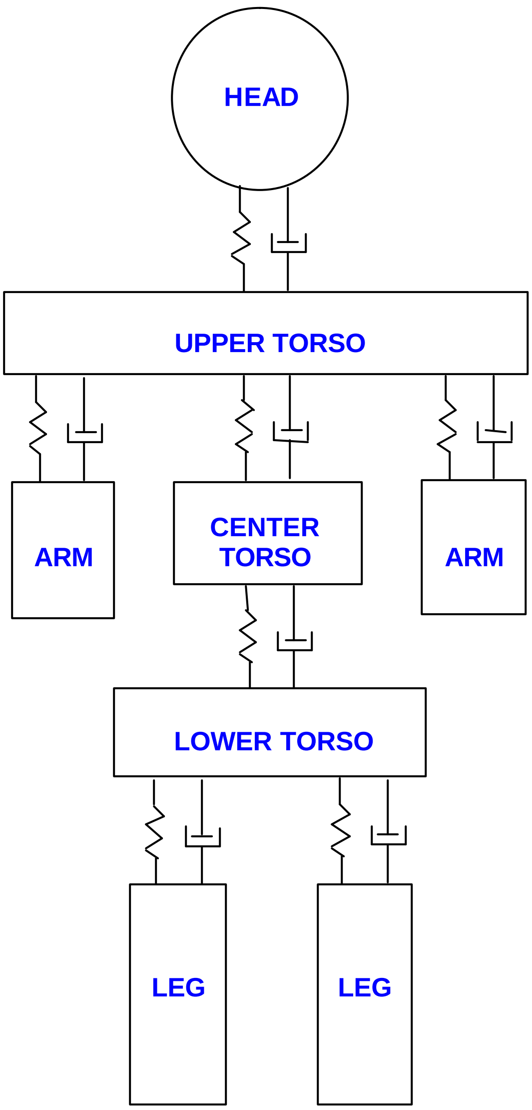

Figure 1.
General lumped parameter human model

many applications, including those involving human exposure to whole-body vibration. Figure 1 illustrates an example of a lumped parameter human model useful in the simulation of human response to vertical (longitudinal) vibration. The head, upper, center, and lower torsos, right and left arms, and right and left legs are modeled as lumped masses. The masses are connected together in the vertical direction by massless springs and dampers that capture human viscoelastic properties.

A chronological review of the lumped parameter models that have been developed to assess vibration exposure is presented below. The models are classified based on: (1) the directions of motion considered in model development (vertical axis only or multi-axis), and (2) the characteristics (linear or nonlinear) of the model spring and damper constitutive equations.

Four model categories are obtained using these criteria: vertical nonlinear models, multi-axis nonlinear models, vertical linear models, and multi-axis linear models. Below, for each model in each category, it provide: (1) a description of the interconnection of the model elements, (2) a description of the methods used to derive parameter values for each model element, and (3) an evaluation of model performance (wherever possible).

It also provide a graphical model description and more detailed model simulation results for the most recently published model within each model category.

II. VERTICAL NONLINEAR MODELS

In 1960, Coermann [1] presented a 6-degree-of-freedom (DOF) model of a human (for standing and sitting postures) used to simulate human dynamic response to longitudinal vibration of very low frequencies. This model included masses for the head, the upper torso, the arm-shoulder, a simplified thorax-abdomen subsystem, the hips, and the legs. A nonlinear spring was connected between the upper torso and the hips in parallel with the thorax-abdomen subsystem to represent the elasticity of the spinal column. Model parameters for each element were estimated from measurements of the mechanical impedance.

The performance of the whole-body model was not published and is therefore difficult to assess. The characteristics of the spine and the thorax-abdomen subsystem, however, were evaluated in detail. Each was modeled with 1 DOF in the whole-body model. Damping was not included in the spine and the performance of the thorax-abdomen subsystem did not match the experimental data particularly well.

In 1971, Hopkins [2] developed a 3-DOF model of a seated human consisting of the upper torso, viscera, and lower torso connected in series. Bilinear springs were used to connect the upper torso with the viscera and to connect the viscera with the lower torso. The vertebral column was represented by a linear spring connecting the upper and lower torsos. The model performance was compared with experimental impedance and transmission data. The model displayed the same number of resonant peaks as the experimental impedance data but had significantly different peak values. The model did not match the experimental transmissibility data, either in shape or in peak values. The model was used exclusively in the analysis of low-frequency vibration.

In 1974, Muksian and Nash [3] presented a 7-DOF nonlinear model dedicated to the analysis of vibration imposed on a seated human. The model included masses

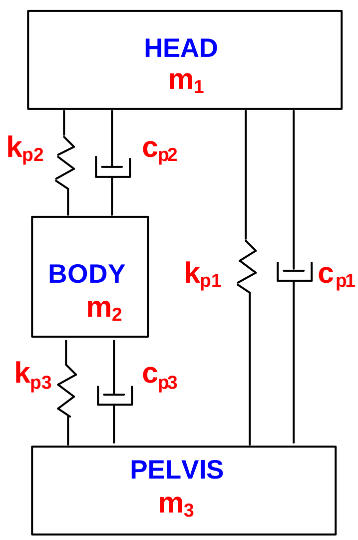

Figure 2. Vertical nonlinear Muksian

associated with the head, back, torso, thorax, diaphragm, abdomen, and pelvis. Linear springs and dampers were used between the head and the back, and between the back and the pelvis. Forces associated with the relative motion of the torso with respect to the back and muscle forces were included in the model as forces acting directly on the masses. The source of the stiffness values was not provided, but the values were similar to the experimental data obtained by Vogt et al. [7]. The damping coefficients were obtained from Coermann et al. [1] and Vogt et al. [7] (except that of abdomen-thorax viscera was an assumed value). The model performance was compared with the experimental data for acceleration ratio (for each mass relative to the input acceleration) given by Goldman and von Gierke [8] and Pradko et al. [9, 10]. At lower frequencies (1 to 10 Hz), the model matched the experimental data by Goldman and von Gierke [8] and Pradko et al. [10] well, but did not compare well with experimental data by Pradko et al. [9]. At higher frequencies, the model performance was significantly different than that observed experimentally.

In 1976, Muksian and Nash [11] presented a 3-DOF model of the human body in the sitting position that contained a parallel connection between the pelvis and the head. Figure 2 shows the model arrangement. It included masses associated with the head (m_1), body (m_2), and pelvis (m3) connected in series, very similar to the model given by Coermann et al. [47]. It neglected the arms and legs, and combined the mass of the upper torso and thorax-abdomen into that of the body. The model was based on the assumption that: (1) all springs (k_{p1}, k_{p2}, and k_{p3}) were linear in the frequency range between 1 and 30 Hz, (2) the damping between the head and body (c_{p2}) was zero, and (3) all other dampers ($c_{p1}$and c_{p3}) were linear between 1 and 6 Hz but nonlinear between 6 and 30 Hz. The values of the masses were obtained from Hertzberg and Clauser [50]. The spring stiffness and damping coefficients were determined by matching existing experimental data at corresponding input frequencies by Magid et al. [58] and Goldman and von Gierke [54]. The parameter values of the model are listed in Table 1. Since two kinds of damper were used for different frequency ranges, the model performed well when compared with experimental data for single-frequency input. However, since the damping values depend on the input frequencies, analysis of the model performance is difficult to assess for conditions involving multiple-frequency input (i.e., random vibration).

No. (i)	Mass m_i (kg)	Stiffness k_{pi} (N/m)	Linear Damping c_{pi}	f (Hz)	Nonlinear Linear Damping c_{31}	c_{33}
1	5.44	27158	1780	6	25462	15403
2	47.17	0	686	7	33949	157173
3	27.22	63318	467	10	11316	2027689
				15	5815	4346462
				20	1044571	7036478
				25	2358	10248780
				30	51533	15088608

Table 1. Parameter Values of the Muksian and Nash Model

III. MULTI-AXIS NONLINEAR MODELS

In 1964, von Gierke [59] described a two-axis, 7-mass model of a human in standing and sitting positions for longitudinal force application and pressures derived from the model presented by Coermann et al. [1]. The thorax-abdomen subsystem was extended to include one additional degree of freedom, the mass of the chest wall. A damper was added between the upper torso and the hips in parallel with the spine spring. Neither the values of the model parameters nor the model simulation performance were provided. This model was applied to the evaluation of motion of the abdominal wall, the diaphragm, and the lung and thorax.

In 1996, Broman et al. [12] described a 2-mass, 3-DOF model of a seated human (as shown in Fig. 3). It included a linear horizontal subsystem (k_1 and c_1), a vertical subsystem (k_2 and c_2), and a rotational subsystem (k_3 and c_3). The horizontal and vertical subsystems were used to represent the coupling between the human and the seat. The rotational subsystem was used to represent the rotation of the upper body relative to the lower body. The model parameters were varied to match the experimental data from Pope et al. [13]. The parameter values are shown in Table 2. The model simulation yields results similar to that of a purely vertical subsystem (the horizontal subsystem spring (k_1) was assumed to have infinite stiffness in the simulation results). In the comparison, the model matched the experimental data very well; however, different values of the model parameters were used when matching the different experimental data, i.e., a single "average human" model was not developed.

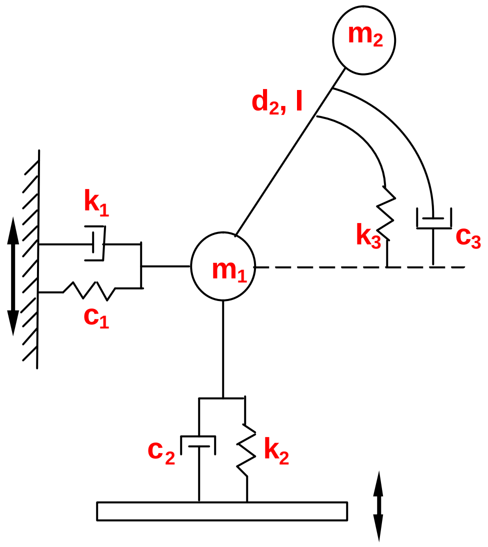

Figure 3.
Multi-axis nonlinear Broman model

Case	m_1 (kg)	m_2 (kg)	I (Nms^2/rad)	d_2 (m)	c_2 (Ns/m)	c_3 (Nms/rad)	k_2 (N/m)	k_3 (Nm/rad)
1	20	45	3.0	0.34	1500	50	60000	10000
2	9	56	3.0	0.39	2000	150	80000	17000

Table 2. Parameter Values of the Broman Model

IV. VERTICAL LOW-AMPLITUDE LINEAR MODELS

Prior to the 1970s, most published models had nonlinear stiffness and damping characteristics to account for the nonlinear behavior observed in the relatively large deformation human tissue studies (necessary in an impact analysis). In 1978, Sandover [14] experimentally investigated the linearity of the human body response to vibration. Results from his investigation indicated that the human body could be modeled as linear when using a 2 m/s 2 rms broadband random vibration stimulus — typical of many transport situations.

In 1981, the International Organization for Standardization (ISO) published a parallel 2-DOF model for both sitting and standing positions [15]. The model was developed to match a composite average driving-point impedance vs. frequency profile (magnitude and phase for the frequency range of 0.5 to 31.5 Hz) derived from existing experimental studies. Since the model had only two suspended masses, it was unable to match the phase response observed in existing experimental seat-to-head acceleration transmissibility studies at moderate to high frequencies [16] (phase angle of approximately 270°).

In 1987, ISO [16] published a 4-mass, 8-DOF model of a human for both sitting and standing positions. No correlation between the elements of the model and anatomical segments was established. Each springdamper set connecting masses included two springs and one damper (one spring parallel to the damper and the other in series). The model was developed to match a composite average seat-to-head acceleration transmissibility vs. frequency profile (amplitude and phase for the frequency range of 0.5 to 31.5 Hz) derived from existing experimental studies. The model matched the experimental data very well except for the transmissibility amplitude in the high-frequency range.

In 1987, Nigam and Malik [17] developed a 15-DOF undamped model for which only a standing posture was considered. It included masses for the head, neck, upper, central, and lower torso, upper and lower arms, upper and lower legs, and feet. The mass of each element was obtained from a previous anthropomorphic body segment study by Bartz and Gianotti [18]. The stiffness was obtained by combining the stiffness of adjacent segments. The model performance was compared with some experimental data such as resonance peaks from Goldman and von Gierke [8], and resonant frequencies for two modes from Greene and McMahon [19]. The natural frequencies of the model were in the range of the experimental resonant data but were relatively high. The leg stiffness was compared with the experimental values from Greene and McMahon [20]. The approximate value of the single leg was 15% larger than the experimental data. As damping was ignored in this study, the model is less realistic and general.

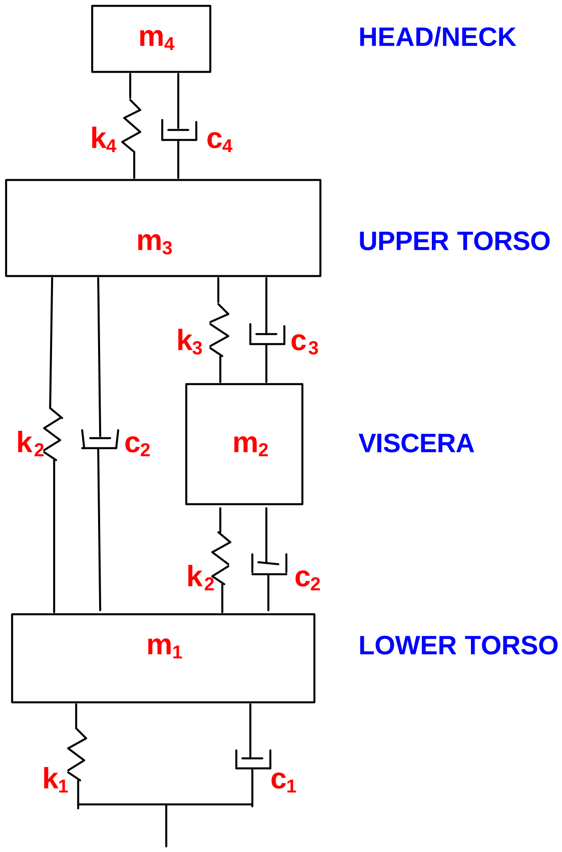

Figure 4.
Vertical Low-amplitude Wan and Schimmels Linear Model

In 1995, Wan and Schimmels [21] developed a series/parallel 4-DOF human dynamic model designed to match the response of seated humans exposed to vertical vibration. Since the model was constructed for subsequent use in optimal seat-suspension design, model simplicity was highly desired. The topology of the 4-DOF model is illustrated in Fig. 4. The model consisted of head/neck (m_4), upper torso (m_3), viscera (m_2), and lower torso (m_1). The model parameters were obtained by comparing simulation results with the results of experimental tests on human subjects to determine: (1) the variation of seat-to-head acceleration transmissibility with frequency, (2) the variation of driving-point impedance with frequency, (3) acceleration ratio from Goldman and von Gierke [8], and (4) the published properties of the human body from Patil and Palanichamy [22].

m_1 (kg)	m_2 (kg)	m_3 (kg)	m_4 (kg)	c_1 (Ns/m)	c_2 (Ns/m)	c_2' (Ns/m)	c_3 (Ns/m)	c_4 (Ns/m)	k_1 (N/m)	k_2 (N/m)	k_2' (N/m)	k_3 (N/m)	k_4 (N/m)
36	5.5	15	4.17	2475	330	909.09	200	250	49341.6	20000	192000	10000	134400

Table 3. Parameter Values of the Wan and Schimmels Model

The parameter values of the 4-DOF model are listed in Table 3. A comparison of simulation results (using the final refined model) with experimental data is presented below.

The 4-DOF model developed by Wan and Schimmels is judged to accurately capture the essential dynamics of a seated human exposed to vertical whole-body vibration. The 4-DOF model simulation results compare well with experimental data for transmissibility, impedance, and acceleration ratio. Also, its parameter values are close to the published properties of the human body. Relative to the results obtained using the previously developed models [15.16.20], the 4-DOF model provides a better match with experimental results for longitudinal vibration, despite its simplicity.

V. MULTI-AXIS LOW-AMPLITUDE LINEAR MODELS

In 1971, Kaleps et al. [23] developed a two-axis, 5-mass model. The masses corresponded to the torso, lung and trachea, chest wall, abdomen, and buttocks. A thoracic cavity was formed by the torso, lungs and trachea, chest wall, and abdomen. The model was used to obtain the body deformation (spinal compression, pressure in the lungs, etc.) under external vertical forces (whole-body vibration and impact used as input) and pressure loads (blast, acoustic pressure, decompression loads as input). The parameters of the model were derived by best approximating the experimental data available. This model combined the mechanical response characteristics of the corresponding body segments. The model performance was compared with the impact impedance measurements from Coermann [24] and the experimental data of the resonance from Clark et al. [25]. The model behavior was similar to the experimental data for each. This model is suitable for obtaining responses of the lung, chest, and abdomen.

In 1978, Jex and Magdaleno [26] developed a 6-mass biomechanical model of a human sitting on a semisupine seat. The model included an active neuromuscular subsystem. It consisted of masses for the head, neck, torso, lower body, upper arm, and forearm. The mass and inertia were obtained from human properties by Braune et al. [27]. The stiffness and damping values were obtained by matching the experimental results from Magdaleno and Jex [28.29]. The model was then validated by comparing model results with the experimental data such as shoulder and head motions under sine and quasirandom vibration obtained by the authors. The model match was quite good except that the phase
of the head response of the model was much different from that of the experimental data. This model was used to simulate an aircraft pilot when in a flying (partially reclined) position.

In 1988, Amirouche and Ider [23] developed a 13-mass, multi-axis model of a human used for both seated and standing postures. It included masses for the head, neck, upper, center, and lower torso, upper and lower arms, and upper and lower legs. The masses were obtained from the Part 572 dummy. The stiffness and damping coefficients in both vertical and rotary directions were obtained by matching the experimental results from Panjabi et al. For the vertical transfer function of the middle torso, the model simulation was very similar to the experimental data in the 5- to 7-Hz range, but was larger in the 2- to 5-Hz range and smaller in the 7- to 15-Hz range. The model phase angle of the vertical acceleration of the middle torso had similar values as the experimental data in the frequency range from 2 to 5 Hz, but had smaller values in the 5- to 13-Hz range. For the rotary transfer function of the middle torso, the model did not match the experimental data in either shape or peak value. The transmission of vibration from the seat to the head was compared with experimental data obtained by Pradko et al. [30], Sandover [14], Donati and Bonthoux [31], Coermann [24], and Griffin et al. [32]. In general, the model did not match the experimental data particularly well. The model compared fairly well with only the shape of the experimental data for relaxed posture by Coermann [24]. Compared with other experimental data, the model had larger transmissibility values in the lower frequencies and had smaller values in the higher frequencies.

In 1994, Qassem et al. [33] presented an 11-mass, two-axis model of a seated human subjected to input forces at the hand and the seat along vertical and horizontal axes. This model was obtained by modifying the model presented by Muksian [3].

Figure 5 shows this model. It included masses for the head, neck (cervical spine), thoracic spine, lumbar spine, torso, upper and lower arms, hand, thorax, diaphragm, abdomen, and pelvis. The mass of each part was obtained from Muksian and Nash [3], and Nigam and Malik [17]. The spring and damper values were obtained from previous studies by Mizrahi and Susak [34], Nigam and Malik [17], and Patil et al. [35]. The seat-to-head, seat-to-torso,and hand-to-lower arm force transmissibility of the model were compared with those of the experimental measurements in the frequency range between 4 and 500 Hz obtained in their study. For the seat-to-torso and hand-to-lower arm force transmissibility, the model matched the experimental data well. For the seat-to-head force transmissibility, the model did not match the experimental data very well, especially in the frequency range from 4 to 40 Hz for which the error was 40 to 100%.

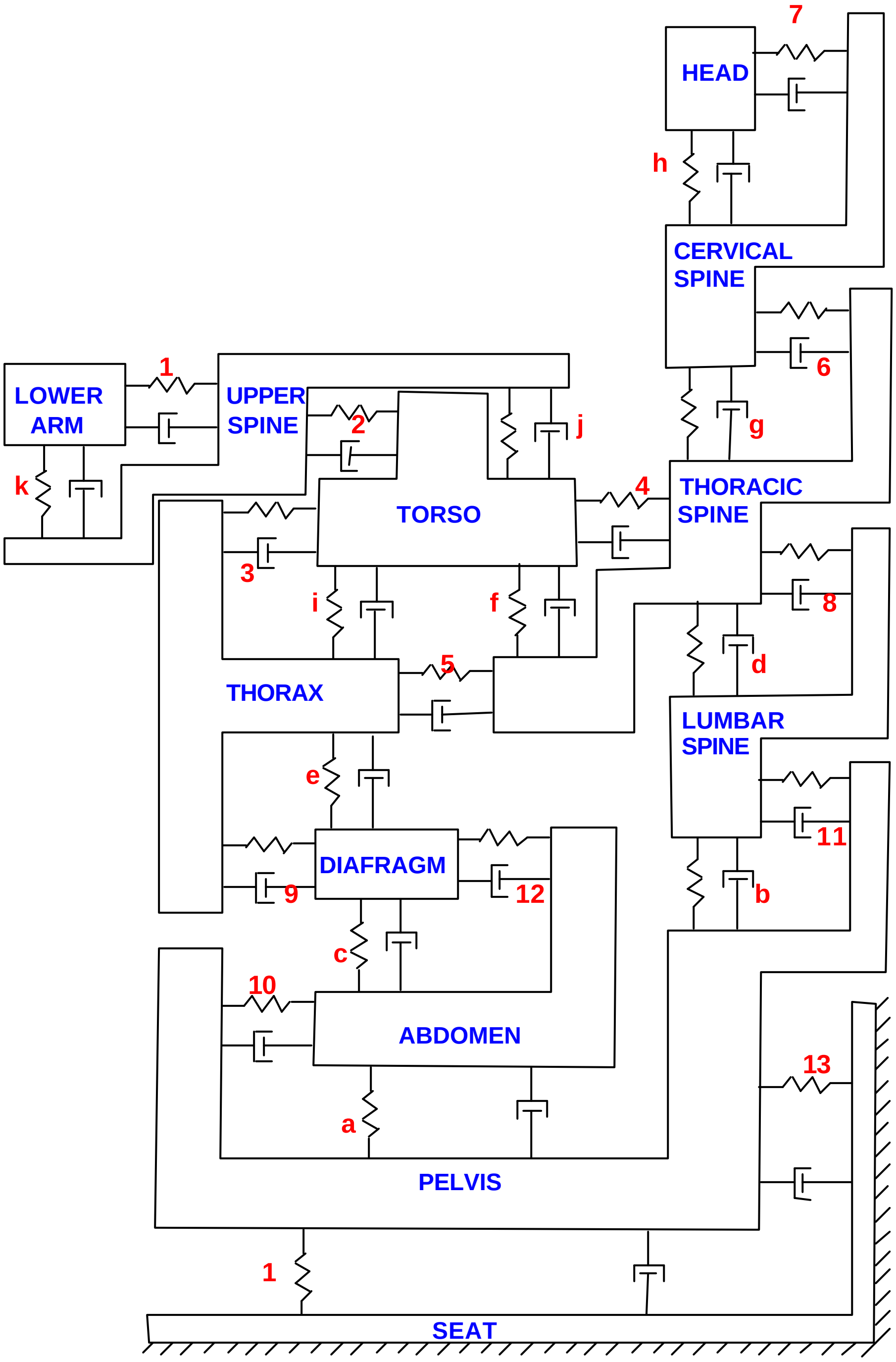

Figure 5.
Multi-axis low-amplitude linear Qassem model

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